Simulation of Dynamic Contour Tonometry compared to in-vitro study revealing minimal influence of corneal radius and astigmatism

The Theoretical Foundations of Dynamic Contour Tonometry

ARVO 2003; Poster#350

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SUMMARY

Purpose

With tonometers currently in use, intra-ocular pressure (IOP) can be determined only indirectly from the force required to perform a specifed deformation of the cornea. An improved theoretical model as well as practical experiments for a direct trans-corneal measurement of IOP shall be presented.

Methods

A theoretical model for studying the forces acting on a cornea when brought into contact with a surface of a given shape is presented. The model allows simulation of measurements using differently contoured contact surfaces and ranges of corneal properties (namely, corneal thickness (CCT) and corneal radius (CR)).

A Dynamic Contour Tonometer (DCT) was constructed based on this theoretical principle and validated on eye bank bulbi against true manometric IOP measured by cannulation.

Results

A contact surface which matches the contour of the cornea is shown to have the desired property of creating an equilibrium between capillary force, rigidity force, appositional force, and the force exercised on the cornea by IOP. A pressure sensor integrated into the contoured surface will therefore measure IOP with no systematic errors being introduced by said forces or by changes in corneal properties.A DCT with an optimized contour tip furnishes IOP results that closely match manometric pressure in eye bank eyes. Results are not affected in a systematic way by variations in CCT, CR, or Astigmatism (AS).

Discussion

DCT permits to eliminate most of the systematic errors arising from physiological variables of the eye, which render all force tonometers inaccurate. The potential of measuring true IOP, combined with the additional capability of measuring dynamic fluctuations in IOP, opens up new opportunities for tonometry as a tool for the differential diagnosis and classification of POAG, OHT, and NTG and for studying vascular effects.

1. PURPOSE

Today, all established methods for estimating intraocular pressure (IOP) use indirect methods. The devices according to Goldmann (GAT), Schiøtz, Perkins, Mackay-Marg, and Draeger, as well as all known non-contact ("air-puff") tonometers might be termed "force tonometers".

To avoid the inherent errors and uncertainties associated with indirect force methods, experiments have been undertaken by several authors [1-4] to use pressure sensors incorporated in flat tonometer tips similar to the Goldmann tip, or in a hand-held diagnostic contact lens [6-7]. For reasons to be detailed later, these types of device tend to furnish IOP readings which are significantly higher than true IOP or the generally accepted "gold-standard IOP" obtained with the Goldmann Tonometer.

In an attempt to find a non-invasive and direct method for continuous IOP measurement, which would be less influenced by inter-individual variations of mechanical properties of the eye (notably the cornea), and therefore be less prone to systematic error, we are presenting an improved theoretical model for direct trans-corneal pressure measurement and have built experimental devices according to this theory. It is based on a pressure-sensing device that closely matches the corneal contour and thus induces the least possible amount of geometrical deformation.

2. METHODS

2.1. Principle of Contour Tonometry

As described in our earlier publications [5], the cornea may be described as a spherical shell constructed from a material that resists stretching but is fairly flexible to bending deformations. The cornea maintains its shape almost irrespectively of intra-ocular pressure [3]. Nevertheless, rigidity forces (i.e. due to bending and buckling) may not been neglected.

The Contour Tonometer features a cylindrical tip with a concave contact surface. Its function may be understood in terms of the following principles:

- The radius $\mathbf{R_t}$ of the tip surface shall exactly match the radius $\mathbf{R_C'}$ assumed by the corresponding corneal segment when pressure on both sides of this part of the cornea is the same.
- Within contact area $\mathbf{A_C}$, the contours of the tip and of the cornea ideally match. This condition shall be termed "**contour match**".
- If a pressure sensor with an active diameter smaller than the contour diameter **d** is integrated flush into the contour surface, it will measure precise IOP.
- The contour diameter d shall be larger than the diameter of the pressure sensor, and the outer diameter of the tear film annulus shall be smaller than the tip diameter.

To arrive at a quantitative description of Contour Tonometry, we shall now discuss and calculate all forces occurring during a contour tonometer measurement on a human cornea.

2.2. Mechanical and Geometrical Calculation

The following assumptions and/or simplifications shall be used:

- The contour of cornea is convex, and the contour of the tip is concave.
- The contours of cornea and tip are spherical.
- The anterior radius Rc of the cornea is always smaller than the curvature radius Rt of the tip.
- The line defined by the centers of curvature of the cornea and the tonometer tip contact surface is defined as the central axis.
- For reasons of symmetry, only force components parallel to the central axis are considered, and the three-dimensional corneal calotte area is replaced by its planar cross-section.

