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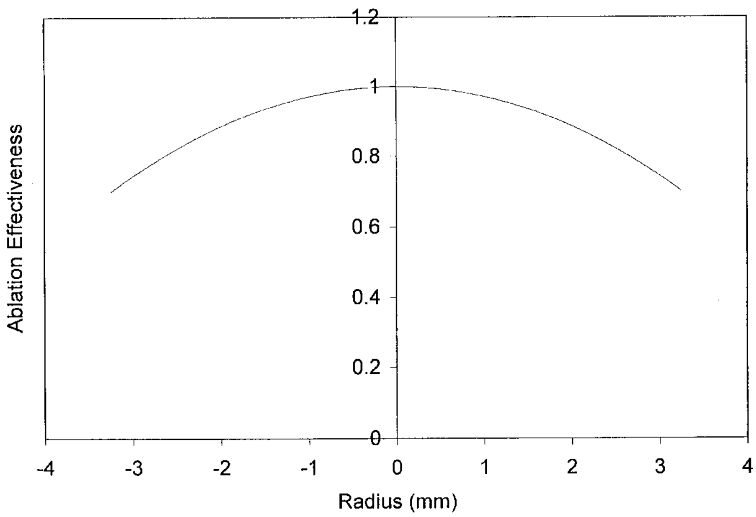

(57) Abstract: An optical correction system for correcting visual defects of an eye includes a wavefront analyzer responsive to a wavefront emanating from an eye for determining an optical path difference between a reference wave and the wavefront. A converter provides an optical correction based on the path difference and on a radially dependent ablation efficiency. The efficiency correction uses a compensating polynomial of the form $A + B\rho + C\rho^2 + D\rho^3 + ... + X\rho^n$, where ρ is a normalized radius measured from a central portion of the cornea, reaching a value of 1 at an outer edge of the optical correction zone. A laser beam is directed to the cornea that has power sufficient for ablating corneal material. The optical correction is achieved by the removal of a selected amount of the corneal material to cerate a desired corneal shape change based on the optical correction.

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OPTIMIZATION OF ABLATION CORRECTION OF AN OPTICAL SYSTEM AND ASSOCIATED METHODS

BACKGROUND OF THE INVENTION

Field of the Invention

The present invention relates to optical aberration measurement and correction, and, more particularly, to a system and method for achieving an empirical optimization of an objective measurement and correction of an optical system such as the human eye.

Description of Related Art

Optical systems having a real image focus can receive collimated light and focus it at a point. Such optical systems can be found in nature, e.g., human and animal eyes, or can be manmade, e.g., laboratory systems, guidance systems, and the like. In either case, aberrations in the optical system can affect the system's performance.

A perfect or ideal human eye diffusely reflects an impinging light beam from its retina through optics of the eye, which includes a lens and a cornea. For such an ideal eye in a relaxed state, i.e., not accommodating to provide near-field focus, reflected light exits the eye as a sequence of plane waves. However, a real eye typically has aberrations that cause deformation or distortion of reflected light waves exiting the eye. An aberrated eye diffusely reflects an impinging light beam from its retina through its lens and cornea as a sequence of distorted wavefronts.

It is known in the art to perform laser correction of focusing deficiencies by photorefractive keratectomy (PRK), which modifies corneal curvature, and LASIK surgery. Such methods typically employ a 193-nm excimer laser to ablate corneal tissue. Munnerlyn et al. (*J. Cataract Refract. Surg.* **14**(1), 46-52, 1988) have presented equations for determining a specific volume of tissue to be removed to achieve a desired refractive correction. Frey (U.S. Pat. No. 5,849,006) teaches a method of using a small-spot laser to remove a desired volume of tissue for effecting a desired refractive correction.

SUMMARY OF THE INVENTION

It is an object of the present invention to provide a system and method for optimizing an ablative correction to a human cornea.

It is a further object to provide such a system and method that accounts for corneal anisotropy.

It is another object to provide such a system and method that includes a radially dependent attenuation of the ablation power.

It is an additional object to provide such a system and method utilizing a mathematical description that can readily be adapted into an ablation algorithm.

These and other objects are achieved by the present invention, an optical correction system for correcting visual defects of an eye. The system comprises a wavefront analyzer responsive to a wavefront emanating from an eye for determining an optical path difference between a reference wave and the wavefront. The system further comprises a converter for providing an optical correction based on the path difference and on a radially dependent ablation efficiency. The efficiency correction uses a compensating polynomial of the form $A + B\rho + C\rho^2 + D\rho^3 + \cdots + X\rho^n$, where ρ is a normalized radius that is optical zone specific and is measured from a central portion of the cornea, reaching a value of 1 at the edge of the optical correction zone.

A laser beam is directed to the cornea that has power sufficient for ablating corneal material. The optical correction is achieved by the removal of a selected amount of the corneal material to create a desired corneal shape change based on the optical correction.

The features that characterize the invention, both as to organization and method of operation, together with further objects and advantages thereof, will be better understood from the following description used in conjunction with the accompanying drawing. It is to be expressly understood that the drawing is for the purpose of illustration and description and is not intended as a definition of the limits of the invention. These and other objects attained, and advantages offered, by the present invention will become more fully apparent as the description that now follows is read in conjunction with the accompanying drawing.

BRIEF DESCRIPTION OF THE DRAWINGS

FIG. 1 is a schematic diagram of a system for determining ocular aberrations.

FIG. 2 is a graph of desired and achieved ablation depths as a function of radial position for a myopic eye.

FIG. 3 is a graph of desired and achieved ablation depths as a function of radial position for a hyperopic eye.

FIGS. 4A and **4B** are graphs of the ablation efficiency function of the present invention: FIG. 4A plots $1 - 0.3r^2$, where $r_{max} = 3.25$ mm; FIG. 4B plots $0.95 - 0.3r^2 - 0.25r^3 + 0.3r^4$.

FIG. 5 is a schematic diagram of a system for delivering an ablative laser beam to an eye.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENTS

A description of the preferred embodiments of the present invention will now be presented with reference to FIGS. 1-5.

The system and method for correcting visual defects of an eye includes a wavefront analyzer, in a preferred embodiment a system **10** (FIG. 1) similar to that described in copending and co-owned application Serial Number 09/664,128, the contents of which are incorporated herein by reference. The apparatus **10** includes a laser **12** for generating optical radiation used to produce a small-diameter laser beam **14**. The laser **12** generates a collimated laser light beam (represented by dashed lines for the beam **14**) of a wavelength and power that is eye-safe. For ophthalmic applications, appropriate wavelengths would include the entire visible spectrum and the near-infrared spectrum. By way of example, appropriate wavelengths may be in a range of from approximately 400–1000 nms, including 550-, 650-, and 850-nm useful wavelengths. While operation in the visible spectrum is generally desired, since these are the conditions in which the eye operates, the near-infrared spectrum may offer advantages in certain applications. For example, the patient's eye may be more relaxed if the patient does not know measurement is taking place. Regardless of the wavelength of the optical radiation, power should be restricted in ophthalmic applications to eye-safe levels. For laser radiation, appropriate eye-safe exposure levels can be found in the U.S. Federal Performance Standard for Laser Products. If the analysis is to be performed on an optical system other than the eye, the examination wavelength range logically should incorporate the intended performance range of the system.

