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ORIGINAL ARTICLE

On-Eye Power Characteristics of Soft Contact Lenses

SOTIRIS PLAINIS, MSc and W. NEIL CHARMAN, DSc

Department of Optometry and Vision Sciences, University of Manchester Institute of Science and Technology, Manchester, United Kingdom

ABSTRACT: *Background.* The power of a soft contact lens on the eye is a function of its off-eye power, the manner in which the lens flexes on the eye, lens hydration changes, and the corneal topography. *Methods.* In the present study, we used a high illumination keratometer, which allowed us to obtain front and back surface keratometric readings when the lenses were in position on the eye. The on-eye power of the lens could then be calculated from these readings, with the assumption that the center thickness and refractive index of the lens corresponded to those *in vitro. Results.* The estimates of the on-eye powers agreed closely with the results indicated by over-refraction. Moreover, comparison of *in vitro* with *in vivo* power estimates indicated that the positive lenses lost power on-eye, whereas the negative ones maintained their power. *Conclusions.* The present study confirms the results of earlier workers, who suggested that soft lenses drape to fit the cornea. Our findings appeared to be in agreement with the predictions of most of the models developed in the past. (Optom Vis Sci 1998;75:44–54)

Key Words: flexure, soft contact lenses, tear lens, power change, keratometer

Before being placed on-eye, a soft contact lens is normally in a fully hydrated state at room temperature and its back optic zone radius (BOZR) and back vertex power (BVP) should be as specified by the manufacturer. On the eye, several changes may occur, all of which may affect the power of the lens. These include lens flexure, tear lens effects, temperature-induced changes, and lens dehydration and centration. It is important that these changes be understood as fully as possible, so that the practitioner can immediately select a lens of appropriate BVP, rather than having to make a series of empirical adjustments before full correction is obtained. In this article, models for predicting the on-eye power changes of soft lenses are first reviewed and then experimental data are compared with the model predictions.

FLEXURE MODELS FOR SOFT LENSES ON-EYE

Flexure

When a hydrogel lens is placed on the eye, because of its flexibility, it bends to take up the same curvature as the central cornea, or almost so. This bending may affect the power of the lens because of the parameter [e.g., BOZR, front optic zone radius (FOZR)] changes that occur. The result of these changes on the lens BVP is called the "lens flexure effect."

A number of researchers have put forward different theoretical hypotheses to explain the optical effect of flexure or bending of a soft lens on the eye. Others have attempted to explain the effects of this flexure by deriving empirical models from clinical data. The formulas describing the more important of these various flexure hypotheses are shown in Table 1. As each author used his/her own symbols for the various parameters of the lenses, the symbols of the formulas have been changed in the following review to agree with conventional usage.

Kaplan[1] was the first to investigate these changes, using an oversimplified model based on the assumption that when the posterior surface of the lens is bent, the anterior surface remains completely parallel to it, with the thickness unchanged throughout. A year later, Wichterle[2] assumed that the lenses flatten to fit a cornea, producing an alignment fit.

Strachan[3] proposed the equal percentage change hypothesis for the change in the radii of the lens; that is, the front surface wraps by the same proportion as the back surface. He also assumed that the refractive index and the center thickness, t, of the lens do not alter and that a tear lens of zero power is formed between the lens and the cornea. Sarver[4] introduced the equal change hypothesis, according to which there is an equal change in the radii of curvature of the posterior and anterior surface of the soft lens. These two hypotheses predict lens flexure effects which are clinically significantly different (Fig. 1).

Bennett[5] based his model on the assumption that the volume of the lens remains constant even though its curvatures change (constant volume hypothesis). He concluded that both positive and negative lenses change power with flexure, and that the power

TABLE 1.
Mathematical statements of some theoretical flexure hypotheses and empirical models which have been put forward in the past.[a]

Name of Model	Author	Mathematical Statement
Equal Percentage Change (EPC)	Strachan[3]	$r_2'/r_2 = r_1'/r_1$
Equal Change (EC)	Sarver[4]	$r_2' - r_1' = r_2 - r_1$
Constant Volume (CV)	Bennett[5]	$\Delta F = -300\,t\left[\frac{1}{(r_2')^2} - \frac{1}{r_2^2}\right]$
Invariant Normals (IV)	Wallace-Williams and Magabilen (cited by Holden et al.)[7]	$r_2' = \frac{y'^2 + [(r_1'^2 - y'^2)^{1/2} - (r_1' - t)]^2 - t_y^2}{2t_y - 2[(r_1'^2 - y'^2)^{1/2} - (r_1' - t)]}$ y': flexed hemichord length t_y: lens thickness at the hemichord length
Constant arc length	Wallace-Williams and Magabilen (cited by Holden et al.)[7]	$\frac{y'}{y} = \frac{r_2' \sin\theta'}{r_2 \sin\theta} = \frac{r_2' \sin a/r_2'}{r_2 \sin a/r_2}$ a: the arc length y and y′: unflexed and flexed hemichord lengths (see Fig. 3a).
Constant Sagittal Change (CSC)	Smith (cited by Weissman and Zisman)[8]	$r_2' - r_2 = (r_1' - r_1)(r_2^2/r_1^2)$
Power (PW)	Weissman and Zisman[8]	$r_2' = \frac{(1-n) + 0.3375}{P_e - \frac{(n-1)/r_1'}{1 - t(n-1)/n\,r_1'} - \frac{-0.3375}{r_c}}$ P_e: Effective Power
Fatt and Chaston (FC)	Chaston and Fatt[12]	$r_1' = r_1\sqrt{r_2'/r_2}$
Alignment (AL)	Holden and Zantos[17]	$r_2' = r_c$
Beam-Bending	Janoff and Dabezies[11]	—
Weissman (W)	Weissman[13]	$r_2' - r_2 = \frac{r_1' - r_1}{1.06 - 0.05\,F}$ F: *In vitro* BVP

[a] Symbols: r_1, unflexed FOZR; r_2, unflexed BOZR; r_1', flexed FOZR; r_2', flexed BOZR; r_c, anterior corneal radius; F, *in vitro* BVP; ΔF, change in BVP; n, refractive index; t, thickness.

change (which is always in the negative direction) is not affected by the initial power of the lens. Voerste[6] provided good clinical evidence in support of Bennett's theoretical work by finding a preponderance of low negative supplemental power effects in a study on over 200 eyes.

