VERGENCE WEIGHTING SYSTEMS AND METHODS FOR TREATMENT OF PRESBYOPIA AND OTHER VISUAL CONDITIONS

FIG. 12A
VERGENCE WEIGHTING SYSTEMS AND METHODS FOR TREATMENT OF PRESBYOPIA AND OTHER VISION CONDITIONS

CROSS-REFERENCES TO RELATED APPLICATIONS

[0001] This application claims the benefit of and priority to U.S. Provisional Patent Application 62/101,436 filed January 9, 2015, the contents of which are incorporated herein by reference in its entirety. Full Paris Convention priority is hereby expressly reserved.


BACKGROUND OF THE INVENTION

[0003] Embodiments of the present invention relate generally to goal functions or visual function diagnostic metrics, and particular embodiments provide methods, devices, and systems for mitigating or treating vision conditions such as presbyopia, often by determining a treatment shape based on selected weighting values for certain viewing distances.

[0004] Presbyopia normally develops as a person ages, and is associated with a natural progressive loss of accommodation, sometimes referred to as "old sight." The presbyopic eye often loses the ability to rapidly and easily refocus on objects at varying distances. There may also be a loss in the ability to focus on objects at near distances. Although the condition progresses over the lifetime of an individual, the effects of presbyopia usually become noticeable after the age of 45 years. By the age of 65 years, the crystalline lens has often lost almost all elastic properties and has only limited ability to change shape. Residual accommodation refers to the amount of accommodation that remains in the eye. A lower degree of residual accommodation contributes to more severe presbyopia, whereas a higher amount of residual accommodation correlates with less severe presbyopia.

[0005] Known methods and devices for treating presbyopia seek to provide vision approaching that of an emmetropic eye. In an emmetropic eye, both distant objects and near objects can be seen due to the accommodation properties of the eye. To address the vision problems associated with presbyopia, reading glasses have traditionally been used by individuals to add plus power dioptr to the eye, thus allowing the eye to focus on near objects and maintain a clear image. This approach is similar to that of treating hyperopia, or farsightedness.

[0006] Presbyopia has also been treated with bi-focal eyeglasses, where one portion of the lens is corrected for distance vision, and another portion of the lens is corrected for near vision. When
peering down through the bifocals, the individual looks through the portion of the lens corrected for near vision. When viewing distant objects, the individual looks higher, through the portion of the bi-focals corrected for distance vision. Thus with little or no accommodation, the individual can see both far and near objects.

5 [0007] Contact lenses and intra-ocular lenses (IOLs) have also been used to treat presbyopia. One approach is to provide the individual with monovision, where one eye (usually the primary eye) is corrected for distance-vision, while the other eye is corrected for near-vision. Unfortunately, with monovision the individual may not clearly see objects that are intermediately positioned because the object is out-of-focus for both eyes. Also, an individual may have trouble seeing with only one eye, or may be unable to tolerate an imbalance between their eyes. In addition to monovision, other approaches include bilateral correction with either bi-focal or multi-focal lenses. In the case of bi-focal lenses, the lens is made so that both a distant point and a near point can be focused. In the multi-focal case, there exist many focal points between near targets and far targets.

10 [0008] Surgical treatments have also been proposed for presbyopia. Anterior sclerostomy involves a surgical incision into the sclera that enlarges the ciliary space and facilitates movement of the lens. Also, scleral expansion bands (SEBs) have been suggested for increasing the ciliary space. Problems remain with such techniques, however, such as inconsistent and unpredictable outcomes.

15 [0009] In the field of refractive surgery, certain ablation profiles have been suggested to treat the condition, often with the goal of increasing the range of focus of the eye, as opposed to restoring accommodation in the patient’s eye. Many of these ablation profiles can provide a single excellent focus of the eye, yet they do not provide an increased depth of focus such that optimal distance acuity, optimal near acuity, and acceptable intermediate acuity occur simultaneously. Shapes have been proposed for providing enhanced distance and near vision, yet current approaches do not provide ideal results for all patients.

[0010] To evaluate the effectiveness of a refractive correction, such as with a spectacle lens, contact lens, intra-ocular lens, or laser refractive surgery procedure, it may be desirable to consider a merit function, or gauge of optical quality, that can determine such effectiveness. Gauges of optical quality are discussed in copending patent application numbers 60/431,634, filed December 6, 2002, 60/468,303, filed May 5, 2003, and 10/738,358 filed December 5, 2003, the disclosures of which are hereby incorporated by reference. Merit functions may be used in evaluating post-corrective measurements, and in predicting the effect or outcome of a proposed corrective
procedure. While the merit function may be objective, it may also desirable that the merit function have a good correlation with subjective test results such as visual acuity, contrast acuity, and the like. The following optical metrics can be or have been used as possible optical metrics or merit functions: high order (HO) root mean square (RMS) error; Strehl ratio; modulation transfer function (MTF) at specific spatial frequencies; volume under MTF surface up to a certain spatial frequency; compound MTF; encircled energy; and wavefront refractions. Other goal functions or visual function diagnostic metrics are available for characterizing lenses and other optical systems, including visual acuity such as logMAR, refractive error such as sphere and cylinder, and contrast sensitivity (CS). However, many of the currently used goal functions are difficult and cumbersome to implement with current clinical methods, and are insufficient in utilizing currently available clinical data and in providing guidance to the administration and diagnosis of reported visual difficulties.

[0011] In light of the above, it would be desirable to have improved methods, devices, and systems for treatment and/or mitigation of optical defects, based on improved goal functions such as a compound modulation transfer function. The goal functions should be easily implemented with existing clinical data, and with clinical data that is currently being generated by present measurement techniques. Optionally, it would be desirable to have improved methods, devices, and systems for treatment and/or mitigation of presbyopia and other optical defects. It may be desirable to provide improved prescriptions in the form of practical customized or optimized prescription shapes for treating or mitigating vision conditions such as presbyopia in a particular patient.

BRIEF SUMMARY OF THE INVENTION

[0012] Embodiments of the present invention encompass systems and methods for determining optimizer values that involve factoring in the variation of the optical metric over a range of testing points, or vergence, to account for the distance vision, intermediate vision, and near vision. In some cases, various weighting protocols can be implemented to that assign different weighting values for different viewing distances. Such techniques can be used in various types of treatment modalities, including without limitation refractive surgery, contact lenses, intraocular lenses, spectacle lenses, and other vision correction approaches such as inlays, conductive keratoplasty, and the like.

[0013] In some cases, systems or methods as disclosed herein can be used in conjunction with therapeutic protocols that involve providing a patient with a presbyopia treatment if the residual accommodation of the eye exceeds the threshold residual accommodation calculated for the eye,
for example as discussed in U.S. Patent No. 7,762,668, the content of which is incorporated herein by reference. In some cases, systems or methods as disclosed herein can be used in conjunction with multifocal therapeutic protocols for presbyopia. In some cases, systems or methods as disclosed herein can be used for treating pre-presbyopic patients. In some cases, the vergence weighting protocols disclosed herein can be used on conjunction with vision treatment approaches such as those described in U.S. Patent Application No. 10/872,331 filed June 17, 2004, U.S. Patent Application No. 11/156257 filed June 17, 2005, U.S. Patent Application No. 12/126,185 filed May 23, 2008, and U.S. Provisional Patent Application No. 62/035,874 filed August 11, 2014, the contents of each of which are incorporated herein by reference.

[0014] Embodiments of the present invention provide devices, systems, and methods that use improved goal functions for mitigating or treating vision conditions in a patient. The goal function can reflect optical quality throughout a vergence range. The goal function may also comprise a ratio of an optical parameter of the eye with a diffraction theory parameter. Relatedly, the goal function may also comprise at least one parameter selected from the group consisting of Strehl Ratio (SR), modulation transfer function (MTF), point spread function (PSF), encircled energy (EE), MTF volume or volume under MTF surface (MTFV), compound modulation transfer function (CMTF), and contrast sensitivity (CS).

[0015] In some instances, these techniques can be carried out in conjunction with treatments provided by any of a variety of laser devices, including without limitation the WaveScan® System and the STAR S4® Excimer Laser System both by Abbott Medical Optics Inc., the WaveLight® Allegretto Wave® Eye-Q laser, the Schwind Amaris™ lasers, the 217P excimer workstation by Technolas PerfectVision GmbH, the Mel 80™ laser by Carl Zeiss Meditec, Inc., and the like. In some cases, embodiments provide techniques for using laser basis data during refractive surgery treatment procedures which can be implemented in such laser devices.

[0016] In one aspect, embodiments of the present invention encompass systems and methods for treating a vision condition of an eye in a particular patient. Exemplary methods may include receiving a vision requirements specification selected for the particular patient, where the vision requirements specification includes a first weighting value for a first viewing distance within a vergence range and a second weighting value for a second viewing distance within the vergence range, and determining an optical surface shape for the particular patient. The optical surface shape can be based on the vision requirements specification and an optical metric. Methods can also include treating the vision condition of the eye of the particular patient by providing a treatment to the patient, where the treatment includes or is based on a shape that corresponds to the
optical surface shape. In some cases, the first viewing distance is a near vision viewing distance, an intermediate vision viewing distance, or a distance (far) vision viewing distance. In some cases, the second viewing distance is a near vision viewing distance, an intermediate vision viewing distance, or a distance (far) vision viewing distance. In some cases, the first weighting value is different from the second weighting value and the first viewing distance is different from the second viewing distance. In some cases, the first weighting value is greater than the second weighting value. In some cases, the first weighting value is less than the second weighting value. In some cases, the first viewing distance is greater than the second viewing distance. In some cases, the first viewing distance is less than the second viewing distance. According to some embodiments, the optical metric is a composite optical metric. In some cases, the optical metric includes a compound modulation transfer function (CMTF) parameter having a combination of modulation transfer functions (MTF's) at a plurality of distinct frequencies. In some cases, the first and second weighting values are members of a weighting value distribution that is linear across a vergence range that includes the first and second viewing distances. In some cases, the first and second weighting values are members of a weighting value distribution that is non-linear across a vergence range that includes the first and second viewing distances. According to some embodiments, a step of treating the vision condition of the eye of the particular patient can include a procedure such as abrating a cornea of the eye of the particular patient to provide a corneal surface shape that corresponds to the optical surface shape, providing the particular patient with a contact lens or a spectacle lens having a shape that corresponds to the optical surface shape, or providing the particular patient with an intra-ocular lens having a shape that corresponds to the optical surface shape.

[0017] In another aspect, embodiments of the present invention encompass systems and methods for generating an optical surface shape for use in treating a vision condition of an eye in a particular patient. Exemplary methods can include receiving a vision requirements specification selected for the particular patient, where the vision requirements specification includes a first weighting value for a first viewing distance within a vergence range and a second weighting value for a second viewing distance within the vergence range. Further, methods can include generating the optical surface shape for the particular patient, where the optical surface shape is based on the vision requirements specification and an optical metric. In some cases, methods may also include determining a procedure for treating the vision condition of the eye of the particular patient based on the optical surface shape. In some cases, the procedure can include ablating a corneal surface of the eye of the particular patient to provide a corneal surface shape that corresponds to the optical surface shape, providing the particular patient with a contact lens or a spectacle lens having
a shape that corresponds to the optical surface shape, or providing the particular patient with an intra-ocular lens having a shape that corresponds to the optical surface shape. In some cases, the first viewing distance is a near vision viewing distance, an intermediate vision viewing distance, or a distance vision viewing distance. In some cases, the second viewing distance is a near vision viewing distance, an intermediate vision viewing distance, or a distance vision viewing distance. In some cases, the first weighting value is different from the second weighting value and the first viewing distance is different from the second viewing distance. In some cases, the first weighting value is greater than the second weighting value. In some cases, the first weighting value is less than the second weighting value. In some cases, the first viewing distance is greater than the second viewing distance. In some cases, the first viewing distance is less than the second viewing distance. According to some embodiments, the optical metric is a composite optical metric. In some cases, the optical metric includes a compound modulation transfer function (CMTF) parameter having a combination of modulation transfer functions (MTF's) at a plurality of distinct frequencies. In some cases, the first and second weighting values are members of a weighting value distribution that is linear across a vergence range that includes the first and second viewing distances. In some cases, the first and second weighting values are members of a weighting value distribution that is non-linear across a vergence range that includes the first and second viewing distances.

In still another aspect, embodiments of the present invention encompass systems and methods for establishing an optical surface shape for use in treating a vision condition of an eye in a particular patient. Exemplary systems can include an input that receives a vision requirements specification selected for the particular patient, where the vision requirements specification includes a first weighting value for a first viewing distance within a vergence range and a second weighting value for a second viewing distance within the vergence range. Systems can also include a data processing module having a processor and a tangible non-transitory computer readable medium, where the computer readable medium is programmed with a computer application that, when executed by the processor, causes the processor to establish the optical surface shape for the eye of the particular patient. The optical surface shape can be based on the vision requirements specification received by the input and an optical metric. In some cases, the computer application, when executed by the processor, causes the processor to determine a protocol for treating the vision condition of the eye of the particular patient based on the optical surface shape. In some cases, the protocol includes a photodisruption procedure for a corneal tissue of the eye of the particular patient, where the photodisruption procedure is configured to provide a corneal surface shape that corresponds to the optical surface shape. In some cases, the
protocol includes a contact lens or a spectacle lens procedure for the eye of the particular patient, where the contact lens or spectacle lens procedure involves a lens having a shape that corresponds to the optical surface shape. In some cases, the protocol includes an intra-ocular lens procedure for the eye of the particular patient, where the intra-ocular lens procedure involves a lens having a shape that corresponds to the optical surface shape. In some cases, the first viewing distance is a near vision viewing distance, an intermediate vision viewing distance, or a distance vision viewing distance. In some cases, the second viewing distance is a near vision viewing distance, an intermediate vision viewing distance, or a distance vision viewing distance. In some cases, the first weighting value is different from the second weighting value and the first viewing distance is different from the second viewing distance. In some cases, the first weighting value is greater than the second weighting value. In some cases, the first weighting value is less than the second weighting value. In some cases, the first viewing distance is greater than the second viewing distance. In some cases, the first viewing distance is less than the second viewing distance.

According to some embodiments, the optical metric is a composite optical metric. In some cases, the optical metric includes a compound modulation transfer function (CMTF) parameter having a combination of modulation transfer functions (MTF’s) at a plurality of distinct frequencies. In some cases, the first and second weighting values are members of a weighting value distribution that is linear across a vergence range that includes the first and second viewing distances. In some cases, the first and second weighting values are members of a weighting value distribution that is non-linear across a vergence range that includes the first and second viewing distances.

[0019] In yet another aspect, embodiments of the present invention encompass computer program products for generating an optical surface shape for use in treating a vision condition of an eye in a particular patient. In some cases, the computer program product is embodied on a tangible non-transitory computer readable medium and includes code for accessing a vision requirements specification selected for the particular patient. The vision requirements specification can include a first weighting value for a first viewing distance within a vergence range and a second weighting value for a second viewing distance within the vergence range. The computer program product can also include code for generating the optical surface shape for the particular patient, where the optical surface shape is based on the vision requirements specification and an optical metric. In some cases, a computer program product can also include code for determining a protocol for treating the vision condition of the eye of the particular patient based on the optical surface shape. According to some embodiments the protocol can include a photodisruption procedure for a corneal tissue of the eye of the particular patient. The photodisruption procedure can be configured to provide a corneal surface shape that corresponds
to the optical surface shape. According to some embodiments the protocol can include a contact lens or a spectacle lens procedure for the eye of the particular patient. A contact lens or spectacle lens procedure can involve a lens having a shape that corresponds to the optical surface shape. According to some embodiments the protocol can include an intra-ocular lens procedure for the eye of the particular patient. An intra-ocular lens procedure can involve a lens having a shape that corresponds to the optical surface shape. In some cases, the first viewing distance is a near vision viewing distance, an intermediate vision viewing distance, or a distance vision viewing distance. In some cases, the second viewing distance is a near vision viewing distance, an intermediate vision viewing distance, or a distance vision viewing distance. In some cases, the first weighting value is different from the second weighting value and the first viewing distance is different from the second viewing distance. In some cases, the first weighting value is greater than the second weighting value. In some cases, the first weighting value is less than the second weighting value. In some cases, the first viewing distance is greater than the second viewing distance. In some cases, the first viewing distance is less than the second viewing distance. In some cases, the optical metric is a composite optical metric. In some cases, the optical metric includes a compound modulation transfer function (CMTF) parameter having a combination of modulation transfer functions (MTF's) at a plurality of distinct frequencies. In some cases, the first and second weighting values are members of a weighting value distribution that is linear across a vergence range that includes the first and second viewing distances. In some cases, the first and second weighting values are members of a weighting value distribution that is non-linear across a vergence range that includes the first and second viewing distances.

[0020] For a fuller understanding of the nature and advantages of the present invention, reference should be had to the ensuing detailed description taken in conjunction with the accompanying drawings.

BRIEF DESCRIPTION OF THE DRAWINGS

[0021] Fig. 1 illustrates a laser ablation system according to an embodiment of the present invention.

[0022] Fig. 2 illustrates a simplified computer system according to an embodiment of the present invention.

[0023] Fig. 3 illustrates a wavefront measurement system according to an embodiment of the present invention.
Fig. 3A illustrates another wavefront measurement system according to an embodiment of the present invention.

Fig. 4A illustrates an example of the compound MTF (upper panel) versus its corresponding individual MTF curves at 15, 30, and 60 cycles per degree (lower panel).

Fig. 4B illustrates an example of the compound MTF (upper panel) versus its corresponding individual MTF curves at 10, 20, and 30 cycles per degree (lower panel).

Fig. 5 is a flow chart illustrating exemplary method steps for optimizing an optical prescription that treats or corrects a vision condition.

Fig. 6 illustrates a data flow process for shape optimization for correction or treatment of a vision condition.

Fig. 7 illustrates a comparison of Direction Set method and Downhill Simplex method.

Figs. 8A and 8B illustrate alternative prescriptions optimized for an eye of a particular patient, and their characteristics.

Fig. 8C illustrates a comparison of optimizer values using even-term polynomials and all power term polynomials for pupil sizes of 4mm, 5mm, and 6mm.

Figs. 9A-D show alternative presbyopia-mitigating prescriptions optimized for an eye of a particular patient.

Fig. 10 illustrates effects of random noise on prescriptions optimized for an eye of a particular patient.

Figs. 11A-C compare optimized prescriptions to alternative treatments for differing pupil sizes.

Figs. 12A-C compare optimized prescriptions to alternative treatments for a range of viewing distances.

Fig. 13 illustrates simulated viewing charts viewed at differing distances to compare optimized prescriptions to alternative treatments.

Figs. 14-16 illustrate graphical interface computer screen displays for a prescription optimizer and system.

Figs. 17 and 18 illustrate pupil sizes and changes at differing viewing conditions for a particular patient.
Fig. 19 graphically illustrates optimizer values for differing levels of residual accommodation.

Fig. 20 illustrates effects of pupil change and residual accommodation on optimized prescriptions for a particular patient.

Figs. 21A-C illustrate effects of pupil change and residual accommodation on optimized prescriptions for a particular patient.

Figs. 22-24 compare optical properties and results of eyes corrected with an optimized prescription to alternative treatments.

Fig. 25 schematically illustrates a system for determining a prescription for a particular patient and delivering that treatment using laser refractive surgery.

Fig. 26A illustrates a relationship between accommodation and pupil size when healthy eyes adjust to differing viewing distances.

Fig. 26B illustrates one exemplary relationship between effective power of an eye and pupil size for a patient, as can be provided from the presbyopia prescriptions of the present invention by generating an optical shape which effects desired changes in power with changes in pupil size of a particular patient under differing viewing conditions.

Fig. 26C illustrates a relationship between manifest power and pupil diameter, for example, as measured from patients having differing pupil diameters who have been successfully treated with a presbyopia-mitigating prescription. Such a relationship may be used to identify a desired change in optical power with changes in pupil diameter for a specific patient.

Figs. 27A and 27B graphically illustrate optical properties of an eye relevant to presbyopia.

Fig. 28 schematically illustrates a presbyopia-mitigating shape having a central add region.

Figs. 29 and 30 schematically illustrates residual accommodation and presbyopia treatments for increasing a focal range.

Figs. 31-37 graphically illustrate results from presbyopia-mitigating treatments for a population of individual patients.

Fig. 38 graphically illustrates accommodation through a range of differing patient ages.
[0052] Fig. 39 schematically illustrates another system for determining a presbyopia-mitigating prescription for a particular patient and delivering that treatment using laser refractive surgery.

[0053] Figs. 40 and 41 graphically illustrate a presbyopia-mitigating prescription derived so as to provide appropriate effective powers at two differing viewing conditions for a particular patient.

[0054] Figs. 42 and 43 graphically illustrate a presbyopia-mitigating prescription derived so as to provide appropriate effective powers at three differing viewing conditions for a particular patient.

[0055] Figs. 44 and 45 graphically illustrate a presbyopia-mitigating prescription derived so as to provide appropriate effective powers at four differing viewing conditions for a particular patient.

[0056] Figs. 46A and 46B graphically illustrate different presbyopia-mitigating prescriptions which provide differing effective power variation characteristics during pupil size changes under differing viewing conditions.

[0057] Figs. 47 and 48 graphically illustrate effects of different pupil sizes on derived presbyopia-mitigating prescriptions and their optical characteristics.

[0058] Fig. 49 illustrates simulated eye-chart letters as viewed with a presbyopic eye treated with a presbyopia-mitigating prescription derived for a particular patient.

[0059] Figs. 50A and 50B illustrate an exemplary power/pupil correlation and corresponding presbyopia prescription.

[0060] Fig. 51 shows through-focus results for a 20/20 eye chart letter E convolved with certain point spread function models across a vergence range according to embodiments of the present invention.

[0061] Fig. 52 illustrates CMTF value curves according to embodiments of the present invention.

[0062] Figs. 53A and 53B depict point spread functions with ring-type and centrally-concentrated configurations, respectively, according to embodiments of the present invention.

[0063] Fig. 54 shows cross sections of the point spread function images according to embodiments of the present invention.

[0064] Figs. 55A and 55B illustrates cross-sections for point spread functions according to embodiments of the present invention.
Fig. 56 depicts cross sections of point spread functions according to embodiments of the present invention.

Fig. 57 illustrates aspects of a method of evaluating an image quality provided by a vision treatment shape, according to embodiments of the present invention.

Fig. 58 illustrates aspects of a method of determining a compound modulation transfer function (CMTF) threshold value for a CMTF spatial frequency set, according to embodiments of the present invention.

Fig. 59 depicts aspects of methods for determining an optical surface shape and providing a treatment to a patient according to embodiments of the present invention.

Fig. 60 depicts aspects of methods for generating an optical surface shape for a patient according to embodiments of the present invention.

Fig. 61 depicts aspects of a vision requirements specification, according to embodiments of the present invention.

Fig. 62 depicts aspects of a vision requirements specification, according to embodiments of the present invention.

Fig. 63 illustrates aspects of techniques for determining a merit function for a target or treatment shape based on an optical metric value for the shape at the various viewing or testing distances, according to embodiments of the present invention.

Figs. 64A and 64B illustrate aspects of weighting value distributions, according to embodiments of the present invention.

Figs. 65A, 65B, 65C, and 65D illustrate aspects of weighting value distributions, according to embodiments of the present invention.

Figs. 66A and 66B illustrate aspects of weighting value distributions, according to embodiments of the present invention.

Figs. 67A and 67B illustrate aspects of weighting value distributions, according to embodiments of the present invention.

Figs. 68A, 68B, and 68C illustrate example distance, intermediate, and near vision experienced by an eye without correction, respectively.
[0078] Figs. 69A, 69B, and 69C illustrate example distance, intermediate, and near vision experienced by an eye corrected for near vision viewing distances only, respectively.

[0079] Figs. 70A, 70B, and 70C illustrate example distance, intermediate, and near vision experienced by an eye corrected for intermediate and near viewing distances, respectively.

5 [0080] Figs. 71A, 71B, and 71C illustrate example distance, intermediate, and near vision experienced by an eye corrected for distance and intermediate viewing distances, respectively.

DETAILED DESCRIPTION OF THE INVENTION

[0081] Although the methods, devices, and systems of the present invention are described primarily in the context of a laser eye surgery system, it should be understood that the techniques of the present invention may be adapted for use in other eye treatment procedures and systems such as contact lenses, intra-ocular lenses, radial keratotomy, collagenous corneal tissue thermal remodeling, removable corneal lens structures, glass spectacles, corneal ring implants, and the like.

[0082] Exemplary systems and methods disclosed herein can be implemented via a variety of ophthalmic devices or solutions. For example, treatment techniques may be used for any of a variety of surgery modalities, including excimer laser surgery, femtosecond surgery, and the like. A variety of forms of lasers and laser energy can be used to effect a correction or treatment, including infrared lasers, ultraviolet lasers, femtosecond lasers, wavelength multiplied solid-state lasers, and the like. By way of non-limiting example, ophthalmic corrections can involve a cornea or lens reshaping procedure, such as, for example using a picosecond or femtosecond laser. Laser ablation procedures can remove a targeted amount stroma of a cornea to change a cornea's contour and adjust for aberrations. In some cases, a treatment protocol can involve the delivery of a series of discrete pulses of laser light energy, with a total shape and amount of tissue removed being determined by a shape, size, location, and/or number of laser energy pulses impinging on or focused within a cornea. In some cases, a surgical laser, such as a non-ultraviolet, ultra-short pulsed laser that emits radiation with pulse durations as short as nanoseconds and femtoseconds (e.g., a femtosecond laser, or a picosecond laser) can be used to treat the eye of a patient. Other pulse widths may be suitable as well. The laser systems can be configured to deliver near infrared light. Other wavelengths may be used as well. The laser systems can be configured to deliver laser light focused at a focus depth (e.g. within corneal or other ophthalmologic tissue) which may be controlled by the system. Laser surgery with ultra-short pulse lasers such as femtosecond lasers can be used to treat the eye. These pulsed lasers can make very accurate incisions of the eye and can be used in many ways to treat the eye. Additional types of incisions that can be performed
with the short pulse lasers include incisions for paracentesis, limbal relaxing incisions, and refractive incisions to shape the cornea, for example.

[0083] In some cases, vision treatments can include focusing femtosecond laser energy within the stroma so as to ablate a volume of intrastromal tissue. By scanning the focal spot within an appropriate volume of the stromal tissue, it is possible to vaporize the volume so as to achieve a desired refractive alteration. Hence, embodiments of the present invention encompass laser surgical techniques that involve femtosecond laser photodisruption or photoalteration treatments. In some cases, a femtosecond laser can be used to perform the photodisruption, thus providing an easy, precise, and effective approach to refractive surgery.

[0084] According to some embodiments, a femtosecond laser (or other laser) of the optical system can be used to incise the cornea or to cut a flap. A femtosecond laser may be used to make arcuate or other incisions in the cornea, which incisions may be customized, intrastromal, stable, predictable, and the like. Likewise, corneal entry incisions may be made, which are custom, multi-plane, and self-sealing.

[0085] Pulsed laser beams include bursts or pulses of light. Pulsed lasers, such as non-ultraviolet, ultra-short pulsed lasers with pulse durations measured in the nanoseconds to femtoseconds range, can be used in ophthalmic surgical procedures as disclosed herein. For example, a pulsed laser beam can be focused onto a desired area of ophthalmologic material or tissue, such as the cornea, the capsular bag, or the lens of the eye, to photoalter the material in this area and, in some instances, the associated peripheral area. Examples of photoalteration of the material include, but are not necessarily limited to, chemical and physical alterations, chemical and physical breakdown, disintegration, ablation, photodisruption, vaporization, and the like. Exemplary treatment systems can include a focusing mechanism (e.g. lens) and/or a scanning mechanism so as to guide or direct a focus of femtosecond energy along a path within the patient's eye (e.g. at one or more corneal subsurface locations).

