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(54) **Optical coherence tomography method and optical coherence tomography apparatus**

Optisches Kohärenztomographieverfahren und optische Kohärenztomographievorrichtung

Procédé de tomographie à cohérence optique et appareil de tomographie à cohérence optique

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(56) References cited:
DE-T5-112006 003 228 US-A1- 2006 089 548

- **JON HOLMES, SIMON HATTERSLEY, NICK STONE, FLORIAN BAZANT-HEGEMARK AND HUGH BARR: "Multi-channel Fourier domain OCT system with superior lateral resolution for biomedical applications" PROC. SPIE, vol. 6847, 18 February 2008 (2008-02-18), pages 68470O-1-68470O-9, XP007915204 DOI: 10.1117/12.761655**
- **FABER D J ET AL: "Quantitative measurement of attenuation coefficients of weakly scattering media using optical coherence tomography" OPTICS EXPRESS, OSA (OPTICAL SOCIETY OF AMERICA), WASHINGTON DC, (US) LNKD- DOI: 10.1364/OPEX.12.004353, vol. 12, no. 19, 20 September 2004 (2004-09-20), pages 4353-4365, XP002446372 ISSN: 1094-4087**

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Description

[0001] The present invention relates to an optical coherence tomography method and an optical coherence tomography apparatus, and more particularly to an optical coherence tomography method and an optical coherence tomography apparatus, using a coherent optical system for use in the medical field.

[0002] Currently, there are a wide variety of ophthalmic devices using optical devices. Examples of such ophthalmic devices include anterior eye imaging apparatuses, retinal cameras and scanning laser ophthalmoscopes (SLOs). Among them, optical coherence tomography (OCT) apparatuses can obtain tomographic images of objects to be measured at high resolution, and therefore are becoming indispensable devices for outpatient medical treatments specialized for retina.

[0003] An OCT apparatus is disclosed, for example, in Japanese Patent Application Laid-Open No. H11-325849. In an OCT apparatus disclosed in Japanese Patent Application Laid-Open No. H11-325849, low coherent light is used. Light from a light source is divided into measurement light and reference light through a split optical path, such as a beam splitter. The measurement light is applied onto an object to be measured, such as a human eye, through a measurement optical path, and return light from the object to be measured is led to a detection position through a detection optical path. The return light as used herein refers to reflected light or scattered light that includes information on an interface of the object to be measured with respect to the irradiation direction of light. The reference light is led to a detection position through a reference optical path. Input to a detection position is coherent light resulting from interference between the return light and the reference light. Then, the wavelength spectrum of the coherent light is collectively acquired by the use of a spectrometer or the like, and the wavelength spectrum is Fourier transformed, thereby obtaining a tomographic image of the object to be measured. In general, an OCT apparatus that collectively measures the wavelength spectrum is termed a spectral-domain OCT (SD-OCT) apparatus.

[0004] With an SD-OCT apparatus, the depth of focus and a transversal resolution can be adjusted by selecting a numerical aperture (NA) of a lens used for controlling a focusing position of the measurement light in an object to be measured. For example, the larger the numerical aperture is, the smaller the depth of focus is, but the higher the transversal resolution is. On the other hand, if the numerical aperture is reduced, the depth of focus becomes larger, but the transversal resolution becomes lower. In other words, the relationship between the depth of focus and the transversal resolution is a trade-off.

[0005] As a method that overcomes this relationship, dynamic focus OCT is disclosed in "OPTICS LETTERS Vol. 28, 2003, pp. 182-184". In this mode, time domain OCT (TD-OCT) that acquires a tomographic image while changing an optical path length is employed. Then, a tomographic image is acquired while changing the optical path length and moving the focus position of a lens in synchronization with each other. As a result, while the transversal resolution is maintained high, the measurement range of an object to be measured (the range in the irradiation direction of measurement light in an acquired tomographic image) can be increased.

[0006] Further prior art can be found in document "Multi-channel Fourier domain OCT system with superior lateral resolution for biomedical applications" by J. Holmes et al. (XPXP007915204), in document US 2006/089548 A1, disclosing "Correlation of concurrent non-invasively acquired signals", in document DE 11 2006 003 228 T5 disclosing a dental optical coherence tomograph, and in non-patent literature D.J. Faber et al., Optics Express, vol. 12, no. 19, pages 4353-4365, disclosing quantitative measurement of attenuation coefficients of weakly scattering media using optical coherence tomography.

[0007] In TD-OCT, however, measurement is performed while consecutively changing the optical path length. Therefore, it takes more time to acquire (measure) a tomographic image with TD-OCT than with SD-OCT. In order to achieve high-speed acquisition of a tomographic image having a large measurement range of an object to be measured and a high transversal resolution, a method of performing dynamic focusing in a spectral-domain mode is considered. As described above, in the spectral-domain mode, as the transversal resolution increases, the depth of focus decreases. Accordingly, to increase the measurement range, an object to be measured needs to be divided into a plurality of measurement regions adjacent to one another along the irradiation direction of measurement light for the purpose of measurement.

[0008] In SD-OCT, a phenomenon as illustrated in FIG. 7 occurs. FIG. 7 illustrates a relationship of a distance between a coherence gate and a mirror for horizontal axis and a measured intensity for vertical axis (intensity of light; reflected intensity) in the case of using the mirror as an object to be measured. Specifically, intensities (digital values) measured when the position of the mirror is distant from the coherence gate by 50, 100, 150, 200, 300, 400, 500, 600, 800, 1000, 1200, 1600 and 2000 μm are shown. Note that the term "coherence gate" refers to a position that is in the measurement optical path and that has the same optical distance as that of the reference optical path. The dotted line schematically shows the envelop of their results (changes in intensity with respect to the position in the irradiation direction in the measurement region), which is a so-called attenuation function. In FIG. 7, as the position of the mirror is more distant from the coherence gate, the intensity attenuates more. This is called "roll-off" or the like, and occurs because of the resolution of a spectrometer and so on.

[0009] As described above, in the case of the phenomenon occurring, the intensity is stronger as the position is closer to the coherence gate whereas the intensity is weaker as the position is more distant from the coherence gate. Therefore, at a boundary of measurement regions, the intensity is strong in one region whereas the intensity is weak in the other region. This causes a jump in measured intensity between regions adjacent to each other.

[0010] Accordingly, an object of the invention is to provide an optical coherence tomography method and an optical coherence tomography apparatus, that can continuously join a tomographic image acquired from each of a plurality of measurement regions.

[0011] The above object is solved by what is defined in the appended independent claims. Advantageous modifications thereof are set forth in the appended dependent claims.

[0012] According to some aspects of the invention, there can be provided an optical coherence tomography method an optical coherence tomography apparatus, that can continuously join a tomographic image acquired from each of a plurality of measurement regions.

[0013] Further features of the present invention will become apparent from the following description of exemplary embodiments with reference to the attached drawings.

FIG. 1 is a flow chart illustrating an optical coherence tomography method according to this embodiment.
FIG. 2 illustrates a configuration of a Mach-Zehnder interference system used in an OCT apparatus according to this example.
FIG. 3 illustrates widths of measurement regions.
FIG. 4A illustrates an ideal tomographic image of an object to be measured.
FIG. 4B illustrates mirror images reflected in measurement regions.
FIG. 4C illustrates a correction image of each measurement region.
FIG. 4D illustrates a calculated real image of each measurement region.
FIG. 5 is a flow chart illustrating a method of analyzing measurement image data.
FIG. 6 illustrates a method of image adjustments of real images.
FIG. 7 illustrates a relationship of a distance between the coherence gate and a mirror and a measured intensity when the mirror is used as an object to be measured.

[0014] An optical coherence tomography apparatus according to this embodiment will be described below.

[0015] The optical coherence tomography apparatus according to the embodiment divides light from a light source into measurement light and reference light through a split optical path. The measurement light is irradiated through a measurement optical path onto an object to be measured. Return light returning from the object to be measured upon irradiation of the measurement light is led through a detection optical path to a detection position. The focus position of the measurement light in the object to be measured (irradiation direction) can be controlled by a focus drive mechanism. The reference light is led through a reference optical path to a detection position. In the reference optical path, a mirror is disposed, and the position of the coherence gate can be adjusted by a mirror drive mechanism. Since the coherence gate and the focus position can be controlled in synchronization with each other, it is possible to divide the object to be measured into a plurality of measurement regions adjacent to one another along the irradiation direction and sequentially perform measurement for every region. Light led to the detection position (coherent light of the return light and the reference light) is resolved into its wavelength spectrum and is analyzed. Thus, a tomographic image of the object to be measured is acquired.

[0016] When the object to be measured is divided into a plurality of measurement regions, a situation in which the coherence gate needs to be arranged in the interior of the object to be measured occurs. Since the coherence gate refers to a position in the measurement optical path that has the same optical distance as that of the reference optical path, images that reflect each other are formed in adjacent regions across the coherence gate. The two images are equivalent, and therefore either of them may be employed for a tomographic image. Hereinafter, an image to be acquired (i.e., an image employed as the tomographic image in the region) is referred to as a "real image", and the other image is referred to as a "mirror image". In the case of adopting the SD-OCT mode, an image (measurement image) represented by coherent light includes a real image and a mirror image, and therefore separating the real image from the mirror image is indispensable.

[0017] With reference to FIG. 1, an optical coherence tomography method according to this embodiment is described. In the embodiment, an object to be measured is divided into M measurement regions Z(0) to Z(M - 1), and measurement is sequentially performed for every region.

[0018] In step S1, measurement starts.

[0019] In step S2, the coherence gate and the position of the focus are adjusted, and a measurement image of the measurement region Z(i) is acquired. Note that the initial value of i is taken to be 0.

[0020] In step S3, the contrast of a measurement image of the measurement region Z(i) is corrected, so that the corrected measurement image (correction image) of the measurement region Z(i) is acquired.

[0021] In step S4, the correction image is analyzed, and signal processing is performed to acquire a real image of the measurement region Z(i).

[0022] In step S5, real images from the measurement regions Z(0) to Z(i) are joined together.

[0023] In step S6, it is determined whether measurement has been performed for all measurement regions (whether measurement has completed for all measurement regions). If there is a measurement region for which measurement has not been performed (if i < M - 1) (No in step S6), one is added to i, and the procedure returns to step S2. If measurement has been performed for all the measurement regions (if i = M - 1) (Yes in step S6), the procedure proceeds to step S7. Thus, in step S7, a desired tomographic image (an image obtained by joining together real images of all the measurement regions; a tomographic image having a large measurement range of the object to be measured and a high transversal resolution) can be acquired.

