Optical Measurement of the Axial Eye Length by Laser Doppler Interferometry

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A new technique has been developed to determine the axial length of the human eye in vivo. Based on laser interferometry in conjunction with the Doppler technique, it uses partially coherent light. This new technique complies with laser safety regulations. High accuracy is achieved, the optical length (OL) can be determined within ± 30 µm, and the reproducibility of the geometric eye length is greater than ± 25 µm. Possible errors are discussed. First comparisons with the ultrasound technique yield good agreement for emmetropic subjects and for subjects with a myopia of up to 10 diopters. The advantages of the laser doppler interferometry (LDI) technique are high accuracy, high transversal resolution, and more comfort for the patient (it is a noncontact method; no anesthesia is needed). Possible future applications of LDI, like measurements of fundus profiles and of retinal thickness, are mentioned. Invest Ophthalmol Vis Sci 32:616–624, 1991

Determination of intraocular distances, especially of the axial length of the eye, is needed in several cases. For example, the most important of these applications is the determination of refractivity of intraocular lenses after cataract surgery. The standard technique used to measure intraocular distances is the ultrasonic echo-impulse technique (US technique). Since the first measurements as early as 1956, this technique has been steadily improved and is now a standard clinical technique. There are several advantages of the US technique, eg, the ability to measure the axial length and other intraocular distances, or the possibility to measure social length through cataract lenses. Drawbacks include the need for mechanical contact between the transducer and the eye, and therefore the need of anesthesia, or the limited transversal resolution of the ultrasound beam.

In the past, noninvasive optical techniques were only used to measure the depth of the anterior chamber and the thickness of the cornea. New methods for the measurement of intraocular distances were reported. The femtosecond optical ranging technique was used to determine the thickness of the cornea of anesthesized rabbit eyes in vivo. A modification of the slit-lamp technology was proposed to measure the thickness of the human retina. Topographic measurements of the human optic nerve head became available by computerized video- graphic image analysis (see references 6–8). The axial length of the human eye was measured by interferometry with partially coherent light. The advantages of this method were a high longitudinal accuracy and a high transversal resolution. In addition, it is more comfortable for the patient, because it is a noncontact method; thus, no anesthesia is needed. The main drawback of this method is the laborious procedure of measurement. The operator has to shift interferometer plates step by step (1 step about 25 µm) over a distance of about 5 mm. At each step, the operator has to look for an interference pattern of poor contrast. A skilled operator needs about 15 min for one measurement. Because this is as tiresome for the investigator as for the patient (the patient has to look into the laser beam the whole time), this method is not applicable for a larger number of patients, especially elderly people.

In this work, the interferometry technique was improved to avoid its main drawback. To achieve this, the eye length is no longer determined by the interpretation of static images, but by a dynamic approach with use of the laser Doppler principle. This article presents the laser Doppler interferometry (LDI) technique; shows that it works in vivo for emmetropic eyes and varying degrees of myopia; discusses its accuracy; discusses the origin of the signals; and shows first comparisons with the US technique. Additionally, possible future applications, such as the measurement of the retinal thickness and of fundus profiles, are discussed.
Materials and Methods

Description of the Method

Figure 1 shows a sketch of the instrument. A multimode semiconductor laser diode (MMLD) emits a light beam (wavelength $\lambda = 780$ nm, power $\approx 250 \mu$W) with high spatial coherence but short coherence length (CL). This beam passes a Fabry Perot interferometer that splits the beam into two parallel, coaxial beams: a direct beam (1) and a second beam (2) that is reflected once at both interferometer plates and hence is retarded by a path difference of twice the plate spacing $d$.

Both beams illuminate the eye through a beam splitter cube (BSC). They serve as measuring beams as well as a fixation target; the subject with the head fixed by a bite-board looks at the beam (it appears as a weak red spot; the wavelength of 780 nm is just visible). The two coaxial beams are reflected at both the cornea and the retina yielding four reflected beams. This introduces an additional path difference of twice the optical length (OL) of the eye between the two beams reflected at the cornea and the two beams reflected at the retina, respectively. (OL is defined in this technique as the path integral $\int n_g \times ds$ of the group refractive index $n_g$ of the eye media along the geometric path from the anterior surface to the retina.)

