The Optic Nerve Head as a Biomechanical Structure: Initial Finite Element Modeling

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PURPOSE. To study the relationship between intraocular pressure (IOP) and the IOP-related stress (force/cross-sectional area) it generates within the load-bearing connective tissues of the optic nerve head.

METHODS. Thirteen digital, three-dimensional geometries were created representing the posterior scleral shell of 13 idealized human eyes. Each three-dimensional geometry was then discretized into a finite element model consisting of 900 constituent finite elements. In five models, the scleral canal was circular (diameters of 0.50, 1.50, 1.75, 2.00, and 2.56 mm), with scleral wall thickness (0.8 mm) and inner radius (12.0 mm) held constant. In three models, the canal was elliptical (vertical-to-horizontal ratios of 2:1 [2.50 × 1.25 mm], 1.5:1 [2.1 × 1.4 mm], and 1.15:1 [1.92 × 1.67 mm]), with the same constant scleral wall thickness and inner radius. In five additional models, scleral canal size was held constant (1.92 × 1.67 mm), and either scleral wall thickness (three models, 0.5, 1.0, and 1.5 mm) or inner radius (two models, 13.0 and 14.0 mm) was varied. In all models, each finite element was assigned a single isotropic material property, either scleral (modulus of elasticity, 5500 kPa) or axonal (modulus of elasticity, 55 kPa). Maximum stresses within specific regions were calculated at an IOP of 15 mm Hg (2000 Pa).

RESULTS. Larger scleral canal diameter, elongation of the canal, and thinning of the sclera increased IOP-related stress for a given level of IOP. For all models, maximum IOP-related stress ranged from 6 × IOP (posterior sclera) to 122 × IOP (laminar trabeculae). For each model, maximum IOP-related stress was highest within the laminar trabecular region and decreased progressively through the laminar insertion, peripapillary scleral, and posterior scleral regions. Varying the inner radius had little effect on the maximum IOP-related stress within the scleral canal.

CONCLUSIONS. Initial finite element models show that IOP-related stress within the load-bearing connective tissues of the optic nerve head is substantial even at low levels of IOP. Although the data suggest that scleral canal size and shape and scleral thickness are principal determinants of the magnitude of IOP-related stress within the optic nerve head, models that incorporate physiologic scleral canal and laminar geometries, a more refined finite element model meshwork, and anisotropic material properties will be required to confirm these results. (Invest Ophthalmol Vis Sci. 2000;41:2991–3000)

The methodology for clinically determining the susceptibility of an optic nerve head (ONH) to a given level of 24-hour IOP does not exist. At present, not only is it impossible to assess 24-hour IOP exposure in a physiologic manner, but there is also no agreement on the relationship between the measured IOP and the propensity of an eye to develop vision-threatening damage.

We have begun to model the optic nerve head as a biomechanical structure. Our approach evolves from the hypothesis that even at normal levels of IOP, the connective tissues of the ONH are constantly exposed to substantial levels of IOP-related stress (force/cross-sectional area). The level of stress generated by normal levels of IOP is assumed to play a central role in the physiology and, in some eyes, the pathophysiology of all three ONH tissue types: connective (load-bearing connective tissues of the peripapillary sclera, scleral canal wall, and lamina cribrosa), axonal (retinal ganglion cell axons), and cellular (astrocytes, glial cells, endothelial cells, and pericytes along with their basement membranes).

This biomechanical model suggests that a given level of IOP-related stress may be physiologic or pathophysiologic depending on the individual ONH experiencing the stress. Physiologic levels of IOP-related stress are assumed to be capable of inducing a broad spectrum of acute and chronic changes in all three ONH tissue types that are central to normal ONH aging. Pathophysiologic levels of IOP-related stress are assumed to
induce pathologic changes in cell synthesis and tissue micro-architecture that underlie the two governing pathophysiolo-
gies in glaucomatous damage: mechanical failure of the load-
bearing connective tissues of the ONH, and progressive
damage to the adjacent axons (with eventual retinal ganglion
cell death) by a combination of both compressive and ischemic
mechanisms.

