Weak correlation between the aberration dynamics of the human eye and the cardiopulmonary system

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It is fairly well established that the higher-order aberrations of the eye fluctuate over relatively short time periods, but as yet there is no conclusive evidence regarding the origin of these fluctuations. We measured the aberrations and the pulse pressure wave simultaneously for five subjects. The aberrations were measured by using a Shack–Hartmann sensor sampling at 21.2 Hz. We decomposed the aberration data into Zernike coefficients up to and including fifth order and also calculated the rms wave-front error. From the pulse data the heart rate variability signal was also derived. Coherence function analysis showed that for all subjects there was a weak correlation between many of the aberrations and the pulse and the derived heart rate variability. The pulse and the heart rate variability can account for only 11% ±2% and 20% ±2%, respectively, of the aberration dynamics. © 2005 Optical Society of America

1. INTRODUCTION
The higher-order aberrations of the human eye, i.e., those beyond defocus and astigmatism, have been shown to fluctuate over time.1,2 The first study measured the aberrations at 25 Hz and found that all of the aberrations showed power out to ~6 Hz. The latter measured the aberrations at 240 Hz and found measurable power out to ~30 Hz. In both studies, plotting the log of the power spectrum of the total rms wave-front error against the log of frequency revealed a straight line with a slope of ~1.5. A later study has reported fluctuations as great as 70 Hz.3 The origin of the fluctuations has been under investigation. Microfluctuations in accommodation, eye movements, and the heartbeat have been deemed unlikely.1 The possibility of the heartbeat was excluded mainly because reported changes in the axial length of the eye of 4 µm with the heartbeat4 were too small to account for the magnitude of the defocus changes observed.

There are two reasons, however, not to dismiss the pulse. First, the power spectrum of the pulse pressure wave not only has power at the heart rate but, as will be shown later, has significant power out to ~20 Hz (dependent on the subject). Although this does not cover the full range of frequencies that have been observed for the aberration fluctuations, it does cover a large part. Second, many events in the eye are correlated with the heartbeat aside from changes in axial length. One example is corneal pulsation.5 Although fundus pulsation leads to changes in axial length, it is not just defocus that may change, as the amount of pulsation has been shown to be dependent on retinal location.4 The high-frequency component (HFC) of the microfluctuations in accommodation has also been attributed to the heartbeat.6–9 One study9 also found a correlation between components of the microfluctuations in accommodation and harmonics of the pulse rate. Microsaccades,10,11 a component of pupillary unrest,12 and intraocular pressure changes13 have been correlated with the heartbeat. The amplitude of the intraocular pressure change has been found to decrease with increasing heart rate.14

The heart rate is not constant but fluctuates over time, a phenomenon termed heart rate variability (HRV). This is modulated by three main physiological rhythms: thermoregulation, changes in the blood pressure system, and respiration, with frequencies in the range 0–0.03 Hz, 0.03–0.15 Hz, and 0.18–0.4 Hz, respectively.15 Hence, not surprisingly, pupil size fluctuations16 and one of the low-frequency components in the microfluctuations in accommodation8 have been shown to be correlated with respiration. The HRV power spectrum also displays the 1/f frequency characteristic when plotted on a log–log plot,17 as is found for the aberrations.

Knowing the source of the dynamic behavior affects fields such as high-resolution imaging of the retina with the use of adaptive optics; see, for example, Ref. 18. If their origin were known, then techniques for reducing their magnitude could be developed that would complement the performance of adaptive optics systems, giving improved performance and better images. The aim of this investigation was to determine whether the collective
changes associated with the pulse and the components of HRV are responsible for the inverse frequency power law characteristic of the aberrations.

