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The contact and immersion ultrasound methods compared using the ray tracing method

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The goal of this article was to compare the contact and immersion ultrasound methods using simulated data. The pursued changes were interrelated with both the axial length of the eye and with partial biometry parameters. The results were compared with empirical data from real eye samples. The main analysing method was a modified algorithm for the ray tracing method created in the DELPHI programming environment (Borland Enterprise). The sample included 129 eyes (64% women and 36% men) of average age 73.65 (SD 9.33) scheduled for surgical removal of cataract. The average axial length was 23.12 mm (SD 1.31). The methods were compared using the simulated movement of the probe from central and coincident positions. We confirmed the tendency of the contact method to register more scattered beam which provides distorted biometric data from the periphery. This was verified by the real data analysis. The average axial length of the eye was 23.12 mm (SD 1.315), measured by the contact method and 23.26 mm (SD 1.298), measured by the immersion method. The difference between the methods was 0.145 mm. The most important changes occur in the vitreum depth which correlated with the total axial length ($r = 0.89$). The ray tracing method provided evidence of greater accuracy of the immersion method which was more sensitive to probe displacement and provided more accurate data. The axial length of the eye was longer according to the immersion method but this had only secondary influence on the accuracy of the method. Applanation of the cornea is the primary source of the contact method inaccuracy. The vitreum depth was the most influenced.

Keywords: biometry, ultrasound measurement, immersion and contact method, ray tracing.

1. Introduction

Biometrical measurement is an everyday routine in ophthalmological practice. One major reason for the measurement is in planning the surgical removal of cataracts. The biometric parameters obtained from ultrasound equipment describe the inner structure of the eye and these parameters are unique for every eye. In reality, we are unable to measure the inner structure of the eye accurately and some generalisations are necessary. This generalization is very important especially in the intraocular lens implant formula. Each formula uses the biometric data as input parameters and these parameters play the main role in computing the power of the implant dioptric. The accuracy of the biometric data is therefore directly connected with the postoperative refraction of the eye.

We use two different methods to measure the axial position of the eye – ultrasound and optical. The ultrasound method is based on intra-ocular ultrasound wave propagation. The wave impacts a surface which separates two media with different velocity characteristics. A part of the wave is refracted through the surface and the second part is reflected back. The analysis of the reflected wave enables us to describe the axial position. The optical method is based on intra-ocular light propagation. Light rays impact surfaces with different refractive indices. Reflected rays interfere with each other. As in the first case, the analysis of the interference ray provides knowledge of the axial position.

The ultrasound measurement is widely used owing to its simplicity. There are two different approaches for setting the measuring probe. The probe can be placed on the cornea (the contact method) or we can make artificial surroundings where we immerse the probe (the immersion method). The immersion method is more accurate than the contact method. The contact method causes applanation of the cornea. The applanation reduces the anterior chamber depth and also influences a proportion of the axial length [1]. The applanation distorts the measuring process and the calculation of the final dioptric power of the IOL is therefore inaccurate. This error reduces the benefit of the whole implanting process and the possibility of inducing required axial refraction is not used. According to OLSEN [2] the inaccuracy of the IOL determination is 54% caused by the distorted axial length and 38% caused by inaccurate determination of the depth of the anterior chamber.

In various articles it is claimed that using the optic method (IOLMaster) is more accurate than using the immersion method and that the contact method is the least accurate. While these findings have been confirmed by several authors no consensus exists and there are more non-uniform conclusions in biometric practice [3].

HILL [4] and VETRUGNO *et al*. [5] favour all except the contact method. The work of ELEFTHERIADIS [1] compares the optical method with the contact ultrasound method. The result shows that the optical method lengthens the axial length of the eye (23.36 mm, SD 0.85), compared to the contact method (22.89 mm, SD 0.83). HŘEBCOVÁ, Vašků [6] found that the average axial length of the eye was 23.28 mm measured by the contact method and 23.38 mm measured by the immersion method (the difference was 0.1 mm). This was confirmed by WATSON and ARMSTRONG [7], by HOFFMANN *et al*. [8] with a difference of 0.15 mm and by OLSEN and NIELSEN [9] with a difference of 0.14 mm. Comparing the methods, KRONBAUER *et al*. [10] describe the average axial length as 23.19 mm (SD 1.32) using the immersion method and significantly less, 22.93 mm (SD 1.31), using the contact method. BEN-ZION *et al*. [11] evaluated the accuracy of the contact and immersion methods according to the final visual acuity. They found no difference between the two methods. This was confirmed by HENNESSY *et al*. [12] who found a difference of 0.03 mm between them. NARVÁEZ *et al*. [13], PACKER *et al*. [14], KISS *et al*. [15] and HAIGIS *et al*. [16] consider the immersion method and the optic PCI method are equal.