If the CL of the laser is shorter than $2 \times$ OL, as in this case, the wave fronts from the retina and the cornea do not interfere. However, if the two path differences $2d$ and $2 \times$ OL equal each other (within a difference of CL), two of the four reflected beams will interfere: the part of beam 1 that is reflected by the retina (beam 1R) and the part of beam 2 that is reflected by the cornea (beam 2C). The interference pattern that results will consist of concentric fringes. These fringes can be seen with the infrared (IR) scope. In principle, the eye length can be measured by a shift of one of the interferometer plates in steps of $\text{CL}/2$. At the position where the fringes are seen, $d$ is measured and $\text{OL} = d \pm \text{CL}/2$. Actually, the first measurements of the eye length, with the use of a precursor of the technique described here, were carried out in this manner.9,10

An electronic detection of interference fringes to indicate the balance of the two path differences, OL and $d$, is difficult and involves pattern recognition techniques with large computational effort. For this reason, a dynamic approach is used rather than an interpretation of static fringes; instead of shifting one of the interferometer plates in steps and checking whether interference fringes appeared, the plate is shifted with constant speed. This causes a Doppler shift $f_D = 2v/\lambda$ (plate speed $v$) of the light frequency of beam 2. If the path difference between beams 1R and 2C is less than CL, they will interfere (as mentioned above), but the intensity of the interference pattern will now be modulated by the Doppler frequency $f_D$.

A photodetector (PD) measures the intensity of the superimposed reflected beams at the point where the center of the interference fringes appears, if the interference condition $2 \times |\text{OL} - d| < \text{CL}$ is fulfilled. The signal is amplified and passed through a band pass filter that transmits only signals with frequency $f_D$. The intensity $I$ of the filtered signal is recorded as a function of the distance of the interferometer plates $d$ by a personal computer system. The value of $d$, where $I$ is a maximum, is equal to the value of OL.

Two additional lasers are included in the interferometer for alignment purposes. A helium neon laser (Melles Griot, Irvine, CA) HeNe is included for coarse alignment of the subject’s eye. For the fine alignment of PD into the center of the interface...

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Fig. 1. Block diagram of the laser Doppler interferometer for measuring the axial length of the eye. HeNe, helium neon laser; SMLD, single mode laser diode; MMLD, multimode laser diode; BSC, beam splitter cube; FPI, Fabry Perot interferometer; PD, photodetector; AMP, amplifier; PC, personal computer; SM, stepper motor; IR, IR-scope.
fringes, a single mode laser diode (SMLD, Sharp Corp., Osaka, Japan) is used. This laser has the same wavelength ($\lambda = 780$ nm) as the measurement laser (MMLD), but with a CL larger than $2 \times \text{OL}$. In this case, the beams reflected at the cornea and the retina interfere permanently, independent of the plate distance $d$, and since $\lambda$ is the same, the position and size of the interference fringes are equal. Therefore, the photodetector, which is mounted on an x-y translation stage and whose position can be seen together with the interference fringes via the IR scope, can be aligned with the center of the fringes before the measurement. A crude alignment is sufficient because the diameter of the detector ($\approx 50$ $\mu$m) is smaller than the width of the first few fringes off the center. Once the alignment is completed, the SMLD is switched off, the MMLD is switched on, and the measurement is started, all by a single footswitch.

Accuracy of the Method

As mentioned above, a signal with Doppler frequency is obtained in the range $d = \text{OL} \pm \text{CL}/2$. Therefore, the precision of this technique is determined by the CL of the MMLD, about 110 $\mu$m. The amplitude of the signal varies within the range $d = \text{OL} \pm \text{CL}/2$ and is at maximum when $d = \text{OL}$. Therefore, if the maximum amplitude of the signal is searched for, the precision can be higher than CL/2.

