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(54) **METHOD AND DEVICE FOR SYNCHRONOUS MAPPING OF THE TOTAL REFRACTION NON-HOMOGENEITY OF THE EYE AND ITS REFRACTIVE COMPONENTS**

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A61B 3/10 (2006.01)

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See application file for complete search history.

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(74) *Attorney, Agent, or Firm* — Benjamin Aaron Adler

(57) **ABSTRACT**

An instrument for measuring aberration refraction of an eye is provided, having: a lens system defining an instrument optical axis and an alignment device for aligning the visual axis of the eye with the instrument optical axis. A light source produces a probing beam that is projected through the lens system parallel to the instrument optical axis and is selectably positionable partially off-set from the instrument optical axis for entering the eye parallel to the instrument optical axis at a plurality of locations on the cornea of the eye. A first photodetector measures the position of a first portion of the probing beam light scattered back from the retina of the eye to measure aberration refraction of the eye at a plurality of locations. A second photodetector synchronously measures the position of a second portion of the probing beam light reflected back from the cornea of the eye.

30 Claims, 9 Drawing Sheets

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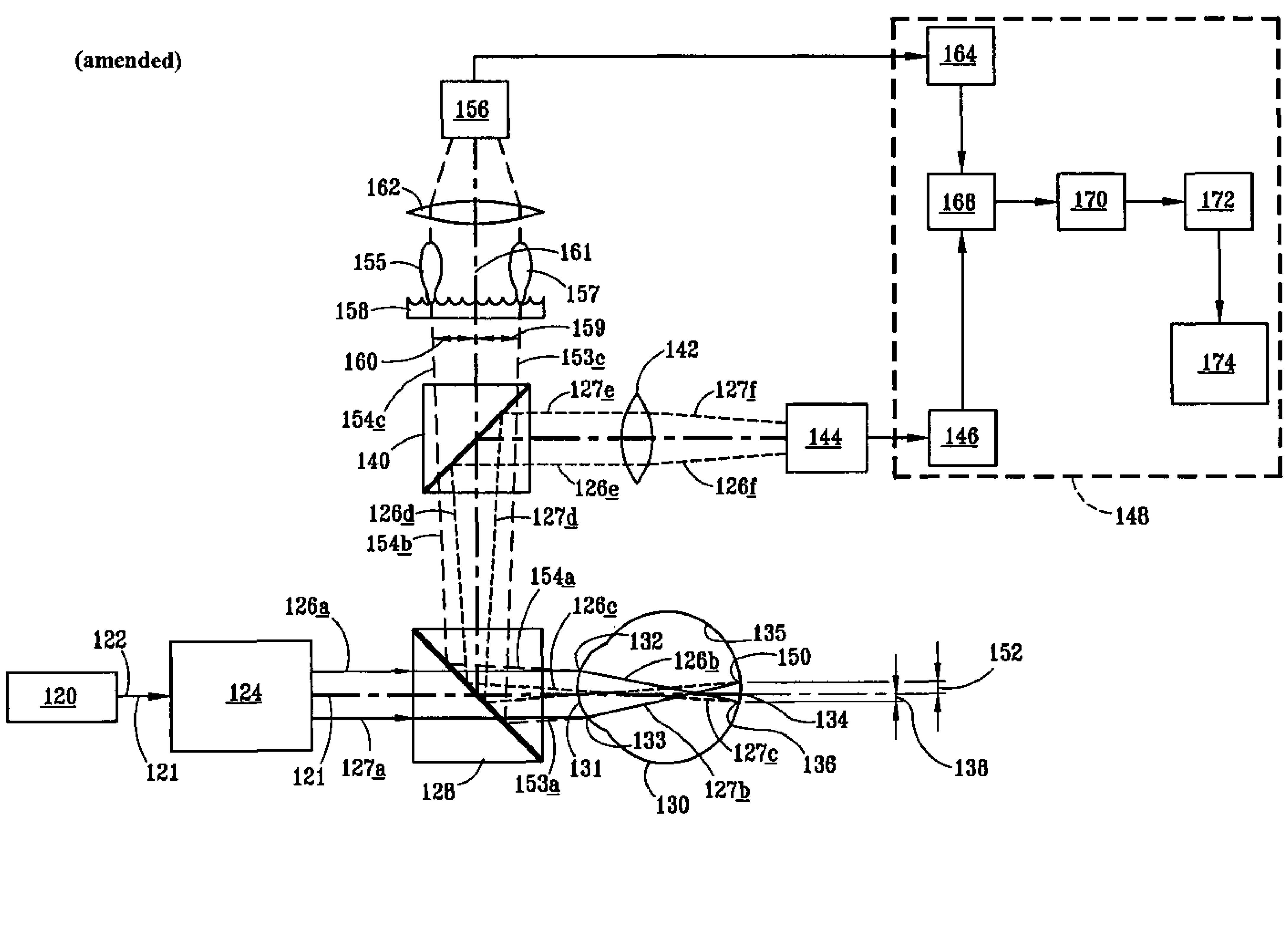

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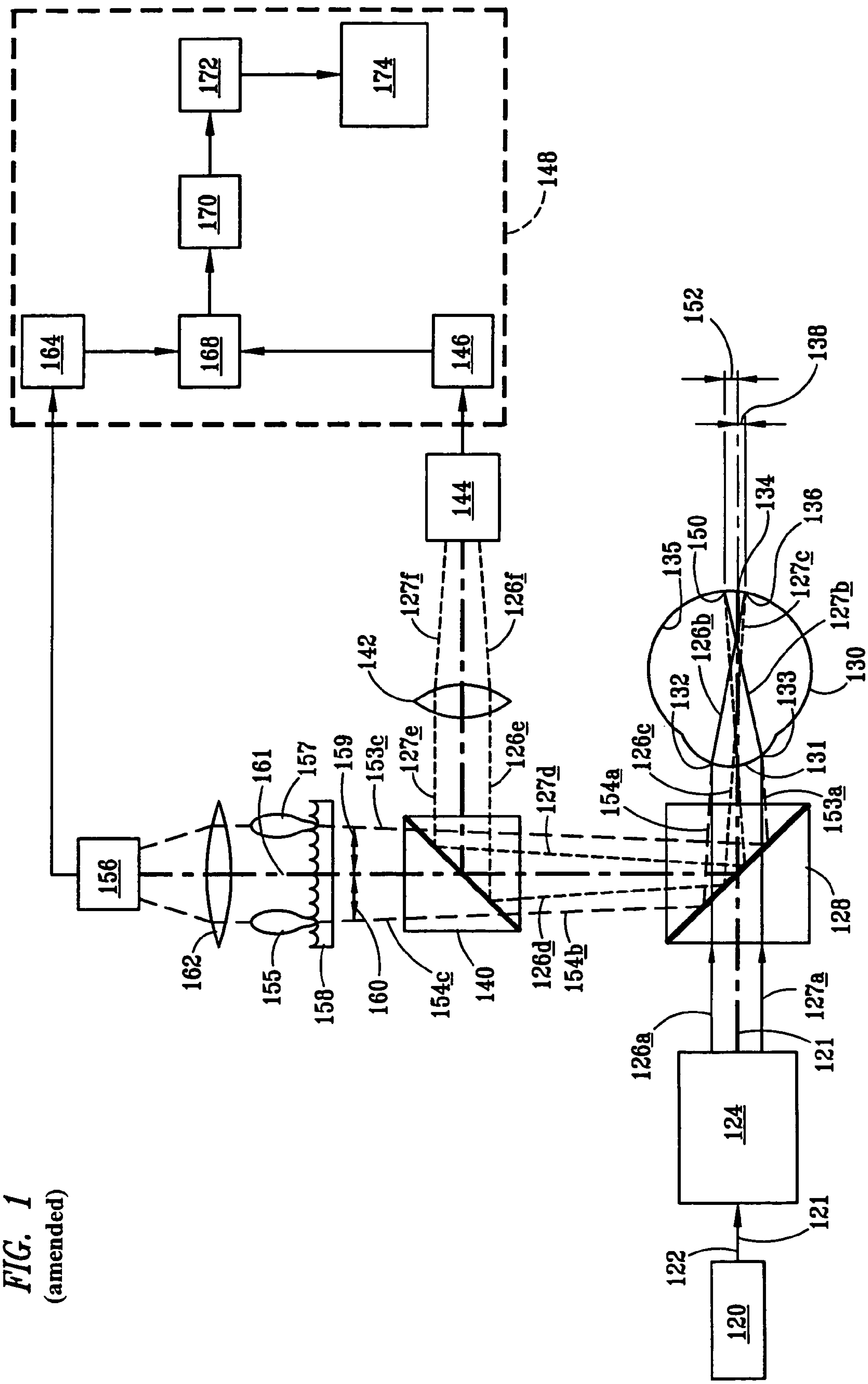

FIG. 1
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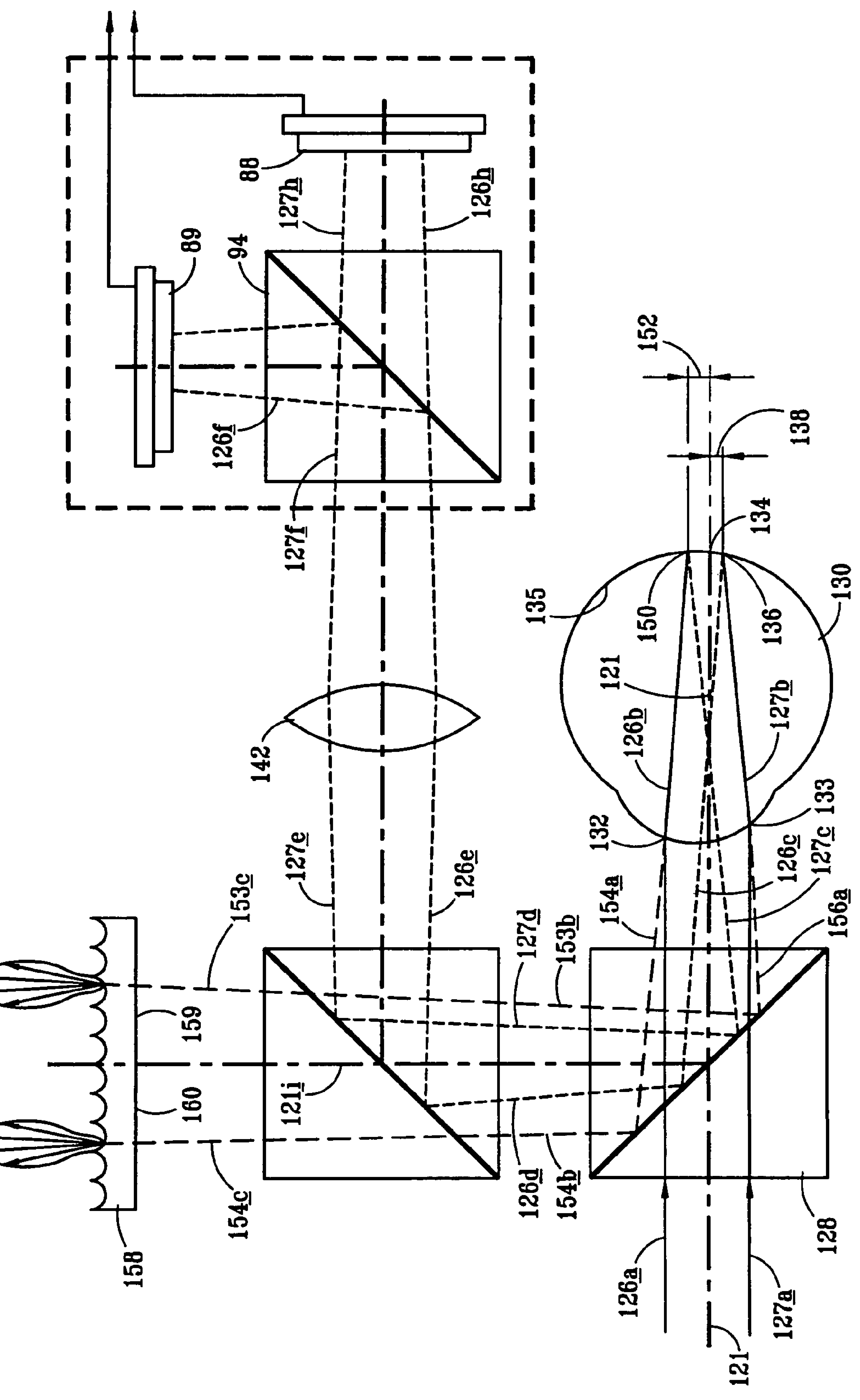