From a basic biomechanical standpoint, stress generates
strain (tissue deformation in response to load) within tissues
that experience load. The magnitude of strain is based on the
material properties of the tissues, including how well the
tissues are able to resist deformations induced by the applied
stress. Tissues are physically damaged (undergo mechanical
failure) in a predictable manner and pattern as the level of
strain exceeds the tissues’ elastic limits.

We hypothesize that, within the ONH, the connective
tissues of the peripapillary sclera, scleral canal wall, and lamina
cribrosa are damaged in a predictable pattern by IOP-related
stress and strain. The model suggests that it is the predictable
pattern of mechanical failure within the connective tissues that
underlies the clinical appearance of a glaucomatous ONH (pro-
gressive posterior bowing and excavation of the ONH tissues
beneath the anterior scleral canal opening). We further hypothe-
size that axons are damaged by both the direct and indirect
effects of IOP-related stress through a variety of mechanisms.
These mechanisms include not only the classic notion of phys-
cal compression (either external compression by intact lam-
nar beams1,2 or spontaneous compression, in which differ-
ces in tissue pressure across the lamina cause axons to
collapse spontaneously)3 but also both acute and chronic ischemia,4,5 which may be induced by the effects of IOP-related
stress and strain on blood flow and diffusion within the con-
nective tissues through which the ONH blood supply must pass.6

The model thus suggests that estimating the maximum
level of IOP-related stress for a given patient’s ONH (for a given
level of IOP) will become an important step in the clinical
estimation of the susceptibility of an individual ONH to glau-
comatous damage at that level of IOP. As a first step toward
building the clinical methodology to make this estimation, we
have begun to use analytical and computational models (finite
element models [FEMs]) to study the relationship between IOP
and IOP-related stress within the load-bearing connective
tissues of the ONH.

Analytical models based on linear elasticity theory are
useful for idealized investigations of the eye as a mechanical
structure. However, because idealized models cannot accu-
rately account for the complex geometry, material property
distributions, and mechanical loading in the eye, approximate
computational methods are needed.

The finite element method is a computer-based engineer-
ning method that has been used to study the mechanical behav-
or of structures with complex geometries and material prop-
erties.7 (Fig 1). An FEM includes the important aspects of
three-dimensional (3D) geometry, material properties, bound-
ary conditions, and mechanical loads. These models are based
on a digital 3D geometry that approximates the structure being
modeled, which is discretized into small, regularly shaped,
“building blocks” (the finite elements) whose boundaries con-
nect points within the geometry (called nodes). Each finite
element of the model is assigned its own shape and material
properties (the characteristics of its connective tissue compo-
sition that determine the element’s behavior under load). The
mechanical behavior of the total structure is then calculated
from the combined behavior of its constituent finite elements.

The principal goal of building an FEM is to determine the
mechanical response (e.g., deformation, constitutive stresses
and strains) of a complex structure subjected to a variety
of loading conditions. Although the finite element method is an
approximate method, the degree of the approximation can be
controlled, and this approach often represents the only prac-
tical solution method for complex structures.7

The modeling of a complex biologic structure is best
performed by starting with simple models that capture the
most important aspects of its structure. The model is then
refined by adding features such as nonlinear material proper-
ties, more physiologically accurate geometries, and secondary
loading conditions, to more accurately capture the structure’s
responses to load.

In this report we present data for the maximum levels of
IOP-related stress within four regions of the posterior scleral
shell for 13 different FEMs, which, taken together, span the
published range of variation in the 3D anatomy of the human
scleral canal and posterior scleral shell.8-10

MATERIALS AND METHODS

Idealized Analytical Model

As a first step, simple analyses of the eye as an idealized
spherical shell were considered. Within the wall of any pres-
surized spherical shell, the two principal stresses reside within
the plane of the vessel wall (the third stress is radial in direc-
tion and minimal in magnitude; Fig 2). For a thin-walled vessel
(defined as having a wall thickness ≤ 0.1 of the radius), linear
elasticity theory predicts that the planar wall stresses are equal
and orthogonal and that each stress can be approximated by the
equation

$$\sigma = \frac{PR}{2t}$$  \hspace{1cm} (1)

where \( P \) is the inner pressure (IOP), \( R \) is the inner radius of the
sphere (approximately one half the axial length), and \( t \) is the
thickness of the vessel wall (scleral thickness).11