This article, in contrast, describes fundamental differences between the immersion and contact methods. It also describes what changes occur when the probe is moved. Three models of the eye – hypermetropic, myopic and emetropic are used and the modified ray tracing method for ultrasound is used to provide a detailed description.

2. Methods

2.1. Measuring devices

The biometrical data were obtained on an OcuScan instrument (Alcon Surgical) set at 20 MHz. This instrument provided the biometrical data on the anterior chamber depth, lens thickness, the depth of the vitreum and axial length of the eye. The following ultrasound velocities at 37 °C [17, 18] were used (see Tab. 1).

T a b l e 1. Ultrasound velocities.

Setting	Ultrasound velocity [m/s]
Cornea	1620
Aqueous humour, vitreum	1532
Lens	1641
Physiological solution (distilled water, 37 °C)	1523

The value of the cataract density was 1629 m/s. The corneal data were measured on an autorefractokeratometer Canon RK-3.

2.2. Ray tracing method applied to ultrasound

The basic ray tracing method developed by SPENCER and MURTY [19] is modified for light propagation through a rotatively symmetrical surface which is described by the multinomial of the arbitrary root. The surface can be tilted in space and can be described by cylindrical notation. We used the ray tracing method for a centred optical system and only the spherical notation of the surface. The aspherical description introduces complexity to no advantage as the probe movement was 1.2 mm in the area. The main principle is based on the iterative calculation for a point represented by coordinates in space and by direction cosines. We started with $A_0[X_0, Y_0, Z_0]$ and we needed to find a point $A[X, Y, Z]$ which was at s_i distance with (k, l, m) direction cosines:

$$X = X_0 + ks_i$$
$$Y = Y_0 + ls_i \qquad (1)$$
$$Z = Z_0 + ms_i$$

Each movement of s_i is determined as the vertex distance of i surface to $(i-1)$ surface. We needed to find the specific impact point where refraction or reflection occurs. We found this point iteratively on each surface. We discovered that the impact point was mathematically complicated for some surfaces. The Newton−Raphson iterative algorithm is used for this purpose. We iteratively found the impact point on the F surface, defined as:

$$F(X, Y, Z) = Z - \frac{c\rho^2}{1 + \sqrt{1 - \kappa c^2 \rho^2}} \tag{2}$$

where ρ is the spherical invariant $(X^2 + Y^2)$, κ defines the surface type (hyperbolic, parabolic, *etc.*) and c is the vertex radius defined as $1/r$ (r – the radius of curvature). As mentioned before, because we use spherical surfaces we set $\kappa = 1$. The iterative algorithm has a convergence to the impact point and we obtained sufficiently precise results from 10 computing cycles. Using j which represents the iterative number, the computing equation is:

$$s_{j+1} = s_j - \frac{F(X_j, Y_j, Z_j)}{F'(X_j, Y_j, Z_j)} \tag{3}$$

where, from the Eq. (1),

$$X_j = X_1 + ks_j$$
$$Y_j = Y_1 + ls_j \tag{4}$$
$$Z_j = Z_1 + ms_j$$

and where

$$F'(X_j, Y_j, Z_j) = \frac{\mathrm{d}F}{\mathrm{d}s_{(s = s_1)}} = \left(\frac{\partial F}{\partial X}\right)_j k + \left(\frac{\partial F}{\partial Y}\right)_j l + \left(\frac{\partial F}{\partial Z}\right)_j m \tag{5}$$