FIG. 2
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FIG. 3
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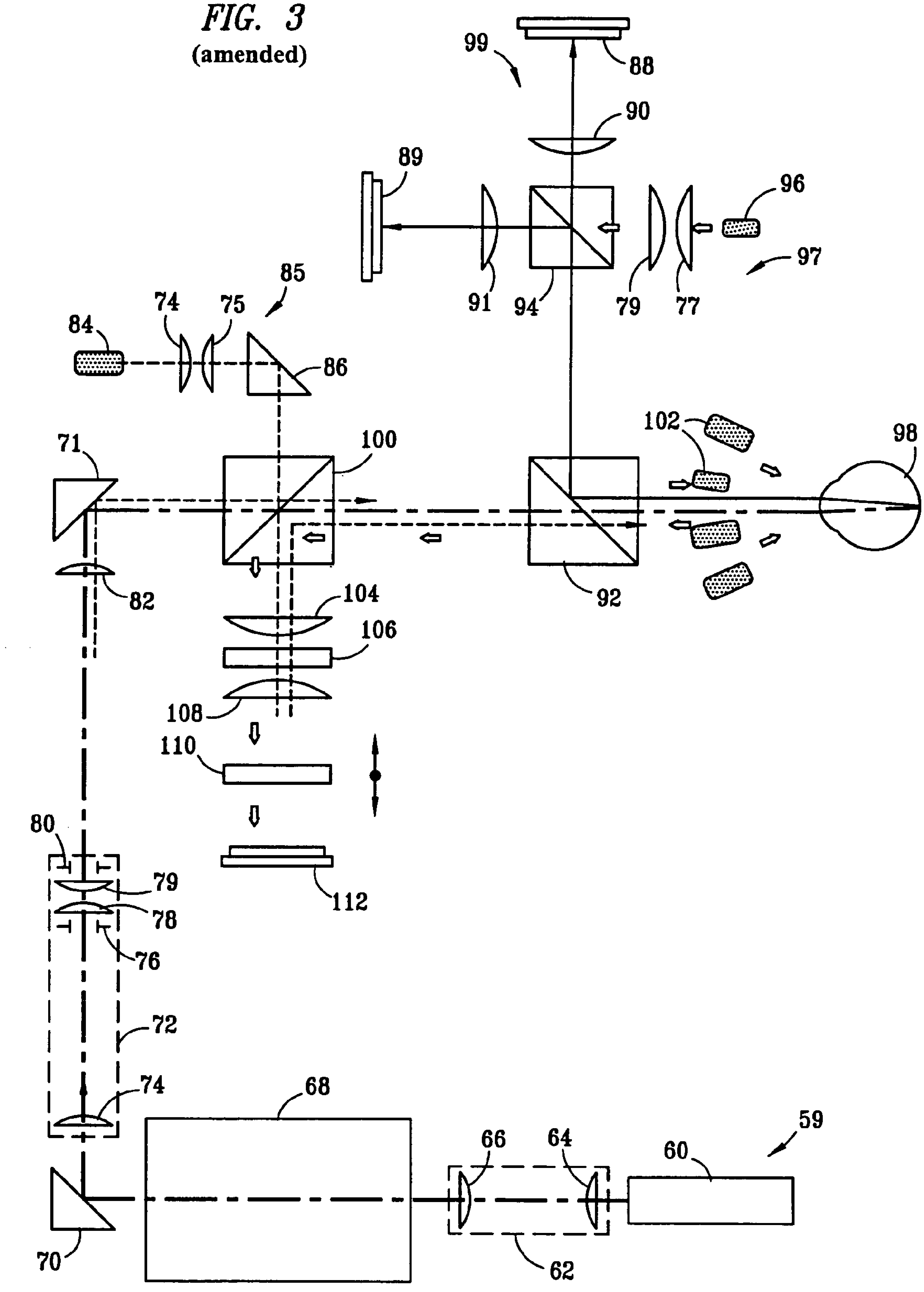

FIG. 4
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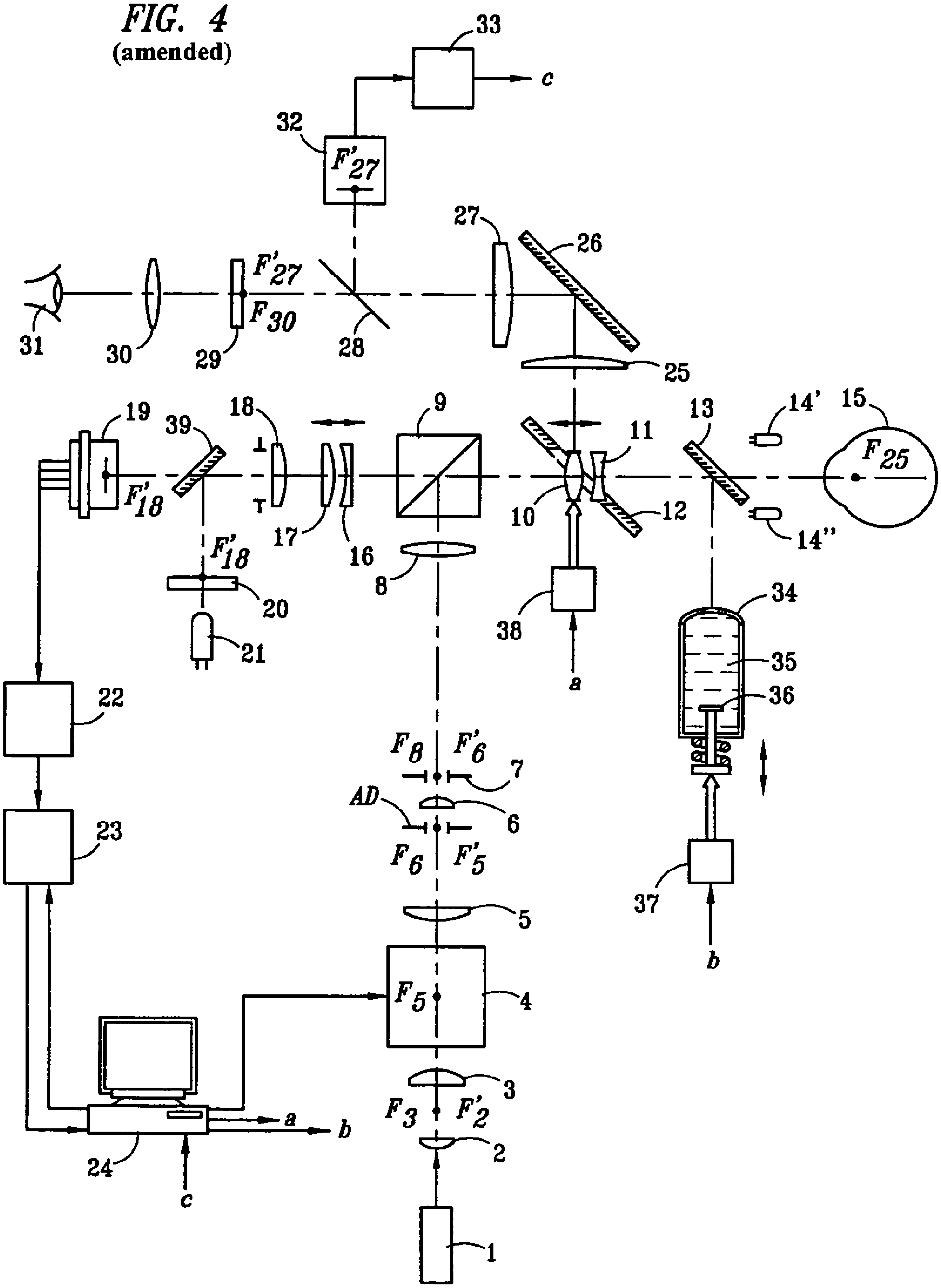

FIG. 5 (amended)

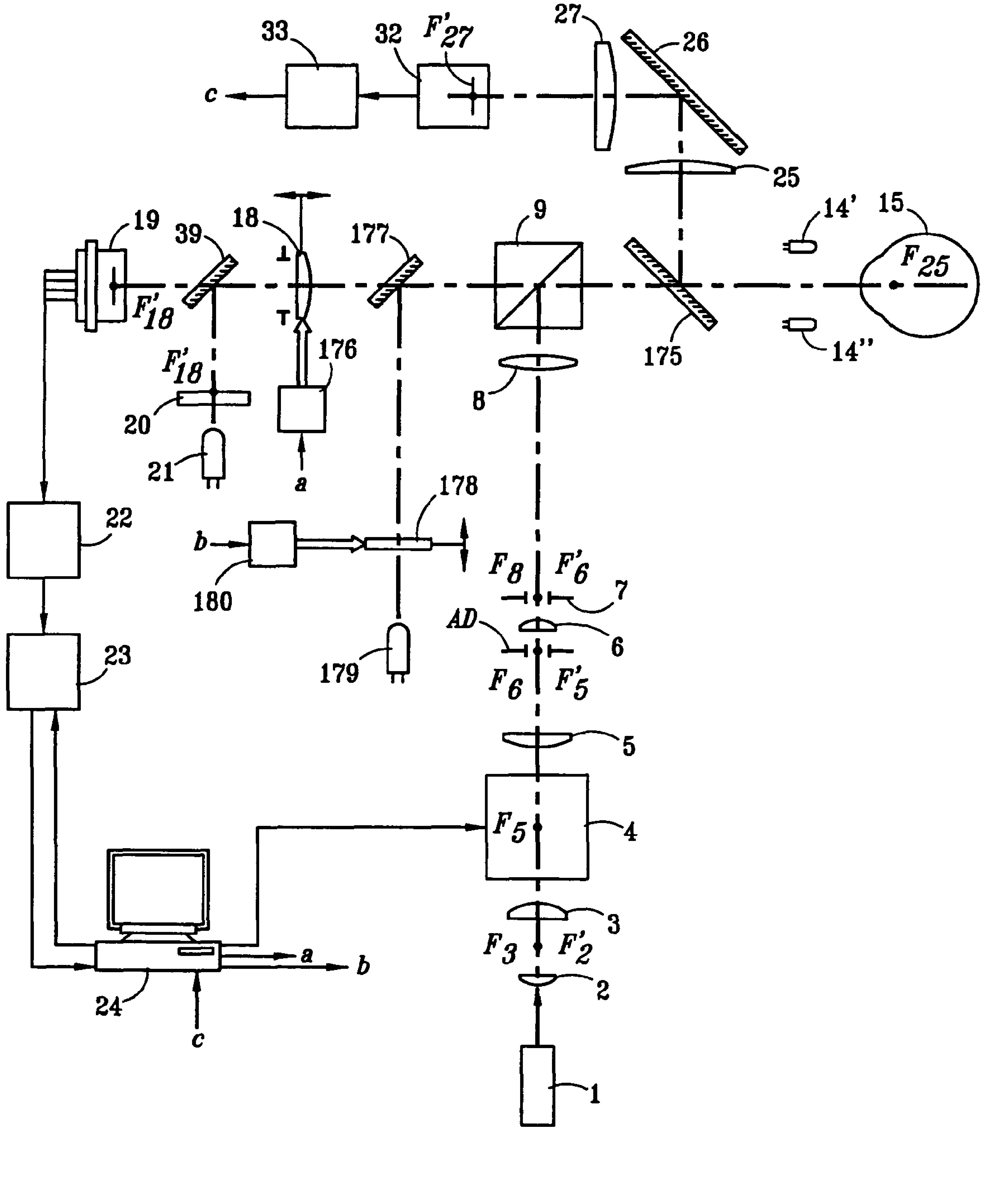

FIG. 6 (amended)

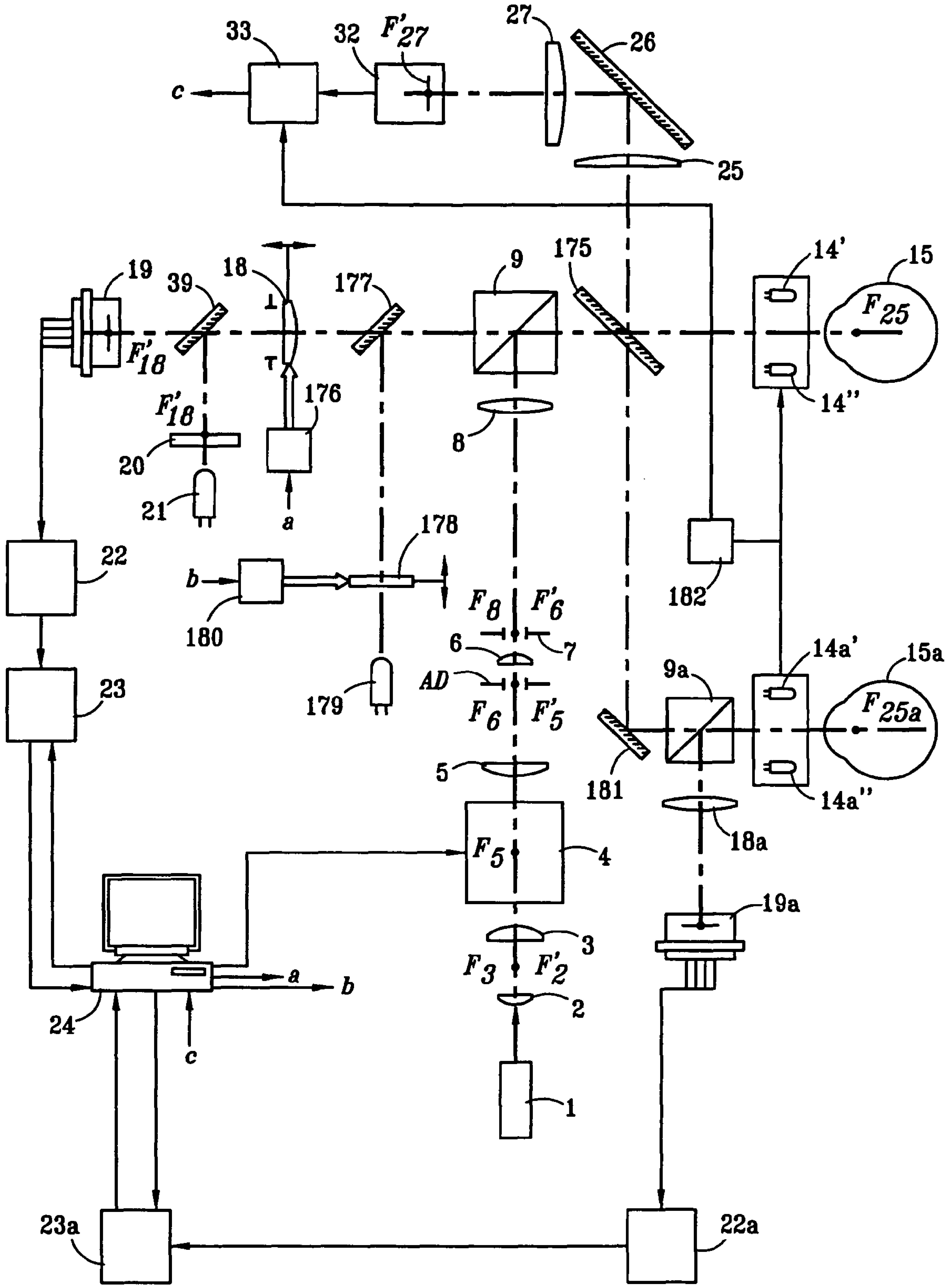

FIG. 7 (amended)

