②

REPORT NO. NADC-85040-60

AD-A155 912

MATCHED FILTERING OF VISUAL EVOKED POTENTIALS TO DETECT ACCELERATION (+Gz) INDUCED BLACKOUT

John G. Nelson, Leonid Hrebien and Joseph P. Cammarota
Aircraft and Crew Systems Technology Directorate
NAVAL AIR DEVELOPMENT CENTER
Warminster, Pennsylvania 18974

3 JANUARY 1985

FINAL REPORT
AIRTASK NO. WR04101001
WORK UNIT NO. DG148

Approved for Public Release; Distribution is Unlimited

Prepared for
NAVAL AIR SYSTEMS COMMAND (AIR-310H)
Department of the Navy
Washington, DC 20360

NOTICES

REPORT NUMBERING SYSTEM — The numbering of technical project reports issued by the Naval Air Development Center is arranged for specific identification purposes. Each number consists of the Center acronym, the calendar year in which the number was assigned, the sequence number of the report within the specific calendar year, and the official 2-digit correspondence code of the Command Office or the Functional Directorate responsible for the report. For example: Report No. NADC-78015-20 indicates the fifteenth Center report for the year 1978, and prepared by the Systems Directorate. The numerical codes are as follows:

CODE	OFFICE OR DIRECTORATE
00	Commander, Naval Air Development Center
01	Technical Director, Naval Air Development Center
02	Comptroller
10	Directorate Command Projects
20	Systems Directorate
30	Sensors & Avionics Technology Directorate
40	Communication & Navigation Technology Directorate
50	Software Computer Directorate
60	Aircraft & Crew Systems Technology Directorate
70	Planning Assessment Resources
80	Engineering Support Group

PRODUCT ENDORSEMENT — The discussion or instructions concerning commercial products herein do not constitute an endorsement by the Government nor do they convey or imply the license or right to use such products.

APPROVED BY: T. J. Gallagher DATE: 28 May 1985

T. J. GALLAGHER
CAPT, MSC, U.S. Navy

Unclassified

SECURITY CLASSIFICATION OF THIS PAGE

AD-A155912

REPORT DOCUMENTATION PAGE

1a REPORT SECURITY CLASSIFICATION Unclassified	1b RESTRICTIVE MARKINGS N/A
2a SECURITY CLASSIFICATION AUTHORITY N/A	3 DISTRIBUTION/AVAILABILITY OF REPORT Approved for Public Release; Distribution Unlimited
2b DECLASSIFICATION/DOWNGRADING SCHEDULE N/A	
4 PERFORMING ORGANIZATION REPORT NUMBER(S) NADC-85040-60	5 MONITORING ORGANIZATION REPORT NUMBER(S) N/A
6a NAME OF PERFORMING ORGANIZATION Aircraft and Crew Systems Technology Directorate 6b OFFICE SYMBOL (If applicable)	7a NAME OF MONITORING ORGANIZATION N/A
6c ADDRESS (City, State, and ZIP Code) Naval Air Development Center Warminster, PA 18974	7b ADDRESS (City, State, and ZIP Code) N/A
8a NAME OF FUNDING/SPONSORING ORGANIZATION Naval Air Systems Command 8b OFFICE SYMBOL (If applicable)	9 PROCUREMENT INSTRUMENT IDENTIFICATION NUMBER N/A
8c ADDRESS (City, State, and ZIP Code) AIR-310H Washington, DC 20360	10 SOURCE OF FUNDING NUMBERS PROGRAM ELEMENT NO: 61153N PROJECT NO: WR04101 TASK NO: WR04101001 WORK UNIT ACCESSION NO: DG148

11 TITLE (Include Security Classification)

Matched-Filtering of Visual Evoked Potentials to Detect Acceleration ($+G_z$) Induced Blackout

12 PERSONAL AUTHOR(S)

John G. Nelson, Leonid Hrebien, Joseph P. Cammarota

13a TYPE OF REPORT	13b TIME COVERED	14 DATE OF REPORT (Year, Month, Day)	15 PAGE COUNT
Final	FROM ______ TO ______	1985 January 3	22

16 SUPPLEMENTARY NOTATION

17 COSATI CODES			18 SUBJECT TERMS (Continue on reverse if necessary and identify by block number)
FIELD	GROUP	SUB-GROUP	Visual Evoked Potential; Acceleration ($+G_z$); Blackout; Matched-Filtering

19 ABSTRACT (Continue on reverse if necessary and identify by block number)

In air-combat maneuvering and on human centrifuges, moderate levels of positive acceleration (+ Gz), coupled with moderate rates of onset, produce visual symptoms which are ordinarily progressive: Decreasing visual sensitivity, dimming of the visual field, peripheral light-loss, and central light-loss (black-out). Since these symptoms provide warning of impending loss of consciousness, subjective visual symptoms are the most commonly used tolerance end point in acceleration research.

20 DISTRIBUTION/AVAILABILITY OF ABSTRACT ☐ UNCLASSIFIED/UNLIMITED ☐ SAME AS RPT ☐ DTIC USERS	21 ABSTRACT SECURITY CLASSIFICATION
22a NAME OF RESPONSIBLE INDIVIDUAL	22b TELEPHONE (Include Area Code) / 22c OFFICE SYMBOL

DD FORM 1473, 84 MAR

83 APR edition may be used until exhausted
All other editions are obsolete

SECURITY CLASSIFICATION OF THIS PAGE

Cont'd

Cont'd

In order to provide an Objective indication of the integrity of the visual system, we have developed (and continue to improve) a method for real-time monitoring of the steady-state visual evoked potential (VEP). Since significant degradation of visual functioning must be recognized in less than approximately four seconds, we required a previously unattained efficiency in producing and measuring the VEP. Using the Fast Fourier Transform (FFT), we developed a method for maximizing the signal-to-noise ratio: A digital, frequency domain, non-white-noise matched filter, with evaluation only at the expected response peak. The coefficients of the matched filter are determined empirically by analysis of test data obtained in a static run, just prior to the dynamic ($+G_z$) run.

Experiments on the Navy's Human Centrifuge demonstrated that the response does progressively decrease, disappearing at black-out. Improved computer facilities have permitted evaluation of alternative methods of processing, and the effectiveness of such processing.

Analysis of new data from a static (non-centrifuge) experiment using four stimulus repetition rates and two electrode positions showed that windowing of the time record prior to FFT does not necessarily improve detection. In addition, it was found that the initial signal-to-noise ratio (S/N) of the VEP was −20 db. Matched-filtering, using a 3-second sliding average, gave an improvement in S/N of 30 to 33 db.

TABLE OF CONTENTS

List of Figures

ABSTRACT

In air-combat manueuvering and on human centrifuges, moderate levels of postitive acceleration ($+G_Z$), coupled with moderate rates of onset, produce visual symptoms which are ordinarily progressive: decreasing visual sensitivity, dimming of the visual field, peripheral light-loss, and central light-loss (black-out). Since these symptoms provide warning of impending loss of consciousness, subjective visual symptoms are the most commonly used tolerance end point in acceleration research.

In order to provide an objective indication of the integrity of the visual system, we have delveloped (and continue to improve) a method for real-time monitoring of the steady-state visual evoked potential (VEP). Since significant degradation of visual functioning must be recognized in less than approximately four seconds, we require a previously unattained efficiency in producing and measuring the VEP. Using the Fast Fourier Transform (FFT), we developed a method for maximizing the signal-to-noise ratio: digital, frequency domain, non-white-noise matched filter, with evaluation only at the expected response peak. The coefficients of the matched filter are determined empirically by analysis of test data obtained in a static run, just prior to the dynamic ($+G_Z$) run.

Experiments on the Naval Air Development Center's Human Centrifuge demonstrated that the response does progressively decrease, disappearing at black-out. Improved computer facilities have permitted evaluation of alternative methods of processing, and the effectiveness of such processing.

Analysis of new data from a static (non-centrifuge) experiment using four stimulus repetition rates and two electrode positions showed that windowing of the time record prior to FFT does not necessarily improve detection. In addition, it was found that the initial signal-to-noise ratio (S/N) of the VEP was -20 db. Matched-filtering, using a 3-second sliding average, gave an improvement in S/N of 30 to 33 db.

INTRODUCTION

During exposure to sufficiently high $+G_Z$ acceleration, subjects on human centrifuges (and pilots of high-performance aircraft) experience the ordinarly progressive symptoms of peripheral light loss, gray-out, black-out, and ultimately loss of consciousness (LOC), due to the reduction or elimination of blood supply to the eye and brain (Duane, 1954, Leverett et al., 1966). With moderate rates of acceleration onset, each stage of visual loss is usually premonitory of the next, providing a margin of safety with respect to inadvertent LOC. Therefore, measures of visual functioning are the usual end-points in acceleration research (Burton et al., 1974; Coburn, 1970; Duane et al., 1962, Krutz et al., 1975) and visual symptoms are also the cue used by pilots to adjust "G-dosage" to tolerable limits. On the human centrifuge, we ordinarily use the NAVAIRDEVCEN Light-Bar (Cohen, 1981), as shown in Figure 1. With this, the subject uses a control stick to keep right and left lights (Light Emitting Diodes in a continuous string) at the edge of what remains of his peripheral vision. The test run is terminated when vision has tunneled to a 60° cone.

Since the experimentally determined end point is the basic datum for our comparions of G-protective devices and methods, an objective indication of the integrity of the visual system is very desirable. The fleet requirement which we address is improved G-tolerance, and this requires the best methods for evaluating proposed improvements. Better methods increase the precision of the experiment, and/or decrease the number of tests required to reach a decision.

In order to provide this objective indication, we have developed a method for real-time monitoring of the steady-state visual-evoked potential (VEP). (Nelson, et al., 1980-1983). The VEP is a small signal measured by surface electrodes on the back of the head, and is produced by viewing a flashing light (Regan, 1972 & 1977). (Figure 1 also shows our strobe light mounted at the center of the light bar.) Since significant degradation of visual functioning must be recognized by us in less than four to five seconds, and since the VEP has a signal-to-noise ratio (S/N) of the order of -20 db, we required a previously unattained efficiency in producing and measuring it. (Note that the noise is the basic background electroencephalogram, not electronic noise.) Time-averaging, which is the most commonly used method for S/N improvements in VEP studies, is definitely not adequate for our requirements for speed of response (Donchin and Lindsley, 1969).