Estimating IOP-related stress within and around the load-
bearing connective tissues of the scleral canal using analytical
models is more complex. A simple closed-form solution of the
thin-walled spherical shell stress equation is no longer feasible,
because the geometry of the ONH and its load-bearing tissues
is too complex. However, a close analogy to this geometry for
which we can compute an analytical solution is the case of an
elliptical hole within a flat plate that is under tension at two
ends. In this idealized case, it can be shown that the maximum
stress always occurs at the edge of the hole along the long axis
of the ellipse, and has a value of

$$\sigma = S\left(1 + \frac{2a}{b}\right)$$  \hspace{1cm} (2)

where \( S \) is the stress applied to the ends of the plate, \( a \) is the
length of the long axis, and \( b \) is the length of the short axis.11
FEM Geometry

For these initial analyses, 13 models (Table 1 and Fig. 3) representing the posterior shell of 13 idealized human eyes were developed by computer with finite element pre-processing software (Truegrid; XYZ Scientific Applications, Livermore, CA). For eight of the FEMs (M1–M8, Table 1), a standard posterior scleral shell with an inner radius of 12.0 mm and wall thickness of 0.8 mm was constructed (Figs. 3A, 3B) and the size and shape of the scleral canal opening were varied (Fig. 4), using dimensions that span the range of values reported for the human scleral canal.8–10 In three additional models (M9, M10, and M11) the scleral canal geometry (1.92 × 1.67 mm) and inner radius (12.0 mm) of model M8 were used, with variable scleral wall thickness (M9, 0.5 mm; M10, 1.0 mm; M11, 1.5 mm). For the final two models (M12, M13), the scleral canal geometry (1.92 × 1.67 mm) and scleral thickness (0.8 mm) of model M8 were used, and the inner radius of the posterior scleral shell was varied (M12, 13.0; M13, 14.0 mm).

For these 13 initial models, the scleral shell consisted of two layers of finite elements (inner and outer layer elements,
of magnitude less to reflect the fact that the axons are likely to be compliant and unlikely to bear significant load.

**FEM Boundary and Operating Conditions**

To prevent rigid body motion of the model, the nodes along the base of the model (representing the transverse or anterior–posterior equator of the eye) were constrained to radial movement only, whereas all other nodes were allowed to move freely in any direction. All simulations were run at an IOP of 2.0 kPa (15 mm Hg) on a computer workstation (Origin; Silicon Graphics, Mountain View, CA) using finite element analysis software (Abaqus ver. 5.8; Hibbit, Karlsson, and Sorensen, Pawtucket, RI). Postprocessing analysis was performed using Abaqus/Post (Hibbit, Karlsson, and Sorensen).

**IOP-Related Stress Calculations by Integration Point and Region**

Within each finite element of each model, the FEM software characterized stress at 27 distinct integration points (Figs. 3D, 3E). We defined the stress magnitude at each integration point to be the magnitude of the vector sum of the three principal stresses acting at that point. For each model, therefore, stress magnitude was determined at a total of 24,300 points (900 elements × 27 integration points/element).

To characterize the maximum stress by region within each model, the finite elements of each model were subcategorized into four regions (Fig. 5). The laminar trabecular region (Fig. 5A) was defined to include all the connective tissue finite elements within the scleral canal. The laminar insertion region (Fig. 5B) was defined to include the single most peripheral element of each laminar beam as well as the single row of elements of the scleral shell immediately adjacent to the scleral canal. The peripapillary scleral region (Fig. 5C) was defined to include all finite elements in the second, third, and fourth rows outside the scleral canal (extending from approximately 1 mm to 3 mm away from the scleral canal). Finally, the posterior scleral region (not shown) was defined to include the remaining finite elements of the posterior scleral shell (extending from the fifth row of elements away from the canal to the elements of the transverse equator).

The stress magnitudes from all the integration points within each region were analyzed and a top 5% value for
maximum stress within a region was defined to be the mean of the highest 5% of stress magnitude values within the region and was intended to conservatively estimate the highest level of stress. Maximum stresses for the laminar insertion and the peripapillary scleral regions were also subcategorized in terms of superior–inferior and nasal–temporal regions (Fig. 5C).

