Abstract: Purpose:
To develop a realistic model of the opto-mechanical behaviour of the cornea after curved relaxing incisions, and compare the astigmatism correction predicted by the model with that of the Lindstrom's nomogram.

Methods:
A three-dimensional finite element model of the anterior hemisphere of the ocular surface was generated, considering three parts: cornea, limbus and sclera. The corneal tissue was modeled as a quasi-incompressible, anisotropic hyperelastic constitutive behaviour strongly dependent on the physiological collagen fibril distribution. Similar models were used for limbus and sclera. With this model we simulated the effects of curved corneal incisions. The resulting geometry of the optical zone was analyzed and finite ray tracing performed to compute refractive power and high order aberrations (HOA).

Results:
The finite element simulation provides the local displacements of the corneal tissue, and from that we obtain the resulting elevation topographies of the surfaces. Results of finite ray tracing show a close agreement between the model and the Lindstrom's nomogram. However, paraxial computations would yield significantly
different results (undercorrection of astigmatism). In addition, arcuates induce important amounts of HOA, mainly coma, trefoil and quadrafoil.

Conclusions:
Finite element models together with finite ray tracing computations permit to perform realistic simulations of the biomechanical and optical changes induced by relaxing incisions. The close agreement found between model and nomogram supports the validity of this approach. This can be a powerful tool for planning the surgery or designing new incisional techniques, whereas commonly used paraxial formulas fail to predict the resulting astigmatism and HOA.
Opto-Mechanical Model of Arcuates for Astigmatism Correction. Low and High Order Aberrations

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ABSTRACT

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Keywords: arcuates, astigmatism correction, finite element model, ray tracing, high order aberrations.
INTRODUCTION

Relaxing incisions, such as arcuates, are often used to correct astigmatism due to the toric shape of the cornea. When incisions are peripheral, the incised meridian flattens (relaxes) losing refractive power. Due to the redistribution of stresses, the perpendicular meridian increases its curvature and hence its power. The planning of arcuates is based on nomograms, but these can hardly be used as a single worldwide reference, and nomograms may require local adaptation. Furthermore results may vary depending on the patient, so that custom treatments should be applied if possible. Instead of using nomograms, the planning of a true custom surgery would involve to acquire all possible data on the cornea (geometry, biomechanical and optical properties) and perform a computer simulation to predict and optimize the changes induced by a given treatment. Apart from reliable data from the patient, the computer-aided design of surgery would require a realistic model of the opto-mechanical behaviour of the cornea.

Geometrical optics models of the cornea have been used since long ago. Nowadays with the widespread use of corneal topographers it is possible to obtain accurate models of individual corneas, to perform numerical ray tracing and compute higher order aberrations (HOA). These models and computations can be then applied to optimize custom surgery. The shape of the cornea depends on the interacting stresses and intraocular pressure so that biomechanics plays an essential role in the optical performance, especially after surgery. Finite element-based biomechanical models of the eye have been presented only recently as a powerful tool for a better prediction of the effects of relaxing incisions and other refractive surgeries. The cornea presents a nonlinear hyperelastic, incompressible and anisotropic constitutive behavior strongly dependent on the physiological collagen fibril distribution. However, hyperelastic anisotropic models were proposed only recently.
In the present study we put together biomechanical\textsuperscript{16, 19} and optical\textsuperscript{4} models of the cornea to simulate arcuates and to predict the resulting optical quality. As a first step, here we have performed theoretical computations on a simple generic model of the cornea, to simulate the effect of arcuates of the Lindstrom’s nomogram\textsuperscript{1}. There is a good agreement between the astigmatism predicted by the model and that of the nomogram for all cases. In addition, our finite element and ray tracing analysis suggests that arcuates would produce a strong increase of HOA. The overall increase of HOA would be comparable to that induced by corneal surgery\textsuperscript{20, 21, 22}, but instead of spherical aberration, arcuates would mainly increase coma, trefoil and quadrafoil.