To select a small-diameter collimated core of laser light beam **14**, an iris diaphragm **16** is used to block all of laser light beam **14** except for the laser beam **18** of a size desired for use. In terms of the present invention, the laser beam **18** will have a diameter in the range of approximately 0.5–4.5 mm, with 1–3 mm being typical, by way of example. A badly aberrated eye uses a smaller-diameter beam, while an eye with only slight aberrations can be evaluated with a larger-diameter beam. Depending on the output divergence of the laser **12**, a lens can be positioned in the beam path to optimize collimating of the beam.

Laser beam **18**, as herein described by way of example, is a polarized beam that is passed through a polarization-sensitive beam splitter **20** for routing to a focusing optical train **22**, which operates to focus the laser beam **18** through the optics of the eye **120** (e.g., the cornea **126**, pupil **125**, and the lens **124**) to the retina **122**. It is to be understood that the lens **124** may not be present for a patient that has undergone a cataract procedure. However, this does not affect the present invention.

The optical train **22** images the laser beam **18** as a small spot of light at or near the eye's *fovea centralis* **123**, where the eye's vision is most acute. Note that the small spot of light could be reflected off another portion of retina **122** in order to determine aberrations related to another aspect of one's vision. For example, if the spot of light were reflected off the area of the retina **122** surrounding the *fovea centralis* **123**, aberrations specifically related to one's peripheral vision could then be evaluated. In all cases, the spot of light may be sized to form a near-diffraction-limited image on the retina **122**. Thus the spot of light produced by laser beam **18** at *fovea centralis* **123** does not exceed approximately 100 μm in diameter and, typically, is on the order of 10 μm.

The diffuse reflection of the laser beam **18** back from the retina **122** is represented by solid lines **24** indicative of radiation that passes back through the eye

120. The wavefront **24** impinges on and is passed through the optical train **22** and on to the polarization-sensitive beam splitter **20**. The wavefront **24** is depolarized relative to the laser beam **18** due to reflection and refraction as the wavefront **24** emanates from the retina **122**. Accordingly, the wavefront **24** is turned at the polarization-sensitive beam splitter **20** and directed to a wavefront analyzer **26** such as a Hartmann-Shack (H-S) wavefront analyzer. In general, the wavefront analyzer **26** measures the slopes of wavefront **24**, i.e., the partial derivatives with respect to x and y, at a number of (x,y) transverse coordinates. This partial derivative information is then used to reconstruct or approximate the original wavefront with a mathematical expression such as a weighted series of Zernike polynomials.

The polarization states for the incident laser beam **18** and the beam splitter **20** minimizes the amount of stray laser radiation reaching the sensor portion of the wavefront analyzer **26**. In some situations, stray radiation may be sufficiently small when compared to the radiation returning from the desired target (e.g., the retina **122**) so that the polarization specifications are unnecessary.

The present invention is able to adapt to a wide range of vision defects and as such achieves a new level of dynamic range in terms of measuring ocular aberrations. Dynamic range enhancement is accomplished with the optical train **22** and/or a wavefront sensor portion of the wavefront analyzer **26**. The optical train **22** includes a first lens **220**, a flat mirror **221**, a Porro mirror **222**, and a second lens **224**, all of which lie along the path of laser beam **18** and the wavefront **24**. The first lens **220** and the second lens **224** are identical lenses maintained in fixed positions. The Porro mirror **222** is capable of linear movement, as indicated by arrow **223** to change the optical path length between the lenses **220** and **224**. However, it is to be understood that the present invention is not limited to the particular arrangement of the flat mirror **221** and the Porro mirror **222** and that other optical arrangements may be used without departing from the teachings and benefits of the present invention.

A "zero position" of the Porro mirror **222** is identified by replacing the eye **120** by a calibration source of collimated light to provide a reference wavefront such as a perfect plane wave **110**. Such a source could be realized by a laser beam expanded by a beam telescope to the diameter that will cover the imaging plane of wavefront analyzer **26** and adjustment of the Porro mirror **222** until the wavefront analyzer **26** detects the light as being collimated. Note that the changes in optical path length brought about by the Porro mirror **222** can be calibrated in diopters to provide an approximate spherical dioptric correction.

In order to empirically determine a treatment efficiency of a particular beam profile in effecting a desired change in refraction, data were collected on the ablation of human corneas *in vivo* with known ablation profiles and known laser beam fluence profiles. The precision and lack of subjectivity of the above-discussed wavefront measurement was used to determine the optical results and hence the effective treatment efficiency of particular ablation profiles. Any deviations from the expected change in aberration content can be attributed to relative differences in ablation effectiveness across the corneal surface.

A single generalized ablation effectiveness function was derived from clinical data using both myopic and hyperopic nominal ablation profiles. The data were collected from nominal ablation profiles obtained using an excimer laser narrow-beam scanning spot such as that disclosed in U.S. Patent Nos. 5,849,006 and 5,632,742, the contents of which are incorporated by reference herein.

The radially symmetric attenuation function of the present invention was determined by analysis of graphs of intended and achieved ablation depth versus normalized radial corneal position for myopic (FIG. 2) and hyperopic (FIG. 3) eyes. In its general form the ablation effectiveness function has the polynomial form $A + B\rho + C\rho^2 + D\rho^3 + \dots + X\rho^n$, as described above. In a specific embodiment the function has the form $A + B\rho + C\rho^2 + D\rho^3 + E\rho^4$, with exemplary coefficients $A \cong 0.95$, $B \cong 0$, $C \cong -0.3$, $D = -0.25$, and $E = 0.3$ for an optical zone radius of 3.25 mm. The ablation effectiveness function includes any radial dependence in the actual ablation rate, that is, for example, micrometers of tissue removed per pulse. However, it also incorporates any biomechanical effect or intrinsic variation in corneal optical properties that can influence the optical outcome in a radially dependent manner.

The attenuation or efficiency function is then used to modify the treatment profile by taking the desired change in corneal depth (the nominal ablation profile) and dividing this by the attenuation function. This yields a new profile that, when ablated, results in the desired change.

