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Dynamic Simulation of the Effect of Soft Toric Contact Lenses Movement on Retinal Image Quality

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Abstract

Purpose—To report the development of a tool designed to dynamically simulate the effect of soft toric contact lens movement on retinal image quality, initial findings on three eyes, and the next steps to be taken to improve the utility of the tool.

Methods—Three eyes of two subjects wearing soft toric contact lenses were cyclopleged with 1% cyclopentolate and 2.5% phenylephrine. Four hundred wavefront aberration measurements over a 5-mm pupil were recorded during soft contact lens wear at 30 Hz using a complete ophthalmic analysis system aberrometer. Each wavefront error measurement was input into Visual Optics Laboratory (version 7.15, Sarver and Associates, Inc.) to generate a retinal simulation of a high contrast log MAR visual acuity chart. The individual simulations were combined into a single dynamic movie using a custom MatLab PsychToolbox program. Visual acuity was measured for each eye reading the movie with best cycloplegic spectacle correction through a 3-mm artificial pupil to minimize the influence of the eyes' uncorrected aberrations. Comparison of the simulated acuity was made to values recorded while the subject read unaberrated charts with contact lenses through a 5-mm artificial pupil.

Results—For one study eye, average acuity was the same as the natural contact lens viewing condition. For the other two study eyes visual acuity of the best simulation was more than one line worse than natural viewing conditions.

Conclusions—Dynamic simulation of retinal image quality, although not yet perfect, is a promising technique for visually illustrating the optical effects on image quality because of the movements of alignment-sensitive corrections.

Keywords

dynamic simulation; soft toric contact lens; retinal image quality; wavefront error; visual acuity

Contact lenses are a commonly used method to correct defocus and astigmatism in the human eye. A toric contact lens has both a spherical and cylindrical component, and is often used to correct patients with astigmatism higher than 0.75 D. Correct alignment of the astigmatic correction with the astigmatic error of the eye is critical for optimal optical and visual performance. With the development of wavefront measurement technology, wavefront-guided customized soft contact lenses has become an area of active research with the aim to correct defocus and astigmatism (lower order aberrations) and higher order aberrations which cannot be corrected by traditional spectacles or contact lenses. Like astigmatic contact lenses, proper alignment is critical for optimal optical and visual performance with the tolerance for misalignment being dependent both on the magnitude and type of aberrations present.

Although astigmatic contact lenses particularly benefit normal eyes with astigmatic errors, wavefront-guided soft customized lenses will particularly benefit those patients with large amounts of higher order aberrations (e.g., keratoconic patients).^{1,2} Results from several studies have shown that lens misalignment (in the form of rotations and translations) can markedly reduce the visual benefit of astigmatic contact lenses and wavefront-guided customized contact lenses. In fact, large amounts of custom lens movement can reduce visual performance as compared with no correction at all.^{3–5}

Guirao et al.⁴ modeling the effects of contact lens translation and rotation concluded that for typical lens decentrations, wavefront-guided corrections could offer a benefit of two to four times (1.5 to 2 times in white light) measured by the modulation transfer function (MTF) as compared with standard correction of only defocus and astigmatism. On the assumption that corneal first surface aberration represented the vast majority of ocular aberrations, Brabander et al. used corneal topography data from keratoconic eyes to model ocular aberrations.³ Similar to Guirao et al., they found that mismatches of the ocular and wavefront-guided contact lens (WGCL) correction translation errors should be smaller than 0.5 mm for a customized soft contact lens correction and rotation up to 10 degrees appeared less critical than translation errors.³

For both the Guirao et al.⁴ and Brabander et al.³ studies, the MTF or metrics calculated from the MTF were used to represent visual performance. The MTF is an objective metric which reveals, by calculation, the contrast loss of the image as a function of spatial frequency. It cannot tell the investigator how a subject perceives a target of interest (e.g., a visual acuity chart).

Applegate's group^{6,7} has been examining the perceptual changes associated with optical aberration by producing optically aberrated high contrast log MAR visual acuity charts for individual aberrations and combinations of aberrations which are read by subjects with best subjective refraction correction. This experimental approach was capitalized on by Menjivar et al.⁸ to determine how the movement of a WGCL influences visual performance of a moderate keratoconic eye when correcting the eye's high and low order aberrations through the fifth radial order. The correction was allowed to translate -0.6 mm to $+0.6$ mm in 0.2 mm steps and rotate -6 degrees to $+6$ degrees in 2 degrees steps. Menjivar et al.⁸ found that simulated WGCL provided improved performance compared with habitual rigid gas permeable correction when translation was 0.2 mm or less and rotation was 6 degrees or less.

These previous studies demonstrate how critical lens stabilization is for eyes with significant ocular aberrations beyond sphere and demonstrate useful methods to conduct simulation studies. However, these studies do not reveal or illustrate how vision is affected as the lens moves from one position to another in a continuous, dynamic fashion.