\textbf{MATERIAL AND METHODS}

\textit{Finite Element Model}

For this study a 3-dimensional model of the anterior hemisphere of the eye was generated (see Figure 1). As other authors in previous studies, we have made the assumption with rotational symmetry in order to simplify the model. A further simplifying assumption was to consider external surfaces spherical, for both cornea and sclera. Therefore, in this simulation, arcuates will induce (instead of correct) astigmatism. To simulate a real surgery to correct different existing amounts of astigmatism, using individual corneal topography maps of patients, would strongly complicate the finite element modelling and is beyond the scope of this study. The assumed geometry for the cornea is anterior radius 7.5 mm\textsuperscript{23}, diameter 12 mm, axial thickness 550 \(\mu\)m and edge thickness 650 \(\mu\)m at the limbus. As a result, the posterior surface is a conicoid with apical radius 6.04 mm and conic constant \(Q = -1.35\). The limbus is a narrow ring with internal and external diameters of 12 to 12.5 mm respectively, and increasing thickness from 650 \(\mu\)m to 1 mm. The sclera is considered spherical with external radius 13 mm and 1 mm thickness. Intraocular pressure was assumed 15 mm Hg\textsuperscript{12, 24}.

\texttt{########## Insert Fig.1 about here ##########}
The living human cornea is highly porous and approximately 80% of the weight of the cornea is water. The stroma forms about 90% of the thickness and is divided into 300-500 sheets of collagen lamellae, distributed in parallel to the corneal surface. Lamellae are about 2-3 mm wide and 1.5-2.5 μm thick, and appear to run uninterruptedly from limbus to limbus. Collagen fibrils lie parallel and run along the whole length of the lamella. In the central region of the cornea, fibrils are disposed predominantly along the superior-inferior and nasal-temporal directions, while near the limbus they follow the circumferential direction to form a pseudo-annulus. In addition, there are other randomly oriented fibrils throughout the cornea. This structure implies an anisotropic behaviour\(^{14,15}\). Our anisotropic model considers two fibril directions, one family along the nasal-temporal direction and another along the superior-inferior direction. The mathematical formulation and implementation details of the finite element model has been published before\(^{17,19}\). Therefore, the main characteristics of the constitutive model of the corneal material are: incompressible (due to the high water content), fibred anisotropic (lamellae), hyperelastic (non-linear behaviour). We apply Holzapfel’s constitutive model\(^{25}\), initially developed to model arterial tissue, to model the corneal behaviour.

\[
\Psi_{iso} = \Psi_{iso}^{matrix} + \Psi_{iso}^{fibrils} = \frac{C_1}{2} (\bar{I}_1 - 3) + \frac{C_2}{2} (\bar{I}_2 - 3) + \frac{k_1}{2k_2} \{\exp[k_2 (\bar{I}_4 - 1)^2] - 1\} + \frac{k_3}{2k_4} \{\exp[k_4 (\bar{I}_6 - 1)^2] - 1\}
\]

The material constants for cornea, limbus and sclera used in this work are shown in Table 1.
Before carrying out the simulations, we have to include real physiological conditions under intraocular pressure (IOP). The stress-free configuration of the cornea is not known \textit{a priori}, but we only know the one already deformed by the effect of the IOP\textsuperscript{16}. An iterative process was conducted to incorporate into the model the initial strains, which is repeated until the displacements (deformation) in the optical zone are negligible. This iterative method guarantees that the geometry of the final configuration under IOP is close to the initial (stress-free) one for the optical zone.

The surgery simulations always depart from that resulting geometry and consider curved incisions, i.e. cuts performed in the clear cornea, about 90% of the thickness deep and perpendicular to the surface. The parameters of the incisions are optical zone diameter, length of arc (in degrees) and number of incisions (one or two). They will depend on the diopters of astigmatism to be corrected, but in all cases they are considered on the horizontal meridian. Figure 2 shows the cases of one and two arcuates for 90° cut length. In order to reproduce the incisions with the established parameters such as optical zone, length, depth and shape, the meshing of the model was designed placing nodes and surfaces between elements exactly where the incision had to be performed. The resulting mesh is composed of 12,928 hexaedral elements and 27,604 nodes, and was created following a radial pattern (16 segments of 5.625°). The nodes belonging to these incision surfaces were initially duplicated but linked by a rough contact. The IOP was then applied to the model and the contact between nodes placed at the surfaces where arcuates had to be made was then removed. The incisions of Lindstrom’s nomogram were simulated for constant optical zone of 6 mm and 90% depth.