In a particular embodiment the attenuation is achieved by computing the Zernike description of the ablation profile and dividing the Zernike polynomial by the attenuation profile that is entered into the laser beam delivery system:

$$P_{input}(\rho,\theta) = P_{desired}(\rho,\theta)/(A + B\rho + C\rho^2 + D\rho^3 + \dots + X\rho^n)$$

In a graph of a simple form of this function, $1 - 0.3r^2$, where $r_{max} = 3.25$ mm (FIG. 4A), the radially dependent ablation efficiency varies from a value of approximately 1 proximate a central location wherein $r \cong 0$ on the corneal surface to a value of approximately 0.7 at a distance from the central location wherein $r \cong 3.25$ mm.

A more detailed version of the attenuation function, $0.95 - 0.3r^2 - 0.25r^3 + 0.3r^4$, which has a more complex shape, is shown in FIG. 4B. The specific function applied for a particular treatment laser system may depend on specifics of that device, such as beam energy, etc. Therefore, the coefficients in the attenuation function polynomial can be adjusted to optimize results for particular treatment conditions.

Preferably the optical correction is further based on refractive indices of media through which the wavefront passes. In a particular embodiment, the converter provides the path difference using a Zernike reconstruction of the wavefront, and the path difference is divided by a difference between an index of refraction of corneal material and an index of refraction of air. The optical correction is a prescribed alteration of corneal surface curvature of the eye, and the optical correction achieved by the reshaping of the corneal surface curvature of the eye is based on the prescribed alteration without regard to a resulting topography of the overall surface of the cornea.

An exemplary laser beam delivery system **5** (FIG. 5) laser beam delivery and eye tracking system may comprise, for example, that taught in U.S. Pat. No. 5,980,513, co-owned with the present application, the contents of which are incorporated herein by reference. The laser beam delivery portion of system **5** includes treatment laser source **500**, projection optics **510**, *X*-*Y* translation mirror optics **520**, beam translation controller **530**, dichroic beamsplitter **200**, and beam angle adjustment mirror optics **300**. The laser pulses are distributed as shots over the area to be ablated or eroded, preferably in a distributed sequence so that the desired shape of the object or cornea is achieved. Preferably the pulsed laser beam is shifted to direct the shots to a plurality of spatially displaced positions on the corneal surface to form a plurality of spatially distributed ablation spots. Each of these spots may have a predetermined diameter, for example, 2.5 or 1.0 mm, and may have an intensity distribution, for example, defined by a Gaussian or a generally flat distribution profile across the spot.

In operation of the beam delivery portion of system **5**, laser source **500** produces laser beam **502** incident upon projection optics **510**. Projection optics **510** adjusts the diameter and distance to focus of beam **502** depending on the requirements of the particular procedure being performed.

After exiting projection optics **510**, beam **502** impinges on *X*-*Y* translation mirror optics **520**, where beam **502** is translated or shifted independently along each of two orthogonal translation axes as governed by beam translation controller **530**. Controller **530** is typically a processor programmed with a predetermined set of two-dimensional translations or shifts of beam **502** depending on the particular ophthalmic procedure being performed. Each of the *X* and *Y* axes of translation is independently controlled by a translating mirror.

The eye tracking portion of system **5** includes eye movement sensor **100**, dichroic beamsplitter **200**, and beam angle adjustment mirror optics **300**. Sensor **100** determines the amount of eye movement and uses that amount to adjust mirrors **310** and **320** to track along with the eye movement. To do this, sensor **100** first transmits light energy **101-T**, which has been selected to transmit through dichroic beamsplitter **200**. At the same time, after undergoing beam translation in accordance with the particular treatment procedure, beam **502** impinges on dichroic beamsplitter **200**, which has been selected to reflect beam **502** (e.g., a 193-nm wavelength laser beam) to beam angle adjustment mirror optics **300**.

Light energy **101-T** is aligned such that it is parallel to beam **502** as it impinges on beam angle adjustment mirror optics **300**. It is to be understood that the term "parallel" as used herein includes the possibility that light energy **101-T** and beam **502** can be coincident or collinear. Both light energy **101-T** and beam **502** are adjusted in correspondence with one another by optics **300**. Accordingly, light energy **101-T** and beam **502** retain their parallel relationship when they are incident on eye **120**. Since *X-Y* translation mirror optics **520** shifts the position of beam **502** in translation independently of optics **300**, the parallel relationship between beam **502** and light energy **101-T** is maintained throughout the particular ophthalmic procedure.

The beam angle adjustment mirror optics consists of independently rotating mirrors **310** and **320**. Mirror **310** is rotatable about axis **312**, as indicated by arrow **314**, while mirror **320** is rotatable about axis **322**, as indicated by arrow **324**. Axes **312** and **322** are orthogonal to one another. In this way, mirror **310** is capable of sweeping light energy **101-T** and beam **502** in a first plane (e.g., elevation), while mirror **320** is capable of independently sweeping light energy **101-T** and beam **502** in a second plane (e.g., azimuth) that is perpendicular to the first plane. Upon exiting beam angle adjustment mirror optics **300**, light energy **101-T** and beam **502** impinge on eye **120**.

The movement of mirrors **310** and **320** is typically accomplished with servo controller/motor drivers **316** and **326**, respectively. In general, drivers **316** and **326** must be able to react quickly when the measured error from eye movement sensor **100** is large, and further must provide very high gain from low frequencies (DC) to about 100 radians per second to virtually eliminate both steady-state and transient error.

More specifically, eye movement sensor **100** provides a measure of the error between the center of the pupil (or an offset from the center of the pupil that the doctor selected) and the location where mirror **310** is pointed.

Light energy **101-R** reflected from eye **120** travels back through optics **300** and beamsplitter **200** for detection at sensor **100**. Sensor **100** determines the amount of eye movement based on the changes in reflection energy **101-R**. Error control signals indicative of the amount of eye movement are fed back by sensor **100** to beam angle adjustment mirror optics **300**. The error control signals govern the movement or realignment of mirrors **310** and **320** in an effort to drive the error control signals to zero. In doing this, light energy **101-T** and beam **502** are moved in correspondence with eye movement while the actual position of beam **502** relative to the center of the pupil is controlled by *X-Y* translation mirror optics **520**.

In order to take advantage of the properties of beamsplitter **200**, light energy **101-T** must be of a different wavelength than that of treatment laser beam **502**. The light energy should preferably lie outside the visible spectrum so as not to interfere or obstruct a surgeon's view of eye **120**. Further, if the present invention is to be used in ophthalmic surgical procedures, light energy **101-T** must be "eye safe," as defined by the American National Standards Institute (ANSI). While a variety of light wavelengths satisfy the above requirements, by way of example, light energy **101-T** may comprise infrared light energy in the 900-nm wavelength region. Light in this region meets the above-noted criteria and is further produced by readily available, economically affordable light sources. One such light source is a high pulse repetition rate GaAs 905-nm laser operating at 4 kHz, which produces an ANSI-defined eye-safe pulse of 10 nJ in a 50-ns pulse. A corneal ablation system using 193-nm ablation in a range of fluences of 100–1000 mJ/cm^2, which uses a small spot (< 2.5 mm) may also be used. One preferred embodiment utilizes a spot < 1.0 mm and 400–600 mJ/cm^2 peak fluences.