To prevent computing oscillation, each iterative algorithm has to be treated using specific conditions or the condition for validity. Further details are described in [19]. This process provides the normal vector which is important for ray refraction. The vector usage can be written as

$$N'\mathbf{S}' \times \mathbf{r} = N\mathbf{S} \times \mathbf{r} \tag{6}$$

The parameter $\mathbf{S}$ is the unit vector of the impact ray with the direction cosines (k, l, m). The parameter $\mathbf{S}'$ is the unit vector of the refracted ray with the direction cosines (k', l', m'). The normal vector $\mathbf{r}$ is situated at the impact point. The refractive index N is for the medium in front of the surface while the refractive index N' is for the medium behind the surface. The Equation (6) can be rewritten as:

$$\mathbf{S}' = \mu\mathbf{S} + \Gamma\mathbf{r} \tag{7}$$

where $\mu = N/N'$ and Γ are the undetermined multiplier which must be found. The Newton–Raphson iterative technique is applied again.

This procedure can be also applied to rays with ultrasound characteristics with minor modifications. According to [17] the light ray can be replaced by the ultrasound ray, provided that the medium is homogeneous and has the ability to transmit ultrasound. The eye complies with these conditions. A modified Snell law for ultrasound refraction can be used, but ultrasound behaves differently when refraction occurs. We must derive the index of refraction from the velocity of the ultrasound for each particular medium. We can write the modified Snell equation as [17]:

$$\frac{\sin \alpha_1}{c_1} = \frac{\sin \alpha_2}{c_2} \tag{8}$$

where α_1 is the impact angle, α_2 is the refraction angle, c_1 [m/s] is the ultrasound velocity in the medium in front of the surface and c_2 [m/s] is the medium behind the surface. We modified Eq. (6) and the optical index of the refraction was substituted by the ratio of ultrasound velocities. We can write:

$$\frac{\mathbf{S}}{N} \times \mathbf{r} = \frac{\mathbf{S}'}{N'} \times \mathbf{r} \tag{9}$$

The Equation (9) can be rewritten with undetermined multiplier as Eq. (7), but the parameter μ has to be defined as:

$$\mu = \frac{c_2}{c_1} \tag{10}$$

where the parameter c_x matches the ultrasound velocity in specific medium. These modifications allow us to use the ray tracing method to describe ultrasound in the eye.

The initial conditions for the contact ultrasound biometry set the medium in front of the cornea as the medium with the same velocity as the cornea – 1620 m/s. For the immersion ultrasound biometry, the probe was placed at a distance 8 mm from the cornea and the probe was immersed in physiological solution with an ultrasound velocity of 1523 m/s. The accuracy of the ray tracing procedure was verified by OSLO software. This professional software allows for ray tracing and we were able to check all computing steps using our own ray tracing procedure. Each step was compared with the result of OSLO. This comparison was only possible in the unmodified version for light. No modification was made for ultrasound and the validity of our ray tracing procedure is assumed.

2.3. Statistical description of the eyes sample

Real biometrical data were used for a practical demonstration. The sample included 129 eyes scheduled for surgical removal of cataract and they were measured by one person using both contact and immersion methods. There were 82 women and 47 men (64% of woman and 36% of men). The average age was 73.65 years (SD 9.33,

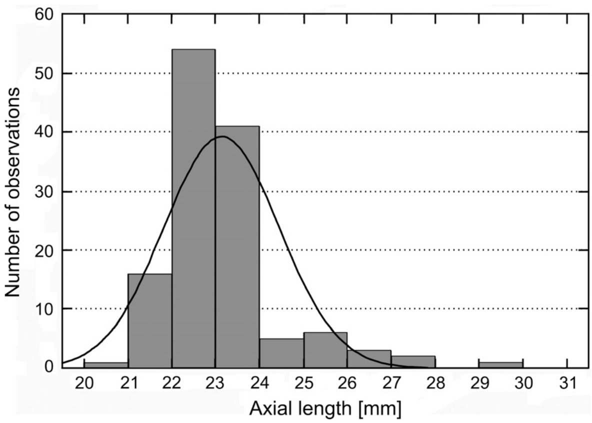

Fig. 1. The histogram of the axial length of the sample n = 129 eyes measured by the contact method.

the youngest proband was 47, the oldest was 95). We measured five values for each parameter – the anterior chamber depth, lens thickness, vitreum chamber depth and axial length of the eye. The average axial length of the sample measured by the contact method was 23.12 mm (SD 1.315, the shortest was 20.92 mm and the longest was 29.32 mm). The histogram of axial lengths is shown in Fig. 1.