[0086] According to some embodiments, the vergence weighting systems and methods disclosed herein can be implemented in connection with software residing in a diagnostic device such as WaveScan® and iDesign™ devices.

[0087] Turning now to the drawings, Fig. 1 illustrates a laser eye surgery system of the present invention, including a laser 12 that produces a laser beam 14. Laser 12 is optically coupled to laser delivery optics 16, which directs laser beam 14 to an eye E of patient P. A delivery optics support structure (not shown here for clarity) extends from a frame 18 supporting laser 12. A
microscope 20 is mounted on the delivery optics support structure, the microscope often being used to image a cornea of eye E.

[0088] Laser 12 generally comprises an excimer laser, ideally comprising an argon-fluorine laser producing pulses of laser light having a wavelength of approximately 193 nm. Laser 12 will preferably be designed to provide a feedback stabilized fluence at the patient's eye, delivered via delivery optics 16. The present invention may also be useful with alternative sources of ultraviolet or infrared radiation, particularly those adapted to controllably ablate the corneal tissue without causing significant damage to adjacent and/or underlying tissues of the eye. Such sources include, but are not limited to, solid state lasers and other devices which can generate energy in the ultraviolet wavelength between about 185 and 205 nm and/or those which utilize frequency-multiplying techniques. Hence, although an excimer laser is the illustrative source of an ablating beam, other lasers may be used in the present invention.

[0089] Laser system 10 will generally include a computer or programmable processor 22. Processor 22 may comprise (or interface with) a conventional PC system including the standard user interface devices such as a keyboard, a display monitor, and the like. Processor 22 will typically include an input device such as a magnetic or optical disk drive, an internet connection, or the like. Such input devices will often be used to download a computer executable code from a tangible storage media 29 embodying any of the methods of the present invention. Tangible storage media 29 may take the form of a floppy disk, an optical disk, a data tape, a volatile or non-volatile memory, RAM, or the like, and the processor 22 will include the memory boards and other standard components of modern computer systems for storing and executing this code. Tangible storage media 29 may optionally embody wavefront sensor data, wavefront gradients, a wavefront elevation map, a treatment map, a corneal elevation map, and/or an ablation table. While tangible storage media 29 will often be used directly in cooperation with an input device of processor 22, the storage media may also be remotely operatively coupled with processor by means of network connections such as the internet, and by wireless methods such as infrared, Bluetooth, or the like.

[0090] Laser 12 and delivery optics 16 will generally direct laser beam 14 to the eye of patient P under the direction of a computer 22. Computer 22 will often selectively adjust laser beam 14 to expose portions of the cornea to the pulses of laser energy so as to effect a predetermined sculpting of the cornea and alter the refractive characteristics of the eye. In many embodiments, both laser beam 14 and the laser delivery optical system 16 will be under computer control of processor 22 to effect the desired laser sculpting process, with the processor effecting (and optionally modifying) the pattern of laser pulses. The pattern of pulses may be summarized in machine readable data of
tangible storage media 29 in the form of a treatment table, and the treatment table may be adjusted
to feedback input into processor 22 from an automated image analysis system in
response to feedback data provided from an ablation monitoring system feedback system.
Optionally, the feedback may be manually entered into the processor by a system operator. Such
feedback might be provided by integrating the wavefront measurement system described below
with the laser treatment system 10, and processor 22 may continue and/or terminate a sculpting
treatment in response to the feedback, and may optionally also modify the planned sculpting based
at least in part on the feedback. Measurement systems are further described in U.S. Patent No.
6,315,413, the full disclosure of which is incorporated herein by reference.

[0091] Laser beam 14 may be adjusted to produce the desired sculpting using a variety of
alternative mechanisms. The laser beam 14 may be selectively limited using one or more variable
apertures. An exemplary variable aperture system having a variable iris and a variable width slit is
described in U.S. Patent No. 5,713,892, the full disclosure of which is incorporated herein by
reference. The laser beam may also be tailored by varying the size and offset of the laser spot
from an axis of the eye, as described in U.S. Patent Nos. 5,683,379, 6,203,539, and 6,331,177, the
full disclosures of which are incorporated herein by reference.

[0092] Still further alternatives are possible, including scanning of the laser beam over the
surface of the eye and controlling the number of pulses and/or dwell time at each location, as
described, for example, by U.S. Patent No. 4,665,913, the full disclosure of which is incorporated
herein by reference; using masks in the optical path of laser beam 14 which ablate to vary the
profile of the beam incident on the cornea, as described in U.S. Patent No. 5,807,379, the full
disclosure of which is incorporated herein by reference; hybrid profile-scanning systems in which
a variable size beam (typically controlled by a variable width slit and/or variable diameter iris
diaphragm) is scanned across the cornea; or the like. The computer programs and control
methodology for these laser pattern tailoring techniques are well described in the patent literature.

[0093] Additional components and subsystems may be included with laser system 10, as should
be understood by those of skill in the art. For example, spatial and/or temporal integrators may be
included to control the distribution of energy within the laser beam, as described in U.S. Patent
No. 5,646,791, the full disclosure of which is incorporated herein by reference. Ablation effluent
evacuators filters, aspirators, and other ancillary components of the laser surgery system are
known in the art. Further details of suitable systems for performing a laser ablation procedure can
be found in commonly assigned U.S. Pat. Nos. 4,665,913, 4,669,466, 4,732,148, 4,770,172,
4,773,414, 5,207,668, 5,108,388, 5,219,343, 5,646,791 and 5,163,934, the complete disclosures of
which are incorporated herein by reference. Suitable systems also include commercially available refractive laser systems such as those manufactured and/or sold by Alcon, Bausch & Lomb, Nidek, WaveLight, LaserSight, Schwind, Zeiss-Meditec, and the like. Basis data can be further characterized for particular lasers or operating conditions, by taking into account localized environmental variables such as temperature, humidity, airflow, and aspiration.

**[0094]** Fig. 2 is a simplified block diagram of an exemplary computer system 22 that may be used by the laser surgical system 10 of the present invention. Computer system 22 typically includes at least one processor 52 which may communicate with a number of peripheral devices via a bus subsystem 54. These peripheral devices may include a storage subsystem 56, comprising a memory subsystem 58 and a file storage subsystem 60, user interface input devices 62, user interface output devices 64, and a network interface subsystem 66. Network interface subsystem 66 provides an interface to outside networks 68 and/or other devices, such as the wavefront measurement system 30.

**[0095]** User interface input devices 62 may include a keyboard, pointing devices such as a mouse, trackball, touch pad, or graphics tablet, a scanner, foot pedals, a joystick, a touchscreen incorporated into the display, audio input devices such as voice recognition systems, microphones, and other types of input devices. User input devices 62 will often be used to download a computer executable code from a tangible storage media 29 embodying any of the methods of the present invention. In general, use of the term "input device" is intended to include a variety of conventional and proprietary devices and ways to input information into computer system 22.

**[0096]** User interface output devices 64 may include a display subsystem, a printer, a fax machine, or non-visual displays such as audio output devices. The display subsystem may be a cathode ray tube (CRT), a flat-panel device such as a liquid crystal display (LCD), a projection device, or the like. The display subsystem may also provide a non-visual display such as via audio output devices. In general, use of the term "output device" is intended to include a variety of conventional and proprietary devices and ways to output information from computer system 22 to a user.

**[0097]** Storage subsystem 56 can store the basic programming and data constructs that provide the functionality of the various embodiments of the present invention. For example, a database and modules implementing the functionality of the methods of the present invention, as described herein, may be stored in storage subsystem 56. These software modules are generally executed by processor 52. In a distributed environment, the software modules may be stored on a plurality of
computer systems and executed by processors of the plurality of computer systems. Storage subsystem 56 typically comprises memory subsystem 58 and file storage subsystem 60.

[0098] Memory subsystem 58 typically includes a number of memories including a main random access memory (RAM) 70 for storage of instructions and data during program execution and a read only memory (ROM) 72 in which fixed instructions are stored. File storage subsystem 60 provides persistent (non-volatile) storage for program and data files, and may include tangible storage media 29 (Fig. 1) which may optionally embody wavefront sensor data, wavefront gradients, a wavefront elevation map, a treatment map, and/or an ablation table. File storage subsystem 60 may include a hard disk drive, a floppy disk drive along with associated removable media, a Compact Digital Read Only Memory (CD-ROM) drive, an optical drive, DVD, CD-R, CD-RW, solid-state removable memory, and/or other removable media cartridges or disks. One or more of the drives may be located at remote locations on other connected computers at other sites coupled to computer system 22. The modules implementing the functionality of the present invention may be stored by file storage subsystem 60.

[0099] Bus subsystem 54 provides a mechanism for letting the various components and subsystems of computer system 22 communicate with each other as intended. The various subsystems and components of computer system 22 need not be at the same physical location but may be distributed at various locations within a distributed network. Although bus subsystem 54 is shown schematically as a single bus, alternate embodiments of the bus subsystem may utilize multiple busses.

[0100] Computer system 22 itself can be of varying types including a personal computer, a portable computer, a workstation, a computer terminal, a network computer, a control system in a wavefront measurement system or laser surgical system, a mainframe, or any other data processing system. Due to the ever-changing nature of computers and networks, the description of computer system 22 depicted in Fig. 2 is intended only as a specific example for purposes of illustrating one embodiment of the present invention. Many other configurations of computer system 22 are possible having more or less components than the computer system depicted in Fig. 2.

[0101] Referring now to Fig. 3, one embodiment of a wavefront measurement system 30 is schematically illustrated in simplified form. In very general terms, wavefront measurement system 30 is configured to sense local slopes of a gradient map exiting the patient's eye. Devices based on the Hartmann-Shack principle generally include a lenslet array to sample the gradient map uniformly over an aperture, which is typically the exit pupil of the eye. Thereafter, the local slopes of the gradient map are analyzed so as to reconstruct the wavefront surface or map.
More specifically, one wavefront measurement system 30 includes an image source 32, such as a laser, which projects a source image through optical tissues 34 of eye E so as to form an image 44 upon a surface of retina R. The image from retina R is transmitted by the optical system of the eye (e.g., optical tissues 34) and imaged onto a wavefront sensor 36 by system optics 37. The wavefront sensor 36 communicates signals to a computer system 22' for measurement of the optical errors in the optical tissues 34 and/or determination of an optical tissue ablation treatment program. Computer 22' may include the same or similar hardware as the computer system 22 illustrated in Figs. 1 and 2. Computer system 22' may be in communication with computer system 22 that directs the laser surgery system 10, or some or all of the components of computer system 22, 22' of the wavefront measurement system 30 and laser surgery system 10 may be combined or separate. If desired, data from wavefront sensor 36 may be transmitted to a laser computer system 22 via tangible media 29, via an I/O port, via an networking connection 66 such as an intranet or the Internet, or the like.

Wavefront sensor 36 generally comprises a lenslet array 38 and an image sensor 40. As the image from retina R is transmitted through optical tissues 34 and imaged onto a surface of image sensor 40 and an image of the eye pupil P is similarly imaged onto a surface of lenslet array 38, the lenslet array separates the transmitted image into an array of beamlets 42, and (in combination with other optical components of the system) images the separated beamlets on the surface of sensor 40. Sensor 40 typically comprises a charged couple device or "CCD," and senses the characteristics of these individual beamlets, which can be used to determine the characteristics of an associated region of optical tissues 34. In particular, where image 44 comprises a point or small spot of light, a location of the transmitted spot as imaged by a beamlet can directly indicate a local gradient of the associated region of optical tissue.

Eye E generally defines an anterior orientation ANT and a posterior orientation POS. Image source 32 generally projects an image in a posterior orientation through optical tissues 34 onto retina R as indicated in Fig. 3. Optical tissues 34 again transmit image 44 from the retina anteriorly toward wavefront sensor 36. Image 44 actually formed on retina R may be distorted by any imperfections in the eye's optical system when the image source is originally transmitted by optical tissues 34. Optionally, image source projection optics 46 may be configured or adapted to decrease any distortion of image 44.

In some embodiments, image source optics 46 may decrease lower order optical errors by compensating for spherical and/or cylindrical errors of optical tissues 34. Higher order optical errors of the optical tissues may also be compensated through the use of an adaptive optic element,
such as a deformable mirror (described below). Use of an image source 32 selected to define a point or small spot at image 44 upon retina R may facilitate the analysis of the data provided by wavefront sensor 36. Distortion of image 44 may be limited by transmitting a source image through a central region 48 of optical tissues 34 which is smaller than a pupil 50, as the central portion of the pupil may be less prone to optical errors than the peripheral portion. Regardless of the particular image source structure, it will be generally be beneficial to have a well-defined and accurately formed image 44 on retina R.

[0106] In one embodiment, the wavefront data may be stored in a computer readable medium 29 or a memory of the wavefront sensor system 30 in two separate arrays containing the x and y wavefront gradient values obtained from image spot analysis of the Hartmann-Shack sensor images, plus the x and y pupil center offsets from the nominal center of the Hartmann-Shack lenslet array, as measured by the pupil camera 51 (Fig. 3) image. Such information contains all the available information on the wavefront error of the eye and is sufficient to reconstruct the wavefront or any portion of it. In such embodiments, there is no need to reprocess the Hartmann-Shack image more than once, and the data space required to store the gradient array is not large. For example, to accommodate an image of a pupil with an 8 mm diameter, an array of a 20 x 20 size (i.e., 400 elements) is often sufficient. As can be appreciated, in other embodiments, the wavefront data may be stored in a memory of the wavefront sensor system in a single array or multiple arrays.

[0107] While the methods of the present invention will generally be described with reference to sensing of an image 44, a series of wavefront sensor data readings may be taken. For example, a time series of wavefront data readings may help to provide a more accurate overall determination of the ocular tissue aberrations. As the ocular tissues can vary in shape over a brief period of time, a plurality of temporally separated wavefront sensor measurements can avoid relying on a single snapshot of the optical characteristics as the basis for a refractive correcting procedure. Still further alternatives are also available, including taking wavefront sensor data of the eye with the eye in differing configurations, positions, and/or orientations. For example, a patient will often help maintain alignment of the eye with wavefront measurement system 30 by focusing on a fixation target, as described in U.S. Patent No. 6,004,313, the full disclosure of which is incorporated herein by reference. By varying a position of the fixation target as described in that reference, optical characteristics of the eye may be determined while the eye accommodates or adapts to image a field of view at a varying distance and/or angles.
[0108] The location of the optical axis of the eye may be verified by reference to the data provided from a pupil camera 52. In the exemplary embodiment, a pupil camera 52 images pupil 50 so as to determine a position of the pupil for registration of the wavefront sensor data relative to the optical tissues.

[0109] An alternative embodiment of a wavefront measurement system is illustrated in Fig. 3A. The major components of the system of Fig. 3A are similar to those of Fig. 3. Additionally, Fig. 3A includes an adaptive optical element 53 in the form of a deformable mirror. The source image is reflected from deformable mirror 98 during transmission to retina R, and the deformable mirror is also along the optical path used to form the transmitted image between retina R and imaging sensor 40. Deformable mirror 98 can be controllably deformed by computer system 22 to limit distortion of the image formed on the retina or of subsequent images formed of the images formed on the retina, and may enhance the accuracy of the resultant wavefront data. The structure and use of the system of Fig. 3A are more fully described in U.S. Patent No. 6,095,651, the full disclosure of which is incorporated herein by reference.

[0110] The components of an embodiment of a wavefront measurement system for measuring the eye and ablations may comprise elements of a WaveScan®, available from AMO Manufacturing USA, LLC in Milpitas, California. One embodiment includes a WaveScan® with a deformable mirror as described above. An alternate embodiment of a wavefront measuring system is described in U.S. Patent No. 6,271,915, the full disclosure of which is incorporated herein by reference. It is appreciated that any wavefront aberrometer could be employed for use with the present invention. Relatedly, embodiments of the present invention encompass the implementation of any of a variety of optical instruments provided by Abbott Medical Optics, Inc., including the iDesign system, and the like.

[0111] Relatedly, embodiments of the present invention encompass the implementation of any of a variety of optical instruments provided by WaveFront Sciences, Inc., including the COAS wavefront aberrometer, the ClearWave contact lens aberrometer, the CrystalWave IOL aberrometer, and the like. Embodiments of the present invention may also involve wavefront measurement schemes such as a Tscherning-based system, which may be provided by WaveFront Sciences, Inc. Embodiments of the present invention may also involve wavefront measurement schemes such as a ray tracing-based system, which may be provided by Tracey Technologies, Corp.

[0112] The present invention is useful for enhancing the accuracy and efficacy of photorefractive keratectomy (PRK), laser in situ keratomileusis (LASIK), laser assisted epithelium
keratomileusis (LASEK), and the like. The present invention can provide enhanced optical correction approaches by improving the methodology for scaling an optical shape, or by generating or deriving new optical shapes, and the like.

[0113] The techniques of the present invention can be readily adapted for use with existing laser systems, including the Excimer laser eye surgery systems commercially available from AMO Manufacturing USA, LLC in Milpitas, California. Other suitable laser systems are manufactured by Alcon, Bausch & Lomb, Wavelight, Schwind, Zeiss-Meditec, Lasersight, Nidek and the like. By providing improved corneal ablation profiles for treating optical defects, the present invention may allow enhanced treatment of patients who have heretofore presented difficult or complicated treatment problems. When used for determining, deriving, and/or optimizing prescriptions for a particular patient, the systems and methods may be implemented by calculating prescriptions for a range of patients, for example, by calculating discrete table entries throughout a range of patient characteristics, deriving or empirically generating parametric patient characteristic/prescription correlations, and the like, for subsequent use in generating patient-specific prescriptions.

[0114] When designing a prescriptive shape for an eye treatment, it is useful to select a mathematical gauge of optical quality appropriate for the vision condition for use as a goal function. This allows for quantification and optimization of the shape, and for comparison among different shapes. The present invention provides methods for establishing a customized optical shape for a particular patient based on a set of patient parameters per the goal function. By incorporating iterative optimization algorithms, it is also possible to generate a shape having an optimized level of optical quality for the particular patient.

[0115] Selecting A Goal Function Appropriate For A Vision Condition

[0116] The goal function relates to optical quality, and it can be, for example, based on, or a function of (or related to) optical metrics such as Strehl ratio (SR), modulation transfer function (MTF), point spread function (PSF), encircled energy (EE), MTF volume or volume under MTF surface (MTFV), or contrast sensitivity (CS); and optionally to new optical metrics which are appropriate to vision conditions such as presbyopia; for instance, compound modulation transfer function (CMTF) as described below. In optical terms, the goal function should make sense. That is to say, minimization or maximization of the goal function should give a predictable optimized optical quality of the eye. The goal function can be a function with a certain number of free parameters to be optimized (minimized) through an optimization, or minimization, algorithm.

[0117] Although there are many types of goal functions available for use with the present invention, the discussion below generally touches on two broad schools of goal functions. In a
Diffraction Theory based approach, the shape is considered as a wave aberration. Typically, a
Fourier transform is employed for calculating optical quality related parameters, such as Strehl
ratio (SR), modulation transfer function (MTF), MTF volume or volume under MTF surface
(MTFV), compound modulation transfer function (CMTF), or contrast sensitivity (CS), encircled
energy (EE) (based on point spread function), as well as special cases that combine one or more of
these parameters, or values of the parameters in specific situations (such as MTF at spatial
frequency or encircled energy at a field of view), or integration of any parameters (volume of MTF
surface at all frequencies or up to a cutoff frequency, for example 60 cycles/degree or 75
cycles/degree, because 60 cycles/degree is the retina cone's limiting spatial frequency). In a
Geometrical Optics approach, or the so-called ray tracing approach, the optical effect is based on
ray tracing. With both the Diffraction Theory and the Geometrical Optics approaches,
polychromatic point spread function with Stiles-Crawford effect, chromatic aberrations as well as
retina spectral response function can be used.

[0118] Monochromatic point spread function (PSF) has been used for describing optical defects
of optical systems having aberrations. Due to the simple relationship between wave aberrations
and the PSF for an incoherent light source, Fourier transform of the generalized pupil function has
been used in the calculation of point spread functions. Most optical applications, however, do not
use a monochromatic light source. In the case of human vision, the source is essentially white
light. Thus, there are limitations associated with the use of monochromatic PSF as a goal function.

[0119] Polychromatic point spread function (PSF) with correct chromatic aberrations, Stiles-
Crawford effect as well as retina response function, can be used for optical modeling of human
eyes. Here, chromatic aberrations arise because light composed of different wavelengths will
focus either in front of the retina or behind it. Only portions of the light will focus exactly on the
retina. This gives the eye an extended depth-of-focus, i.e., if one has focusing error of some
amount, the eye is still capable of focusing at least for some wavelengths. Therefore, chromatic
aberrations in fact help the correction of presbyopia. If the depth-of-focus is sufficiently large,
there would be no presbyopia problem. Unfortunately, the chromatic aberrations are not large
enough and it also varies with the wavelength. Stiles-Crawford effect, also known as pupil
apodization, is due to the waveguide property of the retinal cones. Light from the pupil periphery
has a slightly less chance of being detected by the retina because the ray of light might not reach
the bottom of the cone, due to a slight incident angle. As for the retinal spectral response function,
it is known that the cones, which are responsible for daylight vision, have different sensitivity to
different wavelengths. Only green light is absorbed by the eye almost completely. Both blue light and red light are absorbed by the eye partially.

[0120] Once the PSF is calculated, calculation of the Strehl ratio is straightforward. Strehl ratio can be defined as the ratio of the peak of the point spread function (PSF) of an optical system to the peak of a diffraction-limited optical system with the same aperture size. An example of a Strehl ratio is shown in Fig. 27A. A diffraction-limited optical system is typically a system with no aberrations, or optical errors. It can be an ideal or perfect optical system, having a Strehl ratio of 1.

[0121] The goal function can also be a function of modulation transfer function (MTF).

Modulation transfer function can be used to predict visual performance. Typically, the MTF at one spatial frequency corresponds to one angular extend of features of targets. The modulation transfer function (MTF) can be calculated with the following formulations:

\[ \text{MTF}(u,v) = \text{FT}[\text{PSF}(x,y)] \]

\[ \text{MTF}(u,v) = \text{Re}[\text{GPF}(x,y) \odot \text{GPF}(x,y)] \]

where \( u \) and \( v \) represent spatial frequencies, \( \text{Re} \) represents the real part of a complex number, \( \text{FT} \) represents a Fourier Transform, \( \text{GPF} \) represents a generalized pupil function, and \( x \) and \( y \) represent position or field of view. An example of an MTF is shown in Fig. 27B.

[0122] Modulation transfer function (MTF) is a measure for how much spatial details are transferred from pupil space to imaging space (retina in the case of human eye). MTF can be related to contrast sensitivity (CS). Mathematically MTF can be defined as the Fourier transform of the point spread function as

\[ h(u, v) = \text{FT}[i(x, y) \exp[-i2\pi(ux + vy)] dxdy, \]

where \( i(x,y) \) is the point spread function (PSF). Calculation of PSF can be done with the Fourier transform of the generalized pupil function.

[0123] MTF at a specific spatial frequency can represent the percentage of the sinusoidal wave of a specific spatial frequency that is preserved after going through the optical system. MTF at 30 cycles/degree and at 60 cycles/degree are considered as important because 30 cpd corresponds to 20/20 visual acuity and 60 cpd corresponds to 20/10 visual acuity, the highest spatial resolution the cones in the retinal can process. MTF at other spatial frequencies may also be useful.
The volume under the MTF surface up to a certain spatial frequency (such as 60 cpd) can be meaningful as it includes all spatial frequency information. In some cases, it is desirable to use the volume under MTF surface within a band (i.e. from one specific spatial frequency to another specific spatial frequency).

**Compound Modulation Transfer Function**

Compound MTF can be calculated as a linear combination of MTF at certain spatial frequencies, normalized at diffraction-limited MTF, and can be represented by the following formula

$$\text{CMTF}(v) = \frac{1}{n} \sum_{i=1}^{n} \alpha_i h_i,$$

where $n$ is the number of MTF curves, $a_i$ is the reciprocal of the $i$th diffraction-limited MTF, and $\frac{1}{\alpha_i}$ is the $i$th MTF curve. The selection of certain spatial frequencies can depend on the importance of each frequency. For example, in the case of presbyopia, 20/40 vision may be more important than 20/20 as the distance vision is often compromised by the improved near vision. Figs 4A and 4B show examples of an CMTF curve as well as its individual MTF curves at different specific spatial frequencies. In a perfect optical system, CMTF is equal to one.

In a related embodiment, the compound MTF can be calculated as

$$F(v) = (\alpha_1 \text{MTF}_1 + \alpha_2 \text{MTF}_2 + \alpha_3 \text{MTF}_3)/3$$

where MTFi, MTF2, and MTF3 are the MTF values at 10 cycles/degree, 20 cycles/degree and 30 cycles/degree, respectively. These correspond to Snellen eye chart of 20/60, 20/40 and 20/20 visions, respectively. The weighting coefficients $\alpha_1, \alpha_2, \alpha_3$ can be chosen so that $\alpha_1/\alpha_2, \alpha_2/\alpha_3, 1/\alpha_3$ are the diffraction-limited MTF at these spatial frequencies, respectively. Therefore, in the diffraction-limited case, the compound MTF $F(v)$ can have a maximal value of unity.

Where MTF at one spatial frequency corresponds to one angular extend of features of targets, compound MTF can be calculated as linear combination of MTF at different spatial frequencies normalized by a diffraction-limited MTF, and can similarly be used to predict visual outcome. Another general formula for the calculation of CMTF as a function of visual vergence (nu) is

$$\text{CMTF}(v) = \frac{1}{n} \sum_{i=1}^{n} \alpha_i \text{MTF}_i(v)$$
where \( \frac{1}{n} \) is the reciprocal of the \( n \)-th diffraction-limited MTF. This formula can provide CMTF for all possible vergences. In some cases, three MTF curves at 10, 20 and 30 cycles per degree are used. An ideal value of CMTF can be about 1. Good values can be about 0.2 or about 0.3. In a healthy eye, the spatial frequency limit can be about 60 cycles per degree due to the configuration of retina cones. However, in the treatment of presbyopia, for example, it may not be necessary to provide a treatment corresponding to this limit, as the treatment will often involve a compromise of good distance and near sight. Optionally, a minimum distance vision gauge desired target may be provided, with near sight being optimized and, as needed, compromised.

[0128] Fig. 4A illustrates an example of the compound MTF over a vergence of 3 diopters (upper panel) versus its corresponding individual MTF curves at 15, 30, and 60 cycles per degree (lower panel). Fig. 4B illustrates an example of the compound MTF over a vergence of 3 diopters (upper panel) versus its corresponding individual MTF curves at 10, 20, and 30 cycles per degree (lower panel). Compound MTF can correlate well with visual acuity and contrast sensitivity at the same time, at least optically. In some embodiments, the compound modulation transfer function is determined for individual MTF curves at 30, 45, and 60 cpd. The selection of the individual MTF curve values can involve a linear combination based on the optical response of the eye.

[0129] In general, there can be two different types of cutoff spatial frequencies, and each involves a factors that affect acuity. Cutoff spatial frequency can correspond to the maximum spatial frequency, above which information can no longer be used. Whereas most individuals can discern information from objects having very low spatial frequency, as the spatial frequency increases, it is typically increasingly more difficult for an individual to discern information from such objects. At some threshold, an increased spatial frequency no longer yields increased information.