Thus it can be seen that the present invention provides a system and method for providing a compensating correction function adapted to negate or cancel out the ablation efficiency function to permit the actual desired shape of the corneal removal volume to be obtained, effecting an ideal optical result.

In the foregoing description, certain terms have been used for brevity, clarity, and understanding, but no unnecessary limitations are to be implied therefrom beyond the requirements of the prior art, because such words are used for description purposes herein and are intended to be broadly construed. Moreover, the embodiments of the apparatus illustrated and described herein are by way of example, and the scope of the invention is not limited to the exact details of construction.

Having now described the invention, the construction, the operation and use of preferred embodiment thereof, and the advantageous new and useful results obtained thereby, the new and useful constructions, and reasonable mechanical equivalents thereof obvious to those skilled in the art, are set forth in the appended claims.

THAT WHICH IS CLAIMED IS:

1. An optical correction system for correcting visual defects of an eye, the optical correction system comprising:

a wavefront analyzer responsive to a wavefront emanating from an eye for determining an optical path difference between a reference wave and the wavefront;

a converter for providing an optical correction based on the path difference and on a radially dependent ablation efficiency using a compensating polynomial of the form $A + B\rho + C\rho^2 + D\rho^3 + \cdots + X\rho^n$, where ρ is a normalized radius measured from a central portion of the cornea, reaching a value of 1 at an outer edge of the optical correction zone; and

a laser beam having power sufficient for ablating corneal material, wherein the optical correction is achieved by the removal of a selected amount of the corneal material to create a desired corneal shape change.

2. The system recited in Claim 1, wherein the optical correction means comprise a wavefront analyzer responsive to a wavefront emanating from the eye, and wherein the corneal modification is determined by an optical path difference between a reference wave and the wavefront.

3. The system recited in Claim 2, further comprising:

an energy source for generating a beam of optical radiation; and

focusing optics disposed in the path of the beam for directing the beam through the eye, wherein the beam is reflected back from the retina of the eye as the wavefront of radiation emanating from the eye.

4. The system recited in Claim 1, wherein the polynomial has the form $A + B\rho + C\rho^2 + D\rho^3 + E\rho^4$, with coefficients $A \cong 0.95$, $B \cong 0$, $C \cong -0.3$, $D \cong -0.25$, and $E \cong 0.3$ for an optical zone radius of approximately 3.25 mm.

5. The system recited in Claim 1, wherein the radially dependent ablation efficiency varies from a value of approximately 1.0 proximate a central location wherein $r \cong 0$ on the corneal surface to a value of approximately 0.7 at an outer edge of the optical zone having a radius wherein $r \cong 3.25$ mm.

6. The system recited in Claim 1, wherein the optical correction is further based on refractive indices of media through which the wavefront passes.

7. The system recited in Claim 1, further comprising an eye tracker for monitoring motion of the eye and for adjusting the positions of the laser beam responsive to the motion.

8. The system recited in Claim 1, wherein the optical correction is a prescribed alteration of corneal surface curvature of the eye, and wherein the optical correction achieved by the reshaping of the corneal surface curvature of the eye is based on the prescribed alteration without regard to a resulting topography of the overall surface of the cornea.

9. The system recited in Claim 1, wherein the converter provides the path difference using a Zernike reconstruction of the wavefront, and wherein the path difference is divided by a difference between an index of refraction of corneal material and an index of refraction of air.

10. A system for modifying vision of an eye, the system comprising:

a wavefront analyzer responsive to a wavefront emanating from the eye for determining an optical path difference between a reference wave and the wavefront;

a converter for providing an optical correction based on the optical path difference and a ablation efficiency using a compensating polynomial of the form: $A + B\rho + C\rho^2 + D\rho^3 + \dots + X\rho^n$, where ρ is a normalized radius measured from a central portion of the cornea, reaching a value of 1 at an outer edge of the optical correction zone;

a treatment laser producing a pulsed laser beam for providing a plurality of laser beam shots capable of ablating corneal material; and

beam-shifting means operable with the treatment laser for shifting the pulsed laser beam and for directing the plurality of laser beam shots to a plurality of spatially displaced positions on the corneal surface of the eye as a plurality of spatially distributed ablation spots for providing a desired modification to the cornea thus modifying vision of the eye.

11. The system recited in Claim 10, wherein the beam-shifting means provides a single predetermined shot pattern responsive to the optical correction.

12. The system recited in Claim 10, wherein each of the plurality of ablation spots formed on the surface of the cornea may be defined by a diameter length of approximately 2.5 mm.

13. The system recited in Claim 10, wherein each of the plurality of ablation spots formed on the surface of the cornea includes a length dimension of approximately 1.0 mm.

14. The system recited in Claim 10, wherein each of the plurality of ablation spots formed on the surface of the cornea comprises an intensity profile across the spot defined by an approximately Gaussian distribution.

15. The system recited in Claim 10, wherein each of the plurality of ablation spots formed on the surface of the cornea comprises a generally flat intensity profile across the spot.

16. The system recited in Claim 10, wherein the optical correction is further based on refractive indices of media through which the wavefront passes.

17. The system recited in Claim 10, wherein the polynomial has the form $A + B\rho + C\rho^2 + D\rho^3 + E\rho^4$, with coefficients $A \cong 0.95$, $B \cong 0$, $C \cong -0.3$, $D \cong -0.25$, and $E \cong 0.3$ for an optical zone radius of approximately 3.25 mm.

18. The system recited in Claim 10, wherein the radially dependent ablation efficiency varies from a value of approximately 1.0 proximate a central location wherein $\rho \cong 0$ on the corneal surface to a value of approximately 0.7 at an outer edge of an optical zone having a radius of approximately 3.25 mm.

19. The system recited in Claim 10, further comprising:

an energy source for generating a beam of optical radiation; and

focusing optics disposed in the path of the beam for directing the beam through the eye, wherein the beam is reflected back from the retina of the eye as the wavefront of radiation emanating from the eye.

20. The system recited in Claim 10, further comprising an eye tracker for monitoring motion of the eye and for adjusting the positions of the laser beam responsive to the motion.

21. The system recited in Claim 10, wherein the optical correction comprises a prescribed alteration of corneal surface curvature of the eye, and wherein the optical correction achieved by the reshaping of the corneal surface curvature of the eye is based on the prescribed alteration without regard to a resulting topography of the overall surface of the cornea.

22. The system recited in Claim 10, wherein the converter provides the path difference using a Zernike reconstruction of the wavefront, and wherein the path difference is divided by a difference between an index of refraction of corneal material and an index of refraction of air.

