High rate internal pressurization of the human eye to determine dynamic material properties

J. A. Bisplinghoff[1], C. McNally[1], I. P. Herring[2], F. T. Brozoski[3], S. M. Duma[1]
[1] Virginia Tech – Wake Forest University Center for Injury Biomechanics;
[2] Department of Small Animal Clinical Sciences
[3]United States Army Aeromedical Laboratory

ABSTRACT

Over 1.9 million people suffer from eye injuries in the United States each year, occurring mainly from automobile accidents, sports related impacts, and military combat. As a result of trauma, approximately 30,000 people become blind in one eye every year in the United States. A common injury prediction tool used for eye injuries is computational modeling, which requires accurate material properties to produce reliable results. The purpose of this study is to create a high rate pressurization system to analyze the rupture pressure of human eyes and to determine the dynamic material properties of human sclera. A high rate pressurization system was used to create a dynamic pressure event to the point of rupture in 12 human eyes. The internal pressure was dynamically induced into the eye with a drop tower while the rupture pressure was measured with a small pressure sensor inserted into the optic nerve. Measurements were also obtained for the diameter of the globe, the thickness, and the changing coordinates of the optical markers. A relationship between stress and Green-Lagrangian strain was determined for each test specimen in the x and y direction to show any directional effects. It was found that the average rupture stress was 12.30 ± 4.25 MPa, the average maximum Green-Lagrangian strain in the x-direction was 0.048 ± 0.02, and the average maximum Green-Lagrangian strain in the y-direction was 0.072 ± 0.02. It was also found that the average high rate rupture pressure for the human eye was 0.84 ± 0.13 MPa. In comparing these data with previous studies, it is concluded that the loading rate directly affects the rupture pressure and that the human eye is both anisotropic and viscoelastic.

INTRODUCTION

Over 1.9 million people suffer from eye injuries in the United States each year, occurring mainly from automobile accidents, sports related impacts, military combat and other accidents (McGwin, 2005; Kennedy, 2007). As a result of trauma, approximately 30,000 people become blind in one eye every year in the United States (Parver, 1986). Determining material properties will provide needed information for establishing injury criteria for the human eye. One of the most versatile ways to analyze these injuries is through computer modeling. Several ocular computer models have been created, but most have been designed to study corrective eye surgery and are only accurate for static solutions (Hanna, 1989; Sawusch, 1992; Wray, 1994; Bryant, 1996). A few models have been used for dynamic events, with the most up to date and accurate model being the Virginia Tech Eye Model (VTEM), created by Stitzel (Stitzel, 2002). Previously, uniaxial tensile strip tests were performed on the sclera and cornea for the first dynamic computer model presented by Kisielewicz and Uchio (Kisielewicz, 1998; Uchio, 1999). Because the human eye is an anisotropic viscoelastic material, the material properties must be determined at a higher rate to produce more accurate results. Uchio found peak rupture stress using static material properties to be 9.40 MPa, whereas the VTEM, with more realistic dynamic modeling, found peak rupture stress to be 23.00 MPa (Uchio, 1999; Stitzel, 2002). Unfortunately, all existing models lack accurate material properties of the ocular globe. Therefore the purpose of this study is to determine dynamic material properties for the human sclera.

METHODS

Material properties were determined by inducing a high rate internal pressure into the eye to measure rupture pressure, stress and strain. High rate pressurization was accomplished with a hydraulic system that utilized a drop tower to pressurize the human eye in a dynamic event (Figure 1). To initiate the event, a weight was dropped onto a piston which was inserted into the hydraulic cylinder. Preparation of the system included adding water through the cylinder to act as the medium for pressurization and to produce an approximate initial intraocular pressure of 0.002 MPa. Connecting the eye to the system was a 16-gage intravenous needle inserted into the optic nerve. In order to secure the optic nerve to the needle a medical suture was used while a cylindrical placement guide held the eye in place. To ensure that the optic nerve was sealed, it was covered with a flexible coupling material and then secured with a plastic fastener. Each human eye was stamped with 5 optical markers to provide a method for measuring strain (Figure 2). High speed video was taken at 10,000 frames/sec with a resolution of 512 x 512 to determine the strain and to confirm the location of the rupture. Internal eye pressure measurements for the rupture tests were collected at 30,000 Hz.

