Available at
http://pvamu.edu/aam
Appl. Appl. Math.
ISSN: 1932-9466

Vol. 10, Issue 1 (June 2015), pp. 620 - 633

Applications and Applied Mathematics:
An International Journal
(AAM)

Mathematical Model: Comparative Study of Thermal Effects of Laser in Corneal Refractive Surgeries

Gokul KC*, Dil Bahadur Gurung and Pushpa Raj Adhikary

Department of Natural Sciences (Mathematics)
School of Science
Kathmandu University
Dhulikhel, Kavre, Nepal
*Corresponding Author Email: gokulkc2@gmail.com

Received: June 10, 2014; Accepted: May 19, 2015

Abstract

Lasers have been widely used in ophthalmology. Refractive errors are some of the most common ophthalmic abnormalities worldwide. Laser refractive surgery was developed to correct refractive errors myopia, hyperopia and astigmatism. Two types of laser surgical techniques: lamellar and thermal are available to reshape the corneal curvature. Ultraviolet (UV) emitting argon fluoride (ArF) excimer laser is used to sculpt cornea in lamellar procedures, whereas, infrared (IR) emitting holmium yttrium aluminum garnet (Ho: YAG) laser is used to shrink cornea in thermal procedure. Tissue heating is common in all types of laser surgical techniques. Hence, in this paper, a finite element model is developed to investigate the temperature distribution of cornea in different laser refractive surgeries. Characteristics of optical and thermal processes and influence of the parameters of radiation and tissues on the results of laser action are investigated. The results of mathematical modeling in different surgical techniques are discussed, compared, and validated with experimental results.

Keywords: Temperature, Laser; Human eye; Refractive errors; LTK; LASIK; Finite element method

MSC 2010 No.: 92B05, 80A20, 80M10, 78A60

1. Introduction

1.1. Background of the study

After the development of infrared lasers many researchers performed investigation on their use in ophthalmology. Clinical experiences with applications of ultraviolet and infrared laser radiation for the treatment of different ocular diseases has shown marked advantages. Nowadays, lasers have become so universal that it is difficult to imagine ophthalmology without them. This increased use of lasers and widespread application of different types of eye surgeries have resulted in the need for quantitative understanding of basic interaction between laser radiation and the human eye tissues.

Light tissue interaction results in transmission, reflection, scattering or absorption of the light. Among these interactions absorption is most important because absorption of laser radiation by human eye tissue cause significant changes to the cellular material (Ansari and Mohajerani, 2011). The effects of observed laser photons on biological tissues can be divided into three general categories: photochemical, thermal and ionizing. Generally, thermal effects of laser on tissue is used in ophthalmology to treat retinopathy, macular degeneration etc., ionizing to treat tumors, cancer cells etc. and photochemical to treat refractive errors (Hojlo et al., 2007). Tissue heating is common in photochemical and ionizing effects. Therefore, the greatest concern is with the heating effects to the eye tissues (Ooi et al., 2008).

Lagendijk (1982) used a finite difference method to calculate the temperature distribution in human and rabbit eyes during hyperthermia treatment. The heat transport from the sclera to the surrounding anatomy is described by a single heat transfer coefficient which includes the impact of blood flow in choroid and sclera. Scott (1988), Ng and Ooi (2006), Gokul et al. (2013), Ooi et al. (2009), Ng et al. (2008) utilized finite element method to obtain the temperature profile based on heat conduction using various heat transfer coefficients given by Lagendijk. Flyckt et al. (2006) studied the impact of choroidal blood flow by using three methods: Lagendijk model, bio-heat model and discrete vasculature model in the eye and the orbit. Amara (1995), Narahsimah et al. (2009), Cvetcovic et al. (2011) presented a thermal model of laser in different retinal disease treatments. Ooi et al. (2008), Cvetvovic et al. (2011), Pustovalov and Jean (2006) modeled human eye undergoing Laser thermokeratoplasty.

1.2. Significance and mechanism of corneal refractive surgeries

Refractive errors are some of the most common ophthalmic abnormalities worldwide. Eyeglasses are the mainstream treatment ever since. Refractive surgery was developed so that people could enjoy good vision with no or reduced dependence on glasses or contact lenses (Yu and Jackson, 1999). In normal eye, light is refracted first by the cornea and then by the lens to focus on the retina. The cornea contributes 60 – 70% of refractive power of the eye, the remaining 30 – 40% coming from the lens (Pustovalov and Jean, 2006).

People with refractive error have irregularities in cornea and/or shape of eyeball that cause light to focus too far in front of or too far behind retina. Laser refractive surgery reshapes the corneal curvature, making it flatter or steeper as required to counterbalance the patient's refractive error. There are different laser surgical techniques, each with its own advantages/disadvantages, but they all have in common that they alter the structure, shape and size of the cornea. Laser surgical techniques to correct refractive errors can be divided into 2 categories: lamellar and thermal (Yu and Jackson, 1999). In lamellar procedure, UV emitting argon fluoride excimer laser (ArF laser) is used to remove sub-micrometer amount of tissue from cornea through photo ablation. In thermal procedure, IR emitting holmium yttrium aluminum garnet (Ho: YAG laser) is used to shrink stromal collagen through heating.