\texttt{Insert Fig.2 about here}

\textbf{Optical model}

The finite element simulation provides the displacements of the nodes and from that we obtain the resulting elevation topography. For optical computations we only consider the optical zone of 6 mm
diameter. To build the optical model, we first fit the resulting corneal topography (for both anterior and posterior surfaces) to a parametric model. The mathematical formulation, implementation and application to obtain the optical properties of the average cornea were published in Ref. 4. Basically, the corneal topography is represented by a general non-revolution ellipsoid plus a Zernike polynomial expansion. The melon-shaped ellipsoid is the simplest mathematical way to include two essential properties of the cornea: toricity which is the main cause of astigmatism and conicity which has a major role in spherical aberration. The Zernike polynomial expansion (here we consider up to 7th order) permits us to represent irregularities and departures from that overall ellipsoidal shape. This is especially relevant in this study, since arcuates are expected to induce characteristic deformations strongly dependent on the symmetry (one versus two) and length of cuts. The corresponding software was implemented in Matlab (The Mathworks Inc, Natick, Mass.) This model fit is used as the input for the ray tracing software, but it also provides data for a detailed geometrical analysis, including apical curvature radii and conic constants along the horizontal and vertical meridians, \( R_x \), \( R_y \) and \( Q_x \), \( Q_y \) respectively, as well as Zernike coefficients. Significantly high values of Zernike coefficients represent deformation modes present in the postsurgical cornea, which will cause some amounts of high order aberrations (HOA) with similar symmetries. Finally the ray tracing to compute astigmatism and HOA was performed using the optical design software ZEMAX® (ZEMAX Development Corporation, Bellevue, WA). Further software was necessary (implemented in C++ language) to pass from the ZEMAX to the ANSI standards to report wave aberration data.

RESULTS

The results, corresponding to curved incisions of the Lindstrom’s nomogram for 6 mm optical zone are summarized in Table 2. This includes the cases of one or two mirror-symmetric cuts, and for each case three lengths along the 6 mm diameter circumference in degrees: 45°, 60° and 90°. All the
data in Table 2 and figures have been computed for that 6 mm optical zone. Figure 3 shows the resulting elevation topographies with respect to the initial spherical surface; i.e. they represent the displacements of the points of the corneal surface along the optical axis (or deformations) induced by the simulated surgery. The left panel displays the complete corneal topography (12 mm diameter) for the case of two cuts of 60° length and the right panels correspond to the optical zone (6 mm diameter) only for the 6 cases simulated. The colour scale corresponds to the 90° cuts, with a maximum displacement of about 60 micrometers which occurs at the incisions. These maximum displacements are lower 35 μm and 50 μm, for 45° and 60° cuts respectively, regardless of having one or two incisions.

The upper plot in Figure 4 compares the curvature radii along the horizontal $R_x$ and vertical $R_y$ meridians of the anterior surface induced by the relaxing incisions. As expected, the difference (toricity) increases both with the length and number of cuts. The optical effect of this difference is astigmatism (see Figure 5). The lower panel shows the resulting conic constants $Q_x$ and $Q_y$. Since we have assumed a spherical model for the presurgical corneal, the initial values are zero. Arcuates seem to modify conic constants in the optical zone towards more negative values, which means that the overall shape is a non-rationally symmetric ellipsoid. The ellipsoid becomes more prolate as the number and length of incisions increases, and this will have a strong effect on the spherical aberration. The dotted line represents the theoretical optimal value of the conic constant $Q = -0.528$ providing a minimum values of spherical aberration. Consequently, we expect that arcuates might decrease the spherical aberration of the cornea or even overcorrect it for higher amounts of
astigmatism. Other deformation modes that we can observe in Figure 3 are represented by Zernike coefficients in our surface model, but in this case it is more interesting to analyze their effect on the resulting optical quality. Equivalent changes and similar deformations occur in the posterior surface of the cornea. To avoid redundancy we do not include these data explicitly, but the optical model includes both anterior and posterior surfaces. The posterior surface of the cornea has a minor effect on corneal optical performance (around 10%), but we can not ignore it.

