Interference Fringe Scale for Absolute Ocular Fundus Measurement

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Measurements of the human ocular fundus in the three spatial dimensions have not been absolute because measuring techniques have used the optical system of the eye as part of the total ophthalmoscopic system. The ophthalmoscopic magnification due to the total dioptric power of an individual eye can vary substantially from that of the average eye. A new method has been developed to form interference fringes on the fundus so that the fringe spacing can be calculated within a small error by using measured values. The photographed fringes then act as a scale at the fundus with the fringe spacings serving as the graduations, thus allowing accurate absolute measurements of the fundus through the interfering ocular media. Invest Ophthalmol Vis Sci 24:169-174, 1983

Whenever the ocular fundus is being viewed or photographed, the optics of the eye become part of the viewing or photographic system. Because the optical characteristics of individual eyes can vary substantially, the absolute size of any fundus structure cannot be known for certain. Even a knowledge of refractive correction is not enough to calculate away this difficulty. Emmetropic eyes of varying sizes, which need no correction, can vary substantially in total dioptric power.

The problem of determining fundus dimensions despite the diversity of ocular elements could be solved if a scale were placed at the fundus (Fig. 1). If this were done, then the scale would be magnified the same amount as the fundus. Dimensions for fundus structures, such as the optic nerve head, could be read directly from the scale with an accuracy limited only by that of the scale.

The purpose of this paper is to demonstrate a feasible method using interference fringes to produce a scale for absolute measurement of the fundus and, in particular, of the optic disc and cup.

Materials and Methods

If interference fringes are formed on the fundus by focusing two laser beams at the first principal plane of the crystalline lens, using the lens of Gullstrand's schematic eye for location, then the two point sources will be imaged with unit magnification to the second principal plane of the crystalline lens. The fringe spacing can be calculated (Fig. 2) from a knowledge of the following four factors: the wavelength of the monochromatic coherent light that is used to form the fringes (HeNe \( \lambda = 632.8 \text{ nm} \)); the axial length of the eye from the second principal plane of the crystalline lens to the fundus; the refractive index of the vitreous; and the calculated distance between the two point sources that are brought to a focus at the first principal plane of the model crystalline lens.

In order to form and photograph the interference fringes on the fundus, a Rodenstock retinometer was incorporated with a Donaldson stereofundus camera (Fig. 3). We mounted a thin right-angle mirror between the stereoapertures and used an adjustable positive lens to focus the two laser beams to a plane that is very close to being conjugate with the first principal plane of the crystalline lens. We used Gullstrand's lens for calculation. The system and the eye are at the correct distance apart when one of the focusing lights of the camera is imaged at the anterior surface of the crystalline lens.

To calculate the fringe spacing, several measurements were needed. The axial length of the human eye from the posterior lens surface, which is added to the model posterior principal plane-lens surface distance in the calculation (Fig. 2), was determined by ultrasound using the Digital Biometric Ruler instrument. While the axial length was being measured, the anterior chamber depth was also measured using ultrasound. The corneal curvature was measured with a keratometer (Haag Streit). The magnification of the two focused laser points can be calculated with a knowledge of the anterior chamber.
depth as follows. We used the Gaussian formula for the refraction of spherical surfaces, \( 1/f = (n_1 - n_2)/r \), in which \( f \) = focal length and \( 1/f \) = focal power; \( r \) = radius of curvature of the surface (7.8 mm); \( n_1 \) = effective index of the corneal aqueous humor system (1.330 interpolation for \( \lambda = 632.8 \text{ nm} \)); and \( n_2 \) = refractive index of air. The Gaussian formula for magnification is: \( (d - f)/f = m \), in which \( d \) = anterior chamber depth to the first quasi-principal plane (5.58 mm as calculated from the values given by Gullstrand for his schematic eye\(^2\)), where \( f \) is calculated as above.