2. EXPERIMENTAL SETUP

The aberrations were measured by using a Shack–Hartmann sensor. The experimental setup is shown in Fig. 1. The eye is illuminated by a laser diode operating at 817 nm, which passes through a Badal optometer arrangement consisting of two plane mirrors, PM1 and PM2, on a motorized stage. The beam then passes onto a scanner, which is conjugated with the eye's pupil by lenses L4 and L5. The beam is scanned at ~200 Hz with a scan angle of ~1° to reduce speckle noise. We found that 200 Hz resulted in sufficient homogeneity of the Shack–Hartmann spots. The beam enters the eye off axis, and so corneal reflections are blocked by the spatial filter, SF1. Directly in front of the eye are two rotating cylinders that form a toric to aid in the correction of astigmatism. It is not desirable for the target light to pass via the scanner. The scanner is easily bypassed by using two cold mirrors, CM1 and CM2, at 45° to the beam. These reflect visible light but pass infrared. The light from the fixation target enters the system via a cold mirror, CM3.

The lenslet array of the Shack–Hartmann sensor is conjugated with the eye's pupil via lenses L6 and L9. Taking into account magnification changes through the system, the lenslet array samples the pupil at 0.6-mm intervals. The focal length of the array is 7.5 mm. The CCD camera is a Retiga Ex operating at 21.2 Hz.

A. Spherocylindrical Correction

Shack–Hartmann sensors have a limited dynamic range that is often exceeded by an astigmatic subject. Current Shack–Hartmann sensors for use in the eye correct sphere via a Badal optometer arrangement and apply cylinder with trial lenses; see, for example, Ref. 1. A novel feature of the system is the automated correction of astigmatism by means of a pair of rotating cylinders in combination with a Badal optometer as shown in Fig. 2. Although it is well known that rotating two cylinders produces a variable toric, to the authors' knowledge this is the first time a method to incorporate them into a Shack–Hartmann sensor to correct astigmatism in an automated fashion has been devised. Note that although the stage adjusts the position of the focal point relative to L3, L5 is the Badal lens, as it is the position of the focused spot relative to this lens that is important.

Fig. 1. Experimental setup for aberration measurements. PM, plane mirror; CM, cold mirror; BS, beam splitter; L, lens (focal length is in millimeters); SF, spatial filter.
This system is characterized by the following equations:

\[ P_{\text{cres}} = 2P_c \cos(\theta_2 - \theta_1), \]  

\[ \phi = \frac{\theta_2 + \theta_1}{2}, \]  

\[ P_{\text{sres}} = -z \frac{P_c^2 + 2P_c \sin^2 \left( \frac{\theta_2 - \theta_1}{2} \right)}{2}, \]

where \( P_{\text{cres}} \) and \( \phi \) are the resulting cylindrical power and axis, \( \theta_1 \) and \( \theta_2 \) are the orientations of each cylinder, \( P_c \) is the power of each cylinder, \( P_{\text{sres}} \) is the resultant spherical power, and \( F_s \) is the power of the Badal lens. \( z \) is the shift in the position of the focus before the Badal lens due to the movement of the stage. The cylinders each have a power of \(-3 \) D, and the Badal lens has a power of \(8 \) D. The range of correction is \(\pm6 \) D spherical and \(0\) to \(-6 \) D cylindrical.

The change in the wave front is not linear with the parameters \( \theta_1 \) and \( \theta_2 \). Suitable linear control parameters are obtained by decomposing the effect of the two cylinders into two orthogonal astigmatic terms given by

\[ a = P_{\text{cres}} \cos 2\phi, \]  

\[ b = P_{\text{cres}} \sin 2\phi. \]

The control algorithm used is a simple integrator and is given by

\[ x_{t+1} = gCs + x_t, \]

where \( x \) is a vector containing the coefficients \( a, b, \) and \( z \); \( s \) is a vector containing the Shack–Hartmann slopes; and \( C \) is the control matrix. \( C \) is calculated as a pseudoinverse of the system interaction matrix, which was measured experimentally. The gain \( g \) is normally set to 0.9, and typically 3–5 iterations are sufficient to reduce the residual sphere and cylinder to a few tenths of a diopter. \( P_{\text{cres}} \) and \( \phi \) and hence the cylinder rotations \( \theta_1 \) and \( \theta_2 \) are calculated as the inverse transformation of Eqs. (4) and (5):

\[ \theta_1 + \theta_2 = \arctan \frac{b}{a}, \]  

\[ \theta_1 - \theta_2 = \arccos \frac{P_{\text{cres}}}{2P_c}. \]