Holden et al.,[7] in an experimental study of 88 lenses, concluded that the "equal change" and "equal percentage change" hypotheses were simplistic models and misleading when applied to the type of lenses studied (Fig. 2). Furthermore, they presented two additional hypotheses from an earlier thesis by Wallace-Williams and Magabilen. The first, the constant arc length hypothesis, maintained that the arc length of the back optic zone of the lens remains constant under flexure (see Fig. 3a); this was later simplified to the percentage change hypothesis by Weissman and Zisman.[8]

The other hypothesis proposed by Wallace-Williams and Magabilen (cited by Holden et al.[7]) is the invariant normal hypothesis. This assumes that the front surface of the lens stretches and the back surface compresses as the lens flexes about a cornea steeper

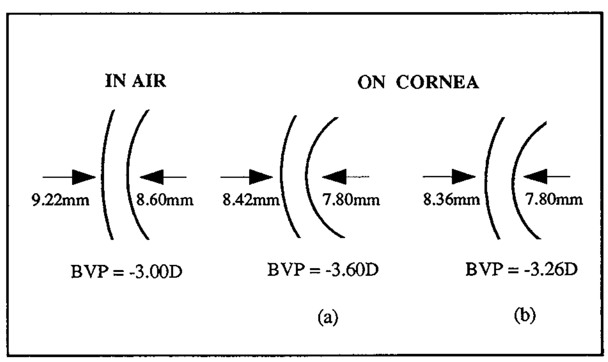

FIGURE 1.
The prediction of on-eye power for a −3.00 D hydrogel lens according to (a) the equal change hypothesis and (b) the equal percentage change hypothesis. The refractive index of the lens is taken as 1.40 and its center thickness as 0.10 mm. Note that in the equal change hypothesis, the change in BVP on-eye for the same lens is higher than the one predicted by the equal percentage change hypothesis.

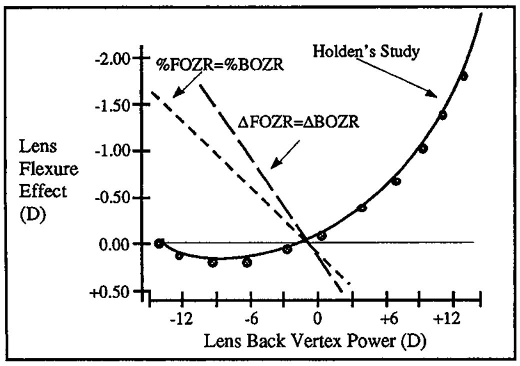

FIGURE 2.

Lens flexure effect observed in Holden et al.'s study[7] and as predicted by the equal change and the percentage change hypotheses. The experimental results indicate that, contrary to equal change and equal percentage change hypotheses, minus powered lenses gain very little power on bending, and plus lenses "lose" power.

than the back radius (Fig. 3b). The hypothesis is similar to beam-bending theory used in engineering.

Bibby[9] used a combination of the invariant normals and constant arc hypotheses to describe an alternative model of lens flexure—the strain-free boundary model. According to this model, between the anterior and posterior surface of the lens there exists a boundary which changes in shape but not in length. Bibby's model was supported by Wechsler et al.'s[10] data. Later, Janoff and Dabezies[11] proposed a variation of the beam-bending hypothesis for plus lenses. Their theory suggests that the back surface does conform to the corneal surface but, because of the greater center thickness of a plus lens in comparison with minus lenses, the front surface cannot stretch enough to allow complete flexure. Inasmuch as the flexure of the back surface creates a minus power effect, and because the front surface does not flex enough to create a sufficient plus power effect, the net result is a loss of power when the positive lens is on the cornea.

Smith (cited by Weissman and Zisman[8]) proposed a constant change in sagittal depth for each surface (constant sag change hypothesis) during flexure. Weissman and Zisman[8] developed the power hypothesis. They calculated the flexed back radius r'_2 by using the effective power of the lens *in situ*, which was considered to be the sum of the flexed power of the contact lens and the tear lens. Chaston and Fatt[12] used Weissman and Zisman's analysis and developed an independent model incorporating their own clinical data for plus lenses.

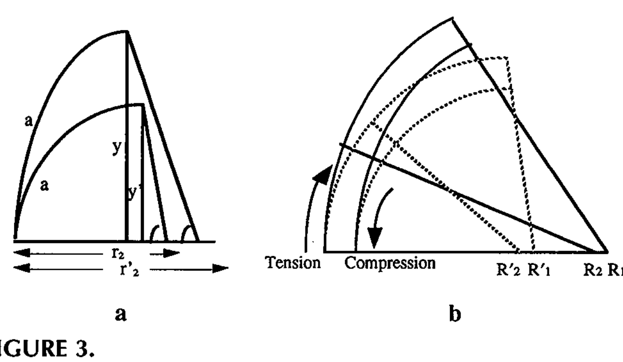

FIGURE 3.

a: Diagram of concept for "constant arc length hypothesis": a is the arc length, y and y' the respective unflexed and flexed hemichord lengths, r_2 and r'_2 the respective unflexed and flexed BOZR (redrawn from Weissman and Zisman[8]). b: Redrawing of figure illustrating "invariant normals hypothesis" from Holden et al.[7]: R_1, R'_1 are the centers of curvature of the front surface radii r_1 and r'_1 unflexed and flexed; R_2, R'_2 the centers of curvature of the back surface radii for the same situations.

Weissman[13, 14] suggested that power change is primarily dependent on original unflexed power (contrary to Bennett's hypothesis), and that minus lens systems (hydrogel lens and potential tear lens) do not generate substantial power changes with flexure, whereas plus lens systems consistently lose plus power. For small values of unflexed power, the relation approximates to the equal change hypothesis.

Weissman and Gardner[15] examined power proposals and suggested that different lens designs may result in different relations between flexing front and back surfaces. Two empirical models were described: $\Delta r_2 \cong 2\ \Delta r_1$ appeared to describe the relation for low plus-powered lenses, and $\Delta r_2 \cong 0.75\ \Delta r_1$ seemed to describe the relation for low minus lenses.