Some common lamellar procedures are photorefractive keractomy (PRK), laser in situ keratomileusis (LASIK) and laser sub epithelial keratomileusis (LASEK). LASIK, PRK and LASEK are photo-ablation techniques where refractive power of the cornea is altered by removing bits and pieces from the corneal surface through ablation (Ooi and Ng, 2009). In PRK the epithelium is removed mechanically and the stromal bed is treated with excimer laser. In LASEK an epithelial flap is detached after application of diluted alcohol solution and laser is applied on it. After laser ablation, the epithelium is peeled back and repositioned. In LASIK, a thin-hinged corneal flap is created with a microkeratome and laser is applied. After photo-ablation, the flap is repositioned (Jr and Wilson, 2003).

Laser thermokeratoplasty (LTK) is the commonly used thermal procedure. LTK is a refractive procedure utilizing heat to reshape the surface of cornea through shrinkage of collagen within the stroma (Pustovalov and Jean, 2004). The increase in refractive power is achieved through this shrinkage of corneal collagen. Shrinkages are induced thermally through laser heating (Ooi et al., 2008). Corneal collagen is known to shrink when heated to temperature on the order of 100°C (Cvetkovic et al., 2011).

Although the result of these refractive surgeries is very predictable (over 90% of eyes will be within ±1.00D of the intended correction), overcorrection and under correction are still possible (Yu and Jackson, 1999). Under correction is more common than overcorrection, occurring in 4 – 10% of eyes. These problems may be permanent or may be corrected through a retreatment. Clinical studies reported that refractive success rate is in between 80 – 90% only (Hojlo et al., 2007). Another major complication is the dry eye, which is more frequent and severe in corneal refractive surgeries. Dry eye may be temporary side effect or become long term permanent (Teneri et al., 2003).

The most significant issue in the eye surgeries with laser is estimation of temperature distribution in eye tissue due to laser radiation intensity. If left unaddressed, the thermal effects may cause tissue damage and potentially reduce refractive outcomes. Therefore, modeling of heat transport is needed in order to explain the actual temperature variation in corneal refractive surgeries.

Most past models focused on modeling heat transport on different retinal diseases and injuries. Although some investigators modeled laser thermokeratoplasty on treating corneal refractive surgery, there is a lack of thermal modeling in excimer laser surgeries and extensive comparison between them. In the present paper, finite element method is used as a tool to find the transient temperature distribution of cornea in lamellar (PRK/LASIK/LASEK) and thermal (LTK) refractive surgeries. The possible effects of temperature increase in refractive outcomes are discussed and compared. The results so obtained are validated with experimental results.

2. Model Formulation

2.1 Discretization

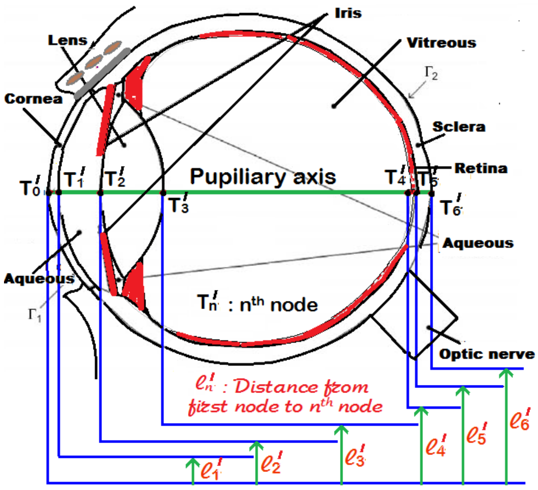

Figure 1: Schematic diagram of human eye

A schematic diagram of human eye is displayed in figure 1. For modeling purpose, the opened eye is divided into six regions: cornea, aqueous humor, lens, vitreous humor, retina, and sclera. The choroid is relatively thin compared to retina, so we modeled choroid together with retina. Iris and cilliary body do not exist on pupillary axis, they are excluded from analysis. Each region is assumed perfectly bonded, homogeneous and the eye is assumed symmetrical about the pupillary axis. The diameter of the eye along pupillary axis is about 25.10 *mm* (Cvetkovic et al., 2011).

Initially, the eye is discretized into 6 regions based on thickness of each layer cornea, aqueous, lens, vitreous, retina and sclera. Furthermore, the analysis is carried out by increasing the mesh size. For convergence study, the mesh size is increased to 500, 5000 and 50000 elements and temperature values are calculated.