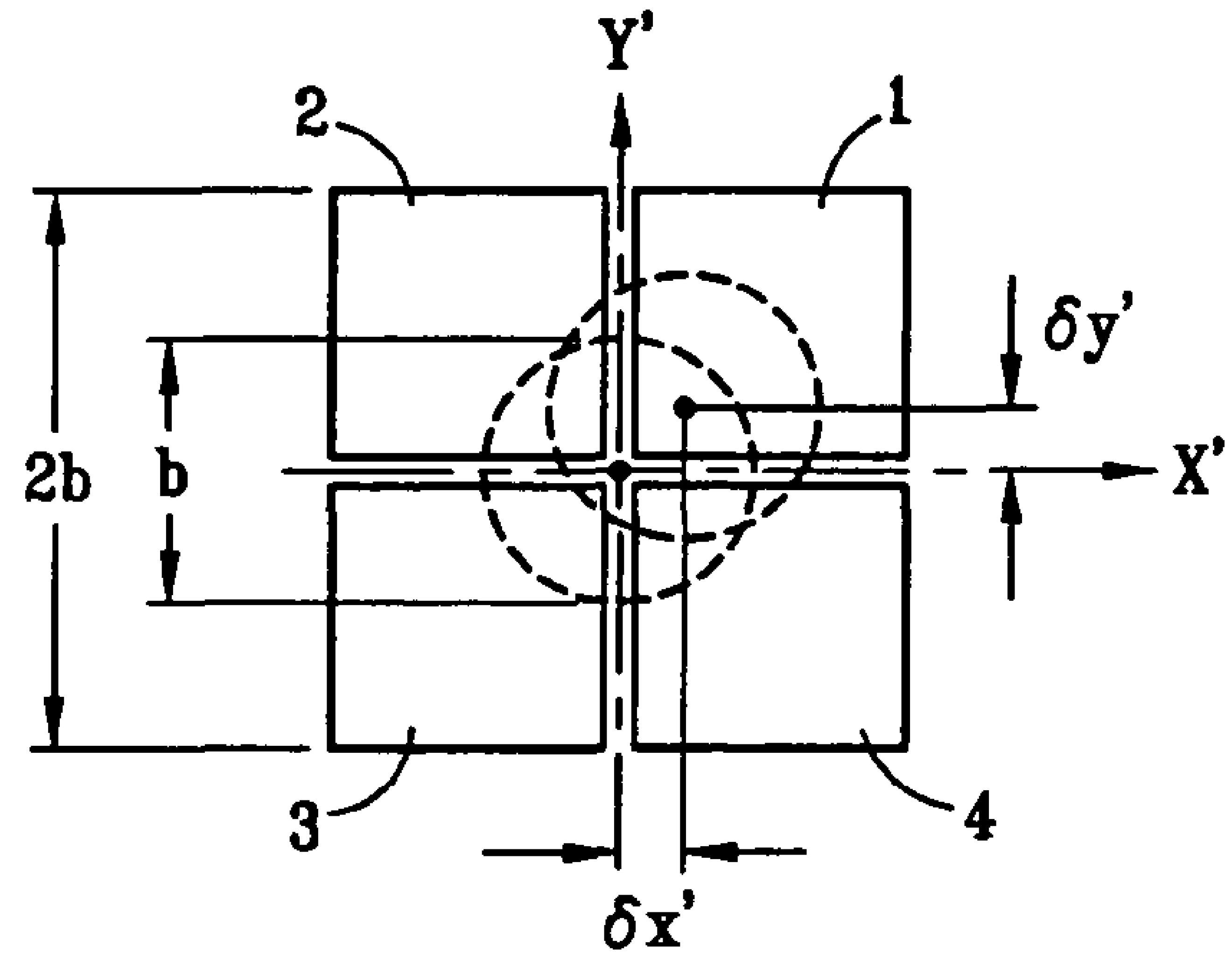

FIG. 8 (amended)

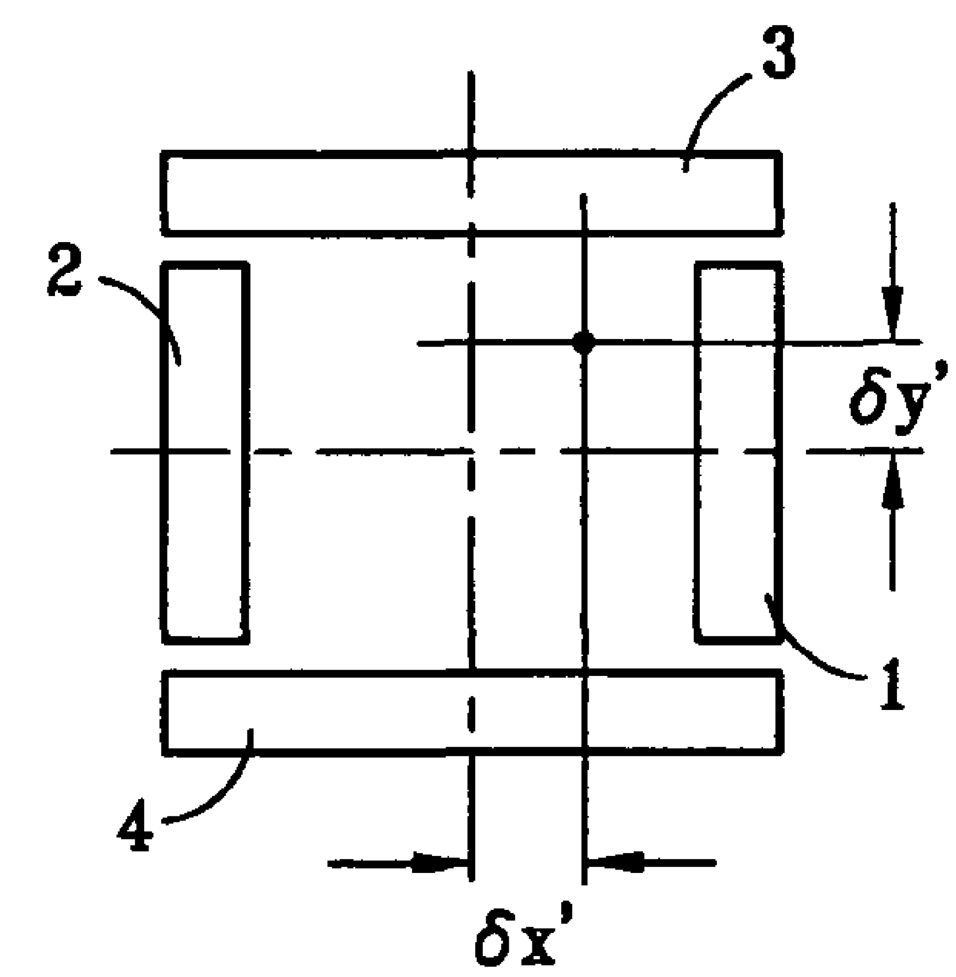

FIG. 9 (amended)

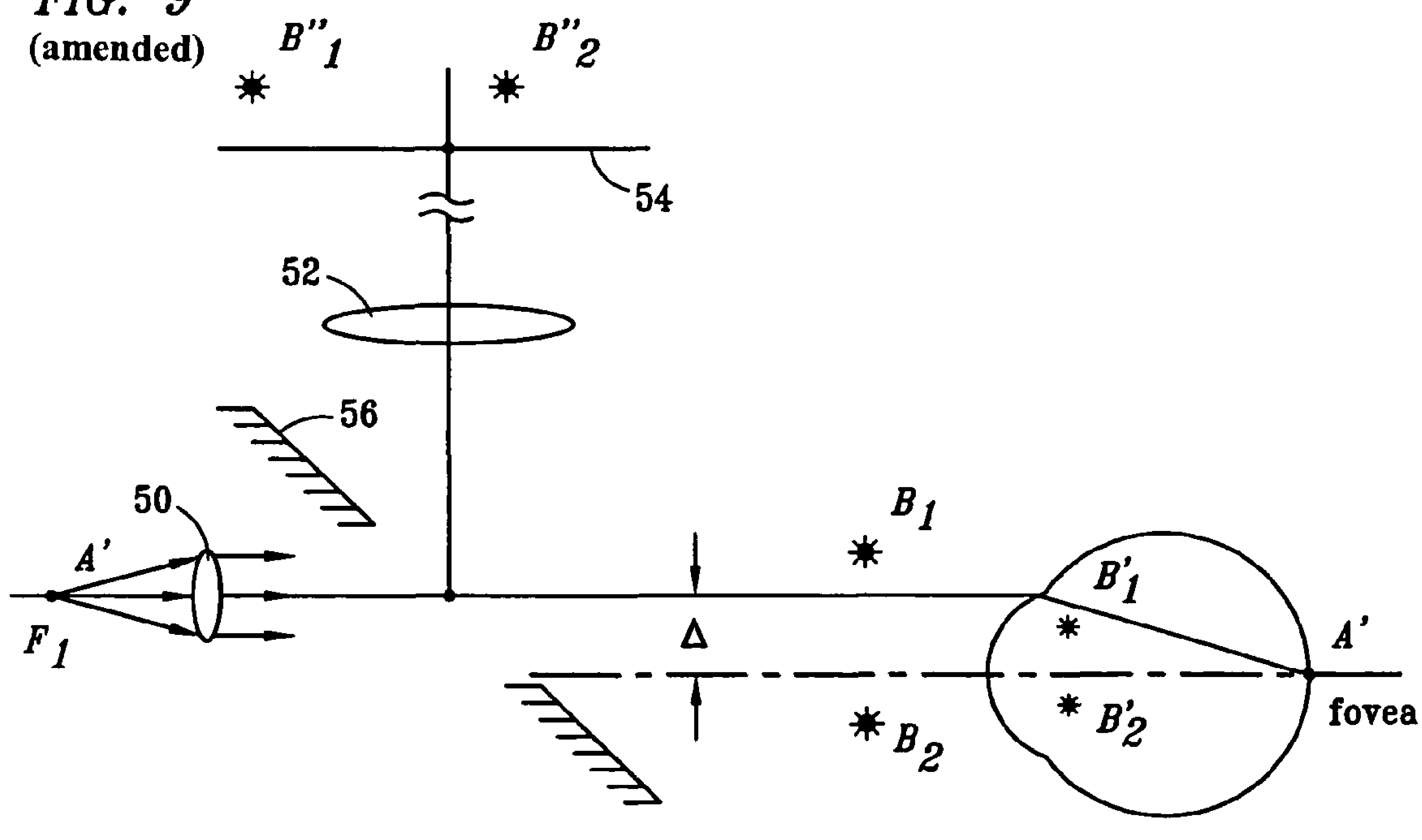

FIG. 10
(amended)