Tear lens

There has been considerable discussion in the literature regarding the possibility that a powered tear lens might be trapped under a soft lens *in situ*. Wechsler et al.[10] presented photographs of various lenses on the eye, with large molecule fluorescein instilled; these showed no noticeable tear lens of a finite power. Weissman and Zisman[16] also attempted to calculate the approximate tear volume trapped under a flexible contact lens *in situ*. They tested a series of standard central thickness (about 0.15 mm) hydrogel lenses and concluded that an optically significant tear lens exists of 10 μl volume. Chaston and Fatt[12] and Holden and Zantos[17] suggested that the anterior corneal surface and posterior lens surface closely align. Michaels and Weissman[18] stated that although some soft contact lenses may closely align with the anterior corneal surface, producing a tear layer of negligible power and minimum volume, many lens-eye situations sometimes trap a tear lens with volume and power of approximately 5 to 10 μl and −0.15 D, respectively.

Weissman and Gardner,[19] however, presented evidence suggesting that thin, low minus lenses may entrap tears of low volume (approximately 5.5 μl) and minimal power, but that thicker, low plus lenses may trap tears of both greater volume (approximately 9.5 μl) and power (up to −2.00 D). This, added to the power change from flexure alone, results in a further decrease in overall plus power, and represents the classic "supplemental power effect."

Note that, in conformity with the usage of earlier authors, the tear fluid has been described in the above discussion as being "trapped" behind the lens. In practice, of course, tear lens exchange will occur with blinking, but it may be expected that any tear lens volume and power will remain approximately constant.

Temperature

As the temperature of the cornea is approximately 32°C and room temperature is about 20°C, there is a change in temperature of the lens when it is put on the eye. Hydrogel lenses undergo changes in all of their parameters with change in temperature (Fatt and Chaston[20]). The important effect of an increase in temperature is the reduction in water content. The loss of water will cause a reduction in lens thickness, total diameter, and FOZR, but no change in BOZR, which remains parallel to the cornea. Tranoudis[21] surveyed a group of lenses from eight contact lens groups and showed that the decrease in water content due to temperature is statistically significant. Fatt and Chaston[22, 23] found that the loss of on-eye power of hydrogel lenses, because of the increase in temperature, can be 0.50 to 2.50 D, depending on the water content of the lens (high water content lenses lose more power).

Dehydration

Other factors, such as the fact that the front surface of the soft lens is exposed to the air, leading to evaporation, may further decrease the water content of the lens. Water loss can occur by several pathways, including evaporation into the atmosphere, drainage into the nasolacrimal system, and possibly absorption into the conjunctival capillaries.[24] Water evaporation from the anterior surface of a hydrogel contact lens will not lead to a totally dry lens because of water replenishment from the tear film at the posterior surface of the contact lens.[25]

Masnick and Holden[26] provided data from which a noticeable water loss can be calculated for lenses of medium water content on the eye when humidity is 60 to 80%. Hamano and Kawabe[27] showed that the base curve of a soft lens steepens with dehydration. Fatt and Chaston[20] postulated that dehydration can increase the refractive index of a lens and decrease its center thickness. Other authors[28–33] showed that dehydration does occur when a soft lens is worn, and the amount varies with polymer type, lens form, length of wearing time, and environmental conditions. The mean data of Wechsler et al.[32] indicated a reduction in hydration of approximately 6.5% at the end of 1 h of lens wear.

It was also found that high water content soft lenses lose a smaller percentage of their water than do those of low water content.[34, 35] Andrasko[30] stated that higher water content lenses are more affected by hydration changes than lower water content ones, and thinner lenses may be more affected than thicker ones. He also reported that the lenses at equilibrium (on-eye) maintain approximately 80 to 93% of their fully saturated water content, and that the lenses dehydrate more in low than in high humidity environments. However, Cohen and Gundel[36] found on average a greater percent of dehydration for the thick high water than for thin low water content lenses, which is inconsistent with Andrasko.[29, 30]

Summary of flexure models

Soft lenses are fitted large and flat. Because the pliable hydrogel lens is flatter than the cornea, it must at least approximately conform to the eye by a change in its radii. Several theories have been proposed to describe the optical and mechanical effects of hydrogel lens flexure *in situ,* but the preceding review shows that none of them has been unanimously accepted.

Flexure not only alters effective power, it also creates a tear lens. Some authors tried to quantify experimentally or theoretically the tear volume trapped under a flexible contact lens,[16, 18] but none of them actually measured the power of such a lens.

Dehydration of a soft lens *in situ* can steepen the back radius of curvature,[27] increase index of refraction, and decrease lens central thickness[20] and thereby can affect on-eye power. Temperature is not the sole factor controlling lens hydration. Individual human and environmental factors (such as wind conditions, humidity, elevation, and season) can also play a role in the changes.

Finally, it should be stressed that over the years there has been a reduction in the typical thickness of soft contact lenses while new materials have been introduced by manufacturers. Thus, models which were appropriate for early lenses of the 1970s may no longer be appropriate for today's much thinner lenses.

EXPERIMENTAL STUDIES OF POWER CHANGES OF SOFT LENSES ON-EYE

To explore on-eye lens power changes further, an attempt has been made to correlate experimental *in vitro* and *in vivo* lens powers with on-eye lens parameter measurements and model predictions. This study used a battery of equipment, including a high illumination keratometer, which enables keratometric readings of both on-eye lens radii to be obtained. Therefore, the actual (and not the hypothetical or predicted) on-eye power of the lens could be calculated, and the power of the entrapped tear lens could be estimated.

METHODS

Subject

Only one eye (right eye) of one subject was used in our study. The subject had no significant refractive error in this eye (plano correction, measured objectively and subjectively). The corneal radii (two meridians) were measured with a Zeiss keratometer. Videokeratographic measurements (EyeSys Corneal Analysis System) showed that peripheral corneal topography was within normal limits.

Lenses

Type. Ten single-vision and eight toric lenses were used for the measurements. The single-vision lenses were Calendar monthly disposable contact lenses (manufactured by Pilkington Barnes-Hind). The torics were two front surface (Optima series, manufactured by Bausch & Lomb), three back surface (C.S.I. torics, manufactured by Pilkington Barnes-Hind), and another three back surface lenses (Hydrocurve II, manufactured by Pilkington Barnes-Hind).

BVP. Five minus and five plus single-vision lenses were considered. The nominal BVP of the minus lenses ranged from −2.50 to −6.50 D, and for the plus lenses ranged from +1.75 to +4.50 D. The sphere for the toric lenses ranged from +3.00 to −3.00 D, and the cylinder from −1.00 to −2.00 D.