3. Simulation of the lateral movement and tilt of the probe

The immersion method is considered to be more accurate according to most studies. The starting position of the probe here was changed intentionally to determine accuracy. The aim was to monitor changes for each position. The lateral movement from the central position was first simulated and then the probe tilts from the initial coincidence position – Fig. 2. The parameters of the eye model were taken from the Gullstrand emetropic eye model. There are several simple eye models, including the Emsley, Le Grand exact eye and the Gullstrand eye. We chose a non-accommodative

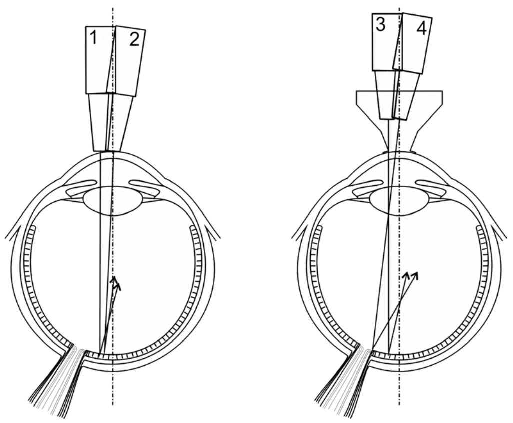

Fig. 2. The probe using the contact method was moved to the lateral position (1) or it was tilted (2). The same simulation was applied when using the immersion method: the lateral movement (3) and the tilt (4).

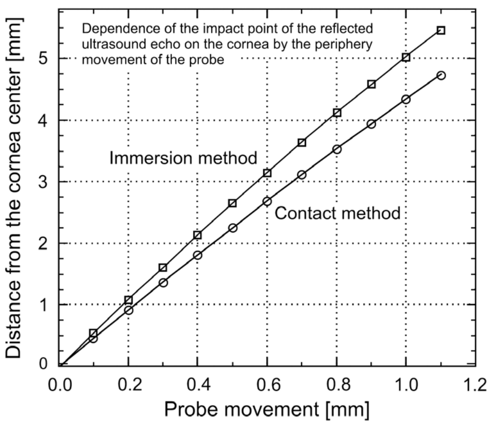

Fig. 3. The results of the ray tracing simulation. The graph shows the distance of the impact point on the cornea as a function of the lateral probe movement.

version of the Gullstrand eye (No. 1) for our purposes. There are more accurate eye models, *e.g.*, LOTMAR [20], DRASDO–FOWLER [21], KOOIJMAN [22], NAVARRO *et al.* [23] or LIOU–BRENNAN [24]. These eye models are very useful for off-axis optical description, but this was not the aim of our study. The well-known applanation effect by the contact method was not taken into account in this simulation.

The results of the first simulation are shown in Fig. 3. The monitored parameter was the impact point on the cornea after the retina reflection. The maximal movement was 1.2 mm. When the probe was decentred further to the lateral position, the reflected ray missed the lens and the iterative algorithm did not find the impact point on the surface.

We can determine how the lateral probe movement influences the ability of the instrument to catch the reflected echo. The wave generated by the ultrasound transducer is propagated with a slightly divergent angle. It is important to consider that the size of the transducer is about 4 mm. The transducer dimensions influence the ability of the device to catch the reflected echo. A bigger transducer permits catching more deflected echo but at the expense of accuracy. The reflected echo includes information from the whole area, not only from one point. For this reason, the data cannot be considered exact. In this study, the ultrasound beam was substituted by the ray. Due to this approximation, exact numerical results were not possible. However, for the purpose of comparing the two methods, this attitude could be used.

The immersion method tended to register the error caused by the probe movement. It was important to consider that the impact point further from the central position by more than 2 mm does not allow us to catch the echo. The method which reaches this limit sooner is more precise owing to the invalidity of the distorted information from the periphery.

Relatively more noticeable differences can be seen from observing the probe tilt – Fig. 4.