If, however, the refractive error of the subject across any meridian is greater than approximately \( \pm2 \) D, the dynamic range of the sensor is exceeded, and so the system is initially set up to correct for an estimate of the full prescription or only the spherical component of the subject and then is refined by using Eq. (6).

The pulse pressure wave was measured by using a commercial device from Contact Precision Instruments. A finger clip measures the pulse pressure by means of photoplethysmography.

### 3. EXPERIMENTAL PROCEDURE

Five subjects were used. Each subject viewed a high-contrast letter at optical infinity with the left eye except for subject SG, who viewed with the right eye out of preference. The other eye was occluded. Accommodation was not paralyzed. Each subject was an experienced observer with good fixation stability. Subjects KH and SG were astigmatic, the prescription of KH being \(-1.00 \) DS and \(-1.25 \) DC and for SG \(-2.00 \) DS and \(-4.00 \) DC. The sphere and cylinder were corrected for these two subjects by the rotating cylinders and the Badal optometer arrangement as discussed previously. Each subject was attached to a bite bar to minimize movement. The wave front was measured at 21.2 Hz over a time period of 71 s and eye pupil size of 4.2 mm. The aberration coefficients up to and including fifth order (excluding piston) and rms wave-front error (excluding piston, tip, and tilt) were calculated for each frame. The Zernike basis convention recommended by the OSA/VSIA Taskforce was used.

The pulse pressure wave was measured simultaneously at 100 Hz. This signal was interpolated by using a cubic spline function and sampled at regular intervals with a time spacing equal to the estimated time between each two Shack–Hartmann measurements. The HRV signal was derived by calculating the time between successive peaks of the pulse pressure wave and converting it to an instantaneous pulse rate in beats per minute. Again this signal was interpolated and sampled at regular intervals corresponding to the sampling of the Shack–Hartmann measurements.

To synchronize the Shack–Hartmann measurements with the pulse measurements a digital signal was sent to the pulse measurement device at the start and end of the aberration data acquisition.

### 4. SPECTRAL ANALYSIS

For each of the signals (pulse, HRV, aberration coefficients, and the total rms wave-front error), the power spectral density function (PSD) was calculated. For a signal \( x \) sampled at a frequency \( F_s \) over a time period \( T \), giving \( N \) data points, the PSD is given by

\[ G_{xx}(f) = \frac{2}{NF_s} |X(f,T)|^2, \]

where \( X(f,T) \) is the discrete Fourier transform of \( x \). The 99% confidence intervals were also calculated as given by
\[
\frac{nG_{xx}}{\chi^2_{n,0.005}} \leq G_{xx} \leq \frac{nG_{xx}}{\chi^2_{n,0.995}},
\]

where \( n \) is the degrees of freedom and \( \chi^2_{n,0.005} \) and \( \chi^2_{n,0.995} \) are the 0.5 and 99.5 percentage points of a \( \chi^2 \) distribution with \( n \) degrees of freedom (dof). The dof is equal to twice the number of records over which the average of \( G_{xx} \) has been formed. In the case of the raw pulse data, rms wavefront error, and each aberration coefficient, the time signals were split into ten sections. Hence the resolution of the PSD was 0.14 Hz and the dof was 20. In the case of the HRV data the time signal was split into five sections (i.e., ten dof) to achieve a higher resolution of 0.07 Hz to better resolve its components. Taking into account the sampling frequency of 21.2 Hz, data could be acquired for frequencies in the range 0.14–11.1 Hz for the pulse and aberration data and 0.07–11.1 Hz for the HRV. Each time signal was extracted by using a Hanning window to reduce spectral leakage. Hence \( G_{xx} \) was calculated by using the Welch method.\(^{21}\)

