Retinal image contrast obtained by a model eye with combined correction of chromatic and spherical aberrations

Kazuhiko Ohnuma,1,* Hiroyuki Kayanuma,1 Tjundewo Lawu,2 Kazuno Negishi,3 Takefumi Yamaguchi,3 and Toru Noda4

1Department of Medical System Engineering, Faculty of Engineering, Chiba University, 1-33, Yayoi-cho, Inage-ku, Chiba 263-8522, Japan
2Hoya Corporation Medical Division, 1-12-11 Funado, Itabashi-ku, Tokyo 174-0041, Japan
3Department of Ophthalmology, Keio University, 2-15-45 Mita, Minato-ku, Tokyo 108-8345, Japan
4Department of Ophthalmology, National Hospital Organization Tokyo Medical Center, 2-5-1 Higashigaoka, Meguro-ku, Tokyo 152-8621, Japan

*oonuma@faculty.chiba-u.jp

Abstract: Correcting spherical and chromatic aberrations in vitro in human eyes provides substantial visual acuity and contrast sensitivity improvements. We found the same improvement in the retinal images using a model eye with/without correction of longitudinal chromatic aberrations (LCAs) and spherical aberrations (SAs). The model eye included an intraocular lens (IOL) and artificial cornea with human ocular LCAs and average human SAs. The optotypes were illuminated using a D65 light source, and the images were obtained using two-dimensional luminance colorimeter. The contrast improvement from the SA correction was higher than the LCA correction, indicating the benefit of an aspheric achromatic IOL.

©2011 Optical Society of America

OCIS codes: (170.0170) Medical optics and biotechnology; (330.0330) Vision, color, and visual optics.

References and Links
1. L. N. Thibos and A. Bradley, Wavefront Customized Visual Correction. The Quest for Super Vision II (SLACK Inc., Thorofare, N.J., 2003), Chap. 10.
2. S. Ravikumar, L. N. Thibos, and A. Bradley, “Calculation of retinal image quality for polychromatic light,” J. Opt. Soc. Am. A 25(10), 2395–2407 (2008).
3. S. Marcos, S. A. Burns, E. Moreno-Barriasop, and R. Navarro, “A new approach to the study of ocular chromatic aberrations,” Vision Res. 39(26), 4309–4323 (1999).
4. S. Manzanera, C. Canovas, P. M. Prieto, and P. Artal, “A wavelength tunable wavefront sensor for the human eye,” Opt. Express 16(11), 7748–7755 (2008).
5. J. S. McLellan, S. Marcos, P. M. Prieto, and S. A. Burns, “Imperfect optics may be the eye’s defence against chromatic blur,” Nature 417(6885), 174–176 (2002).
6. J. Schwiegerling and J. Choi, “Application of the polychromatic defocus transfer function to multifocal lenses,” J. Refract. Surg. 24(9), 965–969 (2008).
7. A. Franchini, “Compromise between spherical and chromatic aberration and depth of focus in aspheric intraocular lenses,” J. Cataract Refract. Surg. 33(3), 497–509 (2007).
8. N. Lópe-Gil and R. Montés-Micó, “New intraocular lens for achromatizing the human eye,” J. Cataract Refract. Surg. 33(7), 1296–1302 (2007).
9. P. Artal, S. Manzanera, P. Piers, and H. Weeber, “Visual effect of the combined correction of spherical and longitudinal chromatic aberrations,” Opt. Express 18(2), 1637–1648 (2010).
10. I. Powell, “Lenses for correcting chromatic aberation of the eye,” Appl. Opt. 20(24), 4152–4155 (1981).
11. L. N. Thibos, M. Ye, X. Zhang, and A. Bradley, “The chromatic eye: a new reduced-eye model of ocular chromatic aberation in humans,” Appl. Opt. 31(19), 3594–3600 (1992).
12. T. Terwee, H. Weeber, M. van der Mooren, and P. Piers, “Visualization of the retinal image in an eye model with spherical and aspheric, diffractive, and refractive multifocal intraocular lenses,” J. Refract. Surg. 24(3), 223–232 (2008).
13. J. Choi and J. Schwiegerling, “Optical performance measurement and night driving simulation of ReSTOR, ReZoom, and Tecnis multifocal intraocular lenses in a model eye,” J. Refract. Surg. 24(3), 218–222 (2008).
14. P. G. Gobbi, F. Pasch, S. Bozza, and R. Brancato, “Optomechanical eye model with imaging capabilities for objective evaluation of intraocular lenses,” J. Cataract Refract. Surg. 32(4), 643–651 (2006).
1. Introduction