<Example>

[0024] Next, a specific example of the optical coherence tomography apparatus according to this embodiment is described. Specifically, an ophthalmic OCT apparatus to which this invention is applied is described below.

<Configuration of Optical Apparatus>

[0025] FIG. 2 illustrates a configuration of a Mach-Zehnder interference system used in an OCT apparatus according to this example. Light emitted from a light source 201 (emitted light) passes through a single mode fiber 202-1 and is led to a lens 211-1. The emitted light is divided into reference light 205 and measurement light 206 by a beam splitter 203-1. After an eye 207, or an object to be measured, is irradiated with the measurement light 206, the measurement light 206 returns as return light 208, which is caused by reflection or scattering. The reference light and the return light pass through a beam splitter 203-2, a lens 211-2 and a single mode fiber 202-3 and are incident on a spectrometer 218. Data such as a wavelength spectrum of light (coherent light of the return light and the reference light) acquired in the spectrometer is input to a computer 219. Note that the light source 201 is a super luminescent diode (SLD), which is a representative, low-coherent light source. Considering the fact that the object to be measured is an eye, it is preferable that the emitted light be infrared light (e.g., light having a center wavelength of 840 nm and a bandwidth of 50 nm).

[0026] A description is given of the reference optical path of the reference light 205. The reference light 205 resulting from division by the beam splitter 203-1 is sequentially incident on mirrors 214-1 to 214-3. The reference light 205 is led to the beam splitter 203-2 and is incident on the spectrometer. Note that the reference light 205 passes through the interior of a dispersion-compensating glass 215-1 between the mirrors 214-1 and 214-2. The length of the dispersion-compensating glass 215-1 is L1, which is preferably equal to twice the depth of a typical eye. This length is preferred so as to compensate the reference light 205 for dispersion caused when the measurement light 206 reflects and scatters in the eye 207. In this example, the length L1 is given to be 46 mm. This length is twice 23 mm regarded as the average diameter of an eyeball of Japanese people. Further, the mirrors 214-1 and 214-2 can be moved in directions indicated by arrows in FIG. 2 by a mirror drive mechanism 213. Bymoving the positions of the mirrors 214-1 and 214-2, the optical path length of the reference light 205 can be adjusted and controlled. The reference light 205 passes through the interior of a dispersion-compensating glass 215-2 between the mirrors 214-2 and 214-3. The dispersion-compensating glass 215-2 is used for dispersion compensating of an objective lens 216 and a scan lens 217 used for scanning an eye.

[0027] A description is given of the measurement optical path of the measurement light 206. The measurement light 206 resulting from division by the beam splitter 203-1 is reflected from a beam splitter 203-3 and is incident on a mirror of an XY scanner 204. The XY scanner 204 performs a raster scan of a retina 210 in a direction perpendicular to the optical axis (irradiation direction). The center of the measurement light 206 is adjusted so as to be in alignment with the center of rotation of a mirror of the XY scanner 204. The objective lens 216 and the scan lens 217 constitute an optical system for scanning the retina 210 (leading the measurement light to various positions of the retina), and are used for scanning the retina 210 with a point in the vicinity of a cornea 209 used as a supporting point. In this example, focal distances of the objective lens 216 and the scan lens 217 are 50 mm and 50 mm, respectively. The focus position of the objective lens 216 (in the irradiation direction) can be adjusted by a focus drive mechanism 212. When the measurement light 206 is incident on the eye 207, the measurement light 206 reflects and scatters by the retina 210, and returns as the return light 208. The return light 208 passes through the same optical path up to the beam splitter 203-3 as the measurement light 206, and passes through the beam splitter 203-3. Then the return light 208 is led by the beam splitter 203-2 to be incident on the spectrometer.

[0028] Note that the focus drive mechanism, the mirror drive mechanism, the XY scanner and the spectrometer are controlled by the computer 219 to perform desired operation. The computer performs data processing, data saving and image processing of the spectrometer.

<Measurement range>

[0029] Next, with reference to FIG. 3, the width (in the irradiation direction) of the measurement region is described. In FIG. 3, the vertical axis indicates the intensity (intensity of light; reflected intensity) and the horizontal axis indicates the position (in the light application direction) in the interior of an object to be measured. FIG. 3 schematically illustrates a case where a coherence gate 301 is placed between the measurement region Z(3) and the measurement region Z(2) adjacent thereto and measurement of measurement region Z (3) is performed. Reference numeral 302 denotes a width of each measurement region, reference character 303 denotes the measurement depth, and reference character 304 denotes the depth of focus. The measurement depth and the depth of focus will be described below.

[0030] The depth of focus (DOF) represents the visible range of an obtained image. The depth of focus is expressed by expression 1 (optical distance) using the numerical aperture (NA) of a lens used for focusing measurement light into an object to be measured and a center wavelength λ of a light source. In FIG. 3, the plus side of the range obtained by expression 1 is indicated by continuous lines and the minus side is indicated by broken lines.

$$\mathrm{DOF} = \pm\lambda/(2\mathrm{NA}^2) \qquad (1)$$

[0031] In cases where an object to be measured is an eye and the object to be measured is divided into six measurement regions, if the width of each measurement region is 500 μm, it is preferable that the depth of focus be longer than the total length of 1000 μm (±500 μm). Note that in a typical SD-OCT apparatus, the whole length of the depth of focus is about 3 mm. As a matter of course, if the number of division increases, the measurement region can be made smaller and therefore the depth of focus can also be decreased. Note that a region exceeding the depth of focus to some extent is not without the possibility of measurement. The focus need not be set at the position of the coherence gate. However, in order to obtain a uniform image, it is preferable that the depth of focus be larger than the width of each measurement region. In the case of an OCT apparatus, the NA can be changed by changing the diameter of a light beam. In general, if the diameter of a light beam incident on an eye increases, the NA increases.

[0032] The measurement depth represents a range in which aliasing does not occur (occurrence of aliasing makes measurement difficult). The measurement depth is expressed by expression 2 (optical distance) using the number N of pixels (even number, typically the powers of 2, such as 1024 and 2048) of a line sensor of a spectrometer and a spectral bandwidth ΔK of the wave number detected by the spectrometer. In FIG. 3, the plus side and the minus side of the range obtained by expression 2 are indicated by continuous lines and broken lines, respectively.

$$\mathrm{Lmax} = \pm N/(4\Delta K) \qquad (2)$$

[0033] Assuming that the center wavelength of measurement light is 840 nm, the bandwidth is 50 nm and the number of pixels of the line sensor of the spectrometer is 1024, the range that can be measured extends up to an optical distance of about ±3.4 mm. Note that the measurement depth represented by expression 2 is a theoretical value, and in fact an actual number of sampling times is less than N because of the optical resolution of a spectrometer. The range that can be accurately replaced (measured) is therefore smaller than the theoretical measurement depth. Accordingly, the width of a measurement region needs to be set to be less than the theoretical measurement depth. In general, the relationship of the width of the measurement region < the theoretical measurement depth is satisfied. Further, in order to obtain a uniform image, it is preferable that the depth of focus (whole length) and the width of a measurement region satisfy the relationship of expression 3. That is, it is preferable that the width of the measurement region be less than one half of the depth of focus when a measurement image of the measurement region in question is acquired.

$$2 \times \text{the width of the measurement region} < \text{the depth of focus (whole length)} \qquad (3)$$

[0034] In discrete Fourier transformation, each element constituting a measurement image has a discrete value which is given by expression 4 (optical distance). Here t is an integer for $0 \le t \le N/2$.

$$L = t/(2\Delta K) \qquad (4)$$

[0035] Numerical depth resolution δ(L) is expressed by expression 5. The numerical depth resolution δ(L) is also an interval per pixel. In this example, the numerical depth resolution δ(L) is an optical distance of about 6.8 μm.

$$Lmin = \delta(L) = 1/(2\Delta K) \qquad (5)$$

<Method of Removing Mirror Image>

[0036] Next, with reference to FIGS. 4A to 4D, a method of acquiring a real image from a corrected measurement image (correction image) (a method of removing a mirror image) is described. In FIGS. 4A to 4D, the vertical axis indicates the intensity, and the horizontal axis indicates the position (in the irradiation direction) in the interior of an object to be measured. Note that according to a method to be described below, a mirror image in one measurement region can be removed by at least one measurement.

[0037] FIG. 4A illustrates an ideal tomographic image of an object to be measured. In this example, the object to be measured is divided into measurement regions Z(0) to Z(5) at regular intervals, and measurement is performed on a region basis. Reference numerals R(0) to R(5) represent real images of the measurement regions Z(0) to Z(5), respectively. In this example, the measurement region Z(0) is disposed as a first measurement region at an end of the object to be measured. A plurality of measurement regions are set so that first to xth measurement regions (x is an integer greater than 1; the measurement regions Z (0) to Z (5) in examples of FIGS. 4A to 4D) are arranged sequentially in a direction of irradiation of measurement light. Note that with an OCT apparatus, a portion having a large difference in refractive index is measured as a large signal. Accordingly, a region at the end of the object to be measured is a region adjacent to a range in which the difference in refractive index can be ignored. Note that even in the interior of the object to be measured, if the difference in refractive index can be ignored in a range equal to or greater than the width of the measurement region, the measurement region in question and a region disposed in the outside thereof can be regarded as different objects. Therefore, such a measurement region may be regarded as a region at the end of the object to be measured.

[0038] FIG. 4B schematically illustrates a mirror image reflected in the measurement region Z(i) (a mirror image to be superimposed on a real image of the measurement region Z(i)) when the coherence gate is placed at the boundary of the measurement region Z(i - 1) and the measurement region Z(i) (i > 1). Since the mirror image reflected in the measurement region Z(i) is a mirror image of the real image of the measurement region Z (i - 1), the mirror image is denoted by a reference character R'(i - 1). Note that a measurement region of i = 0 (the measurement region Z(0)) is a region at the end of the object to be measured, and therefore no mirror image appears.