[0130] A first type of cutoff spatial frequency is related to aperture dimension. In this case, a system having a larger aperture (e.g. an eye with a larger pupil) will correspond to a larger cutoff spatial frequency. Conversely, a system having a smaller aperture (e.g. an eye with a smaller pupil) will correspond to a smaller cutoff spatial frequency. Often, such cutoff spatial frequencies will be linearly dependent on a pupil dimension, for example the pupil diameter. Smaller pupil sizes typically correspond to an extended, or larger, depth of focus. Relatedly, smaller pupil sizes often result in lower resolution. Assuming there are no aberrations, a larger pupil size is thought to confer increased resolution.

[0131] A second type of cutoff spatial frequency typically depends on the spacing of cones on the retina of the eye. With this type of cutoff spatial frequency, the standard value is 30 cpd,
which corresponds to 20/20 vision. Another value, 60 cpd, corresponds to 20/10 vision and is often considered a physiological limit. In such cases, the retinal cones are very closely spaced. The spacing of retinal cones will vary among individuals.

[0132] In the example of presbyopia treatment, it may be desirable to maintain a lower spatial frequency. In some cases, presbyopia will involve a compromise between distance and near vision. It may be difficult to achieve high spatial resolution, thus enhancing the desirability of emphasizing lower and medium spatial frequency information. In other words, high spatial frequency information may be sacrificed in order to improve the combination of near and distance vision.

[0133] As noted above, a compound modulation transfer function can include individual MTF curves at various combinations of spatial frequencies, such as 15, 30, and 60 cycles per degree and 10, 20, and 30 cycles per degree. An individual MTF can have a value ranging from about 5 cycles per degree to about 75 cycles per degree. In many instances, at least one individual MTF of a CMTF will range from about 10 cycles per degree to about 30 cycles per degree, and can often be about 20 cycles per degree. Where a CMTF includes three individual MTFs, a first individual MTF can range from about 5 cycles per degree to about 20 cycles per degree, a second individual MTF can range from about 15 cycles per degree to about 45 cycles per degree, and a third individual MTF can range from about 30 cycles per degree to about 75 cycles per degree. In some circumstances, the upper limit of an individual MTF can be about 60 cycles per degree.

[0134] In some cases, the CMTF will be based on an average of the individual MTF curves. In some embodiments, the present invention provides compound modulation transfer functions that correspond to three, four, five, or any number of individual modulation transfer functions. For example, a CMTF can include from about 2 to about 7 individual MTFs. A CMTF can also include from about 3 to about 6 individual MTFs.

[0135] Individual MTFs can correspond to a curve through a certain vergence. Typically, a target at a far distance corresponds to a small vergence value. As a target moves closer to the eye, the vergence increases. The individual MTFs can be based on a value ranging from about zero to about three diopeters.

[0136] The individual MTFs can be selected based on any number of criteria, such as empirical data or clinical observations. Relatedly, individual MTFs can be chosen for pure testing purposes. The CMTF can provide a parameter to evaluate the effectiveness of a treatment for a vision condition, such as presbyopia. Often, the CMTF will correlate with a particular visual outcome.
To establish an optically optimized shape appropriate for a vision condition, at least one of the goal functions, such as Strehl ratio, encircled energy, or MTF, MTF volume or volume under MTF surface (MTFV), compound modulation transfer function (CMTF), or contrast sensitivity (CS) should be maximized. For improved vision condition treatment, the optical metric can be maximized in all target vergence, that is, for targets at all distances. Furthermore, it is also desirable to minimize the fluctuation of the goal function. Therefore, the goal function, which is incorporated into the optimization algorithm of the optimizer, can be defined as

$$O(c_1, c_2, ..., PAR) = (I + \sigma)(I + PV) \frac{\int_0^{v_o} dv}{\int_0^{v_o} F(v)dv}$$

where $O$ is the goal function; $c_1, c_2, ...$ are the polynomial coefficients; PAR is presbyopia-add to pupil ratio (described below); $v$ is the vergence; $F(v)$ is one of the optical metrics; $\sigma$ is the standard deviation of $F(v)$, PV is the peak-to-valley of $F(v)$; and $v_o$ is the end point of the vergence range, which may be (for example) between 15 and 100 cm, such as 40 cm. Because $\int dv$ is a constant, either a smaller $\sigma$ or a larger $\int F(v)dv$ can minimize the goal function $O$.

The formulas given here are examples of the many formulae that can be used as the goal function. The basic approach will often be to provide a goal function that is optimized to give as practical a solution as possible for correction or treatment of the vision condition.

The compound MTF may reflect to what extent information is being modulated when passing through an optical system. For example, CMTF can represent the percentage of information at different spatial frequencies that is retained.
Selecting An Iterative Optimization Algorithm

[0141] Any of a number of optimization algorithms may be used by the optimizer to maximize, minimize, or otherwise globally or locally optimize the goal function. Because many numerical algorithms use function minimization concept, it is often convenient, but not necessarily required, to use minimization of the goal function. As examples, N-dimensional minimization algorithms such as the Downhill Simplex method, the Direction Set method, and the Simulated Annealing method can be used to optimize the goal function. Likewise, the algorithm described by Press et al., in "Numerical Recipes in C++", Cambridge University Press, 2002 can also be used. Algorithms such as those listed above are often used for function optimization in multi-dimensional space.

[0142] The Downhill Simplex method starts with an initialization of N+1 points or vertices to construct a simplex for an N-dimensional search, and in every attempt tries to reflect, stretch, or shrink the simplex by geometrical transformation so that a close-to-global minimum or pre-defined accuracy can be found. When Gaussian random noise of standard deviation of 0.02 \( \mu \eta \) in optical path difference (OPD) is added, the algorithm still converges, with no degradation.

[0143] In the case of Direction Set method, also known as Powell's method, N one-dimensional vectors are initialized and the N-dimensional search is split in such a way that a one N-dimensional vector is chosen and the minimization is done in that direction while other variables (N-1 dimensions) are fixed. This process is continued until all dimensions are covered. A new iteration is initiated until the pre-determined criterion is met. The Direction Set method can use a separate one-dimensional minimization algorithm such as a Golden section search.

[0144] The Simulated Annealing method, which is useful for dealing with a large number of uncertainties, starts with an initial configuration. The objective is to minimize \( E \) (analog to energy) given the control parameter \( T \) (analog to temperature). Simulated Annealing is analogous to annealing, is a recent, proven method to solve otherwise intractable problems, and may be used to solve the ablation equation in laser ablation problem. This is more fully described in PCT Application No. PCT/US01/08337, filed March 14, 2001, and in U.S. Patent No. 6,673,062, issued January 6, 2004, the entire disclosures of which are incorporated herein by reference. Simulated annealing is a method that can be used for minimizing (or maximizing) the parameters of a function. It is particularly suited to problems with very large, poorly behaved function spaces. Simulated annealing can be applied in the same way regardless of how many dimensions are present in the search space. It can be used to optimize any conditions that can be expressed
numerically, and it does not require a derivative. It can also provide an accurate overall minimum despite local minima in the search space, for example.

[0145] Fig. 5 shows the flow chart of an overall method for shape optimization for a vision condition treatment. Each functional block may contain one or more alternatives. To create an add-on shape W(r) for a vision condition treatment, an iterative function minimization algorithm can be employed such that the goal function, which could be a function of any suitable optical metrics (e.g. CMTF) is itself optimized to solve for an unknown shape. The shape can be expanded into a set of even power term polynomials (EPTP) or non-EPTP (i.e. all power term polynomials). EPTP refers to polynomials that only have the even power terms, for instance, F(r) = ar^2 + br^4 + cr^6. The goal function should have good correlation with visual performance, at least optically. Point spread function can be calculated to obtain additional and/or alternative optical metrics. The vision condition prescription can refer to an optical surface that can be used to treat or mitigate the vision condition. It can correspond to, for example, the shape of a spectacle lens, a contact lens, an intra-ocular lens, a tissue ablation profile for refractive surgery, and the like.

[0146] Another representation of the data flow process is depicted in the flow chart in Fig. 6, which shows data flow for shape optimization for presbyopia correction. Again, each functional block may contain one or more alternatives.

[0147] It is desirable that the optimizer provide satisfactory outcome in terms of attributes such as result, convergence, and speed. Fig. 7 shows a comparison of Direction Set method and Downhill Simplex method for the following inputs: pupil size 5.6 mm, vergence 3D and vergence step 0.1D. Direction Set method uses 17 iterations and Downhill Simplex method uses 152 iterations. Each Direction Set method iteration takes longer than each Downhill Simplex method iteration. The optimizer value for the Direction Set method is 2.8 while that for the Downhill Simplex method is 2.658. Shape for left panel is as -0.9055r^2 + 6.4188r^4 - 2.6767r^6 + 0.5625r^8 with ratio of 0.7418.

[0148] Both algorithms seem to converge to a similar shape, although the depths of the shapes are different. Considering the difference in the pupil ratio, however, the actual shapes within 70% of the pupil radius are quite close. When the vergence step is smaller, each iteration can take a longer time, but the overall number of iterations often tends to become smaller.

**Inputting An Initial Prescription Into an Optimizer**

[0149] The initial prescription, often comprising an optical surface shape, may be defined by an expansion such as a polynomial (EPTP, non-EPTP), a Zernike polynomial, a Fourier series, or a
discrete shape entirety. A discrete shape entirety can also be referred to as a direct surface representation by numerical grid values. The prescription shape may be assumed to be circularly or radially symmetric, with the aim of approaching an emmetropic eye. The symmetric shape can be decomposed into a set of polynomials, such that it has one or more independent variables. One of the variables can be the presbyopia-add to pupil ratio (PAR), or the ratio of the shape diameter to the pupil diameter. When a central power add region is employed (as described below), the PAR can be the ratio of the radius of the presbyopia-add to the radius of the pupil. It will also be appreciated that the ratios discussed herein can be based on area ratios or on diameter or radius ratios. It should be assumed that when diameter or radius ratios are discussed, that discussion also contemplates area ratios. In certain cases, the PAR can range from about 0.2 to about 1.0. Relatedly, in some cases the methods of the present invention can constrain the PAR to range from about 0.2 to about 1.0. The other variables can be the coefficients of each polynomial term. For example,

\[
Shape(r) = ar + br^2 + cr^3 + dr^4 + er^5 + fr^6
\]

**[0150]** The diameter of the shape can be larger than the pupil size, but if so special considerations may need to be taken. For example, it may be necessary to only consider the net shape within the pupil.

**[0151]** The polynomials can be normal polynomials or polynomials with even power terms only. For example, even-power-term polynomials (EPTP) up to the 6th or 8th order can be used to obtain a practically good output, that is, a practical optimal shape for the particular patient. Residual accommodation can also play an active role in presbyopia correction. In a related instance, normal presbyopes can be treated with the prescription obtained in this approach together with a prescription for the correction of the refractive error.

**[0152]** As an example, a circularly or radially symmetric, pupil-size dependent shape for presbyopia-add can be assumed for emmetropic presbyopes. The shape can then be expanded to polynomials up to the 6th or 8th order. With the optimization procedure, it is found that polynomial expansion of the shape up to the 6th or 8th order can be used to obtain a practical optimal shape for presbyopia correction.

**[0153]** In a wavefront with aberrations, denoted by \( W(r, \Theta) \), the wavefront can be thought of as an optimal shape for vision correction. The polychromatic PSF can be expressed as
\[ PSF = \sum \lambda R(\lambda) \left| \text{FFT}\left( P_\infty(r) \exp\left( -j \frac{2\pi}{\lambda} [W(r, \theta) + \alpha \lambda D(\lambda) + V(l) + RA(l)] \right) \right) \right|^2 \]

where \( R(\lambda) \) is the retina spectral response function and can be approximated to
\[ R(\lambda) = e^{-300(\lambda - \lambda_0)^2} \]
and \( P(r) \) is the pupil apodization function (Stiles-Crawford) and can be written as
\[ P_\infty(r) = 10^{-\frac{r^2}{R^2}} \]
and \( D(\lambda) \) is chromatic aberration at wavelength \( \lambda \) and is close to
\[ D(\lambda) = -21.587 + 92.87\lambda - 134.98\lambda^2 + 67.407\lambda^3 \]
and \( V(l) \) is the vergence induced aberration at distance \( l \) meters, and \( RA(l) \) is the residual accommodation induced aberrations with a different sign as compared to \( V(l) \). When there are no aberrations, \( RA(l) \) can cancel \( V(l) \) as long as there is enough residual accommodation in the eye. Here, the central wavelength \( \lambda \) is taken as 0.55 \( \mu \text{m} \) (as all wavelength units in the above formulae are in \( \mu \text{m} \)). The pupil apodization strength parameter \( p \) is taken as 0.06. \( a \) is the conversion factor from diopter to optical path difference (OPD). FFT denotes a fast Fourier transform and \( \sqrt{\cdot} \) indicates the module of a complex number.

[0154] The polychromatic point spread function, or PPSF, can be the point spread function of an eye as calculated with consideration of the polychromatic nature of the incident light. Further, the chromatic aberrations, the Stiles-Crawford effect, as well as the retinal spectral response function can also be considered.

[0155] The vergence induced aberration, or VIA, can be equal to the reciprocal of the vergence distance. When a target at a certain distance is viewed by the eye, it is the same as viewing the target at infinity but the eye has an additional aberration, the vergence induced aberration.

[0156] For emmetropic eyes, it may be desirable that the wavefront that is optimized be circularly symmetric. Therefore, it can be decomposed into a set of polynomials (non-EPTP) as
\[ W(r) = ar + br^2 + cr^3 + dr^4 + er^5 + \ldots \]

[0157] However, if it is desirable that the edge of the shape be smoother, it may be advantageous to decompose the wavefront into a set of even-power-term polynomials (EPTP) as
\[ W(r) = ar^2 + br^4 + cr^6 + dr^8 + \ldots \]

[0158] Using even power term polynomials (EPTP) also can help to establish a surface shape that is more round at the center, which creates certain manufacturing or ablation efficiencies.
It may also be useful to denote another parameter, \( t \), to be the ratio of the radius of the wavefront \( R \) to the radius of the pupil \( R_o \). This is because both \( D(X) \) and \( V(l) \) can have the same size as the pupil and \( W(r) \) usually has a smaller size. When the calculated \( t \) is larger than 1, the shape can become larger than the pupil. In this case, only the portion of the shape up to the pupil size is used for optical quality evaluation.

As depicted in Fig. 8A, although normal polynomials can give slightly better optimizer values than even-power-term polynomials, the prescription may be harder to realize. Fig. 8A illustrates a comparison of shapes with normal polynomials (left panel) and with even-power-term polynomials (right panel). The shape on the right panel can be expanded as
\[
-1.6154r + 1.7646r^2 + 1.2646r^3 + 1.9232r^4 + 0.1440r^5 + 0.1619r^6 \quad \text{with a ratio of 0.8}
\]
and the shape on the left panel can be expanded as
\[
-1.1003r^4 + 8.2830r^5 + 0.7305r^6 - 2.2140r^8 \quad \text{with a ratio of 0.9106.}
\]
Both were determined using Downhill Simplex method for a pupil size of 5.6 mm and vergence of 3D with 0.1D step, without residual accommodation. The left panel shows an optimal shape for 6 normal polynomial terms and the right panel shows an optimal shape with 4 EPTP terms. It has been found that polynomials up to the 8th power (4 EPTP terms) appear to give highly satisfactory results.

Fig. 8B shows another comparison of EPTP and non-EPTP expansions. The left panel shows an optimized shape based on an 8th order expansion (EPTP), whereas the right panel shows an optimized shape based on a 3rd order expansion (non-EPTP). In general, shapes derived from an EPTP have a smoother shape with a flat central zone. This flat central zone can correspond to good distance visual performance.

Another comparison of EPTP and non-EPTP expansions is provided in Fig. 8C, which shows optimized (minimized) values with EPTP and non-EPTP expansion for a 4, 5, and 6 mm pupil over a 3D vergence distance. In general, non-EPTP optimization gives a slightly smaller (more optimized) value than EPTP. Sixth-order EPTP appears to give the smallest value for 4 mm and 5 mm pupils and eighth-order EPTP appears to give the smallest value for a 6 mm pupil.

Third-order non-EPTP appears to give the smallest value for 4 mm and 5 mm pupils and fourth-order non-EPTP appears to give the smallest value for a 6 mm pupil.

Using an even-power-term polynomial (EPTP) expansion can result in a smoother shape than a non-EPTP expansion. This smooth shape can be the minimal requirement for good distance vision. In general, \( e^\lambda \)-order or \( S^\lambda \)-order EPTP expansion and \( 3^\lambda \)-order or 4th-order non-EPTP expansion result in good optimized value. Without residual accommodation, larger pupils can be more difficult to optimize than smaller pupils. This is shown, for example, in Fig. 11A.
The optimized multi-focal shape appears to give much more balanced results for the correction of presbyopia than bi-focal and multi-focal shapes.

In addition to using a general polynomial expansion for the optimal surface, it is also possible to use other means of surface expansion. For example, Zernike polynomial expansions may be used. The following formula presents an example of a Zernike polynomial expansion

\[ W(r) = \sum_{i=1}^{n} c_i Z_i(r, \theta) \]

where radially symmetric terms such as \( Z_4, Z_{12}, \) and \( Z_2 \) can be used, and \( c_i \) are free parameters.

Another way of surface expansion is by means of spectral expansion, or Fourier expansion. The following formula presents an example of a Fourier expansion.

\[ W(r) = \sum_{i=1}^{n} c_i F_i(r) \]

where \( c_i \) are free parameters. Fourier expansion is based on the premise that any surface can be decomposed into a set of sinusoidal harmonics with different spatial frequencies. It may not be necessary to expand the surface to very high spatial frequencies.

Discrete surface, or discrete shape entirely, is another type of expansion that can be used in the present invention. Discrete surface can be represented by the following formula

\[ W(r) = W_{ij}, (i = 1,2,\ldots, M ; j = 1,2,\ldots, M) \]

where \( W_{ij} \) are free parameters (\( M \times M \)).

**Inputting A Set Of Patient Parameters Into an Optimizer**

The set of patient parameters can also be referred to as the set of user input parameters. The input parameters may provide certain patient characteristics, such as pupil size and its variations, desired power, and residual accommodation which can be modeled by factors such as gender, age, and race, or which can be measured by instruments.

Residual accommodation can be measured in diopters. Vergence can also be measured in diopters and typically is inversely related to distance, such that a distance of infinity corresponds to a vergence of zero. Similarly, a normal reading distance of 1/3 meters can correspond to a vergence of 3 diopters, and a farther distance of 10 meters can correspond to 0.1 diopters.

It can be useful to model the residual accommodation in the optimization process. The visual quality of the shape can be optimized given a certain set of conditions such as vergence,
residual accommodation, and chromatic aberrations. However, even without a direct correlation between optical surface and the visual quality, it may be convenient to use the minimum root-mean-square (RMS) error to determine the accommodation during different visual vergence. For instance, if no aberrations are present, and there is 2D of residual accommodation, such a patient uses 0.5 D of residual accommodation when visualizing a target at 2 meters. Relatedly, the patient uses all 2D of residual accommodation to view a target at 0.5 meters. The patient would have difficulty viewing targets closer than 0.5 meters, as the residual accommodation is exhausted and no longer available. People with larger pupils or smaller residual accommodation may be harder to treat.

[0171] When aberrations or additional add-on shapes are present, the amount of residual accommodation for different visual vergence may vary. For example, in a patient having 0.5D residual accommodation, with an add-on shape of exactly 1D added to the eye, the eye may not need to accommodate until viewing a target at a distance of one meter. Here, the 1D add-on can cover the first diopter of visual vergence, either entirely or partially. At a large distance, the visual quality may be worse because the eye cannot accommodate in the reverse direction. The techniques of the present invention can be adapted to enhance an optimizer value at low vergence when residual accommodation is assumed.

[0172] When a more complicated add-on shape is used, one way to determine the accommodation is to calculate the available residual accommodation which would minimize the overall RMS.

[0173] Shape optimization can be customized for a patient. The customization can include the patient's pupil sizes at different lighting and viewing conditions, such as bright far viewing, bright near viewing, dim far viewing, and dim near viewing. The optimization can also be based on the patient's residual accommodation, or predicted residual accommodation based on the patient's age, or the patient's vision preference due to for example, their employment or other requirements. That is to say, the customization can put more emphasis on far, near, or intermediate viewing. Similarly, the customization can put more emphasis on dim lighting condition, bright lighting condition or scotopic lighting condition. Further, the optimization can be based on how long the patient wishes to have the correction last. In many ways, presbyopia correction can be a management of compromise. If a patient needs to have excellent correction, he or she might need re-treatment after a couple of years as he or she gets older, when residual accommodation diminishes and/or the pupil size becomes smaller.
**Inputting A Set Of Initial Conditions Into an Optimizer**

[0174] The output result, or optical surface shape, can be sensitive to the choice of the initial condition. In the case of Downhill Simplex method, the initial condition can be the initial N+1 vertices as well as the corresponding initial optimizer values for an N-dimensional problem. In other words, the conditions can be the initial vertices, as well as the value associated with these vertices, for N independent variables. In the case of the Direction Set method, the initial condition can be the initial N direction's unit vector and an initial point for an N-dimensional problem.

[0175] When both or either the initial values for the polynomial coefficients and the pupil ratio are set low, the resulting actual numbers may often be low, especially for the case of pupil ratio. In one example, the initial condition is chosen to be 1.75 for all coefficients and 0.26 for pupil ratio. Figs. 9A-9D show a variety of shapes determined using different initial conditions, as calculated by the Downhill Simplex method. Pupil size of 5.6 mm and vergence of 3D with 0.1D step are assumed. Shape for Fig. 9A is 4.12r-0.235r^2+0.08r^3-6.9r^4+4.81r^5+2.157r^6; for Fig. 9B it is 2.6165r^2+4.1865r^4+6.9123r^6-9.6363r^8; for Fig. 9C it is 1.7926r^2+5.0812r^4-2.163r^6+3.2766r^8-1.1226r^5+1.6845r^6; and for Fig. 9D it is -1.5178r^2+7.2303r^4-2.4842r^6+1.7458r^8+1.8996r^10.

[0176] For the initial conditions, totally random input and fixed ratios may not necessarily help the algorithm to converge to a global minimum or maximum.

[0177] **Implementing An Optimizer To Establish A Customized Optical Shape For The Particular Patient Per The Goal function So As To Treat Or Mitigate A Vision Condition In The Particular Patient**

[0178] The iterative optimization algorithm can be employed to calculate a shape that optimizes the optical quality for the particular patient. For example, in the case of presbyopia the shape can be calculated to optimize distance vision and near vision. In other words, the corrective optical surface shape corresponds to the set of output parameters provided by the optimizer. The output parameters can be the coefficients of polynomials describing the shape, as well as the ratio of diameter of the shape to that of the pupil diameter. These output parameters can define the final customized or optimized optical surface shape. This approach provides a numerical way for general optimization of the optical surface shape for correction or treatment of a vision condition, such as presbyopia. Whether it is for refractive surgery, contact lens, spectacle lens, or intraocular lens, the approach can be very beneficial. For presbyopes with refractive error, the optimal shape can be combined with the shape that corrects for the refractive error, for example the patient's measured wavefront error.
In order to model such deviation in practice, Gaussian distributed noise can be added into the shape so that when noise is present the stability of the algorithm can be tested. For example, Gaussian noise of standard deviation of 0.02 µm OPD can be introduced. This corresponds to nearly 0.06 µm in tissue depth in the case of laser surgery. This is larger than the general RMS threshold for the Variable Spot Scanning (VSS) algorithm for such a shape. Fig. 10 illustrates a comparison of the shapes calculated with a noise-free (dark) condition and with a 0.02 µm standard deviation of Gaussian random noise in OPD on the wavefront. The noise-free case has an optimizer value of 3.008 with 184 iterations and the noisy case has an optimizer value of 2.9449 with 5000 iterations. Both use Downhill simplex method. Pupil size is 5 mm with 3D vergence and 0.1D step. Noise addition can also help to guarantee the stability of the algorithm.

It is also possible to test how the convergence, optimizer value, and shape work with different input pupil sizes. An example of results from such a test is shown in Table 1. For smaller pupil sizes, the shape can cover the whole pupil. That is to say, the shape can be larger than the pupil size. Also, the depth may tend to become smaller with smaller pupils.

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Table 1. Shapes for pupil dependency with 3D vergence and 0.1D step.

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</tbody>
</table>

[0181] As determined by the approach of the present invention, one desirable optical surface shape has a central un-ablated zone and an outside zone that provides improved near vision or reading capability. Based on the example shown in Fig. 7, the central flat zone can be about 1.96 mm in diameter. Because the healing effect may reduce the central zone, the planned flat ablation may need to go beyond 2 mm in order to get a healed flat zone of about 1.96 mm. This can be for a pupil size of about 5.6 mm (natural size). The present invention can also consider practical pupil dependency in the approach. In one example of the present invention, the optical zone can go to about 0.91 times the size of the pupil size, which is about 5.1 mm. Further, the present invention may also incorporate a transition zone such as the VISX standard transition zone technique, as used in variable spot scanning (VSS). What is more, the present invention can also provide a clear mathematical description for the optical surface shape outside of the un-ablated zone.

[0182] Relatedly, Fig. 11C illustrates that there can be a dependency between optimizer value and pupil size. Fig. 11C also shows a preferred optimizer value (optimal). An optimizer value can be a value of the goal function after it is optimized. Theoretically, this value should not be smaller than unity. An optimization, or minimization, algorithm can be used to find values of free parameters such that the optimizer value is as close to unity as possible.

[0183] The present invention can incorporate varying pupil sizes, although presbyopes may tend to have smaller pupil size variation. Because an optimal shape for a fixed pupil size may no longer be optimized if the pupil size changes, the present invention can provide approaches that can allow for pupil size variations. The final optical surface shape can be one that gives an optimal optical quality over a certain vergence range when the pupil size varies over a range.

[0184] To demonstrate how effective a solution is in terms of optical metrics, the MTF can be shown at different spatial frequencies, as illustrated in Figs. 11A-C, which provides optimizer values for various corrections. Apparently the optimal curve gives the minimum (optimized) value for all pupil sizes. Eyes with larger pupils can be more difficult to optimize. What is more,
carefully designed multi-focal correction can be close to optimal, as further illustrated in Figs. 11A-C. That is, the optimizer value for the multi-focal correction can be close to that of the optimized correction, hence the results are quite similar. This outcome is also illustrated in Fig. 13. The lower regression line in Fig. 11C can set the practical limit for the optimizer value.

[0185] In another approach, to demonstrate how effective a solution is in terms of optical metrics, the compound MTF can be plotted, as shown in Figs. 9A-B. Here, the compound MTF for various treatments for a 5 mm pupil over a 3D vergence is plotted. It can be beneficial to optimally balance the level of compound MTF at every vergence distance or over the desired vergence. Fig. 9C shows a comparison of bi-focal and optimal corrections, with a simulated eye chart seen at different target distances, assuming a 5mm pupil with no accommodation. The eye chart has 20/100, 20/80, 20/60, 20/40, and 20/20 lines, respectively.

[0186] Fig. 10 is a simulated eye chart seen at different target distances, and compares an optimized case (bottom) to no correction (top line); reading glasses (second line); bi-focal lenses (inner half for reading and outer half for distance, third line); and multi-focal lenses (pupil center for reading with maximum power and pupil periphery for distance with zero power and linear power change in between, four line). The effects of the optimization can be clearly seen by the comparison. No accommodation or refractive error is assumed in any of the cases. The eye chart has 20/100, 20/80, 20/60, 20/40, and 20/20 lines.