**RESULTS**

When the standard, human posterior scleral shell geometry with an inner radius of 12.0 mm and a scleral wall thickness of 0.8 mm is used, equation 1 yields principal planar wall stresses (\(\sigma_1\) and \(\sigma_2\), Fig. 2) of approximately 7.5 \(\times\) IOP.

\[
\sigma_1 = \sigma_2 = \frac{PR}{2t} = \frac{\text{IOP(12)}}{2(0.8)} = 7.5 \times \text{IOP} \quad (3)
\]

Combining these two principal stresses gives a total scleral shell wall stress magnitude of 10.6 \(\times\) IOP, regardless of the level of IOP.

\[
\sigma_{\text{sclera}} = \sqrt{\sigma_1^2 + \sigma_2^2} = 10.6 \times \text{IOP} \quad (4)
\]

These analytical equations show that increases in the inner radius and decreases in scleral wall thickness further elevate IOP-related stress within the scleral wall at a given level of IOP.

The stresses (reported in kilopascals) for the posterior scleral shell, peripapillary sclera, laminar insertion zone, and laminar trabeculae (Table 2) are the FEM estimates of the maximum stress magnitudes present within these tissues when the IOP is 15 mm Hg (2.0 kPa). Table 3 summarizes the relationship of maximum IOP-related stress to IOP for each region.
Posterior Scleral Wall Stress

The maximum stress within the posterior scleral shell (away from the peripapillary sclera) for models M1 through M5 (circular scleral canals) and models M6, M7, and M8 (elliptical scleral canals) was approximately 11 times the level of IOP for each model (Tables 2 and 3).

In models M8 through M11, the radius (12.0 mm) and scleral canal geometry (1.92 mm$^3$ 1.67 mm) were held constant, and scleral wall thickness was varied between 0.5 and 1.5 mm. The data suggest that posterior scleral wall stress increased in eyes with thin sclera and decreased in eyes with thick sclera. When scleral wall thickness was held constant at 0.8 mm (models M8, M12, and M13), posterior scleral wall stress increased slightly as the size of the posterior scleral shell increased (from 11 × IOP for an inner radius of 12.0 mm to 13 × IOP for an inner radius of 14.0 mm).

Peripapillary Scleral Wall Stress

Unlike the posterior sclera away from the canal, IOP-related stress within the peripapillary sclera was profoundly influenced by the size and shape of the scleral canal (Table 2). Increasing the size of a circular canal increased peripapillary...
scleral stress from approximately $11 \times IOP$ (0.50 $\times$ 0.50 mm canal, model M1) to $20 \times IOP$ (2.56 $\times$ 2.56 mm canal, model M5; Table 3).

For two canals with similar cross-sectional areas (models M3 and M6, Table 1), peripapillary scleral stress was higher for the more elliptical canal (M6, $21 \times IOP$; M3, $16 \times IOP$; Table 3). Furthermore, within the peripapillary scleral region, maximum stresses near the vertical axis of all elliptical scleral canals were consistently greater than those found nasally and temporally. In the most elongated ellipse (model M6, canal dimension 2.50 $\times$ 1.25 mm), the maximum superior–inferior peripapillary scleral stresses were 49% higher than the nasal–temporal values (data not shown).

For the mean human canal dimensions (model M8), decreasing wall thickness from 1.5 mm to 0.5 mm increased peripapillary scleral stress from 8 to $27 \times IOP$ (models M11–M8, Table 3). However, increasing the size of the posterior shell from an inner radius of 12.0 to 14.0 mm (models M8, M12, and M13) only minimally increased peripapillary scleral stress.

**Laminar Insertion Stress**

Laminar insertion stress was profoundly influenced by the size of the scleral canal (Tables 2 and 3). Increasing the size of a circular canal increased maximum laminar insertion zone stress from approximately $20 \times IOP$ (model M1, 0.50 $\times$ 0.50 mm canal) to $47 \times IOP$ (model M5, 2.56 $\times$ 2.56 canal; Table 3).