2.2. Governing equation and boundary conditions

The governing differential equation representing the bio-heat transfer in the human eye can be written by the well-known Pennes equation addressing the effect of blood perfusion and metabolism is given by (Pennes, 1998):

$$\rho c \frac{\partial T}{\partial t} = k\nabla^2 T + \omega \rho_b c_b (T_b - T) + Q_m + Q_l \,, \tag{1}$$

where, ρ_b = blood density ($Kg\, m^{-3}$), c_b = blood specific heat ($J\, Kg^{-1}\, °C^{-1}$), k = tissue thermal conductivity ($W\, m^{-1} °C^{-1}$), ω = volumetric blood perfusion rate per unit volume (s^{-1}), T_b = blood temperature (°C), T = tissue temperature (°C), Q_m = heat generation due to metabolism (Wm^{-3}) and Q_l = heat generation due to laser (Wm^{-3}).

The heat generation in tissue due to laser radiation depends mainly on absorption coefficient $\alpha\ (m^{-1})$ and fluence rate $I(r,z)(wm^{-2})$ in cylindrical coordinates is given by (Shibib 2013):

$$Q_l(r,z) = \alpha\, I(r,z)\ , \tag{2}$$

where, the fluence rate is given by

$$I(r,z) = I_0\, e^{[-(\alpha+(1-g)\alpha_s)z]}\ , \tag{3}$$

where, α_s is the scattering coefficient, g is anisotropy factor. If absorption dominates scattering then the fluence rate at any depth can be governed by Lambert-Beers law. The heat generation through the tissue due to laser photons absorption is

$$Q_l(r,z) = \alpha\ I_0\, e^{-\alpha z}\ , \tag{4}$$

where, I_0 is the incident value of intensity. For top hat laser beam profile the induced power intensity is given by

$$I_0(r,z) = \frac{4P}{\pi\omega^2}\ , \tag{5}$$

where, P is laser power in watts, ω is the beam waist.

Hence the heat generation due to laser at cornea in one dimension ($r = 0, z = x$) is given by

$$Q_l(x) = \alpha\ I_0\, e^{-\alpha x}\ , \tag{6}$$

Boundary conditions for the system can be defined as follows:

1. On the outer surface of the sclera, the heat flows run into the eye with the complicated network of ophthalmic vessels which are located inside the choroidal layer acting as a heating source to the sclera. This heat exchange between the eye and the surrounding is modeled using the following convection boundary condition:

$$\Gamma_2:\ -\mathrm{k_s}\frac{\partial \mathrm{T}}{\partial \eta} = \mathrm{h_b}(\mathrm{T} - \mathrm{T_b})\ , \tag{7}$$

   where, η is the normal direction to the surface boundary, k_s is the thermal conductivity of sclera, h_b is the heat transfer coefficient between blood and eye (Wm^{-2}°C), and T_b is blood temperature (°C).

2. Since outer surface of the eye (cornea or skin) is exposed to the environment, the heat loss caused via convection, radiation, and evaporation. This loss is modeled using the following boundary condition :

$$\Gamma_1:\ -\mathrm{k_c}\frac{\partial \mathrm{T}}{\partial \eta} = \mathrm{h_a}(\mathrm{T} - \mathrm{T_a}) + \sigma\varepsilon(\mathrm{T}^4 - \mathrm{T_a^4}) + \mathrm{E}\ , \tag{8}$$

   where, h_a presents the convection heat transfer coefficient between the cornea and ambient environment (Wm^{-2}°C), T_a is the ambient room temperature (°C), σ is the Stefan Boltzmann constant ($5.67 \times 10^{-8} W/m^{2\,o}C^4$), ε is the emissivity of the cornea, and E is evaporative heat loss (W/m^2) between cornea and environment.

The nonlinear radiation term in the boundary condition (8) is treated by using simple iterative procedure as follows:

$$-k_c \frac{\partial T_1}{\partial \eta} = [h_a + \sigma\varepsilon(T_1 + T_a)(T_1^2 + T_a^2)](T_1 - T_a) + E\,, \tag{9}$$

$$-k_c \frac{\partial T_1^m}{\partial \eta} = h_{cr}(T_1^m - T_a) + E\,, \tag{10}$$

where,

$$h_{cr} = h_a + \sigma\varepsilon(T_1^{m-1} + T_a)((T_1^{m-1})^2 + T_a^2)\,, \tag{11}$$

$$h_{cr} = h_{convection} + h_{radiation}\,,$$

where, T_1^m are temperature sequences for m = 1, 2, 3, … and T_1^0 represents an initial guess of temperature.

The iteration is completed when the convergent condition is satisfied:

$$||T_1^m - T_1^{m-1}|| < \delta, \tag{12}$$

where, δ is iteration tolerance.

2.3. Numerical solution procedure

The partial differential equation (1) together with laser source equation (6) and boundary conditions (7) and (10) in one-dimensional variational form is:

$$I = \frac{1}{2}\int_L \left[K\left(\frac{dT}{dx}\right)^2 + \omega\rho_b c_b (T_b - T)^2 - 2Q_m T - 2Q_l T + \rho c \frac{\partial T^2}{\partial t}\right] dx + \frac{1}{2} h_b (T - T_b)^2 + \frac{1}{2} h_{cr} (T - T_a)^2 + ET. \tag{14}$$

To minimize I, we differentiate I partially with respect to T_i and equating to zero as follows,

$$\frac{\partial I}{\partial T_i} = 0. \tag{15}$$

Equation (15) is the system of linear equations, which can be written in matrix form as

$$[C]\{\dot{T}\} + [K]\{T\} = \{R\}, \tag{16}$$

where $\{\dot{T}\} = \left\{\frac{\partial T_i}{\partial t}\right\}$, $\{T\} = \{T_i\}$ and $\{R\} = \{R_i\}$ are $N \times 1$ vectors. Similarly, [C] and [K] are $N \times N$ matrices called conductivity and capacity matrices respectively. N is the total number of nodal points in discretization of the domain.