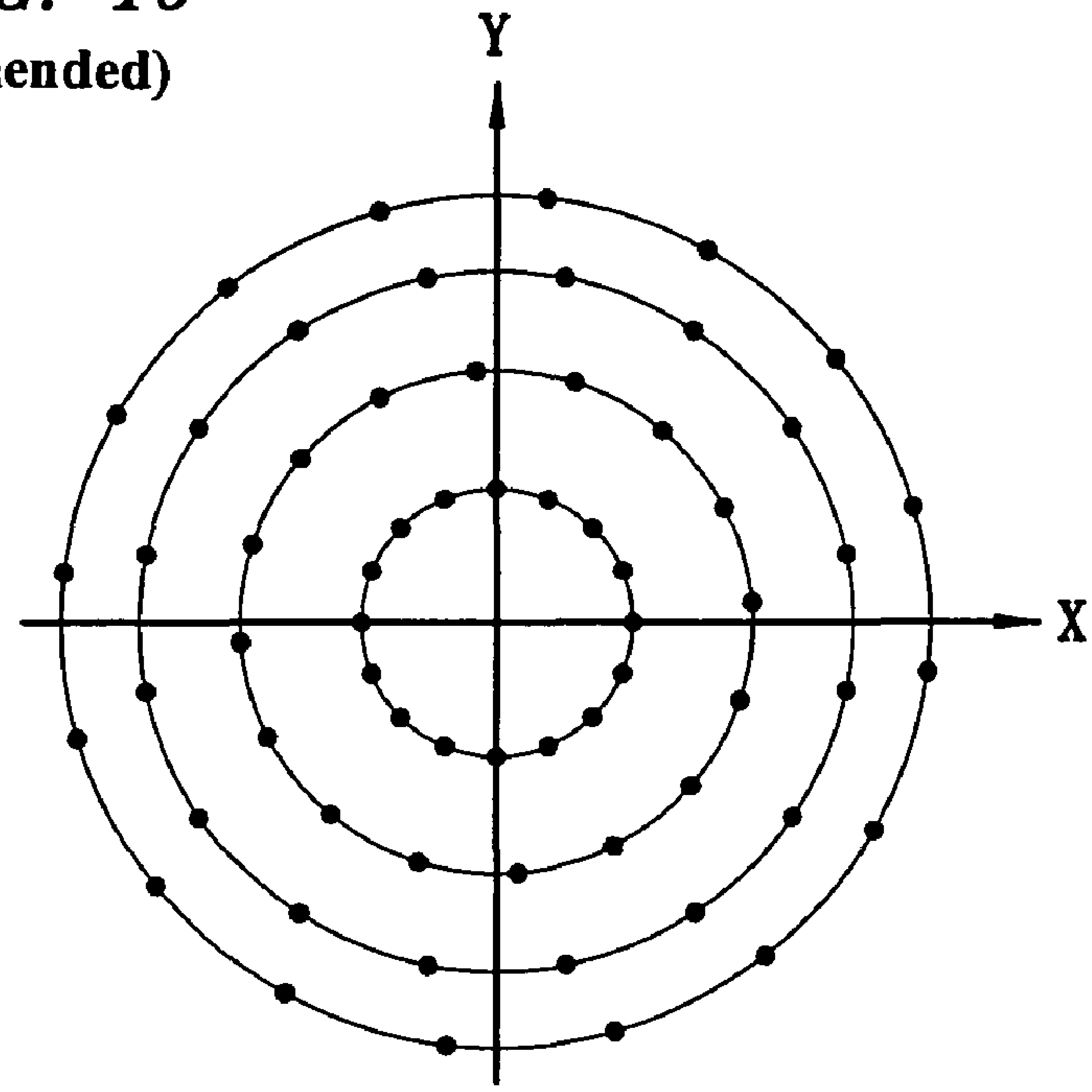

FIG. 11
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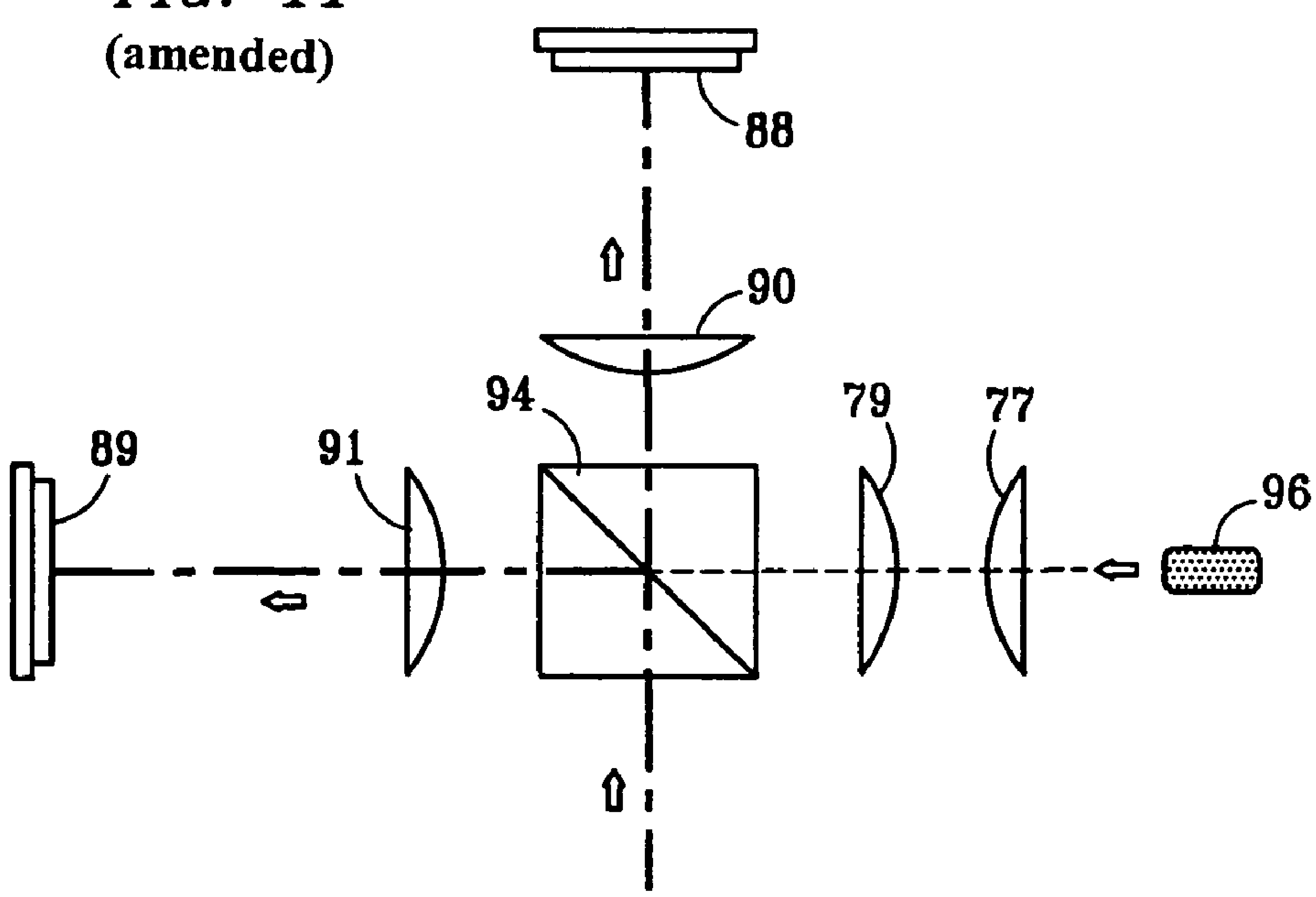

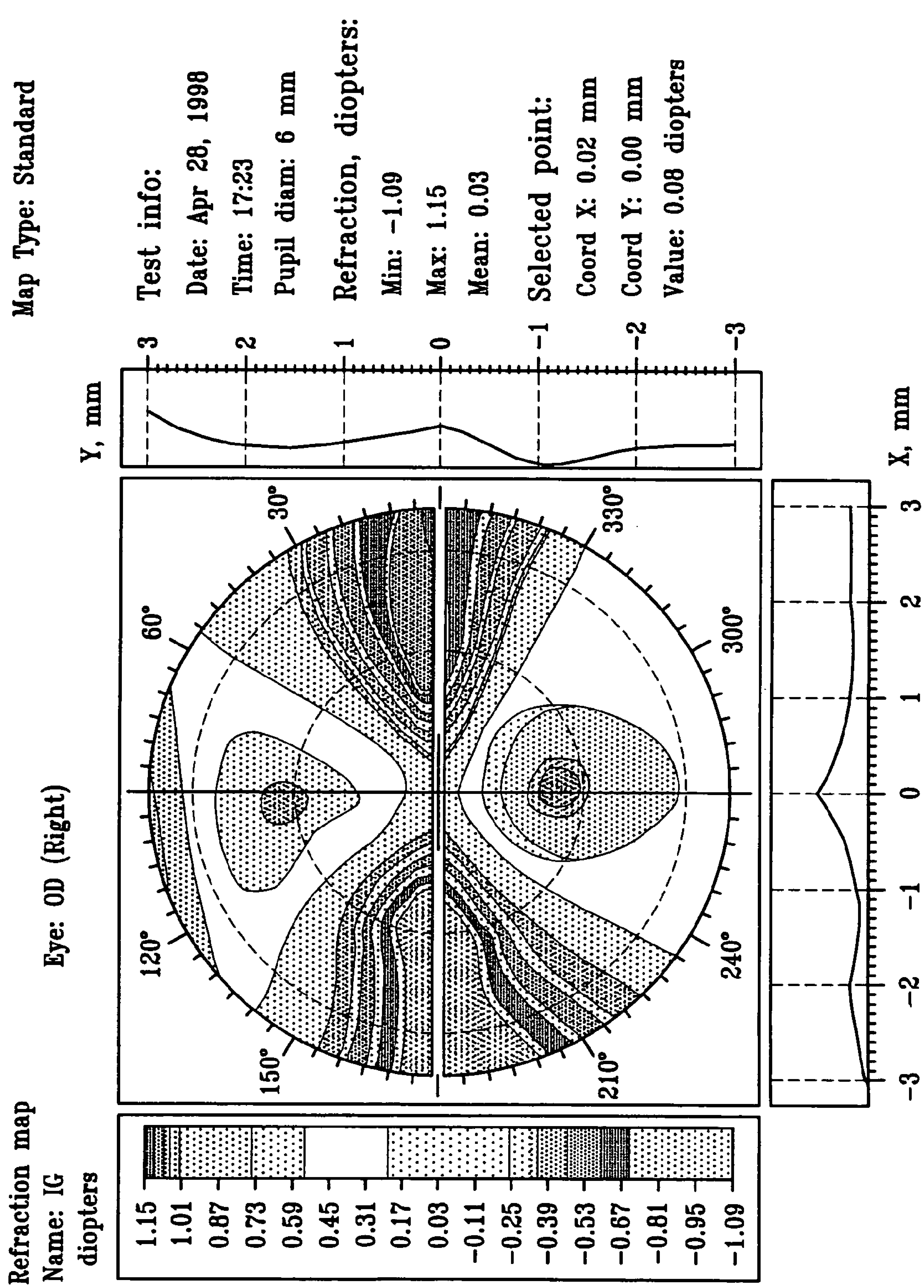

FIG. 12 (amended)

METHOD AND DEVICE FOR SYNCHRONOUS MAPPING OF THE TOTAL REFRACTION NON-HOMOGENEITY OF THE EYE AND ITS REFRACTIVE COMPONENTS

Matter enclosed in heavy brackets [ ] appears in the original patent but forms no part of this reissue specification; matter printed in italics indicates the additions made by reissue.

CROSS-REFERENCE TO RELATED PATENTS AND APPLICATIONS

This is a continuation-in-part of PCT Application No. PCT/US99/23327*, with an international filing date of Oct. 7, 1999,* in which the US is a designated country and which claims *priority to* Ukranian Patent Application No. 98105286[as a priority document]*, with a priority filing date of Oct. 7, 1998.*

TECHNICAL FIELD OF THE INVENTION

The present invention relates to medical ophthalmological equipment, more specifically, it relates to devices for measuring the refraction of the eye as a function of spatial pupil coordinates.

BACKGROUND OF THE INVENTION

A method and a device for mapping the total refraction non-homogeneity of an eye are set forth in prior co-pending PCT Application No. PCT/US99/23327 that includes directing into the eye a narrow laser beam, its axis being parallel to the visual axis of the eye under investigation, scanning the beam over the eye aperture, receiving a portion of light scattered by the retina, analyzing the position of the laser spot projected on the retina, and reconstructing from the data a map of the total refraction of the eye. A discussion of the details of the total refraction non-homogeneity determination have been incorporated herein and set forth below.

In many applications, information on the contribution of other refractive components of the eye may be helpful or necessary, as, for example, for subsequent corrective surgery.

For the purpose of measuring the surface shape of a cornea, a method is known of projecting a regular structure or regular patterns, such as a pattern of concentric disks onto the cornea, analyzing the reflected light and reconstructing from the analyzed data the shape and therefore the refraction distribution caused by the cornea. It has been discovered by Applicants that marrying the techniques for analyzing retina-scattered light and for analyzing cornea reflected light may give very useful information on the contribution to the total eye refraction of such other refractive components of the cornea, and/or the eye lens, that has not heretofore been successfully accomplished.

Measuring devices are known, for the study of the refraction component of the optical system of the eye, which depend on spatial pupil coordinates. These include M. S. [Smimov's] *Smirnov's* device for measuring the wave aberration [1], Van den Brink's device for measuring the transverse aberration [2], N. M. Sergienko's device for measuring the physiological astigmatism [3], and a spatially resolved refractometer [4]. The above devices, based on Scheiner's principle, involve point-by-point investigation over utilizing a number of optical techniques. However, in using all such devices the direct participation of the patient is needed in the preliminary aligning of the eye and in the aberration measurements.

Major disadvantages of the above measuring devices are their low accuracy and productivity, a prolonged measurement process resulting in the patient's fatigue, variations in accommodation, and eye movements while taking measurements, thereby increasing the aberration measurement errors.

More advanced measuring devices are known, which do not require the patient to act as a link in the "measurement chain". These include a device for measuring the aberration by the Foucault's knife method [5], a device for measuring the wave aberration using Hartmann-Shack sensors [6-8], including measurements that incorporate adaptive optics completely compensating the wave aberration [9].

A common disadvantage of the measuring devices with a Hartmann-Shack sensor is the fixed field of view of the raster photoelectric analyzer of transverse aberrations due to the mechanically rigid construction of the lens raster and the invariable mutual spatial arrangement of the photosensitive elements of the charged coupled device or CCD camera. This results in a fixed configuration of grid sites at the pupil plane in which aberrations are measured, with no flexibility of reconfiguring these grid sites for more detailed measurements in separate zones of the pupil depending on their aberration properties.

Other disadvantages of existing devices include: they do not incorporate means for providing an accurate reproducible "linkage" of the patient's eye to the spatial co-ordinates of the measuring device; they do not incorporate a means for adjusting the accommodation of the patient's eye that is necessary for studying the dependence of aberrations on the accommodation characteristics; they are not capable of taking measurements on a dilated pupil without using medicines.

Refraction can also be measured using a spatially resolved objective autorefractometer as disclosed in U.S. Pat. No. 5,258,791 [10]. This device provides spatially resolved refraction data using a closed measuring loop which includes a reference pattern and a measurement beam. In this device, an origin of coordinates of the detector coincides with the center of the fovea image and the detector functions as a zero-position sensor.

The spatially resolved objective autorefractometer disclosed in U.S. Pat. No. 5,258,791 preferably using laser ray tracing, has a number of substantial problems relating to performance in the following basic and auxiliary functions: preliminary alignment of the optical axis of the device relative to the visual axis of the eye; accommodation monitoring of the patient's eye; allocation of points within the pupil at which refraction is measured; and measurement of the angle of laser beam incidence into the patients eye. Respective to the above basic and auxiliary functions, the above drawbacks are inherent to the device disclosed in U.S. Pat. No. 5,258,791.

Preliminary alignment of the optical axis of prior art devices relative to the visual axis of the eye may be problematic for at least the following reasons: first, the visual axis of the eye is assumed to be the line passing through the geometric center of the pupil and the fovea. However, it is known that the geometric center of the pupil does not always coincide with the visual axis due to the misalignment of the pupil opening and the optical axis of the cornea and the crystalline lens. In addition, the pupil may not be symmetrical.

Second, in prior art devices in which alignment is dependant on fixation of the patient's gaze at a focal point, changes in the position of the point at which the patients gaze is fixed results in angular movement of the patient's eye which disturbs the previous alignment. Consequently, both points (on the pupil and on the retina) through which the center line passes do not have a definite location.