[0039] FIG. 4C illustrates correction images H (0) to H (5) of measurement regions when the coherence gate is placed at the boundary between the measurement region Z(i - 1) and the measurement region Z(i). The correction image is obtained by dividing a measurement image S (i) by correction data D(i). The correction data is an attenuation function as illustrated in FIG. 7. The correction images of the measurement regions Z(1) to Z (5) are images in each of which a mirror image is superimposed on a real image. However, as described above, no mirror image appears in the measurement region Z(0), and therefore the correction image H(0) of the measurement region Z (0) is a real image. The correction image H(i) is expressed by expressions 6-1 and 6-2.

$$H(i) = R(i) \qquad i = 0 \qquad (6\text{-}1)$$

$$H(i) = R(i) + R'(i - 1) \qquad i = 1 \text{ to } 5 \qquad (6\text{-}2)$$

[0040] Expression 6-1 represents that the correction image H(0) of the measurement region Z(0) is a real image R(0). Expression 6-2 represents that a real image R(i) of the measurement region Z(i) can be obtained by subtracting a mirror image R'(i - 1) of a real image R(i - 1) from the correction image H(i) of the measurement region Z(i).

[0041] Given that the real image obtained by removing the mirror image from the correction image is denoted by a reference character C(i), the real image C (i) is expressed by expressions 7-1 and 7-2 (reference character C'(i - 1) denotes a mirror image of a real image C(i - 1).

$$C(i) = H(i) \qquad i = 0 \qquad (7\text{-}1)$$

$$C(i) = H(i) - C'(i - 1) \qquad i = 1 \text{ to } 5 \qquad (7\text{-}2)$$

[0042] The mirror image C'(i - 1) can be calculated from the real image C(i - 1). As described above, no mirror image appears in a first measurement region (the measurement region Z(0)). Therefore in this example, the correction image H(0) is employed as a tomographic image (real image) C(0) for the first measurement region. For the second to xth measurement regions in sequence, a Yth (2 ≤ Y≤X) real image is obtained by removing a mirror image of a real image of a (Y - 1)th measurement region from a correction image of a Yth measurement region. That is, in an example of FIG. 4D, the real image C(i) is calculated sequentially for i = 1 to 5. This allows a real image to be acquired for every measurement region. By joining together acquired real images, a desired tomographic image can be obtained (FIG. 4D).

[0043] Note that in this example, the real image C(i) is calculated sequentially from i = 1; however, the calculation method is not limited to that in this example. For example, in cases where the measurement region Z(5) is disposed at the end of the object to be measured and the coherence gate is placed at the boundary between the measurement region Z(I + 1) and the measurement region Z (I) (I is not more than y and not less than 0, and y = 4 in examples of FIGS. 4A to 4D), and the measurement region Z (5) may be the first measurement region. More specifically, in such a case, the correction image H(5) becomes a real image C(5), the mirror image of the real image C (I + 1) of the measurement region Z(I + 1) is reflected in the measurement region Z(I). Therefore, the real image C (I) can be obtained by subtracting a mirror image C'(I + 1) from the correction image H(I). A real image of each measurement region can be obtained by calculating the real image C(I) sequentially for I = 4 to 0.

[0044] It is conceivable that the ends are positioned in the interior of the object to be measured. For example, it is conceivable that the measurement region Z(2) and the measurement region Z (4) are regions at the ends of the object to be measured, and there is no structure in the measurement region Z(3). In this case, if the coherence gate is placed at the boundary between the measurement region Z(i - 1) and the measurement region Z(i), the correction image H(3) becomes the mirror image of the real image C(2), and the correction image H (4) becomes the real image C (4). Therefore, in such a case, real images of the measurement regions Z(0), Z(1) and Z(5) may be calculated in the same way as described above.

[0045] Note that a method of acquiring a real image from a measurement image (a method of removing a mirror image) is not limited to the method described above. For example, as disclosed in Japanese Patent Application Laid-Open No. 11-325849, a real image of a measurement regionmaybe acquired by performing measurement of one measurement region while changing the position of the coherence gate a plurality of times. Any method may be used if a real image of each measurement region can be acquired by it.

<Signal Processing>

[0046] With reference to FIG. 5, a method of analyzing data of measurement images (measurement image data) is described. In this example, a case in which the coherence gate is placed at the boundary between the measurement region Z(i - 1) and the measurement region Z(i), and measurement of the measurement region Z(i) is performed is described. Hereinafter, measurement image data of the measurement region Z(i) is denoted by reference character S(i, k), where i is a region number from 0 to M - 1, and k is an element number from 0 to N - 1 (both i and k are integers). M is the number of regions, and N is the number of pixels of the line sensor. Note that in this example, the element number in a measurement region ranges from 0 to n, and a measurement image is obtained in a range larger than the measurement region. Since n satisfies a relationship of $n < N/2$, $n = 500/6.8$ = about 74 pixels if the width of the measurement region is about 500 μm (because $\delta(L) = 6.8$ μm in this example). The width of a measurement region can be decreased by increasing the number of divisions, and therefore n is decreased with respect to the number of pixels of the line sensor. Similarly, data of the real image of each measurement region (real image data) is denoted by reference character C(i, k).

[0047] In step S1-1, measurement starts. Note that the initial value of i is taken to be 0.

[0048] In step S1-2, in order to perform measurement of the measurement region Z(i), the coherence gate and the position of the focus are adjusted. Because the object to be measured is an eye, the coherence gate is placed at a position on the side of a cornea with respect to a retina. After the coherence gate is placed on the cornea side, the measurement image begins to change as the coherence gate is moved toward the retina. More specifically, the measurement image approaches closer to the coherence gate in synchronization with the movement of the coherence gate.

At a position where a desired state (state where no mirror image is produced) is achieved as a result of movement, the coherence gate and the focus are stopped. This position is determined as the position of the measurement region Z(0). Note that the position of the measurement region Z(i) is a position obtained by adding the width of the measurement region × i to the position of the measurement region Z(0). Ideally, control is performed so that the position of measurement image data S (i - 1, n) is identical to the position of measurement image data S(i, 0).

[0049] In step S1-3, the measurement image data S (i, k) of the measurement region Z(i) is acquired (a measurement image acquisition unit).

[0050] In step S1-4, it is determined whether measurement has completed for desired measurement regions (up to i = 5 in the examples of FIGS. 4A to 4D). If measurement has not completed (No in step S1-4), one is added to i, and the procedure returns to step S1-2. If measurement has completed (Yes in step S1-4), i returns to the initial value and the procedure proceeds to step S1-5.

[0051] In step S1-5, the contrast of measurement image data of the measurement region Z (i) is corrected (a correction unit). Correction is performed, for example, according to a correction function determined depending on an attenuation function that represents changes in intensity with respect to the position in the irradiation direction (of measurement light) in a measurement region. More specifically, an optical coherence tomography apparatus stores in advance or acquires the above-mentioned correction function, and performs correction for every measurement position (element position) using a value of correction function corresponding to the position (a value obtained by substituting the position for the correction function; correction data). Given that data used for correction is correction data D(i, k), the corrected measurement image data (correction image data) H(i, k) is expressed by expression 8.

$$H(i, k) = S(i, k)/D(i, k) \qquad (8)$$

[0052] Note that the correction function may be a correction function itself obtained from a theory or an experiment, may also be an approximate function (a straight line or a secondary curve) of the attenuation function, and may also be a sum or a product of the attenuation function and a given coefficient. Any function may be used if it can eliminate a phenomenon (a phenomenon in which variations in contrast appear) specific to the SD-OCT.

[0053] Note that a single correction functionmaybe used; however, if characteristics (the above-mentioned characteristics; the attenuation function) differ from one measurement region to another, correction functions are preferably prepared for every measurement region (it is preferable that the contrast of a measurement image be corrected using a correction function that differs for every measurement region). For example, in cases where the depth of focus varies depending on the position of the focus, the characteristics vary for every measurement region, and therefore such preparation is effective.

[0054] In step S1-6, a real image of the measurement region Z(i) is calculated (a tomographic image acquisition unit).

[0055] In the measurement region Z(0), no mirror image is produced. Therefore, the relationship between the correction value data H (i, k) and the calculated real image data C(i, k) is expressed by expression 9.

$$C(i, k) = H(i, k) \qquad (9)$$

[0056] Note that correction image data H(0, 0) is not tomographic data (there is no structure at the position of the element), and therefore correction value data H(0, 1) may be used in place of the correction value data H(0, 0).

[0057] If i is larger than 1, a mirror image is produced in the measurement region Z (i). Therefore, mirror image data is removed from the correction image data H(i, k) acquired in step S1-5 to acquire the real image data C(i, k). The removed mirror image data is obtained by reversing relative to the position of the coherence gate (in this example, the boundary between the measurement region and its adjacent region). More specifically, real image data C(i - 1, n - k) as the mirror image data is removed from the correction image data H(i, k). Note that real image data C(i, 0) is data at the position where the coherence gate is placed, and therefore is replaced by real image data C(i - 1, n) (expression 10-1). The calculated real image data C(i, k) is expressed by expression 10-2.

$$C(i, 0) = C(i - 1, n) \qquad k = 0 \qquad (10\text{-}1)$$

$$C(i, k) = H(i, k) - C(i - 1, n-k) \quad 0 < k \leq n \qquad (10\text{-}2)$$

[0058] In step S1-7, it is determined whether real images of desired measurement regions (measurement regions up to i = 5 in examples of FIGS. 4A to 4D) have been acquired. If the acquisition has not completed (No in step S1-7), one is added to i, and the procedure returns to step S1-5. If the acquisition has completed (Yes in step S1-7), i returns to 1 and the procedure proceeds to step S1-8.

[0059] In step S1-8, an image adjustment of the real image of Z(i) is performed. The image adjustment is adjustment of the pixel value (intensity) of a real image and the position of a measurement region (the position in the direction of irradiation of measurement light). As described above, it is desirable that the position of the real image data C(i, 0) and the position of the real image data C(i - 1, n) be identical to each other. However, their positions are displaced from each other because of a position error of the coherence gate, an intensity error of a light source, and the like. In this step, such displacement is adjusted.

[0060] With reference to FIG. 6, the image adjustment is described. In FIG. 6, the vertical axis indicates the reflected intensity, and the horizontal axis indicates the position (in the irradiation direction) in the object to be measured. In FIG. 6, real images of the measurement regions Z(3) and Z(4) adjacent to each other are indicated by a continuous line and a broken line, respectively. A real image of the measurement region Z(i) overlaps a real image of the measurement region Z(i + 1) in the range of k > n. Part or all of data of the overlapping portion is used for the image adjustment. Interpolation is performed between real image data obtained in the range of k > n, and data obtained by the interpolation may be used. Ideally, the real image data is adjusted so that the overlapping portions match each other. Note that assuming that the real image of the measurement region Z (3) has already been adjusted, adjusting the real image of the measurement region Z(4) so as to match the real image of the measurement region Z(3) is described below.