23. An optical correction system for correcting visual defects of an eye, the optical correction system comprising a converter for providing an optical correction based on an optical path difference between a reference wave and a wavefront emanating from an eye and on a radially dependent ablation efficiency using a compensating polynomial of the form $A + B\rho + C\rho^2 + D\rho^3 + \cdots + X\rho^n$, where ρ is a normalized radius measured from a central portion of the cornea, reaching a value of 1 at an outer edge of the optical correction zone, the optical correction useful for determining an optimal amount of corneal material to remove to create a desired corneal shape change.

24. A method of treating a cornea of an eye to effect a refractive correction of the eye, the method comprising the steps of:

determining a corneal modification from a measurement of the eye for providing a desired vision;

providing an optical correction for the eye based on the corneal modification and on a ablation efficiency using a compensating radially invariant polynomial of the form $A + B\rho + C\rho^2 + D\rho^3 + \cdots + X\rho^n$, where ρ is a normalized radius measured from a central portion of the cornea, reaching a value of 1 at an outer edge of the optical correction zone;

directing a laser beam onto the eye for ablating the cornea; and

moving the laser beam in a pattern about the eye, the pattern based on the optical correction.

25. The method recited in Claim 24, further comprising the step of redirecting the laser beam to compensate for eye movement.

26. The method recited in Claim 24, wherein the polynomial has the form $A + B\rho + C\rho^2 + D\rho^3 + E\rho^4$, with coefficients $A \cong 0.95$, $B \cong 0$, $C \cong -0.3$, $D \cong -0.25$, and $E \cong 0.3$ for an optical zone radius of approximately 3.25 mm.

27. The method recited in Claim 24, further comprising the steps of:

selecting an area on the cornea; and

providing a plurality of laser beam spots on the selected area of cornea for ablation thereof, wherein a size of each of the spots is substantially smaller that the selected area, and wherein the spots are in a pattern having a spacing therebetween.

28. The method recited in Claim 27, further comprising the step of forming each of the plurality of ablation spots on the surface of the cornea to be defined by a diameter of approximately 2.5 mm.

29. The method recited in Claim 27, further comprising the step of forming each of the plurality of ablation spots on the surface of the cornea to be defined by a diameter of approximately 1.0 mm.

30. The method recited in Claim 27, further comprising the step of forming each of the plurality of ablation spots on the surface of the cornea to have an intensity distribution defined by a Gaussian profile across the spot.

31. The method recited in Claim 27, further comprising the step of forming each of the plurality of ablation spots on the surface of the cornea to have an intensity distribution defined by a generally flat profile across the spot.

32. The method recited in Claim 24, further comprising the step of providing the optical correction based on refractive indices of media through which the wavefront passes.

33. The method recited in Claim 24, wherein the radially dependent ablation efficiency varies from a value of approximately 1.0 proximate a central location wherein $r \cong 0$ on the corneal surface to a value of approximately 0.7 at an outer edge of an optical zone having a radius of approximately 3.25.

34. The method recited in Claim 24, wherein the optical correction is a prescribed alteration of corneal surface curvature of the eye, and wherein the optical correction achieved by the reshaping of the corneal surface curvature of the eye is based on the prescribed alteration without regard to a resulting topography of the overall surface of the cornea.

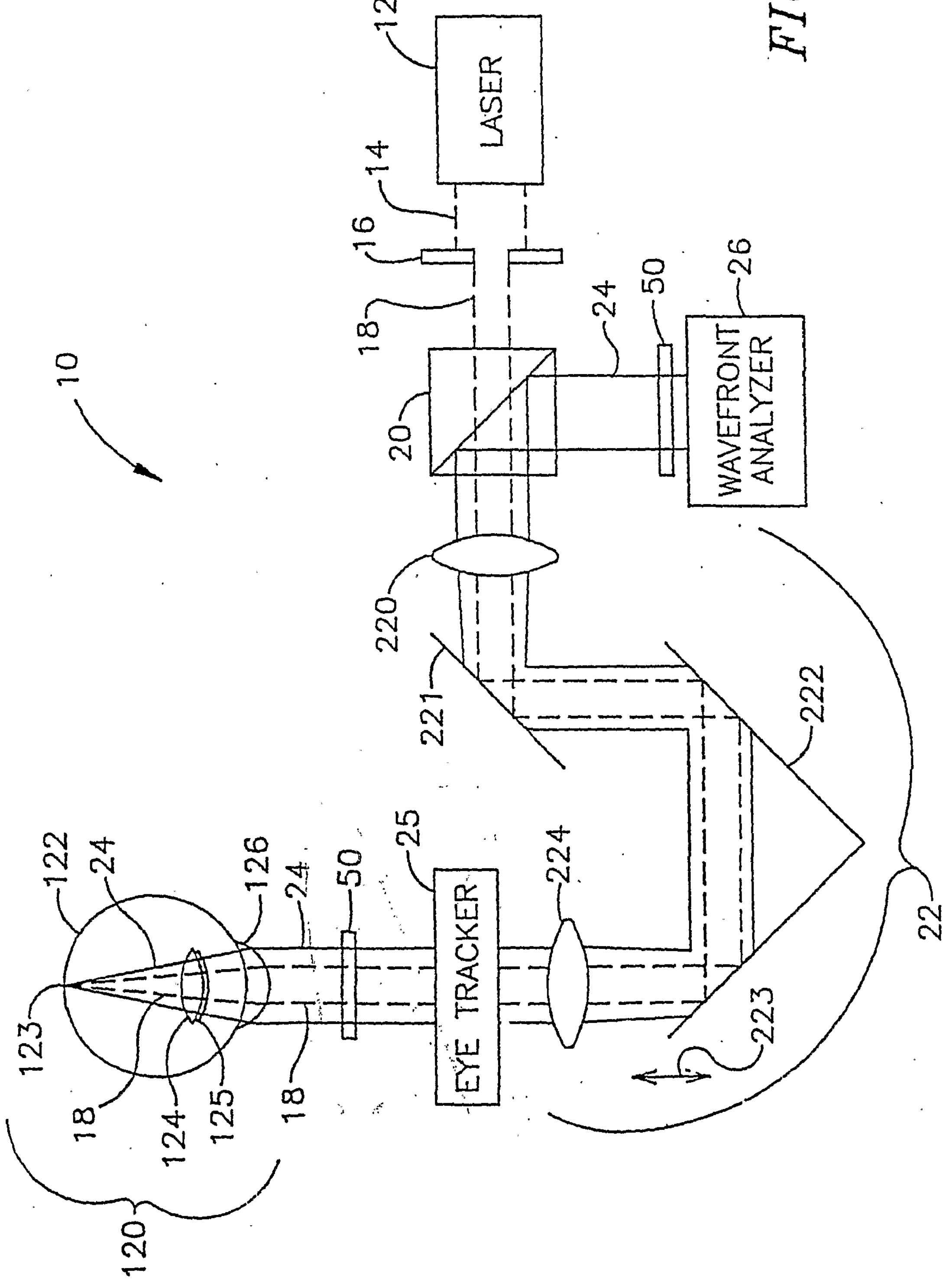
LASER
WAVEFRONT ANALYZER
EYE TRACKER
10
12
14
16
18
20
22
24
25
26
50
120
122
123
124
125
126
220
221
222
223
224