**A. Significance Testing of Spectral Peaks**

Significance testing is important for two reasons. First, it is important to know whether a peak at a particular frequency is a characteristic of the parameter being measured or is a result of noise in the measurement device. The Shack–Hartmann measurements, for example, are particularly susceptible to readout noise. Second, it is important to know whether an increase in the height of a peak between two measurement conditions is a result of the different conditions or merely a result of the statistical properties of the estimator. If the confidence intervals of the two signals in question overlap, then they are not significantly different from each other.

The Shack–Hartmann noise signal was measured by using an artificial eye consisting of a lens and a piece of card as the retina. The artificial eye is not subject to any dynamics, and so the resulting PSD represents noise in the system. The source power was adjusted such that the amount of light coming back from the artificial eye was approximately the same as that for a real eye for the same exposure. Scanning was also employed. The data were processed in the same way as the results from a real eye. Owing to the scanning, the homogeneity of the Shack–Hartmann spots from the real eyes was comparable to that from the artificial eye.

It is possible that there was a small residual amount of speckle noise in the real-eye measurements; however, as the frequency of the scanner was high in comparison with the sampling rate and the spots appeared to be homogeneous, the authors believe that the speckle noise had a negligible effect on the PSDs.

Figure 3 shows the rms wave-front error PSD for the artificial eye and each subject. The lower 99% confidence interval for GK has been plotted, as this subject had the lowest rms in general, and the upper 99% confidence interval for the artificial eye is also shown. It can be seen that for each subject the power across the full measurable frequency range is significant. The average slope among the five subjects was 1.58±0.16 \( \pm 1 \) S.D., which is in agreement with the findings of other studies.\(^{1,2}\) For frequencies up to \( \sim 2 \) Hz the noise PSD is similar in shape to the PSDs of the aberrations for the real eyes. As the difference between the noise PSD and PSDs from the real eyes is approximately 2–3 orders of magnitude in this frequency range, the noise would have had a negligible influence on the shape of the real-eye plots.

For the pulse measurements, the noise signal was the signal obtained when there was no subject. Figure 4 shows the PSD for the pulse noise and pulse for subject KH, who in general had the lowest PSD, and subject GK. The other subjects have been omitted for clarity. There are two main things to note. The first is the striking presence of harmonics of the main pulse rate peak. The second is that the power is not exclusive to these peaks but extends across the full measurable frequency range. This was true for all subjects. Using the pulse data without the interpolation (i.e., 100 Hz sampling rate), revealed significant power out to \( \sim 20 \) Hz for KH and slightly beyond that for the other subjects.
B. Stationarity Testing

Averaging the PSD over segments relies on the assumption that the data are stationary, i.e., their statistical properties are time invariant. Stationarity was tested by means of the reverse-arrangements test, which is a non-parametric trend test. Given $N$ measurements of a variable $x$, the number of reverse arrangements $A$ is given by

$$ A = \sum_{i=1}^{N-1} A_i, $$

$$ A_i = \sum_{j=i+1}^{N} h_{ij}, $$

$$ h_{ij} = \begin{cases} 1 & \text{if } x_i > x_j \\ 0 & \text{otherwise} \end{cases}. $$

If $A$ differs from the expected value for $N$ independent observations of the same random variable, then there is evidence of nonstationarity. Each full-length time-course signal for the pulse, HRV, and rms wave-front error was broken down into ten segments for which the PSDs were calculated. The number of reverse arrangements was then calculated for each frequency for each signal. Stationarity implies time invariance of the autocorrelation function. As spectral density functions and correlation functions are related to each other by a Fourier transform, time invariance of one implies time invariance of the other. For ten segments the upper and lower 99% confidence interval limits are 35 and 9, respectively.

