Journal of Biomechanics 46 (2013) 1321–1327

Contents lists available at SciVerse ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Mechanism of traumatic retinal detachment in blunt impact: A finite element study

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ARTICLE INFO

Article history:
Accepted 5 February 2013

Keywords:
Retinal detachment
Blunt trauma
Eye model
Finite element analysis

ABSTRACT

Retinal detachment typically occurs when the retina is pulled away from its normal position by blunt trauma. It has been estimated that traumatic retinal detachments account for 10–20% of all detachments. Understanding the mechanism of traumatic retinal detachment is helpful for ophthalmologists to make a more accurate diagnosis before the symptoms develop. A finite element eye model, validated through published data, was used to simulate traumatic retinal detachment. Retinal adhesive force was incorporated into the model using breakable bonded contact. Under BB impact, global deformation was divided into four stages: compression, decompression, overshooting and oscillation. Shockwave propagation in the retina produced high strain in the retina. For an impact speed of 50 m/s, the peak strain of 0.138 in ora serrata exceeded the specified threshold for retinal break. When the eye was decompressed, negative pressure occurred around and anterior to the equator, with a minimum of −663 kPa, leading to retinal detachment. The following relative inertia motions between the retina and its supporting tissue extended the detachment. In addition, the simulations of lower shear modulus of vitreous and increased retinal adhesive force also confirm that the extent of retinal detachment is determined by negative pressure and inertial motion. In conclusion, shockwave and negative pressure contribute to retinal detachment. Shockwave propagation in the retina leads to retinal break, while negative pressure and relative inertial motion could pull the retina away from the supporting tissue. The current work would help understand the basic mechanisms underlying blunt trauma.

1. Introduction

The retina is a thin-layer of tissue which covers the inner surface of the human eye. Detachment occurs when the retina is pulled away from its normal position by blunt trauma (Kuhn and Piermici, 2002). Epidemiological studies conclude that the incidence of retinal detachment in a given population could be as much as 0.1 per thousand annually (Brinton and Wilkinson, 2009). Among them, traumatic detachments account for 10–20% (Brinton and Wilkinson, 2009). People engaging in sports such as boxing (Bianco et al., 2005), tennis (Nadeem, et al., 2007), diving (Xu et al., 2006) and bungee jumping (Chorich et al., 1998) are especially at high risk of traumatic retinal detachment. In addition, traumatic detachments are common in people impacted by airbags in vehicle crashes (Kuhn et al., 1995; Manche et al., 1997).

The most common type of retinal detachment secondary to blunt trauma is rhegmatogenous (Johnston, 1991; Sarrazin et al., 2004), whose mechanism will be addressed in this paper. Rhegmatogenous retinal detachment requires two necessary conditions: (1) retinal break and (2) liquid vitreous passing through the break into the subretinal space (Brinton and Wilkinson, 2009). Blunt ocular trauma can lead to both retinal break and vitreous liquefaction, which would increase the chance of subsequent retinal detachment (Kuhn and Piermici, 2002). Diagnosis of traumatic detachment is not always made effectively (Tasman, 1972). It has been reported few detachments can be diagnosed immediately after blunt trauma, while most of them remain undetected clinically until several months later (Cox et al., 1966; Tasman, 1972). Understanding the mechanism of traumatic retinal detachment will assist ophthalmologists in making a more accurate diagnosis before clinical symptoms develop.

Projectile impact experiment is considered as an effective method for studying blunt ocular trauma. Using different objects,

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http://dx.doi.org/10.1016/j.jbiomech.2013.02.006

impact experiments on human or animal eyes have been well studied (Delori et al., 1969; Scott et al., 2000; Pahk and Adelman, 2009; Sponsel et al., 2011). Most researchers use high-speed cameras to record the deformation of the globe. However, *in vivo* high-speed imaging for the interior ocular structures is difficult to obtain. As an alternative method, finite element simulation can provide visualization of the dynamic response of the interior structures. Previous studies have confirmed that finite element method is an effective tool to analyze various ocular injuries in different impact conditions (Uchio et al., 1999, 2003; Stitzel et al., 2002; Power et al., 2002; Gray et al., 2008; Weaver et al., 2011). The pioneering finite element eye model for dynamic analysis was created by Uchio et al. (1999), who studied ocular injuries caused by missile impact under different mechanical conditions. A more sophisticated eye model, Virginia Tech-Wake Forest University (VT-WFU) eye model, including liquid aqueous and vitreous was introduced by Stitzel et al. (2002). With 22 matched experiments, VT-WFU eye model is validated to predict globe rupture accurately. More recent simulation studies relate to traumatic intraocular injuries, including optic nerve damage (Cirovic et al., 2006), retinal detachment (Hans et al., 2009; Rangarajan et al., 2009) and damage (Rossi et al., 2011).

In the current study, a finite element model of the human eye was developed to simulate retinal detachment when impacted with a BB. Retinal adhesion was incorporated into the eye model to provide more realistic interactions between the retina and its support tissue. The simulation results would help understand the basic mechanism of traumatic retinal detachment.

2. Material and methods

2.1. Eye model

An idealized model of a human eye was created according to the VT-WFU eye model (Stitzel et al., 2002) using a three-dimensional CAD software package,

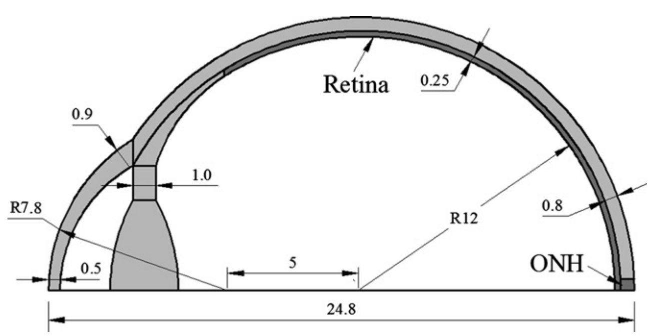

Fig. 1. The structure and dimension of the eye in cross section.