While many other noise reductive methods for real-time monitoring or off-line analysis of the VEP have been used (Regan, 1977, Eisenstein & Morgan, 1982, J.I. Nelson 1984, McGillem and Aunon, 1983), our method appears to be the only one to attempt a formal maximization of S/N, with both signal and noise characterised empirically. In brief, we use the Fast Fourier Transform (FFT) to implement a non-white-noise matched filter to detect the VEP. This method was applied in an experiment on the NAVAIRDEVCEN Dynamic Flight Simulator (Human Centrifuge), and presented at the 1983 Annual Meeting of the Aerospace Medical Association (Nelson, Hrebien, Palumbo, and Cohen, 1983). That experiment showed that the VEP decreases as the tolerance limit is approached, and probably disappears at CLL, confirming earlier work by Duane et al., 1962. Figure 2 shows a specimen real-time run, and Figure 3 shows the weighted mean for all such runs from that study. For slow onset-rate $+G_Z$ ramps to relaxed tolerance, compensation appears adequate to prevent visual problems for the first 60% of the accleration ramp, as shown in Figure 3. beyond that point, increases in acceleration result in a rather regular and progressive decrease in this measure of visual functioning. This decrease occurs over approximately 22 sec. under these conditions. In a previous (static) study (Nelson and Hrebien, 1982 (b)), mechanical occlusion of the stimulus resulted in precipitous loss of response.

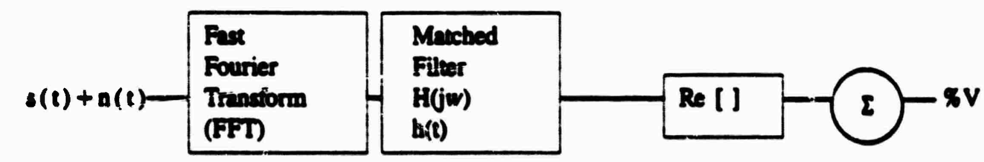
We now have a new computer system which gives us greatly expanded capabilities. It consists of a Digital Equipment Corp MINC (LSI-11/23) computer, two 10.2-megabyte cartridge discs, an 800 & 1600 bpi magnetic tape transport, analog to digital converter (A/D) with direct memory access (DMA), digital to analog conversion (D/A), IEEE-488 interface with DMA, digital plotter, printer, and Sky Computer vector processor. The vector processor is capable of one million floating-point operations per second (1 megaflop), using DMA.

This system permits us to calculate FFT's and other functions much more rapidly, and therefore more frequently, in real-time and off-line. (As a bench-mark example, a 1024 pt. complex floating point FFT requires 60 msec). The discs and magnetic tape permit the recording of all raw data, so that alternative processing methods can be applied to the same data set. The industry-standard mag-tape permits data processing on other systems, such as the NAVAIRDEVCEN Central Computer, which have extensive hardware and software resources.

In an analysis directed toward simplification of the computations required by our VEP signal detection technique (Nelson and Hrebien, (1982 (a)), it became apparent that this method, which was derived from elementary compound variables theory (Guilford, 1950 & 1977), is an implementation of an non-white-noise (NWN) matched filter, which originated in the processing of radar signals. As in many signal detection techniques, we desired to operate on an imput of signal plus noise (VEP + background EEG) with a linear filter to produce an output, whose S/N is maximum at a selected time. It has been shown that this maximization is only achieved by the use of a NWN matched filter (Turin, 1960).

A detailed examination of the application of the rotations of the optimal weights computed in the previous method revealed that they performed the same linear operations that a NWN matched filter would have performed, given the same signal and noise conditions. Realizing this, we save considerable computation time by the direct implementation of the NWN matched filter. The description of the method is also much more compact.

This filtering process is indicated:



With definitions being:

s(t)	=	VEP (as function of time)
n(t)	=	background EEG
h(t)	=	filter impulse response
S(Jw)	=	Fourier transform of s(t)
$\|N(jw)\|^2$	=	Power spectral density of n(t)
H(jw)	=	Filter transfer function
% V	=	percent base vision

The filter transfer function is given by:

$$H(jw) = \frac{K\, S^*(jw)\, e^{-Jwt}}{|N(jw|^2},$$

where K is an arbitrary scaling constant and t is a time delay necessary to make the filter realizable (t is inherent in the sampling process).

A direct implementation of this filter would start with the estimation of $S^*(jw)$, $|N(jw|^2$, using the parameter estimation runs. Computations in the real-time runs then consist of taking the Fourier Transform of the VEP + EEG, multiplying this by H(jw, then summing the real components of this product to obtain a single quantity indicating percent vision.

The matched filter's coefficients are determined by analysis of data obtained in a static run, just prior to the dynamic ($+G_z$) run on the centrifuge, thereby adapting the filter to the unique combination of subject, electrode placement, stimulating frequency, and other variables, including some uncontrolled factors.

Given the new computer capability, we designed and completed an experiment to investigate some of the questions which had accummulated from our prior analysis and experiments: is "windowing" beneficial, what is the underlying S/N, what improvement in S/N is achieved, what is this effect of electrode position, etc. An additional purpose of the experiment was to provide an archived data-base where some new questions might be investigated without the requirement to collect new data.

METHOD

Using a repeated measures design, eight subjects (Ss) were tested with four strobe repetition rates (2,4,8, and 16Hz), with three test runs per S at each rate. The test runs were two minutes in duration, with one minute rest periods interspersed. The visual evoked potential (VEP) was measured at two electrode locations (primary and secondary visual projection areas).

Subjects were male Navy enlisted (seven) and officer (one) volunteers from the laboratory's medical support contingent. Informed consent was obtained as required by the laboratory's Committee for the Protection of Human Subjects. The Committee also approved the experimental protocol.

Tests were conducted in a sound-attenuating chamber, which also provided electrical shielding, and S's wore both ear-plugs and ear-defenders to attenuate strobe sound, in order to avoid contamination of the VEP by an auditory EP. No S's reported being able to see the subjects eyes, face, and posture. While we have never observed strobe-induced seizure, S's were informed of the risk, a medical corpsman was in attendance, and a flight surgeon was on call.

Primary electrodes were placed at positions due to Drasdo (1980), who reported an anatomical study of the location of the visual projection areas. The first electrode was positioned on the midline at 5% (of the inion-nasion distance) above the inion. This is midway between the inoin and Oz in the international 10-20 system (Jasper, 1958). The second primary electrode was located 20% (of the inion-nasion distance) to the right of the first electrode. Both electrodes were referred to Pz (in the 10-20 system), with right mastoid as ground.

The electrodes were Beckman minature silver/silver-chloride, stored in normal saline. These electrodes were modified to permit injection of the electrode paste (Signa gel) after application. This modification was made after observation of EEG lab procedures. Electrode sites were cleaned with Omni-Prep, rubbed with paste, and the electrodes were secured with collidion (non-flexible) and blown dry. This procedure is fast, reliable, and the electrodes are much more securely attached. Electrode-pair impedence was held to less than 4K ohms.

The strobe-light stimulus, a General Radio Model 1538-A, was enclosed in a gounded metal box, with copper screen over the light output. The light was routed through an L-shaped box, painted flat-white internally, to a frosted 14 cm. diameter screen, with a one cm. fixation mark. The eye-to-screen distance was 1.17 meters, giving a visual angle of 7 degrees. When the strobe was set at nominal 4170 flashes per minute, no additional attenuation was required to provide a viewing intensity which was acceptable to the subjects.

An infra-red eye-blink detector was constructed, tested, and abandoned. The strobe-light pulsed the detector, which induced coherent noise in the electrode leads. Without the eye-blink detector, coherent noise was undetectable in runs in which the subject was blindfolded. Figure 4 shows % vision during such a run, while Figure 5 shows a specimen standard run without the blindfold.

Both the data channels were low-pass (100 Hz, 40db/decade), sampled at 1024 sampled/sec. and the data stored on digital magnetic tape. Figure 6 shows the general connections of the system elements: strobe light with trigger generator, the low-pass filtered EEG, the A/D, clock, mag tape, and DECLAB/MNC-23 computer.

RESULTS AND DISCUSSION

In all of our earlier work on the VEP, we made a somewhat arbitrary decision to use a uniform weighting (rectangular window) for the time-function data as opposed to using the von Hann, Hamming, or other window (Harris, 1978). Non-uniform windows are symetric weighting functions with weights smoothly decreasing, closely approached zero as the function nears its beginning and end. The choice of the uniform window as based upon the fact that all of our signal components have been selected to be whole-cycle within the slected time frame, with noise in general not being whole cycle. A uniform window will be maximally sensitive to such signals, and will distort only the spectral estimates of the noise, not the estimates of the signal.

Our new computer system, with its data collection, storage and processing capabilities, permitted us to evaluate our previous choice of window. The data were processed twice, with two solutions per second, with the percent vision variable being output at the same rate, with each % vision estimate the mean of the last six solutions, giving a three-second running average. A 3-second average was chosen, since detection of loss of vision within 3 seconds would be satisfactory for our application. When processing with the uniform window, ½ sec. time frames were used. When processing with the non-uniform window, one-second time frames were used, with 50% overlap.

The non-uniform window selected was the minimum 3-term Blackman-Harris (B-H) (Harris, 1978). This window has an equivalent noise bandwidth of 1.71, with its first sidelobe at -67 db. With 50% overlap, estimates are essentially independent, since the shared variance (correlation squared) is only 1%, since r = .096. Also, the level of the first sidelobe of the window is nearly down to the -72 db resolution noise-floor of our A/D converter.

For each run, for each window, matched filters were derived and used to output % Vision for that run. Coefficients of variation (CV: standard error of % vision, divided by mean % vision, scaled as %) were calculated. A chi-square evaluation of the relative frequency of superiority of the two windows showed the uniform window to be slightly (but quite non-significantly) superior.

Using the ratio of the two CV's resulting from application of the two windows, an analysis of variance showed frequency of stimulation to be marginally significant (p. .05). An examination of the mean values showed that the B-H window improved mean performance for the 2Hz stimulus (10% improvement) and for 4Hz (7% improvement), with the uniform window superior (by only 2%) at 8 and 16Hz. When correlations of CVs between the three runs in each set were examined, the B-H window was found to produce moderately higher correlations between runds. A further ANOVA of CV's, with uniform window for run 1, and B-H windowing for run 2, showed frequency again to be only marginally significant (p .05), but the window-by-frequency effect was signficant at the 1% level.