### Table 2. Maximum Stress Values

<table>
<thead>
<tr>
<th>Model</th>
<th>Posterior Wall Stress</th>
<th>Maximum PPS Stress</th>
<th>Maximum Laminar Insertion Stress</th>
<th>Maximum Laminar Trabecular Stress</th>
</tr>
</thead>
<tbody>
<tr>
<td>Circular</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>M1 (0.50 $\times$ 0.50)</td>
<td>21.5</td>
<td>22.6</td>
<td>40.5</td>
<td>68.1</td>
</tr>
<tr>
<td>M2 (1.50 $\times$ 1.50)</td>
<td>21.5</td>
<td>29.7</td>
<td>61.0</td>
<td>107.2</td>
</tr>
<tr>
<td>M3 (1.75 $\times$ 1.75)</td>
<td>21.5</td>
<td>31.7</td>
<td>68.7</td>
<td>129.5</td>
</tr>
<tr>
<td>M4 (2.00 $\times$ 2.00)</td>
<td>21.5</td>
<td>33.8</td>
<td>76.7</td>
<td>154.1</td>
</tr>
<tr>
<td>M5 (2.56 $\times$ 2.56)</td>
<td>21.4</td>
<td>39.5</td>
<td>94.6</td>
<td>215.0</td>
</tr>
<tr>
<td>Elliptical</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>M6 (2.50 $\times$ 1.25)</td>
<td>21.5</td>
<td>41.1</td>
<td>79.4</td>
<td>144.4</td>
</tr>
<tr>
<td>M7 (2.10 $\times$ 1.40)</td>
<td>21.5</td>
<td>34.1</td>
<td>70.6</td>
<td>130.1</td>
</tr>
<tr>
<td>M8 (1.92 $\times$ 1.67)</td>
<td>21.5</td>
<td>32.3</td>
<td>70.4</td>
<td>133.7</td>
</tr>
<tr>
<td>Wall thickness</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>M9 (0.5)</td>
<td>34.4</td>
<td>53.2</td>
<td>126.5</td>
<td>243.8</td>
</tr>
<tr>
<td>M8 (0.8)</td>
<td>21.5</td>
<td>32.3</td>
<td>70.4</td>
<td>133.7</td>
</tr>
<tr>
<td>M10 (1.0)</td>
<td>17.2</td>
<td>25.4</td>
<td>53.1</td>
<td>99.2</td>
</tr>
<tr>
<td>M11 (1.5)</td>
<td>11.7</td>
<td>16.3</td>
<td>31.8</td>
<td>56.7</td>
</tr>
<tr>
<td>Inner radius</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>M8 (12.0)</td>
<td>21.5</td>
<td>32.3</td>
<td>70.4</td>
<td>133.7</td>
</tr>
<tr>
<td>M12 (13.0)</td>
<td>23.2</td>
<td>33.9</td>
<td>73.0</td>
<td>132.8</td>
</tr>
<tr>
<td>M13 (14.0)</td>
<td>25.0</td>
<td>35.6</td>
<td>75.7</td>
<td>132.8</td>
</tr>
</tbody>
</table>

All dimensions given in millimeters. Data are in kilopascals (15 mm Hg = 2.0 kPa) and are the mean of the highest 5% of stress magnitude values within each region. PPS, peripapillary sclera.