[0187] Using the above approaches, it is possible to obtain a shape that is not only larger than the pupil size, but that can also be practically implemented. Often, only the portion of the shape inside the pupil may be evaluated for optical quality, although this is not a requirement. For example, the entire zone over the pupil can remain un-ablated, but there may be a zone outside the pupil that is ablated. In this way, distance vision is not affected, but for near vision, there can be an advantage from light coming outside of the pupil due to greatly deformed periphery. A goal function based on geometrical optics, or ray tracing, can be useful to determine such shapes.

[0188] Residual accommodation can also affect the optimization result, because it can remove some of the ripples on the combined wavefront at any vergence.

[0189] The approaches of the present invention can be implemented on a variety of computer systems, including those with a 200MHz CPU with 64MB memory, and typically will be coded in a computer language such as C or C++. Simulations have successfully been run on a laptop computer with a 1.2GHz CPU with 256 MB memory. The techniques of the present invention can also be implemented on faster and more robust computer systems.
The present invention includes software that implements the optimizer for practical applications in a clinical setting. The optimizer will often comprise an optimizer program code embodied in a machine-readable medium, and may optionally comprise a software module, and/or a combination of software and hardware. As shown in Figs. 14-16, the software interface can comprise two primary panels: the parameter panel and the display panel. The parameter panel can be split into two sub-panels: optimization and verification. The display panel can also be split into two sub-panels: graph panel and image panel. The software can also include a menu bar, a tool bar, and a status bar. In the tool bar, small icons can be used for easy access of actions.

The optimization sub-panel can include a number of parameter units. For example, a first parameter unit can be the pupil information group. In the examples shown in Figs. 14-16, the user or operator can give four different pupil sizes for a specific eye. More particularly, the pupil information group includes the pupil size in (a) bright distance viewing condition, (b) bright near viewing condition (e.g. reading), (c) dim light distance viewing condition, and (d) dim light near viewing condition (e.g. reading). These different pupil sizes can be used in the optimization process.

A second parameter unit in the optimization sub-panel can be the display group. In the examples shown in Figs. 14-16, the user or operator has three different choices for the display, including (a) none, (b) shape, and (c) metric. The display group can provide instruction to the software regarding what kind of display is desired for each iteration. For instance, none can mean no display, shape can mean displaying the current shape, and metric can mean displaying the current optical metric curve over the desired vergence for this current shape. The choices can be changed during the optimization procedure, and in this sense it is interactive.

A third parameter unit in the optimization sub-panel can be the optical metric group. In the examples shown in Figs. 14-16, the user has five different choices for the metric, including (a) Strehl ratio, (b) MTF at a desired spatial frequency, (c) encircled energy at a desired field of view, (d) compound MTF (CMTF) with a set of specific combinations, which could be any number of MTF curves at different spatial frequencies, and when the "auto" check box is checked, it can use a default CMTF with three frequencies, such as, for example: 10, 20 and 30 cycles/degree, and (e) the MTF volume up to a specific spatial frequency. 25% CMTF over the vergence appears to be an example of a good target value for optimization.

A fourth parameter unit in the optimization sub-panel can be the optimization algorithm group. In the examples shown in Figs. 14-16, the user has three different choices for the optimization algorithm employed by the optimizer, including (a) the Direction Set (Powell's)
method, (b) the Downhill Simplex method, and (c) the Simulated Annealing method. The
optimizer can employ a standard or derived algorithm for function optimization (minimization or
maximization). It can be a multi-dimensional, non-linear, and iterative algorithm.

[0195] A number of other parameters can be included in the optimization sub-panel. As shown
in Figs. 14-16, these other parameters can be implemented separately (optionally as a ComboBox)
with a number of choices for each. These can include parameters such as (a) the number of terms
of the polynomial expansion, (b) the frame size, (c) the PSF type (monochromatic, RGB, or
polychromatic), (d) whether the shape is EPTP or non-EPTP, (e) the vergence requirement, (f) the
vergence step, and (g) the residual accommodation. The software can include a StringGrid table
that displays the polynomial coefficients, the PAR value, the optimizer value, as well as the current
number of iterations. These numbers can be updated every iteration.

[0196] The verification sub-panel can include a number of parameter units. For example, a first
parameter unit can be the "which" group. In the examples shown in Figs. 14-16, the operator can
use this group to select whether to use built-in eye chart letters, or an entire eye chart or a scene. A
second parameter unit in the verification sub-panel can be the left image group. The user can
make a selection in the left image group from PSF and imported scene. A third parameter unit is
the right image group, wherein the user can make a selection from imported scene, and blur at
near. The two image display groups are for the left and right subpanels in the image subpanel.

[0197] As further illustrated in Figs. 14-16, the ComboBox for letter can provide a list of
different eye chart letters, and the VA ComboBox can provide the expected visual acuity, from
20/12 to 20/250. The Contrast ComboBox can provide a list of contrast sensitivity selections,
from 100% to 1%. Two check box can also be included. The Add check box, once checked, adds
the presbyopia to the simulated eye. The Test check box, when checked, performs the distance
(zero vergence). At the bottom, there is a slider with which all the saved images (e.g. PSF and
convolved images) can be reviewed.

[0198] There are many factors that can affect the pupil size, and these factors can be considered
optimization approaches of the present invention. For example, the shape can be customized for
various lighting and accommodation conditions. As shown in Fig. 17, and further discussed in
Table 2, pupil size can change with lighting conditions. Each of the presbyopia-mitigating and/or
treating methods, devices, and systems described herein may take advantage of these variations in
pupil size. A pupil size of a particular patient will often be measured, and multiple pupil sizes
under different viewing conditions may be input for these techniques.
Table 2

<table>
<thead>
<tr>
<th>dim distance</th>
<th>bright distance</th>
</tr>
</thead>
<tbody>
<tr>
<td>5 mm</td>
<td>3.5 mm</td>
</tr>
<tr>
<td>4 mm</td>
<td>2.5 mm</td>
</tr>
</tbody>
</table>

[0199] A patient can also have a task-related vision preference that correlates with lighting conditions, such as those described in Table 3, and the customization can be based upon these task-related preferences.

Table 3

<table>
<thead>
<tr>
<th>cd/m²</th>
<th>lighting condition</th>
</tr>
</thead>
<tbody>
<tr>
<td>30</td>
<td>subdued indoor lighting</td>
</tr>
<tr>
<td>60</td>
<td>display-only workplaces</td>
</tr>
<tr>
<td>120</td>
<td>typical office</td>
</tr>
<tr>
<td>240</td>
<td>bright indoor office</td>
</tr>
<tr>
<td>480</td>
<td>very bright; precision indoor tasks</td>
</tr>
<tr>
<td>960</td>
<td>usual indoors</td>
</tr>
<tr>
<td>1920</td>
<td>bright afternoon</td>
</tr>
</tbody>
</table>

[0200] Fig. 18 illustrates that pupil size can change with accommodation, and Fig. 19 illustrates a comparison of corrections by providing optimizer values for various accommodations. With 3 or more diopters of residual accommodation, the optimizer value can achieve a limit of about 1.0, regardless of the pupil size. Typically, a larger amount of residual accommodation can correspond to a smaller optimizer value after optimization. The limit line can correspond to an optimizer value of about 5.0. In other words, an optimizer value of about 5.0 can be viewed as a good practical limit. Either there can be a smaller pupil, or a larger amount of residual accommodation, in order to optimize such that all vergence distances have good visual performance.

[0201] Figs. 20 and 21 show optimizations under various accommodation conditions. Figs. 21A and 21B show CMTF and optimizer values when pupil size changes and Residual Accommodation (RA) are modeled. Fig. 21C shows simulated eye charts seen at different target distances after optimization, all assuming a 5mm maximum pupil size. Each eye chart has 20/100, 20/80, 20/60, 20/40, and 20/20 lines. The top line simulates no accommodation and no pupil size changes. The middle line assumes no accommodation but the pupil size changes from 5mm (dim distance) to 2.5mm (bright near). In the bottom line, the simulation assumes 1D accommodation with pupil size changes from 5mm (dim distance) to 2.5mm (bright near).

[0202] Fig. 22 shows CMTF values for various corrections. A 5mm pupil eye is assumed, along with a smallest pupil size of 2.5mm (bright light reading condition) and a 1D residual
accommodation. Fig. 23 compares bi-focal, optimal, and multi-focal corrections, under the assumption of a one diopter residual accommodation. These simulated eye charts are seen at different target distances after optimization. 1D accommodation and a 5mm pupil changes from 5mm (dim distance) to 2.5mm (bright near) are assumed. The eye chart has 20/100, 20/80, 20/60, 20/40, and 20/20 lines, respectively. Fig. 24 illustrates a simulated eye chart seen at different target distances. The data in this figure based on the assumption that the pupil size decreases from 5 mm to 2.5 mm, and there is a 1 diopter residual accommodation in all cases.

[0203] The customized shape methods and systems of the present invention can be used in conjunction with other optical treatment approaches. For example, co-pending U.S. provisional patent application number 60/431,634, filed December 6, 2002 (Attorney Docket No. 018158-022200US) and co-pending U.S. provisional patent application number 60/468,387 filed May 5, 2003 (Attorney Docket No. 018158-022300US), the disclosures of which are hereby incorporated by reference for all purposes, describe an approach to defining a prescription shape for treating a vision condition in a particular patient. The approach involves determining a prescriptive refractive shape configured to treat the vision condition, the prescriptive shape including an inner or central "add" region and an outer region. The approach also includes determining a pupil diameter of the particular patient, and defining a prescription shape comprising a central portion, the central portion having a dimension based on the pupil diameter, the inner region of the prescriptive refractive shape, and an attribute of at least one eye previously treated with the prescriptive refractive shape.

[0204] Accordingly, the present invention can include a method for determining a customized shape that includes a scaled central portion as described above, the customized shape giving results at least as good or better than previously known methods.

[0205] Systems

[0206] The present invention also provides systems for providing practical customized or optimized prescription shapes that mitigate or treat vision conditions such as presbyopia in particular patients. The systems can be configured in accordance with any of the above described methods and principles.

[0207] For example, as shown in Fig. 25, a system 100 can be used for reprofiling a surface of a cornea of an eye 150 of a particular patient from a first shape to a second shape having correctively improved optical properties. System 100 can comprise an input 110 that accepts a set of patient parameters, a module 120 that determines an optical surface shape for the particular patient based
on the set of patient parameters, using a goal function appropriate for a vision condition of an eye, a processor 130 that generates an ablation profile, and a laser system 140 that directs laser energy onto the cornea according to the ablation profile so as to reprofile a surface of the cornea from the first shape to the second shape, wherein the second shape corresponds to the prescription shape.

[0208] Referring to Fig. 26A, the present invention will often take advantage of the fact that the eye changes in two different ways with changes in viewing distance: the lens changes in shape so as to provide accommodation, and the pupil size simultaneously varies. Accommodation and pupillary constriction work in unison in normal healthy eyes when shifting from a far to a near viewing distance, and a fairly linear relation may exist between at least a portion of the overlapping constriction and accommodation ranges, but the effect may vary significantly among subjects (from 0.1 to 1.1 mm per dioptr). Moreover, when the stimulus for accommodation is increased beyond the eye's ability to change its refraction, the relationship between accommodation of the lens and pupillary constriction may be curvilinear as shown.

[0209] While they work in unison, pupillary constriction and accommodation are not necessarily linked. These two functions may proceed independently, and may even work in opposite directions, particularly when the patient is simultaneously subjected to large variations in light intensity with changes in viewing distance. Nonetheless, prescriptions for presbyopia can take advantage of the correlation between pupil dimension and viewing distance for a particular patient. The effective time span for a presbyopia-mitigating prescription may also be extended by accounting for gradual changes in pupil dimension over time (such as the gradual shrinkage of the pupil as one ages) with the concurrent gradual decrease in the accommodation. Details regarding constriction of the pupil were published in a book entitled The Pupil by Irene E. Loewenfeld (Iowa State University Press, 1993).

[0210] Referring now to Fig. 26B and 26C, if we assume that we can tailor a beneficial overall optical power for the eye as it changes to different pupil sizes, we may first want to identify a relationship between this desired optical power and pupil size. To determine what powers would be desirable for a particular patient at different viewing conditions, we might measure both the manifest sphere and corresponding pupil sizes of that patient at a variety of different viewing conditions. The manifest sphere may then be used as our desired or effective power to be used for treating presbyopia, as detailed below. The desired optical power might also be determined from the measured manifest, for example, with desired power being a function of the manifest to adjust for residual accommodation and/or anticipated aging effects or the like. In either case, these patient-specific measurements can be the basis for determining desired powers for associated pupil
sizes of that patient, such as at the four points illustrated in Fig. 26B. Fewer or more points might also be used.

[0211] Alternatively, manifest sphere and pupil size for a population of different patients who have been successfully treated with a given presbyopia prescriptive shape may be plotted, and a correlation derived from this empirical data, as schematically illustrated in Fig. 26C. Still further approaches may be employed, including combinations where a population of patients having differing pupil sizes are used to derive an initial correlation, which is subsequently refined with multiple measurements from at least one patient (and often a plurality of patients). Regardless, the relationship between our desired optical power and the pupil size can be determined. As will be clear from the detailed description below, constriction of the pupil at differing viewing distances then allows the overall power of the eye to be altered by the pupillary constriction, despite a loss in the flexibility of the lens. For example, we can employ a peripheral portion of the ocular system having a different power than a central portion. By understanding the variations of these often aspherical optical systems with changing pupil sizes, we can provide good optical performance throughout a range of viewing distances.

[0212] The following description will first provide techniques and devices for iteratively optimizing refraction for treatment of presbyopia. This is followed by a brief review of an exemplary initial laser ablation shape for mitigation of presbyopia, which is in turn followed by an explanation of techniques for optimizing that shape (or other shapes), often using empirical and/or patient-specific information to scale the shape. Generalized analytical and numerical techniques for determining or selecting appropriate presbyopia mitigating prescription shapes will then be provided.

[0213] Defining A Scaled Prescription shape For A Vision Condition

[0214] Determining a prescriptive prescription shape

[0215] Certain prescriptive refractive shapes are effective in treating vision conditions, and it is possible to provide an efficient prescription shape by scaling a shape to the particular patient being treated. Optical shapes can be scaled based on data collected from subjects previously treated with a uniform prescriptive optical shape, such as measured manifest powers for different pupil sizes. Shapes may also be scaled based on the desired overall optical power of the eye under differing viewing conditions.

[0216] It is useful to select or construct an initial prescriptive refractive shape appropriate for the vision condition. For example, prescriptive treatment shapes such as those shown in Fig. 28 have
been found to provide a range of good focus to the eye so as to mitigate presbyopia. This
particular prescriptive shape is the sum of two component shapes: a base curve treatment defining
an outer region having a diameter of about 6.0 mm, and a refractive add defining an inner region
having a diameter of about 2.5 mm. Prescriptive shapes such as this can provide a spherical power
add ranging from about 1.0 diopters to about 4.0 diopters at the inner region. Further, the
spherical power add can be about 3.1 diopters. Combining the inner and outer regions, the overall
prescriptive refractive shape can be aspheric. It is appreciated, however, that the dimensions and
properties of a prescriptive shape can vary depending on the intended purpose of the shape.

[0217] Treatment of presbyopia often involves broadening the focus range of the eye. Referring
to Fig. 29, in an emmetropic eye a focal length of the optical system results in a point of focus 10
that produces a sharp image. At this point, the refractive power of the cornea and lens is matched
to the length of the eye. Consequently, light rays 20 entering the eye converge on the retina 30. If
there is a difference between the refractive power and the length of the eye, however, the light rays
can converge at a point 40 in front of or behind the retina, and the image formed on the retina can
be out of focus. If this discrepancy is small enough to be unnoticed, it is still within the focus
range 50 or depth of focus. In other words, the image can be focused within a certain range either
in front of or in back of the retina, yet still be perceived as clear and sharp.

[0218] As shown in Fig. 30, when an object is at a far distance 60 from the eye, the light rays 20
converge on the retina 30, at focal point 10. When the object is moved to a near distance 70, the
light rays 20' converge at a focal point 80 beyond the retina. Because the image is outside of the
depth of focus 50, the image is perceived to be blurred. Through the process of accommodation,
the lens changes shape to increase the power of the eye. The power increase brings the focal point
80 back toward the retina as the eye attempts to reduce the blur.

[0219] In the presbyopic eye the accommodative mechanism may not work sufficiently, and the
eye may not be able to bring the focal point to the retina 30 or even within the range of focus 50.
In these circumstances, it is desirable to have an optical system having a broadened focus range
50'. One way to achieve this is by providing an optical system with an aspheric shape. The
aspheric shape, for example, can be ablated on a surface of the eye, the surface often comprising a
stromal surface formed or exposed by displacing or removing at least a portion of a corneal
epithelium, or a flap comprising corneal epithelium, Bowman's membrane, and stroma. Relatedly,
the shape can be provided by a correcting lens. In some optical systems, only a portion of the
shape may be aspheric. With an aspheric shape, there is not a single excellent point of focus.
Instead, there is greater range of good focus. The single best focus acuity is compromised, in order
to extend the range of focus. By extending the range of focus 50 to a broadened range of focus 50', there is an improvement in the ability to see both distant and near objects without the need of 3D or more in residual accommodation.

[0220] Without being bound by any particular theory, it is believed that the power add of the inner region depicted in Fig. 28 provides a myopic effect to aid near vision by bringing the near vision focus closer to the retina, while the outer region remains unaltered for distance vision. In this sense the application of this prescriptive shape is bifocal, with the inner region being myopic relative to the outer region. Put another way, the eye can use the inner region for near vision, and can use the whole region for distance vision.

[0221] In a laser ablation treatment, the prescriptive refractive ablation shape can have fairly abrupt changes, but post ablation topographies may show that healing of the eye can smooth the transitions. The shape can be applied in addition to any additional required refractive correction by superimposing the shape on a refractive corrective ablation shape. Examples of such procedures are discussed in co-pending U.S. patent application number 09/805,737, filed March 13, 2001, the disclosure of which is herein incorporated by reference for all purposes.

[0222] Alternative presbyopia shapes may also be scaled using the techniques described herein, optionally in combination with other patient customization modifications, as can be understood with reference to U.S. Provisional Patent Application Nos. 60/468,387 filed May 5, 2003, 60/431,634, filed December 6, 2002, and 60/468,303, filed May 5, 2003, the disclosures of which are herein incorporated by reference for all purposes. Alternative presbyopia shapes may include concentric add powers along a peripheral or outer portion of the pupil, along an intermediate region between inner and outer regions, along intermittent angular bands, or the like; asymmetric (often upper or lower) add regions, concentric or asymmetric subtrace or aspheric regions, and the like. The present application also provides additional customized refractive shapes that may be used to treat presbyopia.

[0223] Determining a pupil diameter of the particular patient

[0224] When scaling a refractive shape to treat a particular patient, it is helpful to determine the pupil diameter of the particular patient to be treated. Several methods may be used to measure the pupil diameter, including image analysis techniques and wavefront measurements such as Wavescan® (AMO Manufacturing USA, LLC in Milpitas, CA) wavefront measurements. The size of the pupil can play a role in determining the amount of light that enters the eye, and can also have an effect on the quality of the light entering the eye. When the pupil is very constricted, a relatively small percentage of the total light falling on the cornea may actually be allowed into the
eye. In contrast, when the pupil is more dilated, the light allowed into the eye may correspond to a greater area of the cornea. Relathedly, the central portions of the cornea have a more dominant effect on the light entering the eye than do the peripheral portions of the cornea.

Pupil size can have an effect on light quality entering the eye. When the pupil size is smaller, the amount of light passing through the central portion of the cornea is a higher percentage of the total light entering the eye. When the pupil size is larger, however, the amount of light passing through the central portion of the cornea is a lower percentage of the total light entering the eye. Because the central portion of the cornea and the peripheral portion of the cornea can differ in their refractive properties, the quality of the refracted light entering a small pupil can differ from that entering a large pupil. As will be further discussed below, eyes with different pupil sizes may require differently scaled refractive treatment shapes.

An inner region of the prescriptive refractive shape

Experimental data from previously treated eyes can provide useful information for scaling a refractive treatment shape for a particular patient. For example, a refractive shape for a particular patient can be scaled based on certain characteristics or dimensions of the shape used to treat the eyes of the subjects. One useful dimension of the above-described presbyopic prescriptive shape is a size or diameter of inner region or refractive add. It is possible to scale a treatment shape for a particular patient based on the diameter of the refractive add of the prescriptive shape. Alternative techniques might scale a power of an inner, outer, or intermediate region, a size of an outer or intermediate region, or the like.

If the refractive add diameter is small, it can occupy a smaller percentage of the total refractive shape over the pupil. Conversely, if the refractive add diameter is large, it can occupy a greater percentage of the total refractive shape over the pupil. In the latter case, because the area of the periphery is relatively smaller, the distance power is diminished. In other words, the area of the add is taking up more of the total refractive shaped used for distance vision.

An attribute of a set of eyes previously treated with the prescriptive refractive shape

As noted above, experimental data from previous prescriptive eye treatments can be useful in scaling a treatment for a particular individual. When scaling a presbyopia treatment shape, it is helpful to identify a pupil diameter measure from among a set of previously treated eyes having a fixed treatment size that corresponds to both good distance and near sight. It is possible to use acuity and power measurements from the set of treated eyes to determine such a
pupil diameter. The fixed treatment size (e.g. 2.5 mm inner region) can then be said to be appropriate for this identified pupil diameter.

**Figs. 31 and 32** illustrate the effect that pupil size can have on distance acuity and near acuity in subjects treated with a prescriptive refractive shape, for example a shape having a 2.5 mm central add zone of -2.3 diopters. Referring to **Fig. 31**, pupil size values were obtained from a group of subjects as they gazed into infinity under mesopic or dim light conditions. The 6-month uncorrected distance acuity values were obtained from the same group of subjects under photopic conditions. Referring to **Fig. 32**, pupil size values were obtained from a group of subjects as they gazed at a near object under mesopic or dim light conditions. The 6-month uncorrected near acuity values were obtained from the same group of subjects under photopic conditions.

One way to determine an optimal pupil diameter measure is by superimposing a near acuity graph over a distance acuity graph, and ascertaining the pupil diameter that corresponds to the intersection of the lines.

Another way to determine a pupil diameter that corresponds to both good distance and near acuity is to define each of the slopes mathematically:

\[
\text{Near acuity} = -2.103 + 0.37879 \times \text{Pupil size (Dim)} \quad \text{(Fig. 27)}
\]

\[
\text{Distance acuity} = 0.40001 - 0.0677 \times \text{Pupil size (Dim)} \quad \text{(Fig. 26)}
\]

By setting the two equations from the graphs equal, it is possible to solve for the intersection point:

\[
-2.103 + 0.37879 \times \text{Pupil size (Dim)} = 0.40001 - 0.0677 \times \text{Pupil size (Dim)}
\]

\[
\text{Pupil size (Dim)} = 2.4/0.45 = 5.33 \text{ mm}
\]

An optimum overlap can occur in a range from between about 4.0 mm to about 6.0 mm. Further, an optimum overlap can occur in a range from between about 5.0 mm to about 5.7 mm. These measurements may correspond to a pupil diameter measure from the set of previously treated eyes that corresponds to both good distance and near vision when the diameter of the central add region is 2.5 mm.

**Defining a refractive shape for treating a particular patient acuity as a function of pupil size**

The present invention provides methods and systems for defining a prescription for treating a vision condition in a particular patient, with the prescription optionally comprising a refractive shape. Such a method can be based on the following features: (a) a prescriptive refractive shape configured to treat the vision condition, including an inner region thereof; (b) a
pupil diameter of the particular patient, and (c) an attribute of a set of eyes previously treated with
the prescriptive shape.

[0238] For example, the prescriptive shape can be the shape described in Fig. 28. The inner
region of the shape can be a refractive add, having a diameter of 2.5 mm. For illustrative purposes,
a pupil diameter of the particular patient of 7 mm is assumed. The attribute of a set of previously
treated eyes can be the pupil diameter of the eyes that corresponds to both good distance and near
vision, such as the exemplary 5.3 mm treated pupil diameter shown in Figs. 31 and 32. Thus, a
ratio of the prescriptive refractive add to treated pupil (PAR) can be expressed as 2.5/5.3.

[0239] The PAR can be used in conjunction with the pupil diameter of the particular patient to
scale the refractive shape. For example, a central portion of the scaled refractive shape can be
calculated as follows.

\[
\text{central portion diameter} = \text{PAR} \times \text{pupil diameter of particular patient}
\]

[0240] Given the example above, the diameter of a central portion of the scaled refractive shape
for treating the particular patient is:

\[
(2.5/5.3) \times 7 \text{ mm} = 3.3 \text{ mm}
\]

[0241] In this example, this scaled central portion can correspond to the diameter of the
refractive add of the defined refractive shape. It should be appreciated that the refractive shape
and the central portion of the refractive shape can alternately be spheric or aspheric. For example,
the refractive shape can be aspherical, and the central portion of the refractive shape can be
aspherical; the refractive shape can be spherical and the central portion of the refractive shape can
be spherical; the refractive shape can be aspherical, and the central portion of the refractive shape
can be spherical; or the refractive shape can be spherical, and the central portion of the refractive
shape can be aspherical.

[0242] As shown above, the PAR can be about 2.5/5.3, or 0.47. It will be appreciated that the
PAR can vary. For example, the PAR can range from between about 0.35 and 0.55. In some
embodiments, the PAR may range from about 0.2 to about 0.8. Optionally, the PAR can range
from about 0.4 to about 0.5. Further, the PAR can range from about 0.43 to about 0.46. It will
also be appreciated that the ratios discussed herein can be based on area ratios or on diameter
ratios. It should be assumed that when diameter ratios are discussed, that discussion also
contemplates area ratios.

[0243] Power as a function of pupil size
In another example, the attribute of a set of previously treated eyes can be the pupil diameter of the eyes that correspond to both good distance and near values for spherical manifest. A group of individuals with varying pupil sizes were treated with the same prescriptive refractive shape, the shape having a constant presbyopic refractive add diameter of approximately 2.5 mm. Pupil sizes were obtained on a Wavescan® device. The Spherical Manifest at 6 months post-treatment is shown as a function of the pupil size in Fig. 33. Here, the spherical manifest represents the effective distance power as the result from the total prescriptive shape, including the inner region and outer regions of the shape.

As Fig. 33 illustrates, for a given prescriptive treatment shape, the effect that the shape has on the individual's manifest can depend on the individual's pupil diameter. Depending on the pupil size of the treated subject, the refractive add will have different relative contribution to the power. And due to the varying pupil sizes, the prescriptive refractive add to treated pupil ratio (PAR) may not be constant. Thus, with the same prescriptive treatment, the effective power can vary among different patients. In a simplified model, the power change from the central portion of the treated eye to the periphery can be assumed to be linear. This simplification can be justified by the data. The change in power can be represented by the following formula, expressed in units of diopters.

\[
MRS \text{ (Effective Distance Power)} = -2.87 + 0.42 \times \text{Pupil size (Dim)} \text{ [diopters]}
\]

The rate change in effective power is 0.42D per mm for distance vision. It has been shown that the pupil diameter can change at a rate of approximately 0.45D per mm. The add power is -2.87 diopters.

Without being bound by any particular theory, it is thought that due to the asphericity of the central add, there can be a linear relationship between the effective distance power and the pupil diameter. Accordingly, is it possible to characterize the ratio of effective distance power versus pupil diameter with the following linear core equation, where \( C_0 \) and \( A \) are constants.