- The sign of forces attracting tip and cornea is defined as negative, the sign of forces pushing tip and cornea away from each other is defined as positive.
- Consistently with Goldmann [8, 9] and own measurements (not shown here), we assume furthermore that during a Goldmann applanation tonometer measurement on a typical human cornea of standard dimensions (i.e., central corneal thickness hc = 537 μm, anterior corneal radius Rc = 7.7 mm), rigidity and capillary forces will be at equilibrium.
- The outer diameter of the tear film annulus is smaller than the diameter of the tip.

(a) Force equilibrium

In agreement with earlier publications in this field, we assume, that once contour matching has taken place, the four major effects acting on the contact surface and on the cornea are at an equilibrium.

The force equilibrium can be described by:

$$\boxed{F_{iop} + F_c + F_r + F_{ap} = 0} \quad (1)$$

where:

F_{iop} is the force exercised by effective IOP, acting on the tonometer's contact surface,
F_c is the capillary force or adhesion force created within the tear film, caused by negative capillary pressure within the tear film,
F_r is the rigidity force responding to deformation of the cornea,
F_{ap} is the appositional force applied externally to the tonometer (e.g. in case of the Goldmann Applanation Tonometer set by the adjustment of the thumb wheel).

The two forces with negative sign, **F_c** and **F_{ap}**, attract tip and cornea, whereas the forces with positive sign **F_{iop}** and **F_r** push tip and cornea away from each other.

Fig. 1 illustrates the geometric setup for the following calculations. The contours of tip and cornea are matched within contact area **A_C**. Due to the difference between radii **R_C** and **R_t**, outside the contact area, the cornea and the tonometer form a gap of which is partially (annulus of width **w_t**) filled with tear fluid.

(b) IOP Force

The force **F_{iop}** represents the reaction between cornea and tip due to the effective intra-ocular pressure **P** acting on the contact area **A_C**.

$$F_{ap} = P \cdot \pi \cdot \frac{d^2}{4} \qquad (2)$$

(c) Capillary Force

The force **F_c** is a capillary force due to adhesion created within the tear film interface between the contact surface of the tonometer and the cornea. **F_c** is defined by equation (3):

$$F_C = F_p + F_S \qquad (3)$$

The first component **F_p** describes a force directly dependent on the capillary pressure **P_c** existing within the tear film volume effective on the area **A_p** of the tear film annulus. Using the Laplace equation for **P_c** and geometrical relationships for **A_p**, the capillary pressure component may be written as

$$F_p = \frac{-\gamma \cdot \pi \cdot R_c^{\,2}}{2 \cdot s} \cdot \cos(\Theta + \frac{\beta - \alpha}{2}) \cdot \left(\sin^2(\beta) - \sin^2(\beta_c)\right) \qquad (9)$$

where Θ is the mean wetting angle at the tear film – air interface.

The second component of the capillary force $\mathbf{F_c}$ is the force $\mathbf{F_s}$ which is related to the surface tensions of the interfaces between tear film, air, cornea, and tonometer. From geometrical relationships and by neglecting minor components, $\mathbf{F_s}$ is obtained as

$$F_s = -\gamma \cdot 2 \cdot \pi \cdot \left(\frac{d}{2} + w_t\right) \cdot \sin(\beta + \Theta) \quad (10)$$

Therefore the capillary force $\mathbf{F_c}$ can be summarized from (9) and (10) to:

$$\boxed{F_c = -\gamma \cdot \pi \cdot \left[\frac{R_c^{\,2}}{2 \cdot s} \cdot \cos(\Theta + \frac{\beta - \alpha}{2}) \cdot \left(\sin^2(\beta) - \sin^2(\beta_c)\right) + \left(d + 2 \cdot w_t\right) \cdot \sin(\beta + \Theta)\right]} \qquad (11)$$

(d) Rigidity Force

The rigidity force $\mathbf{F_r}$ originates as a reaction to distorting the cornea from its initial shape by matching it to the contour shape defined by the tonometer tip. The size of the force is defined by the amount of bending and buckling, expressed as the difference of the asymptotic angles to the surfaces of cornea and tip respectively at point $\mathbf{B_c}$, and the bending rigidity of the cornea, integrated over the circumference of contact area $\mathbf{A_C}$. We assume that the bending rigidity of the cornea is proportional to the third power of its thickness. For a typical cornea of standard rigidity, the bending force can be described as follows:

:

$$\boxed{F_r = \frac{-5.2 \cdot 10^{-3} N}{4.2 \cdot mm^4} \cdot 2 \cdot \pi \cdot R_c \cdot \sin(\beta_c) \cdot (\beta_c - \alpha_c) \cdot h_c^{\,3}} \quad (14)$$