Now Crank-Nicolson method is applied to solve the system (16) with respect to time using the following relation

$$\left(\frac{1}{\Delta t}[C] + \frac{1}{2}[K]\right)\{T\}_{n+1} = \left(\frac{1}{\Delta t}[C] - \frac{1}{2}[K]\right)\{T\}_n + \{R\}, \tag{17}$$

where, Δt is time interval.

The temperature increases from outer surface of the eye towards core, when ambient temperature is less than $37℃$ and vice versa. Hence, we consider the temperature increases/decreases in linear order towards eye core with regard to thickness. For initial nodal temperatures $\{T\}_0$ at time $t = 0$, we assume the following initial condition

$$T(x = l_i, t = 0) = T(0,0) + rl_i\,, \tag{18}$$

where, i ranges from $1,2,3,\ldots,6$, $T(0,0) = 20^{\circ}$C and $r =$ constant to be determined. The equation (17) is repeatedly solved to get the required nodal temperatures.

2.4. The Control Parameters

The cornea is comprised of five distinct layers: epithelium, Bowman's membrane, stroma, descent's membrane and endothelium. The two major physiologic functions of the cornea are as a bio-defense and as a refractive system. The cornea contributes $60 - 70\%$ of refractive power of the eye (Yu and Jackson 1999). Water is the main component of corneal tissue. Collagen is the major structural component of the corneal stroma accounting for $12 - 15\%$ of its weight (Pustovalov and Jean 2004). The cornea causes negligible light scattering (Petit and Ediger 1996). The major absorbers of far UV and IR laser light are proteins, peptides, pigments and water. The cornea absorbs almost all wavelengths greater than $3000nm$, less than $300nm$ and most radiation above $1400nm$ (Voke 2008). The cornea absorbs more than 90% of laser energy in the wavelength between $1.85\mu m$ and $2.1\mu m$ (Ooi et al. 2008) and remaining light transmits and absorbed in aqueous humor.

ArF excimer laser emits high energy ultraviolet light at a wavelength of $193nm$. Common ArF excimer lasers for medical purpose have average laser power $0.64W$, spot size $0.97\ mm^2$, pulse duration $10 - 50ns$, relaxation time $2000 - 10000\mu s$ (Silfvast 2009). An average ablation rate of $10.9\mu m$ per spherical equivalent diopter of correction was recently reported for three clinical refractive surgery systems (Fisher and Hahn 2011). The absorption coefficient of cornea is taken as $39900cm^{-1}$ for $193nm$ wavelength of ArF laser (Petit and Ediger 1996). Fisher and Hahn (2011) developed a functional relationship between laser pulse energy and ablation rates using experimental setup as

$$y = 0.18094\,x + 0.39666, \tag{19}$$

where, $x =$ laser pulse energy (mJ) and $y =$ ablation depth (μm).

Ho: YAG laser emits infrared light at a wavelength of $2070nm$. The general parameter values for this laser are: wavelength(λ) = 2.1μm, laser energy $=25mJ/pulse$, pulse duration = 200μs, relaxation time = 2s, laser power = 150W, beam diameter = 0.6mm (Silfvast 2009). Corneal collagen starts to denature when heated above $45°C$ and shrink when heated to temperatures above $55℃$ and when its temperature exceeds $90℃$, it relaxes to contour the intended corneal shrinkage (Ooi et al. 2008). The absorption coefficient of Ho: YAG laser for corneal tissues lie in the range $25 - 100\text{cm}^{-1}$ (Pustovalov and Jean 2006).

The parameter values for different parts of eye are presented in Table 1.

Table 1. Thermal properties of human eye tissues (Gokul et al. 2013, 2014)

Tissue	Thermal Conductivity (K) ($W\,m^{-1}{}^{\circ}C^{-1}$)	Blood Perfusion(ω) (s^{-1})	Metabolic Rate (Qm) (Wm^{-3})	Density (ρ) (Kgm^{-3})	Specific heat (C) ($J\,Kg^{-1}\,{}^{\circ}C^{-1}$)
Cornea	0.58	0	0	1050	4178
Aqueous	0.58	0	0	996	3997
Lens	0.400	0	0	1050	3000
Vitreous	0.603	0	0	1000	4178
Retina	0.565	0.0222	22000	1050	3680
Sclera	1.0042	0	0	1100	3180

2.5. Methodology

Nowadays up to, $-14D$ of myopia, $+6D$ of hyperopia and $\pm 6D$ of astigmatism can be corrected using ArF excimer laser. For modeling purpose, we take $3D$ correction as a base. Since, $1D$ correction is equivalent to $10.9\mu m$ of ablation; $3D$ correction is equivalent to $32.7\mu m$ of ablation. The average energy per pulse is $1.5mJ$ with maximum $2.5mJ$ and average frequency $100Hz$ with maximum $250Hz$ are currently used in practice. Thus in this study, the temperature distribution are calculated at pulse energies $1.5mJ, 2.5mJ$ and $5mJ$ and at frequencies $100Hz, 250Hz$ and $500Hz$ respectively. The relaxation time between two laser pulses is sampled in $10\mu s$ steps.