SolidWorks 2010 (Dassault systems SolidWorks Corp., SA). A retina layer with uniform thickness of 0.25 mm was modeled covering the inner surface of the sclera shell, and the structure of the choroid was ignored. The optic nerve head (ONH) was assumed to be positioned at the posterior pole of the globe. According to the literature (Sigal et al., 2010), an opening with a 1.8 mm radius was made in the sclera shell at the posterior pole. The optic nerve was passed through the scleral opening. Structure of the eye model in the cross section is shown in Fig. 1. A three-dimensional orbit model was reconstructed based on CT image sequences of a normal skull using Mimics 10.01 (Materialise NV, Belgian). The sequences were obtained from a 28 year old male subject in Beijing Tongren Hospital, Capital Medical University. Written consent was given from the subject for using his CT images for medical research purpose. The space between the eyeball and the orbital wall was filled with fat tissue. Extraocular muscles were excluded from the model because previous studies concluded that the muscles have little influence on the structural mechanical response when an eye is subjected to dynamic impact (Kennedy and Duma, 2008). The retina was assumed as a linear elastic material with Young modulus of 20 kPa (Jones et al., 1992). The same elastic material was assigned to the optic nerve. Vitreous and fatty tissues were both treated as viscoelastic materials (Lee et al., 1992; Schoemaker et al., 2006), whose mechanical properties are detailed in Supplement 1. The orbital bone with Young modulus of 14.5 GPa was also treated as linearly elastic (Robbins and Wood, 1969). The material properties of cornea, sclera, lens, ciliary body, zonules and aqueous humor were identical to those in the VT-WFU eye model (Duck, 1990; Czygan and Hartung, 1996; Uchio et al., 1999; Stitzel et al., 2002; Power et al., 2002). All material properties and mesh types in the eye model are listed in Table 1. The finite element model of the human eye is shown in Fig. 2. The cornea, sclera, lens, zonules, and ciliary were grouped into a "multibody part" to permit node-sharing. The boundary relationships between orbit and fat, optic nerve and retina or sclera, vitreous or aqueous and its neighboring tissues were defined with bonded contacts, obligating them to move together at their respective interface.

2.2. Retinal adhesive force

Contact between the retina and its supporting tissue (the sclera) was defined as a breakable bonded contact. This contact, originally used to simulate welding, was then used to simulate retinal adhesion. Initially, the retina and the sclera were connected with a bonded contact. When subjected to blunt impact, stress can be produced at the contact surface. According to a specified stress criterion, the contact may break and the adjacent structures separate from each other. The criterion was expressed as:

$$\left(\frac{\sigma_n}{\sigma_n^{\text{limit}}}\right)^n + \left(\frac{|\sigma_s|}{\sigma_s^{\text{limit}}}\right)^m \geq 1$$

where, σ_n and σ_s are the normal (only for tensile) and shear stress computed at the breakable contact. σ_n^{limit} and σ_s^{limit} are the specified normal and shear stress limits. The "n" and "m" are the normal and shear stress exponent coefficients, both of which were specified to be 1 in the model.

The normal stress limit σ_n^{limit} was defined as the retinal adhesive force which maintains retinal attachment to the supporting tissue. The shear stress limit was assumed to be equal to the normal. The retinal adhesive force per unit length has been measured by Kita et al. (1990). However, different from Kita's study, we aimed to obtain the force per unit area (pressure). By duplicating Kita's experiment, the retinal adhesive force per unit area was obtained in living rabbits. More details on the measurements and results can be found in Supplement 2. Calculated from 10 measurements, the retinal adhesive force was evaluated to be 340 ± 78 Pa.

Table 1
Mechanical properties and element types in the eye model.

Structures	Element type	Number of elements	Density (Kg m^{-3})	Material model	Material parameters
Cornea	Hexahedron	506	1076	Elastic	Nonlinear stress–strain
Sclera	Hexahedron	5224	1243	Elastic	Nonlinear stress–strain
Lens	Hexahedron	828	1078	Elastic	E=6.88 MPa
Zonules	Hexahedron	384	1000	Elastic	E=357.78 MPa
Ciliary	Hexahedron	1536	1600	Elastic	E=11 MPa
Retina	Hexahedron	2552	1100	Elastic	E=20 kPa
Aqueous	Hexahedron	840	1000	Liquid	Shock EOS linear C_1=1530 m/s, s_1=2.1057
Vitreous	Hexahedron	23,498	950	Viscoelastic	G_0=10 Pa, G_∞=0.3 Pa, β=14.26 1/s, K=2.0 GPa
Fat	Hexahedron	27,501	970	Viscoelastic	G_0=0.9 kPa, G_∞=0.5 kPa, β=50 1/s, K=2.2 GPa
Orbit	Tetrahedron	40,993	1610	Elastic	E=14.5 GPa

Where: E, Elastic modulus; K, Bulk modulus; G_0, Initial shear modulus; G_∞, Infinite shear modulus; β, Viscoelastic decay constant; C_1, Speed of sound through the material; s_1, the coefficient related to the speed of the shocked material.