Since the B-H window did not show clear advantages for this set of data, particularly at those stimulation frequencies giving better detectability, the uniform window was used in further analysis. However, since window choice did affect results, choice of window will continue to be evalulated in future applications to determine what is best for that particular data set.

Using the CV derived with the uniform window, an ANOVA showed frequency of stimulation to be significant (p .01). The mean values of CV for 2, 4, 8, and 16Hz stimulus repetition rate were 70, 51, 44, and 56 respectively, with 8Hz being the best stimulus overall.

In this and other comparisons, electrode position was non-significant. Since even visual inspection of the signals at the two electorde sites indicated gross differences in frequency composition, as shown in Figure 7, the approximate equality of the two sites is encouraging in that combining channels in future applications should give a useful improvement in detection. This is due to the fact that while there are moderate correlations of the noises at the two sites at the same frequencies, the noises at different frequencies are independent.

The formal model of the VEP (signal) combined with background EEG (noise) required by matched filter theory is one of independent additivity. In this, as in previous studies, we have found only small deviations of data from model. Figure 8, from a previous study wiht a strobe repetition rate of 3.2 Hz, shows a slight but consistent elevation of noise power at the harmonic points, relative to the adjacent non-harmonic points. This was also found in the present study. The most parsimonious interpretation of this is that it is due to a small degree of signal instability, contrary to the model. The data definitely do not support any photic-driving or capture theory (Regan, 1972, p. 77), wherein it is postulated that the backgound EEG is brought into syncronization with the stimulus, thereby producing the VEP. If capture occurred, it would decrease the noise at the harmonic points. The additive-independent model appears to provide an adequate approximation.

Since the reason for applying a matched filter to VEP data is to maximize the signal-to-noise ratio (S/N), an analysis in using the unweighted, unprocessed ensemble of time-functions. The overall median for this initial S/N was -19.4 db. The median S/N of the % vision variable, using the 3 second running average, and applying the filter to the data set from which it was derived, was +14.3 db, for an improvement of 33.8 db. When the filter was derived from a run, then applied to the next run, the median shrinkage was 3.8 db, for a net 30 db improvement. This shrinkage indicates that adaptive estimation procedures may provide a worthwhile improvement in detection, particularly when shifts in background EEG are expected.

In this and other comparisons, electrode position was non-significant. Since even visual inspection of the signals at the two electorde sites indicated gross differences in frequency composition, as shown in Figure 7, the approximate equality of the two sites is encouraging in that combining channels in future applications should give a useful improvement in detection. This is due to the fact that while there are moderate correlations of the noises at the two sites at the same frequencies, the noises at different frequencies are independent.

The formal model of the VEP (signal) combined with background EEG (noise) required by matched filter theory is one of independent additivity. In this, as in previous studies, we have found only small deviations of data from model. Figure 8, from a previous study wiht a strobe repetition rate of 3.2 Hz, shows a slight but consistent elevation of noise power at the harmonic points, relative to the adjacent non-harmonic points. This was also found in the present study. The most parsimonious interpretation of this is that it is due to a small degree of signal instability, contrary to the model. The data definitely do not support any photic-driving or capture theory (Regan, 1972, p. 77), wherein it is postulated that the backgound EEG is brought into syncronization with the stimulus, thereby producing the VEP. If capture occurred, it would decrease the noise at the harmonic points. The additive-independent model appears to provide an adequate approximation.

Since the reason for applying a matched filter to VEP data is to maximize the signal-to-noise ratio (S/N), an analysis in using the unweighted, unprocessed ensemble of time-functions. The overall median for this initial S/N was -19.4 db. The median S/N of the % vision variable, using the 3 second running average, and applying the filter to the data set from which it was derived, was +14.3 db, for an improvement of 33.8 db. When the filter was derived from a run, then applied to the next run, the median shrinkage was 3.8 db, for a net 30 db improvement. This shrinkage indicates that adaptive estimation procedures may provide a worthwhile improvement in detection, particularly when shifts in background EEG are expected.

Shifts in the background EEG spectrum are to be expected under conditions of $+G_Z$ acceleration (Berkhout, et al., 1973; Duane, et al., 1962; the Sem-Jacabsens, 1963; Squires, et al., 1964), and the noise term of the matched filter should therefore adjust dynamically to reflect this.

When new data are available from acceleration runs, we will then be able to evaluate the trade-off between lags in dynamic estimates and instability of the estimates. The analysis of the dynamic data will also reveal whether up-dating of filter coefficients must cease as vision begins to be lost. It should also be noted that these shifts in the EEG spectrum will, in themselves, be of interest.

To express the obtained 30 db improvement in S/N in other terms, conventional sampling theory indicates that a sample size of 1000 is required to give this degree of reduction in error variance. In a 3-second time frame, wiht stimulus repetition rates varying from 2 to 16 Hz, the available sample size for time averaging ranges from 6 to 48 (whole cycles of the fundamental).

Looking at the results of the best stimulation frequency for each S we find 8 Hz is best 4 S's, with 4 and 16 Hz best for 2 illustrates that subjects differed in frequency composition of the response, as well as resulting S/N. Selecting the best stimulus for each S gave a 4.4 db difference between the S's with the highest and lowest S/N, while one S showed an 8.2 db difference between this best and worst frequency stimulation. Since it is likely that all of the stimulating frequencies were non-optimal for the individual S's, and in some cases grossly so, the use of a swept-frequency technique to determine a genuine optimum for each S should improve overall results, with radical improvement for S's with highly peaked responses (Spekreijse, 1966). A swept-frequency search is required, since system non-linearities prevent a more analytic approach to determining an optimum. Figure 10, the time averages of the response for one subject at three stimulus repetitions rates, shows the strong non-linearity of the response.

A new stimulating source is being designed to replace the strobe-light stimulus for our future experiments. This will permit simple or harmonically complex sinusoidally modulated light stimulation, with swept-frequency testing to determine the optimal stimulus for the individual S's.

In Figure 11, we show an analysis of all of the runs made wiht an 8 Hz strobe rate, summarizing the total normalized power and coherence as functions of frequency. (Coherence is the proportion of total power that is due to signal (VEP) power)(Bendat and Piersol, 1971). The 18 db range of noise demonstrates the requirement for non-white-noise assumptions when implementing the filter. The distribution of the coherence across many harmonic points demonstrates the advantages of processing the full harmonic content of the response. Single frequency filters, even of the (expensive) precision phase-sensitive lock-amplifier type, (J.I. Nelson, 1984) can only look at one, out of approximately eight, of the significant components of the signal.

The plot of the maximum coherence achieved in any run indicates that individual cases of useable components occur out to 100 Hz, so selection of a frequency cut-off should not be based upon average results. The nulls in the coherence function, which occur at the non-harmonic points, indicate that one of the functions of the matched filter is to serve as a comb filter, having zero gain at the non-harmonic points.

The large S/N improvements achieved to date have provided a measure which is a valuable adjunct to other measures, such as the light-bar or doppler blood flow to the brain. For use as a reliable end-point, the present improvement in S/N of approximately 33 db will need to be increased by an additional 5 to 6 db. This goal may be achieveable.

The signal processing methods developed during this effort, while directed toward the solution of our particular problem, the detection of loss of VEP under acceleration, could be applied to considerable advantage in other VEP research efforts, and to processing of psychophysiological data in general. Improved monitoring of VEP during cranial surgery would be an example of a practical clinical application (Feinsod et al., 1976). The elimination of any frequency components which are entirely noise, and the differential weighting in terms of S/N at the other frequencies would give much more sensitive measures, and could reveal important components of the response which would otherwise remain undetected. Phase information could be retained with only a 3 db penalty, since the matched filter achieves a 3 db improvement by rejection of quadrature noise. The processed signal could be viewed in the time domain, with a specified level of confidence that all of the features of the waveform represent non-random effects.