An alternative method of finding the intralaser point separation, \( "a" \), at the model quasi-principal plane of the crystalline lens can be accomplished using the following procedure. First, measure the in-air laser point separation \( "A" \) by projecting the in-air fringes from the fundus-camera-fringe-projection system to a screen that has a known distance from the two points of focus. The fringe spacings at the screen can be measured accurately using either photographic or mechanical means. The in-air distance between the focused laser points then is expressed by \( A = \lambda L'/F \) where \( \lambda \) = the wavelength of the laser (632.8 nm for HeNe); \( L' \) = the distance from the laser points to the fringe recording screen; and \( F \) = length of one period of the diffraction pattern as measured on the screen. Second, use a dial indicator to measure the camera movement between focusing the laser points on the anterior corneal surface then changing the camera-to-eye working distance, so that the focusing light of the camera is on the anterior lenticular surface and the laser points are at the quasi-principal plane of the crystalline lens. The difference in the dial indicator readings when added to 50.6 microns, which is the distance of the second principal plane from the center of the cornea,\(^2\) will yield the virtual image distance \( D \) of the focused laser points. The virtual image distance is the distance from the corneal system’s principal plane to the virtual image of the focused laser points. The anterior chamber depth from the corneal principal plane (as given above) to the lenticular quasi-principal plane, as measured with ultrasonography and calculated using model principal plane positions, will be designated by \( "d" \). For similar triangles and geometric optics the distance between point sources at the quasi-principal plane is given by \( a = Ad/D \) (where values are defined above).

Figures were taken with a Donaldson stereofundus camera and measurements were made of the eye of a human subject from whom informed consent was obtained. An ultrahigh speed (ASA 1000-5000) black and white Kodak 2475 recording film was used. A model eye was photographed with the Donaldson stereofundus camera and measured directly using very high speed (ASA 400) black and white film (Kodak 5063 Tri-X film). The model eye, which had a water-filled posterior chamber to simulate the vitreous, was measured and photographed with the light of the interference fringes. The model eye consisted of an 18 diopter lens with an adjustable model fundus attached to a micrometer so that the second nodal plane-to-retina axial length could be measured. The nodal planes of the double convex lens were measured using a nodal slide (1.1 mm from surface). The nodal points coincide with the principal points for an air-to-glass-to-air system.

Results

The model eye was photographed, and the fringe spacing was calculated from the measured values,
Fig. 3. Diagram showing the fringe formation and photography apparatus. The Rodenstock retinometer consisting of a laser at L and a variable beam splitter at K focuses two beams using a variable lens at H near a thin mirror, E. The two beams are refocused to a region very close to the principal plane of the crystalline lens when the focusing lights of the Donaldson camera are focused on the anterior lens surface at B by objective lens C and the corneal system. The two beams then interfere to form fringes that can be photographed when they intersect the fundus at A by the fundus camera, C, F, and G.

L = 17 mm; from lens principal plane to model retina: \( \lambda = 0.63 \times 10^{-3} \text{ mm}; n = 1.333; a = 0.38 \text{ mm} \) (Fig. 2). A fringe spacing of 21 \( \mu \text{m} \) was calculated. The measured value fell within one standard deviation (0.4 \( \mu \text{m} \)) of the calculated value allowing for any errors introduced by the film grain size.

Figure 4 is a photograph of the optic nerve head of a normal human eye illuminated by low intensity laser fringes. The fringe spacing was 91 \( \mu \text{m} \) calculated from the measured values \( \lambda = 20.4 \text{ mm}; \lambda = 0.63 \times 10^{-3} \text{ mm}; n = 1.336; a = 0.104 \text{ mm} \) (Fig. 2). Figure 4 shows the fringes at the bottom of a normal cup. The bottom of the cup was in focus so that a slight increase in the axial length for the normal cup depth was added for calculations of the fringe spaces. This increase was \( \frac{1}{2} \text{ mm} \) for cup depth, as estimated from stereophotogrammetric measurements and the depth of field of the camera. Some landmarks show up in the photograph despite the graininess of the high speed film. Laser speckle from the optic nerve also contributes to the speckled appearance. There are vessels on the disc. The margin of the disc is out of focus because the camera has a very shallow depth of field and the focus is at the bottom of the cup. The fringes at the disc rim are very well defined because the camera is focused at this level.

The dimensions for the cup and disc diameter were calculated by comparing the standard fundus photographs taken with the Donaldson camera with the interference fringe photographs. The cup diameter was 16.25 cycles \( \times 0.91 \text{ mm/cycle} = 1.478 \text{ mm} \pm 0.070 \text{ mm} \), and the disc diameter was 21 cycles \( \times 0.091 \text{ mm/cycle} = 1.911 \text{ mm} \pm 0.080 \text{ mm} \). The \( \pm \) error is the maximum error corresponding to 3.29 SD. The errors included a factor of 0.25 cycles because the photographed scale is accurately resolvable only to one-fourth of a cycle.