Two types of chromatic aberrations of the ocular optics have been investigated, i.e., transverse chromatic aberrations and longitudinal chromatic aberrations (LCAs). Thibos and Bradley [1] summarized their results. A transverse chromatic aberration is defined as the transverse or lateral image shift as a function of wavelength. A transverse chromatic aberration is dependent on the centration of each person’s pupil. The frequency distribution of the magnitude of the transverse chromatic aberration is in accordance with the Rayleigh distribution. A LCA is the change in the focusing power with wavelength in the longitudinal axial position, which is about 2 diopters in the visual wavelength range. Ravikumar et al. [2] recently proposed a method to derive the aberration quantity of all wavelengths from infrared wavefront sensor data and calculate the quality of the retinal images. The image qualities with or without LCAs and transverse chromatic aberrations have been compared with the through-focus image quality for model eyes using a polychromatic metric called the visual Strehl ratio. The results showed that the image quality of the ocular optics with monochromatic aberration was almost unaffected by LCAs, whereas the image quality of the ocular optics without monochromatic aberration was affected more by LCAs. Further, the investigators also reported a small effect of the transverse chromatic aberrations. However, the depth of focus becomes larger in the ocular optics with monochromatic aberration. The simulated color images of white objects were shown with LCAs and without monochromatic aberrations. Unfortunately, the simulated images with LCAs and monochromatic aberrations were not presented.

Different approaches have been examined by measuring the ocular aberrations of each wavelength [3,4]. McLellan et al. [5] reported the highest modulation transfer function (MTF) at the 555-nm wavelength in the case of ocular optics without monochromatic aberration and with LCA. The investigators calculated the MTFs of the long (L, 560–580 nm), middle (M, 530–540 nm), and short (S, 420–440 nm) cones and reported almost the same MTFs from the short to long wavelengths with monochromatic aberrations and with LCAs. They concluded that imperfect optics might be due to the ocular defense against chromatic blur. However, the images of the L, M, and S cones were not presented.

In the field of intraocular lenses (IOLs), Schwiegerling and Choi [6] reported that multifocal IOLs have a larger depth of focus with white light illumination compared with monochromatic illumination. Franchini [7] reported the depth of focus of various aspheric IOLs with different SA. A new IOL design to compensate for the LCA has been proposed [8]. Artal and colleagues [9] reported improved contrast and visual acuity by correction of the ocular LCAs and SAs in experiments using human eyes. The authors recommended IOLs with functionality to correct the ocular LCAs.

In the current paper, we designed a model eye comprised of an artificial cornea with a LCA and SA, IOL, and two-dimensional (2-D) luminance colorimeter. The tristimulus values X, Y, and Z of each pixel were obtained by 2-D luminance colorimeter, and the values of L, M, and S cones of each pixel were calculated from the X, Y, and Z values. The optotypes were illuminated by a D65 fluorescent lamp. The contrasts of the gap in the Landolt C chart Y images then were calculated. The images of defocus were obtained by changing the distances of the optotypes, and the depth of focus was recognized easily by observing the images. Moreover, the images without LCA were obtained by applying the LCA compensatory lens [10] in front of the model eye.
Various model eyes using lenses to simulate the cornea to estimate the point spread function, the MTF, and the retinal images have been proposed [11–15]. Choi and Schwiegerling [13] reported a model eye with a LCA and SA. Their artificial cornea had refractive power to focus the incident light 27 to 28 mm from the IOL. The SA of the artificial cornea was 0.27 μm and the LCA was 2.5 diopters in wavelengths ranging from 400 to 700 nm. The LCA was nearly equal to the LCA of the model of Atchison and Smith [16]. The MTFs achieved by focusing three types of multifocal IOLs were obtained by analyzing the images from a slit illuminated by a monochromatic light source. Moreover, the color images were obtained using a commercial digital single-lens reflex camera Nikon D70 (Nikon Corp., Tokyo, Japan) and compared qualitatively, since it is difficult to analyze the color with the commercial digital camera images.

Gobbi et al. [14] made an artificial cornea with 43 diopters using polymethylmethacrylate (PMMA) with a refractive index of 1.490. The collimated incident light was converted into converging light similar to that of human eyes by the artificial cornea. The anterior and posterior curvatures of the surfaces of the artificial cornea were 7.7 and 7.4 mm, respectively. The images provided by this model eye were magnified 2.44 times using two camera lenses to coincide the pixel size of a one-third inch CCD camera with the cone size. However, the authors did not take the chromatic aberrations and color spectrum used in the experiments into consideration.