In order to acquire the internal eye pressure, an in situ pressure sensor was utilized. In this test series, the pressure was measured from a small pressure sensor made by Precision Measurement Company (Model 060, Ann Arbor, MI) that was inserted into the eye through the optic nerve. The pressure transducer was rated for a range of 0-3.45 MPa which was more than adequate for our expected pressure results and had a frequency response in excess of 10 kHz. Because the diameter was also changing with time due to the expansion, motion analysis software was used to measure the movement of the diameter during the test (TEMA, Image Systems, Sweden).

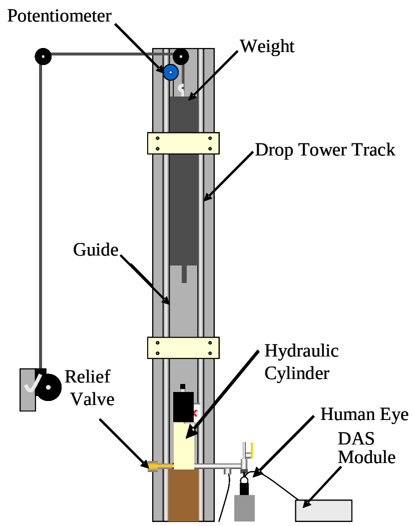

Figure 1: Schematic of the high rate pressurization system used to determine material properties.

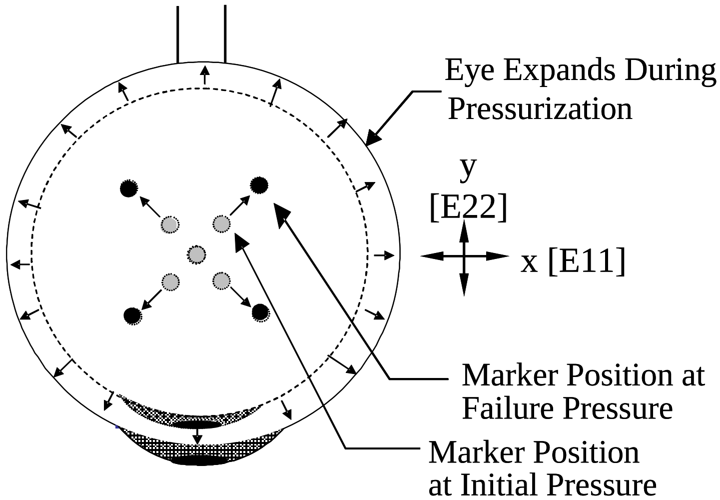

Figure 2: The expansion of a human eye shown with optical markers.

Each specimen was examined with a high accuracy laser to determine the thickness of the sclera after testing. The eyes were sectioned at the optical marker location and kept in saline solution during the test procedure. Measurements were taken with a Microtrak II high speed laser with accuracy of 2.5 micrometers. This method reduced the effect of swelling.

True stress was then calculated using the internal pressure (P), along with the radius (r) and the thickness (T) (Equation 1). Assuming that the eye is a spherical pressure vessel, Equation 1 was used to calculate the true stress. A relationship was derived to relate the radius and the thickness, where R_2 was the radius, R_1 was the original radius, and r_1 was the original thickness subtracted from the original radius (Equation 2). Assuming that the sclera is incompressible, the thickness for each change in the radius was found and used in Equation 1 to find the true stress.

$$\sigma_T = \frac{P(t) \cdot r(t)}{2 \cdot T(r)} \qquad \text{(eq.1)}$$

$$T = R_2 - \sqrt[3]{R_2^3 + r_1^3 - R_1^3} \qquad \text{(eq.2)}$$

Strain of the eye was determined from the high speed video of the events. Motion analysis software was used to track the location of each of the 5 optical markers at each point in time. These data were then analyzed using an original MATLAB code to calculate the Green-Lagrangian strain in both the x (E11) and y (E22) direction. Groups of two optical markers and the origin were created to start the calculation. The deformation gradient tensor (F) for each group was then determined by analyzing the position vectors (Equation 3), where x_i and y_i were the original positions of the markers and X_i and Y_i were the deformed positions of the markers. The Green-Lagrangian strain tensor was found using Equation 4. Using Green-Lagrangian strain reflects the non-linearity of the human eye. A relationship between true stress and Green-

Lagrangian strain was found for each direction using a characteristic average and the corresponding standard deviations. A statistical analysis was performed using a student T-test to determine the significance between the strain results for the x and y directions.