FIG. 1

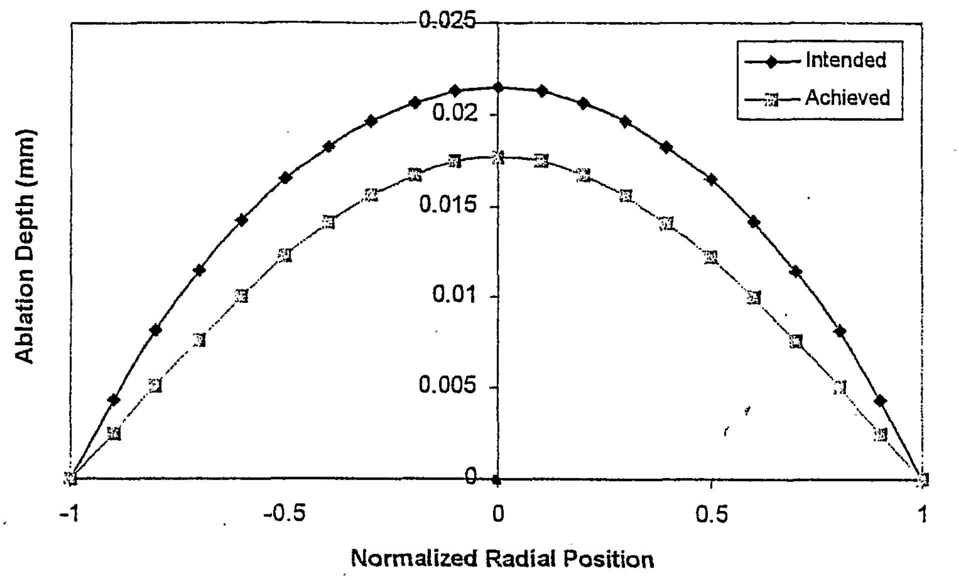
Intended
Achieved
0.025
0.02
0.015
0.01
0.005
0
-1
-0.5
0
0.5
1
Ablation Depth (mm)
Normalized Radial Position

FIG. 2

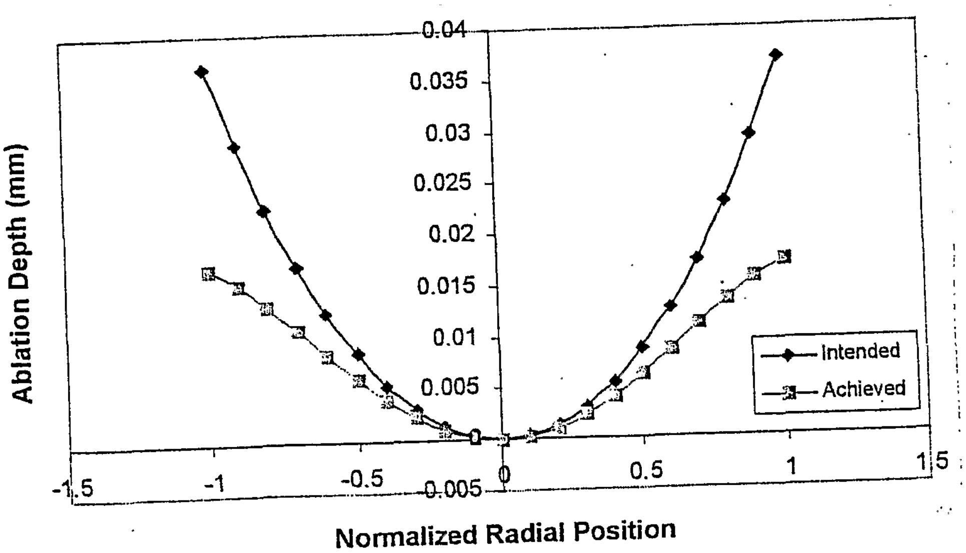
0.04
0.035
0.03
0.025
0.02
0.015
0.01
0.005
-0.005
-1.5
-1
-0.5
0
0.5
1
1.5
Intended
Achieved
Ablation Depth (mm)
Normalized Radial Position

FIG. 3

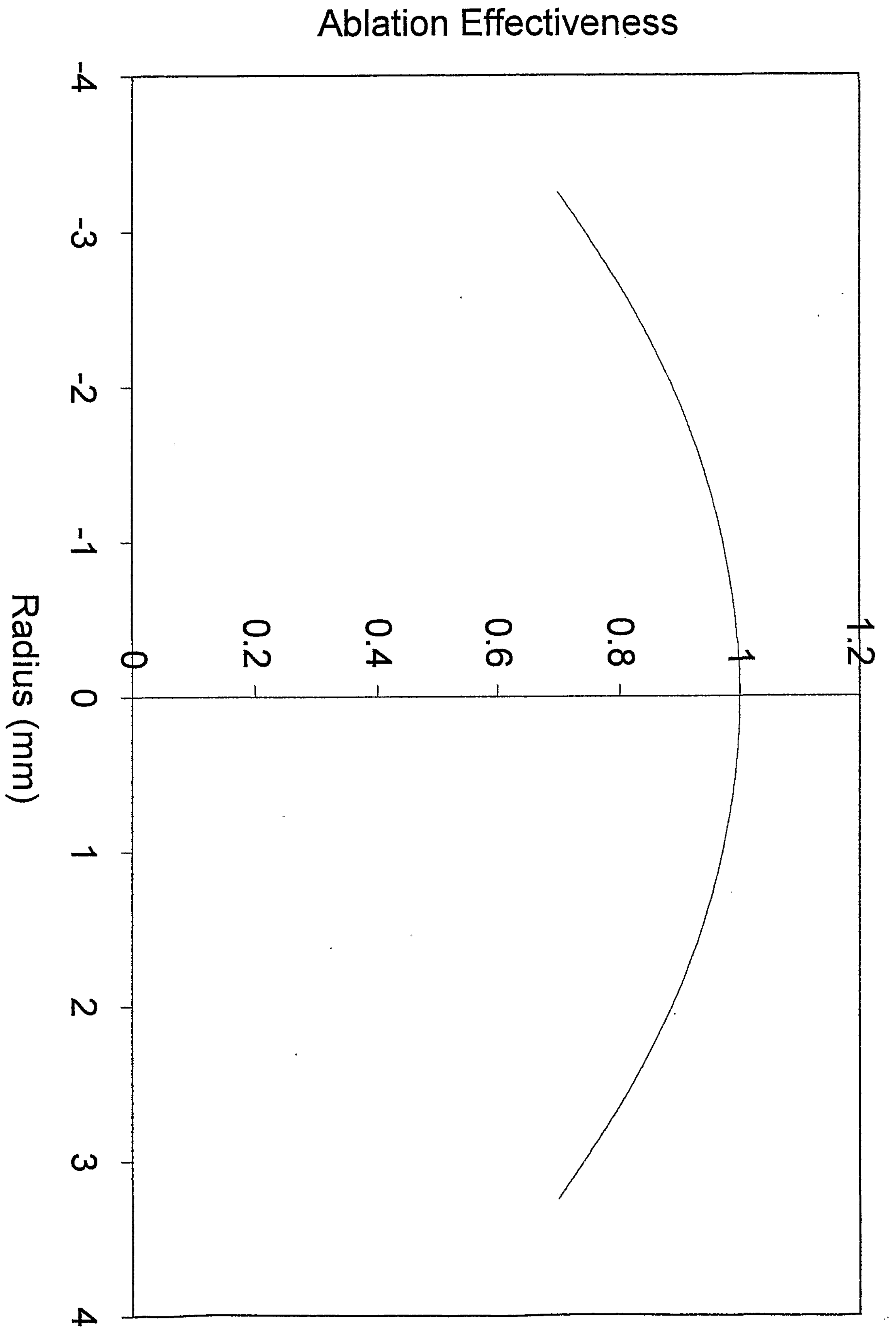
Ablation Effectiveness
1.2
1
0.8
0.6
0.4
0.2
0
-4
-3
-2
-1
0
1
2
3
4
Radius (mm)