Third, the focal points in devices without ametropia compensation can clearly be observed only with an emmetropic or normal eye. When the patient's eye is ametropic, such devices will see a diffused laser beam spot whose width increases with the ametropy. It is obvious that under such conditions the gaze cannot be fixed accurately in a certain direction, which is another factor preventing an accurate alignment. Another drawback of prior art refractometers is that the fovea and the photosensitive surface of the photodetector are optically coupled by the lenses only in the emmetropic or normal eye. In the event of an ametropic eye, the decentering or defocusing of the fovea image on the above-mentioned surface of the photoelectric detector causes additional refraction measurement errors which are not compensated for. The present invention is designed to compensate for this.

Fourth, a sufficiently bright laser radiation may irritate the fovea to such a degree that the eye begins to narrow its pupil reflexly. Therefore, before performance of the eye centering procedure, medicines paralyzing the ciliar body muscles are likely to be required, which changes the refractive properties of the eye as compared with its normal natural state.

The need for accommodation monitoring of the patient's eye has not been satisfied in prior art devices. As a consequence, the patient's eye can be accommodating at any distance. It is known that the refractive properties of the eye depend on the accommodation distance. Because the accommodation is unknown to the operator, it is impossible to correlate the refraction map and the eye accommodation.

It has become apparent to the present applicants that a spatially resolved refractometer should preferably include a device for adjusting to the patient's eye accommodation.

Prior art devices using electromechanical actuators greatly reduce the possibility of ensuring a high-speed scanning of the pupil and the possibility of shortening the duration of the ocular refraction measurement process.

In prior art devices using a disc or movable aperture bearing planar surface to control laser targeting, the aperture occupies only a small portion of the zone in which the laser beam intersects the planar surface. Thus, only that portion of the laser beam which is equal to the ratio of the area of one refraction measurement zone on the pupil to the entire pupil area passes through the aperture. Such a vignetting of the laser beam results in an uneconomical use of laser radiation and should be considered a major drawback of such designs.

Drawbacks in the measurement of the angle of laser beam incidence at which it crosses the necessary measurement zone of the pupil and the center of the fovea are inherent to designs which do not provide a sufficiently high refraction measurement speed. In such designs, the time of measuring the refraction at 10 measurement points of the pupil is up to one minute. During this period the patient's eye can move up to 100 times and change its angular position due to natural tremor, "jumps" and drift.

Systematic instrument errors have plagued prior aberration refractometeis. Due to an irregular distribution of the light irradiance within the light spot on the retina, unequal photosensitivity across the surface photoelectric detector, time instability of the gain of preamplifiers connected to the photoelectric detector elements, and the presence of unsuppressed glares and background illumination the photodetector, the photodetector does not register a "zero" position of the spot on the fovea without systematic errors. Further, as a result of its own aberrations, the optical system providing for eye ray tracing contributes an angular aberration to the laser beam position. The present instrument incorporates structural elements which compensate for such errors and thus increase the refraction measurement accuracy.

What is needed is an improved electro-optical ray tracing aberration refractometer which makes it possible to achieve the following goals: flexibility of allocation of measuring points within the pupil and to pupil and improvement in the effectiveness of using lasers by reducing the vignetting of the laser beam at the aperture diaphragm; reduction in the duration of measuring refraction across the entire pupil to around 10-20 milliseconds; ensuring optical coupling of the photosensitive surface of the photodetector with the fovea even for ametropic eyes as well as for accommodation monitoring at any given distance; synchronously measuring both the total eye refraction aberrations and the component caused by cornea refraction characteristics maintaining incremental accuracy by interposing additional impingement points between impact points with wide variation in refraction characteristics; reduction in instrument errors when measuring aberration refraction; enhancement of the accuracy and definitiveness of instrument positioning relative to the patient's eye; the potential for automation of the positioning and controllability of the working distance between the patient's eye and the device components; and enablement of instrument positioning without medically dilating the pupil. The present invention provides the aforementioned solutions and innovations.

It has also been found that, due to noncoincidence of the points of primary information from the attempted combination of total eye refraction information and the surface shape of a cornea, and the followed approximation, any maps reconstructed from such cornea topography data and separately obtained total eye refraction aberration data would contain significant errors when reconstructing the differences. A workable device and method for synchronous mapping of the total refraction non-homogeneity of the eye and of the refraction components caused by the cornea would be desirable.

SUMMARY OF THE INVENTION

To avoid these and other drawbacks and solve the problems and produce a workable device and method for synchronous mapping of both the total refraction non-homogeneity of the eye and its refractive components, the following inventive technology has been devised. A device for synchronous measuring aberration refraction of an eye and calculation of the component of the total aberration refraction caused by the cornea has been accomplished with a device comprising a polarized light source producing a probing beam along a path to the eye. A beam shifter is provided to rapidly shift the probing beam to a plurality of spaced-apart, parallel paths for impacting the eye at a plurality of points on the cornea. Nonpolarized backscattered light from the retina of the eye is directed through a polarizing beam splitter onto a first position-sensitive detector by which the total eye refraction is determined for each of the plurality of impingement points. Synchronized with each impingement point determination of the total eye refraction, there is a polarized reflection from the impingement point on the cornea directed through the polarizing beam splitter and to a second position-sensitive detector for determining the refraction component caused by the cornea. A comparator can synchronously compare the total refraction for each impingement point to the component of refraction caused by the cornea and by subtraction can determine the component of refraction caused by the parts of the eye other than the cornea. This data can be placed in memory and/or can be supplied to a program device for map reconstruction and to a representative display of the refraction characteristics of the eye and its components.

For additionally improved resolution, a program device may be provided for inserting an additional measuring point or impingement point located between neighboring measuring points that produce a different value higher than a specified maximum difference threshold.

Thus, the object of the present invention is to provide an improved polarized light device for measuring the aberration refraction of the eye. The aberration refractometer allows estimates of the ametropy, astigmatism characteristics, simultaneously allows synchronous determination of component parts of total aberration refraction of the eye contributed by the cornea and therefore the components contributed by other portions of the eye and thereby allowing visual acuity and increased accuracy of calculations of the part of the cornea to be removed by photorefractive keratectomy [11] if necessary to correct eye refraction non-homogeneity or aberration.

In a preferred embodiment, the aberration refractometer comprises a light radiation source, preferably laser light or other polarized light; a telescopic system; a two-coordinate deflector consisting of two single-coordinate deflectors; a deflection angle control unit; an aperture stop; a field stop; a collimating lens; an interferential polarizing beam splitter; a first position-sensitive photodetector with an objective lens for detecting only the position of light backscattered from the retina; a second photosensitive photodetector with an objective for detecting only the position of polarized light reflected from the cornea; a comparator for point-by-point comparison of the retinal backscatter position and the cornea reflection position; and a data processing and display unit consisting of a computer, an analog-to-digital converter and a preamplifier. Use of laser or polarized light as a probing beam in combination with the interferential beam splitter allows the polarized light to be separated from the nonpolarized light reflected back from the retina and prevents it being detected by the first photodetector and further allows polarized light reflected from the cornea to be detected by the second photodetector without detecting the backscattered light from the retina.

The instrument of the present invention is able to reduce the time needed for measuring the refraction, eliminate light beam energy losses at the aperture stop and create a flexible system for locating the measurement points on the pupil by providing the following: the telescopic system is positioned in the probing beam path after the two-coordinate first deflector at a distance corresponding to the coincidence of the entrance pupil of the telescopic system and the gap or zone between the single-coordinate deflectors, the aperture slop or diaphragm is placed between the lenses of the telescopic system at the point of coincidence of their foci, and the field stop or diaphragm is positioned in the plane of the exit pupil of the telescopic system and, at the same time, at the location of the front focus point of the collimating lens situated in front of the interferential polarizing beam splitter at such a distance from the patient's eye which is approximately equal to the focal distance of the collimating lens.

To ensure a constant optical coupling of the photosensitive surface of the first photodetector and the retina for both emmetropic and ametropic eyes, a group of lenses with variable optical power is installed between the interferential polarizing beam splitter and the eye, said group of lenses having the function to adjustably form the retina image of an ametropic eye at infinity regardless of the emmetropic or ametropic condition of the eye. The photosensitive surface of the first photodetector is conjugated with the front focal plane of the objective lens, being inserted following the interferential polarizing beam splitter on the path of the light scattered by the retina.

To provide for fixation of the patient's line of sight along the optical axis of the instrument and to compensate for accommodation of the eye at the required distance while keeping constant optical conjugation of the patient's eye with the photosensitive surface of the detector, a second beam splitter or an optical axis rotation mirror, as well as a plate with a gaze fixing test pattern or a test-target for sight fixation are optically coupled with the photosensitive surface of the photodetector and are located between the photodetector and the objective lens. A second optical group of lenses, with variable negative optical power and which function to form an image for the patient's eye of the test-target at a distance corresponding to the preset accommodation, is positioned between the second beam splitter or optical axis bending mirror and the interferential polarizing beam splitter. When an optical axis bending mirror is used, it is mounted on a movable base making it possible to displace the mirror so as to enable the light radiation scattered by the retina to reach the photodetector during the measurement of the patient's eye characteristics.

In one embodiment, to account for systematic refraction measurement errors, a second mirror, for bending or redirecting the optical axis of the probing laser beam is inserted in the laser beam path after the last optical element before entering the patient's eye. Following the second mirror, an optical calibration unit for simulating an eye is inserted. The optical calibration unit includes an axially movable or stationary retina simulator whose optical characteristics are equivalent to those of the human retina. The second optical axis bending mirror is installed on a movable base so that it can be moved into the probing laser beam path during measurement with the optical calibration unit and moved out when measuring the patients eye refraction.

To align the instrument relative to the patient's eye as well as to enhance accuracy and enable automation of the aligning process, the instrument is provided with a third beam splitter to insert a channel for eye alignment verification of the instrument and the patient's eye. In a preferred embodiment, the co-axial verification channel comprises one or more point of light sources and a TV or electro-optic detecting device, together serving to display the pupil and/or eye image and providing a permission channel to measure eye characteristics when the optical axis of the instrument and the visual axis of the patient's eye coincide. To enable the instrument to be used without dilating the pupil with a medicine, a laser radiation source and/or infrared light sources are incorporated into the coaxial verification alignment mechanism. It is contemplated that the instrument can be used to make refraction measurements under conditions simulating either day or night light conditions.

In an alternate embodiments it is further contemplated that the alignment verification can be done under the control of the operator of the device or can be automated. In one embodiment, the co-axial verification channel provides either visual or acoustic notification that coincidence between the instrument optical axis and the patient's visual axis is proximate or near to "on target" status. Once this status is attained, the instrument is "armed" electronically. Once full coincidence is attained, the measurement controller automatically causes spatially defined parallel light beams, preferably laser beams, to be rapidly fired and enter the eye through the input channel. Light reflecting from the retina is directed to the retinal spot detecting channel for spatial and intensity characterization. This process can permit upwards of at least 5 replicate measurements over 65 spatial locations to be taken within 15 milliseconds without the need for the patient to actively participate in the targeting and alignment process.

BRIEF DESCRIPTION OF THE DRAWINGS

For a more complete understanding of the present invention, including features and advantages, reference is now made to the detailed description of the invention along with the accompanying figures in which like numerals represent like elements and in which:

FIG. **1** is a functional schematic drawing a device for synchronous mapping of the total refraction non-homogeneity of the eye and its refractive components;

FIG. **2** is a schematic depiction of an eye under investigation in which the theoretical aspects of the total eye refraction determination and also the corneal component of refraction may be more fully understood;

FIG. **3** is a schematic diagram of a preferred alternative embodiment of an aberration refractometer according to certain aspects of the present invention;

FIG. **4** is a functional schematic diagram of an instrument for measuring total aberration refraction;

FIG. **5** is a schematic illustration of the operation of one embodiment of a device for measuring the total transverse aberration of a laser beam on the eye retina;

FIG. **6** is a schematic illustration of the operation of another embodiment of a device for measuring the total transverse aberration of a laser beam on the eye retina;

FIG. **7** is a schematic illustration of the operation of yet another embodiment of a device for measuring the total transverse aberration of a laser beam on the eye retina of both of a patient's eyes substantially simultaneously;

FIG. **8** is a schematic illustration of the operation of the lateral position sensing detector having a pair of X direction electrodes and a pair of Y direction electrodes;

FIG. **9** is a schematic illustration of the principle of operation of the means for positioning the measuring device in relation to the patient's eye;

FIG. **10** is an example of a map showing the location of ocular refraction measurement points, constructed with the aid of a computer;

FIG. **11** is a schematic illustration of a photodetector using linear array detector components; and

FIG. **12** is an example map showing the total aberration refraction of the eye, under consideration, constructed with the eye total aberration refractometer portion of the subject invention.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENTS

While the making and using of various embodiments and methods of the present invention are discussed in detail below, it should be appreciated that the present invention provides many applicable inventive concepts which may be employed in a variety of specific contexts. The specific embodiments discussed herein are merely illustrative of specific ways to make and use the invention and do not delimit the scope of the invention.

FIG. **1** shows a schematic functional view of a device and method for synchronous mapping of the total refraction non-homogeneity and components thereof according to one aspect of the present invention. A polarized light source **120** and preferably a laser light source directs a polarized light beam **122** along a polarized light beam path nominally coincident with a central eye axis. The light beam is provided to a beam shifter **124** that is controlled to rapidly shift light beam **122** to any of a plurality of spatially offset parallel light beam paths **126**. The mechanism for light beam shifting will be discussed more fully below in connection with FIGS. **3** and **4**. Beam shifter **124** provides a polarized probing beam **126**. It will be understood that a plurality of such probing beams can be produced by beam shifter **124** within a few milliseconds as, for example, 65 spatially offset parallel polarized probing beams can be produced within 12 milliseconds. For example, only one of the probing beams may be coincident with central axis **121**, another of the plurality of offset polarized probing is depicted as **127**. It will be understood that, for purposes of clarity, the 65 or more parallel polarized probing beams are not depicted in FIG. **1**. Each of the probing beams **126**, as well as the plurality of beams **127**, pass through a beam splitter **128** to the eye **130** under investigation. Each probing beam **126**, **127** impinges upon the cornea **131** at a plurality of impingement points as, for example, point **132** corresponding to probing beam **126** and impingement point **133** corresponding to probing beam **127**, is representative of a plurality of impingement points.