Here we report the development of a tool designed to dynamically simulate the effect of soft toric contact lens movement on retinal image quality, initial findings on three eyes, and the next steps to be taken to improve the utility of the tool.

METHODS

Subjects

Two subjects with good systemic and ocular health were recruited. Both subjects signed a University of Houston Institutional Review Board-approved informed consent. The left eye of subject 1 and both eyes of subject 2 were studied. All study eyes had at least 1.50 D of astigmatism and achieved soft toric contact lens visual acuities of $20/20$ or better and were otherwise healthy. The right eye of subject 1 did not meet the enrollment criteria and was not

included in the study. Each study eye was fitted with a soft toric contact lens. The left eye of subject 1 was fitted with the Metrosoft Toric lens (Metro Optics, Austin, TX) and both eyes of 2 were fit with the SofLens 66 Toric contact lens (Bausch and Lomb, Rochester, NY).

Simulation Strategies

Five simulation strategies of interest, identified by (A) to (E) below, were defined using wavefront error (WFE) data. Later strategies incorporate the advantages of earlier strategies so that the final simulation (E) represents the most sophisticated simulation and simulation level (A) the most basic. Ocular WFE is defined as the difference between the actual wavefront from the ideal point by point across the entrance pupil of the eye (American National Standards Institute standard for ophthalmic wavefront sensing Z80.28).

Simulation A, 5 mm WFE—Simulations were made for subjects based on the WFE measurement wearing their soft toric contact lens correction. The WFE measured every 30th of a second was used to calculate the optical transfer function, which was in turn used in the generation of each frame of a simulated chart movie (Fig. 1).

Simulation B, 5 mm WFE and Stabilized—The simulation described in simulation A was improved upon by accounting for slight variations in aberrations that cause the image to have a slightly different location from frame to frame. In the real world, objects fixed in space (e.g., a clear static acuity chart used in the clinic) do not perceptually jitter. Image jitter in the movie was reduced by cross-correlating the image position in each frame of the image simulation with the original chart serving as a reference.

Simulation C, 5 mm WFE, Stabilized and Temporal Noise reduced—The simulation described in simulation B was improved upon by reducing the temporal noise in the images. WFE was measured at a temporal frequency of 30 Hz. On analysis, this sampling rate turned out to be much higher than the temporal WFE variations found in the study eyes. Thus, to eliminate the possibility of temporal noise not present in the study eyes degrading the simulation, WFE measurements were filtered.⁹ Temporal noise reduction was accomplished by filtering each WFE measurement above 2 Hz with a MatLab program provided by Dr. Heidi Hofer.

Simulation D, 5 mm WFE, Stabilized, Noise Reduced, and Preemphasized for Diffraction Limited 3 mm Pupil—To minimize the effect of viewing the simulations with an aberrated eye, the subjects read the dynamic simulations through a 3-mm artificial pupil. Although a 3-mm pupil is likely to provide the subject with their best spatial vision,¹⁰ it reduces image contrast because of diffraction effects that are not present when viewing through a 5-mm pupil. Simulation C was improved upon by preemphasizing each frame to account for the effect of contrast loss because of the diffraction effects of viewing the simulations through a 3-mm artificial pupil. The preemphasis was accomplished by dividing each measured 5 mm MTF by the diffraction limited 3 mm MTF.

Simulation E, 5 mm WFE, Stabilized, Noise Reduced, Preemphasized for Subjects' Diffraction and Aberrations Over a 3-mm Pupil—Simulation D was improved upon by preemphasizing each frame to account for the effect of contrast loss because of the aberration structure of the eye viewing the simulations through a 3-mm artificial pupil.

A 702FP flat panel color LCD monitor (Dell, Austin, TX) was used to display all acuity charts and movies. The physical size of the monitor was 33.9×27.0 cm (1280×1024 pixels). The test distance from the monitor to the eye was 362.4 cm and was calculated to provide adequate screen resolution to display the smallest desired letter (20/10 letter). Thus, the display screen

subtended $5.3 \text{ degrees} \times 4.3 \text{ degrees}$ at the eye. Gamma correction was applied to correct for the non-linear relationship between input pixel value and display luminance.

Experimental Steps

Step 1—To make the subjects familiar with the experimental procedure, we performed preexperimental training until each subject understood the procedure and was comfortable executing the procedure. The order of letters on the charts used in the training was different from those used in the experiment.

Step 2—Subjects were dilated and cyclopleged with 1% cyclopentolate and 2.5% phenylephrine. Thirty minutes after application of the dilating drops, WFE was measured with the subject wearing their best fitted soft toric lens using the multibuffer mode in the COAS aberrometer (Wavefront Sciences, Albuquerque, NM). The multibuffer mode allows measurement of 30 frames/sec for up to 400 measurements (>13 sec). After data acquisition, the measured WFE were fit with a Zernike expansion through the tenth radial order over a 5-mm pupil.

