Finite element analysis of cornea deformation and curvature change during the keratoplasty suturing process

Jong Won Baek¹ · Seong Jin Park¹

Received: 12 November 2018 / Revised: 16 January 2019 / Accepted: 18 January 2019 / Published online: 23 January 2019
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Abstract

Keratoplasty, which is cornea transplant surgery, is one of the treatment methods for patients with turbidity or keratitis. Recently, keratoplasty using a surgical robot was studied to increase precision. In this study, the effect of surgical factors on the deformation and curvature of the cornea were analyzed in order to improve the accuracy of keratoplasty and derive the optimal surgical factors using finite element method (FEM). Suturing tension and depth were selected as surgical factors. An FEM model, a constitutive equation, and boundary conditions were determined using experiments and reference data. Suturing tension significantly impacted deformation and curvature change, and suturing depth affected the position of the thread-cornea contact point. Both factors have a significant impact on a focal point in the retina and the patient's visual acuity after keratoplasty.

Keywords Cornea curvature · Keratoplasty · Finite element methods (FEM) · Suturing

1 Introduction

The cornea, located at the front of the eyeball with no blood vessels, is the most important part of the eye. Light rays enter through the cornea and are refracted many times. When the cornea becomes turbid, due to diseases such as keratitis, it can cause visual impairment. In these cases, keratoplasty is required for treatment. Figure 1 shows the overall keratoplasty procedure.

However, the results of the surgery were highly dependent on the surgeons [1, 2]. Therefore, robot surgery system for keratoplasty were studied to improve results, including the recovery of visual acuity. In addition, accuracy and prognosis after surgery were expected to improve using a robot system [3, 4].

Robotic surgery has many advantages: increased accuracy, high precision, decreased surgery time, and minimal incision length [4]. As surgical robot technology is rapidly evolving, research on high-precision surgery using a surgical robot is actively being conducted [5].

Many factors influenced the cornea's mechanical behavior, such as deformation and curvature changes in keratoplasty. Subsequently, the optical behavior, refractive index of the cornea, will change. Therefore, to increase the precision of keratoplasty, prediction of the cornea's mechanical behavior change for a robot surgery system is very important. The Finite Element Methods (FEM) can easily apply various boundary conditions and it is very accurate. Additionally, factor analysis can be easily conducted using an FEM simulation tool. FEM is one of numerical analysis methods. General procedure of FEM involves (1) discretizing of complex continuum into several elements, (2) Numerical simulations on each element, (3) Combining all set of simulation results from each element. The stress, deformation or some mechanical behavior can be approximately calculated from these procedure.

Many of studies have investigated the cornea's material properties [6–12]. Rowe et al. [13], Newton and Meek, [14], Boote et al. [15], examined rabbit, bovine, porcine, and human cornea material properties experimentally. Studies on the behavior of the cornea were also carried out. Anderson et al. [16] investigated cornea behavior after a force was applied. Studer et al. [17] attempted to determine how the apex (the center) was deformed using a bulge test. Recently, researches of cornea using finite element method have been performed. Anna Pandolfi et al. [18] simulated

✉ Seong Jin Park
sjpark87@postech.ac.kr

¹ Department of Mechanical Engineering, Pohang University of Science and Technology (POSTECH), Cheongam-ro, Nam-Gu, Pohang, Gyeongbuk 37673, Korea

Fig. 1 Overall process of the keratoplasty

apical displacement under 15 mm Hg intraocular pressure (IOP) using FEM. However, the study only introduced the relationship between IOP and deformation at the apex point.

Therefore, in this study, the effects of suturing tension and depth on the cornea's mechanical behavior were investigated using the proposed cornea standard model during the suturing process in keratoplasty. Furthermore, the changes in corneal deformation and curvature were analyzed in response to the change in both factors.

2 Cornea structure

FEM simulations used ABAQUS 6.13. In order to simulate the cornea's mechanical behavior, the cornea's standard model was implemented using Solidworks, which is a 3D modeling tool, as shown in Fig. 2 [18]. Cornea curvature is not uniform, and the central thickness is larger than the side thickness. The central and side thickness are 0.53 mm and 0.62 mm, respectively.

The suturing case cornea model was established based on the cornea standard model, as shown in Fig. 3. The computational cost of simulating the entire cornea structure was very high. Therefore, a quarter model was implemented in this study.

After the suturing process, a suture must be properly tightened to prevent a wound/incision open. Suture length influenced the corneal diopter and caused many defects such as astigmatism after the surgery [19]. Therefore, suturing tension and depth, which affect suture tightness and length, respectively were simulated and analyzed. The small hole and path were modeled together in the suturing case cornea model (Fig. 3). The corneal surfaces were idealized as being smooth and axisymmetric using an 8-noded solid element, C3D8, with distortion, hourglass mode control. Reduced integration was applied to decrease the computational cost. A total of 52,326 elements were generated with an average aspect ratio of 1.15 (Fig. 2). Hexahedral elements were automatically generated in cornea model without hole and path, but not in suturing case cornea model. The cornea model in the suturing case had a complicated geometry including a hole and a suturing path. In general, hexahedral elements can give more accurate results than tetragonal elements. Therefore, partitioning was used in this case as represented in Fig. 4. Partitioning is the method which divides the shape into several cells. The suturing case cornea model had 23,670 elements with an average aspect ratio of 1.15.