(e) Appositional Force

Solving equation (1) for $\mathbf{F_{ap}}$ and integrating (14), (11), and (2) into equation (1), we obtain equation (15) :

$$\boxed{\begin{aligned} F_{ap} = \gamma \cdot \pi \cdot &\left[\frac{R_c^{\,2}}{2 \cdot s} \cdot \cos(\Theta + \frac{\beta - \alpha}{2}) \cdot \left(\sin^2(\beta) - \sin^2(\beta_c)\right) + \left(d + 2 \cdot w_t\right) \cdot \sin(\beta + \Theta)\right] + \\ &\frac{-5.2 \cdot 10^{-3} N}{4.2 \cdot mm^4} \cdot 2 \cdot \pi \cdot R_c \cdot \sin(\beta_c) \cdot (\beta_c - \alpha_c) \cdot h_c^{\,3} - P \cdot \pi \cdot \frac{d^2}{4} \qquad (15) \end{aligned}}$$

If we introduce the radius of the tip $\mathbf{R_t}$, and the tear film volume $\mathbf{v_t}$, all unknown values, i.e. α, β, α_c, β_c, and **s** may be calculated from geometrical considerations in a straightforward manner. Equation (15) may then be rewritten and solved for **d**, using iterative methods (not detailed here), furnishing

$$d = F(F_{ap}, P, R_c, R_t, h_c, \Theta, \gamma, v_t) \qquad (16)$$

(f) Model Calculations

The results show that the appositional force $\mathbf{F_{ap}}$ in equation 15' and the diameter of contour match **d** are entirely independent of tear film volume $\mathbf{v_t}$.

To examine the expected behavior of the Dynamic Contour Tonometer we chose tip curvature radius $\mathbf{R_t}$=10.5mm and adjusted the appositional force to $\mathbf{F_{ap}}$=1.0 gram. Θ and γ are material constants. Equation (16) thus reduces to

$$d = F(R_C, h_C, P) \quad (16').$$

Figs. 2 – 5 show the calculation of **d** for a physiologically relevant range of values for the variables $\mathbf{R_C}$ (corneal radius), $\mathbf{h_C}$ (corneal thickness) and **P** (IOP). The diameter **d** has no influence on the readout of the pressure sensor, as long as it is larger than the active diameter of the sensor.

2.3. *Comparative Study on Human Cadaver Eyes*

To compare tonometer values with true IOP, a comparative study on human cadaver eyes was performed.

For this *in-vitro* study, sixteen freshly enucleated human eyes were de-epithelialyzed. An intubation needle was placed in the anterior chamber. The corneas were dehydrated with Dextrane 20% from outside by continuous dropping and from inside through the tube until stable CCT was achieved. CCT was monitored using ultrasound pachymetry. Corneal Radius (CR) and Astigmatism (AS) were measured using a keratometer. The tube in the anterior chamber was connected to a pressure transducer and to a bottle system filled with balanced salt solution. The pressure in the eye was then artificially altered between 5 and 58 mmHg in 15 consecutive steps by changing the height of the bottle in relation to the eye. The mean of each 5 consecutive measurements taken at each pressure setting with DCT was compared with the direct manometric readings from the pressure transducer.

Materials and equipment used:

Dextrane 20%: 229’000MW Dextrane, (Sigma, St.Louis MO 62124).

Balanced salt solution BSS (Alcon Surgical, Ft.Worth, TX 76102).

Intubation needle: 22 ga. needle with Y adapter, model Saf-T-Intima, Vialon,
(BD Medical Systems, Franklin Lakes, NJ 07417).

Pressure transducer: Neonate Kit W/30ML, (Abbott Critical Care Systems, Morgan Hill, CA95037).

Manometer: HP Monitor/Terminal Model 78534C (Hewlett-Packard, Palo Alto, CA 94304).

Dynamic Contour Tonometer: experimental device based on a "SmartLens" Dynamic Observing Tonometer
modified by the authors (ODC Ophthalmic Development Company AG, CH-8005 Zurich).

Dynamic Contour Tonometer Tip: contour radius 10.5mm, tip diameter 7mm, appositional force 1 grams

Slit lamp: Model 30 SL M (Zeiss Meditec AG, D-07745 Jena),
equipped with Goldmann Tonometer (Haag-Streit, CH-3098 Köniz).

Keratometer: Orbscan II corneal topography system (Bausch & Lomb, Salt Lake City, UT 84116).

Pachymeter: Ultrasonic Pachymeter model 850, (Humphrey Instruments Inc., Dublin, CA 94568).