FIG. 4A

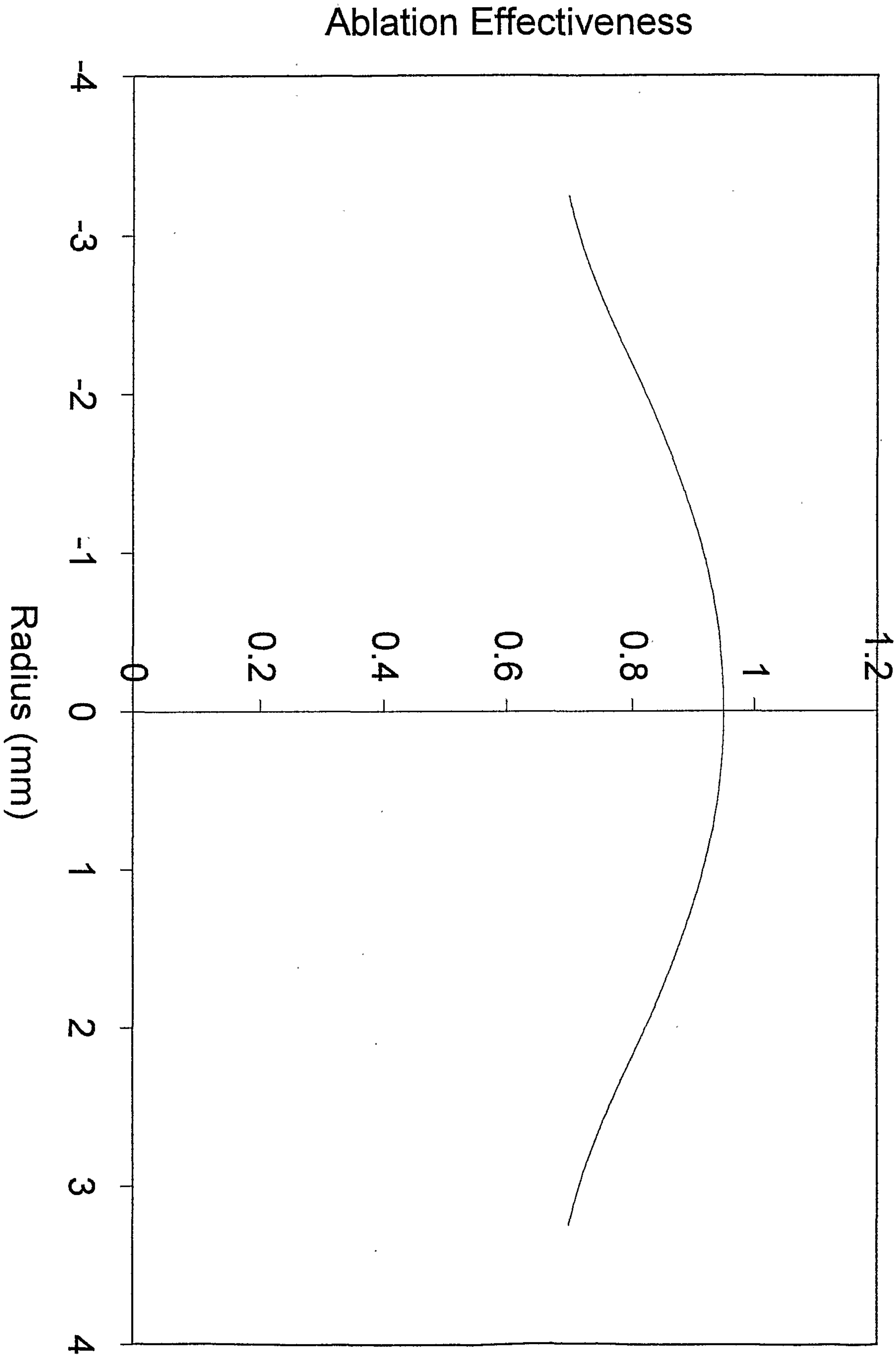
Ablation Effectiveness
-4
-3
-2
-1
0
1
2
3
4
Radius (mm)
0
0.2
0.4
0.6
0.8
1
1.2