The results for subject GK are shown in Fig. 5. For this subject only 1% of the rms wave-front error frequencies and 1% of the pulse frequencies fall beyond the limits. The results were similar for the other subjects. Hence in the main the data are stationary and the averaging over segments is valid. To the authors’ knowledge this is the first time the aberrations of the eye have been tested for stationarity.

C. Coherence Function

The coherence function is a tool to investigate the synergy between two systems in frequency space. It has been used routinely in investigating the correlation between the pulse and other physiological processes including accommodation, miniature eye movements, and thermoregulation. The coherence function is given by

$$ \gamma^2_{xy}(f) = \frac{|G_{xy}(f)|^2}{G_{xx}(f)G_{yy}(f)}, $$

where $G_{xx}$ and $G_{yy}$ are the PSD functions, $G_{xy}$ is the cross-spectral density function, and $\gamma^2_{xy}$ varies between 0 and 1. A value close to 1 indicates a correlation at the frequency in question. As $G_{xy}$ is twice the Fourier transform of the cross-correlation function, the coherence function is essentially a plot of the correlation coefficients over the frequency range of the data. The phase of $G_{xy}$ gives the phase difference between the two signals at the frequency in question. The 95% confidence intervals were calculated with the method proposed by Wang and Tang. As with $G_{xx}$ and $G_{yy}$, the Welch method was applied in the formation of $G_{xy}$.

D. Removal of Blink Artifacts

Owing to the relatively long recording time, eye blinks are inevitable. These cause abrupt changes in the aberration measurements, which can lead to an increase in power across a range of frequencies, and so the eye blinks are effectively a source of noise and must be removed. To locate the blinks we considered the accommodation record (i.e., Zernike mode 4) of each subject. The average time of a blink is 250 ms. We deleted the six data points concerned with the blink, which corresponded to a time span of 284 ms as a result of our sampling time. A cubic spline function was then used to interpolate between the points before and after the blink. This procedure was then repeated for the same points in each coefficient time course.

5. RESULTS

Figure 6 is an example of the pulse data obtained for subject GK, showing the time evolution of the pulse pressure wave, the PSD of the pulse pressure wave, and the time course of the HRV signal and its PSD. The main components of the HRV PSD are labeled. Owing to the signal length, the band nominally associated with thermoregulation has not been resolved. The blood pressure and respiratory peaks are evident, however.

The coherence function between the rms wave-front error and pulse pressure wave is shown in Fig. 7 for subjects SG and JC. The 95% confidence intervals of the coherence function and the normalized pulse PSDs have also been plotted. The results were similar for the other three subjects. The average coherence function value across subjects at the main pulse rate peak for each subject is $0.1 \pm 0.05$ (±1 S.D.). The upper confidence limit gives an idea of the maximum correlation. For a coherence function value of 0.1 the upper 95% limit is 0.37, suggesting that the correlation is weak.

The results for the HRV and rms wave-front error for subjects SG and JB are shown in Fig. 8. The average coherence function values at the blood pressure and respiratory peaks across subjects are $0.1 \pm 0.07$ (±1 S.D.) and $0.23 \pm 0.08$ (±1 S.D.), respectively.
The average coherence function value across all frequencies across all subjects for the pulse and rms wave-front error and HRV and rms wave-front error are $0.11 \pm 0.02 \ (\pm 1 \text{ S.D.})$ and $0.20 \pm 0.02 \ (\pm 1 \text{ S.D.})$, respectively, again suggesting a weak correlation.

Figure 9 shows the coherence function between the first 12 Zernike coefficients and the pulse PSD for one subject, KH. The results for the other four subjects were similar. Again the values are low for each subject. The confidence intervals have been omitted for clarity.