Fitting. The contact lenses used in the study had not been ordered for the subject's eye. The difference between the base curve of the lens and the flattest corneal meridian ranged from 0.21 to 0.91 mm, and the mean of the differences was 0.61 mm. The diameter of the lenses ranged from 13.90 to 14.50 mm, with the mean being 14.23 mm.

Procedure

The power of the lenses *in vitro* was measured with a projection focimeter (Topcon LM-P6 model) having a power resolution of 0.02 D. For the toric lenses, a Tori-Check contact lens measuring aid was used to evaluate if the lens markings and the cylinder axis were correct.[37] Central thickness was measured with an electronic thickness gauge (±0.004 mm accuracy), the measurement of the water content was performed with a hand-held soft contact lens refractometer,[38, 39] and the refractive index (±0.001 accuracy) was calculated from the water content.[40]

The equipment used for the measurements of the BOZR and FOZR both *in vitro* and *in vivo* was the Zeiss keratometer (ophthalmometer) (Carl Zeiss Jena 310 M). The Zeiss ophthalmometer was used because its illumination system is very efficient and images can be detected from weakly reflecting surfaces.[41] The unflexed radii (*in vitro*) of the contact lenses were measured by the wet-cell/keratometer method.[42] Because keratometry is performed through the contact lens back surface, the front surface keratometric readings obtained represent the apparent FOZR. Apparent FOZR was converted to real FOZR by using an appropriate formula (see Appendix A). The on-eye lens radii (which we have termed FOZRE and BOZRE) were also measured with the Zeiss keratometer, as mire reflections can be obtained not only from the front surface of a contact lens (which is brighter), but also from its back surface[17, 43] (Fig. 4). The back surface keratometric readings obtained were not the real ones, because of the refraction of the light rays at the front surface, before and after being reflected by the back surface. Apparent BOZRE was converted to true BOZRE by using an appropriate formula (see Appendix B).

The equivalent power of the lens on-eye was calculated through four measured variables (BOZRE, FOZRE, t, n) by using the thick lens power formula

$$\left(\text{BVPE} = \frac{n-1}{r_1'} + \frac{1-n}{r_2'} + \frac{t}{n}\frac{(n-1)^2}{r_1' \, r_2'}\right).$$

The refractive index (n) and the center thickness (t) of the lens on-eye were assumed to remain constant and to correspond to the values measured *in vitro.* Therefore, the change in power of the lens was determined solely by the changes to BOZRE and FOZRE. Calculation of BVPE from four measured variables (n, t, r_1', r_2'), each of which is subject to error, leads to a greater proportional variability in the estimate of BVPE (≤0.25 D). However, if there was no bias in the way the errors in the four individual variables combined, it would be reasonable to expect that although such errors would cause the reliability of any particular estimate of BVPE to be lower, the average estimate of BVPE would be unaffected.

A streak retinoscope and an autokerato-refractometer (Topcon KR-7500) were used with each lens in place to determine effective power (P_e) of the lens on-eye.

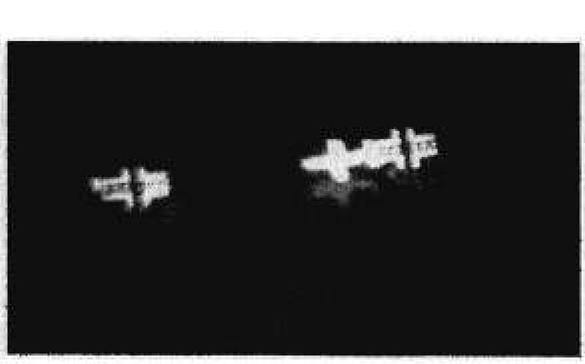

FIGURE 4.
"Sub K" and "Over K" mires when a lens is *in situ.* The bright image comes from the front surface of the lens and the faint secondary image is the one reflected by the back surface.

RESULTS

Tables 2 and 3 give the radii of the lenses, as measured with the Zeiss keratometer, *in vitro* and *in vivo.*

Degree of conformity and tear lens

The power of the tear lens was deduced from the on-eye back radius of the lens, r_2', the anterior corneal radius, r_c, and the tear refractive index (1.336) by using the following equation:

$$\text{tear lens power} = (n_{TL} - 1)\left(\frac{1}{r_2'} - \frac{1}{r_c}\right)$$

It was assumed that the tear lens thickness was negligible, so that the thin lens approximation was adequate.

It can be seen from Table 4, which shows mean results for the different groups of lenses, that most of the lenses conform closely to the shape of cornea. Thus, the calculated mean power of the tear lens formed between the lens and the cornea is small and, when account is taken of the errors in parameter measurement, does not differ significantly from zero.

Change in back vertex power (ΔBVP)

It appears that the behavior of the negative lenses when placed on the eye differs from that of the positive ones. As is illustrated in Fig. 5, because of flexure, the positive single-vision lenses lose power on-eye (mean difference −0.45 and −0.54 D in the two meridians). However, the negative single-vision lenses maintain their power (mean differences 0.09 and −0.07 D). The eight toric lenses also maintain their power; this would be expected, as seven of them are negatively powered. No obvious differences were found in the behavior of the different types of front and back surface toric lenses in this or any other aspect of behavior, although the number of each type of lens was small. All the toric lenses are therefore grouped together.

The Effective Power (Pe, *in vivo* power indicated by over-refraction) behaves similarly to the BVPE. Fig. 6 illustrates the difference (Supplemental Power) between Effective Power and *in vitro* BVP. Again, the positive lenses seem to lose power on-eye (mean difference −0.55 D for both meridians), whereas the power of the negative lenses remains almost the same (mean difference −0.14 D). It should, however, be stressed that the loss appears to increase with increasing unflexed positive power of lens and increasing center thickness (Fig. 7). There is also a decrease in the power of the toric lenses (mean difference −0.46 and −0.64 D in the two meridians).

As can be seen from the comparison of the calculated total on-eye power (BVPE) and the Effective Power (Pe) estimated by over-refrac-

TABLE 2.
In vitro parameters of lenses.