### Table 3. Summary of the Maximum Stress Levels Expressed as Multiples of IOP

<table>
<thead>
<tr>
<th>Model</th>
<th>Posterior Sclera</th>
<th>Peripapillary Sclera</th>
<th>Laminar Insertion Zone</th>
<th>Laminar Trabeculae</th>
</tr>
</thead>
<tbody>
<tr>
<td>Circular</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>M1 (0.50 $\times$ 0.50)</td>
<td>11</td>
<td>11</td>
<td>20</td>
<td>34</td>
</tr>
<tr>
<td>M2 (1.50 $\times$ 1.50)</td>
<td>11</td>
<td>15</td>
<td>31</td>
<td>54</td>
</tr>
<tr>
<td>M3 (1.75 $\times$ 1.75)</td>
<td>11</td>
<td>16</td>
<td>34</td>
<td>65</td>
</tr>
<tr>
<td>M4 (2.00 $\times$ 2.00)</td>
<td>11</td>
<td>17</td>
<td>38</td>
<td>77</td>
</tr>
<tr>
<td>M5 (2.56 $\times$ 2.56)</td>
<td>11</td>
<td>20</td>
<td>47</td>
<td>107</td>
</tr>
<tr>
<td>Elliptical</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>M6 (2.50 $\times$ 1.25)</td>
<td>11</td>
<td>21</td>
<td>40</td>
<td>72</td>
</tr>
<tr>
<td>M7 (2.10 $\times$ 1.40)</td>
<td>11</td>
<td>17</td>
<td>35</td>
<td>65</td>
</tr>
<tr>
<td>M8 (1.92 $\times$ 1.67)</td>
<td>11</td>
<td>16</td>
<td>35</td>
<td>67</td>
</tr>
<tr>
<td>Wall thickness</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>M9 (0.5)</td>
<td>17</td>
<td>27</td>
<td>63</td>
<td>122</td>
</tr>
<tr>
<td>M8 (0.8)</td>
<td>11</td>
<td>16</td>
<td>35</td>
<td>67</td>
</tr>
<tr>
<td>M10 (1.0)</td>
<td>9</td>
<td>13</td>
<td>27</td>
<td>50</td>
</tr>
<tr>
<td>M11 (1.5)</td>
<td>6</td>
<td>8</td>
<td>16</td>
<td>28</td>
</tr>
<tr>
<td>Inner radius</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>M8 (12.0)</td>
<td>11</td>
<td>16</td>
<td>35</td>
<td>67</td>
</tr>
<tr>
<td>M12 (13.0)</td>
<td>12</td>
<td>17</td>
<td>36</td>
<td>66</td>
</tr>
<tr>
<td>M13 (14.0)</td>
<td>13</td>
<td>18</td>
<td>38</td>
<td>66</td>
</tr>
</tbody>
</table>

All dimensions shown in millimeters. Data are the means of the highest 5% of stress magnitude values (from Table 2) expressed as multiples of IOP.
Scleral canal shape also influenced the level of laminar insertion zone stress. For two canals with similar cross-sectional areas (models M3 and M6, Table 1), the maximum stress within the laminar insertion zone was 40 IOP for the elliptical canal (model M6, Table 3) compared with 34 IOP for the circular canal (model M3, Table 3).

In an eye with mean human scleral canal dimensions (model M8, 1.92 × 1.67 mm), decreasing wall thickness from 1.5 mm to 0.5 mm (models M8–M11) increased laminar insertion stress from 16 to 63 IOP (Table 3). However, increasing the size of the posterior shell from an inner radius of 12.0 to 14.0 mm (models M8, M12, and M13) only minimally increased laminar insertion zone stress.

Laminar Stress
Stress within the laminar trabeculae was profoundly influenced by the size of the scleral canal. Maximum laminar stress increased from approximately 34 × IOP in a small circular canal (model M5, 0.50 × 0.50 mm canal) to 107 × IOP in a large one (model M5, 2.56 × 2.56 mm canal; Table 3).

However, unlike peripapillary scleral and laminar insertion zone stress, the maximum level of laminar stress was little influenced by the shape of the scleral canal (Fig. 6). For two canals with similar cross-sectional areas (models M3 and M6, Table 1), the maximum stress within the laminar trabecular region was 72 × IOP for the elliptical canal (model M6, Table 3) and 65 × IOP for the circular canal (model M3, Table 3).

For models with mean human scleral canal dimensions (1.92 × 1.67 mm), decreasing scleral wall (and laminar) thickness from 1.5 mm to 0.5 mm (models M11, M12, and M13) increased maximum laminar stress from 28 to 122 × IOP (Table 3). However, increasing the size of the posterior shell from an inner radius of 12.0 to 14.0 mm (models M8, M12, M13; Table 3) only minimally increased laminar stress.

**DISCUSSION**

In the present studies, we used an engineering technique called finite element modeling to estimate the stresses within 13 idealized human posterior scleral shell geometries at an IOP of 15 mm Hg (2.0 kPa). Our estimates of maximum IOP-related stress achieved levels of 27 to 122 × IOP within the tissues of the peripapillary scleral, laminar insertion, and laminar trabecular regions of these models. The results provide evidence in support of our hypothesis that the load-bearing tissues of the ONH are subject to substantial levels of IOP-related stress even at low levels of IOP.