Fig. 2 Cornea structures (modeling for implementation in FEM simulation)

Fig. 3 Suturing case cornea model **Fig. 4** Meshed geometry **a** before partitioning and **b** after partitioning

3 Simulation conditions

An apex deformation simulation was conducted to validate the cornea standard model and the material properties. IOP was applied to the bottom of the cornea. One study, which simulated corneal tissue deformation, considered the bottom of the cornea to be fixed (no translation and rotation) [20, 21]. The cornea and sclera tissues were restricted by peripheral tissues, so that both could be considered independently. However, when the pressure was applied at the bottom of the cornea, the apex was deformed in the vertical direction (same direction as the pressure application). Both the rotational and the translational deformation occurred at the side of the cornea [16]. Therefore, a pin joint was implemented at the side of the cornea.

$$U_i = 0, \quad i = x, y, z \quad (1)$$

where U_i is the translational deformation.

Generally, IOP is maintained from 10 to 20 mm Hg [20, 22, 23]. When the pressure was applied below the cornea corresponding to a normal IOP range, the average deformed

Table 1 Material properties of the cornea and the thread [20, 26–29]

	Young's modulus (MPa)	Poisson's ratio
Cornea	20.33	0.49 (nearly incompressible)
Thread	1425	0.2

approximately 250 μm [24, 25] at Apex, where is the center point of the cornea. Based on experimental and simulation data of references, the stress–strain curve can be considered to be linear below 15 mmHg IOP. Thus, an elastic material can be used in this simulation. Suture thread was used as 10-0 nylon suture material. Table 1 shows the material properties of the cornea and the thread.

The contact boundary condition was applied between the cornea and the thread [30]. The suturing tension was set to 0.1 N, 0.3 N, and 0.5 N, and the suturing path depth was set to 0.35 mm, 0.45 mm, and 0.55 mm. IOP was set to 15 mmHg. Figure 5 shows the example of the simulation

Fig. 5 Simulation results of y-direction displacement

Fig. 6 Schematic diagram of the surgical factors

results of y-direction displacement with 0.3 N of tension and a 0.45 mm depth.

4 Factors that affected on the mechanical behavior

Factor analysis was conducted to investigate the cornea's mechanical behaviors (displacement, and curvature change) because both factors directly affect optical behavior of the cornea. The focal point was determined by Snell's law as below.

$$\frac{\sin\theta_1}{\sin\theta_2} = \frac{n_2}{n_1} \quad (2)$$

where θ_1 , and θ_2 are the incidence and refraction angle measured from the normal boundary, respectively, and n_1 , and n_2 are the refractive index of each materials. In this cases, n_1 , and n_2 are the air and cornea refractive indices, respectively. Incidence angle was varied by geometrical change and refraction angle then varied because the refractive indices were constant.

Suturing tension and depth were established as the factors that affected the mechanical behaviors. Figure 6 shows the schematic diagrams of the factors.

Fig. 7 Maximum deformation by contact (δ_c) of the cornea with respect to suturing depth

Figure 7 shows the maximum deformation of the cornea by contact (δ_c) with respect to suturing tension and depth, respectively.

The maximum values were obtained at the cornea-suture thread contact point shown in Fig. 6. As a results, deformations increased with high suturing tension. However, suturing depth did not have a significant impact on the results. Deformation slightly decreased as the suturing depth decreased. Although the constitutive equation of cornea and suture thread were elastic, the deformations exhibited nonlinearity (different gradient occurred).

The total deformation was defined as:

$$\delta_{total} = \delta_{IOP} + \delta_c \quad (3)$$

where δ_{IOP} and δ_c are the deformation due to the IOP and contact by tension between the thread and cornea. δ_{IOP} can be represented as:

$$\delta_{IOP} = L_{original} \frac{P}{E_2} \quad (\text{from the constitutive equation}) \quad (4)$$

where $L_{original}$ is the original length, P is the IOP, and E_2 is the Young's modulus of the cornea. The hole and thread had curved surfaces, and were geometrically cylindrical, as shown in Fig. 8.