Lens	t (mm)	n	r_1 (mm)		r_2 (mm)		F (D)	
			180°	90°	180°	90°	180°	90°
SV1	0.094	1.378	9.63	9.66	8.26	8.26	−6.54	−6.54
SV2	0.101	1.378	9.26	9.29	8.27	8.27	−4.72	−4.72
SV3	0.136	1.378	9.13	9.13	8.37	8.37	−3.50	−3.50
SV4	0.152	1.378	9.07	9.07	8.37	8.37	−3.23	−3.23
SV5	0.135	1.378	8.88	8.88	8.39	8.39	−2.50	−2.50
SV6	0.223	1.378	8.74	8.69	8.85	8.85	+1.87	+1.87
SV7	0.230	1.378	8.65	8.69	8.90	8.90	+2.10	+2.10
SV8	0.239	1.378	8.61	8.61	8.84	8.84	+2.24	+2.24
SV9	0.234	1.378	8.70	8.70	8.99	8.99	+2.46	+2.46
SV10	0.274	1.378	8.33	8.33	9.00	9.00	+4.55	+4.55
T1	0.161	1.432	8.39	8.55	8.30	8.30	−0.06	−1.12
T2	0.232	1.432	8.24	8.04	8.29	8.29	+1.47	+2.94
T3	0.076	1.440	9.52	9.53	8.80	8.65	−3.37	−4.76
T4	0.103	1.440	9.25	9.25	8.71	8.59	−2.90	−3.86
T5	0.098	1.440	9.47	9.47	8.77	8.62	−3.15	−4.26
T6	0.098	1.443	9.50	9.50	8.81	8.56	−3.39	−4.77
T7	0.110	1.443	9.57	9.57	8.96	8.68	−2.52	−4.40
T8	0.094	1.443	9.63	9.63	8.95	8.48	−2.67	−4.72

TABLE 3.
Parameters of lenses on-eye as measured with the Zeiss keratometer and calculated from the appropriate formulas.

Lens	r_1' (mm)		r_2' (mm)		BVPE (D)		Tear Lens (D)		Total Power (D)		Pe (D)	
	180°	90°	180°	90°	180°	90°	180°	90°	180°	90°	180°	90°
SV1	9.39	9.17	8.06	7.87	−6.51	−6.67	−0.05	+0.06	−6.56	−6.61	−6.76	−6.76
SV2	8.96	8.75	8.11	7.87	−4.28	−4.68	−0.31	+0.06	−4.59	−4.62	−4.78	−4.78
SV3	8.76	8.57	8.07	7.81	−3.49	−4.08	−0.10	+0.38	−3.59	−3.70	−3.65	−3.65
SV4	8.67	8.49	8.06	7.97	−3.07	−2.67	−0.05	−0.48	−3.12	−3.15	−3.24	−3.24
SV5	8.58	8.36	8.05	7.85	−2.70	−2.72	0.00	+0.16	−2.70	−2.56	−2.75	−2.75
SV6	7.86	7.69	8.06	7.87	+1.56	+1.51	−0.05	+0.06	+1.51	+1.57	+1.50	+1.50
SV7	7.78	7.63	8.01	7.84	+1.78	+1.73	+0.21	+0.22	+1.99	+1.95	+1.50	+1.50
SV8	7.81	7.65	8.01	7.84	+1.60	+1.61	+0.21	+0.22	+1.81	+1.83	+1.75	+1.75
SV9	7.75	7.64	8.03	7.87	+2.09	+1.85	+0.10	+0.06	+2.19	+1.91	+2.00	+2.00
SV10	7.48	7.36	8.03	7.87	+3.93	+3.82	+0.10	+0.06	+4.03	+3.88	+3.75	+3.75
T1	8.15	8.12	8.06	7.90	−0.27	−1.15	−0.05	−0.10	−0.32	−1.25	0.00	−1.50
T2	7.88	7.63	8.03	7.89	+1.50	+2.37	+0.10	−0.05	+1.60	+2.32	+1.00	+1.75
T3	8.57	8.72	8.03	7.92	−3.30	−4.95	+0.10	−0.21	−3.20	−5.16	−3.65	−5.60
T4	8.55	8.57	8.07	7.90	−2.86	−4.15	−0.10	−0.10	−2.96	−4.25	−3.08	−4.18
T5	8.60	8.66	8.08	7.90	−3.10	−4.70	−0.15	−0.10	−3.25	−4.80	−3.65	−4.48
T6	8.66	8.72	8.09	7.97	−3.41	−4.59	−0.20	−0.48	−3.61	−5.07	−4.27	−5.86
T7	8.50	8.66	8.11	7.89	−2.29	−4.77	−0.30	−0.05	−2.59	−4.82	−2.84	−4.73
T8	8.54	8.65	8.11	7.89	−2.57	−4.75	−0.30	−0.05	−2.87	−4.80	−3.78	−5.43

[a] Total Power is the sum of the calculated values of BVPE and tear lens power; Pe is the effective power (*in vivo* power indicated by over-refraction).

tion (Table 3), the Zeiss keratometer slightly overestimates the on-eye power of all the lenses in comparison with the over-refraction. The possible explanations of these discrepancies could be: (1) the uncertainty in the indirect calculation of BVPE by using the thick lens formula. Calculation of the BVPE from four measured variables (refractive index, thickness, front, and back radii of curvature), each of which is subject to error, could lead to a combined variability in the estimate of BVPE of ±0.25 D. And (2) the assumption that the refractive index and the thickness of the lenses remain constant *in vivo*. However, it has been stated that dehydration occurs when a lens is placed on the eye. A decrease in water content of the lens would result in an increase in the refractive index of the lens and, consequently, an increase in its power. This agrees with the findings of negative lenses (SV minus and toric), but cannot justify the findings of the positive ones. Although the positive lenses are thicker, and a decrease in center thickness with dehydration would reduce their power, the effects are likely to be very small. For example, a 5% decrease in water content is likely to reduce lens thickness by approximately 1%, which corre-

TABLE 4.
Mean differences between the BOZR of the lens (r_2') and anterior corneal radius of curvature (r_c), and mean power of tear lens trapped between the two surfaces.