In Figure 5, the astigmatism predicted by the model is plotted against the nominal value taken from the Lindstrom’s nomogram for the different cases simulated. The thick straight line represents nomogram versus nomogram (x = y) as a nominal reference. For each case we consider two different values of astigmatism, paraxial (open circles) and effective for a 6 mm pupil (close squares) equal to the optical zone. In the presence of HOA (see Figures 6 and 7) the effective astigmatism depends on the pupil diameter. Therefore, these two values represent the two extreme cases: Paraxial astigmatism would correspond to a small pupil (close to 0 mm diameter), and the 6 mm case is the effective astigmatism for a pupil diameter equal to that of the optical zone. For intermediate pupil diameters astigmatism will lie between the two lines in Fig. 5. There is a close agreement between simulations and nomogram for the effective astigmatism for a 6 mm pupil, whereas the paraxial values appear to be somewhat undercorrected. Nevertheless, the paraxial approximation is too crude for postsurgical corneas so that the predicted values are unrealistic. Common pupil sizes, 3-4 mm, would have values below but close to the 6 mm case.
The resulting high order aberrations are given in Figures 6, 7, and Table 2 (right columns). Figure 6 shows the wavefronts, plotted as interferogram fringes for the 6 cases analyzed. From left to right: 45°, 60° and 90° cuts; 1 and 2 cuts correspond to the upper and lower interferograms respectively. The number of fringes is high, suggesting high values HOA, with peak-to-valley values of several wavelengths. The 1 cut upper interferograms display a lack of symmetry and a tendency to form triangular patterns suggesting the dominance of coma and trefoil aberrations. The lower interferograms (two cuts) are much more symmetric and show four lobes characteristic of quadrafoil aberration. Here coma and trefoil seem to have lower values. The total amount of HOA does not seem to differ much between the 6 cases analyzed, but there is an important difference between one cut and two cuts. With two cuts (lower interferograms) the optical quality in the central zone (3-4 mm diameter) seems quite good, with a low number of fringes (most fringes tend to concentrate peripherally). Furthermore, this seems to improve with the length of cuts, so that for 90° there is a wider almost aberration-free central zone. Numerical values of the most significant aberration modes are plotted in Figure 7, as well as the total RMS (root mean square) value (black bars). The pre-surgical cornea (no incisions) is a sphere and hence it shows spherical aberration (white bars) only. This is confirmed by the fact that the overall RMS is equal to the spherical aberration. Negative coma and trefoil appear after a single incision, as well as a quadrafoil. For one cut, these aberrations tend to increase with cut length. For two incisions, however, we can see a clear dominance of (negative) quadrafoil. The total RMS is lower than in the one cut case, and the RMS does not increase with length. The lowest amount of HOA corresponds to the longest cuts, as suggested by the lower right interferogram in Fig. 6. In conclusion, arcuates seem to induce important amounts of HOA. The overall RMS increases substantially and varies depending on the number and length of cuts, but this change is not monotonic with the dioptres of astigmatism corrected. The modes of aberrations change with the number of cuts. Coma and trefoil are dominant in the case of one cut, whereas quadrafoil is present in all cases, but is higher in two cuts.
Interestingly spherical aberration seems to decrease as expected from the resulting negative conic constants of the corneal surface.

### Insert Fig. 6 about here ###

### Insert Fig. 7 about here ###

**DISCUSSION**

So far, we have presented a finite element and finite ray tracing simulation of relaxing incisions surgery based on a realistic biomechanical model of corneal material. The corneal material model is incompressible (high water content), hyperelastic to account for the nonlinear behaviour and anisotropic considering the two preferred orthogonal orientations (horizontal and vertical) of the collagen fibres. The finite element simulation considers intraocular pressure but neglects other forces and tensions acting on the eye ball such as gravity or muscles which are little relevant for this study. A potentially more relevant simplifying assumption was to assume a spherical surface for the cornea. This has two consequences: on the one hand our simulation was inversed, that is we considered the case of generating instead of correcting astigmatism. On the other hand, the normal cornea is a prolate ellipsoid (negative conic constant) and this has an important effect on spherical aberration, which is better (lower) than that of a sphere. To evaluate the validity of these simplifying assumptions, we conducted a more realistic simulation considering an ellipsoid corneal surface, for the case of two incisions of 90°. The resulting astigmatism was 4.8 D paraxial and 6.3 D effective for 6 mm pupil, which is slightly lower than values obtained for the sphere (5 D and 6.44 D respectively). These small differences (0.2 D or lower) suggest that the predictions by the spherical model would not change drastically when considering more accurate models of the corneal geometry. Furthermore, larger differences are found in real clinical practice which might be due to large individual differences in corneal topography but also in biomechanical behaviour. The
close agreement found between the simulated effective astigmatism and that of the Lindstrom’s nomogram is the main proof of validity of our model. At this point we want to remark that, in general, models often have free variables in order to fit the data. No fit has been done here whatsoever (except for computing the geometrical parameters R, Q, etc. of the resulting topography after the finite element simulation). The hyperelastic biomechanical behaviour was adjusted once in a previous study to reproduce experimental biomechanical data of normal corneas. Therefore, the mechanical and optical simulations were totally blind without any feedback between them.

