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Contributions of Optical and Non-Optical Blur to Variation in Visual Acuity

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Abstract

Purpose—To determine the relative contributions of optical and non-optical sources of intrinsic blur to variations in visual acuity (VA) among normally sighted subjects.

Methods—Best-corrected VA of sixteen normally sighted subjects was measured using briefly presented (59 ms) tumbling E optotypes that were either unblurred or blurred through convolution with Gaussian functions of different widths. A standard model of intrinsic blur was used to estimate each subject's equivalent intrinsic blur (σ_{int}) and VA for the unblurred tumbling E (MAR_0). For 14 subjects, a radially averaged optical point spread function due to higher-order aberrations was derived by Shack-Hartmann aberrometry and fit with a Gaussian function. The standard deviation of the best-fit Gaussian function defined optical blur (σ_{opt}). An index of non-optical blur (η) was defined as: $1 - \sigma_{opt}/\sigma_{int}$. A control experiment was conducted on 5 subjects to evaluate the effect of stimulus duration on MAR_0 and σ_{int} .

Results—Log MAR_0 for the briefly presented E was correlated significantly with log σ_{int} ($r = 0.95$, $p < 0.01$), consistent with previous work. However, log MAR_0 was not correlated significantly with log σ_{opt} ($r = 0.46$, $p = 0.11$). For subjects with log MAR_0 equivalent to approximately 20/20 or better, log MAR_0 was independent of log η , whereas for subjects with larger log MAR_0 values, log MAR_0 was proportional to log η . The control experiment showed a statistically significant effect of stimulus duration on log MAR_0 ($p < 0.01$) but a non-significant effect on σ_{int} ($p = 0.13$).

Conclusions—The relative contributions of optical and non-optical blur to VA varied among the subjects, and were related to the subject's VA. Evaluating optical and non-optical blur may be useful for predicting changes in VA following procedures that improve the optics of the eye in patients with both optical and non-optical sources of VA loss.

Keywords

visual acuity; Gaussian blur; intrinsic blur; optical blur

Clinical measures of best-corrected visual acuity (VA) can vary substantially among normally sighted individuals,¹ for reasons that have not been fully resolved. A potential factor that may underlie the variation is inter-subject differences in equivalent intrinsic blur, which is an estimate of the intrinsic blur of the visual system. Although the intrinsic blur of the visual system cannot be measured directly, its value can be estimated using the equivalent intrinsic blur paradigm.^{2–6} Under this paradigm, VA is measured for unblurred

targets and for targets blurred by convolution with low-pass Gaussian filters of different standard deviations (σ_{stim}). The use of low-pass Gaussian filters is based on the assumption that intrinsic blur is Gaussian in nature. According to the standard model,⁵ VA is related to σ_{stim} by the relationship:

$$VA = k \sqrt{\sigma_{stim}^2 + \sigma_{int}^2}, \quad (1)$$

where k is a multiplicative constant and σ_{int} (equivalent intrinsic blur) is the amount of σ_{stim} required to reduce VA by $\sqrt{2}$. When σ_{stim} is considerably less than σ_{int} , VA is independent of σ_{stim} and the intrinsic blur of the visual system governs performance. However, when σ_{stim} greatly exceeds σ_{int} , VA is proportional to stimulus blur. Using the equivalent intrinsic blur paradigm, it has been shown previously that normal variation in Vernier acuity,² two-line resolution thresholds,^{3,4} and Landolt C acuity⁶ is related to equivalent intrinsic blur.

As noted previously,² both optical and non-optical sources of blur contribute to σ_{int} . Optical blur refers to blur arising from higher-order aberrations in individuals with optically corrected lower-order aberrations. Non-optical blur refers to the blur that remains once the contributions of optical blur to σ_{int} are accounted for. Non-optical blur can be attributed to neural filtering throughout the visual system.

It is typically assumed that the effects of optical blur on VA are negligible and that individual differences in VA are due to differences in the amount of non-optical blur.⁶ Consistent with this notion, Villegas et al⁷ have shown that VA is not correlated with higher-order aberrations in normally sighted individuals who have optically corrected lower-order aberrations. However, blur generated by the optics of the eye must play a role in determining VA, because VA can be improved by minimizing higher-order aberrations through the use of adaptive optics techniques.^{8,9} Optical blur, non-optical blur, and VA have not been measured in the same subjects. Thus, the contributions of optical and non-optical blur to inter-individual variations in VA are presently unclear.

