Published in final edited form as:

Appl Opt. 2009 July 1; 48(19): 3580–3587.

Astigmatism and defocus wavefront correction via Zernike modes produced with fluidic lenses

Randall Marks¹, David L. Mathine^{1,*}, Jim Schwiegerling^{2,1}, Gholam Peyman^{1,3}, and Nasser Peyghambarian¹

¹College of Optical Sciences, University of Arizona, 1630 East University Boulevard, Tucson, Arizona 86721, USA

²Department of Ophthalmology and Vision Science, College of Medicine, University of Arizona, 655 North Alvernon Way, Suite 108, Tucson, Arizona 86711, USA

³Department of Ophthalmology, 1430 Tulane Avenue—SL69, Tulane University, New Orleans, Louisiana 70112, USA

Abstract

Fluidic lenses have been developed for ophthalmic applications with continuously varying optical powers for second order Zernike modes. Continuously varying corrections for both myopic and hyperopic defocus have been demonstrated over a range of three diopters using a fluidic lens with a circular retaining aperture. Likewise, a six diopter range of astigmatism has been continuously corrected using fluidic lenses with rectangular apertures. Imaging results have been characterized using a model eye.

1. Introduction

Dynamic wavefront compensation has been accomplished in the past using continuous membrane deformable mirrors[1,2]; however, these are expensive enough to preclude their use in many applications. Alternately, liquid crystal spatial light modulators in either a transmissive [3] or reflective [4–9] configuration can be used. Both approaches for liquid crystal wavefront modifiers provide a high degree of flexibility and have demonstrated Zernike corrections up to 15 terms. However, both approaches suffer from pixelation, which causes phase discontinuities at the pixel edges, as well as limited dynamic range. While higher order aberrations can be corrected with these technologies, only limited levels of defocus and astigmatism can be corrected. Reflective systems usually offer higher uniformity in wavefront correction, but suffer from awkward and large configurations when used with the eye due to rerouting the line of sight through the reflective system. The transmissive approach allows direct view, but is the most limited by the number of pixels due to the requirements for electrical connection of the individual liquid crystal pixels, as well as polarization sensitivity.

We describe an ophthalmic application of fluidic lenses, which targets low-order wavefront correction of the defocus and astigmatism. Below, we describe spherical and cylindrical fluidic lenses, which are composed of a flexible membrane and a fluid reservoir. By increasing the volume of fluid in the reservoir the membrane stretches, which alters the optical wavefront of light passing through the membrane. Thus, by continually adjusting the fluid volume in the lens, a continuously variable focal length in a transmissive medium can be obtained while

circumventing pixelation. The fluidic lens also has the advantage of being able to change focal length without mechanically moving the lens, which can provide considerable power savings and eliminate wear associated with moving parts.

A lens with a circular retaining ring was used to generate defocus compensation, while a rectangular retaining ring was used for phase compensation of astigmatism. In both cases the lenses were designed so that the optical power was continuously adjustable between negative and positive optical powers, which are in the typical range of correction for eyeglasses. Prior work on polydimethylsiloxane (PDMS) lenses [10,11] concentrated on spherical focal lengths shorter than 50 mm (greater than 20 diopters (D)), which are not typically used for ophthalmic corrections.

2. Methods

A. Fluidic Lens Fabrication

A key component of the fluidic lens is a PDMS membrane that can be stretched to provide a change in the optical wavefront. The PDMS is fabricated in a machined aluminum mold that contains a 2 in. glass plate on the bottom of the mold to provide an optically flat surface to mold the membrane on. The PDMS (Sylgard 184, Dow Corning) was mixed with a 10:1 ratio of PDMS to curing agent, and approximately 0.75 ml of PDMS was poured into the mold. The PDMS was then out gassed in a vacuum to remove air bubbles. Next, the membrane was baked in an oven at 65 °C for four hours to completely cure the PDMS. The PDMS membrane was then removed from the mold and measured to be approximately 400 μm thick. This thickness of the PDMS is thicker than prior reports that ranged from 30 to 120 μm [11–13]; however, the thicker membrane provided more control for the lower optical powers studied in this paper.