In conclusion the finite element simulations predict changes in the corneal topography with maximum axial displacements from 35 μm to 60 μm depending on the length of incisions. These displacements decrease as we move away from the cuts, which cause a deformation of the corneal surface. As expected, since the incisions are peripheral, the incised meridian flattens, whereas the perpendicular meridian increases its curvature. However, the deformations are more complex and this has a strong impact on the resulting optical quality. Other biomechanical consequences, relevant for planning the surgery, such as the post-surgical distributions of stress and strain, were already discussed in the previous work. The analysis of the resulting topographies reveals changes in toricity, as this was the goal of arcuates, but this surgery induces other deformations of the corneal surface. Interestingly, the optical zone becomes more prolate (more negative Q values). Depending on the symmetry (one versus two cuts) the deformation modes cause odd-symmetrical HOA (coma and trefoil) or even-symmetric HOA (quadrafoil). The fact that corneal incisions induce these types of aberrations, has been recently found for the case cataract surgery with 3.2 mm superior incisions. The overall post-surgical increase of HOA is important. The computed RMS wavefront error for the 6 mm optical zone was always above 1 μm (between 1.09 μm and 1.69 μm depending on the case). Compared to values of normal corneas, arcuates would increase corneal HOA by a factor between 2 and 3. This increase is similar to that found experimentally in other types of refractive surgery such as LASIK. Although the overall amount of induced HOA would
be similar in arcuates and LASIK, the nature of deformations induced is totally different and hence the aberration modes are different too. For instance, LASIK is known to increase spherical aberration and coma, whereas trefoil and quadrafoil seem to be associated to peripheral incisions.

The main conclusion is the close agreement found between the effective astigmatism predicted by the model and that of the Lindstrom’s nomogram. Interestingly, the paraxial estimation of astigmatism provides lower degrees of correction, due to the presence of HOA. This is a well-known effect, totally consistent with many previous studies, which leads us to another important conclusion: The paraxial approximation is too crude for computing refractive power in real eyes; in post-surgical eyes where HOA can be strongly increased it can give wrong predictions. Therefore realistic models and computations (finite element, ray tracing, etc.) are an essential tool to simulate and predict the outcomes of surgical treatments. Here we have reported an example of how the implementation of realistic physical models can be used as an alternative to nomograms of statistical nature. In addition to offer much richer information to plan the surgery, realistic models can be personalized to individual eyes\textsuperscript{28}, as far as specific data of the patient are available, and we believe that this deserves further study.

REFERENCES


**FIGURE CAPTIONS**

**Figure 1.** Finite element model of the anterior hemisphere of the eye ball, composed by cornea, limbus and sclera. The mesh is composed of 12,928 hexaedral elements and 27,604 nodes following a radial pattern.

**Figure 2.** Schematic diagram of 90° arcuates for a 6 mm optical zone: one (left) and two (right) incisions.

**Figure 3.** Elevation topographies resulting from the simulation of arcuates (with respect to the initial spherical surface). The left panel displays the complete corneal surface (12 mm diameter) for the case of two cuts of 60° length; the right panels correspond to the optical zone (6 mm diameter) for the 6 cases simulated. The colour scale corresponds to the 90° cuts.

**Figure 4.** Curvature radii, along the horizontal $R_x$ and vertical $R_y$ meridians (upper panel), and conic constants $Q_x$ and $Q_y$, of the anterior surface of the cornea for the different cases analyzed. The dotted line represents the optimum $Q$ value yielding the lowest spherical aberration.

**Figure 5.** Astigmatism predicted by the model versus nominal value from the Lindstrom’s nomogram for the different cases. Open circles represent paraxial estimations and close squares are the effective astigmatism for a 6 mm pupil. The thick straight line ($x = y$) represents the reference nomogram values;

**Figure 6.** HOA, plotted as interferogram fringes, for the 6 cases analyzed. From left to right: 45°, 60° and 90° cuts; the upper and lower interferograms correspond to 1 and 2 cuts respectively.

**Figure 7.** Resulting values of the most significant aberration modes: spherical aberration, coma, trefoil, quadrafoil and total RMS (black bars).
Opto-Mechanical Model of Arcuates for Astigmatism Correction. Low and High Order Aberrations

Synopsis

A finite element simulation of the opto-mechanical changes induced by arcuates is presented. It reproduces the nomogram with high accuracy and predicts an important increase of high order aberrations.
Figure 1

Cornea

Limbus

Sclera
Conic constants

Curvature radii

Figure 4
High Order Aberrations

No incision
45º x 1
60º x 1
90º x 1
45º x 2
60º x 2
90º x 2

Micrometers

Arcuates

Spherical aberration
Coma
Trefoil
Quadrafoil
RMS HOA

Figure 7
Table 1. Physical parameters and constants of the constitutive biomechanical models of cornea, limbus and sclera.