The difference between the contact and immersion methods is significantly larger in this case. The greater sensitivity of the immersion method was more visible in the case of the probe tilt: if we tilted the probe using the immersion method, we got

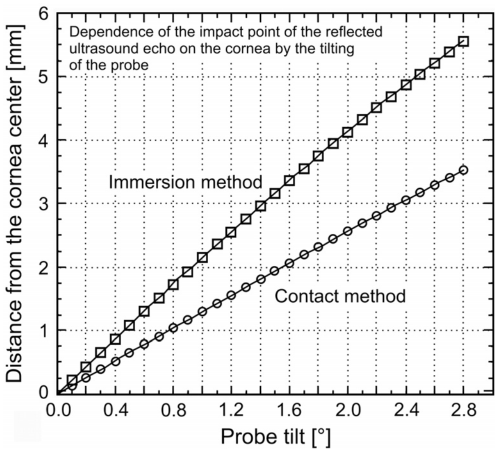

Fig. 4. The results of the ray tracing simulation. The graph shows the distance of the impact point on the cornea as a function of the probe tilt.

a larger deviation of echo than when using the contact method. This allowed us to proceed under the assumption of the greater accuracy of the immersion method. With this method, the transducer catches only echoes from the central area (closer to the optical axis). Therefore these echoes contain precise information. The contact method allows processing from the peripheral area with its associated information distortion

This situation can be compared to the *keyhole effect*. The further from the cornea the probe is (immersion method), the more strictly the condition of the central and coincidence position of the probe has to be kept. Approaching the probe to the cornea allows larger misalignment from the central position and more distorted information – Fig. 5.

To test this assumption, a control sample of 129 eyes was monitored. The main monitored parameter was the standard deviation which shows the bias of each method to provide inaccurate results. Each eye was measured by the contact and by the immersion methods. A standard deviation parameter was recorded as well. The measurements were carried out by one person. It is necessary to point out the correction in the measuring process. If the standard deviation exceeded 0.10, a new

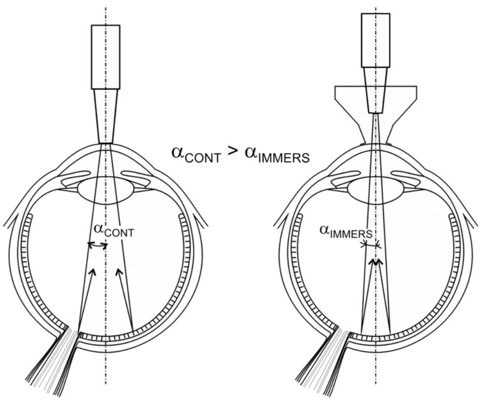

Fig. 5. The contact method (left) catches more deviated echoes due to the closer position to the cornea. The immersion method (right) has a farther probe position and the better condition for accuracy.

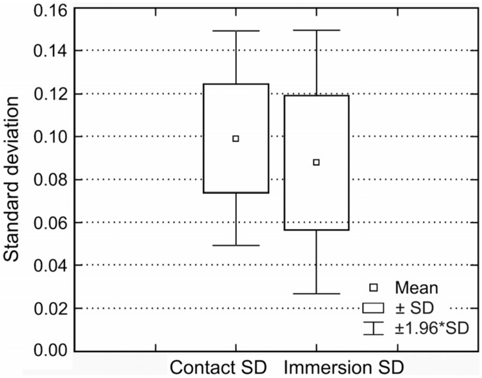

Fig. 6. The box plot diagram of the standard deviation of the specific methods. The immersion method has a smaller standard deviation than the contact method. The immersion method is therefore more accurate and more reliable. The variance is in the lower values. This fact confirms the accuracy of the immersion method.

measurement was carried out to achieve a smaller standard deviation. The same procedure was applied to both methods and hence the resulting deviation had the same effect. Some measurements did not allow us greater accuracy. Therefore, there are some values over the 0.10 limit. If the standard deviation was bigger, the associated method was considered less accurate. The following Fig. 6 shows the box plot diagram of the standard deviation of both methods.

The specific details of the standard deviations for each method are shown in Tab. 2. The statistic of two selected F-tests for scatter proved no significant differences between the methods.

We can also compare the methods. The basic data for each method are shown in Tab. 3, which permits to compare our results with other studies. JIVRAJKA *et al.* [25] measured the mean axial length 23.46 mm (SD 1.03) in the population of 750 eyes. Figure 7 shows the box plot diagrams of the axial length.