Discussion

A scale at the fundus with an accuracy sufficient to overcome the uncertainties imposed by ocular optics is desirable for several reasons. First, dimensions in the fundus would be more useful to the clinician if they were known absolutely rather than just relatively, with respect to an earlier photograph or approximately, using a model eye for calculation. Second, comparisons could be made among individuals in large groups under study. Third, time-lapse measurements could be made on individual eyes with confidence that a change in refraction or a change in...
eye size would not obscure the measurement. It might also be possible to measure aniseikonia with this method.

Eyes can vary in their optical characteristics to a significant degree. In a study of emmetropic eyes done by Sorsby,1 the variation in axial lengths and corneal curvatures of normal emmetropic adults indicated a possible variation of approximately 20% in total dioptric power. This variation would cause an approximate variation of 20% in image magnification and a 44% difference in the calculated fundus area between emmetropic eyes at the extremes of the normal range. When nonemmetropic adults are considered, the variation can be much greater depending on the refractive correction needed and on the method of viewing or photographing the fundus (Table 1). It is interesting to look at the magnification differences between an emmetropic infant's eye6 of 90D total power and an average adult's eye of 60D total power. The infant's eye is 50% more powerful than the adult's eye so that the infant's fundus would be magnified 50% more than the adult's. Differences in magnification would also occur in an aphakic eye that had had its crystalline lens removed (Table 1).

A fundus scale can be calculated if the position and power of the optical components of the eye are known. Although our method using the interference fringe scale requires that some data regarding the photographed eye be incorporated into the calculations, the method has the advantage that direct results can be recorded through the camera, eliminating corrections needed for various refractive states.

It should also be noted that other procedures using interference fringes to form a scale at the fundus are possible. Examples are imaging the laser points to the model principal planes of the eye rather than the model principal plane of the crystalline lens or focusing the laser points to the center of curvature of the cornea (after keratometric determination) to eliminate most of the refractive effects. These procedures seem to be approximately equivalent in accuracy; however, these and other methods for measuring the fundus certainly warrant further study.

Other methods7,8 that claim to obtain absolute measurements of the fundus do not measure the optical elements of the photographed eye. Therefore, these methods obscure the effects on fundus measurements of the total optical power of the eye. Bengtsson and Krakau8 have calibrated the Zeiss camera to compensate for the differences in magnification caused by deviations in refractive error. In addressing the problem of differences in magnification caused by variations in the total dioptric power of the eye, however, they have adopted a “normal value” for total eye power that, they claim, eliminates “about 2/3 of those variations in magnification which are caused by variations in the optical properties of the examined eye.”8 The fractional quantity, “about 2/3,” seems to allude to the fraction of the population that falls within one standard deviation on either side of the mean for a normal distribution. However, in order to cover 99.9% or 99.7% of the population, which is a more reasonable goal if absolute measurement is to have clinical significance, ranges of ±3.29 or ±3.0 standard deviations respectively would be needed. In addition, since magnification is proportional to total eye power in emmetropic eyes,9 even using data reported by Stenström10 in a selective study, we have calculated a percent difference of over 6% in lateral magnification for the ±1 standard deviation range (±1.74D). If three standard deviations are used, a variation of about 20% at the extremes is found in the lateral magnification. Even a 6% variation is significant when measuring the optic cup because area is a squared function of lateral magnification,5 i.e., area magnification = (lateral magnification)2. Therefore, a 6% variation in lateral magnification would yield about 12% in area magnification (1.06)2 = 1.1236. Depth magnification (longitudinal magnification) is also a squared function of lateral magnification. Depth magnification, (fundus to aerial image for the Donaldson camera3,4) equals m2/n, where m is the lateral magnification, and n is the effective refractive index of the eye.11 Cup volume depends on a factor that is the fourth power of the lateral magnification, m, = m4/n. Cup volume = m4/n. Depth magnification, (fundus to aerial image for the Donaldson camera) equals m2/n, where m is the lateral magnification, and n is the effective refractive index of the eye.11 Cup volume depends on a factor that is the fourth power of the lateral magnification, m, = m4/n. For the Donaldson aerial image. We would thus find about a 25% error range for cup volumes when we only consider a range of ±1 standard deviation taken from a selective study, (1.06)4/n/1/n = 1.25.

The errors in the interference fringe method can be classified as precision errors in measuring the eye (ultrasound and keratometry) and errors, both in the fringe formation and recording apparatus, and in the assumed position of the principal planes of the crystalline lens.