Fig. 8 Schematic diagram of contact model between thread and cornea

The thread and hole were in contact, and the contact area and deformation were defined as:

$$b = \sqrt{\frac{4F \left[\frac{1-\nu_1^2}{E_1} + \frac{1-\nu_2^2}{E_2} \right]}{\pi L \left(\frac{1}{R_1} + \frac{1}{R_2} \right)}} \quad (5)$$

$$\delta_c = \frac{2F(1-\nu_2^2)}{\pi L E_2} \left(\frac{2}{3} + \ln \frac{4R_1}{b} + \ln \frac{4R_2}{b} \right) \quad (6)$$

where E_1 and E_2 are the Young's moduli, ν_1 , ν_2 are the Poisson's ratios of the thread and the hole, respectively. L is the contact length, F is the force, and R_1 and R_2 are the radius of the thread and the hole respectively. In this simulation, the thread and hole radii were assumed to be the same, because the thread was so thin that it could be neglected. The above equation could then be simplified as:

$$b = \sqrt{\frac{4F \left[\frac{1-\nu_1^2}{E_1} + \frac{1-\nu_2^2}{E_2} \right]}{\pi L \left(\frac{2}{R_1} \right)}} \quad (7)$$

$$\delta_c = \frac{2F(1-\nu_2^2)}{\pi L E_2} \left(\frac{2}{3} + \ln \left(\frac{4R_1}{b} \right)^2 \right) \quad (8)$$

The contact length L is the half perimeter of a circle with the radius representing the suturing depth (refer to Fig. 6), defined as:

$$L = \pi(SD) \quad (9)$$

Fig. 9 Schematic diagram of the model for curvature change

where SD is the suturing depth. From the above relationship, total deformation can be calculated as:

$$\delta_{total} = L_{original} \frac{P}{E_2} + \frac{2F(1-\nu^2)}{\pi L E_2} \left(\frac{2}{3} + \ln \left(\frac{4R_1}{b} \right)^2 \right) \quad (10)$$

The deformation that occurred by contact between the thread and the cornea was a function of a suturing tension and depth. The above equations were highly non-linear functions, so deformations were represented as non-linear behaviors.

In order to obtain the relationship between local curvature change and suturing tension, the suturing depth was set to 0.45 mm (the mid-value). Similarly, suturing tension was assumed to be 0.3 N to simulate the curvature change and suturing depth relationship. The boundary conditions were the same as in the previous simulation. The x-axis is the distance from the origin point, as shown in Fig. 9. Figure 10 shows the curvature change at each point on top of the cornea.

In Fig. 10, the local curvature change first gradually increased, then increased very sharply at the contact point between the thread and the hole. The contact force and stress were a maximum at this point. The total curvature was proportional to the suturing tension from Eq. 10, even if the IOP, P , was constant. Although the maximum deformation by contact (δ_c) was approximately 1.0 μm with 0.3 N of tension and a 0.45 mm depth, the local curvature change was very large. This was because maximum deformations were obtained at each node point, but curvature changes were not linearly proportional to maximum deformations. In other words, deformation and the curvature have not one-to-one correspondence. The point at which maximum deformation occurred was the same for

Fig. 10 Local curvature changes with respect to suturing tension

Fig. 11 Local curvature changes with respect to suturing depth

all cases, because the suturing depth was held constant. After the contact point, the curvature gradually decreased and converged. However, the curvature change did not become zero because of the effect of IOP.

Figure 11 shows the curvature changes with respect to suturing depth. Suturing tension was set to 0.3 N. Similarly, the curvature gradually converged to a single value. Local curvature changes decreased with increasing suturing depth due to the relationship between deformation and depth (Eq. 10). The main difference between Fig. 11 was the point where the maximum curvature change appeared. If the suturing depth was increased, the contact length also increased. Therefore, the contact point was closer to the origin point (Fig. 9). Curvature changes were not significantly different for the three cases. Thus, the suturing tension effect was much larger than the suturing depth on

curvature changes. However, the location of the contact point was very important optically.

As the location of the contact point was changed, the position at which the light was refracted and the incident angle changed simultaneously. This affected the focal point in the retina and correspondingly, the patient's visual recovery after the keratoplasty.

5 Conclusion

In this study, a simulation of the cornea during the suturing process examined cornea deformation and curvature change which affected optical behavior. The conclusions are as follows:

1. The constitutive equations of the cornea and the suture thread, as well as the boundary conditions were determined. The factors that affected the deformation and curvature of cornea were selected.
2. The effects of suturing tension and suturing depth on cornea deformation and curvature changes were numerically analyzed based on the proposed model. Cornea deformation and curvature increased with increasing suturing tension.
3. As the suturing depth increased, the local curvature did not significantly change. However, the contact position between the thread and the cornea changed. Curvature changes eventually led to changes in light refraction, which greatly affected the recovery of the patient's visual acuity. Both selected factors are important, affecting mechanical and optical behaviors of the cornea.
4. The simulation results can be useful not only in the prediction of deformation and curvature changes but also in optical behavioral change prediction from small deformations of the cornea. In addition, these simulation techniques could be used to predict the recovery of visual acuity of the patient after surgery through the derivation of optimal surgical factors.

Acknowledgements This work was supported by the Industrial Technology Innovation Program (No. 10048358) funded by the Ministry of Trade, Industry & Energy (MI, Korea).

Compliance with ethical standards

Conflict of interest The authors declare that they have no conflict of interest.

Ethical Statement This work does not contain any animal study.

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