3. RESULTS

3.1. *Comparative Study on Human Cadaver Eyes*

De-epithelialyzed human cadaver eyes were dehydrated to a mean CCT of 450 ± 39 µm (range: 381 to 517 µm). Corneal radius (CR) averaged 8.0 ± 0.45 mm (range: 7.42 to 9mm): Mean Astigmatism (AS) 1.34 ± 0.55mm (range: 0.3 to 2.2mm).

(a) DCT measurements vs. manometric IOP; Correlation with corneal properties

DCT readings were strongly correlated to manometrically obtained pressure with a linear regression of (y=1.00x+0.34, R^2=1, P<0.001). Averaged (between 5 consecutive readings for each eye at each pressure step) DCT measurements on human cadaver eyes exhibited a constant bias of +0.38 ± 0.085 mmHg (mean ± 95%CI) relative to the corresponding manometric readings (*Fig. 6*). The 95% limit of agreement for these averaged measurements was ±0.98 mmHg.

No significant correlations were found between CCT and DCT readings (R^2= 0.003; P=0.54), between CR and DCT (R^2= 0.0011; P=0.76), or between AS and DCT readings (R^2=0.014; P=0.28).

3.2. *Quality of IOP measurements by DCT*

To determine the **variability in repeated measurements**, five consecutive readings at identical conditions each were taken on 16 eyes at 8 different manometric pressures and plotted (*Fig. 7*) against their averages. The 95% limit of agreement for repeat measurements was ±1.14 mmHg.

To study the quality of pressure measurement theoretically possible with a contour tonometer, we determined theoretical pressure measurement errors **ΔP*** as the difference of the tonometer reading p* predicted by theory and the true intraocular pressure P. We computed the contact diameter d for each of the 1065 individual measurements taken at different pressures on our set of cadaver eyes, then plotted observed individual measurement deviations (ΔP = p – P) against d. A second-order least-squares fit of the data points thus obtained furnished a polynomial describing the the theoretical measurement error as a function of contour match diameter:

$$\Delta\mathbf{P}^* = \left(\boldsymbol{m} \cdot \mathbf{d}^2\right) + \left(\boldsymbol{n} \cdot \mathbf{d}\right) + \boldsymbol{q} \qquad (17)$$

with m = 0.0632 (p_m < 0.0001), n = -0.6399 (p_n = 0.0002), and q = 1.961 (p_q = 0.0006). The 95% limit of agreement of the experimental points with the polynomial is ± 1.3 mmHg.

With these coefficients, **ΔP*** was calculated for an array of IOP values, corneal radii, and corneal thicknesses. The result is displayed in *Fig. 8*. Computed **ΔP*** is shown at different levels of true IOP ranging from 5 to 30 mmHg. Areas where **ΔP*** is less than 0.5 mmHg are colored green; the intermediate range 0.5 > **ΔP*** >0.8 mmHg is colored yellow; and ranges for which **ΔP*** exceeds 0.8 mmHg are shown in red. Values for **ΔP*** exceeding 1 mmHg were cut off in the graph for **P** = 5 mmHg.

4. DISCUSSION & CONCLUSIONS

4.1. Theoretical Considerations

We have enhanced the theory of Goldmann [8, 9] with respect to the influence of capillary and rigidity forces. Since these forces do not all act on the same area of the cornea, Goldman’s original assumption of a pressure equilibrium must be modified, and a force equilibrium must instead be considered. This allows us to account for the fact that capillary forces are effective only within the area of the tear film annulus and not across the entire area of contact. The equations derived may be solved explicitly for various types of tonometer tips, including the standard, planar Goldmann tip as well as concave tips that match the corneal contour.

The contact surface of the "contoured" tip comes as close as possible to the shape which the cornea assumes if pressure within and outside the eyeball were identical and if no forces were thus acting perpendicularly on the cornea. If this condition of “contour match” is met, a pressure sensing element, integrated into the tip surface, directly measures true IOP. Strictly speaking, each individual cornea would require a custom-made contour-matched tip to fulfill this contour-matching condition and hence furnish a pressure reading corresponding exactly to true IOP. However, our calculations have shown that a "standard contour" may be found which permits faithful measurement of true IOP for a fairly wide range of corneal dimensions. In fact, the condition for a precision measurement is fulfilled as long as the contact area is larger than the pressure sensing element and the outer diameter of the tear film annulus is smaller than the diameter of the tip itself.

In contour tonometry, it is not necessary to know the diameter of contact, or the appositional force. This makes the method much less susceptible to operator error or bias.

The contour match situation establishes itself due to the forces generated by capillary pressure. The pressure measured is therefore independent of the appositional force applied over a range of values from zero to 5 grams.