FIG. 4B

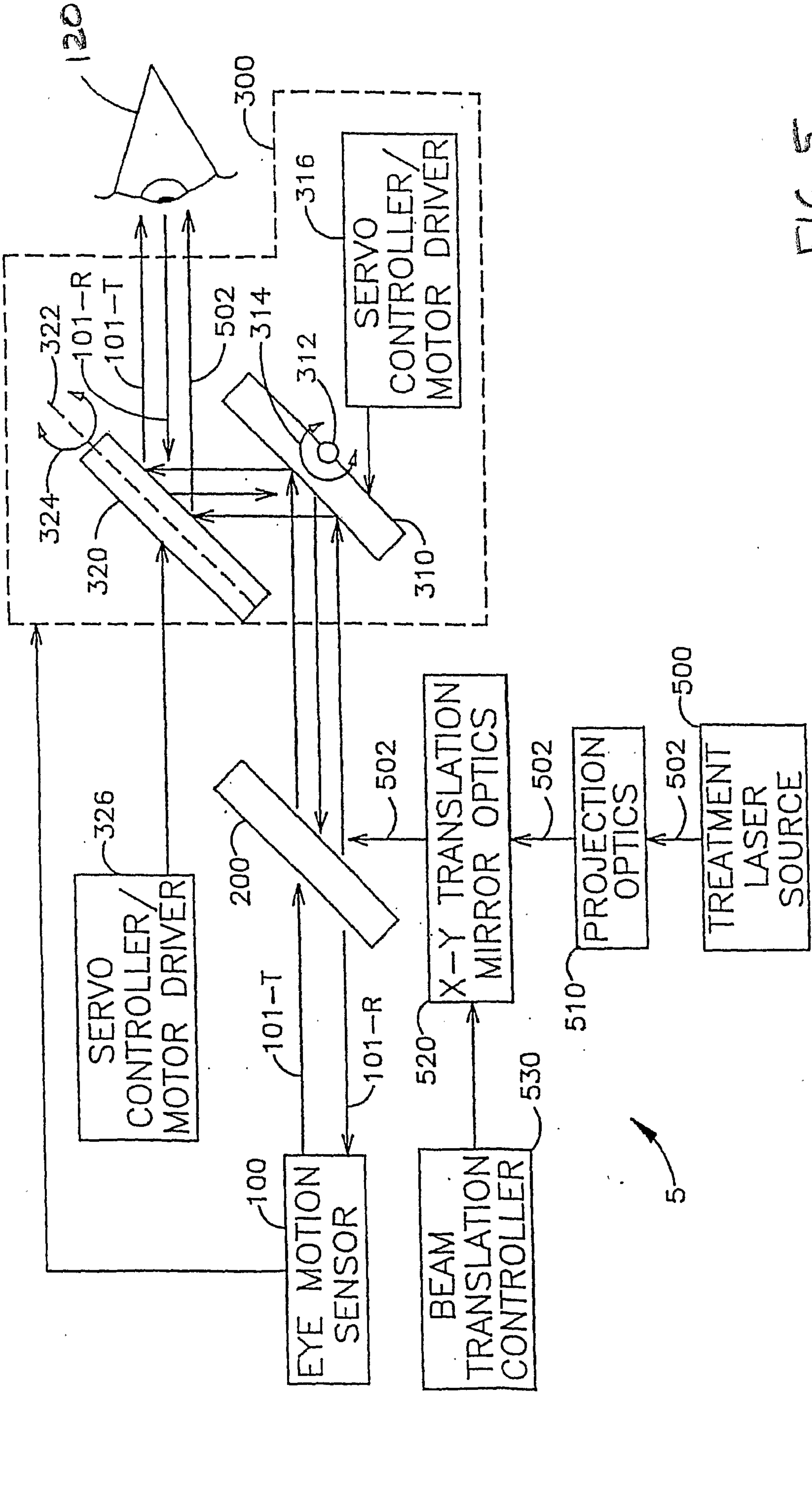
120
300
316
SERVO CONTROLLER/ MOTOR DRIVER
322
101–R
101–T
502
314
312
324
320
310
326
SERVO CONTROLLER/ MOTOR DRIVER
200
502
X–Y TRANSLATION MIRROR OPTICS
502
PROJECTION OPTICS
502
TREATMENT LASER SOURCE
500
101–T
101–R
100
EYE MOTION SENSOR
520
530
BEAM TRANSLATION CONTROLLER
510
5

FIG. 5

INTERNATIONAL SEARCH REPORT

International Application No
PCT/US 01/40352

A. CLASSIFICATION OF SUBJECT MATTER
IPC 7 A61F9/01

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

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)
IPC 7 A61F A61B

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

Electronic data base consulted during the international search (name of data base and, where practical, search terms used)

WPI Data, PAJ, EPO-Internal, INSPEC, COMPENDEX

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category °	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
Y	SCHWIEGERLING JIM ET AL.: "Using corneal height maps and polynomial decomposition to determine corneal aberrations" OPTOMETRY AND VISION SCIENCE, vol. 74, no. 11, November 1997 (1997-11), pages 906-916, XP001028680	1-3, 6-10,16, 19-23
A	page 911, right-hand column, line 3 - page 912, left-hand column, line 17 page 912, right-hand column, line 54 - page 913, right-hand column, line 4	4,5,17, 18

-/--

[X] Further documents are listed in the continuation of box C.

[X] Patent family members are listed in annex.

° Special categories of cited documents :

"A" document defining the general state of the art which is not considered to be of particular relevance

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"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.

"&" document member of the same patent family

Date of the actual completion of the international search	Date of mailing of the international search report
27 September 2001	12/10/2001

Name and mailing address of the ISA	Authorized officer
European Patent Office, P.B. 5818 Patentlaan 2 NL – 2280 HV Rijswijk Tel. (+31-70) 340-2040, Tx. 31 651 epo nl, Fax: (+31-70) 340-3016	Knüpling, M

C.(Continuation) DOCUMENTS CONSIDERED TO BE RELEVANT

Category °	Citation of document, with indication,where appropriate, of the relevant passages	Relevant to claim No.
Y	WO 99 27334 A (AUTONOMOUS TECHNOLOGIES CORP) 3 June 1999 (1999-06-03) page 9, line 24 -page 11, line 3 page 13, line 23 -page 15, line 25 page 32, line 9 - line 24 page 33, line 2 - line 5 ---	1-3, 6-10,16, 19-23
Y	MIERDEL PETER ET AL: "Effects of photorefractive keratectomy and cataract surgery on ocular optical errors of higher order" GRAEFE'S ARCHIVE FOR CLINICAL AND EXPERIMENTAL OPHTHALMOLOGY, SPRINGER VERLAG, vol. 237, 1999, pages 725-729, XP002178635 abstract page 728 - 729, section "Discussion" ---	1-3,6, 8-10,16, 19,21-23
A	WO 98 53881 A (HINKSON STEPHEN J; MUNNERLYN CHARLES R (US); SHIMMICK JOHN K (US);) 3 December 1998 (1998-12-03) page 18, line 2 - line 28 -----	1,10,23

INTERNATIONAL SEARCH REPORT

Information on patent family members

International Application No

PCT/US 01/40352

Patent document cited in search report		Publication date	Patent family member(s)		Publication date
WO 9927334	A	03-06-1999	WO	9927334 A1	03-06-1999
			AU	5459398 A	15-06-1999
			BR	9714878 A	09-01-2001
			EP	1032809 A1	06-09-2000
WO 9853881	A	03-12-1998	AU	7695398 A	30-12-1998
			WO	9853881 A1	03-12-1998