To our knowledge, Greene was the first to suggest that stresses within the peripapillary sclera might be concentrated relative to the more peripheral posterior sclera because of the behavior of stress around any hole (the scleral canal) in a pressurized spherical shell (the posterior scleral shell). Greene used equation 2 to approximate the stresses near the scleral canal and showed that for a circular hole (a/b = 1), the stress concentration factor at the edge of the hole is 3.0. Using that equation alone to estimate the maximum level of stress around elliptical scleral canals of small and large aspect ratios suggests...
that the maximum stress would be approximately $35 \times IOP$ around an elliptical canal with a small aspect ratio (model M8, Fig. 4) and would reach $53 \times IOP$ around an elliptical canal with a large aspect ratio (model M6, Fig. 4).

Although the linear model of a hole in a thin plate provides basic insight into the mechanical conditions surrounding the scleral canal, it is unable to take into account important aspects of the complex 3D anatomy of the ONH—e.g., the curvature of the scleral wall, the variable 3D architecture of the connective tissues within the canal, and the nonhomogeneous material properties of the scleral and laminar extracellular matrix. In our initial attempt to explore these issues, 13 FEMs were used that were based on 13 idealized representations of the human posterior scleral shell.

Within the 13 FEMs of this report, we found that peripapillary scleral stress exhibited maximum levels that varied from $11 \times IOP$ to $20 \times IOP$ when a circular scleral canal was adjusted from $0.50 \times 0.50$ mm to $2.56 \times 2.56$ mm, was consistently highest near the superior and inferior poles of vertically elliptical scleral canals, and was additionally increased by elongation of an elliptical canal, increasing the inner radius of the sphere, and decreasing wall thickness. Maximum laminar insertion zone stresses were approximately twice the magnitude of peripapillary scleral stresses in each model and were also influenced by changes in canal geometry, inner radius, and wall thickness. Finally, laminar trabecular stresses were consistently higher than stresses in any other region of the model; were also affected by changes in canal geometry, inner radius, and wall thickness; and demonstrated maximum values as high as $122 \times IOP$ (model M9, Tables 2 and 3).

The magnitude of IOP-related stress within the connective tissues of the ONH has been only indirectly addressed in previous reports. Zeimer et al. discussed the implications of stress within the lamina cribrosa and its relation to the viscoelastic properties of the laminar trabeculae and scleral canal. Donggi and Zeqin recently published a mathematical model predicting the displacement of a thin circular plate representing an idealized lamina cribrosa. Yablonski and Asamoto have suggested how the interplay between IOP and radius of curvature of the lamina may affect stresses within the load-bearing tissues of the ONH. Yan et al. proposed a laminar model in which stresses (predominantly shearing stresses) are responsible for the observed posterior displacement of the lamina cribrosa at elevated IOP.

Our computational simulations strongly suggest that the level of IOP-related stress for a given level of IOP is principally determined by the 3D geometry of the load-bearing tissues of the ONH. Specifically, ONHs that have large and/or elliptical scleral canals or that are bounded by thin peripapillary and posterior sclera should have higher levels of IOP-related stress for the same level of IOP.

The literature regarding scleral canal size as a risk factor for glaucomatous damage is inconclusive. Several researchers have reported that glaucomatous eyes measured clinically and after death do not have larger disc diameters, and some have suggested that this implies that scleral canal size is not a risk factor for glaucomatous damage. Others have reported a larger disc area in blacks than in whites. Chi et al. have postulated that this may be one reason why blacks may have a higher susceptibility to glaucomatous damage; however, others have suggested that it is not clear that the ONH in blacks is, in fact, more susceptible to glaucomatous damage. Nesterov and Egorov have suggested that, on a theoretical basis, disc size should be one determinant for susceptibility, and Cahu and Bartov have explored the theoretical implications of both axial length and scleral thickness.