Step 3—Not all image acquisitions yielded WFE measurements. The major cause for a defective measurement was a partial Shack/Hartmann image due to a blink. Before generating the retinal image movie, each data set was cleansed of defective measurements.

Step 4—*Movie frame generation*: Each acceptable WFE measured from step 2 was imported into Visual Optics Laboratory (version 7.15, Sarver and associates, Inc.) in a batch file operation to generate each frame for the dynamic retinal image simulation.

Step 5—We devised our simulations in MatLab (version 7.1, MathWorks, Natick, MA), using the Psychophysics Toolbox extensions.^{11,12} Movie frames for each simulation of interest generated in step 4 were imported into a custom MatLab psychtool box program to produce the continuously looping movies. Three different movies (each using a different ordering of the letters in the eye chart) were produced for each test eye and each simulation conditions A–E.

Step 6—Each subject read each of the three dynamic retinal image simulations for each simulation level of interest through a 3-mm artificial pupil with best cycloplegic spectacle correction. Visual acuity was recorded until the subject missed a total of five letters.

Step 7—As a comparison, normal visual acuity (a clear high contrast log MAR acuity chart) was measured for three different acuity charts through a 5-mm artificial pupil with the best fitting soft toric lens and the best fitting soft toric lens plus an overcorrection. For discussion purposes the best simulation will be the simulation that produces a measured acuity most similar to the measured acuity wearing their best fitting toric lens.

Step 8—Log MAR visual acuity in each simulation level was plotted as bar graphs and compared with those in natural viewing conditions (in step 7). The simulation which most closely mimicked the natural reading condition was deemed the best simulation.

RESULTS

After compensating for the minification effect of using best cycloplegic spectacle correction when subjects read simulated chart movies,¹³ log MAR visual acuity was plotted for all three test eyes under each experimental condition.

Fig. 2 shows the visual acuity result for the left eye (OS) of subject 1. The best simulation was simulation E. Average acuity for this simulation level of interest was nine letters worse than the natural contact lens viewing condition, and 6.5 letters better than the most basic simulation (5 mm WFE).

For subject 2's right eye (OD) all simulations were within two letters of either of the free-viewing contact lens wear conditions (Fig. 3). Test-retest precision is 2 letters. Simulation level E for this eye produced an artificial ringing at the letter edges. As a result, data for simulation E although similar to simulation D are not reported. This ringing is presumably attributed to the Gibbs phenomenon. The Gibbs phenomena commonly referred to as ringing artifacts in a Fourier analysis results from trying to fit discontinuous functions (e.g., such as a sharp edge of a letter) with a partial sum of continuous functions.^{14,15}

For subject 2's left eye (OS) the best simulation was simulation C (Fig. 4). Average acuity was 6.5 letters worse than the natural contact lens viewing condition. There was little difference in visual acuity among four simulation levels of interest. As in the right eye of subject 2 simulation E produced an artificial ringing at the letter edges in the simulation. As a result, data for simulation E is not reported.

DISCUSSION

Clinically, visual acuity is the most popular method for measuring visual performance. The measurement of visual acuity relies on the response of the patient and is relatively insensitive to a wide variation in retinal image quality (Fig. 5). When measuring visual acuity, clinicians have no way of visualizing what the patient is seeing, particularly when the patient is wearing an astigmatic or WGCL correction. Retinal image quality can vary as the alignment varies between the optical errors of the eye and the contact lens designed to correct these errors.

As shown in Fig. 5, these two subjects can both read 20/20⁺², but it is obvious that the retinal image of the patient without complaints (left panel) has a retinal image with higher fidelity than the patient with complaints (right panel). A simulation displaying the dynamic variation in retinal image quality will allow the clinical and basic researcher to visualize for themselves changes in retinal image quality with contact lens movement.

In this investigation, subject 2's right eye simulated visual performance most closely matched actual visual performance for simulation level A (the 5 mm WFE, stabilized, noise reduced, preemphasized for diffraction limited 3 mm pupil simulation). All simulations were essentially the same and within the two letter average standard deviation of the measurement. The two letter average standard deviation of the measurement is essentially identical to the average standard deviation of the measurement for static simulation experiments across subjects reported previously^{6,7} and for the measurement of log MAR acuity in general.¹⁶ However, for the other two test eyes (left eyes of subjects 1 and 2), average visual acuity in the contact lens simulation was more than one line worse than the natural viewing condition. Possible explanations for the discrepancies between eyes include:

1. Natural viewing visual acuity was not measured at the same time as the WFE. Consequently the natural viewing and simulated conditions may not be directly comparable. To illustrate, consider the possibility that during the WFE measurements the lens was misaligned but was not during the natural viewing condition. An indication that this may have in fact happened was found when questioning subject 2 after data analysis was complete. Subject 2 commented that during WFE measurement the red probe beam appeared blurry and smeared out. Typically, the probe beam appears reasonably clear once the Badal assembly of the complete ophthalmic analysis system has adjusted for spherical errors. A probe beam that appears blurry or smeared

out is likely due to the increased aberration caused by lens movement, tear breakup, etc. Lens misalignment may also occur when visual acuity was measured under the natural contact lens viewing conditions. Unlike the WFE measurement condition, it is obvious to the subjects that the acuity chart is blurry and they can reduce the transient error by blinking to realign the lens. However, if the transient blurriness is unknowingly captured in the WFE measurement, the adverse effects would be built into the all simulations leading to blurry simulated acuity movies.

2. In the simulated movies based on 5mmWFE (prism terms are set to zero), the location of each frame changes slightly from one another, which makes the movies look like they are moving. To reduce the effect of slightly different location from frame to frame (even when prism terms are set to zero), stabilization was used in four simulation levels (B, C, D, and E) by cross-correlating image position in each frame of the image simulation with the original chart serving as a reference. Even with the stabilization procedure, simulated images still jittered slightly. In natural viewing conditions the chart does not jitter.

Given the variability of the results in the three test eyes the dynamic simulation has not been completely optimized to routinely provide results equivalent to the free viewing condition. Nonetheless the better simulations visually illustrates to well-sighted observers how contact lens movements dynamically influence retinal image quality. It is anticipated that this promising dynamic simulation method once perfected will be as effective as well as an objective method to examine patient complaints and determine through modeling just how stable an individual correction has to be to increase performance over standard lens designs.

This first use of the dynamic simulation technique points to different steps to improve both the simulation and the methods used to validate the simulations. During validation, measurements of the WFE for generating the simulations need to be taken simultaneously with the natural viewing visual performance measurements to eliminate the possibility that WFE measurements do not reflect the actual WFE at the time of the visual performance measurement. Secondly, as the technique becomes more refined to improve the sensitivity of the validation procedures, a more sensitive metric of visual performance such as low contrast low luminance visual acuity should be used in place of high contrast acuity.¹⁷

To improve the dynamic simulation itself four steps will be taken. First, a thorough focus set of simulations will be made to determine if the wavefront sensor properly defined defocus. The reason being it is assumed in Shack/Hartmann wavefront sensing that the reflection of the probe beam takes place in the plane of the entrance aperture of the photoreceptors. A displacement of this reflection by as little as 0.08 mm can cause 0.25 D change in defocus, which in turn has a large effect on simulated image quality. A 0.25D defocus error is equivalent to 0.325 μm of WFE over a 6-mm pupil, which has been shown to decrease high contrast acuity by approximately 14 letters.⁶ A 14-letter loss is greater than the difference between the free viewing and simulated viewing conditions for all three of the simulations reported here by a factor of 2. Second, the efficiency of simulation generation can be increased by a factor of 6 by performing wavefront sensing at 5 Hz as opposed to 30 Hz with essentially no loss in simulation fidelity. Third, WFE measurements obtained with the complete ophthalmic analysis system are monochromatic in nature; real viewing is polychromatic. Generating the simulations using a polychromatic point spread function weighed by the spectral luminosity function will improve simulation accuracy and allows research teams to explore the importance of ocular chromatic aberration. Finally, artificial ringing because of the Gibbs phenomena can be reduced by carefully selecting the range of spatial frequencies used to generate the simulations.

CONCLUSION

Dynamic simulation of retinal image quality is a promising technique for visually illustrating the optical effects on retinal image quality because of movements of alignment-sensitive corrections.

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FIGURE 1.

A road map for image creation. Visual Optics Laboratory (VOL) software (Sarver and Associates, Carbondale, IL) was used to generate each chart image used in the simulation movie. VOL calculates the image of any given object using the frequency domain path.

FIGURE 2.

Average soft toric contact lens log MAR visual acuity data for the left eye of subject 1 for each experimental condition after compensating for the minification effect caused by using best cycloplegic spectacle correction when subject 1 read simulated chart movies.

FIGURE 3.

Average soft toric contact lens log MAR visual acuity data for the right eye of subject 2 for each experimental condition after compensating for the minification effect caused by using best cycloplegic spectacle correction when subject 2 read simulated chart movies.

FIGURE 4.

Average soft toric contact lens log MAR visual acuity data for the left eye of subject 2 for each experimental condition after compensating for the minification effect caused by using best cycloplegic spectacle correction when subject 2 read simulated chart movies.

Both patients have better than 20/20 acuity

FIGURE 5.

Two different simulated retinal images with the same visual acuity of 20/20⁺² (log MAR – 0.04).