The purpose of the present study was to determine the relative contributions of optical and non-optical sources of intrinsic blur to variations in VA among normally sighted subjects. An equivalent intrinsic blur paradigm was used to measure VA and σ_{int} for a briefly presented tumbling E optotype. Although the brief stimulus duration minimizes the potentially confounding effect of eye movements, VA is known to improve with increasing stimulus duration.^{10,11} Similarly, estimates of σ_{int} may also be dependent on stimulus duration. Thus, in the present study, VA and σ_{int} were also measured in a subset of subjects at a longer duration (590 ms) for which temporal integration should be nearly complete.¹⁰ To determine the contribution of optical blur to variations in VA among normally sighted subjects, an estimate of optical blur was derived from the optical point spread function (PSF) measured using Shack-Hartmann aberrometry. A Gaussian function was fit to the PSF and the standard deviation of the best-fit Gaussian function defined optical blur (σ_{opt}). Because non-optical blur cannot be measured directly, we defined non-optical blur as: $1 - \sigma_{opt}/\sigma_{int}$.

METHODS

Subjects

Sixteen individuals (9 males and 7 females, ages 23 to 59 years) participated in the study. No subject had a history of visual abnormalities and all had VA better than 20/25, as measured with the Lighthouse Distance Acuity Chart viewed through the best optical correction and a 3.0 mm artificial pupil. Table 1 provides the characteristics of the subjects,

including sex, age, refraction, and chart acuity. Fourteen of the 16 subjects (all but subjects 7 and 16) participated in the main experiment that evaluated the relative contributions of optical and non-optical sources of intrinsic blur to inter-individual variations in VA. Five subjects (subjects 6, 7, 10, 12, 16) participated in a control experiment that examined the effects of duration on VA and σ_{int} . The study conformed to the tenets of the Declaration of Helsinki and the experiments were approved by an institutional review board at the University of Illinois at Chicago. Written informed consent was obtained from each subject prior to testing.

Psychophysical Measurement of Equivalent Intrinsic Blur

Stimuli and Instrumentation—The test stimuli were tumbling E optotypes, which have advantages over stimuli with curved features, such as the Landolt C, for use with video displays.¹² The tumbling E has been used previously to examine whether correcting higher-order aberrations affects VA.^{e.g., 9,13} The E was constructed according to the principles of the Sloan font,¹⁴ such that the stroke width was 1/5 of the overall optotype size and the three bars were of equal length. The E was either unblurred or blurred through convolution with 2D Gaussian functions of different standard deviations (σ_{stim}). Stimuli were generated by a Macintosh G4 computer using MATLAB software and the Psychophysics Toolbox extensions.¹⁵ The stimuli were displayed on a 22" NEC monitor (FE2111SB) with a screen resolution of 1024 × 768 and an 85-Hz refresh rate, driven by an ATI video card (Radeon 9000 Pro) with 10-bit DAC resolution.

The experiments required the presentation of a broad range of optotype sizes, so an appropriate combination of test distance and σ_{stim} values was selected to avoid potential floor and ceiling effects imposed by the size of the display. For the main experiment, the tumbling E stimuli were presented for approximately 59 ms (5 video frames) at a test distance of 4.5 m. The short exposure duration was selected to minimize the effects of eye movements. The short duration also minimized the likelihood that the log MAR values would fall outside of the range that could be produced by the display at this test distance (stroke widths of 0.6 to 20 arcmin). Under these conditions, the values of σ_{stim} were of 0.8, 3.2, and 12.8 arcmin. Figure 1 illustrates the effect of these values of Gaussian blur on four different stimulus sizes, defined in terms of stroke width.

In a control experiment that examined the effect of stimulus duration on VA and σ_{int} , log MAR and σ_{int} were measured at stimulus durations of both 59 ms and 590 ms (50 video frames). For this experiment, the test distance was increased to 9 m by the use of a front surface mirror. The possible range of stroke widths at this further test distance was 0.3 to 10 arcmin. To ensure that VA for the blurred stimuli would be measurable under these conditions, σ_{stim} was reduced to 0.2, 0.8, and 3.2 arcmin.