To our knowledge, apart from scleral canal size, scleral canal shape has not previously been considered as a risk factor for glaucomatous damage. Quigley and Addicks have characterized the difference in laminar beam and axonal fenestration dimensions within the superior-inferior versus the nasal-temporal peripheral scleral canal. Both groups attributed early superior and inferior axonal loss in glaucoma to the relative lack of connective tissue support in these regions. The idealized geometries in our report do not address these differences in 3D laminar anatomy; however, it is hoped that ongoing studies in our laboratories involving geometries derived from 3D reconstructions of serial histologic sections eventually will do so.

Although the current literature has not clearly shown scleral canal size and shape to be risk factors for glaucomatous damage, we believe that conclusions should not be drawn from the group of studies that compare the size of the optic disc in glaucomatous versus nonglaucomatous eyes. Our data do not suggest that glaucomatous eyes should necessarily have larger scleral canals than nonglaucomatous eyes. Rather, our data suggest that among groups of eyes with similar age, severity of disease, and levels of 24-hour IOP exposure, but differing rates of onset and/or progression of glaucomatous damage, those groups with greater rates of damage may demonstrate differences in the size and/or shape of the scleral canal and the thickness of the posterior sclera.

Chisholm et al. reported differences in eye wall stress calculations (but did not study scleral canal size and shape differences) between hypertensive glaucoma suspects in whom disease did or did not progress to glaucomatous damage. However in their study, 24-hour IOP exposure was not longitudinally assessed.

The accuracy of the predictions of an FEM depends on the accuracy of its 3D geometry, the accuracy of its material properties, and the refinement of its element mesh (i.e., the relative size of the individual elements within the model). The FEMs in this report should be considered preliminary and will improve as digital geometries derived from 3D reconstructions of serial histologic sections (Fig. 1F) are implemented and discretized into larger numbers of elements and as the assignment of material properties becomes more realistic.

Within our 13 FEMs, the constituent elements of each model were grouped into posterior scleral, peripapillary scleral, laminar insertion, and laminar trabecular regions (Fig. 5). Although in this report we assigned the same scleral material strength to all connective tissue finite elements of all 13 models, future models will assign separate, nonisotropic material properties to elements within each of these regions, based on their profound extracellular matrix differences.

Because the geometry of these models does not adequately reflect the complexity of the lamina cribrosa, the location and magnitude of the model’s predictions for laminar stress may not be accurate. Within the human scleral canal, the connective tissues that make up the laminar sheets are densely fused into a thick connective tissue coat that surrounds the central retinal vessels. The laminar sheets then progressively...
attenuate into individual beams that are thinnest at their insertion into the scleral canal wall.25–26 Because of their thin cross-sectional area, we expect that stresses will be highest within these peripheral laminar beams. However, this basic aspect of the 3D geometry of the lamina cribrosa has not been incorporated into our initial models. In fact, within these models, the smallest beams are found within the central scleral canal, and the stress concentrations found there may thus be artifactual (Fig. 6).

The simple laminar geometry in these models may be responsible for two additional forms of error (Figs. 4 and 6). First, within the elliptical models, the horizontally aligned beams are slightly thinner than those that are vertically aligned, and the resultant differences in laminar beam cross-sectional area may artifactually elevate stresses along the nasal–temporal axis. Second, in all the models, there is slight asymmetry of the beams within the scleral canal that may not be physiologic and may alter the concentration of stress.

Finally, in these linear models, there is a direct relationship between IOP and the magnitude of IOP-related stress. Therefore, the absolute magnitudes of IOP-related stress for IOPs of 30, 45, and 60 mm Hg can be estimated by multiplying the values in Table 2 by 2 (IOP of 30 mm Hg), 3 (IOP of 45 mm Hg), and 4 (IOP of 60 mm Hg), respectively. These linear relationships are approximations and result from the fact that the material properties of these models were deliberately kept linear, elastic, and isotropic. Although we believe that this was a fair first approximation, biologic soft tissues by their nature are viscoelastic and nonisotropic14 and do not show this linear relationship.

The elastic and viscoelastic material properties of posterior sclera are currently under study.30 Future inclusion of these viscoelastic material properties along with their nonisotropic assignment will improve the accuracy of the models’ predictions and allow the study of risk factors associated with the distribution of connective tissues within the scleral canal that are separate from canal size and shape and scleral wall thickness.

## References