The **systematic pressure measurement error ΔP*** is always positive; i.e. measured values are equal to or slightly larger than **P** (true IOP). It varies as a complex function of true IOP and of corneal characteristics (radius CR and thickness CCT) . However, **ΔP*** is less than 1 mmHg over the entire range of CR and CCT values shown, (with the exception of CR>9 at **P** = 5). Hence the DCT may be expected to be adequately accurate for practical clinical purposes in the IOP range critical for Glaucoma diagnosis and management (19 – 30 mmHg), for corneal radii ranging from 6 to 9.5 mm and for corneal thicknesses from 300 to 800 microns.

With increasing IOP, the point of optimum performance moves from steeper to flatter corneae. Dependence on corneal thickness is minimal and all but disappears at IOPs exceeding 30 mmHg. It is noteworthy that the case of thin, flattened corneas common after refractive surgery, is handled very well by the DCT for eyes with IOPs in the critical 15 to 25 mmHg range.

4.2. Performance of our experimental Contour Tonometer

The results obtained with our DCT device in our *in-vitro* study corroborate the theoretical considerations. Over the entire range of physiologically relevant pressures and geometrical variations, we found good agreement between DCT and manometer values, with little variance and marginal bias. We did not find any correlation with terms relating to corneal geometry (CCT, CR, or AS), and the relationship between manometric IOP and the DCT measurement was perfectly linear. Repeatability within 5 consecutive measurements was significantly better, and the limit of agreement of measurement on different bulbi at the same pressure was significantly narrower than what is typically found with force tonometers.

4.3. *Clinical Outlook*

Elimination of the major physiological variables of the eye which influence tonometry and which render all force tonometers inaccurate, enables non-invasive contour tonometry to furnish IOP values close to manometric pressure, and to reach a precision which was hitherto accessible only by invasive measurement. We may conclude that only variables modulating the true IOP can influence DCT measurement. The most important one is the cardiac cycle which causes a volume displacement of the choroidal bed and hence a pulsatile pressure change giving rise to the well known Ocular Pulse Amplitude (OPA). Precise determination of the OPA is possible with DCT, since it records 100 samples per second and is capable of detecting static pressure as well as all dynamic pressure fluctuations on a time scale from approximately 0.1 to 10 seconds.

In fact, the discussion about a new “true IOP” standard for tonometry will have to include the question of a more stringent definition of what should be quoted as “the IOP”: systolic IOP, diastolic IOP, mean IOP, or median IOP? The difference can be between 1 to 10 mmHg depending on the patient, and it can be unambiguously measured with the DCT. For a proper distinction of clinically relevant (pathological) IOP levels from “normal” levels, particularly for a proper (and hopefully better than hitherto possible) classification of different forms of glaucoma, consensus on this issue must be reached.

Further planned studies are planned to validate DCT measurements against manometric perssure in living eyes and to confirm the promising results already obtained in exploratory studies with healthy volunteers and patients.

5. LITERATURE REFERENCES

[1] E. S. Perkins, J. Edwards, R. C.Saxena: A new recording tonometer; Trans. Ophthal. Soc. U.K.; (1977), 97, 679.

[2] E. S. Perkins: The ocular pulse; Current Eye Res.; (1981), 1, 19-23.

[3] E. S. Perkins: The ocular pulse and intraocular pressure as a test for carotid artery stenosis; British Journal of Ophthalmology, (1985), 69, 676-680.

[4] E. Linner, E. Rumberger, J. Draeger: Die Möglichkeit der flächenunabhängigen Tonometrie. Klin. Mbl. Augenheilk; (1987),190, 30-33.

[5] H. Kanngiesser, Y. Robert: Dynamic Contour Tonometry. ARVO 2002, Poster.

[6] Entenmann B, Robert Y, Pirani P, Kanngiesser H, Dekker P. Contact Lens Tonometry – Application in humans. Invest Ophthalmol Vis Sci. 1997;38:2447-2451.

[7] P.W. Dekker, Y.C.A. Robert, H.Kanngiesser, P. Pirani, B. Entenmann. Principles of Contact Lens Tonometry. International Ophthalmology 22: 105-111, 1999.

[8] H. Goldmann, Th Schmidt: Über Applanationstonometrie; Ophthamologica Vol. 134 (4), (1957), 221-242.

[9] H. Goldmann, Th. Schmidt: Weiterer Beitrag zur Applanationstonometrie; Ophthalmologica 141; (1961), 441-456.