Equation A: \[
\text{Effective Distance Power} = C_0 + A(\text{pupil\_diameter})
\]

In individuals having smaller pupil diameters, the contribution of the outer region of the prescriptive shape is diminished; the manifest refraction is more myopic and the effective power is smaller. And whereas a lower MRS value can correspond to a more myopic refraction, a higher MRS value can correspond to a less myopic refraction. The manifest refraction, which can be expressed in terms of power, is often proportional to distance vision, which can be expressed in terms of acuity or logarithm of the minimum of angle of resolution (logMAR).
As discussed above, a PAR can be determined based on acuity measurements as a function of pupil size. In an analogous manner, it is possible to determine a PAR based on power measurements as a function of pupil size.

Skewing

The Effective Distance Power Equation A above represents one approach to finding a good approximation to customize the refractive shape size. In sum, the intersection of a distance version of the equation and a near version of the equation is solved to determine a pupil diameter measure, which forms the denominator for the PAR (prescriptive shape add diameter/pupil diameter of treated eye). By adjusting the PAR, it is possible to adjust the shape to achieve emmetropia or other refractive states.

Altering the size of the prescriptive shape add

Referring to Fig. 33, a treated pupil diameter of about 5.4 mm has a spherical manifest of about -0.6 diopters. If the size of the prescriptive shape add is made bigger, the line can be shifted downward. Consequently, the effect in a particular patient treated with the scaled refractive shape would be a more myopic spherical manifest of -2.0, for example. On the other hand, if the size of the add is made smaller, the line can be shifted upward, and the effect would be a spherical manifest of -0.2, for example. As the diameter of the add decreases, the manifest of the particular patient treated with the scaled refractive shape becomes more skewed to better distance sight. As the diameter of the add increases, the manifest becomes more skewed to better near sight.

Fixing the PAR

It is possible to set the near manifest for all patients by fixing the PAR. Referring to the example of Figs. 31 and 32 (where the Equation A intersection is about 5.3 mm), a ratio of 2.5/5.3 mm can rotate these near and distance lines toward horizontal, about the 5.3 mm point. In other words, an analysis of particular patients treated with a PAR of 2.5/5.3 is expected to result in manifest versus pupil size plots having lines that are more horizontally oriented. Thus each patient would be expected to have similar near manifest. Alternatively, it is possible to choose a different point of rotation to optimize distance manifest over near manifest, or vice versa. For example, by choosing a 5.0 mm point for rotation, better near manifest can be provided at the expense of the distance manifest.

When comparing the graphs of Figs. 31 and 32 the distance acuity and near acuity slopes can vary. As shown in these figures, near vision changes at a slightly higher rate than distance vision. In other words, near vision appears to be more sensitive to changes in pupil diameter than
distance vision. An adjustment was made to near measurements in Fig. 32 to offset a distance correction used during the measurement.

[0257] Non-linear models

[0258] The effective distance power versus pupil diameter can also be expressed by the following non-linear equation.

Equation B : Power = C_o + A(pupil_diameter) + B(pupil_diameter)^2 + C(pupil_diameter)^3 + ...

[0259] where Co, A, B, and C are constants. This equation is only one of many that can be used to model the desired relationship. Similar non-linear equations can be used to model desired effective power, as discussed below. Also, both linear and non-linear equations can be used to model target manifest, as discussed below.

[0260] Target manifest (acuity as a function of power)

[0261] The target manifest or desired power at a particular viewing distance may or may not be emmetropic (0 diopters). For example, near sight may be improved by a manifest which is slightly myopic. Following an analysis similar to that discussed above for pupil size dependency, an optimum target refraction can be calculated based on acuity as a function of power in a set of eyes treated with the prescribed refractive shape. Figs. 34 and 35 show the distance and near acuity as a function of manifest, respectively. Distance and near acuity versus manifest can be expressed by the following non-linear equations.

Near Acuity = A_o + A(Manifest) + B(Manifest)^2 + C(Manifest)^3 +...

Distance Acuity = A_o + A(Manifest) + B(Manifest)^2 + C(Manifest)^3 +...

[0262] Applying a first order approximation to the above equations, and using measurements from previous data, the near and distance acuity as a function of manifest can be expressed as follows.

Near Acuity = 0.34 + 0.67 (Manifest)

Dist Acuity = -0.04 - 0A3(Manifest)

[0263] The intersection between the two functions can be solved as follows.

0.34 + 0.67 (Manifest) = -0.04 - 0.13 (Manifest)

Manifest = (-0.04 - 0.34) / 0.67 + 0.13 = - 0.48 [Diopters]

[0264] The point where the two lines meet is about -0.5D. Therefore, it can be useful to set the target manifest to -0.5D. The target manifest equations can be refined based on additional data.
collected from those patients that are treated with the refractive shape. As noted above in reference to Fig. 28, a prescriptive shape may be the sum of a base curve treatment and a central refractive add. It is possible to change the base shape to compensate for any power offset contributed by the central refractive add to the central manifest.

5 [0265] PAR refinements applied to particular patients

[0266] As additional data is accumulated, it is possible to calculate the higher order terms of Equation B. More particularly, it is possible to calculate the higher order terms from additional subjects who have been treated with refractive shapes corresponding to constant and linear term adjustments. For example, a group of patients can be treated according to the PAR of 2.5/5.3 discussed above, and based on their results, the PAR can be further refined.

[0267] A group of patients had adjustments made to their prescriptive presbyopic shape based on results from the analysis discussed above. The patients were treated with shapes based on a constant PAR of 2.5/5.6 as applied to the central add shape, with a target manifest of -0.5D. These adjustments rotate the equation about the 5.6 mm line toward horizontal because the near effect is a constant. For example, a 5 mm pupil patient has the same near correction as a 6 mm pupil patient, which means that their near acuity should be the same, i.e. a plot of the near acuity versus pupil size will be a substantially flat line. Figs. 36 and 37 show the result of these adjustment on this group of patients. As predicted, the lines rotated. The distance acuity of 7 of 8 of these patients was 20/20 (logMAR 0) or better, and the 8th was 20/20+2. Their near acuity slopes have also flattened, with 7/8 patient having simultaneous 20/32 -2 acuity or better, and the 8th 20/40. Table 4 summarizes the acuity and power measures.

Table 4

<table>
<thead>
<tr>
<th>Acuity</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Near acuity</td>
<td>0.19 ± 0.1</td>
</tr>
<tr>
<td>Distance acuity</td>
<td>-0.08 ± 0.08</td>
</tr>
<tr>
<td>MRS</td>
<td>-0.19 + 0.26</td>
</tr>
</tbody>
</table>

[0268] This PAR adjusted group has, which is a good result for a presbyopia treatment.

[0269] Optimizing A Refractive Shape For A Vision Condition

25 [0270] It is possible to define customized refractive shapes such that they are optimized to treat a particular patient. In one approach to defining an optimized refractive shape, the power of the refractive shape may be based on the central power add of a prescriptive shape, and the power change requirement of the particular patient. Other approaches may involve deriving an
appropriate prescription so as to provide a desired overall effective power of the eye at different viewing conditions, again by taking advantage of the changes in pupil size.

[0271] Determining a desired central power add of a prescriptive refractive shape configured to treat the vision condition

5 [0272] A prescriptive shape can be selected for treating the vision condition of the particular patient. For example, the prescriptive shape shown in Fig. 28 can be selected for treating a particular patient having presbyopia. As previously discussed, the central power add of this exemplary prescriptive shape can be about -3.1 diopters.

[0273] Determining a power change of a particular patient

10 [0274] The desired power change of a particular patient can vary widely, and often depends on the patient’s desired treatment or a recommendation from a vision specialist. For example, the desired power change of a particular patient having presbyopia can be about -2.5 diopters. The desired power change may be linear or non-linear.

[0275] Determining a pupil diameter parameter of the particular patient

15 [0276] When defining a refractive shape for treating a vision condition in a particular patient, it is helpful to determine the pupil diameter parameter of the particular patient. Pupil diameters can be measured by, for example, a pupillometer. Pupil diameter parameters can involve, for example, the patient’s pupil diameter as measured under certain distance and lighting conditions, such as under photopic conditions while the patient gazes at infinity (distance-photopic). Pupil diameter parameters can also involve pupil diameter measurements under other conditions such as distance-mesopic, distance-scotopic, near-photopic, near-mesopic, or near-scotopic. Still further additional measurements at other viewing conditions, such as at intermediate distances and/or moderate lighting conditions, may also be measured. Often, pupil diameter parameters will be based on two pupil diameter measurements. For example, a pupil diameter parameter can be the value of the particular patient's pupil diameter at distance-photopic minus the patient's pupil diameter at distance scotopic. According to this example, if the distance-photopic pupil diameter is 0.7 mm and the distance-scotopic pupil diameter is 0.2 mm, then the pupil diameter parameter is 0.7 mm minus 0.2 mm, or 0.5 mm.

[0277] Defining a refractive shape configured to treat the particular patient, the power of the refractive shape at a given diameter based on: the central power add of the prescriptive refractive shape, the power change requirement of the particular patient, and the pupil diameter parameter of the particular patient
When defining the refractive treatment shape, it can be beneficial to base the power of the refractive shape (Power/Shape Requirement) at a given diameter based on the central power add of the prescriptive refractive shape, and on the power change requirement of the particular patient. For example, the power of the refractive shape can be a function of a given diameter, as expressed in the following formula.

\[ \text{Power/Shape\_Requirement} = C_0 + A(\text{pupil\_diameter}) \]

where Power/Shape Requirement is the power of the refractive shape at a particular Pupil_Diameter, \( C_0 \) is the central power add of the prescriptive refractive shape, and \( A \) is calculated as

\[ A = \frac{(\text{PRC} - C_0)}{\text{PDP}} \]

where PRC is the power change requirement for the particular patient, and PDP is the pupil diameter parameter (obtained, for example, by subtracting the diameter of the pupil measured when the patient is gazing at infinity from the diameter of the pupil measured when the patient is looking at a near object under identical light conditions). Given the values discussed above, the Power/Shape Requirement (PSR) can be calculated as follows.

\[ \text{PSR} = -3.1 \text{ diopters} + \left[ \left( -2.5 \text{ diopters} - 3.1 \text{ diopters}/0.5 \text{ mm} \right) \right](\text{pupil\_diameter}) \]

or

\[ \text{PSR} = -3.1 \text{ diopters} + 1.2(\text{pupil\_diameter}) \]

Other Pupil Diameter Parameters

It is also possible to calculate a pupil diameter parameter based on a pupil diameter change slope as measured under certain distance and lighting conditions, for example, as the patient gazes at infinity while the lighting conditions change from photopic to scotopic (distance-photopic to scotopic). Pupil diameter parameters can also involve pupil diameter change slopes such as near-photopic to scotopic, photopic-distance to near, mesopic-distance to near, or scotopic-distance to near.

The Effective Power

The effective power (e.g., linear power model or higher order model) can be used to calculate or derive a presbyopic shape, optionally based on the following parameters.

F.1. Emmetropic at distance (photopic and mesopic lighting conditions)

a. This can determine a maximum diameter of the add

F.2. Near can have an effective power of -2.5D (or more, if desired by the patient
F.3. The rate of change of power for the add-treatment combination can have one of the four:

   i. The same power rate of change as the photopic - Distance to near
   ii. The same power rate of change as the mesopic - Distance to near
   iii. The same power rate of change as the scotopic - Distance to near
   iv. Non-linear rate of change similar to the above, but is optimized to give better simultaneous distance and near vision.

[0283] For an eye gazing into infinity, under photopic conditions, the theoretical pupil size at emmetropia can vary within the population. Moreover, the pupil diameter can further vary when the eye is used for different tasks. For example, the pupil diameter can decrease as the eye's gaze changes from infinity to a near object. As the eye changes from a distance gaze to a near gaze, the typical pupil diameter decreases. This change in pupil diameter may be linear with convergence and sigmoid with accommodation. In an eye treated with an exemplary prescriptive shape, the pupil diameter at near gaze can typically have the inner region of the prescriptive shape as the dominant refractive component. Consequently, the change of pupil size from larger to smaller (distance gaze to near gaze) can be equivalent to a change in power. In comparison, the distance gaze pupil will have an effective power based on the combination of the inner region add and the outer region of the prescriptive shape, with the outer region becoming a more dominant refractive component. Therefore, each refractive shape can be customized to each particular individual because of the many different combinations available. By changing the power of the cornea, for example, from emmetropia at the "distance" pupil size to within a range of about -1.0 diopters to about -4.0 diopters myopic for "near" pupil size, it may be possible to mitigate presbyopia.

[0284] A general prescription may go as follows. First, measure the continuous pupil size and/or size change at different distances and lighting conditions, such as for at least one (optionally two or more, in some cases all) of: Distance - Photopic; Distance - Mesopic, Distance - Scotopic, Near - Photopic, Near - Mesopic, and/or Near - Scotopic. The pupil size can be affected by the lighting conditions as well as viewing distances. The refractive shape can also include adjustments and/or optimization for lighting. In photopic conditions, the pupil is typically constricted. In scotopic conditions, the pupil is usually dilated. Under mesopic conditions, the pupil can be variably dilated or constricted depending on the specific type of mesopic condition. Second, calculate the pupil diameter continuous rate of change for the following combinations: Distance - photopic to scotopic, Near - photopic to scotopic, Photopic - Distance to near, Mesopic - Distance to near, and/or Scotopic - Distance to near. It is possible to design a shape and ablation size such
that patient is substantially emmetropic as pupil size goes from larger (distant) to smaller (near), typically within a range.

[0285] The presbyopic lens power can compensate focus such that the lens is the inverse of the rate of pupil change. To do this, the power can change (for example -3D) for different pupil diameters.

\[
\text{Power Change Requirement} = C_0 + A(pupil \_diameter) + B'(pupil \_diameter)^2 + C(pupil \_diameter)^3 + ... 
\]

[0286] The Power/Shape Requirement in the above equation may be effective power, and/or may be manifest power. The power can change with changes in pupil diameter. For a linear power shape, the coefficient A can be calculated as follows.

\[
\frac{d(\text{power})}{d(pupil \_diameter)} = A
\]

[0287] Solving for the linear coefficient,

\[
A = \frac{\text{Power Change Requirement} - C_0}{pupil \_diameter \_rate \_of \_change}
\]

[0288] The target manifest can be targeted to the patient's request or a doctor's recommendation by using the effective distance power equation as described above in the "target manifest" section.

[0289] Multifocal shapes

[0290] A good refractive shape (including a multi-focal shape) may be at or near an optimum compromise between distance and near sight. The near add has an "effective" power - it may not have a single power because of the multi-focal shape. The sum of the peripheral and central add may give the distance power - again it may not have a single power because of the multi-focal shape.

[0291] The Age Dependent Presbyopic Shape

[0292] As discussed above, as one ages, accommodation decreases. This is shown in Fig. 38. At 60, accommodation can decrease significantly, even to nearly zero. Studies have shown that pupil sizes decrease as one gets older. As seen in the figure, the slope or rate of change in accommodation also changes with age. It is possible to optimize the pupil dependencies to the age
related change in accommodation. The rate of distance and near acuities for a central add shape can be

\[
\text{Near\_acuity} = -2.103 + 0.37879 \times \text{Pupil size (Dim)}
\]
\[
\text{Distance acuity} = 0.40001 - 0.0677 \times \text{Pupil size (Dim)}
\]

According to these equations, as the pupil size decreases, the near acuity gets better, at a rate of 0.37 lines per millimeter. The distance acuity gets worse, but at much slower rate of 0.07 lines per millimeter. Therefore, it is possible to optimize the treatment parameters for the patient's age by targeting the treatment for less myopia. It is possible to allow a shift in the centering of the "range" by taking the residual accommodation into account in the customization of the treatment.

It is possible that the optimum shape may be on a "linear" power approximation as discussed above, but it may consist of higher orders. Though the effective power can be given by the equation above, the shape can be constant over, for example, a central 2.5mm and have a curvature gradient that will blend the central add to the peripheral region. With this shape it may be beneficial to choose the diameter of the central add to match the patients near pupil such that the near pupil will encompass only the central add when it's at its smallest, and the gradient will be customized to the patient's pupil size rate of change.

Hence, by modeling the residual accommodation, the range of pupil change may be shifted to optimize the "life" long presbyopic correction.

Systems

The present invention also provides systems for scaling refractive shapes and providing practical customized or optimized refractive shapes that mitigate or treat presbyopia and other vision conditions in particular patients. The systems can be configured in accordance with any of the above described methods and principles.

For example, as shown in Fig. 39, a system 1000 can be used for reprofiling a surface of a cornea of an eye 1600 of a particular patient from a first shape to a second shape having correctively improved optical properties. System 1000 can comprise an input 1100 that accepts a prescriptive shape specific for treating the vision condition, an input 1200 that accepts a pupil dimension of the particular patient, a module 1300 that scales a dimension of a central portion of a refractive shape based on the pupil dimension of the particular patient and an attribute of at least one eye previously treated with the prescriptive shape, a processor 1400 that generates an ablation profile, and a laser system 1500 that directs laser energy onto the cornea according to the ablation
profile so as to reprofile a surface of the cornea from the first shape to the second shape, wherein the second shape corresponds to the refractive shape.

[0299] Calculating of Presbyopia Mitigating Prescriptions

[0300] Methods, Systems, and Devices described herein can be used to generate prescriptions for treatment of refractive errors, particularly for treatment of presbyopia. Such treatments may involve mitigation of presbyopia alone, or may treat a combination of presbyopia with other refractive disorders.

[0301] As described above, presbyopia is a condition where the degree of accommodation decreases with the increase of age. Most people have some degree of presbyopia by the age of about 45.

[0302] Treatments of presbyopia may involve passive and/or active procedures. In passive procedures, treatment or mitigation is performed in such a way that an improved balance between near vision and distance vision is provided and maintained. In an active procedure, restoration of full or partial accommodation is a goal. So far, active procedures for the correction of presbyopia have not been fully successful.

[0303] With passive procedures, it is desirable to provide an improved and/or optimal balance between near vision and distance vision. In order to do that, patients may sacrifice some of their distance vision to gain improved near vision. In addition, they may sacrifice some contrast sensitivity because of the introduction of the asphericity of the new optics of the eye. Fortunately, the sacrifice of distance vision and contrast sensitivity may be mitigated by taking advantage of a pupil shrinkage when the eye accommodates.

[0304] As described below, an analytical solution for a presbyopia shape can be achieved based on a desire for different powers at different pupil sizes. In order to understand this, we can take advantage of a concept of optical power that depends on the change of pupil size and might also depend on wavefront aberrations other than defocus terms. We will concentrate on the pupil size dependency in this description.

[0305] The following approach considers the correction as a "full pupil" correction rather than "partial pupil" correction as employed with a central add. Healing effect, flap effect as well as how the effective power correlates with the manifest refraction may be addressed with empirical studies, allowing these effects to be fed back into the following calculations and/or a laser ablation planning program as appropriate so as to provide optimized real-world results.
Effective Power and Its Application to Presbyopia

used herein, "effective power" means the optical power that best matches the manifest sphere at a certain pupil size. With wavefront based ocular aberrations, the defocus-dependent effective power can be written as

\[ P_{ef} = -\frac{4\sqrt{3}c^0_2}{R^2}, \]  

(1)

where \( R \) stands for the pupil radius in mm when \( c^0_2 \) is the Zernike coefficient given in microns in order to get the effective power in diopters, and \( P_{ef} \) is effective power. When a wavefront map is defined in radius \( R \) with a set of Zernike polynomials, when the pupil shrinks the smaller map, if re-defined with a new set of Zernike polynomials, will have a different set of Zernike coefficients than the original set. Fortunately, analytical as well as algorithmatical solutions of the new set of Zernike coefficients exist. If the original set of Zernike coefficients is represented by \( \{c_i\} \) that corresponds to pupil radius \( r_2 \), then the new set of Zernike coefficients \( \{b_i\} \) that corresponds to pupil radius \( r_2 \) can be expressed by a recursive formula as

\[ b^0_{2i} = e^{2i} \sum_{j=0}^{\frac{n-1}{2}} (-1)^j c^0_{2(i+j)} \frac{2(i+j)+1}{2i+1} \frac{(2i+j)!}{j!(2i)!} - \sum_{k=2(i+1)}^{n} b^0_{k} \sqrt{\frac{k+1}{2i+1}} \frac{(-1)^{k/2-1}(k/2+i)!}{(k/2-i)!(2i)!} \]

where \( e = r_2/r_1 \), \( n \) is the maximum radial order. As an example, if we set \( i = 1 \), and \( n = 4 \), we have the following formula

\[ b^0_2 = |e^2 - 15(1 - e^2)c^0_2|^2 e^{-2} \]

Therefore, a power profile with pupil size can be given as a condition to obtain an optical surface for presbyopia correction.

In order to obtain a presbyopia prescription (which will here be an optical shape), let's assume that we know the power profile or desired effective optical powers for different viewing conditions so as to mitigate presbyopia. From the power profile, we can in general do an integration to calculate the wavefront shape. In the following, we consider three cases where two, three, or four power points (different desired effective optical powers for different associated viewing conditions, often being different viewing distances and/or pupil diameters) are known.

Two-Power-Point Solution
Let's consider radially symmetric terms $Z^0_2$ and $Z^4_4$, when the pupil radius is changed from $R$ to $eR$, where $e$ is a scaling factor not larger than 1, since the new set of Zernike coefficients for the defocus term can be related to its original coefficients as

$$b^0_2 = [c^0_2 - \sqrt{15}(1-e^2)c^0_4]e^2. \quad (2)$$

Substituting $c^0_2$ with $b^0_2$, and $R^2$ with $e^2R^2$ in Equation 1 using Equation 2, we have

$$4\sqrt{3}c^0_2 - 12\sqrt{5}(1-e^2)c^0_4 = -R^2P. \quad (3)$$

Suppose we request power $p_0$ at radius $e_0 R$ and $p_1$ at radius $e_1 R$, an analytical solution of the original wavefront shape, which is represented by $c^0_2$ and $c^0_4$, can be obtained as

$$c^0_2 = \frac{(1-e^2)p_0 - (1-e^2)p_1}{4\sqrt{3}(e^2 - e_1^2)} R^2$$
$$c^0_4 = \frac{-p_0 - p_1}{12\sqrt{5}(e^2 - e_1^2)} R^2. \quad (4)$$

As an example, let's consider a pupil with a dim distance size of 6mm, requesting effective power of 0D at pupil size 6mm and bright reading pupil size of 4.5mm, requesting effective power of -1.5D. Substituting $e_0 = 6/6 = 1$, $\beta = 4.5/6 = 0.75$, and $p_0 = 0$ and $p_1 = -1.5$, we get $c^0_2 = 0$ and $c^0_4 = -1.15$. Figs. 40 and 41 show the presbyopia shape and effective power as a function of pupil size. It is very close to a linear relationship.

Three-Power-Point Solution

Let's consider radially symmetric terms $Z^0_2$, $Z^0_4$ and $Z^0_6$, when the pupil radius is changed from $R$ to $eR$, where $e$ is a scaling factor not larger than 1, since the new set of Zernike coefficients for the defocus term can be related to its original coefficients as

$$b^0_2 = [c^0_2 - \sqrt{5}(1-e^2)c^0_4 + 421(2 - 5e^2 + 3e^4)c^0_6]e^2. \quad (5)$$

Substituting $c^0_2$ with $b^0_2$, and $R^2$ with $e^2R^2$ in Equation 1 using Equation 5, we have

$$4\sqrt{3}c^0_2 - 12\sqrt{5}(1-e^2)c^0_4 + 12\sqrt{7}(2 - 5e^2 + 3e^4)c^0_6 = -R^2P. \quad (6)$$
Suppose we request power $p_0$ at radius $eR$, $P_i$ at radius $gR$, and $p_2$ and radius $gR$, an analytical solution of the original wavefront shape, which is represented by $c_2^0$, $c_4^0$ and $c_6^0$, can be obtained as

\[
c_2^0 = \frac{(1-e_2^2)(1-e_2^2)(e_2^2-e_2^2)p_e}{4\sqrt{3}(e_1^2-e_1^2)(e_1^2-e_1^2)(e_1^2-e_1^2)} + \frac{(1-e_2^2)(1-e_2^2)(e_2^2-e_2^2)p_e}{R^2}.
\]

\[
c_4^0 = \frac{(5-3e_2^2-3e_2^2)(e_2^2-e_2^2)p_e}{36\sqrt{5}(e_1^2-e_1^2)(e_1^2-e_1^2)} - \frac{(5-3e_2^2-3e_2^2)(e_2^2-e_2^2)p_e}{R^2}.
\]

\[
c_6^0 = \frac{(e_2^2-e_2^2)p_e}{36\sqrt{7}(e_1^2-e_1^2)(e_1^2-e_1^2)}.
\]

(7)

As an example, let's consider a pupil with WaveScan pupil size of 6mm, and dim distance pupil size of 6mm, requesting effective power of 0D and bright reading pupil of 3.5mm, requesting effective power of -1.5D. In between are the dim reading and bright distance, with combined pupil size of 4.5mm with effective power of -0.5D. Substituting $e_0 = 6/6 = 1$, $e_1 = 4.5/6 = 0.75$, and $e_2 = 3.5/6 = 0.583$ as well as $p_0 = 0$, $p_1 = -0.6$ and $p_2 = -1.5$, we get $c_2^0 = 0$, $c_4^0 = -0.31814$ and $c_6^0 = 0.38365$. Figs. 42 and 43 shows the presbyopia shape and the effective power as a function of pupil sizes.

**Four-Power-Point Solution**

Let's consider radially symmetric terms $Z_{2}^0$, $Z_{4}^0$, $Z_{6}^0$, and $Z_{8}^0$, when the pupil radius is changed from $R$ to $eR$, where $e$ is a scaling factor not larger than 1, since the new set of Zernike coefficients for the defocus term can be related to its original coefficients as

\[
b_2^0 = [c_2^0 - VL5(1-i^2)c_4^0 + \sqrt{2}1(2-5e^2+3e^4)c_6^0 - V3(10-45e^2-2Se^6)c^0]e^2
\]

(8)

Substituting $c_2^0$ with $b_2^0$ and $R^2$ with $e^2R^2$ in Equation 1 using Equation 8, we have

\[
4V3c_2^0 - 12V5(1-e^2)c_4^0 + 12V7(2-5e^2+3e^4)c_6^0 - 12(10-45e^2+63e^4-2Se^6)c^0 = -R^2P
\]

(9)

Suppose we request power $p_0$ at radius $eO$, $P_i$ at radius $eR$, $p_2$ and radius $eR$, and $p_3$ and radius $eR$, an analytical solution of the original wavefront shape, which is represented by $c_2^0$, $c_4^0$, $c_6^0$, and $c_8^0$, can be obtained as:
\[ \lambda = \frac{e_0^2 - e_1^2}{e_2^2 - e_3^2} \] (11)
\[ \alpha_0 = (e_0^2 - e_1^2)(e_2^2 - e_3^2) \] (12)
\[ \beta_0 = (e_0^2 - e_1^2)(e_2^2 - e_3^2) \] (13)
\[ \gamma_0 = [9 - 4(e_0^2 + e_1^2 + e_3^2)] \gamma_0 \] (17)
\[ \delta_0 = [9 - 4(e_0^2 + e_1^2 + e_3^2)] \delta_0 \] (19)
\[ \alpha_1 = [45 - 35(e_1^2 + e_2^2 + e_3^2) + 21(e_1^2 e_2^2 + e_1^2 e_3^2 + e_2^2 e_3^2) \beta_0] \alpha_0 \] (20)
\[ \beta_1 = [45 - 35(e_0^2 + e_1^2 + e_2^2 + e_3^2) + 21(e_0^2 e_1^2 + + e_2^2 e_3^2) + e_2^2 e_3^2)] \beta_0 \] (21)
\[ \gamma_1 = [45 - 35(e_0^2 + e_1^2 + e_2^2 + e_3^2) + 21(e_0^2 e_1^2 + e_2^2 e_3^2 + e_3^2 e_1^2) + e_3^2 e_1^2)] \gamma_0 \] (22)
\[ \delta_1 = [45 - 35(e_0^2 + e_1^2 + e_2^2 + e_3^2) + 21(e_0^2 e_1^2 + e_2^2 e_3^2 + e_3^2 e_1^2) + e_3^2 e_1^2)] \delta_0 \] (23)

As an example, let’s consider a pupil with WaveScan pupil size of 6mm, and dim distance pupil size of 6mm, requesting effective power of 0D and bright reading pupil size of 3.5mm, requesting effective power of -1.5D. We also request that the bright distance pupil size to be 5mm and dim reading pupil size of 4.5mm, with effective power of -0.2D and -0.5D, respectively. Substituting \( e_0 = 6/6 = 1 \), \( e_1 = 5/6 = 0.833 \), \( e_2 = 4.5/6 = 0.75 \) and 
\( e_3 = 3.5/6 = 0.583 \) as well as \( p_0 = 0 \), \( p_1 = -0.2 \), \( p_2 = -0.5 \) and \( p_3 = -1.5 \), we get \( c_3^0 = -0.2919 \), \( c_5^0 = 0.3523 \) and \( c_7^0 = -0.105 \). Figs. 44 and 45 show the presbyopia shape and the effective power as a function of pupil sizes. Note that both the presbyopia shape and the effective
power are similar to those shown in Figs. 42 and 43. However, the shape and power given with 4-term solution is smoother and have a flatter power at larger pupil sizes.