[0061] An adjustment of the positions of the measurement regions (i.e., an adjustment in the horizontal axis direction of FIG. 6) is performed so that the intensity difference of the overlapping portion of tomographic images (the continuous line and the broken line) of the measurement region and its adjacent region is equal spacing. That is, in order to cause the intensity difference of the overlapping portion of the continuous line and the broken line to be equal spacing (e.g., to minimize the dispersion of intensity differences of the overlapping portion), the broken line is shifted in the horizontal axis direction. If in the overlapping portions, there is a specific peak in each of the real images, adjustment may be performed so that their peak positions match each other. Intensity adjustment (i.e., an adjustment in the vertical axis direction of FIG. 6) is performed so that the intensity difference of the overlapping portion of tomographic images (the continuous line and the broken line) of the measurement region and its adjacent region is minimum. That is, in order to cause the intensity difference of the overlapping portion of the continuous line and the broken line to be minimum (e.g., to make the total of absolute values of intensity differences of the overlapping portion minimum), the broken line is shifted in the vertical axis direction. Note that in the image adjustment, only one of the position and the intensity of the measurement region may be adjusted. If both the position and the intensity of the measurement region are adjusted, it is preferable that the intensity be adjusted after the position is adjusted.

[0062] In step S 1-9, it is determined whether the image adjustment of the real image of desired measurement regions (measurement regions up to i = 5 in examples of FIGS. 4A to 4D) has completed. If the image adjustment has not been completed (No in step S1-9), one is added to i, and the procedure returns to step S1-8. If the image adjustment has completed (Yes in step S1-9), the procedure proceeds to step S1-10.

[0063] In step S1-10, real images acquired for all the measurement regions are joinedtogether. Thus, in step S1-11, the desired tomographic image can be acquired. Note that when the real images are joined together, for the overlapping portions, their average values may be used, and an element whose number is greater than n may be ignored.

[0064] It should be noted that calculation is made with the coherence gate placed at the boundary of the measurement regions in this example; however, an error due to the spectrum of a light source is sometimes mixed to a component of S (i, k) with i in a lower order. In such a case, when a measurement image is acquired, the position of the coherence gate may be set on a side of the adjacent region with respect to the boundary between the measurement region and the adjacent region. For example, when measurement of the measurement region Z(i) is performed, the coherence gate shouldbe shifted from the boundary between the measurement region Z(i- 1) and the measurement region Z (i) toward the measurement region Z(i - 1) by several to several tens of elements. The number of shifted elements may be determined depending on the coherence function of a light source, or the like.

[0065] As a result, data for every measurement region can be smoothly connected. This enables a more accurate tomographic image to be obtained.

[0066] As described above, with an optical coherence tomography apparatus according to this embodiment, the contrast of a measurement image is corrected, and a tomographic image (real image) is acquired from the corrected measurement image. Thus, the tomographic image acquired from each of a plurality of measurement regions can be continuously joined together.

[0067] While the present invention has been described with reference to exemplary embodiments, it is to be understood that the invention is not limited to the disclosed exemplary but is defined by the claims.

Claims

1. An optical coherence tomography method that divides light from a light source (201) into measurement light (206) and reference light (205) and acquires a tomographic image of an object (207) on a basis of a wavelength spectrum of interfering light of the reference light (205) and return light (208), the return light (208) returning from the object (207) upon applying the measurement light (206) onto the object (207) , the optical coherence tomography method comprising the steps of:

   acquiring a measurement image based on the wavelength spectrum of the interfering light at each of a plurality of measurement regions (Z(0)-Z(i)) of the object (207) adjacent to one another in a direction of irradiation of the measurement light (206), thereby adjusting the position of the coherence gate and the focal position to positions corresponding to the respective measurement region (Z(0)-Z(i));
   correcting, for every measurement region (Z(0)-Z(i)), a contrast of the measurement image based on an intensity change in the measurement region (Z(0)-Z(i)) corresponding to the measurement image in the direction of irradiation; and
   acquiring, for every measurement region (Z(0)-Z(i)), a tomographic image from the corrected measurement image.

2. The optical coherence tomography method according to claim 1, wherein in the step of correcting the contrast of the measurement image, the contrast of the measurement image is corrected with a correction function differing for every measurement region (Z(0)-Z(i)).

3. The optical coherence tomography method according to claim 1 or 2, wherein in the step of correcting the contrast of the measurement image, the contrast of the measurement image is corrected for every measurement region (Z(0)-Z(i)) according to a correction function determined based on an attenuation function representing a change in intensity with respect to a position in the measurement region (Z(0)-Z(i)) in the direction of irradiation.

4. The optical coherence tomography method according to any one of claims 1 to 3, further comprising a step of adjusting, for every measurement region (Z(0)-Z(i)), an intensity of the tomographic image and/or a position of the measurement region (Z(0)-Z(i)) in the direction of irradiation.

5. The optical coherence tomography method according to claim 4, wherein
   the measurement image is acquired in a range larger than the measurement region (Z(0)-Z(i)); and
   in the step of adjusting the intensity of the tomographic image and/or the position of the measurement region (Z(0)-Z(i)) in the direction of irradiation, for every measurement region (Z(0)-Z(i)), the position of the measurement region in the direction of irradiation is adjusted so that a difference in intensity of an overlapping portion of the tomographic images of the measurement region (Z(0)-Z(i)) and an adjacent region thereto is equal spacing.

6. The optical coherence tomography method according to claim 4 or 5, wherein
   the measurement image is acquired in a range larger than the measurement region (Z(0)-Z(i)), and
   in the step of adjusting the intensity of the tomographic image and/or the position of the measurement region in the direction of irradiation, for every measurement region (Z(0)-Z(i)), the intensity of the tomographic image is adjusted so that a difference in intensity of an overlapping portion of the tomographic images of the measurement region (Z(0)-Z(i)) and an adjacent region thereto is minimum.

7. The optical coherence tomography method according to any one of claims 1 to 6, wherein in the step of acquiring the measurement image, the position of the coherence gate is adjusted on a side of an adjacent region to the measurement region (Z(0)-Z(i)) with respect to a boundary between the measurement region (Z(0)-Z(i)) and the adjacent region.

8. The optical coherence tomography method according to any one of claims 1 to 7,
   wherein, in the step of acquiring the tomographic image, the tomographic image for every measurement region (Z(0)-Z(i)) is acquired by removing a mirror image of the tomographic image of the adjacent region from the corrected measurement image.

9. The optical coherence tomography method according to any one of claims 1 to 8, wherein, in the step of correcting the contrast of the measurement image, for every measurement region, a contrast change due to an intensity change depending on a distance from the coherence gate is corrected.

10. The optical coherence tomography method according to any one of claims 1 to 9, wherein the object is a retina.

11. An optical coherence tomography apparatus configured to divide light from a light source (201) into measurement light (206) and reference light (205) and to acquire a tomographic image of an object (207) on a basis of a wavelength spectrum of interfering light of the reference light (205) and return light (208), the return light (208) returning from the object (207) upon irradiating the measurement light (205) onto the object (207), the optical coherence tomography apparatus comprising:

a measurement image acquisition unit configured to acquire a measurement based on the wavelength spectrum of the interfering light at each of a plurality of measurement regions (Z(0)-Z(i)) of the object (207) adjacent to one another in a direction of irradiation of the measurement light, and to adjust, for every measurement region (Z(0)-Z(i)), the position of the coherence gate and the focal position to positions corresponding to the respective measurement region;
a correction unit configured to correct, for every measurement region (Z(0)-Z(i)), a contrast of the measurement image based on an intensity change in the measurement region (Z(0)-Z(i)) corresponding to the measurement image in the direction of irradiation; and
a tomographic image acquisition unit configured to acquire, for every measurement region (Z(0)-Z(i)), a tomographic image from the corrected measurement image.

12. The optical coherence tomography apparatus according to claim 11, wherein, the correction unit is configured to correct for every measurement region, a contrast change due to an intensity change depending on a distance from the coherence gate.

13. The optical coherence tomography apparatus according to claim 11 or 12, wherein the apparatus is configured to measure a retina as the object.

Patentansprüche

1. Optisches Kohärenztomographieverfahren, das Licht von einer Lichtquelle (201) in Messlicht (206) und Referenzlicht (205) teilt und ein tomographisches Bild eines Objekts (207) auf einer Basis eines Wellenlängenspektrums von interferierendem Licht des Referenzlichts (205) und zurückkehrenden Lichts (208) erhält, wobei das zurückkehrende Licht (208) von dem Objekt (207) zurückkehrt, nachdem das Messlicht (206) auf das Objekt (207) angewendet wurde, und wobei das optische Kohärenztomographieverfahren die Schritte aufweist:

Erhalten eines Messbildes basierend auf dem Wellenlängenspektrum des interferierenden Lichts an jeder einer Vielzahl von Messbereichen (Z(0)-Z(i)) des Objekts (207), die zueinander in einer Richtung des Bestrahlens mit dem Messlicht (206) benachbart sind, wodurch die Position des Kohärenzgates und die Fokuslage an Positionen entsprechend dem jeweiligen Messbereich (Z(0)-Z(i)) angepasst werden;
Korrigieren eines Kontrasts des Messbildes basierend auf einer Intensitätsänderung in dem Messbereich (Z(0)-Z(i)) entsprechend dem Messbild in der Richtung der Bestrahlung für jeden Messbereich (Z(0)-Z(i)); und
Erhalten eines tomographischen Bildes aus dem korrigierten Messbild für jeden Messbereich (Z(0)-Z(i)).

2. Optisches Kohärenztomographieverfahren nach Anspruch 1, wobei in dem Schritt des Korrigierens des Kontrasts des Messbildes der Kontrast des Messbildes mit einer Korrekturfunktion korrigiert wird, die für jeden Messbereich (Z(0)-Z(i)) unterschiedlich ist.