The difference in the means is 0.145 mm, which is in the agreement with other studies [8] and [9]. We used the *t*-test for dependent samples. No significant differences

T a b l e 2. Additional data on standard deviations of axial lengths of the eyes measured by the contact and immersion methods.

	Mean [mm]	SD
SD of the axial length measured by the contact method	0.0987	0.0236
SD of the axial length measured by the immersion method	0.0904	0.0331

T a b l e 3. The basic data of axial lengths measured by the contact and immersion methods.

	Mean [mm]	SD	Pearson coefficient
Axial length (contact method)	23.12	1.315	0.976
Axial length (immersion method)	23.266	1.298	0.976

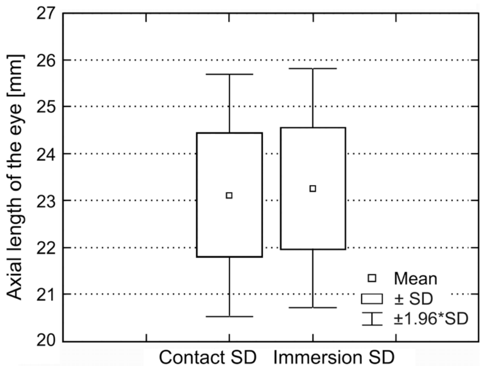

Fig. 7. The box plot diagram of axial lengths of the contact and the immersion methods. The immersion method slightly tends to measure the axial length of the eye as longer.

were found between the two methods. However, a difference of 0.145 in addition to systematic error can cause conditions leading to a significant refraction error.

4. Changes in the biometrical data

We used three models of the eye treated by the ray tracing method: hypermetropic (HY), emetropic (EM) and myopic (MY). To use the ray tracing method, all eye parameters have to be known. We designed a model of the eye based on keratometry data and ultrasound data (the anterior chamber depth, lens thickness and the axial length of the eye) [26]. This eye model allows us to reconstruct other important eye data, especially radius of curvature. The final axial refraction of this eye model was compared with the axial refraction from autorefractokeratometer. The accuracy of this eye model was confirmed statistically. Real biometry data, which are the least different from the real axial refraction, were used for this eye model. The details are presented in Tab. 4. The eye model difference means the deviation of the specific eye model from the real axial refraction recalculated on spherical equivalent.

On the basis of these data, we reconstructed the radius of curvatures and analysed the data. The negative radius of curvature of the retina was set at –10 mm in all cases. We tilted and moved the probe in the same way to compare the accuracy of two methods. The ideal measurement is performed in the central position coincident with the visual axis. The emitted ultrasound ray is propagated to the retina; it is partly

Table 4. The data of real eyes treated by the ray tracing method.

	Hyperopic eye	Emetropic eye	Myopic eye
Dioptric power of the eye [D]	43.375	41.375	44.435
Anterior chamber depth [mm]	2.552	2.72	3.026
Lens thickness [mm]	4.92	4.553	4.593
Axial length of the eye [mm]	21.768	23.283	25.583
Calculated axial refraction [D]	+3.442	–0.089	–7.14
Eye model difference [D]	0.075	–0.4	+0.15

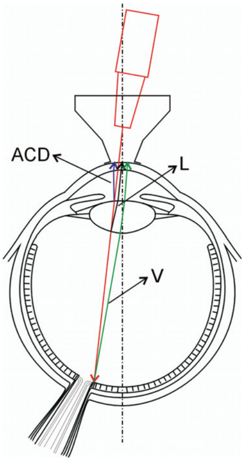

Fig. 8. The schematic figures of the ray tracing of the ultrasound ray when the probe is tilted. The figure shows the immersion method.

reflected and returns to the cornea – Fig. 8. On the basis of the technical data for specific settings and time, we can determine the distance of the surface to the probe. If we tilt or decentre the probe from a central position, then the ray travels for a longer time. It is also true for the reflected ray.

The data from the tilted probe do not match the data from the central position. The degree of difference is shown in Tab. 5. The largest possible differences from the central nominal values were observed – data from Tab. 4. The maximal probe decentration depended on the eye model used.