[0326] It is also possible to use the same approach to obtain analytical solutions for conditions that use more than four power points. For example, when we use five power points, we could use up to 10th order of Zernike coefficients to describe the aspheric shape that satisfies the power profile defined with five power points. Similarly, six power points can define an aspheric shape using 12th order of Zernike coefficients. Because more power points can in general make the analytical solution more difficult, another way of approaching the solution is by more complex numerical algorithms. Due to the availability of the recursive formula, the equations that lead to analytical solutions might be converted to an eigen system problem, which does have numerical solutions, optionally making use of the methods of William H Press, Saul A. Teukolsky, William Vetterling, and Brian P. Flannery, in *Numerical Recipes in C++*, (Cambridge University Press, 2002). Such a solution may be more accurate than use of discrete power point.

[0327] Discussion

[0328] It is helpful to address how many terms to use in determining the presbyopia shape. In the two-power-term solution, we use the pupil sizes as well as the corresponding desired powers. We can use this solution for a somewhat “bifocal” design with one distance pupil size and power (which should be zero to keep the eye at emmetropia) and one reading pupil size and its corresponding power. From Figs. 40 and 41, the effective power follows a rather linear relationship with pupil size changes. This may not be ideal in that the distance power may tend to become myopic. With a 3-power-term solution, we have one more freedom to choose the power in a middle pupil size and in fact the solution is rather close to a 4-power-term solution when carefully designed. Unfortunately, with a 3-power-term solution, the bright distance pupil and the dim reading pupil tend to be averaged and so do the corresponding powers. This may become too inflexible to design an ideal shape. Therefore, the 4-power-term solution, which tends to give a more favorable reverse Z-curve, should be used in the practical implementation. The reverse Z-curve such as that shown in Fig. 46A, a positive power gradient region between two lower slope (or flat) regions within a pupil size variation range for a particular eye, may be a beneficial effective power characteristic for presbyopia mitigation.

[0329] Even with a 4-power-term solution, choosing effective powers in-between dim distance pupil and bright reading pupil should be carefully considered. For instance, in order to satisfy restaurant menu reading, we might want to increase the power for dim reading. In this case, an unfavorable S-curve would exist, as is also shown in Fig. 46A. Presbyopia-mitigation shapes
corresponding to the S-curve and Z-curve shapes are shown in Fig. 46B. These results were generated for a 6mm pupil with the dim distance pupil at 6mm with a power of 0D, the bright distance pupil at 5mm with power of -0.2D and -0.7D, the dim reading pupil at 4.5mm with a power of -1.2D and the bright reading pupil at 3.5mm with a power of -1.5D. To reduce the fluctuation of effective power, we can also increase the power in bright distance and in this case the distance vision can be affected (in addition to the contrast drop due to asphericity).

[0330] Another parameter we can set is desired reading power. Optionally we can give the patient full power; say 2.5D, so the treatment can be sufficient to treat presbyopia for the patient's life span. However, the natural pupil size decreases with increasing age. Therefore, a shape well suited to a patient at the age of 45 could become deleterious at the age of 60. Secondly, not everyone easily tolerates asphericity. Furthermore, too much asphericity can reduce the contrast sensitivity to a level that distance vision would deteriorate. As such, measurement of a patient’s residual accommodation becomes beneficial in the success of presbyopia correction. In addition, the various pupil sizes at different lighting conditions and accommodation can be measured systematically and more accurately. Such measurements may employ, for example, a commercially available pupilometer sold by PROCYON INSTRUMENTS LIMITED of London, United Kingdom, under the model number Procyon P-2000 SA. A wide variety of alternative pupil measurement techniques might be used, including visual measurements, optionally using a microscope displaying a scale and/or reticule of known size superimposed on the eye, similar to those employed on laser eye surgery systems commercially available from AMO Manufacturing USA, LLC in Milpitas, California.

[0331] The influence of high order aberrations on the effective power, as described above regarding the power map, may also be incorporated into the presbyopia-mitigating shape calculations. This may involve integration over the entire power map, i.e., the average power, with appropriate adjustment so as to avoid overestimating power (that may otherwise not agree with the minimum root-mean-square (RMS) criterion) and so as to correlate with patient data. The influence of high order spherical aberrations on effective power calculation should not be entirely ignored. In particular, the influence on the depth of focus, and hence to the blur range during manifest refraction test, can be determined using clinical testing.

[0332] Taking advantage of the ability to calculate presbyopia shapes based on effective power, presbyopia-mitigating shapes can be derived and/or optimized based on the following considerations. First, image quality of the presbyopia shape at different viewing conditions can be evaluated. To do so, optimization of the shape itself can be pursued. This can be done in several
ways, such as using diffraction optics (wave optics) or geometrical optics (ray tracing). Because we are dealing with aberrations of many waves, it may be impractical to use point spread function based optical metrics. However, since the aberration we introduce belongs to high orders only, wave optics may still work well. In fact, a comparison of Zemax modeling with three wavelengths and using verification tools (wave optics), as shown in Fig. 16, with 7-wavelengths show almost identical results in both point spread function (PSF) and modulation transfer function (MTF). Fig. 47 shows some derived shapes for a 5mm and a 6mm pupil, while the corresponding MTF curves are shown in Fig. 48. The simulated blurring of eye chart letter E for both cases is shown in Fig. 49. These letters graphically illustrate verification of presbyopia shape using a goal function with 7-wavelengths polychromatic PSF and a 20/20 target. The first image shows a target at 10m. The second to the last image shows targets from 1m to 40cm, separated by 0.1D in vergence. One diopter of residual accommodation is assumed for each. Even without optimization, the optical surface shown gives almost 20/20 visual acuity over 1.5D vergence.

The above approach is valid to apply in contact lens, intra-ocular lens, as well as spectacles, as well as refractive surgery. Such calculations for refractive surgery may be adjusted for the healing effect as well as the LASIK flap effect based on empirical studies and clinical experience.

As established above, it is possible to obtain analytical expressions for the Zernike coefficients of the first few spherical aberrations of different orders to create an aspheric shape for presbyopia correction based on one or more desired effective powers. Healing effect, flap effect, and the correlation of effective power with manifest refraction will benefit from additional patient data and empirical studies to further refine the presbyopia shape so as to (for example) more accurately plan the shape for future ablation.

Figs. 50A and 50B illustrate exemplary desired power curves and treatment shapes for mitigating presbyopia of a particular patient. The four power point solution was used to establish these shapes. For a 6mm pupil, the following Table 5 describes the four conditions or set points from which the shape was generated:
Table 5

<table>
<thead>
<tr>
<th>Conditions</th>
<th>6mm Pupil</th>
<th>5mm Pupil</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Effective power</td>
<td>Pupil size (mm)</td>
</tr>
<tr>
<td>1</td>
<td>0D</td>
<td>6</td>
</tr>
<tr>
<td>2</td>
<td>-0.5D</td>
<td>5</td>
</tr>
<tr>
<td>3</td>
<td>-1D</td>
<td>4.5</td>
</tr>
<tr>
<td>4</td>
<td>-1.5D</td>
<td>3.4</td>
</tr>
</tbody>
</table>

[0336] Fig. 50A shows the effective power profiles, while Fig. 50B shows the corresponding presbyopia shapes. To model the healing and LASIK flap effect, we uniformly boost the shape by 15%. In addition to the added presbyopia shape, we also used -0.6D physician adjustment in the wavefront prescription generation to offset myopic bias to aim emmetropia at normal viewing condition (bright distance) after surgery.

CMTF Based On PSF

[0337] As discussed elsewhere herein, the compound modulation transfer function (CMTF) provides a useful optical metric for evaluating any optical design, and in general provides an effective tool for simulation, evaluation, optimization, and design of optical surface shapes for laser refractive surgery and other vision treatment modalities. In exemplary cases, CMTF can be used to evaluate shapes for treating presbyopia of the eye, in the context of laser refractive surgeries, contact lens prescriptions, intraocular lenses, and the like.

[0338] A CMTF value can be calculated in a variety of ways. For example, a CMTF value can be calculated based on a monochromatic point spread function (PSF). It is also possible to calculate a CMTF value based on a polychromatic point spread function (PSF).

[0339] Techniques for obtaining a polychromatic PSF are discussed elsewhere herein, as well as in US 2010/0103376, which is incorporated herein by reference. In some cases, Fourier transform of the phase screen may be used for determining a point spread function, according to diffractive theory. In other cases, ray tracing may be used to determine a point spread function. As noted elsewhere here, a point spread function can be used for diagnostic purposes, treatment purposes, analyzing optical acuity, and optimizing optical treatment shapes, such as shapes for treating presbyopia in a patient.

[0340] Given a particular polychromatic PSF, it is possible to perform a Fourier transform of the PSF to obtain a modulation transfer function, which may involve determining the modulus. Multiple modulation transfer functions obtained based on different spatial frequencies can be used
to determine a CMTF. Exemplary techniques for performing a Fourier transform of a PSF to obtain an MTF are described in J. W. Goodman, Introduction to Fourier Optics, 3rd ed (Roberts & Company, 2005), the content of which is incorporated herein by reference. Similarly, in some cases an MTF can be determined from a PSF in a manner which does not involve a Fourier transform of the PSF. For example, it is possible to calculate the integration of the overlap area of a pupil function (e.g. overlap area of two pupil apertures), to obtain an optical transfer function (OTF). The MTF can be based on a modulus, or magnitude, of the OTF. Exemplary integration techniques are describe in J. W. Goodman, Introduction to Fourier Optics, 3rd ed (Roberts & Company, 2005), previously incorporated. Where there are aberrations, it may be desirable to obtain the MTF based on the PSF Fourier transform approach, instead of the integration approach. According to these techniques, it is possible to obtain an MTF or CMTF based on a PSF or PPSF.

[0341] Because modulation transfer functions can be calculated based on point spread functions, it is desirable to obtain or provide point spread functions which are accurate and appropriate for the intended use. As described elsewhere herein, according to some embodiments polychromatic point spread functions are particularly well suited for use for applications involving the optics of the human eye.

**CMTF Threshold**

[0342] Embodiments of the present invention encompass system and methods for evaluating an image quality provided by a vision treatment shape. For example, techniques may include obtaining a plurality of through-focus compound modulation transfer function (CMTF) values for the vision treatment shape, comparing these CMTF values to a CMTF threshold value, and evaluating the image quality based on the comparison.

[0343] According to some aspects of the present invention, it is useful to determine a CMTF threshold value for use in evaluating the image quality. A CMTF threshold may represent a minimal CMTF value, below which the image quality may not be considered to be acceptable or desired. The CMTF threshold can be used to evaluate whether a particular optical surface or shape may be effective for treating a particular vision condition. In some instances, a CMTF threshold can have a value of about 0.1. In some instances, a CMTF threshold can have a value of about 0.3.

[0344] **FIG. 51** illustrates through-focus results for a 20/20 eye chart letter E convolved with certain point spread function models (e.g. monochromatic, polychromatic, and no aberration polychromatic) across a vergence range (e.g. -1.0 D to 3.0 D) for a 5.0 mm pupil size. Such through-focus results can be obtained by moving a vision target throughout a range of vergence,
for example, from a distant location (e.g. 0 D, which corresponds to infinity) to a near location
(e.g. 3.0 D).

[0345] For FIG. 51, through-focus (in terms of vergence) wavefront maps are used to generate
the point spread function, using both the monochromatic and polychromatic PSF models. A no
aberration situation, which uses a polychromatic PSF model, can be compared to the
monochromatic and polychromatic PSF results. As seen here, the polychromatic PSF provides
improved image results over the monochromatic PSF at 0.5D (distant), and between 1.5D and 3.0
D (near).

[0346] As shown here, visual inspection of the results at 1.0 D vergence indicates that the
convolved eye chart letter image is difficult to discern. Relatedly, visual inspection of the results
at 0.5 D vergence indicates that the eye chart letter image associated with the PPSF model is
discernible (e.g. visibly apparent borders of letter E indicate acceptable acuity or resolution, albeit
with low contrast to surrounding background), whereas the eye chart letter image associated with
the MPSF model is not so discernible. Surgeons, diagnostic device operators, and other personnel
can visually inspect such results to determine discernibility, or to compare discernibility between
different images, and to make selections or determination based on the visual inspection.

[0347] Whereas FIG. 51 provides convolution images for the three different treatment shapes,
FIG. 52 provides corresponding CMTF value curves for the three same shapes. Specifically, FIG.
52 shows the results of CMTF through-focus CMTF curves for one presbyopic correction shape
using monochromatic and polychromatic PSF models, as compared with a diffraction-limited (no
aberration) case. The CMTF values here were calculated using spatial frequencies of 10, 15, 20,
and 30 cpd (cycles per degree). For each of (i) the diffraction limited case, (ii) the presbyopic
correction shape using monochromatic PSF, and (iii) the presbyopic correction shape using
polychromatic PSF, the peak CMTF is at or near the 0 Diopter vergence. The presbyopia shapes
were optimized with the noted set of cpd spatial frequencies.

[0348] For the purpose of presbyopia corrective treatments, for example, the through-focus
CMTF curves of FIG. 52 can be directly compared with the convolved eye chart letters shown in
FIG. 51, in order to compare the various correction shapes at selected vergences. Based on the
comparison, it is evident that CMTF values of 0.1 or more correspond with visually discernible
letters. For example, at 0.5D as shown in FIG. 51, the monochromatic PSF convolved eye chart
letter is not discernible, whereas the polychromatic PSF convolved eye chart letter is discernible.
Relatedly, at 0.5D as shown in FIG. 52, the monochromatic PSF CMTF is below 0.1, whereas the
polychromatic PSF CMTF is about 0.1. With reference to FIGS. 51 and 52, exemplary
embodiments may include calculating a through-focus curve, or through-vergence curve, with a particular optical metric, such as the compound modulation transfer function. Such techniques are well suited for use in identifying, generating, or evaluating treatment shapes.

[0349] Through-focus CMTF curves such as these are useful to evaluating whether a particular treatment may be helpful for a particular vision condition. For example, a treatment shape may be evaluated with a through-focus CMTF curve to determine whether it might provide a suitable bifocal correction. If the through-focus CMTF curve indicates that the treatment shape provides CMTF values at or above a threshold (e.g. 0.1) for both distance vision (e.g. 0 D) and near vision (e.g. 3.0 D), it may be concluded that the treatment shape is a good candidate for the bifocal correction. In this case, the treatment shape may generate two peaks on the through-focus CMTF curve, one peak for distance vision and one peak for near vision. In some cases, a treatment shape may provide a through-focus CMTF curve that meets or exceeds the threshold across an entire vergence range (e.g. 0 D to 3.0 D), in which case the treatment shape may be considered to provide a very desirable outcome.

[0350] Based on an evaluation of FIGS. 51 and 52, it can be concluded that for the presbyopic correction shapes (e.g. at noted cpd spatial frequency sets), image quality for the convolved E is quite acceptable where the CMTF curve value is greater than about 0.3. Where the CMTF curve value is lower than about 0.1, however, the image quality is not as acceptable.

[0351] When calculating a goal function or merit function, it is possible to do so with a through-focus approach. In some cases, it is desirable to obtain an optimized goal function, which may be calculated through a set of target distances, or a set of vergence values. For example, it may be possible to separate a 3.0 D vergence into 30 bins, and CMTF values can be obtained for each of the 30 incremental 0.1 D vergence locations, and the merit or goal function can be determined accordingly.

[0352] Table 6 shows the range of vergence below the CMTF threshold, for each of the shapes depicted in FIG. 52. As depicted here, the presbyopic correction shape using polychromatic PSF provides the least vergence range (about 0.8D) below threshold.

<table>
<thead>
<tr>
<th>CMTF threshold</th>
<th>Vergence Range below CMTF Threshold</th>
<th>Diffraction Limited</th>
<th>Monochromatic PSF</th>
<th>Polychromatic PSF</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.1</td>
<td>0.3 D to 3 D (total: 2.7 D)</td>
<td>0.2 D to 1.2 D (total: 1.0 D)</td>
<td>0.5 D to 1.3 D (total: 0.8 D)</td>
<td></td>
</tr>
</tbody>
</table>

Table 6
<table>
<thead>
<tr>
<th>OD</th>
<th>0 D to 0.1 D (total: 0.1 D)</th>
<th>0 D to 0.2 D and 1.3 D to 1.7 D (total: 0.6 D)</th>
<th>0 D to 0.3 D and 1.5 D to 3.0 D (total: 1.8 D)</th>
</tr>
</thead>
</table>

[0353] It is apparent from FIG. 52 that the CMTF for the presbyopic correction shape using monochromatic PSF is greater than the CMTF for the presbyopic correction shape using polychromatic PSF, at around the 0 D vergence value. It is also apparent that the CMTF for the presbyopic correction shape using monochromatic PSF and the presbyopic correction shape using polychromatic PSF diverge significantly at around 1.5 D vergence. For vergence values higher than 1.5 D, the presbyopic correction shape using polychromatic PSF provides markedly higher CMTF values than does the presbyopic correction shape using monochromatic PSF.

[0354] According to embodiments depicted here, CMTF values below 0.1 D correspond to little or no image discemibility, whereas CMTF values above 0.3 correspond to good image discemibility. Beneficial optical correction surfaces provide good image discemibility (e.g. CMTF above threshold) across a large vergence, or across the through-focus range. As illustrated in Table 6, with respect to the 0.1 CMTF threshold, the PPSF provides the narrowest vergence range (0.8 D) below threshold. And with respect to the 0.3 CMTF threshold, the PPSF provides the broadest vergence range (1.8 D) above threshold. Hence, PPSF can be considered to correspond to a more beneficial result, because it provides the least vergence range below 0.1 CMTF threshold (e.g. little image discemibility) and the greatest vergence range above 0.3 CMTF threshold (e.g. good image discemibility). A CMTF through-vergence curve with values all above threshold is considered to provide a beneficial vision result.

[0355] In general, the 0 D vergence value corresponds to distance vision (e.g. infinity), and the 1.5 D to 3.0 D vergence values (or in some cases, 2.0 D to 2.5 D) correspond generally to near vision. For example, 3 diopters corresponds to a viewing distance of about 33 cm, which is considered a true reading distance, 2 diopters corresponds to a viewing distance of 0.5 m, and 1 diopter corresponds to a viewing distance of 1.0 m.

[0356] It can be seen from FIG. 52 that the polychromatic point spread function provides improved CMTF for near vision across a larger vergence range, while providing reduced CMTF for distance vision across a smaller vergence range. Hence, a polychromatic point spread function can provide benefits over a monochromatic point spread function in terms of accuracy, particularly for near vision. The polychromatic point spread function can also provide benefits over a monochromatic point spread function in terms of a wider depth of field, or greater through-focus range, for higher CMTF values. For at least these reasons, it has been discovered that CMTF
values based on such polychromatic point spread function provide enhanced features over CMTF
values based on a monochromatic point spread function.

[0357] In some instances, through-focus CMTF values can be compared with a CMTF threshold
value of 0.1 for purposes of evaluating image quality. In some instances, through-focus CMTF
values can be compared with a CMTF threshold value of 0.3 for purposes of evaluating image
quality. According to the embodiments illustrated in FIGS. 51 and 52, a CMTF value of 0.1 may
be considered a minimum threshold, below which it is difficult or impossible to discern the
images, and above which may provide satisfactory results for most patients. Relatedly, a CMTF
value of 0.3 may be considered a sufficient threshold, above which provides very good results.

Once a threshold is established, evaluations can be made based on the CMTF values alone (e.g. as
shown in FIG. 52), without relying upon visual inspection of the convolution images (e.g. as
shown in FIG. 51).

[0358] In addition to the spatial frequency combinations shown in FIGS. 51 and 52,
embodiments of the present invention encompass the use of other spatial frequency combinations
to generate other CMTF curves. Such through-focus CMTF curves can be compared with their
corresponding convolved eye chart letters to determine at what level a CMTF value for a particular
CMTF spatial frequency set can be considered a good threshold for 20/20 letter discernibility.

[0359] Hence, embodiments of the present invention encompass system and method for
obtaining acceptable or threshold CMTF values for evaluating visual acuity based on an input set
of spatial frequencies for an optical system. In one particular embodiment, the optical system is a
human eye, and a threshold CMTF value of 0.1 for discerning a 20/20 letter E is obtained based on
input spatial frequencies of 10, 15, 20, 30 cpd. Correlations between the cpd frequency values and
the threshold CMTF can be based on visual inspection of the convolved eye chart.

[0360] Such threshold CMTF values can be used for determining whether a particular shape
design provides an acceptable vision treatment result for a patient. Threshold CMTF values can be
used for any general shape design, including shapes designed specifically for a presbyopia
treatment. According to some embodiments, a threshold CMTF value may be determined by a
surgeon, optionally with feedback provided by a patient, or with computer models.

[0361] According to some embodiments, it is possible to modify or generate optimization
algorithms, once a threshold CMTF value is determined. In certain cases, including some laser or
intraocular lens treatment modalities, multifocal corrections may involve a compromise between
distance and near vision. For example, some corrections may provide enhanced distance vision characteristics, while other corrections may provide enhanced near vision characteristics.

[0362] As noted elsewhere herein, in some cases a treatment shape may provide a through-focus CMTF curve that meets or exceeds the threshold CMTF across an entire vergence range (e.g. 0 D to 3.0 D). Such treatment shapes may provide a true multifocal or omnifocal correction for a patient, where the patient has satisfactory vision throughout the vergence range, at all target distances. In such cases, the threshold may be considered to be the minimal acceptable range. In some cases, a treatment shape may provide a through-focus CMTF curve that meets or exceeds a threshold CMTF of 0.3 across an entire vergence range. In some cases, the threshold CMTF can have a value within a range from about 0.1 to about 0.3.

Depth Of Field

[0363] In general, treatment shapes based on a polychromatic point spread function (PPSF) are better and more accurate than treatment shapes based on a monochromatic point spread function (MPSF). For example, the through-focus (or depth of field) for a treatment shape based on PPSF is wider than the through-focus for a treatment shape based on MPSF. For example, as depicted in Table 6, the vergence range above a 0.3 CMTF threshold is 0.6 D for MPSF and 1.8 D for PPSF (e.g. 1.8 D > 0.6 D).

Additional Features Of Monochromatic and Polychromatic Point Spread Functions

[0364] Both monochromatic and polychromatic point spread functions can be used for simulating and evaluating the effects of optical systems. A MPSF may not accurately reflect certain features associated with the human eye, such as the effect of chromatic aberrations, the Stiles-Crawford effect, and the retinal response function effect, for example. A PPSF, however, may be better suited for capturing such features.

[0365] FIGS. 53A and 53B illustrate a point spread function (PSF) with the use of monochromatic and polychromatic models, respectively. These figures show PSF images for a -0.12 D myopic eye with 6 mm pupil calculated using the monochromatic and polychromatic models. With such a small amount of myopia, this particular eye is considered to have a visual acuity that exceeds 20/20 and a crispy, compact PSF. These images correspond to a point-source light. For example, the image profile on the focal plane for a point source can define the point spread function. However, the monochromatic PSF of FIG. 53A shows a ring type configuration. In contrast, the polychromatic PSF of FIG. 53B shows a centrally concentrated configuration.
Hence, a polychromatic PSF model may be considered well suited for use with the optics of the human eye. FIG. 54 depicts the cross sections of the PSF images of FIGS. 53A and 53B.

[0366] As shown in FIG. 54, the normalized intensity for the PPSF has a peak value near the center, or zero arc minute field of view. In contrast, the normalized intensity for the MPSF is more diffused across the field of view, and presents a valley value near the center, or zero arc minute field of view. The concentrated peak result associated with PPSF, as compared with the ring result associated with MPSF, is evidence that PPSF provides improved results over MPSF.

[0367] FIGS. 55A and 55B show the cross-sections for point spread functions with increasing defocus, using monochromatic and polychromatic models, respectively. As depicted here, the range of defocus extends from a diffraction-limited situation (0 defocus) to a 0.200 D (maximum focusing error), with 0.025D increments disposed therebetween. The PPSF of FIG. 55B provides smooth and regular transitions of the observed normalized intensity curve, when stepping from one level of defocus to an adjacent level of defocus. In contrast, the MPSF of FIG. 55A provides erratic transitions of the observed normalized intensity curve, when stepping from one level of defocus to an adjacent level of defocus.

[0368] As additional focusing error is introduced, the polychromatic point spread function of FIG. 55B is observed to become less and less crispy (e.g. shallower and broader peak of normalized intensity). The monochromatic point spread function of FIG. 55A, however, is observed to become wider and wider very quickly initially as additional focusing error is introduced, subsequently changing from a peak to a ring, and thereafter reverting back from a ring to a peak. This effect can also be seen in FIG. 56, where the curves are re-normalized.

[0369] FIG. 56 depicts cross sections of point spread functions calculated using a monochromatic model with re-normalization. This result is not consistent with normal human eye optics, which do not present a monochromatic situation. Instead, the human eye provides features such as chromatic aberrations, the retina response, and the Stiles-Crawford effect. Relatedly, white light has a broad spectral range, encompassing light rays of many different wavelengths (e.g. red, yellow, blue, green, indigo, and violet). These features affect the realization of the point spread function. The polychromatic point spread function, as shown in FIGS. 54 and 55B, provide a more centrally defined zone of normalized intensity, which is more realistic optically. Hence, it can be seen that for visual simulation or evaluation, use of a polychromatic PSF may be more appropriate than use of a monochromatic PSF. Exemplary techniques for implementing the point spread function are discussed elsewhere herein, as well as in previously incorporated

[0370] According to embodiments of the present invention, polychromatic aberrations can be used to calculate point spread functions. What is more, polychromatic point spread functions can be used to convolve resolution targets.

[0371] The use of polychromatic point spread functions can provide an improvement in accuracy as compared with the use of monochromatic point spread functions. This can be the case with the use of polychromatic aberrations, as compared with monochromatic aberrations, for the calculation of the point spread function. This can also be the case with the use of monochromatic point spread functions, as compared with polychromatic point spread functions, for convolving images.

[0372] As described elsewhere herein, it is possible to use monochromatic point spread functions as well as polychromatic point spread functions. Once a point spread function has been determined, it is possible to evaluate acuity.