Type of Lens	Meridian	No.	r_2'-r_c (mm)		Tear Lens (D)	
			Mean	± SD	Mean	± SD
Plus SV lenses	H	5	−0.020	0.020	+0.11	0.10
	V	5	−0.022	0.016	+0.12	0.08
Minus SV lenses	H	5	0.020	0.023	−0.10	0.12
	V	5	−0.006	0.059	+0.04	0.31
All SV lenses	H	10	0.000	0.003	+0.00	0.15
	V	10	−0.014	0.041	+0.08	0.22
Toric lenses	H	8	0.023	0.031	−0.11	0.15
	V	8	0.028	0.027	−0.14	0.15
All lenses	H	18	0.012	0.032	−0.06	0.16
	V	18	0.007	0.041	−0.03	0.22

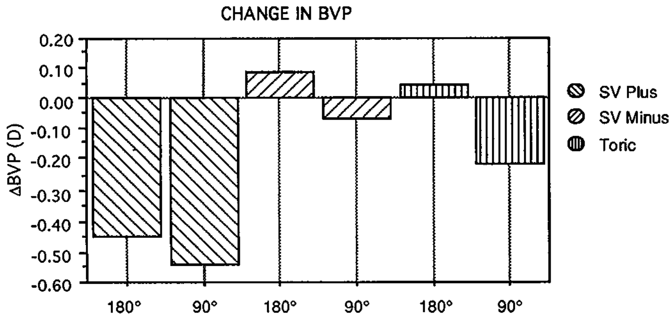

FIGURE 5.
Plot of the mean difference (ΔBVP) between calculated on-eye power (BVPE) and measured *in vitro* BVP because of lens flexure, for all types of the lenses used. The BVPE was calculated from keratometer measurements of on-eye lens radii and *in vitro* measurements of lens thickness and index.

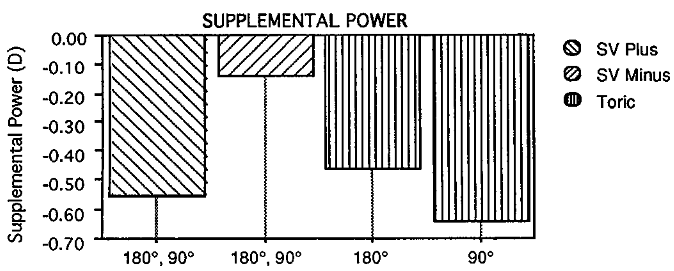

FIGURE 6.
Plot of Supplemental Power (Effective Power-BVP) for all types of lenses used. The Effective Power is the on-eye power as indicated by over-refraction; the BVP is the lens power measured *in vitro* (off the eye).

sponds to a thickness change of only about 1 μm for a center thickness of 0.12 mm.

DISCUSSION
Comparison with theoretical predictions

As was discussed earlier, various flexure models of soft contact lenses were developed in the past. These models were applied to the lenses used in the present study. With the aid of the mathematical statements displayed in Table 1 and the parameters measured with the Zeiss keratometer, on-eye predicted power (BVPE) can be

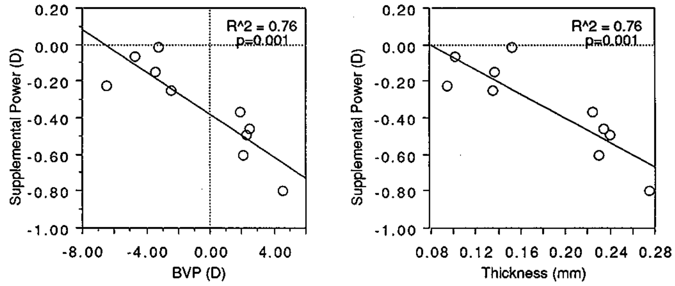

FIGURE 7.
Plots of Supplemental Power (difference between on-eye power indicated by over-refraction and *in vitro* power) for single-vision lenses as a function of their BVP and thickness. The full line is the least squares regression fit to the data points. The R^2 is the coefficient of determination, and the p value refers to the regression analysis.

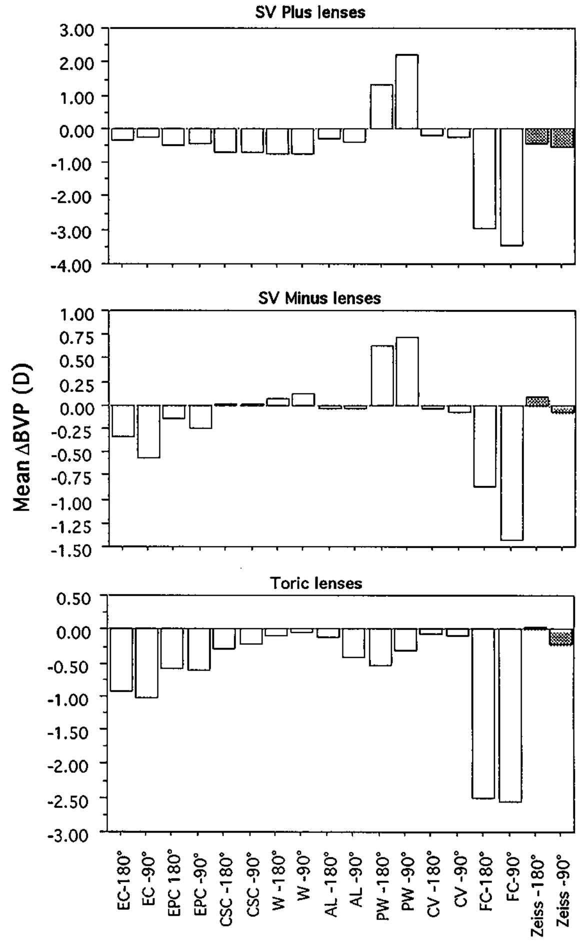

FIGURE 8.
Mean ΔBVP for all types of lenses as predicted by the various hypotheses and as calculated by the present study (Zeiss). The calculated values for the present study are based in Zeiss keratometer radii measurements, and the assumption that center thickness and refractive index are unchanged.

calculated. In Fig. 8 the mean ΔBVP (change in lens power because of flexure) predicted from these hypotheses is compared with the corresponding results from the present study. Table 5 gives a breakdown of the statistical significance of the differences between the predictions of ΔBVP by the various models and the experimental data for the three types of the lenses studied.