6. FIGURES

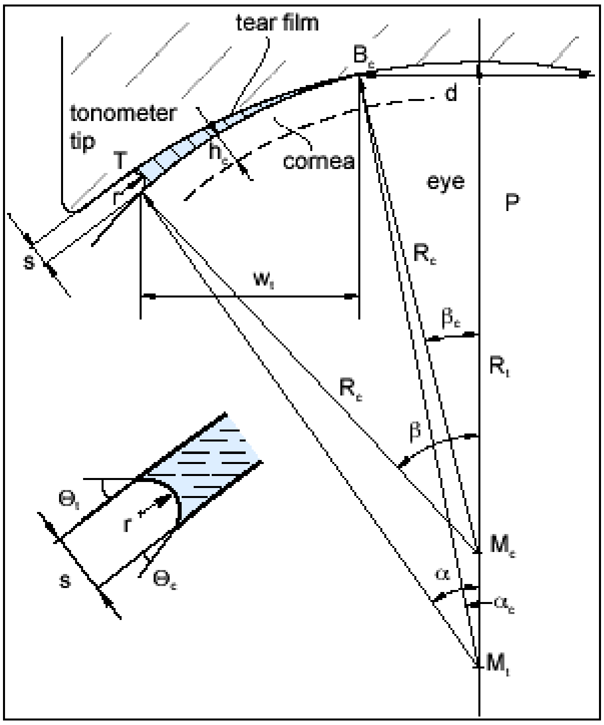
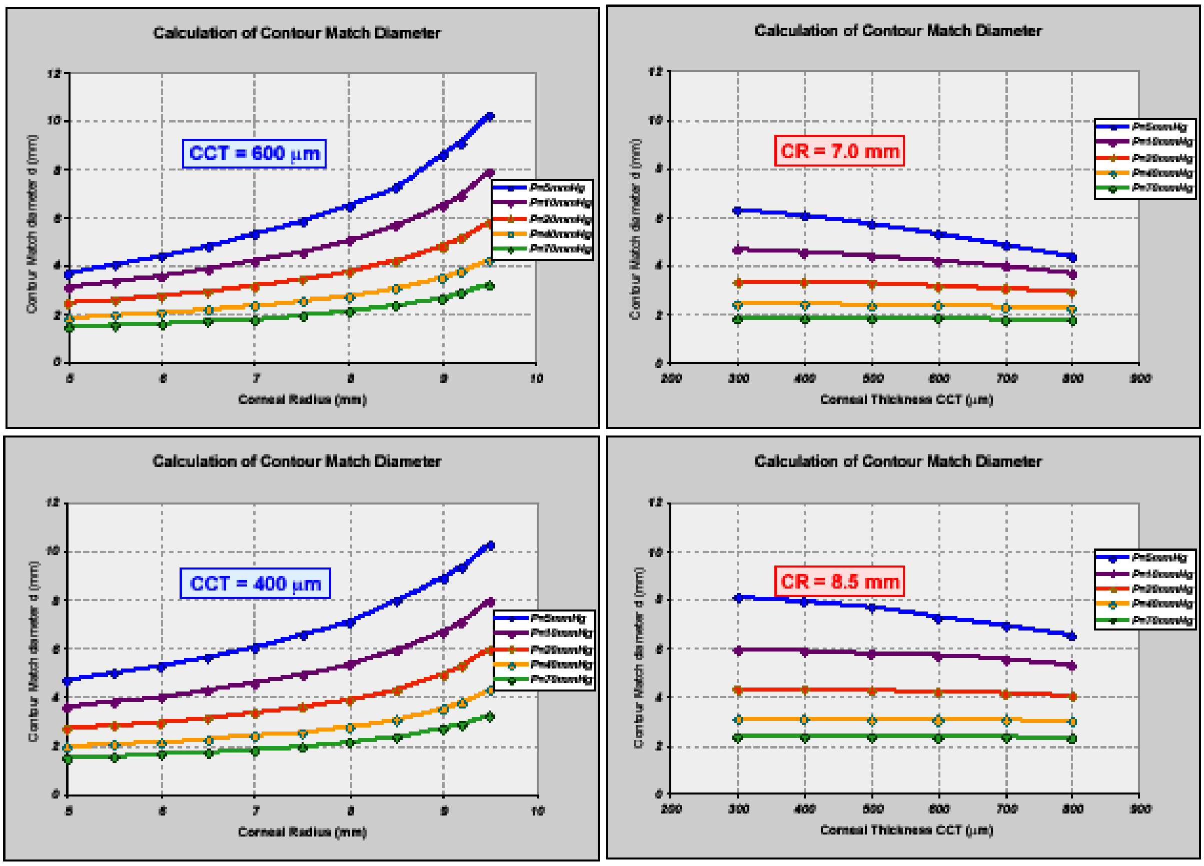


Fig. 1 Geometric parameters used for describing the interaction of contour tonometer tip, tear film, and cornea. In the calculations, Θ_C and Θ_t are substituted by the mean wetting angle Θ.