The stimuli were presented in the center of an adapting field that subtended 3.4° horizontally and 2.6° vertically at the 4.5-m test distance and 1.7° horizontally and 1.3° vertically at the 9-m test distance. The luminance of the adapting field was 106 cd/m², and the luminance of the unblurred test stimulus was 1.4 cd/m², yielding a Weber contrast of ~99%. The filtered optotypes were presented without rescaling the contrast. The display luminance was calibrated with a photometer (Minolta LS 110) and the temporal characteristics of the display were confirmed using an oscilloscope and photocell.

Procedure and Analysis—Prior to all measurements, the pupil of the tested eye was dilated with 2.5% phenylephrine hydrochloride. The best optical correction for each subject was determined through a 3.0 mm artificial pupil that was mounted on the phoropter. The artificial pupil was used to control the retinal illuminance and also to optimize the optical quality of the eye by minimizing the effects of higher-order aberrations and diffraction. The

dilated pupil of the tested eye was centered on the artificial pupil using a two-dimensional, two-color alignicator.¹⁶ The alignicator was presented between acuity measurements to ensure that the subject maintained proper alignment throughout the test session.

The subject's task was to judge the orientation of the tumbling E, which was randomly facing either to the right or up on each trial. The alternatives were limited to these two so that judgments were based on orientation rather than phase, which can be used as a cue for left-right and up-down judgments.^{17,18} A brief warning tone signaled the start of each stimulus presentation, and the subject verbally reported the orientation, which was recorded by the examiner. The subjects were given a brief practice session to become familiar with the task.

Log MAR for each value of σ_{stim} was determined using a two-alternative forced-choice staircase procedure. An initial estimate of log MAR was obtained by presenting the optotype at a suprathreshold size and then decreasing the size in steps of 0.1 log unit until an incorrect response was recorded. Following this initial search, log MAR was determined using a two-down, one-up decision rule, which provides an estimate of the 71% correct point on a psychometric function.¹⁹ Each staircase continued until 12 reversals had occurred and the mean of the last 8 reversals was taken as log MAR. The staircase length was typically 40–50 trials, which produced stable measurements (the standard error of the mean of the last 8 reversals was typically less than 0.05 log MAR). One staircase measurement of log MAR was obtained from each subject for each value of σ_{stim} , with the stimuli presented in order of increasing σ_{stim} .

Log MAR values were plotted as a function of log σ_{stim} and were fit with the log form of the following equation:³

$$MAR = MAR_0 \left(1 + (\sigma_{stim} / \sigma_{int})^2 \right)^{1/2} \quad (2)$$

where MAR_0 represents the estimated VA for the unblurred target. As noted previously,³ equation 2 is related to equation 1 by substituting MAR_0 for $k \cdot \sigma_{int}$. MAR_0 and σ_{int} were free parameters that were adjusted to minimize the mean squared error using the generalized reduced gradient algorithm, which was used for all curve fitting.

Shack-Hartmann Aberrometry

Instrumentation—A Shack-Hartmann (SH) aberrometer, designed according to previously published specifications,²⁰ was used to measure the wavefront aberration function of the eye. Laser light at a wavelength of 780 nm was focused on the retina by the optics of the eye. The light returned from the retina was sampled by a lenslet array placed at a plane conjugate with the pupil plane. A charge-coupled device (CCD) camera captured the SH image, which was composed of a matrix of spots produced by the lenslet array.

Procedure—During SH image acquisition, the subject fixated on a spot of light produced by a laser light source while the examiner aligned the instrument with the center of the dilated pupil. SH images were obtained after stabilization of the tear film following a blink. Images of poor quality due to artifacts such as eye movements or blinks were not included in the analysis. Seven to nine SH images were obtained for each subject and were averaged for analysis. Partial derivatives of the wavefront aberration function were estimated based on the displacement of the centroid of each spot in the SH image from a perfect grid. The wavefront aberration function was represented by the sum of the third- to sixth-order Zernike polynomials. Calculations were restricted to a 3.0 mm pupil to match the artificial

pupil size used for the psychophysical experiments. The two-dimensional optical point spread function (PSF) was derived from the wavefront aberration function using standard transformations.²¹

The PSF was radially averaged to provide a one-dimensional line profile. Although phase information contained in the PSF was lost by radial averaging, the impact on the PSF was minimized due to the small pupil size. The one-dimensional line profile was normalized to unity and then fit with a Gaussian function, which has been shown previously to provide a good description of the central region of the PSF,²² which is of primary importance in determining visual resolution.²³ The standard deviation of the best-fit Gaussian function (σ_{opt}) defined optical blur.