$$F = \left(\begin{bmatrix} X_1 & X_2 \\ Y_1 & Y_2 \end{bmatrix}_{t=1} \right) \left(\begin{bmatrix} x_1 & x_2 \\ y_1 & y_2 \end{bmatrix}_{t=0} \right)^{-1} \qquad \text{(eq.3)}$$

$$E = \begin{bmatrix} E_{11} & E_{12} \\ E_{21} & E_{22} \end{bmatrix} = \frac{1}{2}(F^T F - I) \qquad \text{(eq.4)}$$

In this study, 12 human eyes were procured and kept in a saline solution in glass jars and refrigerated for no longer than 15 days before they were tested, similar to previous studies. As stated in previous literature, the time after death prior to testing does not correlate to the degradation of the corneo-scleral shell with an R^2 value of 0.14 (Kennedy, 2004). A statistical analysis was done using a student T-test to determine the significance of the differences between the current study and previous rupture pressure studies.

RESULTS

The high rate pressurization of 12 human eyes resulted in a mean rupture pressure of 0.84 ± 0.13 MPa (Table 1). The time to rupture was measured and resulted in a mean of 30.78 ± 6.87 ms. The dynamic loading rate ranged from 13.15 MPa/s to 34.41 MPa/s with an average loading rate of 28.35 ± 6.60 MPa/s. The strain for the x-direction ranged from 0.022 to 0.80 and the y-direction ranged from 0.032 to 0.112. In comparison, the meridional direction showed a significant difference in strain when compared to the equatorial direction (p=0.02) (Figure 3). The true stress for the twelve tested eyes resulted in a range from 6.73 MPa to 21.37 MPa (Table 1). The stress-strain relationships determined using the above methods show that the general trends for both directions are similar but demonstrate significant directional effects relative to strain (p=0.02). The characteristic average for both directions and the corresponding standard deviations show the directional effects (Figure 3).