An aluminum holder was machined (Fig. 1) to mount the membrane for the lens. The membrane plate in the lens was designed to provide a beveled edge for the PDMS to rest on. A retaining ring was used to mechanically secure the PDMS to the membrane plate. The retaining ring also caused the PDMS membrane to stretch when secured over the beveled edge, which was important to provide a uniform optical surface for the PDMS membrane without any unwanted waves or bumps. The clear aperture and restraint for the circular lens was 22.4 mm, while a rectangular aperture of 16 mm $\times$ 30 mm was used to produce the astigmatism cell. The fluidic chamber was nominally 8.58 mm thick for the circular lens and contained an inlet hole for fluid and an exit hole to remove unwanted gas during the initial filling of the lens. A circular glass aperture was used for the other side of the lens. This glass was 2 mm thick with a diameter of 35 mm (Edmund Optics NT45-642) and was mechanically secured to the aluminum holder.

The PDMS membrane was then placed on top of a 1.25 in. diameter glass plate so that there were no bubbles between the plate and the membrane. This ensured that the outward facing surface of the membrane, which was molded against the optically flat surface, was relatively free of any unwanted waves or bumps. The membrane was then submerged in deionized water and attached to the beveled edge of the aluminum holder. Next, the 1.25 in. diameter glass plate was removed, and the aluminum holder was pressed and restrained against a flange on the clear aperture of the lens cell to provide additional uniform tension to the PDMS membrane. The assembly was then removed from the deionized water.

Deionized water was used for the initial demonstrations of the lens because of its simplicity. A two-syringe system (Fig. 1) was used to provide fluid to the lens. The larger syringe was used to fill the lens chamber, while the smaller syringe was used to provide more control over the amount of water injected into the lens.

B. Shack–Hartmann

The fluidic lenses were measured using a Shack–Hartmann test to provide a refractive measurement of the lens. The experimental setup used a 532 nm laser that was expanded and collimated to a beam that was approximately 1 cm in diameter. The lenslet array consisted of a 64×64 array of lenses that were separated on $250 \mu\text{m}$ centers with a focal length of 19 mm. The image was recorded with a 1/3 in. CCD camera that has $7.6 \mu\text{m}$ and a resolution of 640×480 pixels. A relay lens with a $0.5\times$ magnification was used to image the output of the lenslet array onto the CCD camera. A reference beam was used to calibrate the sensor prior to insertion of the fluidic lenses. The Shack–Hartmann wavefront sensor is sensitive to slopes of approximately $0.08 \mu\text{m}/\text{mm}$ with a maximum detectable slope of $24 \mu\text{m}/\text{mm}$. The recorded images were analyzed on a personal computer, and the measured wavefronts were decomposed into Zernike polynomials.

C. Optical Profiler

A WYKO NT 9800 commercial optical profiler was used to characterize the surface of the PDMS membranes. The instrument is a white-light interferometer that uses the fringe visibility of a low coherence source to obtain surface information from the sample. The optical profiler field of view was maximized for our instrument, using the smallest field lens of $0.55\times$ and the smallest magnification of $5\times$. This resulted in a field of view of approximately 2 mm. The optical profiler was operated in a vertical scanning mode, which obtains two-dimensional images during a vertical scan and then uses these images to obtain the surface profile. Twelve images were taken in a three by four pattern and then stitched together to produce a 5 mm image.

D. Model Eye

Imaging experiments were performed through the fluidic lenses using a model eye (Fig. 2). A saline-filled chamber with a 20D intraocular lens was used as the eye model. Next, a 6 mm pupil is used to approximate the pupil of the eye. Then a 40D lens is used to approximate the cornea. Finally, ophthalmic trial lenses were used to generate spherical and cylindrical refractive error when needed.

The fluidic lenses were placed in front of the model eye. The 6 mm pupil in the eye became the system pupil and generated a footprint of approximately 7.6 mm on the PDMS membrane. With the field of view performed in these tests, approximately 10.6 mm of the membrane is examined in the images.

E. Spherical Lens Model

A simple relationship between the optical power and the changes in fluidic volume of the lens can be derived from the geometrical design (Fig. 3) of the lens with the assumption that changes in fluidic volume will alter the curvature of the lens [14]. The optical power can be related to the lens curvature (r_c) using the relationship

$$r_c = \frac{n_{\text{fluid}} - n_{\text{air}}}{D}, \quad (1)$$

where the index of refraction of the PDMS is assumed to be 1.33 and that of air is 1.00. The optical power from the water/PDMS interface is negligible due to the membrane thickness. The optical power (D) is given in diopters. The curvature is then used to calculate the lens sag (h) according to

$$h = r_c \pm \sqrt{r_c^2 - d^2/4}, \quad (2)$$

where d is the diameter of the restraining aperture of the lens. It follows that the change in fluid volume (V) can be determined according to

$$V = \pi h^2 \frac{(3r_c - h)}{3}. \quad (3)$$

3. Results and Discussion for Variable Focus Lens

A. Spherical Lens

A lens with a circular aperture was constructed and characterized. The theoretical plot (Fig. 4) shows excellent agreement with the optical power determined from the surface shape measured using the optical profiler. No noticeable hysteresis is seen in the data, which is because the liquid is incompressible and the membrane has a pretension. The slight difference between the theory and experimental data is attributed to the finite thickness of the PDMS membrane and edge effects around the periphery of the lens aperture.