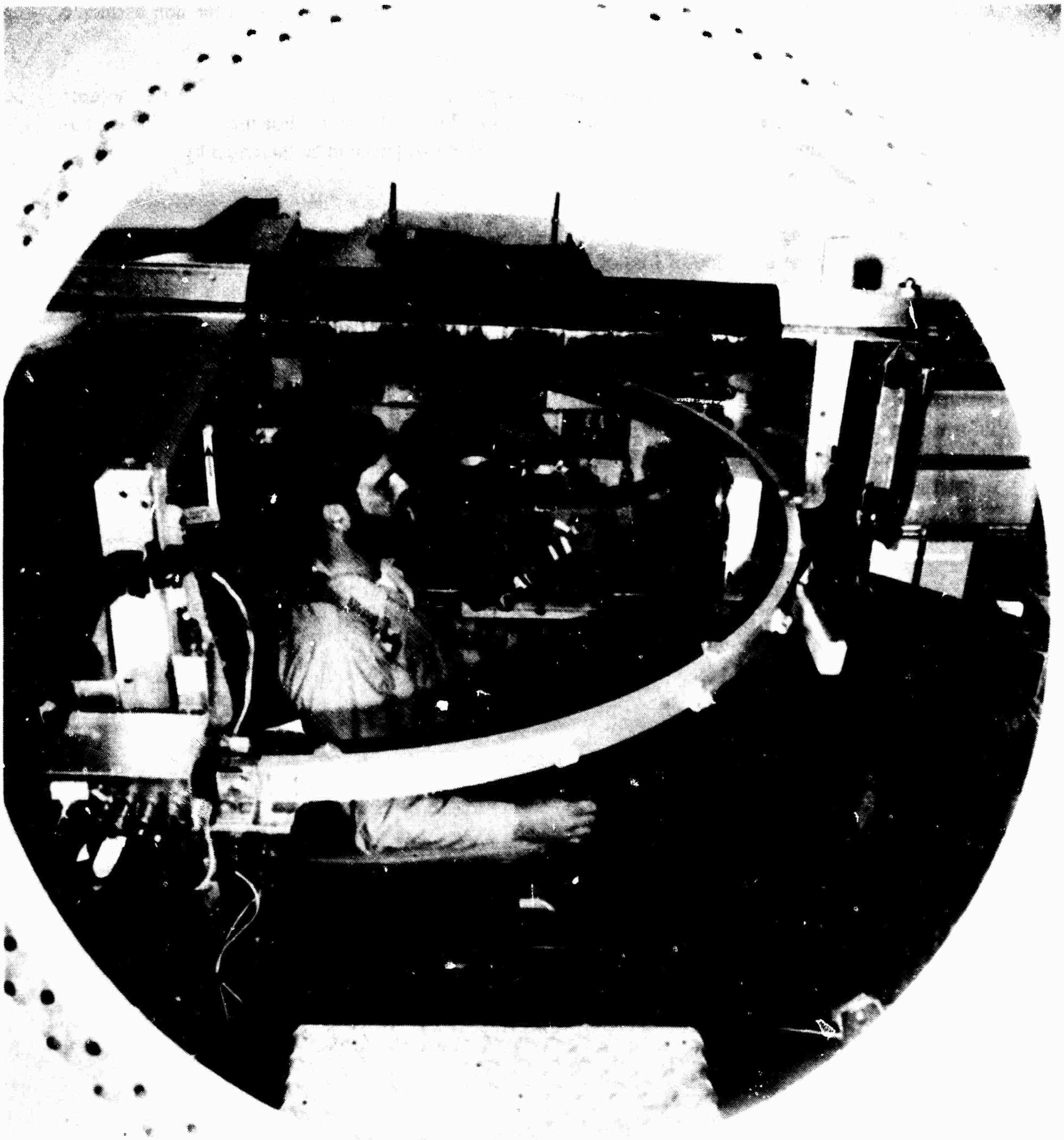


Figure 1. Naval Air Development Center light bar, used to dynamically track the loss of peripheral vision under +Gz acceleration.

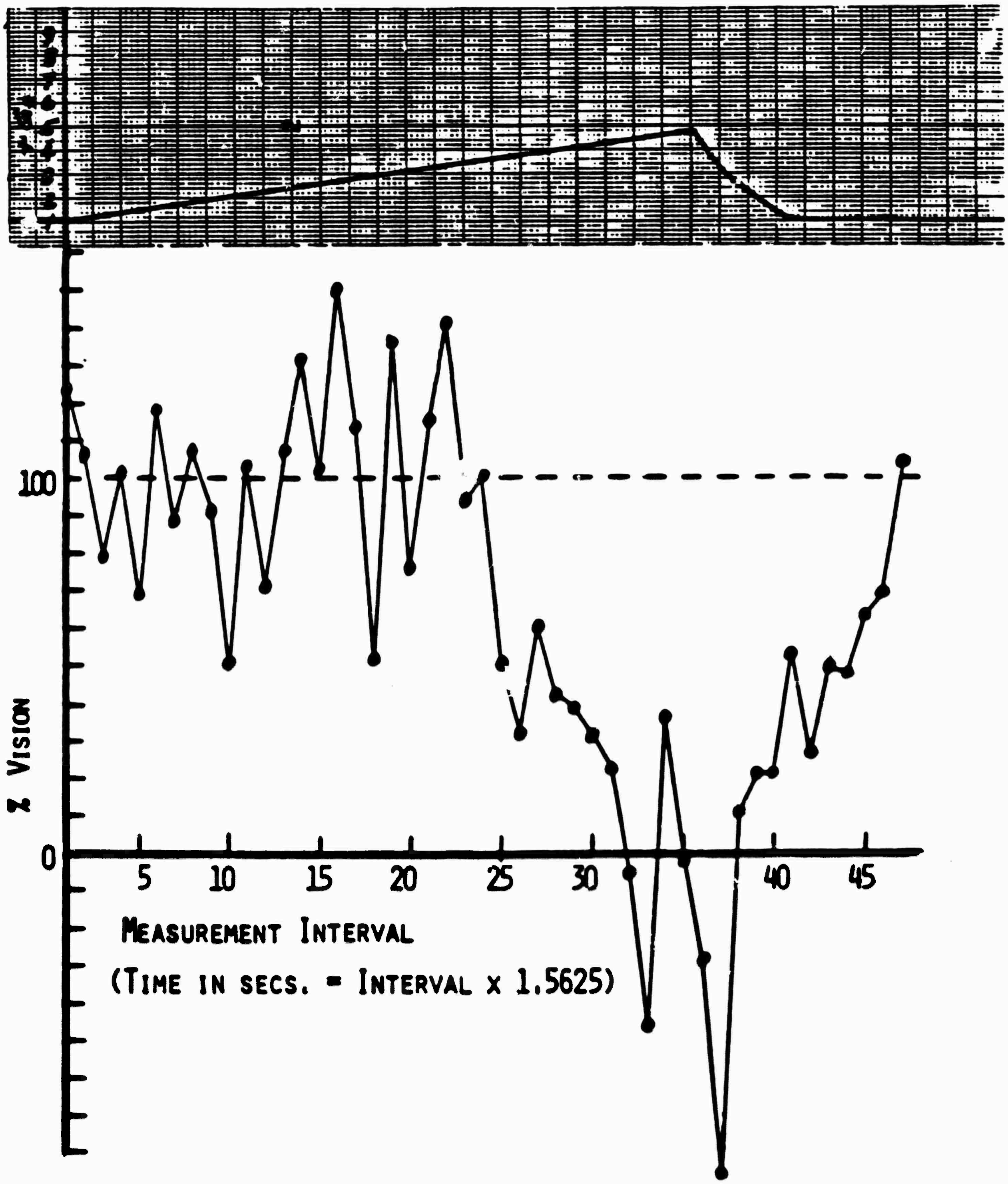


Figure 2. Specimen real-time run showing % vision as a function of $^{+}G_z$ acceleration.

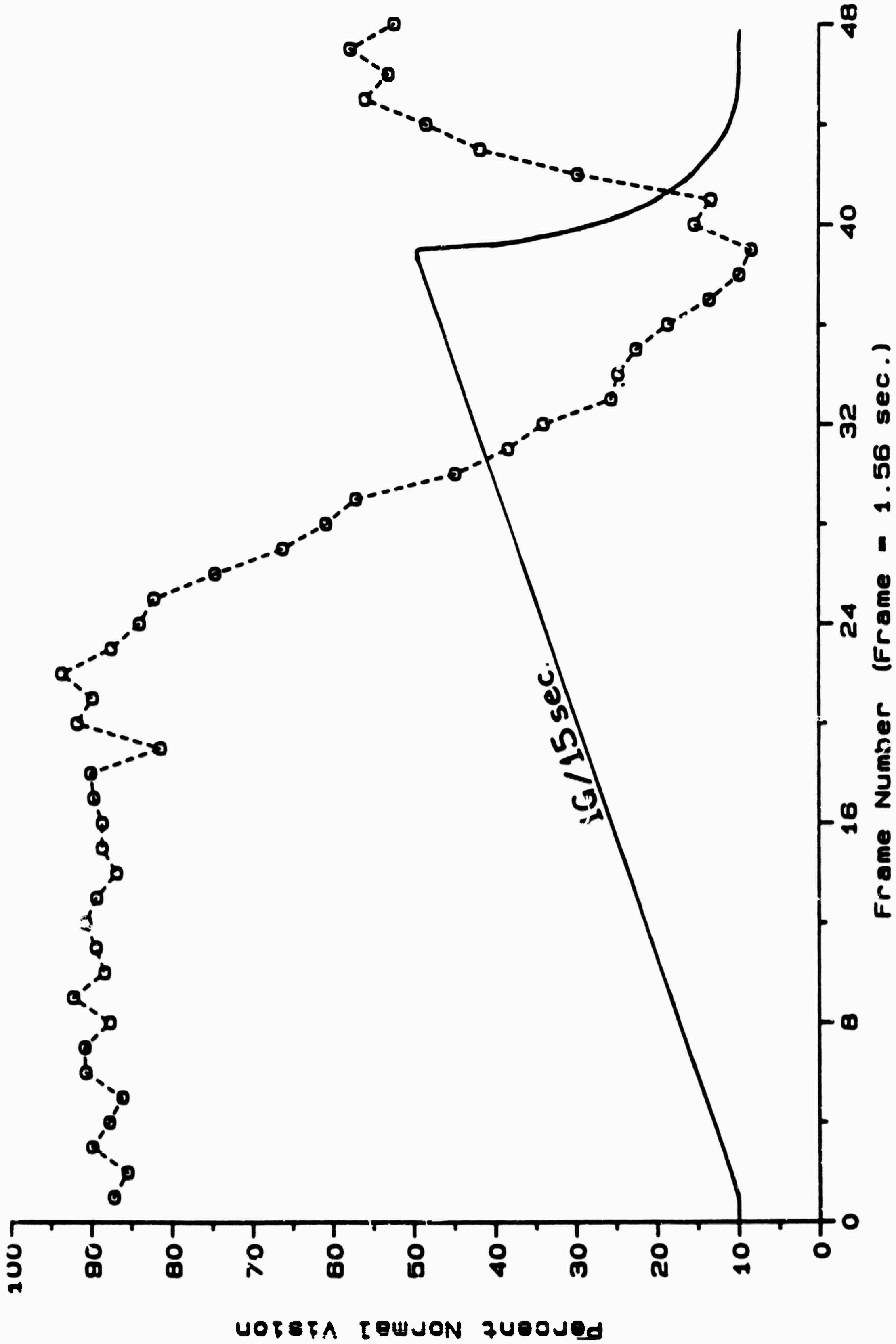


Figure 3. Weighted mean of % Base Vision for all +Gz acceleration runs, with 1 G/15 sec. acceleration ramp.