[0373] **FIG. 57** illustrates a method 5700 of evaluating an image quality provided by a vision treatment shape according to embodiments of the present invention. As shown here, method 5700 includes obtaining a plurality of through-focus compound modulation transfer function (CMTF) values for the vision treatment shape, as depicted by step 5710, comparing the plurality of through-focus CMTF values to a CMTF threshold value, as depicted by step 5720, and evaluating the image quality based on the comparison between the through focus CMTF values and the CMTF threshold value, as depicted by step 5730.

[0374] **FIG. 58** illustrates a method 5800 of determining a compound modulation transfer function (CMTF) threshold value for a CMTF spatial frequency set. As shown here, method 5800 includes obtaining a plurality of through-focus CMTF values for a vision treatment shape, where the CMTF values are based on the CMTF spatial frequency set, as depicted by step 5810, obtaining a plurality of through-vergence convolved images based on the vision treatment shape and a point spread function, as depicted by step 5820, and determining the CMTF threshold value for the CMTF spatial frequency set based on the plurality of through-focus CMTF values and the plurality of convolved images, as depicted by step 5830.
Goal Functions Having Multiple Metrics or Parameters

[0375] Treatment modalities for presbyopia and other vision conditions can be based on goal functions having multiple metrics or parameters. For example, composite optical metrics may include various combinations of metrics selected from a Strehl ratio, a modulation transfer function (MTF), an encircled energy, a compound modulation transfer function (CMTF), a point spread function (PSF), a volume under MTF surface (VMTF), a contrast sensitivity (CS), and the like. In this way, combinations of multiple optical metrics for the calculation of a goal function or optimizer value can be used in treatments such as intraocular lenses, contact lenses, spectacle lenses, refractive surgery and other laser photoalteration procedures, corneal inlays, conductive keratoplasty procedures, and the like.

[0376] According to some embodiments, composite optical metrics may include linear combinations of individual optical metrics. In certain embodiments, a composite optical metric may include individual weighting coefficients or functions associated with respective individual parameters of the composite optical metric. For example, a goal function or composite optical metric may be represented by the following formula,

\[ m(l) = \sum_{i=1}^{n} k_i M_i(l) \]

where \( k_i \) is a weighting coefficient or function for an \( i \) th optical metric \( M_i(l) \), \( n \) represents the number of individual metrics or parameters, \( l \) is the vergence, and \( m(l) \) is the composite metric. The weighting function may be a weighting coefficient (a number, or a constant), or it may be a function, such as a two-dimensional function. Relatedly, an individual metric of the composite metric may be two-dimensional. According to some embodiments, it is possible to treat the weighting function as apodization functions to eliminate certain spatial frequency information as needed or desired. In some instances, a weighting function can be represented as \( k_i(\rho, \Theta) \) in a polar coordinate system. In some instances, a weighting function can be represented as \( k_i(x,y) \) in a Cartesian coordinate system. According to some embodiments, a weighting function may be represented by a spatial function.

[0377] Where an optical metric is a single number (e.g. Strehl Ratio) then it may be desirable that the weighting function also be a number (e.g. as opposed to a two dimensional function). Where an optical metric is in the frequency domain (e.g. CMTF) then it may be desirable that the weighting function also be in the frequency domain, or optionally as a constant. Where an optical metric is in the spatial domain (e.g. PSF or encircled energy) then it may be desirable that the
weighting function also be in the spatial domain, or optionally as a constant. Both frequency
domain and spatial domain metrics can be expressed in two dimensional representations, such as \( k_1 \)
\((\rho, \Theta)\) in a polar coordinate system or \( k_1 \) \((x,y)\) in a Cartesian coordinate system.

[0378]  An optical metric may be represented by a spatial function. For example, optical metrics
such as a two dimensional point spread function may be represented in normal space \((x,y)\), and
optical metrics such as an optical transfer function or compound modulation transfer function may
be represented in Fourier space or frequency domain \((k_x, k_y)\). In some embodiments, optical
metrics represented in normal or real space may be represented in distance units, such as
millimeters or microns. In some embodiments, optical metrics represented in the frequency space
may be expressed in cycles per degree (e.g. an indication of oscillation behavior). In some
instances, an optical metric may not be represented by a spatial function. For example, optical
metrics such as a Strehl Ratio may be represented as a number or a constant. When summing or
combining optical metrics, it may be helpful to have the optical metrics be represented in the same
space or expressed in similar units. In some cases, it may be possible to convert an optical metric
from one space or representation to another, so that it can conveniently be combined with other
optical metrics. For example, a composite optical metric that combines Strehl Ratio and
Compound Modulation Transfer Function may be straightforward, because Strehl Ratio can be
expressed as a single number. In comparison, a composite optical metric that combines Point
Spread Function and Compound Modulation Transfer Function may involve a conversion of one
of the individual parameters, so that both parameters are represented in the same space. In some
embodiments, a composite optical metric may include parameters that are originally in the Fourier
or frequency domain. According to some embodiments, a weighting function may be normalized.
In some cases, a weighting function may not be normalized. Goal functions or composite optical
metrics, or individual components thereof, can be minimized or maximized as discussed elsewhere
herein. In some instances, optimizer values may be used, where the optimizer value is minimized
(e.g. determining a minimum value in two dimensional space) such that the optical metric is
maximized. In some cases, an optimizer value can correspond to an optical metric, as the mean of
the reciprocal of the optical metric.

[0379]  In certain embodiments, a goal function or composite optical metric may include a
CMTF parameter as one of the individual parameters of the composite. For example, a
combination may include CMTF with Strehl ratio, or CMTF with encircled energy. The following
equation represents a composite optical metric that includes CMTF and Strehl ratio.

\[
m(l) = \ ^{\text{CMTF}}(l) + k_2 \text{SR}(l)
\]
Similarly, the following equation represents a composite optical metric that includes CMTF and visual Strehl ratio,

$$mil) = k_1CMTF(l) + k_2VSR(l)$$

where $VSR(l)$ stands for visual Strehl ratio. In some embodiments, composite optical metrics disclosed herein can include any of the individual metrics described in Thibos, et al, "Accuracy and precision of objective refraction from wavefront aberrations," J. Vision 4, 320-351 (2004), the content of which is incorporated herein by reference.

**Vergence Weighting And Testing Points**

For the treatment or correction of presbyopia or other vision conditions, different individuals may have different needs or desires with regard to their vision. For example, people such as accountants and lawyers, whose day to day professional activities may include reading documents, may require or desire the very best near vision and may be willing to sacrifice some intermediate and distance vision. On the other hand, people such as truck drivers and golfers, whose professional activities may benefit from crystal-clear far vision, may require or desire the best distance vision and may be willing to sacrifice intermediate vision and near vision. Further, engineers who spend large amounts of time viewing a computer screen may wish to optimize their intermediate vision and may be willing to sacrifice some near vision and distance vision.

In the context of vision treatment and correction modalities, the term optical vergence can refer to certain vision testing conditions, for example calculated as the reciprocal of the testing or viewing distance in meters. Such testing conditions can also be referred to as testing points. According to some embodiments of the present invention, optimization of an optical shape may involve factoring in such testing point preferences for individual customization.

**FIG. 59** depicts aspects of an exemplary method 5900 for treating a vision condition of an eye in a particular patient. As shown here, the method includes receiving a vision requirements specification selected for the particular patient, as indicated by step 5910. As discussed elsewhere herein, the vision requirements specification can include a first weighting value for a first viewing distance within a vergence range and a second weighting value for a second viewing distance within the vergence range. The method also includes determining an optical surface shape for the particular patient, as indicated by step 5930. The optical surface shape can be based on the vision requirements specification and an optical metric 5920. Further, the method can include treating the vision condition of the eye of the particular patient by providing a treatment to the patient, as indicated by step 5940. The treatment can include a shape that corresponds to the optical surface
shape. Relatedly, embodiments of the present invention encompass systems and methods for
determining a procedure for treating a vision condition of an eye of a particular patient based on an
optical surface shape. In some cases, a procedure can include ablating a corneal surface or
subsurface of the eye of the particular patient to provide a corneal surface shape that corresponds
to the optical surface shape. In some cases, a procedure can include providing the particular
patient with a contact lens or a spectacle lens having a shape that corresponds to the optical surface
shape. In some cases, a procedure can include providing the particular patient with an intra-ocular
lens having a shape that corresponds to the optical surface shape.

[0386] Relatedly, FIG. 60 depicts aspects of an exemplary method for generating an optical
surface shape for use in treating a vision condition of an eye in a particular patient. As shown
here, the method includes receiving a vision requirements specification selected for the particular
patient, as indicated by step 6010. As discussed elsewhere herein, the vision requirements
specification can includes a first weighting value for a first viewing distance within a vergence
range and a second weighting value for a second viewing distance within the vergence range. The
method also includes generating the optical surface shape for the particular patient, as indicated by
step 6030. The optical surface shape can be based on the vision requirements specification and an
optical metric 6020.

[0387] FIG. 61 depicts aspects of a vision requirements specification 6100, according to
embodiments of the present invention. The vision requirements specification can include a first
weighting value V1 for a first viewing distance D1 within a vergence range 6110 and a second
weighting value V2 for a second viewing distance D2 within the vergence range. In some cases,
the first viewing distance can correspond to a near vision viewing distance 6120 (or near vision
viewing distance range), an intermediate vision viewing distance 6130 (or intermediate vision
viewing distance range), or a distance vision viewing distance 6140 (or distance vision viewing
distance range). Similarly, in some cases, the second viewing distance can correspond to a near
vision viewing distance 6120 (or near vision viewing distance range), an intermediate vision
viewing distance 6130 (or intermediate vision viewing distance range), or a distance vision
viewing distance 6140 (or distance vision viewing distance range). According to some
embodiments, the first weighting value V1 can be different from the second weighting value V2,
and the first viewing distance D1 can be different from the second viewing distance D2. In some
cases, the first weighting value V1 can be greater than the second weighting value V2. In some
cases, the first viewing distance D1 is less than the second viewing distance D2. In some cases,
the first viewing distance D1 is greater than the second viewing distance D2.
In some cases, as depicted in FIG. 61, the first weighting value \( V_1 \) can be less than the second weighting value \( V_2 \). Hence, the vision requirements specification 6100 depicted here may be suitable for an individual who needs or desires excellent distance (or far) vision, and where good near vision is not as important. In contrast, FIG. 62 depicts a vision requirements specification 6200 that is suitable for an individual who needs or desires excellent near and intermediate vision, and where distance vision is not as important. As shown here, the weighting values corresponding to near and intermediate distances are greater in value, as compared to the weighting value corresponding to far distance. In some embodiments, the weighting values corresponding to intermediate and far distances are greater in value, as compared to the weighting value corresponding to near distance. In some embodiments, the weighting values corresponding to near and far distances are greater in value, as compared to the weighting value corresponding to intermediate distance. In some cases, weighting values corresponding to different distances (e.g. near and far) can be balanced to have the same value.

In some cases, weighting values corresponding to different distances (e.g. near and far) can be unbalanced so as to have the different values.

In some cases, weighting values can be selected or customized according to a particular patient preference or treatment protocol. As an example, for a patient who works as an engineer, the weighting can be 0.5 for near, 2.0 for intermediate (e.g. at 0.5 meters or a range including 0.5 meters), 0.5 for distance, or the like. Such a weighting regime can be well suited for use for treating an individual who desires good vision for viewing a computer screen (e.g. higher weighting values for the viewing distance). The weighting values for other occupations may vary. For example, for truck drivers, the weighting can be 0.5 for near, 0.5 for intermediate, and 2.0 for distance. For lawyers, the weighting can be 2.0 for near, 0.5 for intermediate, and 0.5 for distance. The weighting values can be provided in a normalized format (e.g. the values have a mean of 1.0). In some cases, the weighting can correspond to a linear function of viewing distance (or vergence), or to a nonlinear function, depending on the need or desired treatment.

According to exemplary embodiments of the present invention, the weighting values for different viewing distances can be used in conjunction with an optical metric for evaluating or determining an optical surface shape (which may correspond to a target or a treatment shape), for example based on a merit function or optimizer value. As shown in FIG. 63, a merit function 6310 for a target or treatment shape can be determined based on an optical metric value for the shape at the various viewing or testing distances, for example 6320a, 6320b, and 6320c, as applied to the various weighting values for the various viewing or testing distances, for example [VI, DI],
[V2, D2], and [V3, D3]. In this way, a merit function can capture information about the optical surface shape, where the information varies as a function of (e.g. CMTF values at different distances/vergence), and the optical metric information can be modified according to weighting values for various distances/vergences.

Optical metrics such as a compound modulation transfer function (CMTF) can provide a measure of optical quality for the shape throughout a range of vergence or at different spatial distances (l). Exemplary optical metrics (e.g. such as composite optical metrics) that can be used in conjunction with embodiments of the present invention are discussed in US Patent Publication No. 2014/0016091, the content of which is incorporated herein by reference. In some cases, an optical metric can include a compound modulation transfer function (CMTF) parameter having a combination of modulation transfer functions (MTFs) at a plurality of distinct frequencies.

In some cases, the optimizer value or merit function 6310 for an optical surface shape can be calculated based on various parameters associated with the optical metric.

For example, if the distance values are selected for every 0.1 Diopter or 0.1 meter and the vergence range is 3.0 Diopeters, then there will be 30 data points or values to be considered (e.g. one optical metric value for each of the distance values). Based on those 30 values, it is possible to calculate the mean, standard deviation, and peak-to-valley for the optical metric.

The merit function value may be calculated as:

\[ f = \frac{(1+\sigma)(1+\beta)}{\hat{Q}(l)} \]  

(Eq. 1)

where \( \hat{Q}(l) \), \( \sigma \), and \( \beta \) are the mean, standard deviation, and peak-to-valley (maximum to minimum), respectively, of the optical metric for CMTF or other metrics like Strehl Ratio (SR), modulation transfer function (MTF), point spread function (PSF), encircled energy (EE), MTF volume or volume under MTF surface (MTFV), contrast sensitivity (CS), or various combinations thereof. Exemplary metrics are described in US Patent Application No. 13/732,124 filed December 31, 2012, the content of which is incorporated herein by reference.

According to some embodiments, once an optical metric is defined, a merit function can be constructed so that a function minimization algorithm can be applied to maximize the optical metric. In some cases, the merit function is inversely proportional to the optical metric. In some cases, it is possible to maximize the optical metric while ensuring the optical metric curve over the range does not fluctuate excessively. An exemplary merit function or optimizer value is provided by Eq. 1.
As noted above, different weighting values can be associated with the different distances/vergence. The different weighting values can be applied to the optical metric values when calculating the merit function. Hence, whereas Eq. 1 in some embodiments uses the mean \( \hat{Q}(l) \) of the optical metric over the entire vergence range, it is possible to construct a merit function that replaces the mean \( \hat{Q}(l) \) with the following expression:

\[
\hat{Q}(l) = \frac{1}{\sum_{i=1}^{t} S_i} \sum_{i=1}^{t} S_i C_i^{(m)}(l)
\]

(Eq. 2)

where \( s_i \) is the weighting coefficient or value for the given testing or viewing distance (vergence). This expression for \( \hat{Q}(l) \) can be substituted in the merit function calculation depicted in Eq. 1. The optical metric \( C_i^{(m)} \) can be any desired optical metric (e.g., CMTF), \( (m) \) can be the number of spatial frequencies (e.g., for calculating MTF), and \( l \) can refer to the vergence.

The summation \( \sum \) in the denominator indicates that the weighting coefficients or values \( s_i \) are summed. Here, different testing distances can correspond to different viewing distances, which can be throughout the vergence range \( (i = 1 \) to \( t) \). At a distance of infinity, the vergence can be considered as zero. As indicated here, \( t \) can represent the number of samples. For example, for \( t = 10 \), each individual sample can represent 1/10 of the range. Hence, \( t \) can refer to the degree of granularity for which the metrics are calculated. In some cases, the sampling points can correspond to a linear distribution. In some cases, the sampling points can correspond to a bell curve distribution. In some cases, the sampling points can correspond to a quadratic distribution. Where the distribution is non-linear, it may be desirable to use a greater number of sampling points, according to some embodiments.

For example, certain approaches for the treatment of presbyopia and other vision conditions can involve the use of a shape that is optimized based on the optimizer value as a function of an optical metric such as Strehl ratio, modulation transfer function (MTF), encircled energy, compound modulation transfer function (CMTF), and the like. Aspects of such optical metrics are disclosed in U.S. Patent Nos. 7,320,517; 7,475,986; 7,862,170; 8,029,137; and 8,220,925, the contents of which are incorporated herein by reference. Related vision treatment modalities may include the use of one or more combinations of multiple optical metrics for the calculation of an optimizer value, for example as described in U.S. Patent Publication No. 2014/0016091, the content of which is incorporated herein by reference.
In some cases, as illustrated in FIGS. 64A and 64B, first and second weighting values (e.g. VI, V2) can be members of a weighting value distribution (e.g. 6410a, 6410b) that is linear across a vergence range that includes first and second viewing distances (e.g. D1, D2).

As discussed elsewhere herein, either linear or nonlinear functions can be used for a weighting value distribution. For example, quadratic functions, quartic functions, Gaussian distribution functions, and the like, may be used.

In some cases, as illustrated in FIGS. 65A, 65B, 65C, and 65D, first and second weighting values (e.g. VI, V2) can be members of a weighting value distribution (e.g. 6510a, 6510b, 6510c, 6510d) that is non-linear across a vergence range that includes first and second viewing distances (e.g. D1, D2).

FIGS. 66A and 66B depict aspects of various weighting value distributions (e.g. 6610a, 6610b) according to embodiments of the present invention. As shown in FIG. 66A, a weighting value distribution 6610a can be defined by a function having one or more peaks (e.g. 6620a, 6630a) within a vergence range. In some cases, the weighting value distribution 6610a can be defined by two sub-distributions or sub-functions 6612a, 6614a. As shown in FIG. 66B, a weighting value distribution 6610b can be defined by a step function having one or more steps (e.g. 6620b, 6630b, 6640b) within a vergence range.

According to some embodiments, the weighting value distribution depicted in FIG. 67B is preferred over the weighting value distribution depicted in FIG. 67A, for example because of lower peak weighting value, and broader distribution across vergence range. Put another way, a more gentle change in the weighting function or weighting value distribution can provide less fluctuation of the values, thereby reducing the possibility of having too low or insufficient weighting for a given vergence. For example, if a 2.8 weight is assigned for intermediate and 0.1 weights are assigned for near and distance, respectively, then both near and distance vision may not be sufficient, and the patient may not effectively be able to read and see distance objects. As noted elsewhere herein, weighting values can be provided in a normalized format (e.g. the values have a mean of 1.0).

According to some embodiments, a scaling operation can be applied. Accordingly, the ratio or relationship between different weighting values for different distances may be more relevant that the magnitude of the individual weighting values themselves. In some cases, it is possible to use a multiplication scaling factor to compare to the weights at different distances (vergence). For example, a weighting at intermediate can be 2 times (double) a weighting at near
and far, or a weight at far can be 3 times (triple) a weight at near and 2 times (double) a weight at intermediate.

[0407] FIGS. 68-71 provide example distance, intermediate, and near vision experienced by an eye pre- and post-treatment according to some embodiments of the present invention. For example, FIGS. 68A, 68B, and 68C illustrate example distance, intermediate, and near vision experienced by an eye without any vision correction, respectively. In particular, FIG. 68A illustrates an example distance view down a street, FIG. 68B illustrates an example intermediate view of a computer screen, and FIG. 68C illustrates an example near view of news text. As can be seen in each of FIGS. 68A-68C, each of distance, intermediate, and near vision experienced by the eye will be out of focus or otherwise blurry without any vision correction.

[0408] FIGS. 69A, 69B, and 69C illustrate example distance, intermediate, and near vision experienced by an eye corrected with an optimization or emphasis for near vision viewing distances with less emphasis on far and intermediate viewing distances, respectively. Put in another way, the distance, intermediate, and near viewing distances are each corrected for, however the distance and intermediate viewing distances are corrected with less optimization or emphasis compared to the optimization or emphasis for near vision viewing distances. As illustrated in FIG. 69C, the example near view of the news text is now in focus with the optimization for near vision correction. The example intermediate view of the computer screen in FIG. 69B for the eye corrected with optimization for near vision is more in focus or less blurry than the intermediate view of the computer screen shown in FIG. 68B for the eye without correction, but is still slightly out of focus compared to the example near view of the news text shown in FIG. 69C. Lastly, the example distance view of the street in FIG. 69A for the eye corrected with the optimization for near vision distance only is more in focus or less blurry than the example distance view of the street in FIG. 68A for the eye without correction, but is also out of focus and more out of focus than the example intermediate view of the computer screen in FIG. 69B for the eye corrected with optimization for near vision correction only.

[0409] FIGS. 70A, 70B, and 70C illustrate example distance, intermediate, and near vision conditions experienced by an eye corrected with an optimization or emphasis on both intermediate and near viewing distances and with less of an emphasis on far viewing distances, respectively. The optimization for both intermediate and near viewing distance correction may correspond to the vision requirements specification graph shown in FIG. 62, where the weighting values corresponding to near and intermediate distances are greater in value, as compared to the weighting value corresponding to far distance. Put in another way, the distance, intermediate, and
near viewing distances are each corrected for, however the far viewing distances are corrected with less optimization or emphasis compared to the optimization or emphasis for both intermediate and near vision viewing distances. The weighting values for intermediate and near vision viewing distances may be the same or may be different in some embodiments. As shown in FIG. 70C, the example near view of the news text when the eye is corrected with optimization or emphasis for both intermediate and near viewing distances is less in focus compared to the example near view of the news text shown in FIG. 69C when the eye is corrected with optimization for near vision viewing distances only; however, the example near view of the news text when the eye is corrected with optimization for both intermediate and near viewing distances in FIG. 70C is still more in focus than the example near view of the text shown in FIG. 68C, when the eye is without any vision correction. While the example near view of the text is slightly less in focus when the eye is corrected with optimization for both intermediate and near viewing distances compared to when the eye is corrected with optimization for near vision viewing distances only, the example intermediate view of the computer screen shown in FIG. 70B where the eye is corrected with optimization for both intermediate and near viewing distances is slightly more in focus or less blurry compared to the example intermediate view of the computer screen shown in FIG. 69B where the eye is corrected with optimization for near viewing distances only (i.e., less emphasis for far and intermediate viewing distances). Further, as shown in FIG. 70A, the example distance view of the street for the eye corrected with emphasis for both intermediate and near vision is more out of focus that the example intermediate view of the computer screen in FIG. 70B for the eye corrected with optimization for both intermediate and near vision viewing distances only, but is in more focus or otherwise less blurry than the example distance view of the street in FIG. 68A for the eye without any correction. The example distance view of the street for the eye corrected with optimization for both intermediate and near vision as shown in FIG. 70A may be, in some instances, of similar quality as the example distance view of the street for the eye with optimization for near vision correction only shown in FIG. 69A.

[0410] FIGS. 71A, 71B, and 71C illustrate example distance, intermediate, and near vision conditions experienced by an eye corrected with optimization or emphasis for both distance and intermediate viewing distances and less emphasis on near viewing distances, respectively. The optimization for both distance and intermediate viewing distance correction may be associated with a vision requirements specification where the weighting values corresponding to distance and intermediate are greater in value, as compared to the weighting value corresponding to near distance. As illustrated in FIG. 71A, the example distance view of the street for the eye corrected with optimization for both distance and intermediate viewing distances is more in focus than the
example distance view of the street for the eye corrected with optimization for near viewing distances only as shown in FIG. 69A and is more in focus than the example distance view of the street for the eye corrected with optimization for both intermediate and near viewing distances as shown in FIG. 70A. Further, the example intermediate view of the computer screen experienced by the eye corrected with optimization for both distance and intermediate viewing distances as shown in FIG. 71B is more in focus than the example intermediate view of the computer screen experienced by the eye corrected with optimization for near viewing distances only as shown in FIG. 69B. In some embodiments, the intermediate view of the computer screen for the eye corrected with optimization for both distance and intermediate viewing distances as shown in FIG. 71B may be comparable to the example intermediate view of the computer screen for the eye corrected with optimization for both intermediate and near vision shown in FIG. 70B. As shown in FIG. 71C, the example near view of the news text for the eye corrected with optimization for both distance and intermediate viewing distances may be more blurry than the example near view of the news text experienced by the eye corrected with optimization for both intermediate and near distances shown in FIG. 70C and by the eye corrected with optimization for near distances only shown in FIG. 69C. While the example near view of the news text for the eye corrected with optimization for both distance and intermediate viewing distances may be more blurry than the example new views of the news text experienced by the eye corrected with optimization for near distances or corrected with optimization for both intermediate and near distances, the example near view of the news text for the eye corrected with optimization for both distance and intermediate viewing shown in FIG. 71C may nevertheless be less blurry or more in focus than the example near view of the news text experienced by the eye that has no correction as shown in FIG. 68C.

[0411] In further embodiments, a weighting function may not be applied to distance/far, intermediate, and near vision specification. Instead, the correction may be based on a function that applies to a portion of or the entire range of the vision field or vergence range. As an example using target distance, the weighting function may have values at various target distances. For example, the weighting function may have values associated with one or more of 100 m, 10 m, 5 m, 4 m, 3 m, 2 m, 1 m, 50 cm, 40 cm, 30 cm, 20 cm, etc. As another example using vergence range, the weighting function may have values at various vergences. For example, the weighting function may have values associated with one or more of 0D, 0.2D, 0.4D, 0.6D, 0.8D, 1D, 1.5D, 2D, 2.5D, 3D, 3.5D, 4D, etc. Therefore, the application of the vergence weighting is more general than just 2 or 3 target distances. This idea may be applied to both eyes similarly or differently.
It should be understood that these example views are provided by way of illustration only and that eye corrections may be customized with preferred optimization or emphasis in many other manners per the embodiments disclosed herein. For example, an eye may be corrected with an emphasis for far viewing distances only or with an emphasis for intermediate distances only. Additionally as set forth above, in some embodiments, the weighting values corresponding to near and far distances may be greater in value, as compared to the weighting value corresponding to intermediate distance. In some cases, weighting values corresponding to different distances (e.g., near and far) can be balanced to have the same value. Additionally, it should be appreciated that the same correction may be applied to both eyes of a patient or different corrections with different emphases may be applied to each eye to provide additional treatment customization. For example, one eye may be optimized for distance viewing while the other eye may be optimized for near viewing distances (e.g., monovision). Alternatively, one eye may be optimized for distance, and the other eye may be optimized for both near and intermediate viewing distances (e.g., FIGS. 70A-70C). Still further, one eye may be corrected with an emphasis for near viewing distances (e.g., FIGS. 69A-69C), and the other eye may be corrected with an emphasis for both distance and intermediate viewing distances (e.g., FIGS. 71A-71C). Thus, in some embodiments, one eye may be optimized for a particular vergence weighting function and the other eye may be corrected for another particular vergence weighting function. For example, one eye may be corrected with an emphasis for both distance and intermediate viewing distances (e.g., FIGS. 71A-71C) and the other eye may be corrected with an emphasis for both intermediate and near viewing distances (e.g., FIGS. 70A-70C). Further, it some embodiments, a correction applied to one eye may be customized based on eye dominance. Also, it should be appreciated that the patient eye may be corrected with a weighting function having values associated with a number of target distances or vergences. With such an application the vergence weighting may be more general than just 2 or 3 target distances or vergence ranges.

Accordingly, a method for treating a vision condition of a particular patient may be provided in some embodiments. The method may include receiving a vision requirement specification selected for the particular patient. The vision requirement specification may include a first weighting function for a first viewing distance within a first vergence range for the first eye and a second weighting function for a second viewing distance within a second vergence range for a second eye. The method may further include determining an optical surface shape for each eye of the particular patient. The optical surface shape may be based on the vision requirements specification of the particular eye and an optical metric. Thereafter, the method may include treating the vision condition of the eyes of the particular patient by providing a treatment to each
eye of the patient. The treatment may include a shape that corresponds to the optical surface shape.