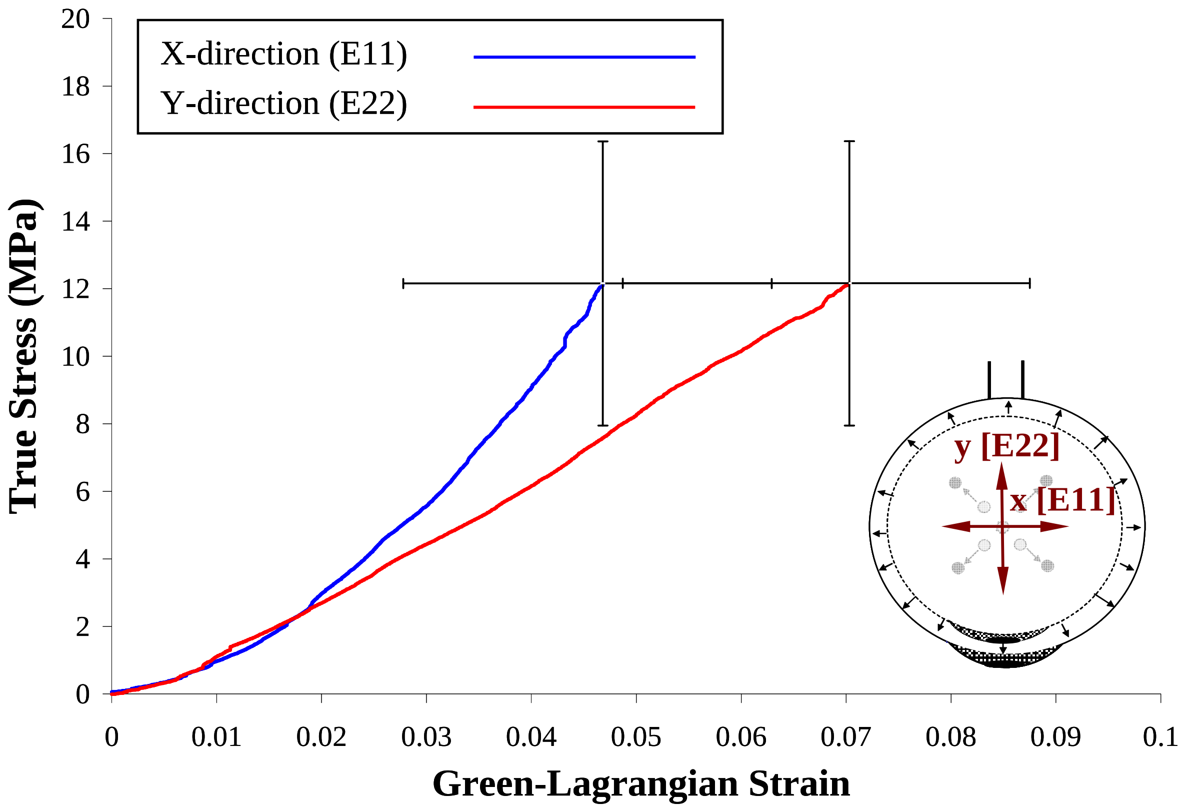

Figure 3: Stress-strain response for the x and y directions represented by a characteristic average and the corresponding standard deviations.

Table 1: Rupture pressure and stress-strain results for 12 human eye tests

	Rupture Pressure (MPa)	**Time to Rupture (ms)**	**Loading Rate (MPa/s)**	**Thickness (mm)**	**Maximum Stress (MPa)**	**Maximum Strain in x-direction**	**Maximum Strain in y-direction**	**Rupture Direction**	**Rupture Location**
HRH-1	0.70	21.23	32.78	0.46	11.07	0.033	0.071	equatorial	equator
HRH-2	0.80	23.80	33.66	0.56	11.94	0.034	0.047	equatorial	equator
HRH-3	0.71	26.13	27.13	0.81	6.73	0.032	0.080	equatorial	equator
HRH-4	0.78	28.07	27.72	0.59	10.46	0.038	0.112	equatorial	equator
HRH-5	0.91	28.67	31.72	0.47	17.53	0.037	0.098	equatorial	equator
HRH-6	0.96	28.77	33.48	0.51	15.03	0.063	0.070	meridional	equator
HRH-7	1.01	29.33	34.41	0.62	14.62	0.080	0.080	equatorial	equator
HRH-8	0.98	29.77	32.92	0.44	21.37	0.070	0.047	equatorial	equator
HRH-9	0.94	31.47	30.01	0.54	10.00	0.047	0.083	equatorial	equator
HRH-10	0.88	37.33	23.69	0.85	8.11	0.022	0.058	equatorial	equator
HRH-11	0.76	38.97	19.51	0.49	12.51	0.059	0.083	equatorial	equator
HRH-12	0.60	45.87	13.15	0.57	8.17	0.056	0.032	equatorial	equator
Average	0.84	30.78	28.35	0.58	12.30	0.048	0.072		
Standard Deviation	0.13	6.87	6.60	0.13	4.25	0.018	0.023		

DISCUSSION

The purpose of this study was to determine high rate material properties for the human sclera. Previous material property studies have been performed on the eye but none with the entire globe present (Battaglioli, 1984; Ahearne, 2007). Ahearne used an indentation technique to characterize the material properties for human and porcine corneas (Ahearne, 2007). Battaglioli tested the compressive properties of sclera after bisecting the equator and removing the lens, iris and choroid (Battaglioli, 1984). These material properties represent not only a section of the eye rather than the entire globe but also a static analysis rather than high rate.

For this study, the primary limitation in the stress calculation was the measurement of the thickness. Thickness measurements were taken using a laser but even with this precise instrument, variability is still present. With the use of true stress the thickness is more accurately represented. However, because the thickness varies across the eye it still can only be approximated. When determining the Green-Lagrangian strain of the human eye, the driving limitation was the accuracy with which the software tracked the optical markers. The change in the coordinates during a static event was divided by the total change in location throughout the dynamic event to find the percent error in the software. The average tracking error for the strain calculations was found to be 0.54%. Another consideration lies within the difference between a two dimensional strain calculation and a three dimensional calculation. The optical markers were created from a stamp to ensure that they were no more than 3.81 mm apart which produced an average error of 1.23% when using the two dimensional calculation. Although the previous ocular material property studies were conducted on sections of eyes rather than the entire globe, the strain results from the current study are consistent with previous results (Yamada, 1970; Battaglioli, 1984).