The wavefronts produced from the spherical lens surface are shown in Fig. 5 as the lens fluid is adjusted so that the optical power of the lens changes from positive to negative values. The line scan in the y direction shows a small amount of aberrations near zero optical power. We attribute this to errors in mounting of the PDMS membrane, and reduction of this error can be achieved with improved membrane alignment.

The optical quality of the lens was also characterized using a Shack–Hartmann test with a 6 mm pupil. The fluidic lens was relayed to a lenslet array to assess the emerging wavefront. Images of the Shack–Hartmann spot pattern were recorded by a personal computer as a function of fluid volume, and the resultant wavefront was reconstructed from these images. The resultant images were used to compute the coefficients for the Zernike polynomials (Fig. 6). As expected, the defocus term $Z(2,0)$ dominates the Zernike modes with the other coefficients less than $0.35 \mu\text{m}$. The dominant aberration is the astigmatism term $Z(2,2)$, which starts out at $0.163 \mu\text{m}$ for 3.03D and monotonically increases to $0.35 \mu\text{m}$ RMS as the optical power changes to -2.98D . All the other coefficients are less than $0.1 \mu\text{m}$ RMS, which demonstrates the high optical quality of the lens.

The overall residual error was determined by subtracting the Zernike terms $Z(0,0)$, $Z(1,-1)$, $Z(1,1)$, and $Z(2,0)$ from the Shack–Hartman wavefront profile. The resultant residual plots are shown in Fig. 7 for images from approximately +3D to -3D . The plots show a systematic progression of the residual error, which includes the higher order Zernike terms, that we have attributed to artifacts of the membrane alignment and possibly to membrane nonuniformity. In all cases, the maximum residual error is less than $1 \mu\text{m}$ peak-to-valley and produces negligible effects as perceived by the human eye.

To simulate the performance of these lenses for ophthalmic uses, a model eye was used to show the effect of the circular lens. Initially, an ophthalmic one-diopter spherical trial lens was placed in the optical path to simulate defocus and an image of Siemen's star was taken [Fig. 8(a)]. One can observe the characteristic circle of confusion where the optical transfer function becomes negative and the dark lines are inverted. There is a slight ellipticity in the circle, which is due to the small amount of astigmatism present in the lens as determined in the Shack–Hartmann tests. Next, the fluidic lens was adjusted to correct for the defocus generated by the ophthalmic

trial lens and an image was recorded [Fig. 8(b)]. The corrected image shows that the large majority of aberrations are corrected with a small amount of field aberrations are attributed to a slightly shifted pupil in the model eye, as well as the alignment astigmatism. Finally, similar images using a 2D lens were taken for the uncorrected [Fig. 8(c)] and corrected [Fig. 8(d)] fluidic lens. The uncorrected image shows the contrast reversal has moved outward as expected, while the corrected image is similar to the previous corrected image.

The optical quality of the lens was assessed using the visual Strehl ratio computed in the frequency domain (VSOTF) [15], using the formula

$$\text{VSOTF} = \frac{\int_{-\infty}^{\infty} \int_{-\infty}^{\infty} \text{CSF}_N(f_x, f_y) \text{OTF}(f_x, f_y) df_x df_y}{\int_{-\infty}^{\infty} \int_{-\infty}^{\infty} \text{CSF}_N(f_x, f_y) \text{OTF}_{\text{DL}}(f_x, f_y) df_x df_y}. \quad (4)$$

The VSOTF is the ratio of the measured optical transfer function (OTF) to that of the diffraction limited case. However, this metric is weighted by the contrast sensitivity function (CSF_N) to account for the human visual response [16].

The VSOTF was calculated from the optical profile (Fig. 9) and found to be almost 0.8, which is quite acceptable for ophthalmic applications. The VSOTF was limited by the small amount of residual astigmatism present in the lens. This could potentially be improved by better membrane alignment techniques that we are currently pursuing.