From Fig. 8 and Table 5 it is clear that the prediction of the Fatt

TABLE 5.
Significance table for the difference between the values of ΔBVP calculated from the various models and the Zeiss keratometer results obtained from the present study.[a]

Name of Model	Plus SV Lenses		Minus SV Lenses		Toric Lenses	
	180°	90°	180°	90°	180°	90°
Constant Volume (CV)	—	—	—	—	—	—
Power (PW)	S[b]	S	S	S	—	—
Alignment (AL)	—	—	—	—	—	—
Weissman (W)	—	—	—	—	—	—
Fatt and Chaston (FC)	S	S	S	S	S	S
Constant Sagittal Change (CSC)	—	—	—	—	—	—
Equal Percentage Change (EPC)	—	—	—	—	—	—
Equal Change (EC)	—	—	S	—	S	—

[a] The statistical test used was the Ancova test with Dunnett's two-tailed *post hoc* comparison (at significance level of 5%). Note that predictions of the Power (PW) and Fatt and Chaston (FC) models differ significantly from the experimental findings.
[b] S, significantly different at 5%.

and Chaston[12] (FC) and Power[8] (PW) hypotheses consistently differs significantly from those of other models and the findings of the present work for all types of lenses. Also, the equal change hypothesis (EC)[4] yields different results for the minus single-vision and toric lenses.

The results of the Fatt and Chaston flexure hypothesis appear unreasonable (Fig. 8, Table 5). Although the Fatt and Chaston model was developed from data, it should be stressed that only high-powered soft lenses, which were fitted on simulated eyes (polymethyl methacrylate domes), were used in their study.

The Power (PW) hypothesis[8] results also show a statistically significant difference from the present study for the single-vision lenses. This can be explained by the fact that observed power itself (measured by over-refraction) was used to generate predicted flexed lens system parameters.

The predictions of the equal change (EC) hypothesis are reasonably close to the findings of the present study for plus single-vision lenses, but exhibit a considerable difference for the negative single vision and toric lenses (Fig. 8), which is statistically significant in the horizontal meridian (Table 5).

All the other models in most cases are in close agreement and show no statistical difference, although closer examination of the predictions of the equal percentage change (EPC) hypothesis shows systematic discrepancies.

The alignment[17] (AL) and Weissman[13] (W) models closely agree with the present study. This could be predicted as they also were developed from empirical data and involved essentially the same assumptions.

The models which were developed from conjecture (i.e., no data were used) and closely predict the performance of both positive and negative lenses are the constant sagittal change (CSC)[8] and the constant volume (CV)[5] ones. Both hypotheses predict little change in the power of the negative lenses *in situ* and a considerable decrease (greater for the constant sagittal change hypothesis) in the on-eye power of the positive lenses.

An alternative, simpler comparison between observed changes and several model predictions can be made in terms of the radius changes. Writing $\Delta r_1 = r_1' - r_1$, $\Delta r_2 = r_2' - r_2$, it can easily be shown that the model predictions given in Table 1 can be rewritten as:

$$\frac{\Delta r_2}{\Delta r_1} \left\{ \begin{array}{ll} = r_2/r_1 & \text{(EPC, Strachan}^3\text{)} \\ = 1 & \text{(EC, Sarver}^4\text{)} \\ = (r_2/r_1)^2 & \text{(CSC, Smith}^8\text{)} \\ = 2(r_2/r_1) & \text{(FC, Chaston and Fatt}^{12}\text{)} \\ = \dfrac{1}{1.06 - 0.05F} & \text{(W, Weissman}^{13,14}\text{)} \end{array} \right.$$

Fig. 9 shows the observed mean values of $\Delta r_2/\Delta r_1$ for single-vision lenses as a function of lens BVP in comparison with the model predictions using the r_1 and r_2 values of Table 2. It is evident that the Chaston and Fatt, FC, model is inadequate and that the equal change hypothesis, EC, is very approximate. The other models all match the data reasonably well and, at least over the power range studied, appear to be reasonably good descriptions of the on-eye radius changes.

It is, however, desirable that further studies are carried out with a wider range of lens materials and parameters to further refine the available models. It would, for example, be interesting to explore the effect of lens thickness in more detail, in view of the apparently strong correlation between supplemental power and thickness in Fig. 7. We note, too, that our data were obtained with a single subject. Although this has advantages in producing mutually consistent results, it can be criticized in that there were inevitably variations in the fitting relationship of the different lenses to the cornea (Table 2). It could therefore be desirable to extend the study to explore both the effect of the fitting relationship on flexure and the possibility that marked inter-subject differences may occur due, perhaps, to differences in corneal topography.

CONCLUSIONS

The present study confirms the results of earlier workers, who suggested that soft lenses drape to fit the cornea. Tear lens powers

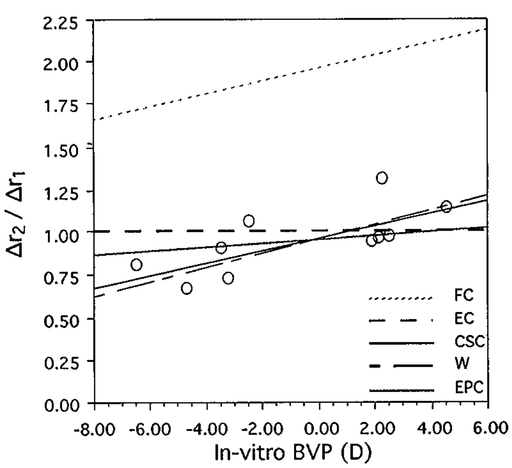

FIGURE 9.
Plot of observed mean values of $\Delta r_1/\Delta r_2$ (O) for single-vision lenses as a function of lens BVP in comparison with the model predictions (see also Table 1).

were found to be very low (≤0.25 D) with both single-vision and toric lenses.

The observed small changes in on-eye power are compatible with the predictions of several models (constant sagittal change, constant volume, alignment, and Weissman models). They are, however, poorly predicted by the Fatt and Chaston, Power, and Equal Change hypotheses, and the predictions of the Equal Percentage Change hypothesis also show discrepancies.

It is interesting that in practice several apparently different models in fact predict very similar results, and that experimental data cannot distinguish between the success of their predictions. This may suggest that the different models are simply statements of the same basic relationship in slightly different forms, at least for today's thin lenses.

APPENDIX A
Calculation of real FOZR from apparent FOZR obtained with a keratometer

The back surface readings (in saline) of a contact lens obtained with a keratometer represent the real BOZR of the lens (when converted to air values). Because keratometry is performed through the contact lens back surface, the front surface keratometric readings obtained represent the apparent FOZR. Therefore, measurements of the front surface through the back surface do not give a true value of the FOZR. This happens because the light rays from the keratometer mires undergo refraction at the back surface before being reflected by the front surface (Fig. 10), and then undergo a second refraction at the back surface before reaching the keratometer.