The instrument is calibrated by the measurement of the width of the first few fringes off the center. Once the alignment is searched for, the precision can be higher than CL/2. The sound velocities were taken 1532 m/sec for the aqueous and the vitreous and 1641 m/sec for the lens. These data are first results of a detailed comparative investigation that was carried out in collaboration with Professor HD Gnad, Department of Ophthalmology, Lainz Hospital, Vienna, Austria.

In Vivo Measurements

The laser safety regulations have to be met to carry out in vivo measurements. The HeNe alignment laser delivers a power of about 1 $\mu$W to the fundus, equivalent to an intrabeam viewing, with a pupil of 7 mm diameter, of a beam with a power density of less than 3 $\mu$W/cm$^2$. This is below the limit of permanent illumination of 18 $\mu$W/cm$^2$. Since the coarse alignment takes about 20 sec, the safety regulations are met. The SMLD delivers a power of 70 $\mu$W to the fundus and yields a power density of about 180 $\mu$W/cm$^2$, if averaged over an aperture of 7 mm diameter. This is permitted for several hours for $\lambda = 780$ nm. The fine alignment takes about 20–30 sec.

The MMLD, turned on during the measuring period only, has a higher intensity of about 250 $\mu$W or 650 $\mu$W/cm$^2$ (average over 7 mm aperture) that is allowed for about 4 min. During one measurement, a distance of 5 mm is scanned with the interferometer plates at a speed of 1.85 mm/sec, so the duration of the measurement and the duration of laser illumination is less than 3 sec, ie, well below the limit.

In vivo measurements were carried out on seven volunteer subjects (the right eye in each case), from whom full informed consent was obtained. To their knowledge, they had no eye diseases. Three of them were emmetropic, 2 had a low degree of myopia (1 or 2 diopters), 1 had myopia of 4 diopters, and 1 had myopia of 10 diopters. In the last two cases, the subjects had to wear their spectacles, otherwise, no signal could be obtained. (If the beam is not focused on a spot of the retina but is smeared out, the reflected beam will consist of several individual beams reflected from different points of the retina, and no interference pattern can be seen.) The spectacles do not influence the measurement if no specular reflections from their surfaces are reflected into the photodetector.

Ultrasound Measurements

A first comparison with the usual ultrasound technique was performed. A Kretztechnik 7200 MA Hochfrequenz Echograph (Kretztechnik, A-4871, Zipf, Austria) was used, employing a water-immersion technique. The sound velocities were taken 1532 m/sec for the aqueous and the vitreous and 1641 m/sec for the lens. These data are first results of a detailed comparative investigation that was carried out in collaboration with Professor HD Gnad, Department of Ophthalmology, Lainz Hospital, Vienna, Austria.

Results

Calibration of the Instrument

Figure 2 shows an example of a calibration measurement. The optical thickness of a plane parallel glass plate of known thickness and refractive index is measured. A distance of 5 mm is scanned by the interferometer plates, from 30.2–35.2 mm. The signal intensity I (ordinate) is plotted as a function of the interferometer plate spacing $d$ (abscissa). The width of the signal at half maximum height is about 110 $\mu$m, but the position of the maximum can be determined with a higher precision. A cursor readout on the computer is used to determine the position of the peak, and a standard deviation of $\pm 3$ $\mu$m is obtained for five consecutive measurements. The cursor readout is then adjusted to show the known value of the optical length at the position of the signal peak. This calibration is used for the subsequent in vivo measurements.
Fig. 2. Plot of intensity I vs. optical length OL in the case of a plane parallel glass plate. The peak of the signal is at 33.120 ± 0.003 mm. The subsidiary peaks in distances of multiples of 1.1 mm from the main peak are artefacts of the MMLD.

The additional small peaks in distances of multiples of 1.1 mm from the main peak are artifacts of the laser diode and are caused by a periodic repetition of the coherence function of the laser. Since the distance from the main signal is constant, confusion can easily be avoided.