Although low-order aberrations were optically corrected for each subject using the phoropter, small residual low-order aberrations will likely remain because of small errors in the subjective refraction and the relatively coarse steps (0.25 D) of the phoropter. To determine the effect of any residual low-order error, the PSF for each subject was recomputed with the 2nd order Zernike coefficients included. The 2nd order Zernike coefficients were estimated as the difference between the refractive error derived by aberrometry and the vertex-corrected clinical refraction (i.e., the correction used to measure $\log MAR_0$ and σ_{int}). Including the residual low-order aberrations increased σ_{opt} by only 0.02 log units, on average. Consequently, the low-order aberrations obtained from the SH aberrometry were not included in the PSF used to quantify σ_{opt} .

RESULTS

Psychophysical Measurement of Equivalent Intrinsic Blur

Figure 2 plots $\log MAR$ as a function of $\log \sigma_{stim}$ for one representative subject, obtained for the 59-ms exposure duration. For reference, the right y-axis shows the corresponding Snellen equivalents of the $\log MAR$ values. For small values of $\log \sigma_{stim}$, $\log MAR$ was approximately constant. However, for larger values of $\log \sigma_{stim}$, $\log MAR$ increased in proportion to $\log \sigma_{stim}$. The curve represents the least-squares best fit of equation 2 to the data. This function transitions from a slope of 0 at low values of $\log \sigma_{stim}$ to a slope of 1 at high values. $\log MAR_0$, the VA for the unblurred stimulus predicted by the fit, was -0.11 for this subject. $\log \sigma_{int}$, indicated by the vertical dashed line, is the value of $\log \sigma_{stim}$ that increased $\log MAR_0$ by 0.15 log units (i.e., $\log \sqrt{2}$, as indicated by the horizontal dashed line). For this subject, σ_{int} was approximately 1 arcmin.

Figure 3 shows the relationship between $\log MAR_0$ and $\log \sigma_{int}$ for the individual subjects for the 59-ms exposure duration (see Table 2 for the values for each subject). As in Figure 2, the corresponding Snellen equivalents are shown on the right y-axis for reference. $\log MAR_0$ (estimated VA for the unblurred stimulus) increased systematically with $\log \sigma_{int}$ (equivalent intrinsic blur), as has been reported previously for two-line resolution,^{3,4} and there was a statistically significant correlation between the two parameters ($r = 0.95$, $p < 0.01$). The data were fit with a line with unit slope, which corresponds to the slope of the linear function used previously to describe the relationship between two-line resolution and equivalent intrinsic blur.^{3,4}

The effect of stimulus duration on MAR_0 and σ_{int} is shown in Figure 4. In this figure, mean $\log MAR$ is plotted as a function of $\log \sigma_{stim}$ for the five subjects who participated in the control experiment. The 59-ms and 590-ms exposure durations are represented by the filled circles and open squares, respectively. The primary effect of duration was to shift the function vertically. For the one log unit increase in duration, mean $\log MAR_0$ decreased by 0.16 log units (indicated by the horizontal dashed lines), which was a statistically significant

decrease ($t = 5.33, p < 0.01$). The increase in duration also resulted in a small decrease in the mean value of σ_{int} , (indicated by the vertical dashed lines), but this decrease was not statistically significant ($t = 1.91, p = 0.13$).

Optical Blur

Figure 5 shows the one-dimensional PSF (solid line) derived from SH aberrometry for the same subject whose log MAR data are shown in Figure 2. The dashed line represents a Gaussian function fit to the derived PSF. The standard deviation of the Gaussian function (σ_{opt}) was 0.9 arcmin for this subject. For all subjects, a Gaussian function provided a good description of the PSF over the central region from the peak to 1.5 arcmin (mean $R^2 = 0.96$), consistent with previous work.²²

The relationship between log MAR_0 and log σ_{opt} for the individual subjects is shown in Figure 6 (the value of σ_{opt} for each subject is given in Table 2). The correlation between log MAR_0 and log σ_{opt} was not statistically significant ($r = 0.46, p = 0.11$), although there was a weak trend for log MAR_0 to increase as log σ_{opt} increased.