There are additional possible sources of noise. One might be movement of the subject relative to the system. This was assessed in two ways. The first method made use of the phenomenon of eye retraction. It has been shown that when one eye blinks, the fellow eye retracts by 0.5–1.5 mm.\textsuperscript{24} Subject JC was instructed to blink his

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**Fig. 6.** All pulse data for subject GK. A, time evolution of the pulse pressure wave; B, derived HRV time course; C, pulse PSD; D, HRV PSD.

**Fig. 7.** Coherence function for the pulse pressure wave and rms wave-front error for two subjects.
right eye every 5 s (i.e., at 0.2 Hz) while measurements were made on his left. If this caused any of the aberrations to change, then there would be an expected significant increase in the rms wave-front error at this frequency. The results revealed that there were no significant changes in the rms or in each individual aberration coefficient at the blink frequency.

Second, measurements were taken on an artificial eye with and without subject JB biting on a bite bar attached to the artificial eye mount. For some frequencies there were significant differences in the rms wave-front error for the artificial eye when the subject was attached to the mount. However, even though bite bar instabilities appear to have an effect, the resulting power was nearly 3 orders of magnitude smaller than the rms wave-front error PSD for this subject. Hence the overall effect of instabilities in the bite bar on this subject’s and other subjects’ rms wave-front error is likely to be negligible.

6. DISCUSSION
The coherence function values between the rms wave-front error and the pulse were low for each subject, generally lying in the range 0–0.4. When we considered the HRV and the rms wave-front error as well as the pulse and each individual aberration, again the coherence val-
ues were low. When low values are observed there are three possible reasons. One is extraneous noise in the measurements. We believe this to be unlikely because each signal was tested against its noise signal, which revealed that all PSD values were significantly different from noise. The second possibility is that the mechanism relating the two signals is not linear: One can envisage that there may be nonlinear limits to the deformation of the eye.

The third possibility is that the aberrations are not affected by the pulse alone but there is some other input. We think that this is the most likely possibility. For example, only a component of pupil size fluctuations is correlated with the pulse, and hence there is another input in this mechanism that may also be affecting the aberrations. One way the coherence function values can be interpreted is to say that the value at a given frequency is a measure of the proportion of one signal that is due to the other. Hence one can say that the dynamics of the pulse and HRV account for 11% ±2% and 20% ±2%, respectively, of the rms wave-front error dynamics.

The lack of high coherence is perhaps not surprising when one is considering the shapes of the PSDs. Figure 10 shows the pulse, HRV, and rms wave-front error PSDs for one subject. The plots have been displaced for clarity. It can be seen that the slopes are noticeably different. The HRV PSD is considerably steeper than that of the rms. The pulse PSD is also slightly curved. The shapes of the pulse and HRV PSDs were similar for all subjects.

It must be pointed out that a nonzero coherence function value does not imply a causal relationship between the aberrations and the pulse. A further experiment was carried out on one subject to see whether significant changes in the HRV and pulse PSD produced significant changes in the PSD of the rms wave-front error. Subject JC was instructed to breath deeply at a constant rate of 0.1 Hz (one breath per 10 s). Comparison of this case with the normal-breathing case revealed many significant changes in the pulse and HRV, as expected. Significant we mean that the two PSD signals differed by more than their 99% confidence intervals. Nevertheless, there were no significant changes in the rms wave-front error. This further demonstrates the “weakness” of the influence of the pulse for this subject.

A. Comparison with Other Studies

As discussed in the Introduction, numerous investigations have shown evidence of a correlation between the HFC of accommodation fluctuations and the pulse rate. Defocus is represented by the term $Z_4$. The coherence-function value for defocus at the main heart rate peak was only 0.1 on average for the subjects used in this study. When we compare the PSD of defocus and the pulse there is a peak in the accommodation spectra that roughly corresponds to the heart rate for some subjects. However, the peaks are slightly displaced in frequency, resulting in a low coherence-function value. Hence the findings here do not support the findings of these studies.