Norrby et al. [15] examined a modified ISO model PMMA eye. They also adopted a prolate aspheric anterior curvature of 7.8 mm with a Q value of −0.0205, and spherical posterior curvature of 7.02 mm and called the proposed model a physiologic model eye. The collimated incident light converged 27.65 mm from the iris pupil in both models, which agrees with the current ISO model. However, the modified ISO model had lower refractive power with a 39.03-mm focal length compared with the focal length of the human cornea. However, the physiologic model eye had similar refractive power with a 24.2-mm focal length similar to the average focal length of the human cornea. The investigators compared the MTFs of the two model eyes at the artificial 3.0- and 4.5-mm pupils and found that the MTFs were equivalent at the focal point or at the different position of the image plane in the case of the collimated incident light. However, for near object measurements, only the physiologic model eye provided relevant results. We made the artificial cornea using PMMA with an aspheric anterior surface and flat posterior surface. The artificial cornea had a 0.27-μm positive SA at 6-mm entrance pupil size. The refractive power of the artificial cornea was slightly smaller than the refractive power of the ocular cornea. The images constructed by the model eye were magnified using two camera lenses to coincide the pixel size with the cone size in the fovea. Therefore, the image contrast was affected by both camera lenses to give a slightly lower MTF at high spatial frequency. Fortunately, correcting the contrast data was unnecessary because we only analyzed the lower spatial frequency which was almost not affected by both camera lenses.

2. Materials and methods

The model eye had the dimension and refractive index of each surface shown in Table 1. This design was optimized to provide a SA and LCA similar to that of the modified Gullstrand model (Table 2) using the ZEMAX optical design software (ZEMAX Development Corporation, Bellevue, WA, USA). The artificial cornea had an aspheric anterior surface with a 1.19 conic constant corresponding to a 0.27-μm positive SA in a 6-mm entrance pupil. The posterior surface of the artificial cornea was laid on the planar surface of the glass cell made of BK7. The glass cell was filled with pure water, and the collimated incident light converged in air 25.58 mm from the iris pupil, which was slightly shorter than the 26.30 mm of the modified Gullstrand model (with a 4.0-mm distance from the posterior cornea to the iris pupil) and 27.65 mm of the model eye by Norrby et al. [15]. The IOL was set in a black IOL holder with a central circular hole substituting as the iris pupil. The IOL holder was inserted into the glass cell with the anterior surface of the IOL facing the artificial cornea. The chromatic aberration of our model and the modified Gullstrand model with the 0.27-μm positive SA were compared.
Figure 1 shows the coincidence of the two chromatic aberration curves with the LCA refractive error of about 2.0 diopters. Figure 2 shows the diagram of the entire model eye including 2-D luminance colorimeter CA-2000 with 1000 × 1000 pixels black and white CCD camera (Konica Minolta, Tokyo, Japan) [19]. The images of the model eye were magnified to 50/16 = 3.1 times by two camera lenses with 16- and 50-mm focal lengths, respectively, to coincide with the pixel sizes of 8.8 × 6.6 μm, which was larger than the cone size. The XYZ color values were obtained by 2-D luminance colorimeter. The values of the L, M, and S cones then were calculated from the XYZ values. The photograph of the optical arrangement is shown in Fig. 3. The Landolt C optotypes were set at 5, 2.5, 1.66, 1.25, and 1.0 m and illuminated by a D65 fluorescent lamp (FLR40S-D-EDL-D65/M Toshiba, Tokyo, Japan). The best focus was set at 5 m. The 20 diopeters spherical IOL VA-60BB (HOYA, Tokyo, Japan) and the 20 diopeters aspheric IOL with 0.27-μm negative SA Tecnis ZA9003 (AMO, Santa Ana, CA) were used to compare the image contrast of the large SA (about 0.4 μm) and 0 SA in the whole optics, respectively. To correct the LCA, the lens designed by Powel [10] with the negative LCA set in front of the model eye. The following four cases of experiments were examined: case 1, correction of the LCA and the SA; case 2, correction of only the SA; case 3, correction of only the LCA; and case 4, no correction of the LCA or the SA.