B. Astigmatic Lens

The optical power from the astigmatic lens was measured for different fluid volumes between 120 and $-90 \mu\text{L}$ from a baseline volume (Fig. 10). The resultant ophthalmic optical power ranged from approximately 0D to 3D of cylinder, while the axis changed approximately 90° near the crossover point of the lens. The spherical term ranged from -0.95D to -5.45D .

The wavefront for the astigmatic lens was analyzed using the Shack–Hartman test with a 6 mm pupil. The resultant Zernike coefficients (Fig. 11) were extracted from the measured wavefronts for a range of optical powers. The defocus and astigmatism ($Z(2,0)$ and $Z(2,2)$) were produced in approximately equal quantities. The other oblique astigmatism term, $Z(2,-2)$, had a small finite value because the lens was slightly rotated during the test, which produced this orthogonal astigmatism term. The higher-order terms are also present but at a much lower value than those produced with the circular restraint for the elastic membrane. We hypothesize that the restraint process creates tensions that are symmetric with the restraint mechanism and tends to generate astigmatism instead of other Zernike terms.

The residual error was calculated by mathematically removing the Zernike terms up to and including $Z(2,2)$ from the original wavefront. The two-dimensional plots of the residual error are shown in Fig. 12. There is some evidence of systematic changes in the residual error as the fluid levels are changed; however, a large proportion of the error terms appears to behave randomly across the dynamic range of the tests. However, the total error terms produce negligible degradation of the vision quality as discussed later in the VSOTF section.

Image quality of the astigmatic lens was tested using the model eye. When testing the fluidic lens with a rectangular aperture, a cylindrical ophthalmic trial lens was placed in the optical path to simulate an aberration of the model eye. An image of Siemen's star is shown in Fig. 13 when the fluidic lens was adjusted for zero power. Next, the fluidic lens was adjusted to correct for the astigmatism of the model eye. When adjusting the model eye a defocus component is also generated, so a second trial lens was used to correct the spherical component generated

by the fluidic lens. In practice this spherical component could be generated with a second fluidic lens with a circular aperture.

The VSOTF was calculated for the astigmatic lens using the optical profile and is plotted as a function of cylinder (Fig. 14). The value of the axis term changes by approximately 90° as the fluid volume is varied and both values of cylinder are plotted. An extremely high value of optical quality of the lens is seen since the VSOTF does not drop below 0.98. The high optical quality of the astigmatic lens is attributed to the low residual error and the low values of the higher order Zernike terms.

4. Conclusion

In summary we have presented fluidic lenses that are capable of generating spherical and astigmatic Zernike wavefronts. The first type of lens uses a circular retainer for an elastic membrane and is capable of producing only defocus ($Z(2,0)$) alterations in the wavefront. The second type of lens uses a rectangular aperture and is capable of consistently and predictably generating second order ($Z(2, -2)$, $Z(2,0)$, and $Z(2,2)$) Zernike wavefronts. Both wavefronts were used with a model eye to demonstrate image correction that is suitable for ophthalmic applications. The lens design is small and compact, which potentially eliminates the need for mechanically moving or swapping lenses in a phoropter.

Acknowledgments

The authors acknowledge support from the Technology Research Infrastructure Fund Photonics program at the University of Arizona.

References

1. Kennedy GT, Paterson C. Correcting the ocular aberrations of a healthy adult population using microelectro-mechanical (MEMS) deformable mirrors. *Opt. Commun* 2007;271:278–284.
2. Liang J, Williams DR, Miller DT. Supernormal vision and high-resolution retinal imaging through adaptive optics. *J. Opt. Soc. Am. A* 1997;14:2884–2892.
3. Love GD. Wave-front correction and production of Zernike modes with a liquid-crystal spatial light modulator. *Appl. Opt* 1997;36:1517–1524. [PubMed: 18250829]
4. Schmidt JD, Goda ME, Duncan BD. Aberration production using a high-resolution liquid-crystal spatial light modulator. *Appl. Opt* 2007;46:2423–2433. [PubMed: 17429453]
5. Wang X, Wang B, Pouch J, Miranda F, Anderson JE, Bos PJ. Performance evaluation of a liquid-crystal-on-silicon spatial light modulator. *Opt. Eng* 2004;43:2769–2774.
6. Bagwell BE, Wick DV, Batchko R, Mansell JD, Martiez T, Terstaino SR, Payne DM, Harriman J, Serati S, Sharp G, Schwiegerling J. Liquid crystal based active optics. *Proc. SPIE* 2006;6289:628908.
7. Bagwell, BE.; Wick, DV.; Schwiegerling, J. Multi-spectral foveated imaging system. 2006 IEEE Aerospace Conference (IEEE; 2006;
8. Wick DV, Martinez T, Restaino SR, Stone BR. Foveated imaging demonstration. *Opt. Express* 2002;10:60–65. [PubMed: 19424331]
9. Wick DV, Martinez T. Adaptive optical zoom. *Opt. Eng* 2004;43:8–9.
10. Gunasekaran RA, Agarwal M, Singh S, Dubasi P, Coane P, Varahramyan K. Design and fabrication of fluidic controlled dynamic optical lens system. *Opt. Lasers Eng* 2005;43:686–703.
11. Zhang D-Y, Justis N, Lien V, Berdichevsky Y, Lo Y-H. High-performance fluidic adaptive lenses. *Appl. Opt* 2004;43:783–787. [PubMed: 14960070]
12. Tsai FS, Cho SH, Lo Y-H, Vasko B, Vasko J. Miniaturized universal imaging device using fluidic lens. *Opt. Lett* 2008;33:291–293. [PubMed: 18246158]
13. Zhang D-Y, Justis N, Lo Y-H. Fluidic adaptive zoom lens with high zoom ratio and widely tunable field of view. *Opt. Commun* 2005;249:175–182.