Optical Measurements

Figure 3 shows a measurement of the OL of a subject with a myopia of about 2 diopters. The width of the signal peak is about the same as that of the plane parallel glass plate (cf. Fig. 2). The signal-to-noise ratio (S/N) is less than that obtained from the glass plate. This leads to a greater error when the position of the peak is determined. A greater reproducibility can be achieved if the center of the signal peak is determined instead of the position of the signal maximum, since the overall shape of the signal is less influenced by statistical noise than the position of the maximum. Therefore, the center of the peak was determined throughout all the optical measurements. The center of the peak is at about 33.55 mm (the exact position is determined by a cursor readout: 33.534 mm). Five consecutive measurements yield a standard deviation (SDEV) of about ± 22 μm. The OL of the eyes of the other six subjects was determined in a similar way. No major difficulties were encountered when the subjects were measured who had a myopia of 4 and 10 diopters, respectively, if they wore their spectacles. Only the S/N ratio was slightly poorer than in the case of emmetropic sub-

Fig. 3. Plot of I vs. OL in the case of an eye with a myopia of 2 diopters. Position of peak center: 33.534 mm.
Table 1. Eye length determined optically and acoustically

<table>
<thead>
<tr>
<th>Subject</th>
<th>Ametropia (diopters)</th>
<th>OL (LDI) (mm)</th>
<th>GL (LDI) (mm)</th>
<th>GL (US) (mm)</th>
<th>Difference (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>G.G.</td>
<td>0</td>
<td>31.19</td>
<td>23.01</td>
<td>22.82</td>
<td>0.19</td>
</tr>
<tr>
<td>A.F.</td>
<td>-2</td>
<td>33.56</td>
<td>24.78</td>
<td>24.64</td>
<td>0.14</td>
</tr>
<tr>
<td>C.H.</td>
<td>-1</td>
<td>34.53</td>
<td>25.50</td>
<td>25.54</td>
<td>-0.04</td>
</tr>
<tr>
<td>H.S.</td>
<td>0</td>
<td>32.75</td>
<td>24.17</td>
<td>24.36</td>
<td>-0.19</td>
</tr>
<tr>
<td>M.J.</td>
<td>0</td>
<td>31.91</td>
<td>23.55</td>
<td>23.60</td>
<td>-0.05</td>
</tr>
<tr>
<td>L.S.</td>
<td>-4</td>
<td>33.91</td>
<td>25.04</td>
<td>25.04</td>
<td>0.00</td>
</tr>
<tr>
<td>K.L.</td>
<td>-10</td>
<td>38.46</td>
<td>28.42</td>
<td>28.37</td>
<td>0.05</td>
</tr>
</tbody>
</table>

OL, optical length; GL, geometrical length; LDI, determined by laser Doppler interferometry; US, determined by ultrasound.

Standard deviations: OL (LDI) ≤ 30 μm; GL (LDI) ≤ 25 μm; GL (US): accuracy = 120 μm13 to 200 μm14.

Measurement of a Fundus Profile

In one case, the fundus profile was measured, i.e., the distance from the cornea to different points of the retina. A second beam of green light that could be tilted about two perpendicular axes acted as a fixation target for the second eye. Figure 4 shows OL as a function of the angle between the vision axis and the direction of measurement. The measurement was carried out in steps of 2° from 24° nasal to 22° temporal (larger angles yielded no acceptable signal quality). Each spot corresponds to the mean value of five consecutive measurements. Error bars are indicated. The solid line indicates the theoretic fundus profile (see Discussion).

Observation of Double Peaks

In some cases, two peaks occur during a measurement. Figure 5A shows such a case. The second peak is the usual one, but the first one appears sometimes. With some subjects, the first peak occurs frequently; with others, seldom or not at all. The distance between the two peaks equals an optical length of 256 ± 27 μm. If the measurement is done with an angle of 5° (Fig. 5B) between vision axis and measurement axis, the distance between the peaks increases to OL = 384 ± 25 μm. At angles greater than 8°, the first peak disappears.