Non-Optical Blur

As described in the Introduction, non-optical blur (η) was defined as: $1 - \sigma_{opt}/\sigma_{int}$. The value of η provides an index of blur that is introduced by non-optical (i.e., neural) sources, with larger values of η representing higher levels of non-optical blur. The relationship between log MAR_0 and log η is shown in Figure 7 (the value of η for each subject is given in Table 2). Of note, log MAR_0 was independent of log η for subjects with small values of log η . For subjects with larger values of log η , however, log MAR_0 increased in direct proportion to log η . To characterize these data, a piecewise fit was used, consisting of two linear functions with slopes constrained to be 0 and 1, respectively. This piecewise fit provided a good description of the data ($R^2 = 0.87$).

DISCUSSION

The purpose of this study was to determine the relative contributions of optical and non-optical sources of intrinsic blur to variations in VA among normally sighted individuals. VA for the briefly presented tumbling E, represented by log MAR_0 , varied by more than a factor of two (0.3 log units) among our subjects. This variation in VA is consistent with that observed previously for a similar brief exposure duration using Sloan letters.¹¹ Equivalent intrinsic blur, represented by σ_{int} , also varied by more than a factor of two among our subjects. Log MAR_0 was highly correlated with log σ_{int} , consistent with previous studies that used other forms of acuity measurement.^{2,3} Thus, our results confirm that inter-subject differences in equivalent intrinsic blur play a fundamental role in the variation in VA observed among normally sighted individuals.

In contrast to previous studies that have used the equivalent intrinsic blur paradigm, we distinguished between optical and non-optical sources of equivalent intrinsic blur. Our measure of optical blur was obtained using Shack-Hartmann aberrometry. Among our subjects, log MAR_0 was not correlated significantly with optical blur. The lack of a significant relationship between log MAR_0 and higher-order optical aberrations is consistent with previous work using other metrics of optical blur. For example, Villegas et al.⁷ found no significant correlation between VA for the tumbling E optotype and the root mean square (RMS) value for higher-order optical aberrations derived by Shack-Hartmann aberrometry.

However, our results demonstrate a significant, nonlinear relationship between VA and our index of non-optical blur (η). For subjects with log MAR_0 values less than approximately 0, log MAR_0 was independent of log η . For these subjects, the contribution of non-optical blur

to equivalent intrinsic blur was small, and there was minimal variation in $\log MAR_0$. By contrast, for subjects with $\log MAR_0$ values greater than approximately 0, $\log MAR_0$ was proportional to $\log \eta$. For these subjects, individual differences in $\log MAR_0$ were primarily driven by non-optical blur. This finding is consistent with the suggestion of Coppens and van den Berg,⁶ who proposed that individual differences in VA are due to differences in neural processing among individuals. However, we report that this is the case only for those individuals on the lower end of normal VA (i.e. $\log MAR_0$ values greater than 0 for the briefly presented tumbling E). Thus, the results demonstrate that the relative contributions of optical and non-optical sources of intrinsic blur to inter-subject differences in VA are related to the subject's VA level.

Although a brief stimulus presentation duration was used in the present study to minimize the effects of eye movements, the control experiment showed that our basic findings would not be altered significantly if a longer stimulus duration had been used. The primary effect of duration was to produce a vertical shift of the function relating $\log MAR$ to $\log$ stimulus blur (Figure 4). This vertical shift is due to a similar degree of temporal integration for the different levels of stimulus blur. The mean improvement in $\log MAR_0$ due to the one log unit increase in stimulus duration was 0.16 log units, which is consistent with the improvement in VA reported for Landolt C acuity measurements¹⁰ and letter acuity measurements¹¹ using similar stimulus durations. In comparison, there was only a minimal (0.08 log units) and statistically non-significant decrease in σ_{int} . Thus, the primary effect of the increase in duration was to improve VA without significantly reducing σ_{int} .

There are two considerations in the interpretation of our results. First, the wavelength of the laser used to estimate optical blur (780 nm) was different from the effective wavelength of the tumbling E stimulus (presented on the video display). However, this difference would not affect the nature of the relationship between $\log MAR_0$ and $\log$ optical blur (Figure 6) or between $\log MAR_0$ and $\log$ non-optical blur (Figure 7). Second, some information about the width of the PSF may have been lost by modeling the PSF as a Gaussian function. However, the loss would not likely alter the results significantly, because the data were well characterized by the Gaussian functions fit to the PSFs (Figure 5).

It would be of interest to determine whether our index of non-optical blur could be used to predict VA improvements following correction of ocular aberrations with adaptive optics. For example, minimizing optical blur may be most effective in improving visual acuity for subjects whose equivalent intrinsic blur is dominated by optical blur (e.g. subjects whose data fall along the horizontal portion of the function in Figure 7). From a clinical perspective, deriving measures of optical and non-optical blur could potentially be useful for evaluating VA loss in individuals with both optical and non-optical sources of vision loss.