Figs. 2 – 5 (below) Plots of contact diameter d for selected values of CCT and CR (hc = 400 µm and 600 µm; RC = 7.0 mm and 8.5 mm); at manometric presures ("true IOP") of 5, 10, 20, 40, and 70 mmHg. Results of iterative calculation using eq. (16').



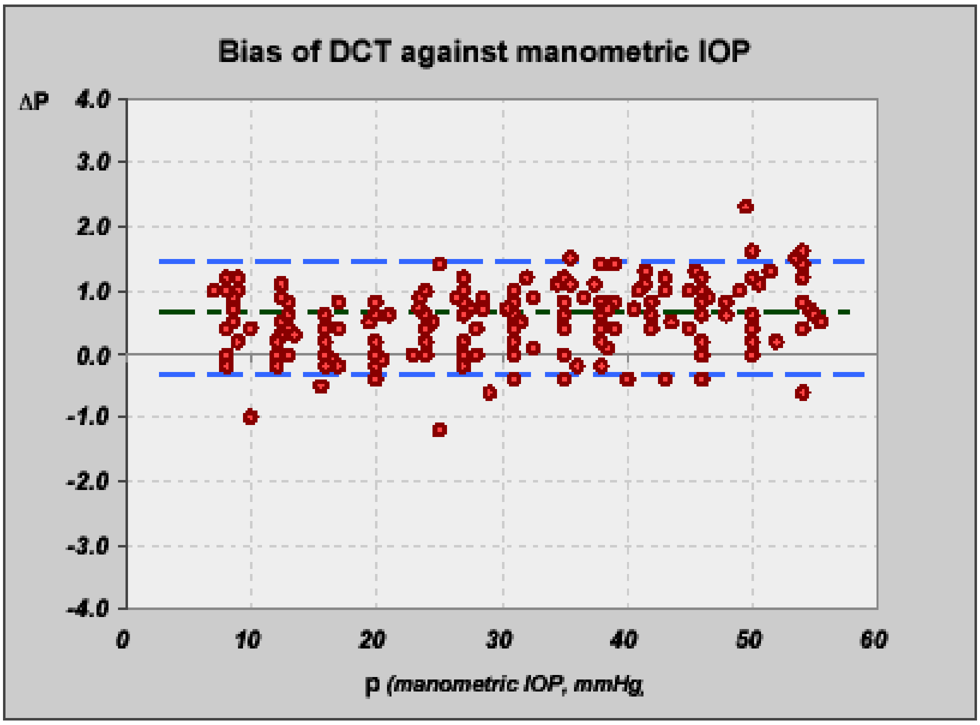


Fig. 6 Bias of tonometer readings against manometric IOP (cadaver eyes). Difference ΔP of contour tonometer reading (mean of 5 consecutive measurements) and manometric IOP is plotted against manometric IOP. Dashed line: mean bias; dotted line: 95% prediction interval of tonometer bias.

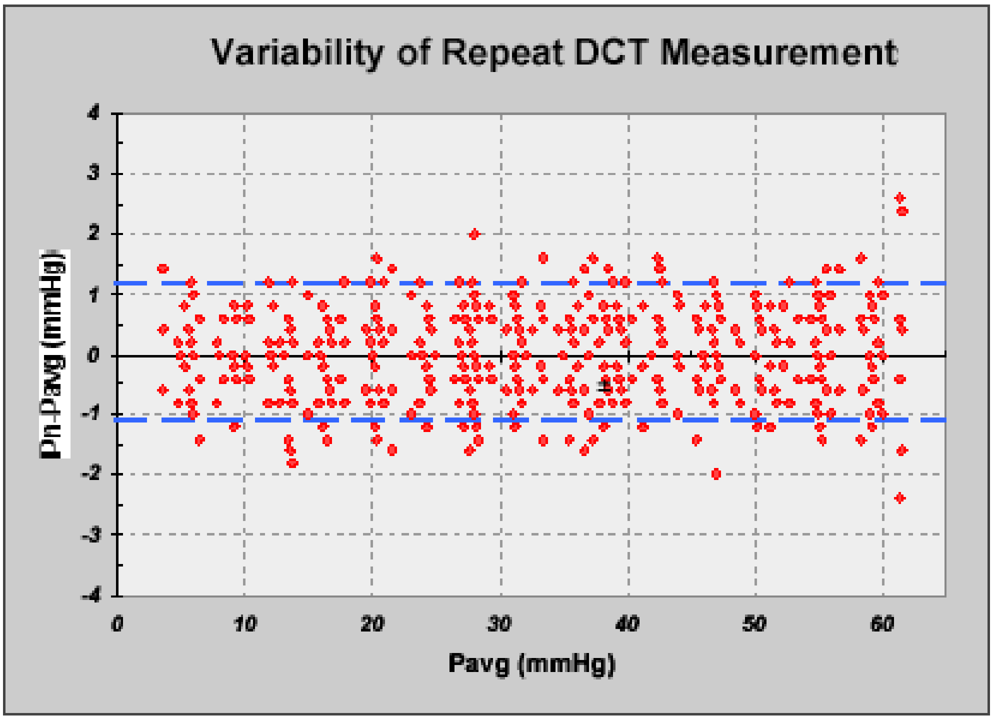


Fig. 7 Variability of repeated measurements with contour tonometer (cadaver eyes). Difference of individual measurements from mean of 5 consecutive repeat measurements is plotted against mean of 5 measurements. Dashed line: 95% limit of agreement of all measurements.