3. Optisches Kohärenztomographieverfahren nach Anspruch 1 oder 2, wobei in dem Schritt des Korrigierens des Kontrasts des Messbildes der Kontrast des Messbildes für jeden Messbereich (Z(0)-Z(i)) gemäß einer Korrekturfunktion korrigiert wird, die basierend auf einer Abschwächungsfunktion bestimmt wird, die eine Änderung in einer Intensität mit Bezug auf eine Position in dem Messbereich (Z(0)-Z(i)) in der Richtung der Bestrahlung repräsentiert.

4. Optisches Kohärenztomographieverfahren nach einem der Ansprüche 1 bis 3, ferner mit einem Schritt des Anpas-

sens einer Intensität des tomographischen Bildes und/oder einer Position des Messbereichs (Z(0)-Z(i)) in der Richtung der Bestrahlung für jeden Messbereich (Z(0)-Z(i)).

5. Optisches Kohärenztomographieverfahren nach Anspruch 4, wobei
das Messbild in einem größeren Bereich als dem Messbereich (Z(0)-Z(i)) erhalten wird; und
in dem Schritt des Anpassens der Intensität des tomographischen Bildes und/oder der Position des Messbereichs (Z(0)-Z(i)) in der Richtung der Bestrahlung für jeden Messbereich (Z(0)-Z(i)) die Position des Messbereichs in der Richtung der Bestrahlung so angepasst wird, dass eine Differenz in einer Intensität eines überlappenden Abschnitts der tomographischen Bilder des Messbereichs (Z(0)-Z(i)) und eines dazu benachbarten Bereichs abstandsgleich ist.

6. Optisches Kohärenztomographieverfahren nach Anspruch 4 oder 5, wobei
das Messbild in einem größeren Bereich als dem Messbereich (Z(0)-Z(i)) erhalten wird; und
in dem Schritt des Anpassens der Intensität des tomographischen Bildes und/oder der Position des Messbereichs in der Richtung der Bestrahlung für jeden Messbereich (Z(0)-Z(i)) die Intensität des tomographischen Bildes so angepasst wird, dass eine Differenz in einer Intensität eines überlappenden Abschnitts der tomographischen Bilder des Messbereichs (Z(0)-Z(i)) und eines dazu benachbarten Bereichs minimal ist.

7. Optisches Kohärenztomographieverfahren nach einem der Ansprüche 1 bis 6, wobei in dem Schritt des Erhaltens des Messbildes die Position des Kohärenzgates auf einer Seite eines dem Messbereich (Z(0)-Z(i)) benachbarten Bereichs mit Bezug auf eine Grenze zwischen dem Messbereich (Z(0)-Z(i)) und dem benachbarten Bereich angepasst wird.

8. Optisches Kohärenztomographieverfahren nach einem der Ansprüche 1 bis 7,
wobei in dem Schritt des Erhaltens des tomographischen Bildes das tomographische Bild für jeden Messbereich (Z(0)-Z(i)) durch Entfernen eines Spiegelbildes des tomographischen Bildes des benachbarten Bereichs von dem korrigierten Messbild erhalten wird.

9. Optisches Kohärenztomographieverfahren nach einem der Ansprüche 1 bis 8,
wobei in dem Schritt des Korrigierens des Kontrasts des Messbildes für jeden Messbereich eine Kontraständerung auf Grund einer Intensitätsänderung abhängig von einem Abstand von dem Kohärenzgate korrigiert wird.

10. Optisches Kohärenztomographieverfahren nach einem der Ansprüche 1 bis 9,
wobei das Objekt eine Netzhaut ist.

11. Optische Kohärenztomographievorrichtung, die konfiguriert ist, Licht von einer Lichtquelle (201) in Messlicht (206) und Referenzlicht (205) zu teilen und ein tomographisches Bild eines Objekts (207) auf einer Basis eines Wellenlängenspektrums von interferierendem Licht von dem Referenzlicht (205) und zurückkehrendem Licht (208) zu erhalten, wobei das zurückkehrende Licht (208) von dem Objekt (207) bei einer Bestrahlung auf dem Objekt (207) mit dem Messlicht (205) zurückkehrt, und die optische Kohärenztomographievorrichtung aufweist:

eine Messbilderhalteinheit, die konfiguriert ist, eine Messung basierend auf dem Wellenlängenspektrum des interferierendem Lichts an jeder einer Vielzahl von Messbereichen (Z(0)-Z(i)) des Objekts (207) zu erhalten, die zueinander in einer Richtung einer Bestrahlung mit dem Messlicht benachbart sind, und für jeden Messbereich (Z(0)-Z(i)) die Position des Kohärenzgates und der Fokuslage an Positionen anzupassen, die dem jeweiligen Messbereich entsprechen;
eine Korrektureinheit, die konfiguriert ist, für jeden Messbereich (Z(0)-Z(i)) einen Kontrast des Messbildes basierend auf einer Intensitätsänderung in dem Messbereich (Z(0)-Z(i)) entsprechend dem Messbild in der Richtung der Bestrahlung zu korrigieren; und
eine Tomographischebilderhalteeinheit, die konfiguriert ist, für jeden Messbereich (Z(0)-Z(i)) ein tomographisches Bild aus dem korrigierten Messbild zu erhalten.

12. Optische Kohärenztomographievorrichtung nach Anspruch 11,
wobei die Korrektureinheit konfiguriert ist, für jeden Messbereich eine Kontraständerung auf Grund einer Intensitätsänderung abhängig von einem Abstand von dem Kohärenzgate zu korrigieren.

13. Optische Kohärenztomographievorrichtung nach Anspruch 11 oder 12,
wobei die Vorrichtung konfiguriert ist, eine Netzhaut als das Objekt zu messen.

Revendications

1. Procédé de tomographie par cohérence optique qui divise la lumière provenant d'une source de lumière (201) en une lumière de mesure (206) et une lumière de référence (205) et acquiert une image tomographique d'un objet (207) sur la base d'un spectre de longueurs d'onde de lumière d'interférence entre la lumière de référence (205) et une lumière renvoyée (208), la lumière renvoyée (208) étant renvoyée par l'objet (207) lors de l'application de la lumière de mesure (206) sur l'objet (207), le procédé de tomographie par cohérence optique comprenant les étapes consistant à :

   acquérir une image de mesure sur la base du spectre de longueurs d'onde de la lumière d'interférence dans chacune d'une pluralité de régions de mesure (Z(0)-Z(i)) de l'objet (207), adjacentes les unes aux autres dans une direction d'exposition de la lumière de mesure (206) pour ainsi ajuster la position de la porte de cohérence et la position focale à des positions correspondant à la région de mesure respective (Z(0)-Z(i)) ;
   corriger, pour chaque région de mesure (Z(0)-Z(i), un contraste de l'image de mesure sur la base d'une variation d'intensité dans la région de mesure (Z(0)-Z(i)) correspondant à l'image de mesure dans la direction d'exposition ; et
   acquérir, pour chaque région de mesure (Z(0)-Z(i)), une image tomographique à partir de l'image de mesure corrigée.

2. Procédé de tomographie par cohérence optique selon la revendication 1, dans lequel, lors de l'étape consistant à corriger le contraste de l'image de mesure, le contraste de l'image de mesure est corrigé par une fonction de correction qui est différente pour chaque région de mesure (Z(0)-Z(i)).

3. Procédé de tomographie par cohérence optique selon la revendication 1 ou 2, dans lequel, lors de l'étape consistant à corriger le contraste de l'image de mesure, le contraste de l'image de mesure est corrigé pour chaque région de mesure (Z(0)-Z(i)) conformément à une fonction de correction déterminée sur la base d'une fonction d'atténuation représentant une variation d'intensité par rapport à une position dans la région de mesure (Z(0)-Z(i)) dans la direction d'exposition.

4. Procédé de tomographie par cohérence optique selon l'une quelconque des revendications 1 à 3, comprenant en outre une étape consistant à ajuster, pour chaque région de mesure (Z(0)-Z(i)), une intensité de l'image tomographique et/ou une position de la région de mesure (Z(0)-Z(i)) dans la direction d'exposition.

5. Procédé de tomographie par cohérence optique selon la revendication 4, dans lequel
   l'image de mesure est acquise dans une gamme de distances supérieure à la région de mesure (Z(0)-Z(i)) ; et
   lors de l'étape consistant à ajuster intensité de l'image tomographique et/ou la position de la région de mesure (Z(0)-Z(i)) dans la direction d'exposition, pour chaque région de mesure (Z(0)-Z(i)), la position de la région de mesure dans la direction d'exposition est ajustée de façon qu'une différence d'intensité d'une partie de chevauchement entre des images tomographiques de la région de mesure (Z(0)-Z(i)) et une région qui lui est adjacente soit un même espacement.

6. Procédé de tomographie par cohérence optique selon la revendication 4 ou 5, dans lequel
   l'image de mesure est acquise dans une gamme de distances plus grande que la région de mesure (Z(0)-Z(i)) ; et
   lors de l'étape consistant à ajuster intensité de l'image tomographique et/ou la position de la région de mesure dans la direction d'exposition, pour chaque région de mesure (Z(0)-Z(i)), l'intensité de l'image tomographique est ajustée de façon qu'une différence d'intensité d'une partie de chevauchement entre les images tomographiques de la région de mesure (Z(0)-Z(i)) et une région qui lui est adjacente soit minimale.

7. Procédé de tomographie par cohérence optique selon l'une quelconque des revendications 1 à 6, dans lequel, lors de l'étape consistant à acquérir l'image des mesure, la position de la porte de cohérence est ajustée d'un côté d'un région adjacente à la région de mesure (Z(0)-Z(i)) par rapport à une frontière entre la région de mesure (Z(0)-Z(i)) et la région adjacente.

8. Procédé de tomographie par cohérence optique selon l'une quelconque des revendications 1 à 7,
   dans lequel, lors de l'étape consistant à acquérir l'image tomographique, l'image tomographique correspondant à chaque région de mesure (Z(0)-Z(i)) est acquise en éliminant une image miroir de l'image tomographique de la région adjacente de l'image de mesure corrigée.

9. Procédé de tomographie par cohérence optique selon l'une quelconque des revendications 1 à 8,
dans lequel, lors de l'étape consistant à corriger le contraste de l'image de mesure, pour chaque région de mesure, une variation du contraste due à une variation d'intensité en fonction d'une distance par rapport à la porte de cohérence est corrigée.

10. Procédé de tomographie par cohérence optique selon l'une quelconque des revendications 1 à 9,
dans lequel l'objet est une rétine.