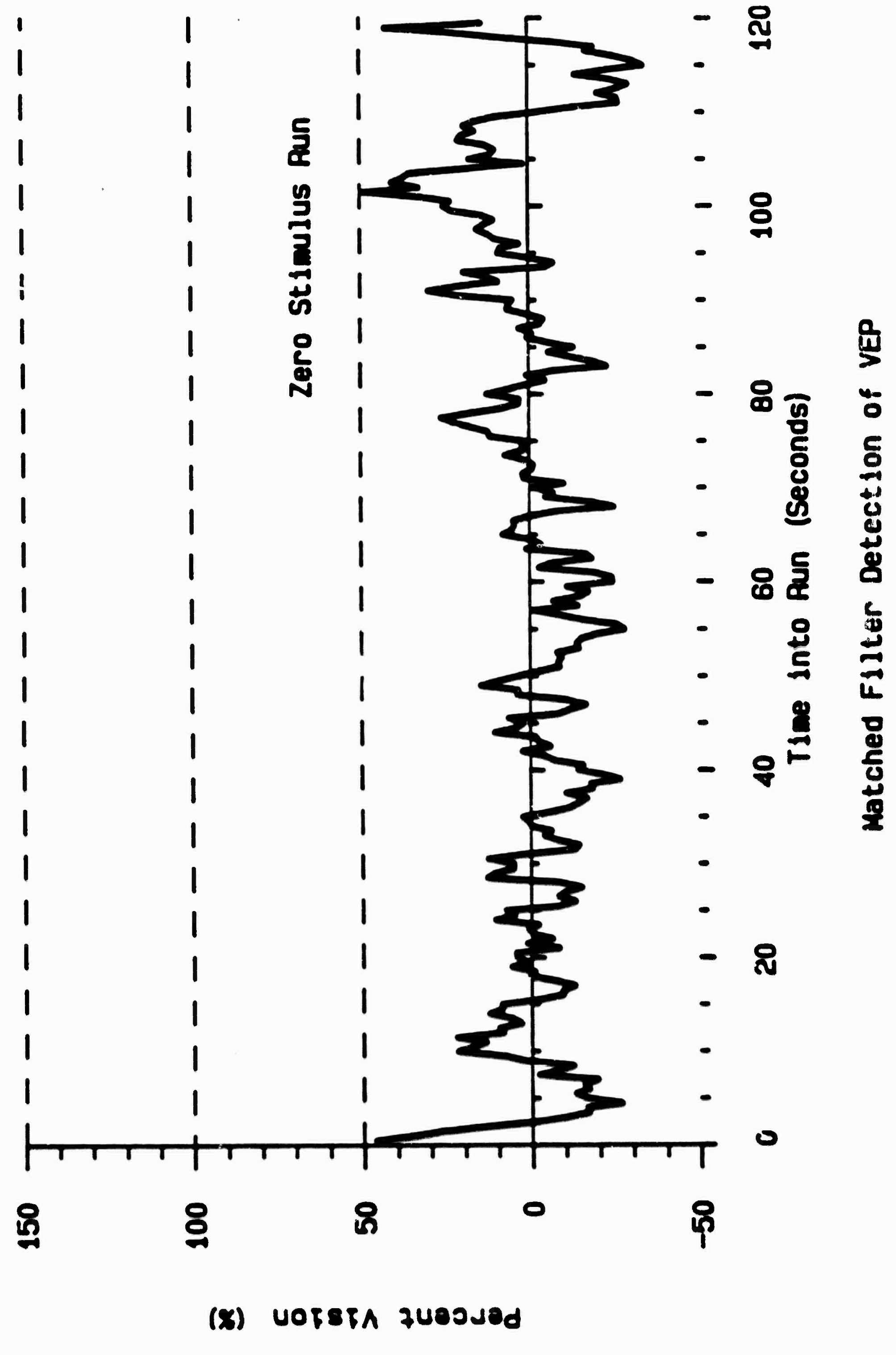


Figure 4. Blindfolded (zero-stimulus) static run, showing absence of response.

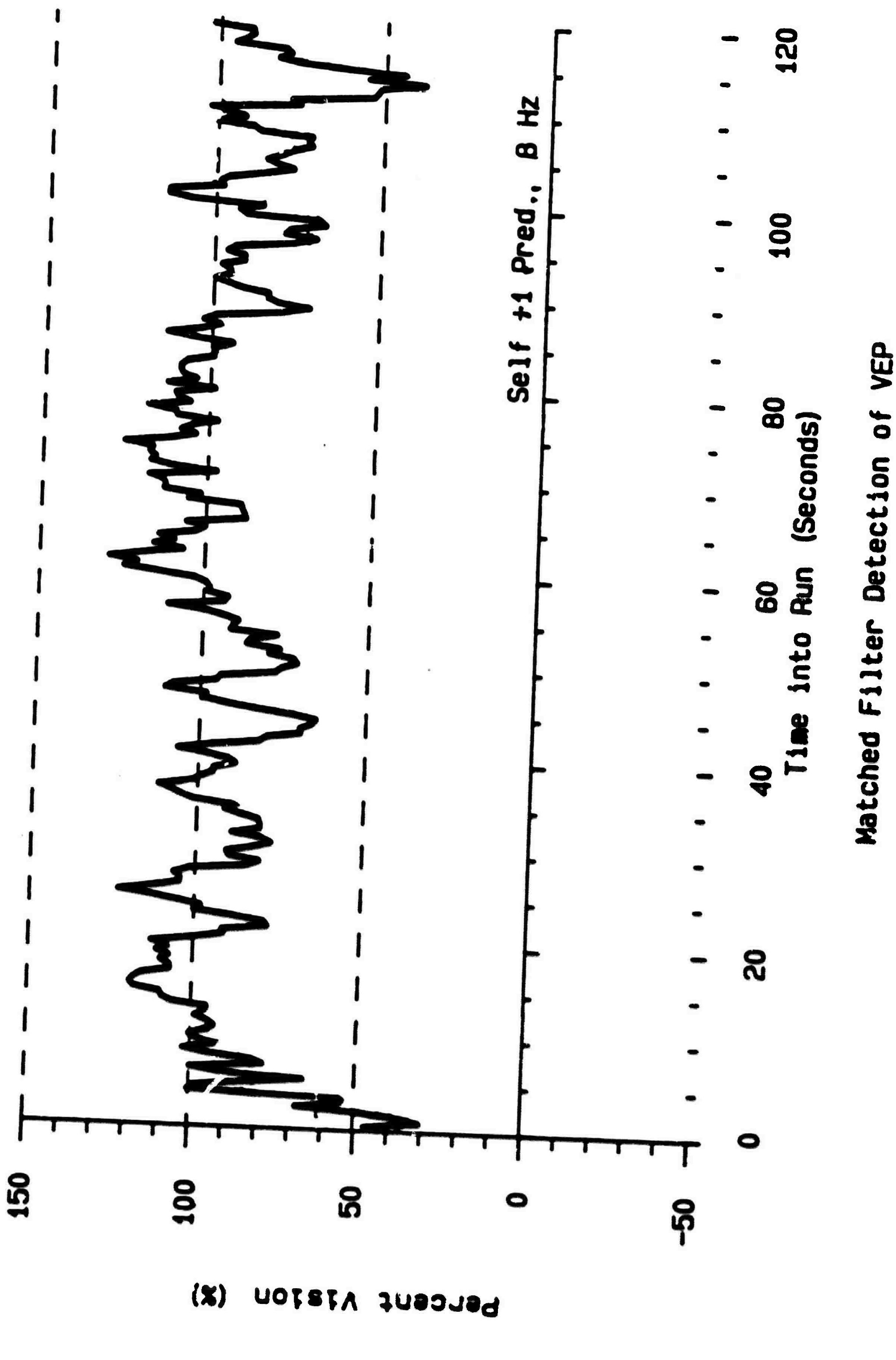


Figure 5. Static run, showing presence of response.