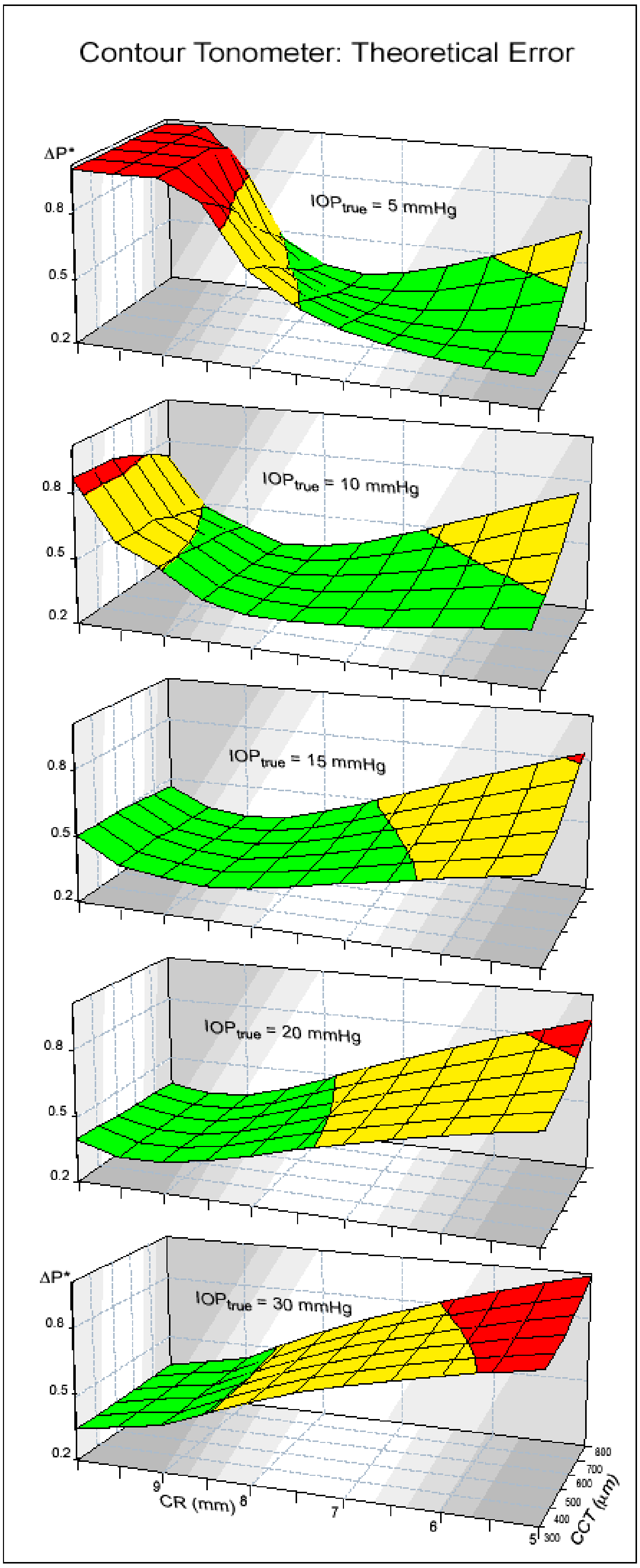


Fig. 8 (left) Theoretical pressure measurement error ΔP* of contour tonometer, plotted against corneal radius CR and corneal thickness CCT for 5 selected values of true IOP (P = 5, 10, 15, 20, 30 mmHg).

Areas where ΔP* < 0.5 mmHg: green;
0.5 > ΔP* > 0.8 mmHg: yellow,
ΔP* > 0.8 mmHg: red.

(Areas where ΔP*>1 were cut off in the topmost graph).

Fig. 9 (below) Dynamic Contour Tonometer constructed by the authors (experimental device). Custom-built contour tonometer tip is mounted on the base of a Goldmann Tonometer, which is set to provide 1 gm appositional force. Pressure measured by built-in sensor is read from LCD display and may be transferred to a connected PC.