11. Appareil de tomographie par cohérence optique configuré pour diviser la lumière provenant d'une source de lumière (201) en une lumière de mesure (206) et une lumière de référence (205) et pour acquérir une image tomographique d'un objet (207) sur la base d'un spectre de longueurs d'onde de lumière d'interférence entre la lumière de référence (205) et une lumière renvoyée (208), la lumière renvoyée (208) étant renvoyée par l'objet (207) lors de l'exposition à la lumière de mesure (206) de l'objet (207), l'appareil de tomographie par cohérence optique comprenant :

une unité d'acquisition d'image de mesure configurée pour acquérir une mesure sur la base du spectre de longueurs d'onde de la lumière d'interférence dans chacune d'une pluralité de régions de mesure (Z(0)-Z(i)) de l'objet (207), adjacentes les unes aux autres dans une direction d'exposition de la lumière de mesure (206) pour ainsi ajuster la position de la porte de cohérence et la position focale à des positions correspondant à la région de mesure respective ;
une unité de correction configurée pour corriger, pour chaque région de mesure (Z(0)-Z(i)), un contraste de l'image de mesure sur la base d'une variation d'intensité dans la région de mesure (Z(0)-Z(i)) correspondant à l'image de mesure dans la direction d'exposition ; et
une unité d'acquisition d'image tomographique configurée pour acquérir, pour chaque région de mesure (Z(0)-Z(i)), une image tomographique à partir de l'image de mesure corrigée.

12. Appareil de tomographie par cohérence optique selon la revendication 11, dans lequel l'unité de correction est configurée pour corriger, pour chaque région de mesure, une variation de contraste due à une variation d'intensité en fonction d'une distance par rapport à la porte de cohérence.

13. Appareil de tomographie par cohérence optique selon la revendication 11 ou 12,
dans lequel l'appareil est configuré pour mesurer une rétine en tant qu'objet.

FIG. 1

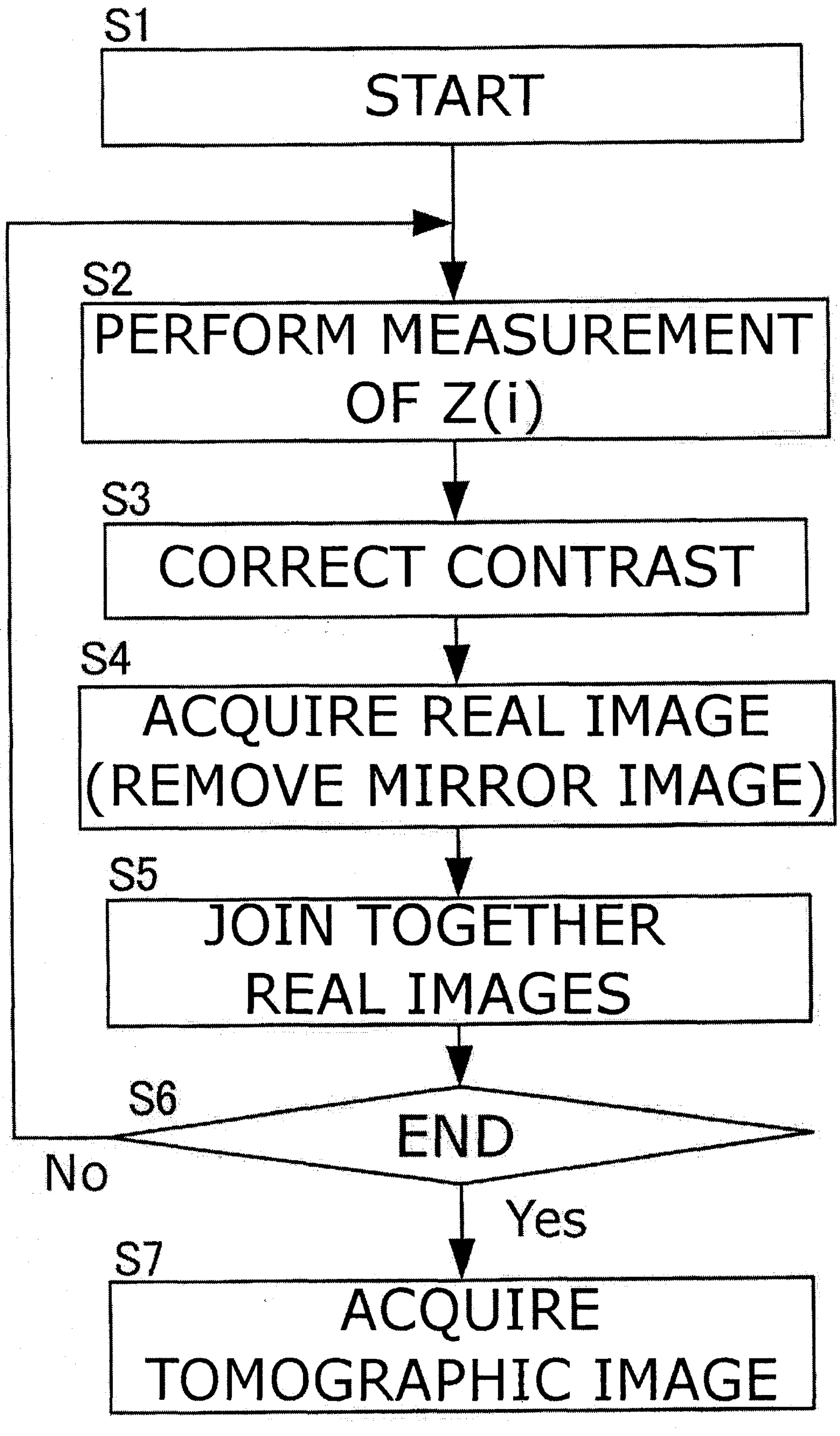
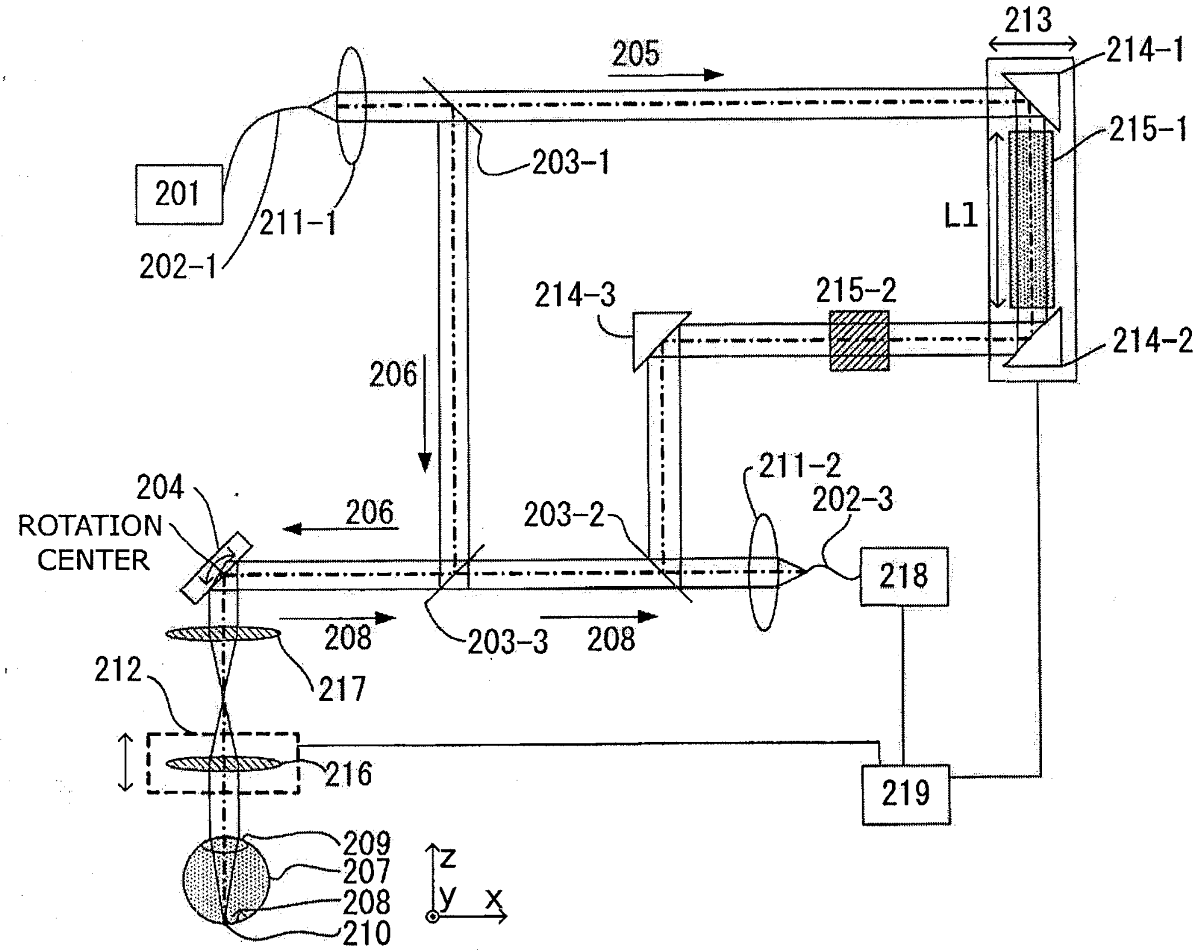
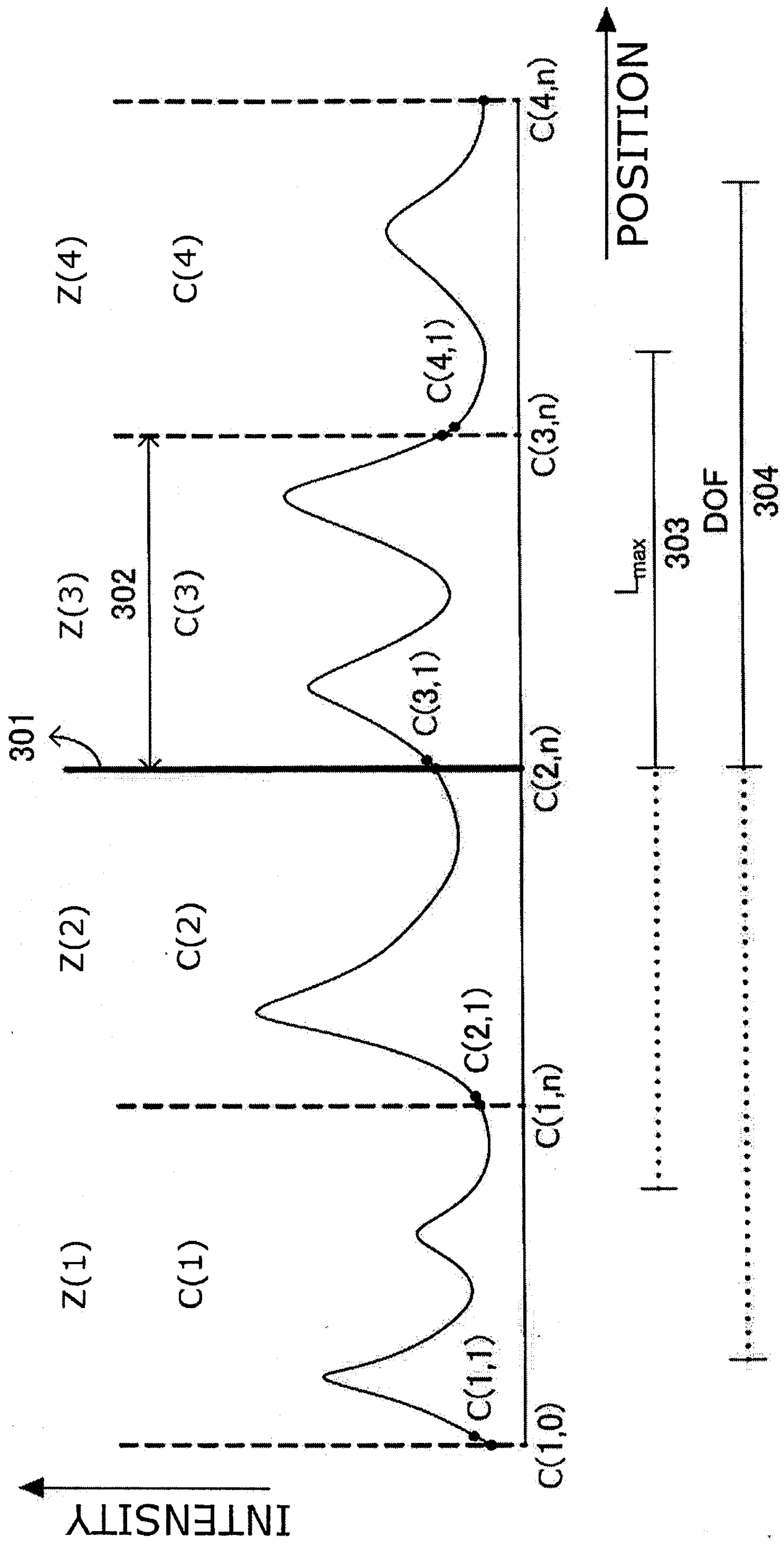