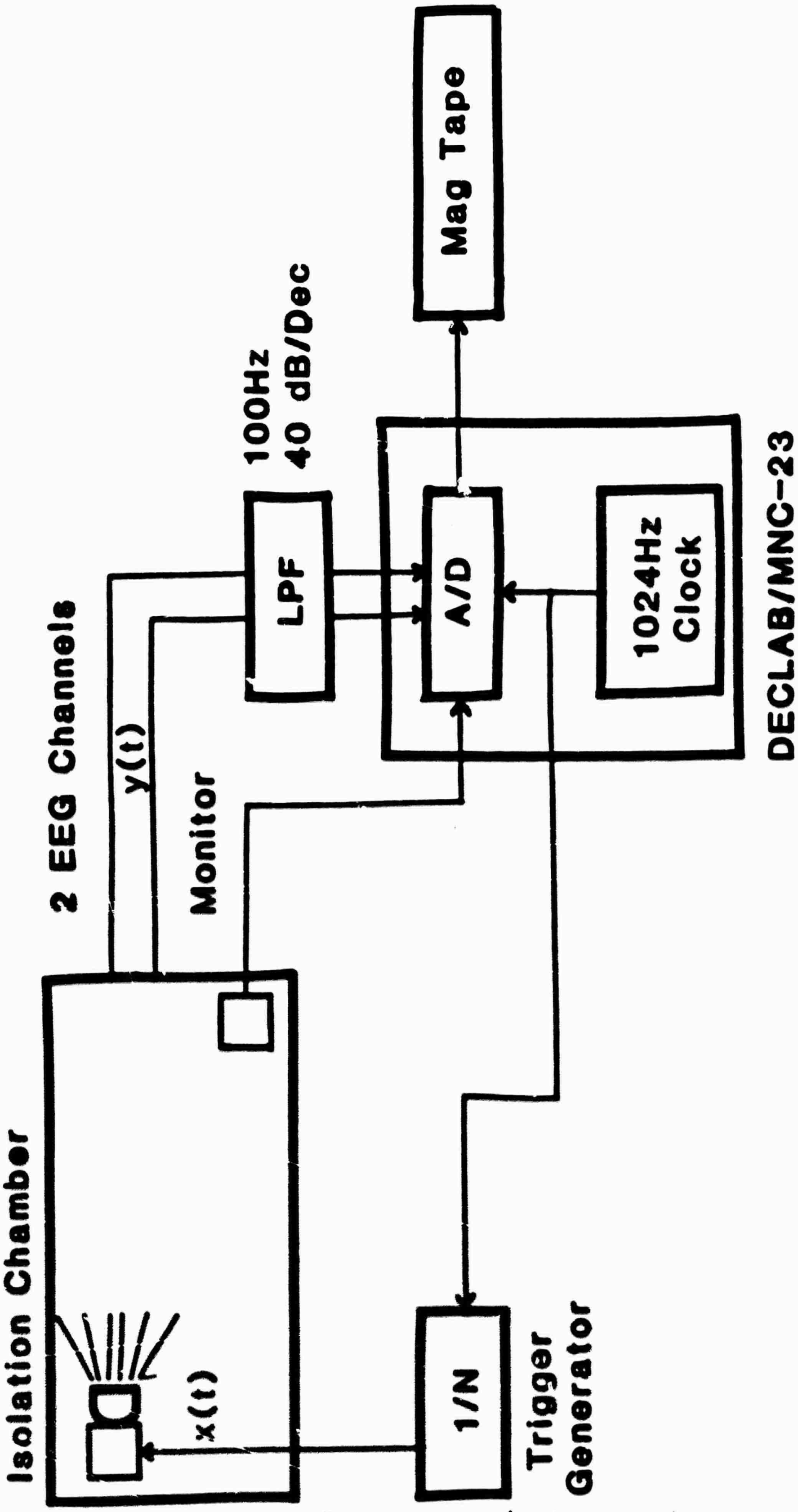
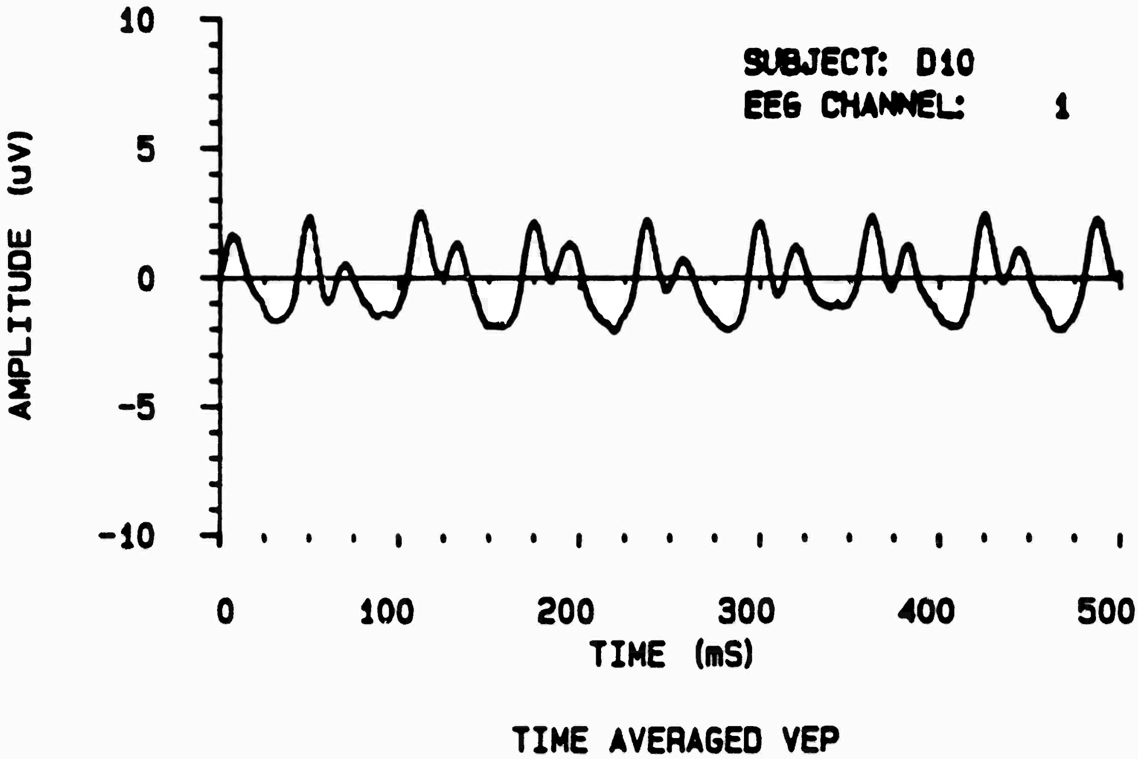


Figure 6. Interconnection of system elements.



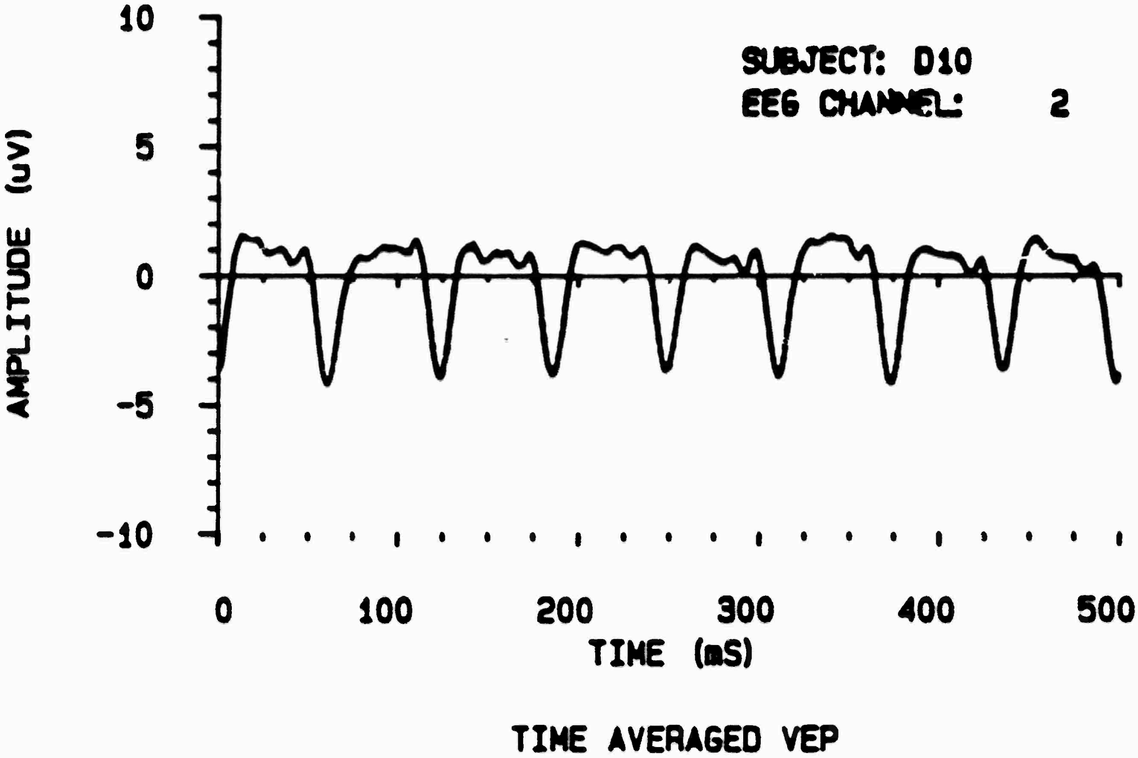


Figure 7. Time average of VEP for one subject, 16 Hz stimulus, showing difference in signals at two electrode sites.

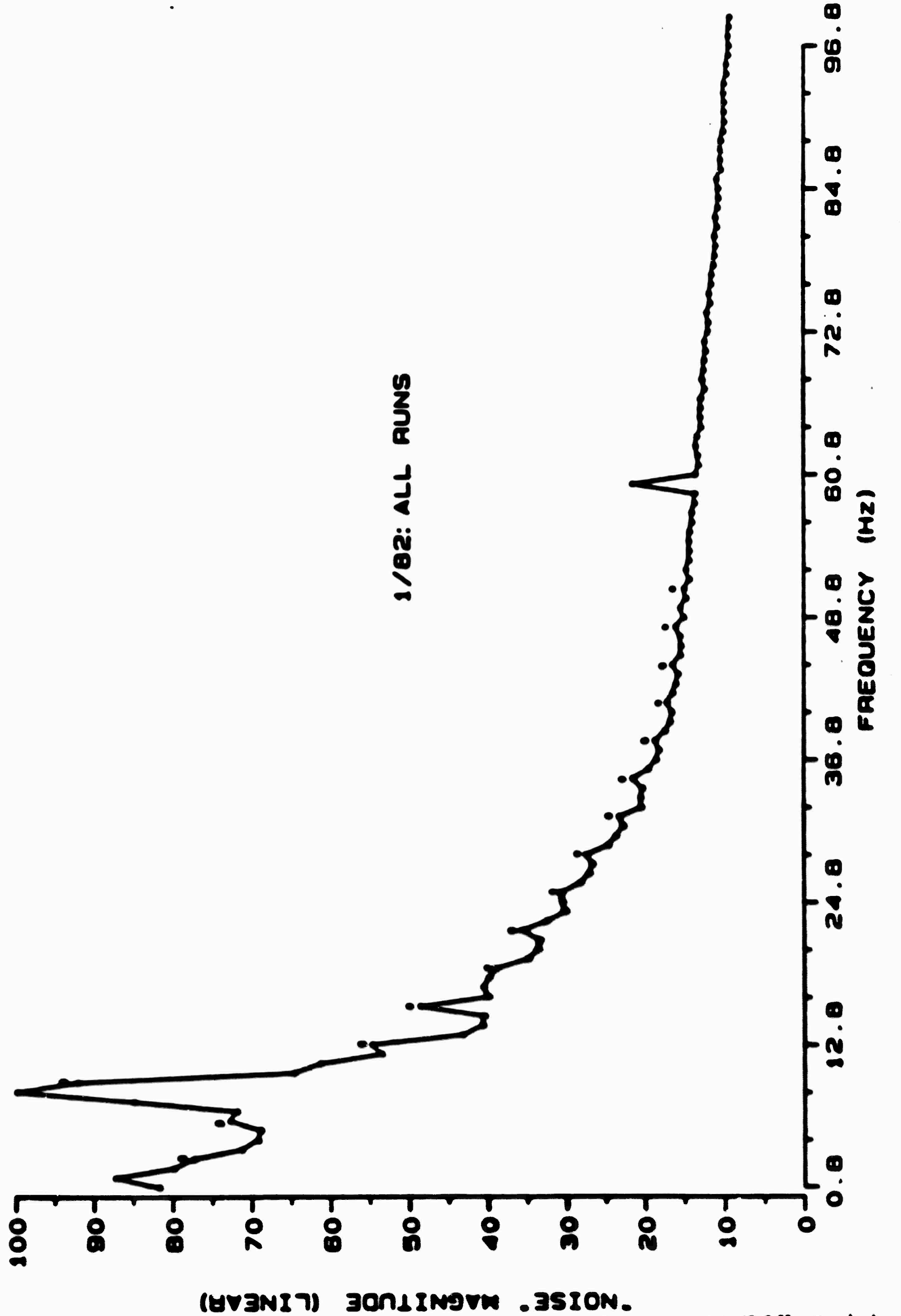
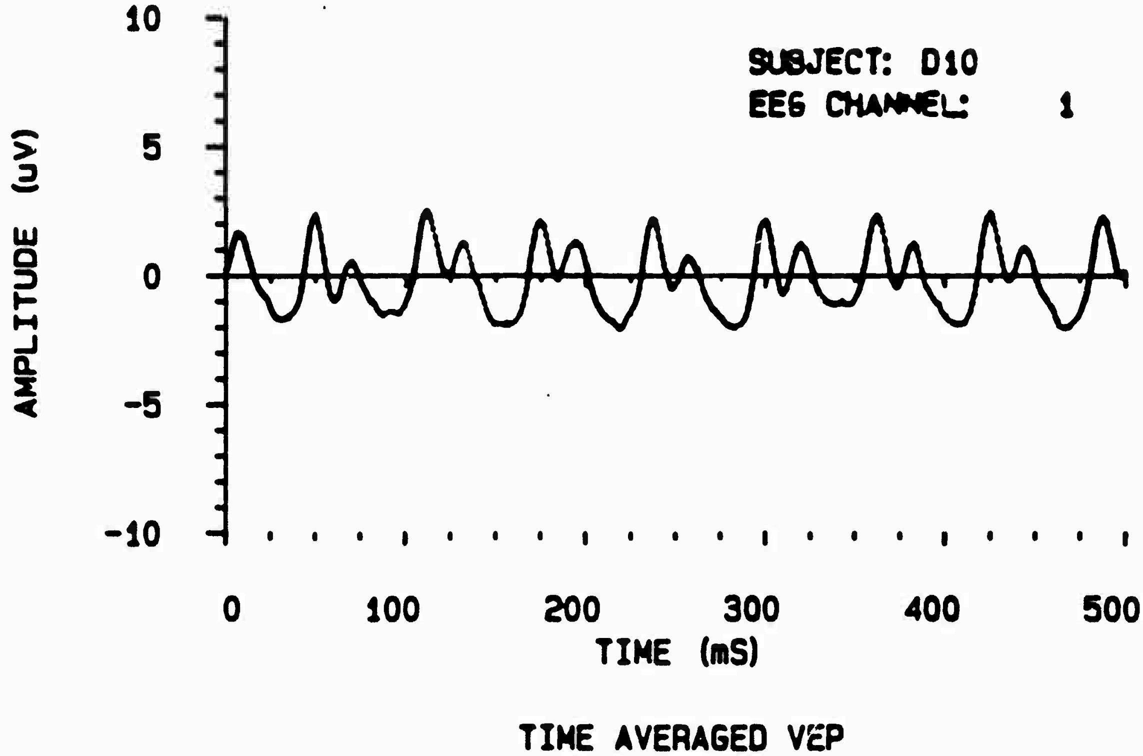