CONCLUSIONS

This study used a high rate pressurization system and motion analysis software to determine the dynamic material properties of the human sclera and the rupture pressure for the human eye. Measurements of internal eye pressure, thickness, diameter and optical marker location were found and used to calculate stress and Green-Lagrangian strain for each specimen. A relationship between stress and strain was found along with the rupture pressure for each of the 12 human eyes tested. Results show a significant difference in the maximum strain between the x and y directions ($p=0.02$), while still maintaining a similar trend. Stress, strain and rupture pressure values from the current study are consistent with previous trends, which concludes that the human eye is both anisotropic and viscoelastic. This study presents dynamic material properties of the human sclera and a mean globe rupture pressure that can be used for establishing injury criteria to help prevent eye injuries in the future.

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REFERENCES

AHEARNE, M., YANG, Y., THEN, K. Y., et al. (2007). An indentation technique to characterize the mechanical and viscoelastic properties of human and porcine corneas. Annals of Biomedical Engineering 35(9): 1608-1616.

BATTAGLIOLI, J. L. and KAMM, R. D. (1984). Measurements of the compressive properties of scleral tissue. Investigative Ophthalmology and Visual Science 25: 59-65.

BRYANT, M. R. and MCDONNELL, P. J. (1996). Constitutive laws for biomechanical modeling of refractive surgery. Journal of Biomechanical Engineering 118: 473-481.

HANNA, K. D., JOUVE, F. E., WARING, G. O., et al. (1989). Computer simulation of arcuate and radial incisions involving the corneoscleral limbus. Eye 3: 227-239.

KENNEDY, E. A., INZANA, J. A., MCNALLY, C., et al. (2007). Development and validation of a synthetic eye and orbit for estimating the potential for globe rupture due to specific impact conditions. Stapp Car Crash Journal 51: 381-400.

KENNEDY, E. A., VOORHIES, K. D., HERRING, I. P., et al. (2004). Prediction of severe eye injuries in automobile accidents: static and dynamic rupture pressure of the eye. Association for the Advancement of Automotive Medicine 48: 165-179.

KISIELEWICZ, L. T., KODAMA, N., OHNO, S., et al. (1998). Numerical prediction of airbag caused injuries on eyeballs after radial keratotomy. SAE International Congress and Exposition.

MCGWIN, G., XIE, A. and OWSLEY, C. (2005). Rate of eye injury in the united states. Archives of Ophthalmology 123: 970-976.

PARVER, L. M. (1986). Eye trauma: the neglected disorder. Archives of Ophthalmology 104: 1452-1453.

SAWUSCH, M. R. and MCDONNELL, P. J. (1992). Computer modeling of wound gape following radial keratotomy. Refractive and Corneal Surgery 8: 143-145.

STITZEL, J. D., DUMA, S. M., CORMIER, J. M., et al. (2002). A nonlinear finite element model of the eye with experimental validation for the prediction of globe rupture. Stapp Car Crash Journal 46: 81-102.

UCHIO, E., OHNO, S., KUDOH, J., et al. (1999). Simulation model of an eyeball based on finite element analysis on a supercomputer. British Journal of Ophthalmology 83: 1106-1111.

VOORHIES, K. D. (2003). Static and dynamic stress/strain properties for human and porcine eyes. Mechanical Engineering Master of Science: 100.

WRAY, W. O., BEST, E. D. and CHENG, L. Y. (1994). A mechanical model for radial keratotomy: toward a predictive capability. Journal of Biomechanical Engineering 116: 56-61.

YAMADA, H. and EVANS, F. G. (1970). Strength of biological materials. Baltimore, Williams & Wilkins.

AUTHOR LIST

1. Jill Bisplinghoff (primary author)
    Center for Injury Biomechanics, Virginia Tech
    100F Randolph Hall
    Blacksburg, VA 24061
    Phone: (540)-231-3414
    Fax: (540)-231-2953
    Bisplinj@vt.edu

2. Craig McNally
    Center for Injury Biomechanics, Virginia Tech
    100F Randolph Hall
    Blacksburg, VA 24061
    Phone: (540)-231-1617
    cmcnally@vt.edu

3. Ian Herring
    Department of Small Animal Clinical, Phase II
    Blacksburg, VA 24061
    Phone: (540)-231-3016
    iherring@vt.edu

4. Fred Brozoski
    US Army Aeromedical Laboratory
    Fort Rucker, AL
    Phone: (334) 255-6889
    Frederick.Brozoski@US.Army.mil

5. Stefan Duma
    Center for Injury Biomechanics, Virginia Tech
    100F Randolph Hall
    Blacksburg, VA 24061
    Phone: (540) 231-3945
    duma@vt.edu