### Table 1. Model Eye Design

| Medium | Radius (mm) | Conic Constant | Thickness (mm) | Refractive Index (wavelengths in nm) |
|--------|-------------|----------------|----------------|------------------------------------|
| Air    | Infinity    | 0              | Infinity       | 1                                  |
| PMMA   | 13.83       | 1.19           | 4.55           | 1.503145 1.497763 1.493795 1.491680 1.489199 |
| Air    | Infinity    | 0              | 0.1            | 1                                  |
| BK7    | Infinity    | 0              | 0.5            | 1.527204 1.522379 1.518722 1.516728 1.514321 |
| Water  | Infinity    | 0              | 3.5            | 1.340583 1.337123 1.334468 1.332990 1.331151 |
| BK7    | Infinity    | 0              | 0.5            | 1.527204 1.522379 1.518722 1.516728 1.514321 |
| Air    | Infinity    | 0              | 21.580447      | 1                                  |

*The refractive indices of each wavelength of water and PMMA are quoted from Handbook of Optics Vol. I Chap. 43 and Vol. II Chap. 34 [17].

### Table 2. Design of the Modified Gullstrand Model

| Medium | Radius (mm) | Conic Constant | Thickness (mm) | Refractive Index (wavelengths in nm) |
|--------|-------------|----------------|----------------|------------------------------------|
| Air    | Infinity    | 0              | Infinity       | 1                                  |
| Cornea | 7.7         | −0.25          | 0.5            | 1.385997 1.381694 1.378571 1.376922 1.374993 |
| Vitreous | 6.8        | 0              | 30.387030      | 1.344365 1.340394 1.337479 1.335926 1.334093 |

*The refractive indices of each wavelength of cornea and vitreous are obtained by a polynomial fit of the refractive indices given by Le Grand [18].

The values of L, M, and S cones are derived from the X, Y, and Z values obtained by 2-D luminance colorimeter according to the following equation.

\[
\begin{bmatrix}
  L \\
  M \\
  S
\end{bmatrix} =
\begin{bmatrix}
  0.4002 & 0.7076 & -0.0808 \\
  -0.2263 & 1.1653 & 0.0457 \\
  0 & 0 & 0.9182
\end{bmatrix}
\begin{bmatrix}
  X \\
  Y \\
  Z
\end{bmatrix}
\]

(1)
This equation represents the RLAB color appearance model, which used the Hunt-Pointer-Estevez transformation matrix for conversion from CIE XYX to LMS normalized to D65. The transformation matrix from the XYZ to LMS in Eq. (1) depended on the color appearance models. Therefore, the values of L and M depended on the transformation matrix. However, S was nearly equal to Z in any color appearance models.

The contrasts of the Landolt C gap in the Y images were calculated. Moreover, the effect of the monochromatic aberration and the LCA, obtained by McLellan et al. [5] were verified using the images of the L, M, and S cones.

![Graph showing displacement of focus position between the model cornea and the modified Gullstrand model. The 546.07-nm (e-ray) wavelength is used for reference.](image1)

**Fig. 1.** Comparison of the displacement of focal position between the model cornea and the modified Gullstrand model. The 546.07-nm (e-ray) wavelength is used for reference.

![Diagram of the whole model eye including the 2-D luminance colorimeter CA-2000 (Konica Minolta, Tokyo, Japan). The images of the model eye are magnified to 50/16 = 3.1 times by two camera lenses: L1 with a 16-mm and L2 with a 50-mm focal length.](image2)

**Fig. 2.** Diagram of the whole model eye including the 2-D luminance colorimeter CA-2000 (Konica Minolta, Tokyo, Japan). The images of the model eye are magnified to 50/16 = 3.1 times by two camera lenses: L1 with a 16-mm and L2 with a 50-mm focal length.

![Photograph of the optical arrangement.](image3)

**Fig. 3.** Photograph of the optical arrangement.
3. Results

Figure 4 shows the color images of the four cases. The color Landolt C images at 5, 2.5, 1.66, 1.25, and 1.0 m were composed from the X, Y, Z values in 2-D luminance colorimeter. The Landolt C in the central images of the optotypes at 5 and 2.5 m had no color at the edge of the C. However, the peripheral Landolt C images had colored edges due to imperfect compensation of the transverse chromatic aberration. Therefore, only the contrasts of the Cs in the center of the images were estimated in the analysis. Figure 5 shows the enlarged images of the Landolt Cs of logarithm of the minimum angle of resolution (logMAR) 0.4, 0.3, 0.2, and 0.1 at 2.5, 1.66, and 1.25 m for case 2. The color distributions of these images were not dependent on the direction. These verified the absence of the transverse chromatic aberration.