Ultrasound Measurements

US measurements were performed on the same subjects as the LDI technique. One measurement was performed on each eye during this preliminary study. The results are shown in Table 1. The reported accuracies for this technique range from ± 120 μm for measurements of aphakic eyes13 to ± 200 μm for measurements of cataractous eyes14 (the latter used the technique in this work).

Discussion

A new method to determine the axial length of the eye has been developed. It is based on interferometry...
methods in conjunction with the laser Doppler technique. The method is noninvasive, and no contact between the eye and the sensor is needed. The method is based on laser intensities in compliance with laser safety regulations. Reproducible results were obtained with emmetropic subjects and with subjects who had up to 10 diopters of myopia. Hyperopic subjects have not been studied but there should be no major difficulties if these subjects wear spectacles with proper refractivity.

Accuracy of the OL

The accuracy of the OL of the eye depends on the calibration of the instrument and on the accuracy of the center of the signal peak. Since the calibration can be done with an accuracy of about ± 3 μm and the center of the peak can be determined within about ± 30 μm, the error that results will not exceed ± 30 μm (standard deviation of single measurements in each case). Errors due to small deviations of the measurement direction from the optical axis (because of saccadic movements of the eye between the individual measurements) are included in this standard deviation. Larger deviations could occur if the patient is unable to cooperate and look into the measuring beam, but this case can be recognized by a larger standard deviation.

Conversion of OL to Geometric Length (GL)

For the comparison with the US measurements and for ophthalmologic purposes, the OL has to be converted into the GL of the eye. The OL is usually
the eye, consist mainly of water. Therefore, the dispersion of water at 20°C that was used for the OL-to-GL conversion throughout this work. Secondly, only the total OL of the eye was measured, not the values of the individual eye media (although this is possible in principle).

Approximations were made to overcome these difficulties. First, the eye media aqueous and vitreous, which contain the major part of the light path within the eye, consist mainly of water. Therefore, the dispersion of water is used to calculate the ng values for these media for λ = 780 nm. Because of a lack of dispersion data for cornea and lens, the water dispersion was also used for these media, although their water content is considerably smaller. Since their GL values are only about 0.5 and 3.6 mm, respectively (values from Gullstrand’s schematic eye), the possible error is small. If these media had no dispersion in the range 550–780 nm, the GL values were only about 35 μm shorter than in the case of the assumed water dispersion. Table 2 shows the ng values (calculated from the values of 550 nm at 37°C and from the water dispersion [at 20°C]) that were used for the OL-to-GL conversion throughout this work. Second, the OL-to-GL conversion is based on Gullstrand’s schematic eye. The schematic eye has a value of GLS = 24 mm and a mean group refractive index (nGS = 1.3549) that yields OLs = 32.518 mm. It was assumed that individual deviations from this value are due to variations of the length of the vitreous. To calculate the deviation δGL of the actual GL from GLs, the difference between the actually measured OLm and the schematic value OLs) is divided by the group refractive index of the vitreous nGV. δGL = (OLm - OLs)/nGV. GL is then obtained simply by addition: GL = GLs + δGL.

The GL values calculated in this manner are also shown in Table 1; their standard deviation is less than ±25 μm in each case.

Table 2. Group refractive indices ng of the eye media for λ = 780 nm

<table>
<thead>
<tr>
<th>Cornea</th>
<th>Aqueous</th>
<th>Lens</th>
<th>Vitreous</th>
<th>Mean value of schematic eye</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.3856</td>
<td>1.3459</td>
<td>1.4070</td>
<td>1.3445</td>
<td>1.3549</td>
</tr>
</tbody>
</table>

Since the thickness of the cornea is nearly constant in different eyes and the refractive index of aqueous and vitreous is almost the same, the only major error can arise from the uncertain value of the lens thickness tL. The US measurements in this work yielded tL values that are up to 0.9 mm larger than the value 3.6 mm of the schematic eye. These US values of tL might be too high; the eyes were possibly accommodated since the subjects might have tried to look at the US transducer (in contrast to the LDI measurements, where the subjects look at a point at virtual infinity). In spite of this, the US values can be used to estimate the largest possible error from the differences of the ng values of lens and vitreous (Table 2): if tL is 0.9 mm longer than the value of the schematic eye, a GL value calculated as above has to be reduced by about 40 μm.