LTK shows much promise in correcting the refractive errors of hyperopia and astigmatism. Nowadays, LTK can correct mild hyperopia, astigmatism or combination of them without involving corneal epithelium or central visual axis. Typical application consists of delivering several laser pulses in an annular pattern in order to induce local shrinkage of the corneal collagen. According to Ooi et al. (2008), a typical pulsed LTK treatment consists of seven laser pulses, which we consider here. The temperature distribution during Ho: YAG laser surgery is calculated at different absorption coefficients $2500, 4000$ and $5000m^{-1}$. The relaxation time between two laser pulses is sampled in $1\mu s$ steps.

The governing equation (1) with boundary conditions (7) and (10) is solved without laser source in steady state using different parameter values as tabulated in table 1. The obtained steady state results are utilized as initial condition for the transient analysis when laser is subjected to corneal stroma. To approximate the solution, finite element method with Lagrange linear basis function is used for space and finite difference method (Crank – Nicholson) is used for time. The nonlinear term in boundary condition (8) is linear zed using iterative procedure as discussed in section 2.2. The tolerance value(ε) is chosen as 10^{-5}.

3. Results

The convergence study has been carried out by varying the mesh size of the domain in steady state case without using laser. Our initial mesh consists of 6 elements representing each eye tissues. The mesh size is increased to $500, 5000$ and 50000 elements and temperature values are calculated. The obtained temperature values are tabulated in table 2.

Table 2. Convergence study

Mesh Type	Mesh size	Temperature values
Coarse	6	32.1745℃
Normal	500	32.1754℃
Fine	5000	32.1768℃
Extremely fine	50000	32.1782℃

Figure 2 shows the peak temperature distribution of single pulse during the application of different laser pulse energies in ArF laser surgery. The peak temperature (at single pulse) obtained are 140℃, 156.93℃ and 234.14℃ respectively. Figure 3 shows the peak temperature distribution during the application of different absorption coefficients in LTK. The peak temperature obtained are 91.29℃, 105.49℃ and 152.08℃ respectively. The temperature values on the order of 100℃ are required to initiate shrinking of corneal tissues.