The negative values refer to reduction in parameter during the increase in the decentration of the probe; analogically, if the value is positive, the parameter grows with the probe decentration. The following conclusions can be deduced:

– There is no great difference in the parameters of the anterior chamber depth and lens thickness between the contact and immersion methods. All errors are

T a b l e 5. The data here show the differences in millimetres when we moved the probe from the central position. The degree of decentration is limited by the eye model. The maximal peripheral movement was from 0.9 mm (myopic) to 1.2 mm (emetropic) and the maximal tilt was from 2.1° (myopic) to 3° (hypermetropic). All distances are in millimetres.

		Hypermetropic (HY)		Emetropic (EM)		Myopic (MY)	
		Contact	Immersion	Contact	Immersion	Contact	Immersion
Movement	Anterior chamber depth	0.09834	0.10826	0.07606	0.08356	0.04582	0.05105
	Lens thickness	−0.08619	−0.10129	−0.06867	−0.07972	−0.03945	−0.04737
	Axial length eye	**−0.3517**	**−0.4686**	**−0.2896**	**−0.3812**	**−0.3787**	**−0.4412**
Tilt	Anterior chamber depth	0.00637	0.04367	0.00552	0.0375	0.00339	0.02046
	Lens thickness	−0.01043	−0.05132	−0.0085	−0.4373	−0.00537	−0.02402
	Axial length eye	**−0.1318**	**−0.4169**	**−0.1449**	**−0.3233**	**−0.198**	**−0.426**

proportionally different (HY anterior chamber depth: 0.00637 against 0.04367). These errors have no major impact.

– The above average lens thickness (HY eye model: L = 4.922 mm) led to changes in decimal order. The lens with an average thickness 4.42 mm (the average value measured by the contact method, n = 129 eyes) led to changes of about one twentieth of a millimetre. If we move the lens by 0.1 mm further into the eye then the final axial refraction is changed by 0.12 D. This fact was discovered by the light ray tracing using the Gullstrand model of the eye in paraxial space.

– The probe movement caused a change in the axial length of the eye, relative to the vitreum length. A change of 0.3 mm could lead to a difference in dioptric power of 1 D.

– The biggest difference in the axial length of the eye between the methods was caused by the probe tilt. From this point of view, the contact method was more accurate. The error caused by the lateral movement was similar for both methods.

The reason for the accuracy of the immersion method is described in Section 3. The error caused by the probe tilt was expected. It was interesting to discover that the contact method in this case was three or four times more accurate than the immersion method. During the visualization of this case, the rays approach the optical axis. This case is not resistant to errors which increase with the axial length of the eye. We must mention the so far neglected influence of the applanation of the cornea in the contact method. The conclusion can be drawn that considering numeric ray tracing calculation there is no difference in the anterior chamber depth distance between the two methods. The inaccuracy of the contact method is caused by the physical contact of the probe – Fig. 9. The immersion method is more accurate, primarily due to the absence of the applanation and secondarily due to the probe position further from the eye.

The most important problem in the intraocular lens (IOL) calculation is in inaccurate determination of the axial length of the eye. The longer axial length of the eye causes the bigger measuring error of vitreous depth [27]. The vitreous depth directly correlates with the axial length of the eye. The Pearson correlation coefficient for vitreum depth and axial length of the eye is similar for both methods – r = 0.89 (high correlation). The calculation formula for the IOL dioptric power is valid when the axial length is the most important parameter. If we take the longer axial length

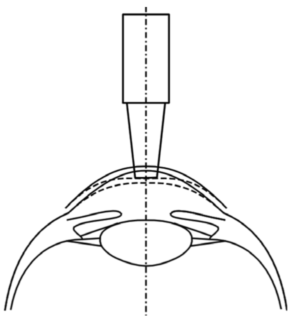

Fig. 9. The schematic figure of the applanation of the cornea in the contact method.

(over 25 mm) into account, we can consider the resulting refraction error is caused mainly by an error in the axial length measurement. In case of the short axial length we can talk about a systematic error of the calculation formula.

The results following from the numeric simulation allow us to find a correlation between the axial length of the eye and standard deviation. We cannot proceed with this analysis due to the required data selection. During the biometry, it is common to correct measurements if the standard deviation is over 0.10. Measurements exceeding this value are discarded and the measurements are repeated. This approach has a negative influence on the independent selection of a sample and the statistical result is invalid. This correction was made on the axial length of the eye.