Origin of Reflections and Determination of Retinal Thickness

If different techniques for the measurement of the eye length are compared, it is important to know from which histological structures the signals originate. These could vary for different physical phenomena—in this case, light and sound.

In the case of the LDI technique, sometimes two signals are seen. Their widths are about the same as those of the glass plate; hence, they must originate from single, discrete interfaces. The distance of these signals varies with the angle between vision axis and direction of measurement (Fig. 5). This can indicate the origin of the signals. Two reflections from the retina were shown to originate from structures anterior, and posterior to retinal blood vessels, respectively. Therefore, the main signal of LDI could originate from the retinal pigment epithelium (RPE), whereas the occasional first signal could be caused by the internal limiting membrane. In this case, the distance between the two signals should have a minimum at the fovea centralis, which is in accordance with the experimental findings. If the reflection from the internal limiting membrane is specular (which can be assumed if it is caused by a difference of refractive index at the interface), it is reasonable that it does not always occur. Its occurrence depends on the momentary orientation of the eye during a saccadic movement. At greater angles, the specular reflected light would not be transmitted back through the pupil which is in accordance with the present findings. This specular reflection is stronger with dark-pigmented subjects. On the other hand, the main reflection assumed to originate at the RPE is not specular. It consists of a directed and a diffused part; the latter dominates outside the fovea centralis. Therefore, it can pass through the pupil at higher angles.
The observation of two signals is not only important for the determination of the origin of the reflections, but it can also be used to measure the thickness of the retina. Demands for the measurement of the retinal thickness noninvasively led to a new approach. The reproducibility of in vitro tests with this technique, based on slit-lamp technology, was ± 9 μm. The reproducibility of the LDI method in vivo is ± 30 μm for OL. Since the accuracy of LDI is determined by the CL of the laser, the reproducibility can be improved by the replacement of the MMLD by a super radiant diode (SRD), whose CL is about five times shorter. The accuracy of the retinal thickness measurement by LDI can be improved to about ± 5 μm.

An advantage of LDI would be that no contact lens is needed. However, the fact that the first signal is not seen in every subject causes problems. This problem may be overcome with the use of a photodetector with a greater S/N ratio. A problem common to the LDI and slit-lamp technology is the unknown refractive index of the retina, which prevents an exact determination of the geometric thickness of the retina. However, this factor seems to be of minor importance, since a distinction between normal and damaged retina could also be based on OL values.

In some cases, a third signal peak is seen. It is smaller but broader than the other two peaks and is located behind the main peak. It is assumed that this peak is caused by structures of the fundus behind the RPE. Further studies are necessary to clarify the origin of this signal.

Comparison with the Ultrasound Technique

The comparison of the two techniques shows good agreement. The maximum deviation is about 200 μm and is within the range of accuracy limits reported for the US technique. Since no systematic deviation is seen, it can be concluded that the two signals, sound and light, are reflected at the same layer. Furthermore, the assumptions made in the calculation of the GL values from the OL values are supported. However, since the US values are based on single measurements, this comparison has to be regarded as preliminary. A detailed comparison of both techniques that will yield statistically more relevant data was carried out in collaboration with Professor HD Gnad, Department of Ophthalmology, Lainz Hospital, Vienna, Austria. The results will be published.

In this section, the advantages and drawbacks of the two methods are discussed. One advantage of the US technique is the ability to measure additional intraocular distances, especially the depth of the anterior chamber and the thickness of the lens. These measurements were not carried out by LDI because of the low S/N ratio of the photodetector used. However, they are possible in principle with the improvement of the S/N ratio, eg, first results with a slightly modified experimental setup showed the feasibility of the measurement of the thickness of the cornea by LDI. Another advantage of the US technique is the ability of the sound waves to penetrate mature cataract lenses. Cataract lenses scatter light and make it difficult to use laser interferometry. These difficulties might be overcome partly, because the LDI technique used near infrared light of λ = 780 nm. The transmittance of this long wavelength is greater than that of short wavelengths through old lenses, which can be assumed to be comparable with early stages of cataract. Therefore, the LDI measurements should work at least with early stages of cataract. By use of longer wavelengths of about 1100 nm, the penetration of cataract lenses could be further improved because of the lower light scattering in this case.