The total refractive eye aberration for a thin beam of light entering at impingement point **132** is determined by locating point **136** on the retina and determining the spatial position **138** of that illuminated point on the retina **136** relative to the fovea **134** aligned along central axis **121**. This position may be indicated relative to central axis **121** by coordinates (dx_1, dy_1). Light impacting the retina is backscattered off the retina. The backscattering is not a reflection per se and therefore the entering beam **126**b is depolarized by the retinal surface. The backscattered light, having its optical axis represented by path **126**c, is non-polarized light. The light **126**c is directed in beam splitter **128** along path **126**d to a polarizing beam splitter **140**. Beam splitter **140** is a polarizing beam splitter so that it reflects non-polarized light and allows polarized light to pass through it. Thus, beam **126**d is directed along path **126**e through a lens **142** that focuses it along path **126**f to a first photo detector **144**. As will be discussed more fully below with respect to FIGS. **2**, **3** and **8**, photo detector **144** may be provided with a further beam splitter **94** and x and y photodetection arrays **88** and **89** to determine the position (dx_1 and dy_1). From this position and based upon the standard length of the eye, the total refraction characteristics can be calculated in a total refraction calculator **146**. The total refraction calculator **146** may, for example, be circuitry and/or computer software within a multifunction computer **148**.

Returning now to FIG. **1**, each of the plurality of offset probing beams of which **127**a is a sample, passes through beam splitter **128** and impinges upon the anterior surface of the cornea **131** at a plurality of impingement points of which **133** is a representative sample. Entering at impingement point **133**, the beam **127**b is projected due to the total refractive characteristics of the eye along the path of beam **127** to a point **150** on the retina **135** of eye **130**. The backscattered light **127**c from the retina **135** is projected back out through the eye aperture and is directed by beam splitter **128** along path **127**d to the polarizing beam splitter **140**, where it is directed along path **127**e through lens **142** and onto a first photodetector **144** for determining the offset location **152** represented by (dx_2, dy_2) away from central axis **121**. Through the rapid and repeated operation of beam shifter **124**, a plurality of times within a fraction of a second, an entire grid pattern of impingement points, see, for example, the grid pattern of FIG. **10**. As will be more fully discussed below, mappable data, with respect to the total eye aberration is provided at the total refraction calculator **146**.

Synchronously with each of the polarized probing beams **126**a and the plurality of additional beams **127**a, the component of aberration caused by aberrations in the cornea surface **131** may simultaneously be determined. Because beams **126**a and **127**a are polarized light, they will partially reflect off of the cornea surface **131** as a polarized light beam **154**a, in the case of probing beam **126**a, at an impingement point **132** and as reflected polarized light **153**a in the case of probing beam **127**a and at impingement point **133**. Because the light reflects from the cornea at an angle corresponding to the angular position of the anterior cornea surface **131** at the point of impingement **132**, reflected beam **154**a diverges from beam **126**a, depending upon the refractive characteristics of the cornea **131** at the impingement point. Beam **154** is directed by beam splitter **128** to polarizing beam splitter **140** along path **154**b. The beam **154**b passes through polarizing beam splitter **140** because it is polarized light that was reflected from the cornea surface and travels along path **154**c so that its position may be detected by a second photodetector **156**. To facilitate determination of the reflective angle of beam **154** off of the cornea, the distance from the cornea to a semitransparent scattering screen **158** is a known quantity so that the offset distance **160** of the beam **154**c impacting scattering screen **158** is indicative of the topography of the cornea surface **131**.

The scattering screen **158** causes, the light beams **154**c and **153**c, to scatter, as schematically depicted with scattering diagrams **155** or **157**. The position **160** or **159**, with respect to the optical centerline **161**, is imaged by a lens **162** onto the second photodetector **156**. Once again, the second photodetector **156** may comprise an array of x and y photodetectors using a beam divider to determine the x-y position **160** for the reflected light from the cornea. This information is provided to a cornea cause refraction calculator **164**.

The data from both total refraction calculator **146** and from **[**he**]** *the* cornea cause refraction calculator **164** is fed into a comparator **168** and also to memory **170**. The comparator information produces data, including the total refraction for each point, the cornea cause refraction for each impingement, i.e., for each shifted probing beam and may also determine the component of the refraction aberration due to components of the eye other than the cornea. From this information, a map of refractive characteristics of the eye is reconstructed in a map reconstruction unit **172**. The reconstruction map produced at **172** may be displayed at a display **174**, such as a CRT screen or a color printout. All of the total refraction calculator **146**, the cornea cause refraction calculator **164**, the comparator **168**, the memory **170**, the map reconstruction unit **172** and the display **174** may be separately provided or alternatively may be included in a computer system and display screen and/or printer schematically represented by system dash lines **148** in FIG. **1**.

FIG. **2** is an enlarged schematic view of a portion of the device for synchronous mapping of the total refraction and its component parts, better schematically depicting the paths of the probing beams **126**a and **127**a, as well as the backscatter light path **126**c, **126**d, **126**e and **126**f to a photodetector **146**. The photodetector **146** is shown comprising an x component detector **88** and a y component detector **89**. Further, the beams respectively directed or passing through the polarizing beam splitter **140** are more clearly depicted and the points of impingement on the semitransparent light scattering screen **158** are more clearly demonstrated.

The semitransparent light scattering screen **158** may, for example, be milk glass or translucent fluorescent light cover material having a substantially homogeneous characteristics so that polarized light beams impacting at any point produce the same relative intensity and same relative diffusion by which the position of such light beams may be detected with position sensor **156**. The second position sensor **156**, although not depicted, may also be constructed similarly to position sensor **146** so that x component sensor array **88** and y sensor component array **89** are used in combination to get an x-y position sensor.

FIG. **3** schematically depicts the optical channels of one embodiment of the total aberration portion of the refractometer of the subject invention. A spatially defined parallel beam input channel **59** extends from a light source such as a laser or other low diffusion light source up to the eye of a patient **98**. In one preferred embodiment a 650λ laser was employed. Along the spatially-defined parallel beam input channel is a cylindrical telescope **62** including two lenses **64** and **66**. Light from the cylindrical telescope enters the deflector **68**. The deflector **68** is preferably an acousto-optical deflector electronically controlled by a control unit such as a computer. Alternatively a galvanometric mirror deflector or the like could be used. Two coordinate deflectors or angular direction mechanisms may be used as a deflector **68**. A reflection mirror or mirror prism **70** reflects the light beam through a telescopic system **72**, including preferably, but not necessarily, a lens **74**, an entrance aperture **76**, lenses **78** and **79** and an exit aperture or field stop **80**. The polarized light beam passes from the field stop **80** to collimating lens **82** and is deflected by mirror **71** and passes transparently through beam splitter **100** and interferential beam splitter **92** en route to the eye **98**.

Light sources placed in front of the eye are used to align the visual axis of the eye with the optical axis of the instrument. Preferably a plurality of orthogonally placed light emitting diodes (LEDs) **102**, for example emitting at a λ of 940 mm could be employed. Light produced by LEDs **102** is reflected off the cornea and imaged by camera **112**. When the reflected light aligns with preset targeting parameters, the instrument is in the proper alignment and therefore in the permissive mode for firing of the spatially resolved parallel beams formed along channel **59**.

The illuminated eye is then ultimately imaged by camera **112** as the image passes through the beam splitter prism **92** and is redirected at beam splitter **100** to pass through optical elements **104**, **106**, **108** and **110** to finally fall upon the CCD camera **112**.

A retinal spot position detecting channel **99** is used to detect the position of reflected spots from the retina of eye **98** created by the input channel and includes a interferential polarization beam splitter **92** that directs non-polarized reflected light from the retina of eye **98** to a position sensor.

In one embodiment of a photodetection position sensor as shown in FIG. **3**, there is a beam splitter **94** that splits the image directing one component of the nonpolarized retina image through an optical lens **90** to a "x-coordinate" photodetector **88** and directs another component of the image through optical lens **91** to a "y-coordinate" photodetector **89**. Preferably, the orthogonally placed photo detectors **88** and **89** are high resolution linear array photodetectors and the position measurement created on those detectors may be used directly to provide XY coordinates for the measurement of the position of reflected spots on the retina of eye **98**. Instead of using linear array detectors, an actual XY matrix photo detector or a CCD detector with its own objective lens can be used to replace the beam splitter **94** lenses **90** and **91** and the linear array photodetectors **88** and **89**. One benefit of the linear arrays is that they provide for a large range of aberration detection that exceeds the range of a simple quadrant photodetector. For example, a typical quadrant photodetector may be useful for detecting aberrations of a range of about ±3 diopters while linear arrays can accommodate a range of approximately ±10 diopters. Another option is to use lateral position sensing detectors. The drawback of using a quadrant detector is in the dependence on the shape and size of the light spot imaged on its surface. Multi-element detectors like 1D or 2D arrays (linear arrays or CCD) as well as lateral position sensing detectors are free of this drawback. In yet another embodiment, the photo detector may be a 2-dimensional or x-y photodetection matrix or a CCD sensory matrix.

Details of the embodiment depicted in FIG. **3** are further explained with reference to FIG. **11** below. The position of a spot of targeting light reflected back from a reflection spot on the fovea of the retina can be determined using reflection beam splitter **94** to direct a first portion of the reflected non-polarized light from the retina spot through lens **90** to an x-direction linear array photodetector **88** for measuring changes in position only in one direction, for example in a x-direction. A second portion of the reflected **[**nonpolarzed light**]** *nonpolarized*, substantially identical to the first portion, is directed through lens **91** to a y-direction linear array photodetector **89** for detecting changes in position only in a direction at ninety degrees to the first direction, for example the y-direction. The change in the x-y position is measured by calculating the position of the center of light intensity of the light spot projected on the linear array **88** (x direction) and linear array **89** (y direction).

Light source **96** and condenser lenses **77** and **79** enable homogeneous irradiating of the linear arrays **88** and **89**, thus checking their homogeneity at servicing. Light emitting diode **96** and condenser lenses **77**, **79** form a wide beam for calibrating photodetectors **88** and **89**. If any of the elements is out of tolerance, its output can be corrected at signal processing procedures.

A fixation target channel **85** preferably comprises a light source. In a preferred embodiment the light source is a green 565 λ LED **84**. The light may be transmitted through lenses **74** and **75** and directed by prism **86** and through beam splitter **100** which has wavelength differentiating optical coatings. Fixation target is positioned on the optical element **106**. The light beam from LED **84** passes through lenses **104** and **108** and fixation target **106** and is reflected off of the mirror **110**. The fixation target light passes back through the lens **104** and is redirected by beam splitter **100** at 90 degrees out toward the eye for the patient to visualise the image as coming from the location of the surface **110** which image can be moved from near fixation to far fixation or adjustable anywhere in between and this may be used for changing the eye accommodation over a period of time and simultaneously taking a series of measurements including spatially resolved aberration refraction measurements as well as pictures on the CCD camera **112**. This produces a time lapse imaging of the eye and measurements of the aberration refraction as it cycles through different fixation target distances. The different target fixation distances may be automatically moved or adjusted from near to far using electro mechanical adjustment means that may be synchronized with the measurements and/or images taken on a time lapse basis.

The instrument described herein was developed to provide a total aberration refractometer able to accurately and quickly provide a refractive map of either emmetropic or ametropic eyes without accommodation error.

FIG. **4** shows a functional diagram of another embodiment of the subject instrument for measurement of the total aberrations in the eye and total refraction non-homogeneity. A light source whose radiation is used for the ray tracing of the eye is provided, as for example, by laser **1**. A telescopic expander comprising for example lenses **2** and **3** provides a normal functioning of a two-coordinate acousto-optic deflector **4** preferably consisting of two single-coordinate deflectors. A telescopic laser-beam narrower is formed by lenses **5** and **6** with an aperture stop or diaphragm AD located at the common foci of the lenses **5** and **6**. A field stop or diaphragm **7** is placed at the back focus of lens **6** so that its image formed by the telescopic narrower in the back-pass is located between the single-coordinate deflectors. With this placement, the redistribution of the light illuminance in the light spot on the pupil is minimized when the angular position of the laser beam is varied at the exit of the single-coordinate deflectors. The front focus of a collimating lens **8** is aligned with the center of the field stop or diaphragm **7** to ensure telocentric passage of rays through interferential polarizing beam splitter **9**.

An ametropia compensator is schematically depicted as a varifocal group of lenses **10** and **11**, adjustable to compensate for the patient's eye ametropia. One of the lenses is mounted on a movable base connected to actuator drive **38**. An accommodation controller is schematically depicted as lenses **16** and **17** that constitute a varifocal group of lenses for accommodation control of the patients eye.

An objective lens **18**, at whose focal point the photosensitive surface of a position-sensitive photodetector **19** is located, is intended to form an image of the irradiated retina in the plane of the photosensitive surface of the position-sensitive photodetector. The photosensitive elements of the photodetector are connected through a preamplifier **22** and an analog-to-digit converter **23** to a computer **24**. A beam coupler **39** is movably mounted between the objective lens **18** and the photodetector **19** to optically conjugate the plane of the test-target or plate **20** with the photosensitive surface of the first photodetector **19** as well as with the fovea surface. The plate **20** is needed to ensure the fixation of the patient's gaze. Located behind the plate **20** is a light source or radiator **21** serving to illuminate the plate.