Figure 8. Noise (RMS) residue after removal of signal, with markers at harmonic points (3.2 Hz stimulus).



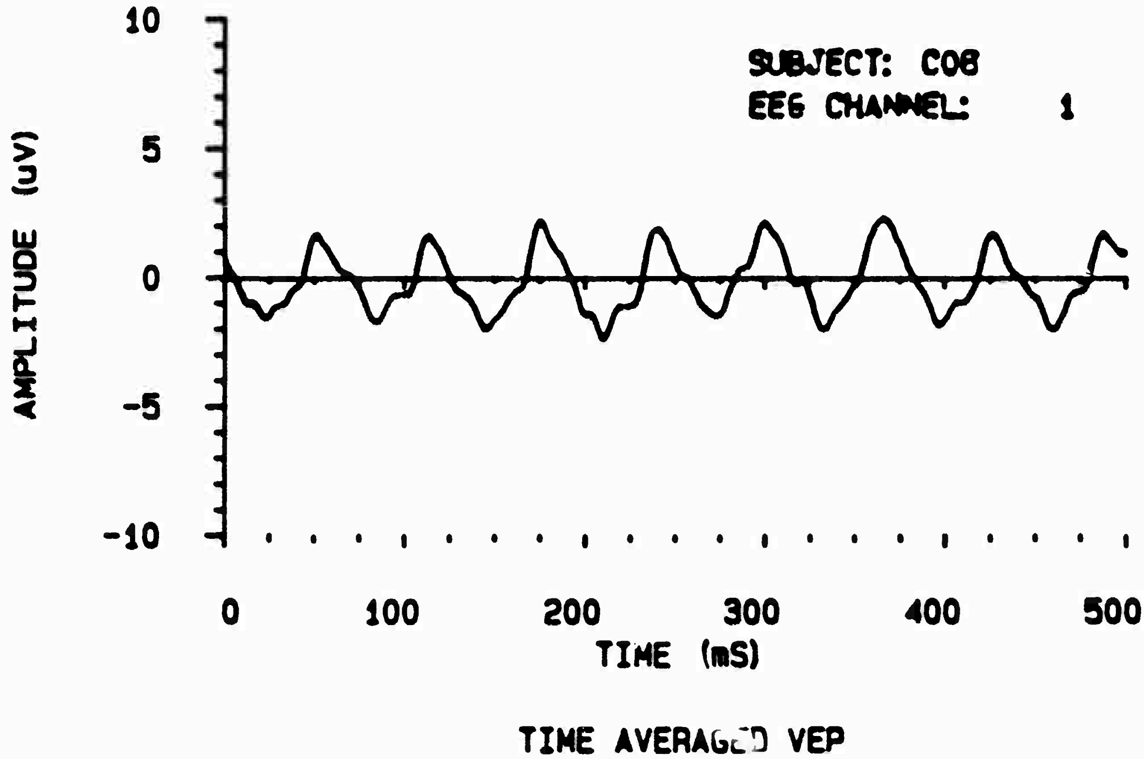


Figure 9. Time averages of VEP for two subjects, same stimulus and electrode sites for both.

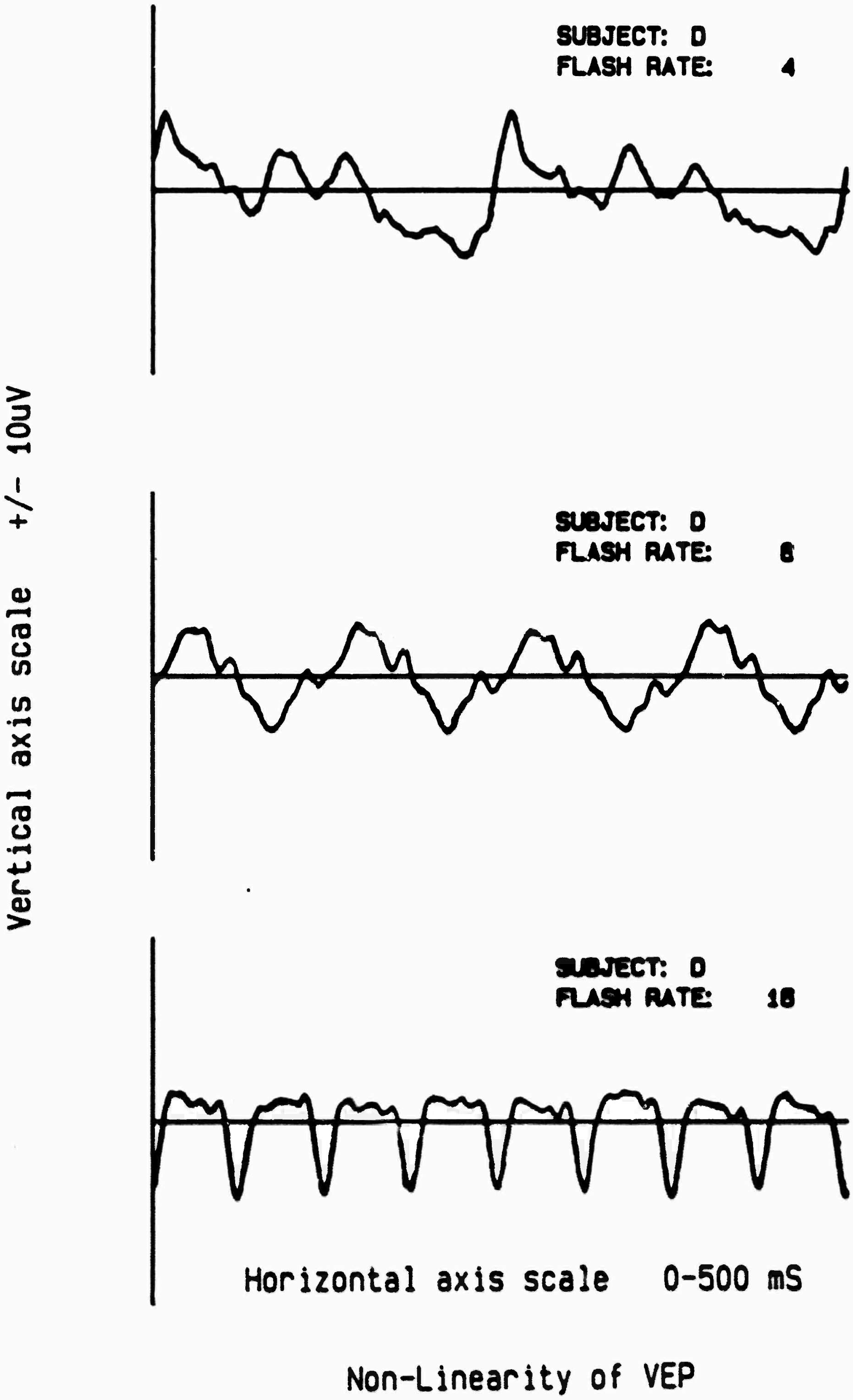


Figure 10. Time averages of VEP for one subject at three flash rates, showing non-linearity of response.