S1
START
S2
PERFORM MEASUREMENT OF Z(i)
S3
CORRECT CONTRAST
S4
ACQUIRE REAL IMAGE (REMOVE MIRROR IMAGE)
S5
JOIN TOGETHER REAL IMAGES
S6
END
No
Yes
S7
ACQUIRE TOMOGRAPHIC IMAGE

FIG. 2



213
214-1
205
215-1
201
211-1
203-1
202-1
L1
214-3
215-2
214-2
206
211-2
204
202-3
ROTATION
CENTER
206
203-2
218
208
203-3
208
212
217
216
219
209
207
208
210
z
y
x

FIG. 3



INTENSITY
POSITION
Z(1)
Z(2)
Z(3)
Z(4)
C(1)
C(2)
C(3)
C(4)
301
302
C(1,0)
C(1,1)
C(1,n)
C(2,1)
C(2,n)
C(3,1)
C(3,n)
C(4,1)
C(4,n)
L_{max}
303
DOF
304

FIG. 4A

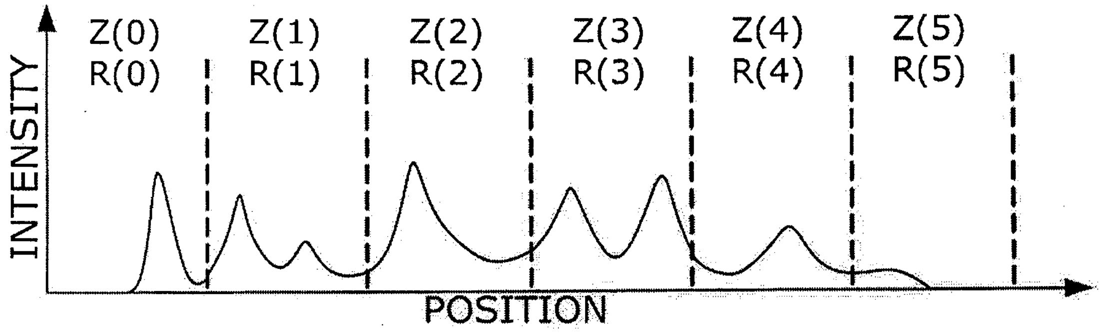


FIG. 4B

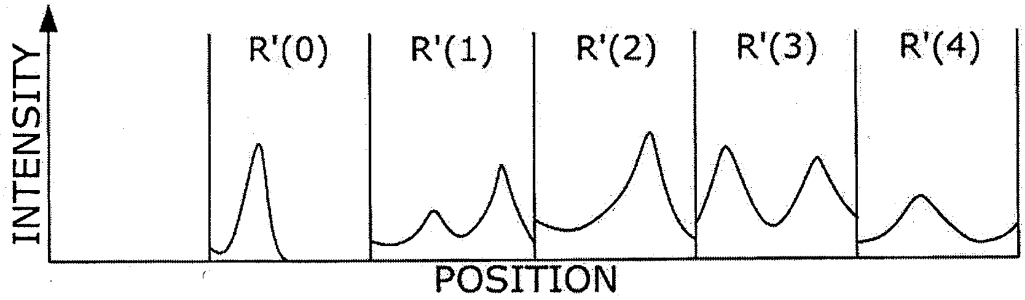


FIG. 4C

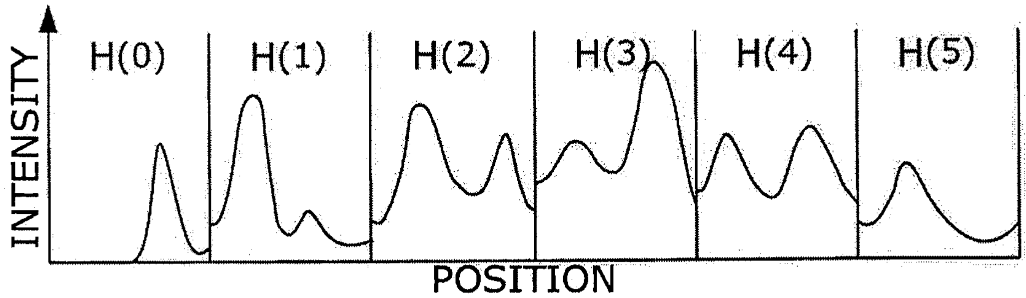


FIG. 4D

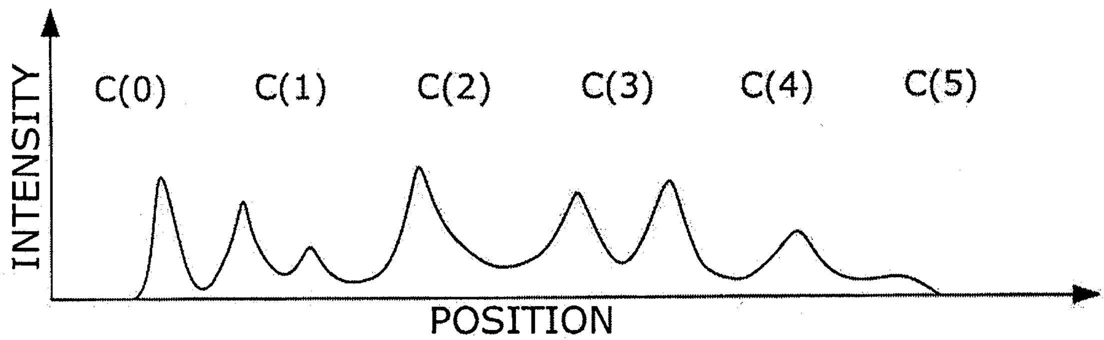


FIG. 5

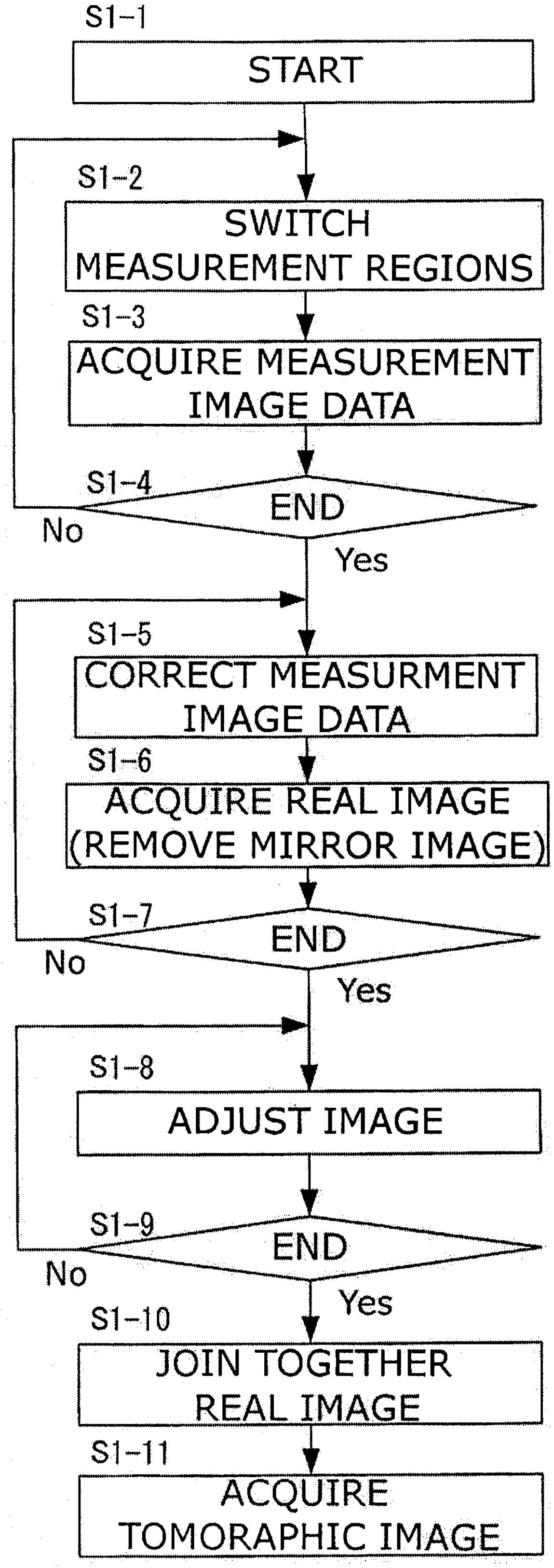

S1-1
START
S1-2
SWITCH MEASUREMENT REGIONS
S1-3
ACQUIRE MEASUREMENT IMAGE DATA
S1-4
END
No
Yes
S1-5
CORRECT MEASURMENT IMAGE DATA
S1-6
ACQUIRE REAL IMAGE (REMOVE MIRROR IMAGE)
S1-7
END
No
Yes
S1-8
ADJUST IMAGE
S1-9
END
No
Yes
S1-10
JOIN TOGETHER REAL IMAGE
S1-11
ACQUIRE TOMORAPHIC IMAGE