Elements **25** through **30** comprise a microscope whose objective lens consists of lenses **25** and **27** together with mirror **26**. A plate **29** with first coordinate-grid is preferably located at the back focal plane of a lens **27**. A lens or a group of lenses **30**, the front focal point of which coincides with the back focal point of the lens **27**, comprises an eyepiece of the microscope. The beam splitter **28** serves to optically couple the retinal plane with the photosensitive plane of a TV camera **32** connected to the computer through a video signal conversion and input board, alternatively termed a frame grabber board, **33**.

By means of a mirror **12** provided with an opening, the optical axis of the microscope is aligned with the optical axes of the ray tracing channel (elements **1-11**) and the photoelectric arrangement for measuring the transverse ray aberration on the retina (elements **16-19**).

In a preferred embodiment, four light-emitting diodes (LEDs) **14** are installed in a cross-wise configuration in front of the patient's eye. Each LED is preferably located in the same plane as each other LED, at an equal distance from the optical axis and perpendicular with the axis. The microscope and the LEDs comprise a system for the visual and television positioning of the instrument relative to the patient's eye. The microscope is installed so that the front focal plane of lens **25** coincides with the plane, where imaginary or virtual images of the LEDs **14**, mirrored by the anterior corneal surface, are located.

Before the total refraction measurement process is commenced, the instrument is positioned relative to the patient's eye and the instrument is calibrated using the optical calibration unit **34-36**. Movably mounted between the lens **11** and the LEDs **14** is a mirror **13** which serves to join the optical axes of the instrument and the optical calibration unit **34-36**. In one preferred embodiment of an optical calibration unit, it comprises a meniscus or cornea simulator **34**, liquid medium or vitreous simulator **35**, and retina simulator **36**. The retina simulator **36** is preferably movably mounted so that it can be moved along the optical axis by means of actuator or drive **37**.

The instant measuring instrument incorporates a computer **24** or like device for controlling the acousto-optic deflector **4**, analog-to-digital converter **23**, and actuators or drives **37** and **38**. The computer **24** or like device or devices may perform additional duties including, for example, mathematical processing and data storage, calculation and display of aberration parameters and the ocular refraction characteristics as well as provide setting measurement modes and implementation of automatic instrument alignment.

The instrument for measurement of the total eye refraction, in its preferred embodiment, functions in the following way. The light beam emitted, for example by laser **1**, is expanded, collimated and directed to the acousto-optic deflector **4**, which changes its angular position in accordance with the corresponding computer program. The telescopic narrower **5** and **6** reduces the beam thickness to the requisite magnitude. The center of the stop or diaphragm **7** is a point of angular "swinging" of the beam exiting from the telescopic narrower. Due to its positioning in the front focal plane of the lens **6**, the aperture stop or diaphragm AD has its image in the back focal plane of the lens **8** which is aligned with the eye pupil. Further, because the stop or diaphragm **7** is positioned in the front focal plane of the collimating lens **8**, angular swinging of the laser beam with the angle vertex located on the stop or diaphragm **7** is converted into parallel shifting of its optical axis after passing the lens **8**.

If the patient's eye is ametropic, the axial movement of the lens **10** (or **11**) converts the telocentric beam into a beam which diverges (in the case of myopia), or converges (in the event of hyperopia), so that the image of the diaphragm **7** is optically conjugated with the retina. This also ensures parallelism of the rays reflected by the retina in the zone in front of the beam splitter **9**, which is necessary for its normal functioning.

The light entering the eye **15** of the patient is polarized in the plane shown in FIG. **4**. Only that component of the returning beam depolarized by interaction with the retina is allowed by the beam splitter **9** to pass to the first photodetector **19**. This protects the first photodetector from the polarized light reflected by the surfaces of the lenses **10** and **11** and by the cornea or the eye and which can produce an illuminance incompatible with determining the total refraction according to normal functioning of the instrument.

Lenses **16** and **17** and the objective lens **18** produce an image of the illuminated area of the retina in the plane of the first photodetector **19**. In FIG. **4** the foci locations are designated as follows: F_3, F_5, F_6, F_8, F_{25}, and F_{30}, designating points of front foci of the corresponding lenses while F_2', F_5', F_6', F_{18}', and F_{27}', designating points of back foci of the lenses.

In one more embodiment, presented in FIG. **5**, the function of ametropia compensator is combined in the component **18** that is an objective lens for the photodetector **19**. In this embodiment, target object **20** and photodetector **19** are positioned at equal distances from the lens **18**. LED **21** irradiates the target object **20**. The lens **18** is positioned by the patient in such a way that a clear image of target object **20** is seen. In this position, focal planes of the photodetector **19** and the eye **15** are conjugated. Positioning of the lens **18** can be implemented with the electric drive **176**. This positioning can be done automatically. Accommodation control is executed with another target object **178**, irradiated by the LED **179**, and positioned by an electronic drive **180**. Both drives, **176** and **180**, are connected with computer **24**. Accommodation target object **178** is coupled with optical axis by the mirror **177**. Colors of LEDs **21** and **179** should be different for their easy identification. For example, conjugation is made with red LED **21**, and accommodation follows with green LED **179**.

Still another embodiment of the invention, schematically shown in FIG. **6** and based generally on the configuration presented in the FIG. **5**, enables simultaneous investigation of both eyes **15** and **15**a. According to the design shown in FIG. **6** this embodiment beneficially contains an additional channel for the second eye **15**a. The implementation is such that the common laser probing is used for both eyes due to two beam splitters **175** and **181**. Optical and electrical components **9**a, **14**a, **18**a, **19**a, **22**a, and **23**a have the same roles as the corresponding components **9**, **14**, **18**, **19**, **22**, and **23** at measurement of the first eye **15**. Control unit **182** switches the infrared LEDs **14** and **14**a to alternatively get irradiated first and second eye **15** and **15**a respectively. In this way, images of the eyes can be displayed simultaneously, for example, on the left and right parts of the monitor's screen. Described implementation enables separate adjustment and advantageously provides simultaneous measurements on both eyes.

In the various embodiments of FIGS. **4**, **5**, and **6** the laser beam is positioned by the computer and the acousto-optic deflector so as to enter the pupil within the requisite refraction measurement zone. If the optical system of the eye has aberration refraction, the light image of the stop or diaphragm **7** on the retina is displaced from the axis, which results in the corresponding displacement of the illuminated zone image on the photosensitive surface of the position-sensing photodetector **19**.

If photodetector **19** is a four-quadrant photodiode, as, for example, that shown diagrammatically in FIG. **7**, having quadrants **1**, **2**, **3** and **4**, an aberration displacement of the light spot δx, δy on the retina can be given by the formula:

$$\delta x = \frac{\beta}{2}\left[\frac{(U_1 + U_4) - (U_2 + U_3)}{U_1 + U_2 + U_3 + U_4}\right]\cdot b,$$

$$\delta y = \frac{\beta}{2}\left[\frac{(U_1 + U_2) - (U_3 + U_4)}{U_1 + U_2 + U_3 + U_4}\right]\cdot b,$$

where β is the transverse magnification in the plane of the first photodetector as related to the plane of the retina, b is a coefficient depending on the size of the light spot in the plane of the photodetector, and U_1, U_2, U_3 and U_4 are the photoelectric signals coming from the corresponding photodetector channels.

If photodetector **19** is a lateral position sensing detector, as shown in FIG. **8**, having a pair of x-direction electrodes **1** and **2**, and a pair of Y-direction electrodes **3** and **4**, an aberration displacement of the light spot δx, δy on the retina can be described as follows:

$$\delta x = \beta\left[\frac{U_1 - U_2}{U_1 + U_2}\right]\cdot a,$$

$$\delta y = \beta\left[\frac{U_3 - U_4}{U_3 + U_4}\right]\cdot a,$$

where β is the transverse magnification between the planes of photodetector and retina, U_1, U_2, U_3 and U_4 are the signals coming from the electrodes, **1**, **2**, **3** and **4** correspondingly, and a is a scaling coefficient depending on the electrical parameters of the lateral detector.

The principle of operation relating to the positioning the instrument in relation to the patient's eye is illustrated in FIG. **9**. The collimating system **50** corresponds to the elements **39**, **18** and **17**, **16** of FIG. **4** (if the eye is accommodated at a finite distance). Point A is the light radiator and gaze fixation point and is formed by the elements **20** and **21** in FIG. **4**. The mirror **56** with an opening corresponds to element **12** of FIG. **4**. The microscope objective lens **52** is comprised of the elements **25**-**27** of FIG. **4** while the microscope objective image plane **54** (FIG. **4**) corresponds to the element **29** (FIG. **4**). B_1 and B_2 are light radiators corresponding to the LED **14** of FIG. **4**. B'_1 and B'_2 are primary images of the radiators while B''_1 and B''_2 are secondary images of the radiators.

As can be seen from FIG. **9**, the fixation of the gaze on the point A, located on the optical axis of the instrument, does not guarantee the coincidence or alignment of the visual axis of the eye and the optical axis of the instrument because the eye sees the point A on the fovea even when $\Delta \neq 0$. The fixation of the gaze on the point A is ensured only when the above axes are parallel.

Taking into account that the largest contribution to the optical power of the eye is made by the anterior surface of the cornea, the visual axis line is assumed to be the line passing through the fovea center and the vertex of center of curvature of the front surface of the cornea. If the radiator B_1 is positioned in front of the patient's eye, then, due to reflection of the light from the anterior or front surface of the cornea, this surface functioning as a convex mirror, forms an imaginary or virtual image B'_1 of the radiator, located symmetrically to the axis in accordance with the laws of geometric optics.

When several radiators, such as for example, B_1 and B_2, are positioned in front of the patient's eye symmetrically to the optical axis of the instrument (FIG. **9**), their secondary images B''_1 and B''_2 will be shifted in the image plane of the microscope objective lens aside from the axis if $\Delta \neq 0$.

Thus, to make the optical axis of the instrument and the visual axis of the eye coincide, two conditions must be satisfied: the patient's gaze is fixed on the point A and the images B''_1 and B''_2 are centrally positioned in relation to the axis of the objective lens **52**. The positioning can be checked using the coordinate grid provided on the plate **29** (FIG. **1**) or using the monitor screen when the TV channel is utilized. When the image of the eye is aligned at all points with concentric locations on the grid or the TV screen, the measurement controller is armed for taking a spatially resolved set of refraction measurements. The operator can than activate the measurement that can take only a few milliseconds. The measurements are "grabbed" in the grabber board and stored for producing an aberration refraction map as in FIG. **12**. The measurements can also be activated automatically when the proper alignment is detected. Further, according to one embodiment of the invention, a plurality of measurements can be made sequentially during the occurrence of a predetermined event, such as through a sequence of movement of the eye target from a "near" accommodation distance to a "far" or infinity accommodation distance. A plurality of measurement images can be captured or automatically grabbed and stored over a time period or while any other changes are occurring for which eye measurements might indicate a dynamic change in the refraction of the eye.

The coincidence of the points B''_1 and B''_2 with the surface or plane **54** is indicative of setting the fixed working distance between the instrument and the eye which is the result of the focusing of the images B''_1 and B''_2 on the surface **54**.

The point of gaze fixation is created by locating the mirror **39** (FIG. **4**) on the optical axis of the instrument. The radiators **14'** and **14"** play the part of the radiators B_1 and B_2 shown in FIG. **9**.

Another embodiment of eye instrument alignment can be implemented using manually or automatically operated measurement of the pupil edges; forming a figure, approximately a circle. Its center does not coincide usually with the center of symmetry of four reflexes, two of which B''_1 and B''_2 are shown in FIG. **9**. This non-coincidence can be taken into consideration at further signal processing.

The calibration of the instant aberration refraction instrument may be effected using the optical calibration unit. The optical calibration unit can be made to incorporate known aberrations at the corresponding cornea simulator **34** measurement points. For example, the aberration may be determined by the computer using special optical design programs. If, for example, the front surface of the lens **34** is ellipsoidal, then the aberration refraction at all the points of the pupil is equal to zero.

When an ametropy compensator is used, nonparallel laser beams will enter the optical calibration unit. This will result in a standard aberration of defocusing; to compensate for this aberration, the retina simulator can be moved along the optical axis by means of the actuator **37** to the focus point. Thereby, the fovea can be optically conjugated with the retina simulator.

Systematic errors of measurements of the transverse aberration will be evidenced by the deviation of the measurement results from the estimated data. Such determinable systematic errors can be taken into account when measuring actual total ocular aberrations.

The calibration by comparison with the optical calibration unit is preferably performed automatically before measuring the ocular aberrations by locating the mirror **13** on the optical axis of the instrument.

Prior to the ray tracing of the patient's eye the mirrors **13** and **39** are withdrawn from the light path entering the eye and then the light passes to the photodetector. The aberration displacement of the image of the light spot on the fovea is measured at a set of points on the cornea corresponding to an ocular ray tracing grid chosen by the operator. An example of a grid or an allocation of measurement points on the pupil is shown in FIG. **10**.

The data on measurement of the transverse aberrations on the retina $\delta x(\rho, \phi)$ and $\delta y(\rho, \phi)$ are used for further calculations of the coefficients of the Zernike polynomials by means of the least squares method in order to approximate the function of the total wave aberration of the eye. The wave aberration function is then used to calculate the local total refraction at any point of the pupil. In addition, the approximation makes it possible to determine or reconstruct the nature of local aberration refraction in that small axial zone of the pupil, where it is impossible make accurate direct measurement of refraction.

In one experiment conducted using this instrument in which five replicate tests were performed and the results averaged, the laser beam total aberration on the retina at 65 points of the pupil was been performed in within 12 milliseconds with no more that 5 mW of light radiation entering the eye.

FIG. **12** is an example of an ocular aberration refraction map constructed with the use of the total eye aberration refractometer portion of the invention.