<table>
<thead>
<tr>
<th>Material</th>
<th>$C_1$ (MPa)</th>
<th>$D$ (MPa$^{-1}$)</th>
<th>$C_2$ (MPa)</th>
<th>$k_1$ (MPa)</th>
<th>$k_2$</th>
<th>$K_3$ (MPa)</th>
<th>$k_4$</th>
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</thead>
<tbody>
<tr>
<td>cornea</td>
<td>0.1</td>
<td>1e-5</td>
<td>0.0</td>
<td>0.234</td>
<td>29.917</td>
<td>0.234</td>
<td>29.917</td>
</tr>
<tr>
<td>Limbus</td>
<td>0.1</td>
<td>1e-5</td>
<td>0.0</td>
<td>0.234</td>
<td>29.917</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>sclera</td>
<td>35</td>
<td>1e-5</td>
<td>-32</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
</tbody>
</table>
Table 2. Summary of results. The first row (no incision) corresponds to the initial presurgical values. The different columns show geometrical data (R, Q); induced astigmatism (nominal, paraxial and effective) and HOA.

<table>
<thead>
<tr>
<th>Incisions</th>
<th>Rx (mm)</th>
<th>Ry (mm)</th>
<th>Qx</th>
<th>Qy</th>
<th>Astigmatism (nomogram)</th>
<th>Astigmatism (paraxial)</th>
<th>Astigmatism (6 mm zone)</th>
<th>Spherical aberr. (µm)</th>
<th>Coma (µm)</th>
<th>Trefoil (µm)</th>
<th>Quadrafoil (µm)</th>
<th>RMS HOA (µm)</th>
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</thead>
<tbody>
<tr>
<td>No incision</td>
<td>7.50</td>
<td>7.50</td>
<td>0</td>
<td>0</td>
<td>0 D</td>
<td>0 D</td>
<td>0 D</td>
<td>0.446</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0.446</td>
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<tr>
<td>45º arc x 1</td>
<td>7.59</td>
<td>7.4</td>
<td>-0.13</td>
<td>-0.096</td>
<td>1.5 D</td>
<td>1.3 D</td>
<td>1.43 D</td>
<td>0.37</td>
<td>-0.247</td>
<td>-0.81</td>
<td>-0.576</td>
<td>1.21</td>
</tr>
<tr>
<td>60º arc x 1</td>
<td>7.64</td>
<td>7.33</td>
<td>-0.18</td>
<td>-0.083</td>
<td>2.5 D</td>
<td>1.96 D</td>
<td>2.31 D</td>
<td>0.35</td>
<td>-0.451</td>
<td>-1.24</td>
<td>-0.766</td>
<td>1.66</td>
</tr>
<tr>
<td>90º arc x 1</td>
<td>7.76</td>
<td>7.32</td>
<td>-0.396</td>
<td>-0.129</td>
<td>3 D</td>
<td>2.65 D</td>
<td>3.27 D</td>
<td>0.251251</td>
<td>-0.73</td>
<td>-1.38</td>
<td>-0.474</td>
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</tr>
<tr>
<td>45º arc x 2</td>
<td>7.69</td>
<td>7.31</td>
<td>-0.25</td>
<td>-0.21</td>
<td>3 D</td>
<td>2.53 D</td>
<td>2.81 D</td>
<td>0.297</td>
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<td>0</td>
<td>-1.127</td>
<td>1.29</td>
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<tr>
<td>60º arc x 2</td>
<td>7.83</td>
<td>7.23</td>
<td>-0.41</td>
<td>-0.28</td>
<td>4.5 D</td>
<td>3.64 D</td>
<td>4.41 D</td>
<td>0.206</td>
<td>0.039</td>
<td>0.362</td>
<td>-1.421</td>
<td>1.55</td>
</tr>
<tr>
<td>90º arc x 2</td>
<td>8.03</td>
<td>7.16</td>
<td>-0.8</td>
<td>-0.29</td>
<td>6 D</td>
<td>5.01 D</td>
<td>6.44 D</td>
<td>0.070</td>
<td>0</td>
<td>0</td>
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