FIG. 6

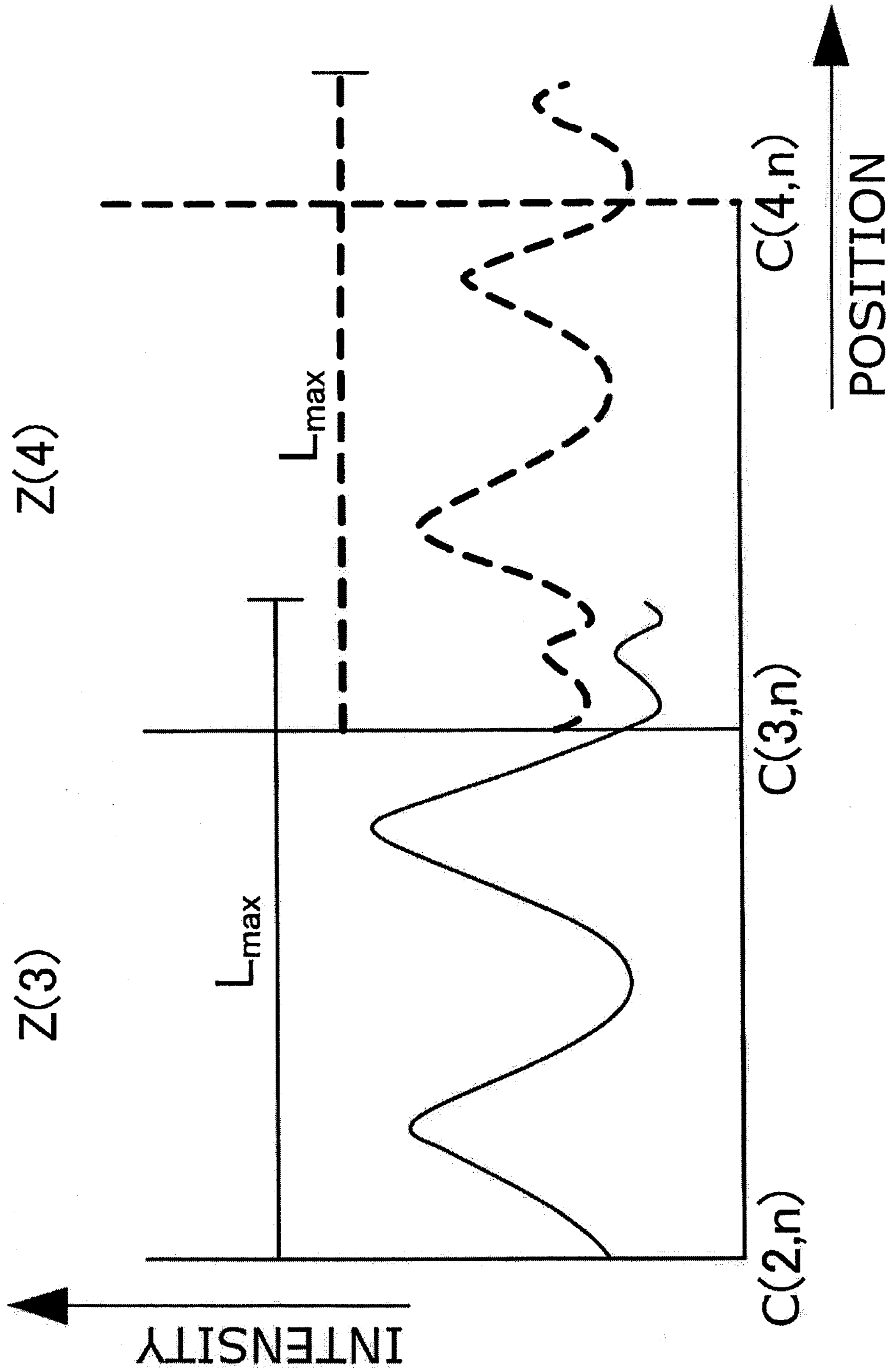
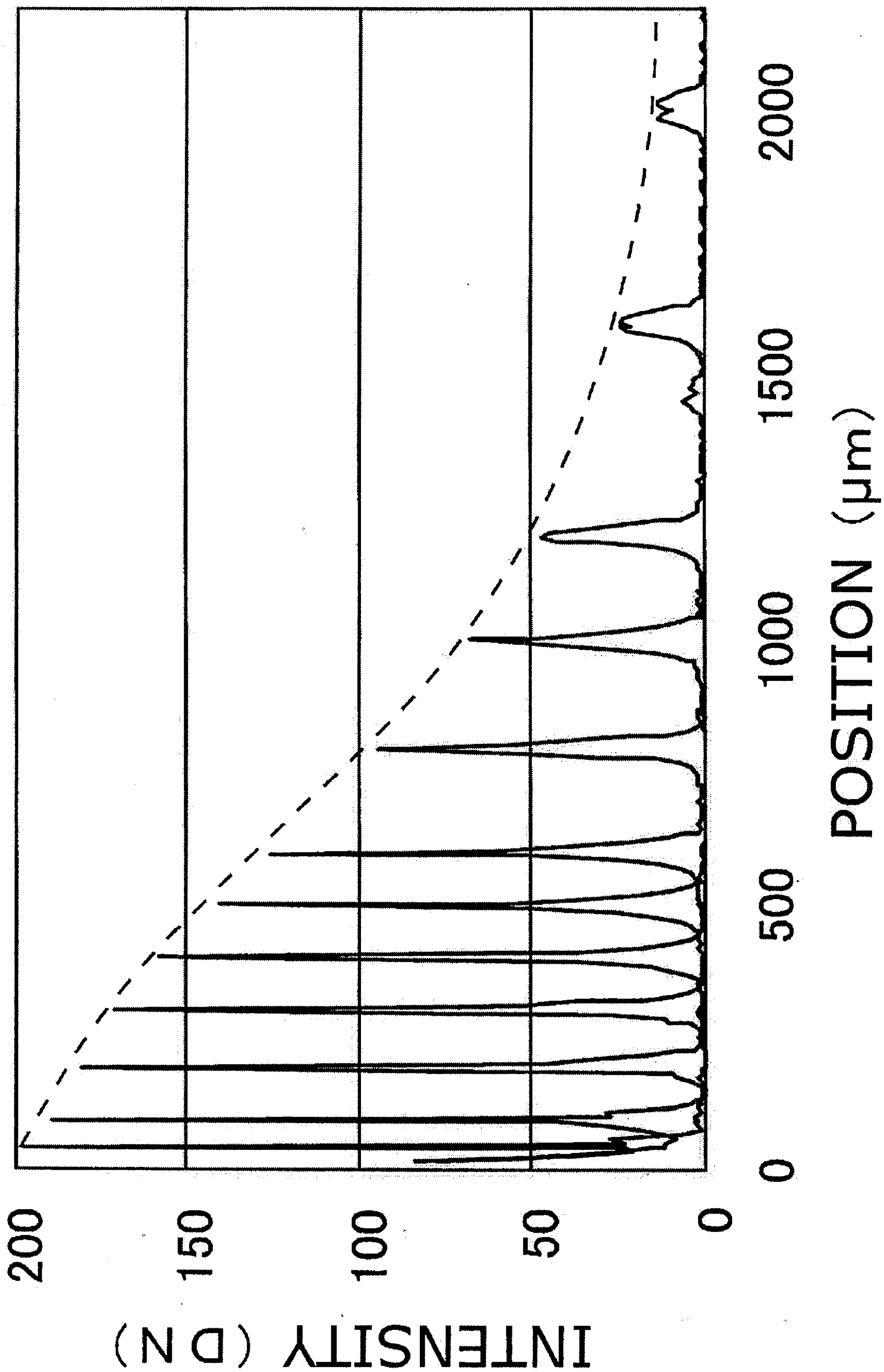


FIG. 7



REFERENCES CITED IN THE DESCRIPTION

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Patent documents cited in the description

- JP H11325849 B **[0003]**
- US 2006089548 A1 **[0006]**
- DE 112006003228 T5 **[0006]**
- JP 11325849 A **[0045]**

Non-patent literature cited in the description

- *OPTICS LETTERS,* 2003, vol. 28, 182-184 **[0005]**
- **D.J. FABER et al.** *Optics Express,* vol. 12 (19), 4353-4365 **[0006]**

专利名称(译)	光学相干断层扫描方法和光学相干断层扫描设备		
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当前申请(专利权)人(译)	佳能株式会社		
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其他公开文献	EP2251637A1		
外部链接	Espacenet		

摘要(译)

根据本发明的光学相干断层摄影方法包括以下步骤：将测量对象划分为在测量光的照射方向上彼此相邻的多个测量区域，并且基于每个测量区域获取测量图像。在相干光的波长谱上;对每个测量区域校正测量区域的测量图像的对比度;并且，对于每个测量区域，从校正的测量图像获取断层图像。

$$DOF = \lambda/(2NA^2) \quad (1)$$