However, we could still analyze the influence of the applanation on the cornea. The control sample included $n = 129$ eyes. Each eye was measured by the contact and immersion methods and we recorded 5 values of the anterior chamber depth for each measurement. More details are shown in Tab. 6 together with the box plot diagram in Fig. 10.

As expected, the average anterior chamber depth measured by the contact method was shallower. This error was not evident when the eye was measured by the ray tracing method. We can conclude that this inaccuracy is caused mainly by the practical approach and by the systematic error of the method. The standard deviation of the contact method has larger scatter which can be explained by the variable pressure of the probe on the eye. The negative influence of this pressure is in conformity with the results of ELEFTHERIADIS [1]. However, we cannot exclude the influence of the pressure found in the immersion method by NEMETH *et al.* [28].

T a b l e 6. Details of anterior chamber depth measured by the contact and immersion methods.

	Contact method	Immersion method
Mean	2.945721	3.127783
Standard deviation	0.103236	0.078716
Minimal value	1.678	1.7
Maximal value	3.886	3.91

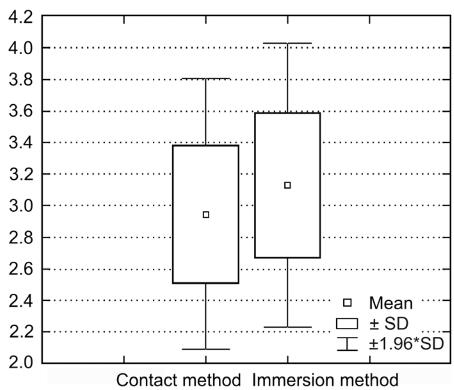

Fig. 10. The box plot diagram of the anterior chamber depth in millimetres. The data was measured by the contact and the immersion method.

Ultrasound biometry is based on the propagation and reflection of the ultrasound wave from specific eye structures. Partial coherency interferometry measures the axial length with the propagation and reflection of the infrared light beam (λ = 780 nm) [1]. Propagation of the ultrasound wave in specific settings is described with the velocity of the sound in the settings while the propagation of the light beam is described by the optical index of refraction. In the case of a mature cataract we observed the lamella structure and this effected the propagation in both cases. The ultrasound wave is less influenced by the cataract than the optical method – the difference of the acoustic velocity is lower than the difference of the index of the refraction. High density cataract prevents the propagation of the light and influences the measured parameters. According to Hill, we cannot use the optical method in 8% to 10% cases [4]. However, the optical method is very accurate and in special eye situations it provides very good results (pakofobie, silicon oil, *etc*., [29]). In special cases we still have to rely on the ultrasound immersion method.

5. Conclusions

Simulation using the ray tracing method for ultrasound confirms the greater sensitivity of the immersion technique based on movement or tilt of the probe. This has positive impact on the accuracy of measurement since if the deviation exceeds a certain threshold, it is impossible to capture the reflected ray. The limit determining the ability to catch the echo is higher for the contact method, which is the reason for its inaccuracy. The probe distance from the cornea is the main determining parameter. This situation can be compared to the keyhole effect. This result was proved by statistical testing of real sample data. The contact method has a larger scatter of the measured parameters.

We also showed that the probe tilt changes all parameters. The anterior chamber depth and the lens thickness are influenced by this decentration. The possible degree of resulting error is −0.12 D of the axial refraction. The biggest differences can be found in the axial length parameter. The resulting inaccuracy depends on the axial length of the eye. The tendency to error is smaller for the eye with the short axial length. The longer the axial length is, the larger the tendency is. A difference up to 0.3 mm can be predicted for eyes with an axial length over 25 mm.

The data provided possibilities for analyzing the anterior chamber depth. The ray tracing simulation did not reveal any significant methodical differences between the immersion and contact methods. However, the practical data confirmed the shallower anterior chamber depth by the contact method. We add this inaccuracy to the applanation effect of the cornea. The accuracy of the immersion method is primarily caused by the absence of direct contact of the probe with the cornea and secondarily by the farther position of the probe.

The results favour the characteristics of the ultrasound beam. The density of the cataract influences the ultrasound beam less than the light beam as the changes in light refraction are greater. This fact gives advantage to the ultrasound method despite the fact that the optical method is more accurate.

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