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Figure 1.

Examples of tumbling E optotypes that were either unfiltered (left column) or low-pass Gaussian filtered (right three columns) using the values of σ_{stim} indicated at the bottom, for the four stroke widths indicated at the right.

Figure 2.

Log MAR as a function of $\log \sigma_{stim}$ for one representative subject. The right y-axis indicates the Snellen equivalents of the log MAR values. The solid line represents the least-squares best fit of equation 2 to the data. The arrow indicates $\log \sigma_{int}$, which corresponds to the value of $\log \sigma_{stim}$ (vertical dashed line) necessary to elevate log MAR by 0.15 log units ($\log \sqrt{2}$; indicated by the horizontal dashed line) above $\log MAR_0$.

Figure 3.

Log MAR_0 as a function of log σ_{int} . The right y-axis indicates the Snellen equivalents of the log MAR_0 values. The line has unit slope and the vertical position was determined by minimizing the mean squared error of the fit.

Figure 4.

Mean log MAR as a function of $\log \sigma_{stim}$ for stimulus durations of 59 ms (circles) and 590 ms (squares). Error bars represent the standard errors of the means. The right y-axis indicates the Snellen equivalents of the log MAR values. The solid curves represent the least-squares best fits of equation 2 to the data. The vertical and horizontal dashed lines indicate the values of $\log \sigma_{int}$ and the corresponding log MAR value for the two durations.

Figure 5.

The radially averaged point spread function (solid line) derived from Shack-Hartmann aberrometry for the same subject whose log MAR data are shown in Figure 2. The dashed line represents the least-squares best fit of a Gaussian function. The arrow indicates the value of σ_{opt} .

Figure 6. Log MAR_0 as a function of log σ_{opt} . The right y-axis indicates the Snellen equivalents of the log MAR_0 values.

Figure 7.

Log MAR_0 as a function of $\text{log } \eta$ (defined as: $1 - \sigma_{opt}/\sigma_{int}$). The solid lines represents a piecewise linear fit to the data as described in the text.

Table 1

Subject characteristics

Subject No.	Sex	Age (Years)	Refraction sphere (diopter)	Refraction cylinder (diopter $\times$ angle)	Chart VA (log MAR)
1	F	23	-8.00	0.00	0.00
2	M	24	-1.00	+1.00 $\times$ 2°	0.00
3	M	26	-4.25	0.00	-0.09
4	M	26	-2.50	+1.00 $\times$ 92°	0.01
5	M	26	-2.00	0.00	-0.09
6	M	28	-4.25	+2.00 $\times$ 180°	-0.02
7	M	29	0.00	0.00	-0.13
8	M	34	0.00	+0.25 $\times$ 180°	-0.08
9	M	39	-2.25	0.00	-0.12
10	F	48	-1.00	+0.50 $\times$ 90°	-0.08
11	F	53	-6.50	+0.75 $\times$ 91°	-0.04
12	F	56	0.00	0.00	-0.08
13	F	57	-4.25	+0.75 $\times$ 170°	0.07
14	F	57	+3.25	+1.50 $\times$ 100°	0.01
15	F	58	+1.25	+0.25 $\times$ 140°	0.05
16	M	59	0.00	0.00	-0.07

Table 2Log MAR_0 , log blur values, and log η for the 59-ms stimulus

Subject No.	Log MAR_0	Log σ_{int} (arcmin)	Log σ_{opt} (arcmin)	Log η
1	0.16	0.23	-0.12	-0.26
2	-0.10	-0.12	-0.15	-1.10
3	-0.01	0.04	-0.09	-0.61
4	0.11	0.20	0.00	-0.44
5	0.08	0.14	-0.09	-0.38
6	-0.11	0.00	-0.14	-0.58
7	-0.24	-0.21	n/a	n/a
8	-0.04	0.09	0.06	-1.24
9	-0.05	0.01	-0.05	-0.90
10	-0.03	0.02	-0.10	-0.62
11	0.15	0.20	-0.03	-0.39
12	0.16	0.18	-0.07	-0.36
13	0.30	0.35	-0.03	-0.24
14	-0.06	-0.02	-0.09	-0.81
15	0.12	0.12	-0.15	-0.33
16	0.17	0.12	n/a	n/a