The extremely fast measurement permits the computer control program to cause a plurality or spatially resolved aberration measurements to be made in a very short period of time. The control program in one embodiment automatically activates a plurality of measurements coordinated with a series of adjusted accommodation fixation distances and automatic determination of proper eye alignment to receive a series of data measurements from the retinal spot position detecting channel. A series of refraction measurements for a dynamic eye refraction system is produced. Spatially resolved refraction measurements can be automatically programed and automatically made during a variety of dynamic changes such as varying accommodation or during normal functioning of the eye under a variety of predetermined conditions and internal or external changing conditions.

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While this invention has been described with reference to illustrative embodiments, this description is not intended to be construed in a limiting sense. Various modifications and combinations of illustrative embodiments, as well as other embodiments of the invention, will be apparent to persons skilled in the art upon reference to the description. It is therefore intended that the appended claims encompass such modifications and enhancements.

Other alterations and modifications of the invention will likewise become apparent to those of ordinary skill in the art upon reading the present disclosure, and it is intended that the scope of the invention disclosed herein be limited only by the broadest interpretation of the appended claims to which the inventors are legally entitled.

What is claimed is:

1. A device for measuring aberration refraction of an eye, comprising: a polarized light source producing a probing beam along a path to the eye; a telescopic system of lenses having an entrance pupil and an exit pupil; a two-coordinate deflector consisting of two single-coordinate deflectors separated by a zone; a deflection control unit; an aperture stop; a field stop; a collimating lens; a first position-sensitive photodetector with an objective lens operatively positioned for receiving non-polarized light reflected from the retina of the eye, a second position-sensitive photodetector for simultaneously receiving polarized light reflected from the cornea of the eye; and a data processing and display unit including a computer coupled to said first and second photodetectors for calculating total eye refraction aberration and the component caused by cornea aberration.

2. The measuring device of claim **1** further comprising an interferential polarization beam splitter positioned along the path to the eye between the collimating lens and the eye for directing non-polarized light backscattered from said retina to said first position-sensitive photodetector and for directing polarized light reflected from said cornea to said second position-sensitive photodetector.

3. The measuring device of claim **2**, wherein the light source producing the probing beam comprises a laser light, the laser light probing beam being directed by said interferential polarization beam splitter and separately directs non-polarized light as scattered back from the retina to said photodetector and polarized light reflected from the anterior surface of the cornea to said second photodetector.

4. The measuring device recited in claim **3**, wherein, between the interferential polarization beam splitter and the eye in the path of the probing beam a group of lenses with a variable optical power is installed, forming a retina image, focused at infinity for the ametropic eye, and the photosensitive surface of the photodetector is coincident with the front focal plane of the objective lens and is located behind the interferential polarization beam splitter in the path of light scattered by the retina.

5. The measuring device recited in claim **3**, further comprising an optical axis bending mirror and a plate with a gaze fixation test pattern optically coupled with the photosensitive photodetector and located between the photodetector and the objective lens and another varifocal group of lenses with a variable optical power, serving to form the test pattern image visible to the eye at a distance corresponding to a variably selected accommodation, another varifocal group of lenses is positioned between the optical axis bending mirror and the interferential polarization beam splitter and the optical axis bending mirror being mounted on a movable base making it possible to displace the mirror so as to enable the light radiation scattered by the retina to reach the photodetector during the measurement of the patients eye characteristics.

6. The measuring device recited in claim **3**, wherein another optical axis bending mirror is located in the laser beam path after the last optical element, another mirror being followed by an optical calibration unit with a retina simulator whose optical characteristics are equivalent to those of the human eye retina.

7. The measuring device recited in claim **6**, wherein the another optical axis bending mirror is mounted on a movable base so that it can be located in the probing laser beam path for measuring the retina standard characteristics and can be withdrawn therefrom before measuring eye aberration refraction.

8. The measuring device recited in claim **6**, wherein the another optical axis bending mirror further comprises an optical coating which permits selective transmission of light such that it needs not be removed from the probing laser beam path when measuring eye aberration refraction.

9. The measuring device recited in claim **3**, wherein there is an additional beam splitter leading to a channel for verifying the alignment of the device with the eye.

10. The measuring device of claim **9**, wherein the channel for verifying the alignment of the device and the patients eye comprises one or more point light source and a TV or optoelectronic photodetector serving to display an image of the eye and produce a signal enabling the measurement of the ocular characteristics when the optical axis of the measuring device and the visual axis of the patients eye coincide.

11. The measuring device of claim **10**, wherein the channel produces an automatic permission signal via an electronic circuitry or a computer electronically linked to said channel contains a program recognizing said automatic permission signal and directing measurement of the ocular characteristics directly upon achieving preset or derived alignment criteria according to the permission signal.

12. An instrument for measuring aberration refraction of an eye having a visual axis, the instrument comprising: a lens system defining an instrument optical axis; an alignment device for aligning the visual axis of the eye with the instrument optical axis; a light source producing a probing beam projected through the lens system parallel to the instrument optical axis and selectably positionable partially off-set from the instrument optical axis so that said probing beam **[**imates**]** *impacts* upon the eye parallel to the instrument optical axis at a plurality of impingement locations on the cornea of the eye; a first photodetector for measuring the position of probing beam light scattered back from the retina of the eye to measure aberration refraction of the eye at plurality of locations corresponding to each of said plurality of impingement locations; a second photodetector for measuring the position of probing light beam reflected from the cornea at said plurality of impingement locations, and a computer for receiving input from said first and second photodetectors to calculate said total refraction aberration and said component thereof due to said cornea.

13. An instrument for measuring aberration refraction of an eye according to claim **12** wherein said first and said second photodetectors are selected from the group comprising: a quadrant photodetector, lateral position sensing detector, orthogonally placed linear array photodetectors, XY matrix photodetectors, or CCD detector.

14. An instrument for measuring aberration refraction of an eye as in claim **12** wherein said polarized light comprises laser light and further comprising an interferential polarization beam splitter positioned along said probing beam path for directing the probing beam polarized laser light into the eye and positioned along the visual axis of the eye between the eye and said first and said second photodetectors for permitting only non-polarized light scattered back from the retina of the eye to be detected by said first photodetector and for permitting only said polarized light reflected from said cornea to be detected by said second photodetector.

15. An instrument for measuring aberration refraction of an eye as in claim **12** further comprising: an illumination source producing a target image collimated and projected to the eye along and coincident with the instrument optical axis for fixing the gaze of the eye at a predetermined image focus distance and for aligning the visual axis of the eye coincident to said instrument axis and a varifocal telescopic lens group adjustable to regulate eye accommodation at a fixed focal distance for the target image.

16. An instrument for measuring aberration refraction of an eye as in claim **12** further comprising: a varifocal ametropia compensator including an adjustable telescopic lens system for focusing the probing beam to compensate for ametropia of the eye.

17. An instrument for measuring aberration refraction of an eye as in claim **16** further comprising: a calibration unit for calibration of the instrument to correct for lens system induced errors and to correct for errors induced by the adjustment of the varifocal telescopic lens group to compensate for eye ametropia.

18. An aberration refractometer comprising:

a) an optical beam input channel which generates parallel light rays along an optical axis at spatially defined points onto an eye of a patient, either simultaneously or sequentially, including at least one light source and at least one lens;

b) an eye alignment verification channel for confirming alignment of the visual axis of the eye as the patient maintains fixation with the optical axis of the input channel;

c) a fixation target channel including at least one visual target which can be adjusted manually or in an automatic fashion to simulate different focal distances and which is in optical communication with the eye;

d) a retinal spot position-detecting channel in optical communication with the eye including at least one first photodetector for recording at least one characteristic of a first portion of input channel light from at least one of said spatially defined points and scattered back from the retina of the eye; and

e) a cornea reflection position detecting channel in optical communication with the eye including at least one second photodetector for recording at least one characteristic of a second portion of input channel light from said at least one of said spatially defined points and reflected back from said cornea.

19. The aberration refractometer of claim **18** further comprising a computer, electronic circuitry and a program for monitoring alignment of the visual axis of the eye and the input channel and automatically directing the generation of probing light rays upon eye alignment verification.

20. The aberration refractometer as in claim **18** further comprising computer electronic circuitry for receiving position data from said first and second photodetectors and for comparing said data for each of said plurality of impingement points for determining the total eye refraction aberration and the components due to non-homogeneity of the cornea.

21. The aberration refractometer as in claim **18** further comprising computer electronic circuitry for producing a spatially resolved map for all of said spatially defined impingement points.

22. A method for synchronous mapping of the total refraction non-homogeneity of the eye and selected refractive components of the total refraction non-homogeneity, said method comprising steps of:

a) directing a measuring beam of light into said eye, the cross-section of beam being narrower than the entrance aperture of said eye, said beam being centered and its axis coinciding with the visual axis of said eye, said beam having a first portion thereof passing through an anterior surface of said eye to impinge upon the cornea of said eye, as well as having a second portion thereof reflecting from the corneal surface;

b) changing the position of a measuring point of beam entrance into said eye keeping said beam parallel to said visual axis;

c) measuring an offset between the location on the retina at which said beam impinges on the retina at the central position of said beam and the changed position of said measuring point;

d) determining the total refractive characteristic of said eye from the measured offset in step **22**(c);

e) determining the angle of reflection of said second portion of said measurable beam reflected from said cornea;

f) determining the partial refractive characteristic, caused by said cornea, from the determined angle of reflection in step **22**(e);

g) comparing, in each measuring point, the total refractive characteristic determined in step **22**(d) with the partial refractive characteristic determined in step **22**(c) by means of subtraction of said partial characteristic from said total refractive characteristic;

h) storing refractive characteristics determined in steps **22**(d), **22**(f) and **22**(g) for each measuring point;

i) transforming the discrete sets of said refractive characteristics into continuous surfaces by means of approximation; and

j) presenting said surfaces as color coded charts on the screen of a display or in any other way of presentation of the three-dimensional information.

23. The method recited in claim **22** wherein the step of determining the angle of reflection from the **[**comer**]** *cornea* further comprises the steps of:

a) projecting the reflected part of said beam from the cornea at said measuring point onto a semitransparent light scattering screen positioned out of the zone of said entrance aperture of said eye; and

b) measuring the coordinates of the location of the beam projection on said screen.

24. The method recited in claim **22** further comprising:

a) limiting a spatial zone of determining mean value of refraction of said eye to a cross-sectional area of said eye less than or equal to the cross-sectional area of said entrance aperture of said eye; and

b) determining the zonal refraction by averaging the sets of values determined in steps **22**(d), **22**(f), or **22**(g) for the measuring points inside said spatial zone.

25. The method recited in claim **22** further comprising:

a) comparing the refraction characteristic in each measuring point determined in steps **22**(d), or **22**(g) with the corresponding refraction characteristic for the neighbor measuring point by means of extracting the value of the refractive characteristic in said measuring point from the value of the refractive characteristic in the measuring point neighbor to said measuring point;

b) inserting an additional measuring point, located between the neighboring measuring points used for the comparison in step **25**(a), if the difference value determined in step **25**(a) is higher than a specified threshold; and

c) repeating the steps **22**(a)-**22**(k) until there is no more value determined in step **25**(a) higher than said specified threshold.

26. A device for synchronous mapping of the total refraction non-homogeneity of the eye and its refractive components comprising:

a) means for providing a narrow beam of light to be directed into said eye;

b) means for shifting said beam over the entrance aperture of said eye in parallel to itself and positioning said beam in a measuring point;

c) means for detecting the light back-scattered from the retina of said eye;

d) means for measuring an offset between the location on the retina at which said beam impinges on said retina for the first position when the beam's axis coincides with the visual axis of said eye and for the second position when the beam is displaced to said measuring point;

e) means for calculating the total refractive characteristic of said eye in said measuring point from the measured offset by means of **26**(d);

f) a semitransparent light-scattering screen inserted in outer zone of said entrance aperture;

g) means for detecting the light reflected from the cornea of said eye and projected on said semitransparent light-scattering screen;

h) means for measuring the coordinates of the location of said beam reflected from said cornea in said measuring point and projected on said screen;

i) means for calculating the partial refraction characteristic of said eye in said measuring point, caused by said cornea, from the coordinates measured by means of **26**(h);

j) means for calculating the difference between the values of said total refractive characteristic calculated by means of **26**(d) and the value of said partial refractive characteristic caused by said cornea and calculated by means of **26**(i);

k) means for storing refractive characteristics calculated by means of **26**(d), **26**(i), and **26**(j) for each measuring point;

l) means for transforming the discrete sets of said refractive characteristics into continues three-dimensional distributions of said refractive characteristics; and

m) means for displaying said three-dimensional distributions of said refractive characteristics.

27. The device recited in claim **26** comprising means for calculating the averaged values of refraction within a spatially limited zone of the entrance aperture of said eye, said zone being smaller or equal to said entrance aperture of said eye.

28. The device recited in claim **26** comprising:

a) means for calculating the absolute difference between the values of the same refractive characteristic in each pair of neighboring measuring points; and

b) means for producing a control command for repeating the measurements of said refractive characteristic in an additional measuring point inserted between said neighboring points.

29. The device recited in claim **26** comprising said screen having the shape of the figure of rotation of straight or curved line about the optical axis of said beam in the central position of said beam.

30. A device for measuring aberration refraction of both of a patient's eyes, comprising: a polarized light source producing a probing beam along paths of a patient's eyes; a telescopic system of lenses having an entrance pupil and an exit pupil; a two-coordinate deflector consisting of two single-coordinate deflectors separated by a zone; a deflection control unit; an aperture stop; a field stop; a collimating lens; a first position-sensitive photodetector with an objective lens operatively positioned for receiving non-polarized light reflected from the retina of both eyes, one eye at a time, a second position-sensitive photodetector for simultaneously and selectively receiving polarized light reflected from the cornea of both eyes, one eye at a time, coordinated to receive polarized light from the cornea of the same eye as the non-polarized light is received from the retina; and a data processing and display unit including a computer coupled to said first and second photodetectors for calculating total eye refraction aberration and the component caused by cornea aberration separately and substantially simultaneously for both of the patient's eyes.

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