We need to calculate the position of A_1 through r_2, but C_1 is anterior to lens.

For image of A_1 through r_1 (Fig. 10):

$$\frac{1.336}{l'} - \frac{n}{-t} = (1.336 - n)\frac{1}{r_2} = F_2,$$

where r_2 is taken as +ve and F_2 is taken as −ve

$$\text{i.e., } \frac{1.336}{l'} = F_2 - \frac{n}{t} = \frac{F_2 t - n}{t} \Leftrightarrow$$

$$\Leftrightarrow l' = \frac{1.336\ t}{F_2 t - n}$$

Hence, apparent radius $r'_{1\alpha}$ equals:

$$r'_{1\alpha} = (r_1 - t) - \left(\frac{1.336\ t}{F_2 t - n}\right) \Leftrightarrow r_1 = r'_{1\alpha} + t\left[1 + \frac{1.336}{F_2 t - n}\right]$$

where F_2 is taken as −ve and thickness (t) as +ve, and:

n = lens refractive index
t = center thickness
r_1 = real FOZR
$r_{1\alpha}$ = apparent FOZR
F_2 = back surface power [$F_2 = (1.336 - n)\ 1/r_2$]
r_2 = real BOZR
C_1 = real center of curvature
$C_{1\alpha}$ = apparent center of curvature
l' = image distance of A_1.

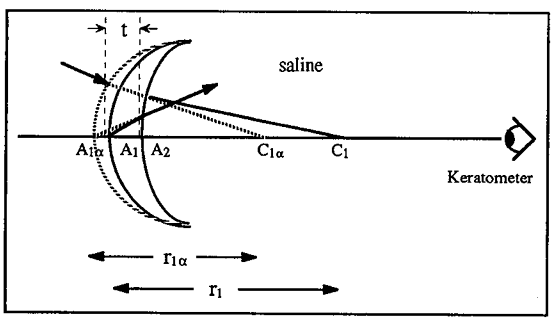

FIGURE 10.
Due to refraction by the lens back surface (A_2), the radius of curvature (r_1) of the lens front surface appears to be slightly steeper ($r_{1\alpha}$) to an observer using the keratometer. C_1 and $C_{1\alpha}$ are, respectively, the real and apparent positions of the center of curvature of the lens front surface.

APPENDIX B
Calculation of real BOZRE from apparent BOZRE

In Fig. 11, the keratometer gives direct measurement of the radius of curvature of the front surface (A_1) of the contact lens. However, measurement of the back surface (A_2) through the front surface does not give a true value of the radius of curvature of the back surface. This is because the light rays from the keratometer mire targets undergo refraction at the front surface (A_1) before being reflected by the back surface (A_2), and then undergo a second refraction at the front surface before reaching the keratometer.

We need to calculate the image positions A'_2 and C'_2 as seen through the appropriate surfaces, taking r'_1, r'_2, t as being +ve numbers.

Image distance for A'_2 is given by:

$$\frac{n'}{l'} - \frac{n}{l} = (n' - n)\frac{1}{r} \Leftrightarrow \frac{1.0}{l'} - \frac{n}{-t} = (1 - n)\frac{1}{-r'_1} = F_1 \Leftrightarrow$$

$$l' = \frac{t}{F'_1 t - n}$$

Also, for image of C_2:

$$\text{1st surface } (r'_2) \quad \frac{n'}{l'} - \frac{1}{-r'_2} = (n - 1)\frac{1}{-r'_2} \Leftrightarrow$$

$$l' = -r'_2 \text{ (coincident object/image at } C_2)$$

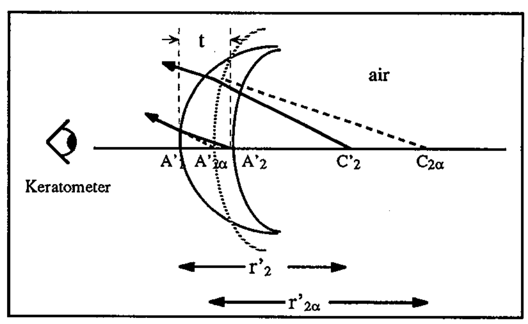

FIGURE 11.
Due to refraction by the lens front surface (A'_1), the position (A'_2) and the radius of curvature (r'_2) of the lens back surface, appear to be, respectively, $A_{2\alpha}$ and $r_{2\alpha}$, to an observer using a keratometer. C'_2 and $C_{2\alpha}$ are, respectively, the real and apparent positions of the center of curvature of the lens back surface.

2nd surface (r_1') $\frac{1}{l'} - \frac{n}{(-t - r_2')} = (1 - n)\frac{1}{-r_1'} = F_1' \Leftrightarrow$

$$l' = \frac{t + r_2'}{F_1'(t + r_2') - n}$$

i.e., apparent radius of curvature is:

$$A_{2\alpha}' - C_{2\alpha}' = \frac{t}{F_1' t - n} - \left[\frac{t + r_2'}{F_1'(t + r_2') - n}\right] = r_{2\alpha}' \quad (1)$$

$$r_{2\alpha}' = \left[\frac{1}{F_1' - n/t} - \frac{1}{F_1' - n/(t + r_2')}\right]$$

which is Holden and Zantos's[17] expression

$$\text{Setting } l' = \left(\frac{1}{F_1' - n/t}\right) - r_{2\alpha}' \quad (2)$$

From equations 1 and 2

$$l' = \frac{1}{F_1' - n/(t + r_2')} \Leftrightarrow r_2' = \left(\frac{+n}{F_1' - 1/l'} - t\right)$$

n = lens refractive index
t = center thickness
r_1' = real FOZRE (i.e., FOZR on eye)
F_1' = front surface power
r_2' = real BOZRE (i.e., BOZR on eye)
$r_{2\alpha}'$ = apparent BOZRE
C_2' = real center of curvature
$C_{2\alpha}'$ = apparent center of curvature
l' = image distance of A_2.

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Sotiris Plainis
Department of Optometry & Vision Sciences
UMIST P.O. Box 88
Manchester M60 1QD United Kingdom
email: S.Plainis@umist.ac.uk