14. Sinzinger, S.; Jahns, J. *Microoptics*. Wiley-VCH; 1999. p. 86-87.
15. Thiobos L, Hong X, Bradley A, Applegate RA. Accuracy and precision of objective refraction from wavefront aberration. *J. Vision* 2004;4:329–351.
16. Westland S, Owens H, Cheung V, Paterson-Stephens I. Model of luminance contrast-sensitivity function for application to image assessment. *Color Res. Appl* 2006;31:315–319.

Fig. 1. Photograph of the fluidic lens. The larger syringe is used to fill the lens, while the smaller syringe is used to fine tune the optical properties of the lens.

Fig. 2.

Schematic diagram of the imaging system, which is composed of a model eye and the fluidic lens. An ophthalmic correction lens is shown before the fluidic lens and is used to correct the spherical portion of the optical power for the astigmatic lens. The details of the system are discussed in the paper.

Fig. 3.
Model diagram for the refractive nature of the membrane with changes in fluidic volume.

Fig. 4. Graph of the optical power as a function of the injected fluidic volume. The experimental points are plotted on the diagram, while the solid line corresponds to the simple theoretical model.

Fig. 5. Measured wavefront for the circular retaining ring is plotted for two orthogonal directions. The deflection was measured using an optical profiler.

Fig. 6.

Zernike coefficients were extracted from the Shack-Hartman measurements of the fluidic lens with the circular aperture. The inset has an expanded scale for the higher order aberrations.

Fig. 7.

Two-dimensional plots of the residual error after subtracting the Zernike polynomials from the wavefront determined from Shack–Hartman measurements. A 6 mm pupil size was used for the images. The distribution progresses with optical power and is attributed to artifacts of the membrane restraint.

Fig. 8. Images of a Siemens star were taken with the model eye. A 1 diopter ophthalmic trial lens was used to generate defocus in the model eye. (a) The uncorrected images and (b) the image corrected with the fluidic lens are shown. Likewise, (c) is the uncorrected image for 2 diopters of defocus, and (d) is the image is corrected by the fluidic lens.

Fig. 9. VSOTF as a function of the optical power for a fluidic lens with a circular restraining aperture.

Fig. 10. Optical powers produced by the astigmatic fluid lens as the fluid volume is changed. (a) Spherical power and (b) cylindrical power. The cylinder is given in the ophthalmic convention where the axis varies between 0° and 180° .

Fig. 11.

Zernike wavefronts extracted from Shack–Hartman measurements for different fluid volumes. The defocus component, $Z(2,0)$, increases in almost equal proportions to the astigmatic component, $Z(2,2)$. The inset shows very small values for the higher order Zernike terms.

Fig. 12. Residual error terms produced after subtracting the Zernike terms from the measured wavefront. A 6 mm pupil size was used for calculation of the Zernike terms. The values of the cylinder and axis are also shown for each plot.

Fig. 13. Image taken with a model eye. (a) An ophthalmic trial lens of 2D cylinder was used to produce the astigmatism in the model eye. (b) The fluidic lens was adjusted to correct the astigmatism in the fluidic eye. A second ophthalmic trial lens was used to remove the defocus generated by the fluidic lens.

Fig. 14. Measured VSOTF for the astigmatic lens. The high values of the VSOTF are indicative of the high optical quality of the lens.