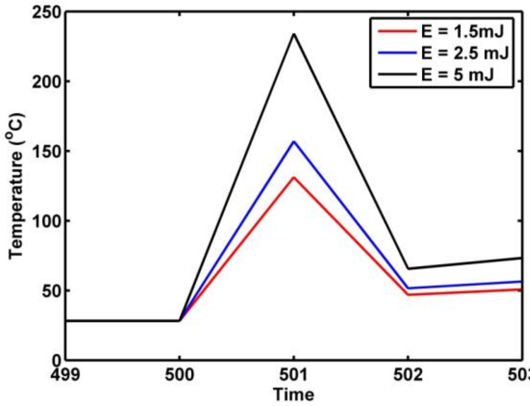

Figure 2. Peak temperature for different laser power in ArF refractive surgery

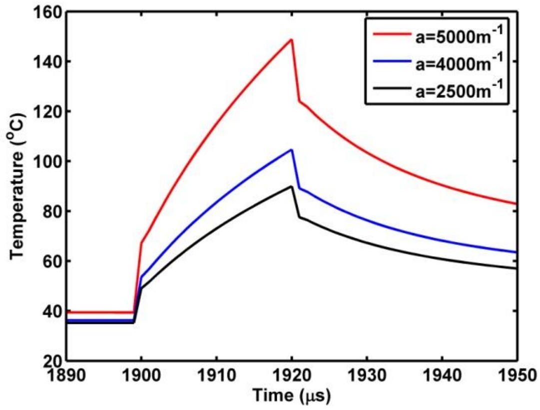

Figure 3. Peak temperature for different absorption coefficients in LTK

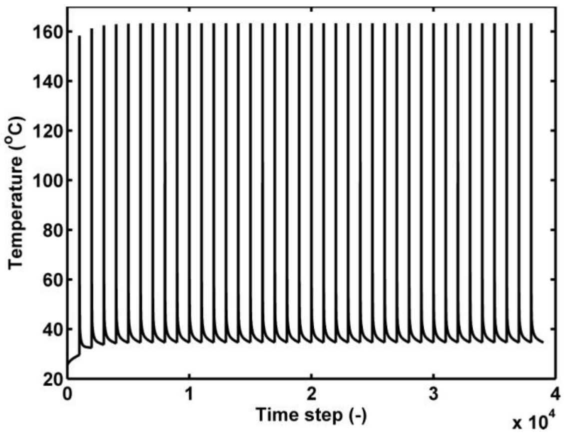

Figure 4. Temperature distribution of cornea at frequency 100Hz

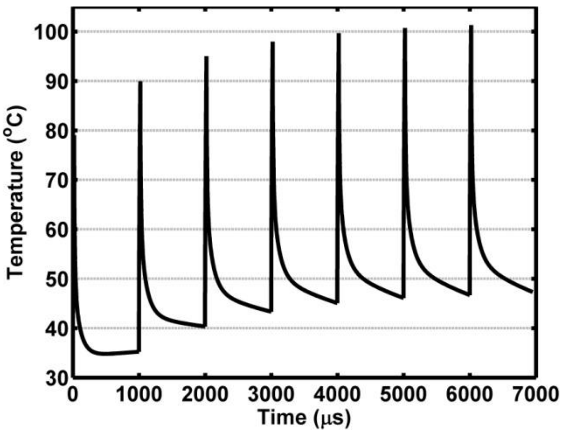

Figure 5. Temperature distribution of cornea at absorption coefficient $2500m^{-1}$

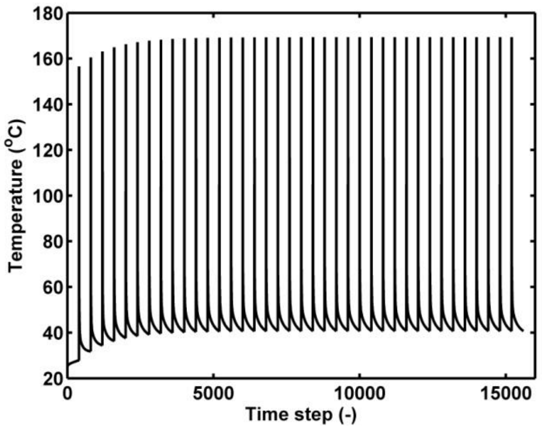

Figure 6. Temperature distribution of cornea at frequency 250Hz

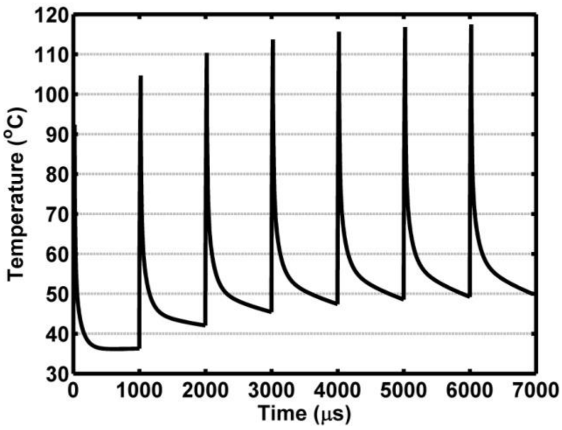

Figure 7. Temperature distribution of cornea at absorption coefficient $4000m^{-1}$

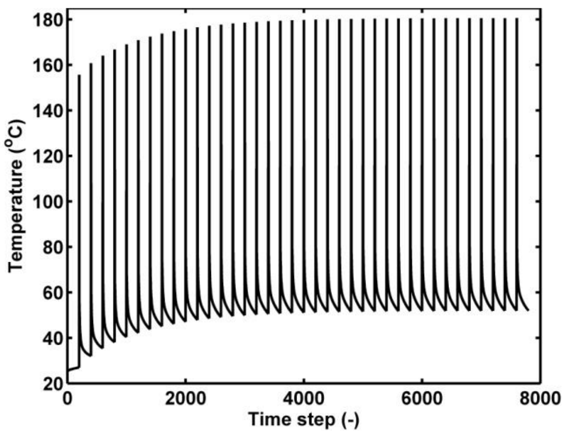

Figure 8. Temperature distribution of cornea at frequency 500Hz

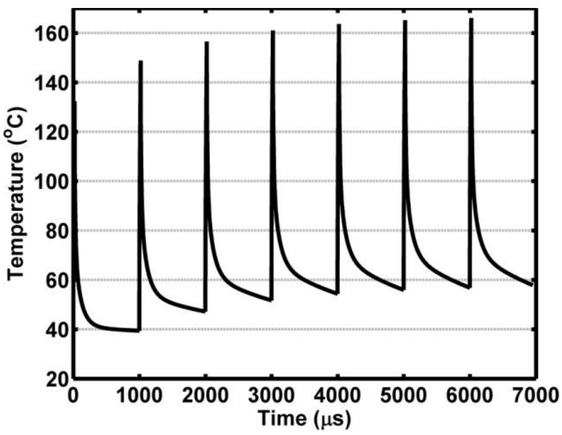

Figure 9. Temperature distribution of cornea at absorption coefficient $5000m^{-1}$

According to the functional relationship between ArF laser pulse energy and ablation rates in PRK developed by Fisher and Hahn, the ablation rate at $2.5mJ$ pulse energy is $0.85\mu m$ per pulse. To ablate $32.7\mu m$ of corneal stroma, approximately 38 pulses are needed.