[0414] In some embodiments, the first weighting function includes a first weighting value associated with a far distance and a second weighting value associated with a near distance. The first weighting value of the first weighting function may be greater than the second weighting value of the first weighting function. The second weighting function may include a first weighting value associated with the far distance and a second weighting value associated with the near distance. The second weighting value may optionally be greater than the first weighting value.

[0415] In further aspects, the first weighting function may include a first weighting value associated with a far distance, a second weighting value associated with an intermediate distance, and a third weighting value associated with a near distance. The first weighting value of the first weighting function may be greater than the second and third weighting value of the first weighting function. The second weighting function may include a first weighting value associated with the far distance, a second weighting value associated with the intermediate distance, and a third weighting value associated with the near distance. The first weighting value of the second weighting function may be less than the second and third weighting values of the second weighting function in some embodiments. Optionally, the second and third weighting values of the second weighting function may be the same.

[0416] In certain aspects, the first weighting function may include a first weighting value associated with a far distance, a second weighting value associated with an intermediate distance, and a third weighting value associated with a near distance. The third weighting value of the first weighting function may be greater than the second and third weighting value of the first weighting function. The second weighting function includes a first weighting value associated with the far distance, a second weighting value associated with the intermediate distance, and a third weighting value associated with the near distance. The third weighting value of the second weighting function may be less than the first and second weighting values of the second weighting function. Optionally, the first and second weighting values of the second weighting function may be the same. Thus, in some embodiments, the first vergence weighting function and the second vergence weighting function may be different.

[0417] As discussed elsewhere herein, embodiments of the present invention encompass systems and methods that involve determining an optical surface shape for a particular patient (e.g. where the optical surface shape is based on a vision requirements specification and an optical metric), and treating a vision condition of an eye of the particular patient by providing a treatment to the
patient, where the treatment is based on a shape that corresponds to the optical surface shape.

Relatedly, in some cases exemplary systems and methods can involve generating a treatment shape for treating the eye of the patient, where the treatment shape is based on the optical surface shape. In some cases, the treatment shape can be an intraocular lens treatment shape. In some cases, the treatment shape can be a contact lens shape. In some cases, the treatment shape can correspond to a laser ablation or photodisruption target shape.

[0418] Each of the above calculations may be performed using a computer or other processor having hardware, software, and/or firmware. The various method steps may be performed by modules, and the modules may comprise any of a wide variety of digital and/or analog data processing hardware and/or software arranged to perform the method steps described herein. The modules optionally comprising data processing hardware adapted to perform one or more of these steps by having appropriate machine programming code associated therewith, the modules for two or more steps (or portions of two or more steps) being integrated into a single processor board or separated into different processor boards in any of a wide variety of integrated and/or distributed processing architectures. These methods and systems will often employ a tangible media embodying machine-readable code with instructions for performing the method steps described above. Suitable tangible media may comprise a memory (including a volatile memory and/or a non-volatile memory), a storage media (such as a magnetic recording on a floppy disk, a hard disk, a tape, or the like; on an optical memory such as a CD, a CD-R/W, a CD-ROM, a DVD, or the like; or any other digital or analog storage media), or the like.

[0419] As the analytical solutions described herein some or all of these method steps may be performed with computer processors of modest capability, i.e., a 386 processor from Intel™ may be enough to calculate the Zernike coefficients, and even 286 processor may be fine. Scaling of Zernike coefficients was described by Jim Schweigerling, "Scaling Zernike Expansion Coefficients to Different Pupil Sizes," J. Opt. Soc. Am. A 19, pp 1937—1945 (2002). No special memory is needed (i.e., no buffers, all can be done as regular variables or using registers). Also, it can be written in any of a wide variety of computer languages, with the exemplary embodiment employing C++. This exemplary embodiment comprises code which performs the Zernike coefficient calculation, shape combination (combining a regular aberration treatment prescription as well as the presbyopia shape), and provides graphical output for reporting purpose. It was written in C++ with Borland C++ Builder™ 6, and it is run with a laptop of 1.13GHz CPU having 512Mb of memory.
As noted above, a variety of output data can be generated by the systems and methods of the present invention. Such outputs may be used for a variety of research, comparison, prediction, diagnostic, and verification operations. The outputs may be evaluated directly, or they may be used as input into the system for further analysis. In some embodiments, the outputs will be used to model the effect of an ocular treatment prior to application. In other embodiments, the outputs will be used to evaluate the effect of an ocular treatment after application. The outputs may also be used to design ocular treatments. Relatedly, it is possible to create treatment tables based on outputs of the instant invention.

While the exemplary embodiments have been described in some detail, by way of example and for clarity of understanding, those of skill in the art will recognize that a variety of modification, adaptations, and changes may be employed. Hence, the scope of the present invention should be limited solely by the claims.
WHAT IS CLAIMED IS:

1. A method for treating a vision condition of an eye in a particular patient, the method comprising:
   receiving a vision requirements specification selected for the particular patient, the vision requirements specification comprising a first weighting value for a first viewing distance within a vergence range and a second weighting value for a second viewing distance within the vergence range;
   determining an optical surface shape for the particular patient, the optical surface shape based on the vision requirements specification and an optical metric; and
   treating the vision condition of the eye of the particular patient by providing a treatment to the patient, the treatment comprising a shape that corresponds to the optical surface shape.

2. The method according to claim 1, wherein the first viewing distance comprises a member selected from the group consisting of a near vision viewing distance, an intermediate vision viewing distance, and a distance vision viewing distance.

3. The method according to claim 2, wherein the second viewing distance comprises a member selected from the group consisting of a near vision viewing distance, an intermediate vision viewing distance, and a distance vision viewing distance.

4. The method according to claim 1, wherein the first weighting value is different from the second weighting value and the first viewing distance is different from the second viewing distance.

5. The method according to claim 1, wherein the first weighting value is greater than the second weighting value.

6. The method according to claim 1, wherein the first weighting value is less than the second weighting value.

7. The method according to claim 1, wherein the first viewing distance is greater than the second viewing distance.

8. The method according to claim 1, wherein the first viewing distance is less than the second viewing distance.
9. The method according to claim 1, wherein the optical metric is a composite optical metric.

10. The method according to claim 1, wherein the optical metric comprises a compound modulation transfer function (CMTF) parameter comprising a combination of modulation transfer functions (MTF's) at a plurality of distinct frequencies.

11. The method according to claim 1, wherein the first and second weighting values are members of a weighting value distribution that is linear across a vergence range that comprises the first and second viewing distances.

12. The method according to claim 1, wherein the first and second weighting values are members of a weighting value distribution that is non-linear across a vergence range that comprises the first and second viewing distances.

13. The method according to claim 1, wherein the step of treating the vision condition of the eye of the particular patient comprises a procedure selected from the group consisting of:

   ablating a cornea of the eye of the particular patient to provide a corneal surface shape that corresponds to the optical surface shape,
   providing the particular patient with a contact lens or a spectacle lens having a shape that corresponds to the optical surface shape, and
   providing the particular patient with an intra-ocular lens having a shape that corresponds to the optical surface shape.

14. A method for generating an optical surface shape for use in treating a vision condition of an eye in a particular patient, the method comprising:

   receiving a vision requirements specification selected for the particular patient, the vision requirements specification comprising a first weighting value for a first viewing distance within a vergence range and a second weighting value for a second viewing distance within the vergence range; and
   generating the optical surface shape for the particular patient, the optical surface shape based on the vision requirements specification and an optical metric.

15. The method according to claim 14, further comprising determining a procedure for treating the vision condition of the eye of the particular patient based on the optical surface shape.
16. The method according to claim 15, wherein the procedure comprises a member selected from the group consisting of:
   ablating a corneal surface of the eye of the particular patient to provide a corneal surface shape that corresponds to the optical surface shape,
   providing the particular patient with a contact lens or a spectacle lens having a shape that corresponds to the optical surface shape, and
   providing the particular patient with an intra-ocular lens having a shape that corresponds to the optical surface shape.

17. The method according to claim 14, wherein the first viewing distance comprises a member selected from the group consisting of a near vision viewing distance, an intermediate vision viewing distance, and a distance vision viewing distance.

18. The method according to claim 17, wherein the second viewing distance comprises a member selected from the group consisting of a near vision viewing distance, an intermediate vision viewing distance, and a distance vision viewing distance.

19. The method according to claim 14, wherein the first weighting value is different from the second weighting value and the first viewing distance is different from the second viewing distance.

20. The method according to claim 14, wherein the first weighting value is greater than the second weighting value.

21. The method according to claim 14, wherein the first weighting value is less than the second weighting value.

22. The method according to claim 14, wherein the first viewing distance is greater than the second viewing distance.

23. The method according to claim 14, wherein the first viewing distance is less than the second viewing distance.

24. The method according to claim 14, wherein the optical metric is a composite optical metric.
25. The method according to claim 14, wherein the optical metric comprises a compound modulation transfer function (CMTF) parameter comprising a combination of modulation transfer functions (MTF's) at a plurality of distinct frequencies.

26. The method according to claim 14, wherein the first and second weighting values are members of a weighting value distribution that is linear across a vergence range that comprises the first and second viewing distances.

27. The method according to claim 14, wherein the first and second weighting values are members of a weighting value distribution that is non-linear across a vergence range that comprises the first and second viewing distances.

28. A system for establishing an optical surface shape for use in treating a vision condition of an eye in a particular patient, the system comprising:

   an input that receives a vision requirements specification selected for the particular patient, the vision requirements specification comprising a first weighting value for a first viewing distance within a vergence range and a second weighting value for a second viewing distance within the vergence range; and

   a data processing module comprising a processor and a tangible non-transitory computer readable medium, the computer readable medium programmed with a computer application that, when executed by the processor, causes the processor to establish the optical surface shape for the eye of the particular patient, the optical surface shape based on the vision requirements specification received by the input and an optical metric.

29. The system according to claim 28, wherein the computer application, when executed by the processor, causes the processor to determine a protocol for treating the vision condition of the eye of the particular patient based on the optical surface shape.

30. The system according to claim 29, wherein the protocol comprises a member selected from the group consisting of:

   a photodisruption procedure for a corneal tissue of the eye of the particular patient, the photodisruption procedure configured to provide a corneal surface shape that corresponds to the optical surface shape,

   a contact lens or a spectacle lens procedure for the eye of the particular patient, the contact lens or spectacle lens procedure comprising a lens having a shape that corresponds to the optical surface shape, and
an intra-ocular lens procedure for the eye of the particular patient, the intra-ocular lens procedure comprising a lens having a shape that corresponds to the optical surface shape.

31. The system according to claim 28, wherein the first viewing distance comprises a member selected from the group consisting of a near vision viewing distance, an intermediate vision viewing distance, and a distance vision viewing distance.

32. The system according to claim 31, wherein the second viewing distance comprises a member selected from the group consisting of a near vision viewing distance, an intermediate vision viewing distance, and a distance vision viewing distance.

33. The system according to claim 28, wherein the first weighting value is different from the second weighting value and the first viewing distance is different from the second viewing distance.

34. The system according to claim 28, wherein the first weighting value is greater than the second weighting value.

35. The system according to claim 28, wherein the first weighting value is less than the second weighting value.

36. The system according to claim 28, wherein the first viewing distance is greater than the second viewing distance.

37. The system according to claim 28, wherein the first viewing distance is less than the second viewing distance.

38. The system according to claim 28, wherein the optical metric is a composite optical metric.

39. The system according to claim 28, wherein the optical metric comprises a compound modulation transfer function (CMTF) parameter comprising a combination of modulation transfer functions (MTF’s) at a plurality of distinct frequencies.

40. The system according to claim 28, wherein the first and second weighting values are members of a weighting value distribution that is linear across a vergence range that comprises the first and second viewing distances.
41. The system according to claim 28, wherein the first and second weighting values are members of a weighting value distribution that is non-linear across a vergence range that comprises the first and second viewing distances.

42. A computer program product for generating an optical surface shape for use in treating a vision condition of an eye in a particular patient, the computer program product embodied on a tangible non-transitory computer readable medium, comprising:
   code for accessing a vision requirements specification selected for the particular patient, the vision requirements specification comprising a first weighting value for a first viewing distance within a vergence range and a second weighting value for a second viewing distance within the vergence range; and
   code for generating the optical surface shape for the particular patient, the optical surface shape based on the vision requirements specification and an optical metric.

43. The computer program product according to claim 42, further comprising:
   code for determining a protocol for treating the vision condition of the eye of the particular patient based on the optical surface shape.

44. The computer program product according to claim 43, wherein the protocol comprises a member selected from the group consisting of:
   a photodisruption procedure for a corneal tissue of the eye of the particular patient, the photodisruption procedure configured to provide a corneal surface shape that corresponds to the optical surface shape,
   a contact lens or a spectacle lens procedure for the eye of the particular patient, the contact lens or spectacle lens procedure comprising a lens having a shape that corresponds to the optical surface shape, and
   an intra-ocular lens procedure for the eye of the particular patient, the intra-ocular lens procedure comprising a lens having a shape that corresponds to the optical surface shape.

45. The computer program product according to claim 42, wherein the first viewing distance comprises a member selected from the group consisting of a near vision viewing distance, an intermediate vision viewing distance, and a distance vision viewing distance.

46. The computer program product according to claim 45, wherein the second viewing distance comprises a member selected from the group consisting of a near vision viewing distance, an intermediate vision viewing distance, and a distance vision viewing distance.
47. The computer program product according to claim 42, wherein the first weighting value is different from the second weighting value and the first viewing distance is different from the second viewing distance.

48. The computer program product according to claim 42, wherein the first weighting value is greater than the second weighting value.

49. The computer program product according to claim 42, wherein the first weighting value is less than the second weighting value.

50. The computer program product according to claim 42, wherein the first viewing distance is greater than the second viewing distance.

51. The computer program product according to claim 42, wherein the first viewing distance is less than the second viewing distance.

52. The computer program product according to claim 42, wherein the optical metric is a composite optical metric.

53. The computer program product according to claim 42, wherein the optical metric comprises a compound modulation transfer function (CMTF) parameter comprising a combination of modulation transfer functions (MTF’s) at a plurality of distinct frequencies.

54. The computer program product according to claim 42, wherein the first and second weighting values are members of a weighting value distribution that is linear across a vergence range that comprises the first and second viewing distances.

55. The computer program product according to claim 28, wherein the first and second weighting values are members of a weighting value distribution that is non-linear across a vergence range that comprises the first and second viewing distances.

56. A method for treating a vision condition of a particular patient, the method comprising:

   receiving a vision requirement specification selected for the particular patient, the vision requirement specification comprising a first weighting function for a first viewing distance within a first vergence range for the first eye and a second weighting function for a second viewing distance within a second vergence range for a second eye;
determining an optical surface shape for each eye of the particular patient, the optical surface shape based on the vision requirements specification of the particular eye and an optical metric; and

treating the vision condition of the eyes of the particular patient by providing a treatment to each eye of the patient, the treatment comprising a shape that corresponds to the optical surface shape.

57. The method of claim 56, wherein the first weighting function includes a first weighting value associated with a far distance and a second weighting value associated with a near distance and wherein the first weighting value of the first weighting function is greater than the second weighting value of the first weighting function; and wherein the second weighting function includes a first weighting value associated with the far distance and a second weighting value associated with the near distance, and wherein the second weighting value is greater than the first weighting value.

58. The method of claim 56, wherein the first weighting function includes a first weighting value associated with a far distance, a second weighting value associated with an intermediate distance, and a third weighting value associated with a near distance, and wherein the first weighting value of the first weighting function is greater than the second and third weighting value of the first weighting function; and wherein the second weighting function includes a first weighting value associated with the far distance, a second weighting value associated with the intermediate distance, and a third weighting value associated with the near distance, and wherein the first weighting value of the second weighting function is less than the second and third weighting values of the second weighting function.

59. The method of claim 58, wherein the second and third weighting values of the second weighting function are the same.

60. The method of claim 56, wherein the first weighting function includes a first weighting value associated with a far distance, a second weighting value associated with an intermediate distance, and a third weighting value associated with a near distance, and wherein the third weighting value of the first weighting function is greater than the second and third weighting value of the first weighting function; and wherein the second weighting function includes a first weighting value associated with the far distance, a second weighting value associated with the intermediate distance, and a third weighting value associated with the near distance, and wherein
the third weighting value of the second weighting function is less than the first and second weighting values of the second weighting function.

61. The method of claim 60, wherein the first and second weighting values of the second weighting function are the same.

62. The method of claim 56, wherein the first vergence weighting function and the second vergence weighting function are different.
FIG. 4B
User Input
1. Pupil size
2. Residual accommodation
3. Vergence need

Initialization
Initial shape + initial result

Optical Metric
1. Strehl Ratio
2. MTF
3. Encircled energy
4. CMTF
5. MTFV
6. CS

Optimizer
1. Downhill Simplex
2. Direction Set (Powell’s)
3. Simulated Annealing

Is the result good enough?

Output Shape and result

FIG. 5
Trail Shape Input

\[ W(r) = ar + br^2 + cr^3 + dr^4 + \ldots \]

Total Wavefront Aberrations
1. Presbyopia shape
2. Chromatic aberrations
3. Vergence induced aberrations

Retina Image Generation (PSF)
1. Stiles-Crawford effect
2. Retinal spectral response
3. Five-point polychromatic PSF

Optimizing Function
1. Three-point MTF normalization
2. Strehl Ratio
3. N-point MTF normalization
4. Encircled energy at specific field of view
5. MTF at specific frequency

Optimizer
1. Area under optimizing function over vergence
2. Fluctuation adjusted area under optimizing function over vergence

Result

FIG. 6
FIG. 8C
OPTIMIZER VALUE FOR VARIOUS CORRECTIONS

FIG. 11A

OPTIMIZER VALUE FOR VARIOUS CORRECTIONS

FIG. 11B
FIG. 15 (Cont.)
FIG 16 (Cont.)

MTF Variation

VISX Research Projects Presbyopia Optimal eye chart1.bmp
FIG. 21A

FIG. 21B
FIG. 22

CMTF for Various Corrections

- No Correction
- Bi-focal
- VISX
- Multi-focal
- Optimal

Distance vergence in diopters
FIG. 23
Emmetropic Eye

Bi-Focal

Central Add

Multi-Focal

Optimized

SIMULATED EYE CHART

FIG. 24
Input that accepts a set of patient parameters

Module that determines an optical surface shape for the particular patient based on the set of patient parameters, using a gauge of optical quality appropriate for presbyopia of an eye

Processor that generates an ablation profile

Laser system that directs energy laser onto the cornea according to the ablation profile

Eye of particular patient

FIG. 25
SCATTERPLOT: PUPIL SIZE (DIM) vs. 6M UC NePhLM ADJUSTED
6M UC NePhLM ADJUSTED = -2.103 + .37879 * PUPIL SIZE (DIM)
CORRELATION: r = .90387

FIG.32

SCATTERPLOT: PUPIL SIZE (DIM) vs. 6M MRS
6M MRS = -2.871 + .42262 * PUPIL SIZE (DIM)
CORRELATION: r = .79275

FIG.33
Scatterplot: 6M MRSE vs. 6M UC PhHILM
6M UC PhHILM = -0.441 - 1.342 * 6MMRSE
Correlation: -0.6033
Scatterplot: 6M MRSE vs. 6M UC NePhLM adjusted

6M UC NePhLM adjusted = \(0.34483 + 0.67328 \times 6M\text{MRSE}\)

Correlation: \(r = 0.92963\)

FIG. 35
Scatterplot WS Pupil vs. 1M UA NePhLM
Correlation: $r = 0.55797$

1M UA NePhLM = 1.1664 + 0.6287 * WS Pupil
FIG. 38

Mean Accommodation Amplitude

\[ y = 0.0001x^3 - 0.0157x^2 + 0.2319x + 11.007 \]

\[ R^2 = 0.9975 \]
Module that scales a dimension of a central portion of a refractive shape based on the pupil dimension of the particular patient and an attribute of at least one eye previously treated with the prescriptive shape

Processor that generates an ablation profile

Laser system that directs energy laser onto the cornea according to the ablation profile

eye of particular patient

FIG. 39
2-Term solution: Effective power over a range of pupil sizes (4.5 mm to 6 mm)

FIG. 41

2-Term solution: Presbyopia shape

FIG. 40
FIG. 43
3-Term solution: Effective power over a range of pupil sizes (3.5mm to 6mm)

FIG. 42
3-Term solution: presbyopia shape
4-Term solution: Effective power over a range of pupil sizes (3.5mm to 6mm)
Unfavorable S-curve and the favorable reverse Z-curve: effective power

Unfavorable S-curve and the favorable reverse Z-curve: Presbyopia Shapes

FIG. 46A

FIG. 46B
**FIG. 50A**

Effective Power

- Power in dioplers
- Pupil Size in mm

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**FIG. 50B**

Presbyopia Shapes

- Ablation depth in microns
- Distance from pupil center in mm

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SUBSTITUTE SHEET (RULE 26)
PSF CROSS SECTION WITH DIFFERENT DEFOCUS (RENORMALIZED)

FIG. 56
obtaining a plurality of through-focus compound modulation transfer function (CMTF) values for a vision treatment shape

comparing the plurality of through-focus CMTF values to a CMTF threshold value

evaluating the image quality based on the comparison between the through-focus CMTF values and the CMTF threshold value

FIG. 57
obtaining a plurality of through-focus CMTF values for a vision treatment shape, where the CMTF values are based on a CMTF spatial frequency set

obtaining a plurality of through-vergence convolved images based on the vision treatment shape and a point spread function

determining the CMTF threshold value for the CMTF spatial frequency set based on the plurality of through-focus CMTF values and the plurality of convolved images

FIG. 58
FIG. 59

RECEIVING A VISION REQUIREMENTS SPECIFICATION SELECTED FOR A PARTICULAR PATIENT

OPTICAL METRIC

DETERMINING AN OPTICAL SURFACE SHAPE FOR THE PATIENT

PROVIDING A TREATMENT TO THE PATIENT (TREATMENT IS BASED ON OPTICAL SURFACE SHAPE)

FIG. 60

RECEIVING A VISION REQUIREMENTS SPECIFICATION SELECTED FOR A PARTICULAR PATIENT

OPTICAL METRIC

GENERATING AN OPTICAL SURFACE SHAPE FOR THE PATIENT
FIG. 65A

FIG. 65B
**FIG. 65C**

- **WEIGHTING VALUE**
- **VERGENCE RANGE**
- **VIEWING DISTANCE**
- **(HIGH) (INTERMEDIATE) (NEAR) (FAR) (LOW)**
- 

**FIG. 65D**

- **WEIGHTING VALUE**
- **VERGENCE RANGE**
- **VIEWING DISTANCE**
- **(HIGH) (INTERMEDIATE) (NEAR) (FAR) (LOW)**
- 

**[V1, D1]**

**[V2, D2]**

**[6510c]**

**[6510d]**
FIG. 66A

FIG. 66B
Without Correction

Near
FIG. 68C

Intermediate
FIG. 68B

Distance
FIG. 68A
Optimized for Near Only

STOCKS SANK Wednesday, with tech issues leading the drop after Netel said it fired several top executives. Worries about higher interest rates continue to pressure indices. The Dow industrials sank 135.56, or 1.3%, to 10342.60, while the Nasdaq composite tumbled 42.99, or 2.1%, to 1989.54.

CAMCAST DROPPED its bid for Walt Disney, saying Disney's lack of interest in the deal would make it unsuccessful. Camcast also swung to a first-quarter profit and revived plans for a $1 billion stock buyback.

NETEL FIRED its CEO, chief financial officer and controller and said it will delay the release of its first-quarter results. The telecom-equipment maker also plans more restatements.
Optimized for Intermediate/Near

Distance
FIG. 70A

Intermediate
FIG. 70B

Near
FIG. 70C

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NOTEL FIRED its CEO, chief financial officer and controller and said it will delay the release of its first-quarter results. The telecom-gear maker also plans more restatements.
Optimized for Distance/Intermediate

Distance
FIG. 71A

Intermediate
FIG. 71B

Near
FIG. 71C
**INTERNATIONAL SEARCH REPORT**

**International application No**

PCT/US2015/067363

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**A. CLASSIFICATION OF SUBJECT MATTER**

INV. A61F9/008 G02C7/02 G02C7/04 G02C7/06

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**ADD.**

According to International Patent Classification (IPC) or to both national classification and IPC

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**B. FIELDS SEARCHED**

Minimum documentation searched (classification system followed by classification symbols)

A61F G02C

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**DOCUMENTATION SEARCHED**

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

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**EPO-Internal , WPI Data**

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**C. DOCUMENTS CONSIDERED TO BE RELEVANT**

<table>
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<tr>
<th>Category*</th>
<th>Citation of document, with indication, where appropriate, of the relevant passages</th>
<th>Relevant to claim No.</th>
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<td>Y paragraphs [0022], [0112] - [0114], [0204], [0208], [0250], [0253], [0266]</td>
<td>25-27, 39-41, 53-55</td>
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* Special categories of cited documents:

- **A** document defining the general state of the art which is not considered to be of particular relevance
- **E** earlier application or patent but published on or after the international filing date
- **L** document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)
- **O** document referring to an oral disclosure, use, exhibition or other means
- **P** document published prior to the international filing date but later than the priority date claimed

- **T** later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention
- **X** document of particular relevance: the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone
- **Y** document of particular relevance: the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art
- **A** document member of the same patent family

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Date of the actual completion of the international search: 22 April 2016

Date of mailing of the international search report: 02/05/2016

Name and mailing address of the ISA:

European Patent Office, P.B. 5818 Patentlaan 2
NL - 2280 HV Rijswijk
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Fax: (+31-70) 340-3016

Authorized officer: Buchler Costa, Joana

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Form PCT/ISA/210 (second sheet) (April 2005)
**INTERNATIONAL SEARCH REPORT**

### Box No. II  Observations where certain claims were found unsearchable (Continuation of item 2 of first sheet)

This international search report has not been established in respect of certain claims under Article 17(2)(a) for the following reasons:

1. **X** Claims Nos.: 1-13, 16, 56-62 because they relate to subject matter not required to be searched by this Authority, namely:
   
   A meaningful search is not possible on the basis of claims 1-13, 16 and 56-62 because these claims are directed to Rule 39.1(iv) PCT - Method for treatment of the human or animal body by therapy Rule 39.1(iv) PCT - Method for treatment of the human or animal body by surgery.

2. □ Claims Nos.: because they relate to parts of the international application that do not comply with the prescribed requirements to such an extent that no meaningful international search can be carried out, specifically:

3. □ Claims Nos.: because they are dependent claims and are not drafted in accordance with the second and third sentences of Rule 6.4(a).

### Box No. III Observations where unity of invention is lacking (Continuation of item 3 of first sheet)

This International Searching Authority found multiple inventions in this international application, as follows:

1. □ As all required additional search fees were timely paid by the applicant, this international search report covers all searchable claims.

2. □ As all searchable claims could be searched without effort justifying an additional fees, this Authority did not invite payment of additional fees.

3. □ As only some of the required additional search fees were timely paid by the applicant, this international search report covers only those claims for which fees were paid, specifically claims Nos.:

4. □ No required additional search fees were timely paid by the applicant. Consequently, this international search report is restricted to the invention first mentioned in the claims; it is covered by claims Nos.:

**Remark on Protest**

- □ The additional search fees were accompanied by the applicant's protest and, where applicable, the payment of a protest fee.

- □ The additional search fees were accompanied by the applicant's protest but the applicable protest fee was not paid within the time limit specified in the invitation.

- □ No protest accompanied the payment of additional search fees.
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