Fig. 4. Color images of the optotypes of the Landolt C chart at 5, 2.5, 1.66, 1.25, and 1.0 m in four cases: case 1, correction of both the LCA and the SA; case 2, correction of only the SA; case 3, correction of only the LCA; and case 4, no correction of the LCA or the SA.

Fig. 5. Enlarged images of the Landolt Cs of logMAR 0.4, 0.3, 0.2, and 0.1 at 2.5, 1.66, and 1.25 m in case 2 (correction of only the SA). This shows no transverse chromatic aberration in the center of the images.
Fig. 6. The X, Y, and Z images of the optotypes of the focus position for all cases.

Fig. 7. The contrasts of the brightest point and the darkest point around the gap of the Landolt Cs of logMAR 0.4, 0.2, 0, and −0.2 for four cases. (a) Case 1, (b) case 2, (c) case 3, and (d) case 4.
in the center of the images. The colors of the Landolt Cs by reducing the distance changed from violet and magenta, to orange. Figure 6 shows the X, Y, and Z images of the optotypes at 5 m for all cases. The highest Y image contrast was seen for case 1 (correction of both the LCA and SA). Furthermore, the contrasts of the brightest point and the darkest point around the gap of the Landolt C charts for logMAR 0.4, 0.2, 0, and −0.2 were calculated, and the relations between the contrast and object vergence for all cases are shown in Fig. 7. The order of the contrast from high to low was cases 1, 2, 3, and 4 in the images of the optotypes at 5 m. However, the highest contrast was in case 3 in the optotype image at 1.66 m (−0.6 diopters), whereas the lowest contrast was in case 1. These results showed that the SA and the LCA play the same role, i.e., extending the depth of focus. This phenomenon also was verified in case 4 (without correction of the LCA or SA), which had the largest depth of focus due to the combination of LCA and SA. Again, these results were similar to the results obtained by Thibos and Bradley and Ravikumar et al. [1,2].

Figure 8 shows the L, M, and S cone images obtained by substituting the X, Y, and Z images of the optotypes at 5 m into Eq. (1) for the four cases. The S cone images had low contrast in case 2 (correction of only the SA) and case 4 (no correction of LCA or SA); however, the S cone image in case 3 had higher resolution than the image in case 4. The results verified that imperfect optics might be due to an ocular defense against chromatic blur as McLellan et al. concluded [5].

![Images of optotypes at 5 m for all cases](image)

Fig. 8. The L, M, and S cones images. These images are obtained by substituting the X, Y, and Z images of the optotypes at 5 m into Eq. (1) in four cases. (a) Case 1, (b) case 2, (c) case 3, and (d) case 4.

4. Discussion

The highest contrast can be obtained by correcting both the SA and the LCA, and the largest depth of focus is obtained without correcting the SA and the LCA. This shows that the LCA
plays a similar role to that of the SA. The results were similar to the analytic simulation of Thibos and Bradley and Ravikumar et al. [1,2] and the estimation of McLellan et al. [5]. Their estimation that the SA might be the ocular defense against chromatic blur is considered true. However, it is better and natural to consider that the LCA has a similar role to that of the SA.

Ordinarily, we see objects in the presence of the LCA, and the images of black and white objects have the colored edges as shown in Fig. 4. However, the colored edges of the Landolt Cs are not usually seen. This suggests the role of the neural system of disturbed recognition of the colored edge. Therefore, the images of the L, M, and S cones become a good resource for confirming the role of the neural system. Mullen has shown the contrast sensitivity of human color vision to red-green and blue-yellow chromatic gratings [20]. In his report, these channels have cut off spatial frequency at 10 cycles/deg. In other words, the color effect seen in many of these images would be filtered out and only the luminance information would be perceived.

Correcting the SA and the LCA of the IOL as recommended by Artal et al. [9] also is considered reasonable from our results.

5. Conclusion

We evaluated the effects of the SA and the LCA using IOLs and the model eyes with 2-D luminance colorimeter. The largest contrast images obtained by correcting both the SA and the LCA were similar to the results of Artal et al. [9]. In contrast, the largest depth of focus was obtained without correcting both aberrations. The SA as the monochromatic aberration was used. However, other monochromatic aberrations such as astigmatism are considered to have a different effect. To evaluate the effect of astigmatism, a new model eye will be proposed.

Acknowledgments

This study was supported by Prof. Dr. Hirohisa Yaguchi and Dr. Yoko Mizokami of the Department of Informatics and Imaging Systems, Chiba University.