The common element in the investigations mentioned above is that accommodation was measured with an objective infrared optometer. The basic principle is that the eye is illuminated by an infrared source and then its vergence is measured after reflection from the retina and refraction by the cornea and lens. The advantage of the Shack–Hartmann sensor over this type of optometer, however, is that it can distinguish between different aberration terms. With the objective infrared optometer, it is possible that changes in other aberration terms that may have been correlated with the pulse, such as spherical aberration, were interpreted as fluctuations in accommodation. Fluctuations in astigmatism, for example, are evident during steady-state viewing. However, as the coherence function for all aberrations is low, this cannot explain the difference.

Another consideration is the sensitivity of the Shack–Hartmann sensor used. The amplitude of the HFC when the eye is focused for infinity is ±0.01 D. For the pupil size of 4.2 mm used in this study, 0.01 D is ~0.006 μm. Figure 11 shows the PSD of the defocus noise and the PSD for a theoretical 1-Hz sinusoidal fluctuation with an amplitude of 0.006 μm. As can be seen, the power in the theoretical signal is nearly two orders of magnitude above the noise. Hence we believe it is unlikely that sensitivity was an issue.
Several authors have investigated the changes in the HFC with pupil size. Campbell et al. found that for one of their subjects the HFC disappeared when the pupil size was small.\textsuperscript{27} Stark and Atchison, however, found that the HFC for one of their subjects became larger with decreasing pupil size.\textsuperscript{28} These researchers suggest that for reasons unknown, the accommodation system occasionally becomes unstable and exhibits these HFC fluctuations. Hofer et al.\textsuperscript{1} comment in their study on the dynamics of the aberrations that the spectral signature of the pulse rate was seen only occasionally; however, it is unclear whether they measured the pulse rate of their subjects. This finding suggests nonstationarity. Hence in the study here, assuming that the effect of the pulse is to cause occasional instability, then averaging over records would tend to average this peak out.

Figure 12 shows the normalized pulse PSD and normalized defocus PSD for subject KH for each individual data segment. In no case is there a peak in the PSD of defocus corresponding to the main heart rate peak. Furthermore the reverse-arrangements test indicated that the data obtained were stationary. Hence it is unlikely that the low coherence-function value at the pulse rate peak of each subject was due to nonstationarity and the averaging out of peaks.

The stationarity of the aberrations was tested on a signal with a limited length of 71 s. Hence the results from this study cannot rule out the fact that the aberrations may not be stationary over longer periods of time.

With respect to the rms wave-front error, the average slope of $1.58 \pm 0.16$ is in agreement with the results of other studies.\textsuperscript{1,2}

B. Other Possible Contributors to the Aberration Dynamics

The use of eye retraction to assess eye movement relative to the system revealed no significant changes in the aberrations at the blink frequency. This suggests that the system can tolerate movements relative to the system of a few tenths of a millimeter. When assessing bite bar instabilities by measuring the aberration dynamics of an artificial eye with and without a subject on a bite bar attached to the artificial eye mount, the results suggest that even with a bite bar there are some movement artifacts. However, these were extremely small in comparison with the subject's rms wave-front error, and so it is unlikely that they have a significant effect. Hence movement artifacts did not contribute to the shape of the rms wave-front error PSD.

In considering what may be contributing to the dynamics of the rms wave-front error, an obvious line of thought is what other mechanisms in the body display the $1/f$ power law behavior. One is the modulation of action-
potential impulses. Hence it could be that the effect of these impulses on the behavior of the muscles in the eye causes the 1/f behavior of the aberrations. Other contributing factors could be the aqueous flow or tear-film fluctuations.

7. CONCLUSION

In summary, by simultaneously measuring the pulse-pressure wave and aberrations, we have shown that the pulse and the derived HRV share frequency components in common with the aberrations. The low coherence-function values indicate that the correlation is weak and that the pulse is not the only contributor to the power law nature of the rms wave-front error but that there are other contributors yet to be determined.

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