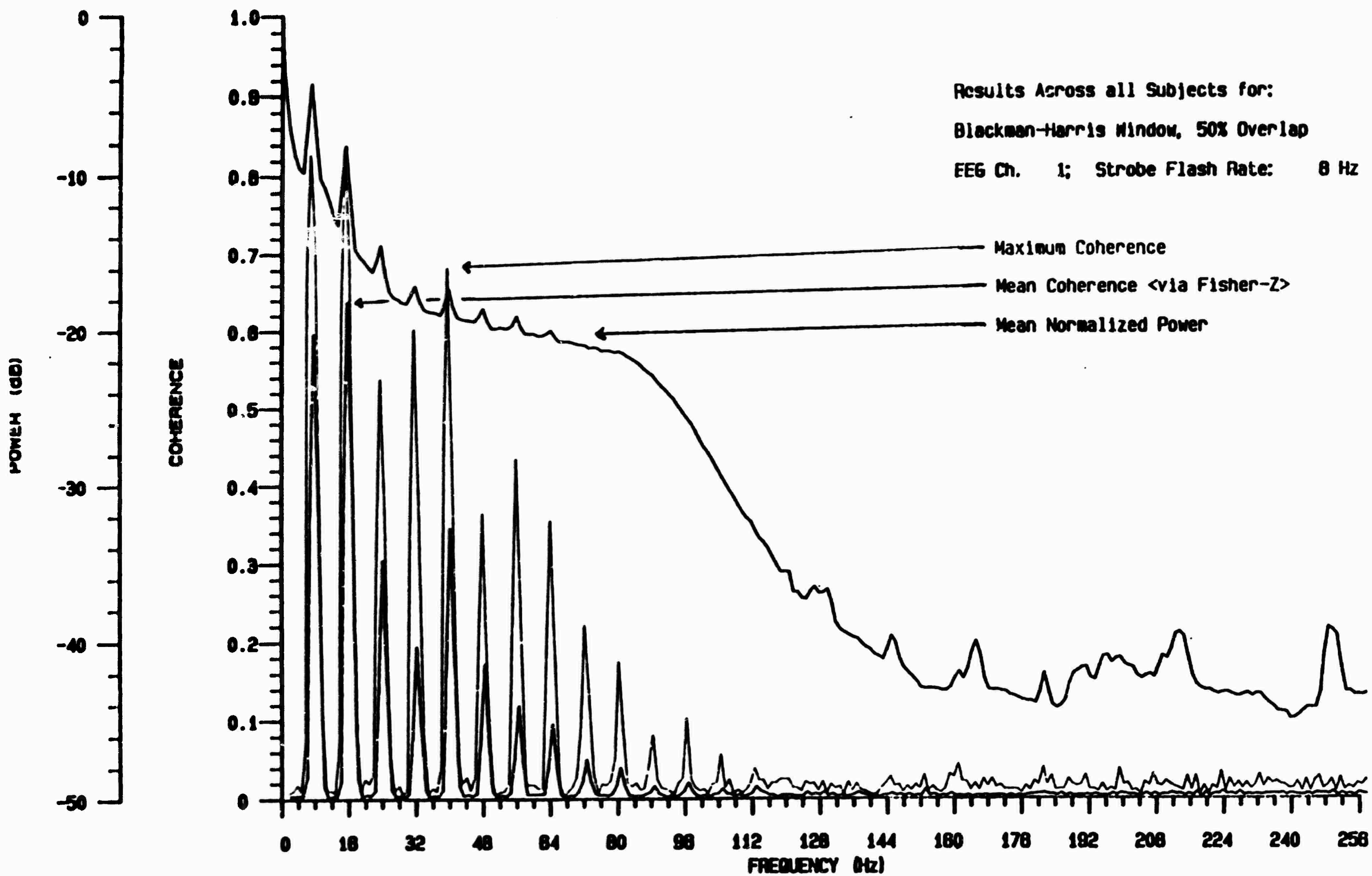


Figure 11. Power, mean coherence, and maximum coherence, for all 8 Hz flash rate runs.

REFERENCES

1. Bendat, J. S., and Piersol, A. G. 1971. *Random data: Analysis and measurement procedures.* New York: Wiley - Interscience.

2. Berkhout, J., O'Donnell, R. D., and Leverett, S. D. Jr. 1973. Changes in electroencephalogram spectra during repeated exposure to $+G_Z$ acceleration. USAF-AMRL-TR-72-123. Aerospace Medical Research Laboratory, Wright-Patterson Air Force Base, Ohio.

3. Burton, R. R., Leverett, S. D. Jr., and Michaelson, E. D. 1974. Man at high sustained $+G_Z$ acceleration: a review. *Aerospace Med.*, 45, 10, 1115-1136.

4. Cohen, M. M. 1981. On the additivity of acceleration protection, in *Aerospace Medical Assoc. Annual Meeting Preprints*, pp. 9-10.

5. Coburn, K. R. 1970. Physiological end-points in acceleration research. *Aerospace Med.*, 41, 1, 5-11.

6. Donchin, E., and Lindsely, D. B. (Eds.) 1969. *Averaged evoked potentials: Methods, results, and evaluations.* NASA-SP-191, Washington, DC: National Aeronautics and Space Administration.

7. Drasdo, N. 1980. Cortical potentials evoked by pattern presentation in the foveal reggion. In Barber, C. (Ed.) *Evoked potentials*, Baltimore: Univ. Park Press, 167-174.

8. Duane, T. 1954. The effect of anoxia. Observations on the fundus oculi during blackout. *A.M.A. Arch. Ophthal.* 51:343.

9. Duane, T. D., Lewis, D. H., Weeks, S. D., and Toole, J. F. 1962. The effects of applied ocular pressure and of positive acceleration on photic driving in man. NADC-MA-6214. Naval Air Developement Center, Warminster, PA.

10. Eisenstein, B. A., and Morgan, E. B. 1982. A dual processor system for the analysis of evoked potentials. *IEEE Frontiers of Computers in Medicine (COMPMED 82)*, 2.2.1-2.2.4.

11. Feinsod, M., Selhorst, J. B., Hoyt, W. F., and Wilson, C. B. 1976. Monitoring optic nerve function during craniotomy. *J. Neurosurg.* 44, Jan, pp 29-31.

12. Guilford, J. P. 1950. *Fundamental statistics in psychology and education.* (2nd ed.), New York: McGraw-Hill.

13. Guilford, J. P. and Fruchter, B. 1979. *Fundamental statistics in psychology and education* (6th ed.), New York, McGraw-Hill.

14. Harris, F. J. 1978. On the use of windows for harmonic analysis wiht the discrete fourier transform. *Proceedings of the IEEE*, 66, -1, pp. 51-83.

15. Jasper, H. H. 1958. The ten-twenty electode system of the International Federation. *Electroenceph. Clin. Neurophysiol. 10*, 371-375.

16. Krutz, R. W., Rositano, S. A., and Mancini, R. E. 1975. Comparison of techniques for measuring G_Z in man. *J. Appl. Physiol.*, 38: 1143-1145.

17. Leverett, S., Kirkland, V. E., Schermerhorn, T. J., and Newson, W. A. 1966. Retinal circulation in man during G_Z blackout. *Proc. Aerospace Med. Ass'n.*

18. McGillem, C. D., and Aunon, J. I. 1983. New techniques for measuring single event related brain potentials. Air Force Office of Scientific Research/NL, AFOSF-TR-82-0901.

19. Nelson, J. G., Palumbo, J. J., and Cohe M. M. 1980. Use of the coherence function in analysis of steady-state visual evoked potential (VEP), in *Research Program Review, Aircrew Physiology*, Naval Air Systems Command Research and Technology Group, pp. 15-22.

20. Nelson, J. G., and Hrebien, L. May 1982. Real-time monitoring of the visual evoked potential, in *Aerospace Medical Assoc. Annual Meeting Preprints*, pp. 169-170. Also issued as NADC-82205-60. Jul 1982, Naval Air Development Center, Warminster, PA 18974.

21. Nelson, J. G., Hrebien, L., and Cohen, M. M. 1982. Progress in real-time monitoring of the visual evoked potential, in *Research Program Review: Aircrew Physiology*, Naval Air Systems Command, Report No. NADC-82232-60, Jun 1982, Naval Air Development Center, Warminster, PA 18974, pp. 15-27.

22. Nelson, J. G., and Hrebien, L. 1982. Real-time detection of steady-state evoked potentials. *Frontiers of Computers in Medicine: COMPMED* 82, pp. 24-27. Also issued as NADC-82206-60. Jul 1982, Naval Air Development Center, Warminster, PA 18974.

23. Nelson, J. G., Hrebien, L. Palumbo, J. J., an Cohen, M. M. 1983. Loss of visual evoked potential as an end-point in acceleration research. *Proceedings of Aerospace Medical Association Meeting*, pp. 77-78.

24. Nelson, J. G., Hrebien, L., and Palumbo, J. J. 1983. Real-time monitoring of the visual evoked potential: progress and plans. In *Research Program Review, Aircrew Physiology*, Naval Air Systems Command, Research and Technology Group, May 1983, pp. 45-58.

25. Nelson, J. G., Hrebien, L., and Cammarota, J. P. 1984. Objective detection of acceleration induced loss of vision. *NAVAIRDEVCEN Technical Review*, Winter 1985.

26. Nelson, J. I. 1984. Assessment of human visual performance with a swept evoked potential technique. Naval Air Development Center, Report No. NADC-84101-60, Warminster, PA 18974.

27. Regan, D. 1972. *Evoked potentials in psychology, sensory physiology and clinical medicine*. London: Chapman and Hall/New York: Wiley.

28. Regan, D. 1977. Steady-state evoked potentials. *J. Opt. Soc. Am.*, 67: 1475.

29. Rositano, S. A., Gillingham, K. K., Sandler, J., and Skrettingland, K. R. 1975. Correlation of eye level blood flow velocity and peripheral ligh loss during $+G_Z$ stress. *Proceedings of Aerospace Medical Association*, pp. 123-124.

30. Sem-Jacobsen, C. W., and Sem-Jacobsen, I. E. 1963. Selection and evaluation of pilots for high performance aircraft and spacecraft by inflight EEG study of stress tolerance. *Aerospace Med.*, 34, 7, 605-609.

31. Spekreijse, H. 1966. *Analysis of EEG responses in man evoked by sine wave modulated light*. The Hague: Dr. W. Junk.

32. Squires, R. D., Jensen, R. E., Sipple, W. C., and Gordon, J. J. 1964. Electroencephalographic changes in human subjects during blackout produced by positive acceleration. NADC-MA-6402, Naval Air Development Center, Warminster, PA.

33. Turin, G. L. 1960. An introduction to matched filters. *IRE Trans. on Info. Theory*, IT6, 3, 311-329.