Figure 4 shows the temperature distribution of cornea in LASIK at laser frequency $100Hz$ and constant pulse energy $2.5mJ$. Initially temperature rises very fast and after some pulses, it reaches in steady state. The steady state temperature obtained after 38 pulses is $34.60℃$. Similarly, steady state temperature obtained after 38 pulses at laser frequencies $250\ Hz$ and $500Hz$ are $40.70℃$ and $51.99℃$ as shown in figures 6 and 8 respectively.

Figure 3 shows the temperature distribution pattern of 7 pulses and at absorption coefficient $2500m^{-1}$ in LTK. The corneal temperature increases above $100℃$ with steady state temperature $47.31℃$. Figure 5 and 7 show the temperature distribution pattern of 7 pulses and at absorption coefficient $4000m^{-1}$and $5000m^{-1}$ respectively. The steady state temperature obtained after 7 pulses at absorption coefficients $4000m^{-1}$ and $5000m^{-1}$ are 53.72℃ and 59.27℃ respectively.

4. Discussion

In ArF laser surgeries, corneal epithelium is removed first and laser is applied on the bed of stroma. If the stromal temperature reaches greater than 45℃, collagen denaturation may starts. Even in 40℃ stromal temperature in presence of environmental oxygen, several chemical reactions like oxidation may possible. These effects may cause tissue damage and potentially may reduce refractive outcomes. Hence the increase in temperature and chemical reactions may damage (in the sense of ablation) stroma, which may be the cause of over/under correction in LASIK.

After removal of epithelium flap in LASIK, due to high evaporation, the steady corneal temperature is achieved at 30℃, which is closed to the experimental studies. Based on this normal value, the temperature is increased up to 5℃, 15℃ and 33℃ during $100\text{Hz}, 250\text{Hz}$ and 500Hz frequency used for $3D$ correction. Moldonado-Codina et al. (2001) investigated the temperature changes occurs during photorefractive keratectomy (PRK) when performed in 19 bovine corneas. They observed an average temperature rise at the corneal surface is 8℃. Morchen et al. (2009) observed a maximum temperature increase of 15℃ in bovine corneas for $-9.00D$ myopia treatment and 11℃ for $+6D$ hyperopia correction. Betney et al. (1997) observed maximum temperature increase of 9.5℃ during PRK surgery. This shows that our results in LASIK are in good agreement with past experimental results.

We discussed above the effects of temperature increase in ArF surgeries. The major side effect discussed in ArF surgeries is under and over correction. Over and under correction are due to the denaturation of proteins and collagen shrinkage. Hence in ArF surgeries collagen shrinkage is the major side effect, however in LTK collagen shrinkage is the intended effect. In LTK, temperature should reach to that threshold value from where corneal shrinkage starts. In ArF surgeries, high energy UV laser's very short pulses heats up stroma for very small time $(10 - 30ns)$ and break intermolecular bonds. This process vaporizes stromal collagen precisely and sculpts the cornea. In LTK, IR laser's very long $(200\mu s)$ pulses (compare to ArF laser) heats up cornea constantly to a certain degree and evaporates water from stroma/epithelium.

After the epithelium is removed from cornea, exposure to environmental air dehydrates the corneal tissue and establishes a hydration gradient. The majority of corneal stroma consists of lamellae composed of type I collagen fibers that are of uniform dimension and are evenly spaced by a matrix of proteoglycans. Proteoglycans are more efficient at holding water and plays important role in corneal hydration. Evaporation of water from matrix of proteoglycans, leading to decreased spacing of collagen fibers. This precisely shrinks and makes steeper cornea in LTK. However, Slight variation in corneal hydration is considered a potential source of variation in stromal ablation rates in LASIK. The stromal ablation rate increases with increasing hydration and hydration increases with corneal depth (Fisher and Hahn 2011). Therefore, one possible variable behind over/under correction may be stromal hydration.

The main problem of LTK is related to the selection of laser radiation parameters that provide effective heating and shrinkage of the entire volume of stroma irradiated by laser beam. The use of radiation with $\lambda = 1.0 - 1.35\mu m$ with very low value of α in the range $0.1 - 5cm^{-1}$

is dangerous because of very deep penetration of radiation in intraocular tissues. The use of radiation with wavelength $> 2.5\mu m$ and absorption coefficient $\alpha = 100 - 1000 cm^{-1}$ leads to very shallow penetration and heating of corneal tissues.

The results show that it is possible to achieve the tissue heating above $100°C$ for collagen shrinkage. The level of shrinkage very strongly depends on temperature. Ooi et al. (2008) investigated the temperature distribution of cornea using boundary value method. They obtained the maximum temperature of $110°C$ after 7 pulses. Cvetkovic et al. (2011) obtained significantly higher temperature values $230.12°C$ after LTK. Pustovalov et al. (2004) found relatively lower temperature values $70 - 85°C$ inside the stroma after LTK. Our results lie within the results from past literatures and show the best temperature values for corneal shrinkage and hence for LTK.