Coleman12 reported that his axial length and anterior chamber depth measurements by ultrasound were both accurate to within 30 microns. An error

Table 1. Image size

<table>
<thead>
<tr>
<th>Condition</th>
<th>Variation</th>
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<tbody>
<tr>
<td>Emmetropia (adult population)</td>
<td>20%</td>
</tr>
<tr>
<td>Ametropia (±6D)</td>
<td>50%</td>
</tr>
<tr>
<td>Age (90D infant vs 60D adult)</td>
<td>50%</td>
</tr>
<tr>
<td>Aphakia (15D)</td>
<td>25%</td>
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Source: Percent variation for emmetropia was calculated from the extremes (±SD) of optical powers found in a normal adult population by Sorsby.1 The data for infant eyes were taken from Davson.8

20% variation
25% variation
50% variation
50% variation
20% variation

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of this magnitude will yield two errors: one in fringe spacing of 0.6% because of the limitations of the axial length measurement and the other of 0.15% because of anterior chamber tolerance.

In our study keratometric measurement was limited by the precision of the scale, so that it was accurate to \( \frac{1}{4} \)D. The power of the corneal-anterior chamber system (43.08D) divided by \( \frac{1}{4} \)D equals approximately 0.6%, which would also be the error in the fringe spacings.

If optimal precision is achieved, errors attributable to the limitations of ultrasound and the keratometer are on the order of 1.5%.

Our method uses a model lens for principal plane calculations so that large errors might be expected from this source. This is not the case, however. For example, if the model lens principal planes differ from the real principal plane by as much as 1 mm, there would be a negligible error in the fringe spacing arising from the use of model linear dimensions rather than real linear dimensions. This error would be about 0.1% and can be expressed as \( \frac{\Delta d + \Delta L}{L} \) (Fig. 5), where \( \frac{1}{\Delta d} + \frac{1}{\Delta L} = \frac{1}{50} \) mm and where \( \Delta L \) is a negative quantity. This error would result from the images of the laser points being at a focus on what could be thought of as quasi-principal planes (Fig. 5), which would differ from the model by only a small amount. There is, however, a small but not negligible error in the fringe spacing caused by the difference in magnification of the quasi-principal plane images. This error would be about 2% if the model lens were 1 mm different from the real lens in principal plane location. If the principal plane is missed by \( \Delta d \), errors can be calculated as follows: if \( \frac{1}{\Delta d} + \frac{1}{\Delta L} = \frac{1}{f} \) for \( f = 50 \) mm and \( d = 1 \) mm, then the error in fringe spacing is \( |\Delta L/\Delta d| - 1 \approx 2\% \).

The optic nerve head is about 15° off the eye’s optical axis. This angle will cause a small, but not negligible, change in the fringe spacing, about 2%, as calculated from a model eye. We can, however, compensate for this by increasing the fringe spacing by this amount, by using this correction factor in the fringe-spacing equation.

The method combines standard ophthalmological devices and measurements that are all accepted as safe. The Rodenstock retinometer is certified in accordance with the Radiation for Health and Safety Act of 1968 and has such a low intensity (10 microwatts) that it is well within the standards set by the American National Standards Institute (ANSI). The laser points are restricted from focusing on the retina by the mechanical travel of the camera, but even if they did, the intensity is low enough to be well within the ANSI standards. The low light levels of the device required large aperture size, fairly long exposure times, and high film speeds (\( \frac{1}{2} \) sec and 1000 ASA, respectively). The fundus camera had a very shallow depth of field because the apertures were opened up. This effect should not be attributed to the fringes’ being out of focus because they have an infinite depth of focus. The long exposure time was tolerable because the fringes were fixed with the fundus camera so that the fringes would not double expose with small eye movements. The photographic equipment was set up to test the feasibility of the method. We were able to obtain promising results; however, the procedure required very cooperative subjects, and we do not believe it is ready for clinical use. Exposure times could be lowered further by use of photonic amplification or high speed charge-coupled-device (CCD) imaging sensors in place of the photographic film.

In conclusion, although the method is still in the developmental stage, it has so far shown itself to be feasible and safe with the promise of becoming a reliable way to obtain absolute measurements of fundus dimensions.

**Key words:** fundus, fundus measurement, interference fringes, interference fringe scale, optic nerve head measurement, absolute fundus measurement.

**Acknowledgment**

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**References**