The advantages of the LDI technique are that it is a noncontact method; therefore, the eye is not deformed during the measurement; this excludes an additional error source. Also, the necessity to anesthetize the cornea is avoided; thus, the patient is more comfortable. In addition, high longitudinal and transversal resolution is achieved; a relative accuracy of better than 25 μm is achieved for the geometric eye length that can further be improved with use of an SRD. Since the laser beam can be focused at a spot of about 10 μm diameter on the retina, a transversal resolution of this magnitude seems possible. The diameter of the US beam at the retina is difficult to determine, since the change from a near-field to a far-field pattern occurs at a distance from the transducer of about 20–40 mm, ie, in the range of the axial eye length. However, from the diffraction theory it can be estimated that, for a wavelength of 150 μm and 5 mm diameter of the transducer, the minimum diameter of the US beam in a distance of 25 mm is about 1 mm, ie, two orders of magnitude more than the laser spot (smaller diameters would require a shorter wavelength and/or a larger transducer diameter). Therefore, measurements of fundus profiles with significantly better resolution should be possible with the LDI method. Finally, the LDI method allows measurements of the thickness of the retina.

Fundus Profile Measurements

A first feasibility proof of a fundus profile measurement, although still with low resolution, is shown in Figure 4. In addition to the measured values, a theoretical fundus profile is plotted in the figure ( ). It is calculated on the basis of Gullstrand's schematic eye: the radius of curvature of the cornea and the location of the nodal points of the schematic eye were used. Only the radius of curvature of the retina was
matched to yield an overall optical length equal to the OL of the subject's eye at an angle of 0° between vision axis and direction of measurement. The \( n_g \) values of cornea, lens, aqueous, and vitreous were taken from Table 2.

The overall agreement between measured values and the theoretic curve is good. The deviations at an angle of about 12°–14° nasal could be caused by an excavation of the optic disk. It should be mentioned, however, that in the region of the optic disk, the signal looks different from the usual narrow signal peaks. The narrow peak is replaced by a broad maximum that indicates that the light is reflected from several layers, displaced about 230 \( \mu \)m along the measuring direction. For this reason, the error bars for these angles are larger. The additional halfwidth of the peak is taken into account in these cases. At 16° nasal, no reproducible result was obtained.

The main purpose for fundus profile measurements is their possible use in glaucoma diagnostics. In this case, the topography of the optic nerve head has to be determined. The most advanced technique for this purpose is the computerized videographic image analysis (CVIA) (see references 6–8). Extensive studies of normal, glaucoma suspect, and glaucomatous eyes were carried out with this technique, and the results showed that new parameters that describe the surface contour of the peripapillary nerve fiber layer are better than standard structural parameters in their ability to distinguish normal from glaucomatous eyes. 8

No comparison between normal and glaucomatous eyes has been carried out by LDI. Therefore, it is difficult to determine whether LDI will be able to compete with CVIA, especially with regard to the speed of the measurement. It should be possible to measure 20 \( \times \) 20 points in about 100 sec by increasing the speed of the stepper motor and the computer program and by the automatic tilt of the measuring beam. In this case, LDI could have advantages; a higher depth resolution (10 \( \mu \)m or better with an SRD); the ability to measure through undilated pupils (because the beam diameter can be reduced to 1 mm); and possibly the direct measurement of the thickness of the nerve fiber layer (if the width of the broad signal peak seen at the optic disk corresponds to the thickness). These features will be available after more study.

**Key words:** biometry, oculometry, eye length measurement, laser interferometry, laser Doppler interferometry

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