The variation of the absorption coefficient of corneal tissue leads to significant changes of the shape, position and dimension of the heated region. Within a given temperature range, thermal denaturation is associated with collagen shrinkage parallel to the axis of the collagen. This is the intended effect in LTK. Long term stability depends on the parameters of laser radiation and cornea tissue, all of which contribute to control stable collagen shrinkage. There are many cases of failure and of unwanted adverse effects and applications for LTK.

Lamellar procedures (PRK/LASIK/LASEK) are the most widely used methods in refractive surgery. Because they are stable and gives long lasting results until presbyopia starts. Excimer laser surgeries show more promise over LTK in correcting mild to severe myopia, mild hyperopia and astigmatism but not the combination of them. The thermal procedure (LTK) is a repeatable procedure demonstrating more stable results over five to eight years. LTK is effective for hyperopia and astigmatism but not for myopia because LTK does not perform in the central part of the cornea. LTK makes more spherical and natural shape of the corneal surface than excimer laser surgery. LTK shows more promise over excimer laser surgeries in correcting hyperopia, astigmatism and combination of them.

5. Conclusion

We developed a finite element mathematical model based on time dependent Pennes bioheat equation of human cornea during PRK/LASIK/LASEK and LTK. The temperature increase by ArF excimer laser in lamellar procedures and by Ho: YAG laser in thermal procedures of corneal refractive surgeries are discussed. The temperature values obtained by varying absorption coefficients, laser power and frequencies show good agreement with experimental results. Heating due to absorption of laser photons by corneal tissues causes collagen shrinkage, protein denaturation, and more evaporation of water and vaporization of stroma. In lamellar procedure these are the side effects and in thermal procedures these are the intended effects. However, the temperature control for better results in both surgical techniques is still a challenge. Although refractive surgery techniques are safe and effective, temperature increase should be considered as a risk factor. It is possible to increase success rate of refractive surgeries by selecting the appropriate laser parameters in different spectral ranges. We can extend this model to predict the thermal effects due to laser radiation by various lasers in different tissues of human body. It is expected that, the information generated from this model are useful to laser surgeons, biomedical engineers, medical scientists and future researchers.

REFERENCES

Amara E. H. (1995). Numerical investigations on thermal effects of laser-ocular media interaction, Int. J. Heat Mass Transfer, Vol. 38.

Ansari M. A. and Mohajerani E. (2011). Mechanisms of Laser Tissue Interaction: I. Optical Properties of Tissue, J. Lasers. Med. Sci., Vol. 2.

Betney S., Morgan P. B., Doyle S. J. and Efron N. (1997). Corneal temperature changes during photorefractive keratectomy, Cornea, Vol. 16.

Codina M. C., Morgan P. and Efron N. (2001). Thermal consequences of photorefractive keratectomy, Cornea, Vol. 20.

Cvetkovic M., Poljak D., Peratta A. (2011) FETD Computation of the Temperature Distribution Induced into a Human Eye by a Pulsed Laser, Prog. in Electromagnetics, Vol. 120.

Fisher B. T. and Hahn D. W. (2011). Real time measurement of ArF excimer laser corneal tissue ablation rates using cross – correlation of laser waveforms, Optics Express, Vol. 19.

Flycket V. M. M., Roaymakers B. W., Lagendijk J. J. W. (2006). Modeling the impact of blood flow on the temperature distribution in the human eye and the orbit: fixed heat transfer coefficient versus the Pennes bioheat model versus discrete blood vessels, Phys. Med. Biol., Vol. 51.

Gokul, K. C., Gurung, D. B. and Adhikary, P. R. (2013). Effect of Blood Perfusion and Metabolism in Temperature Distribution in Human Eye, Advances in applied mathematical Biosciences, Vol. 4.

Gokul, K. C., Gurung, D. B. and Adhikary, P. R. (2013). FEM Approach for Transient Heat Transfer in Human Eye, Applied Mathematics, Vol. 4.

Gokul, K. C., Gurung, D. B. and Adhikary, P. R. (2013). Transient thermal model of airflow effects in human eye temperature, Kathmandu University journal of science, engineering and technology, Vol. 9.

Gokul, K. C., Gurung, D. B. and Adhikary, P. R. (2014). Modeling airflow effects in human eye temperature with and without eyelids, International journal of applied mathematics and mechanics, Vol. 10.

Gokul, K. C., Gurung, D. B. and Adhikary, P. R. (2014). Thermal effects of eyelid in human eye temperature, Journal of applied mathematics and informatics, Vol. 32.

Hojlo M. M., Berkowska P. K. and Bator A. H. (2007). Therapeutic Application of lasers in ophthalmology, Adv. Clin. Exp. Med., Vol. 16.

Jr R. A., Wilson S. E. (2003). LASIK vs LASEK vs PRK: Advantages and indications, Seminars in Ophthalmology, Vol. 18.

Lagendijk J. J. W. (1982). A mathematical model to calculate temperature distributions in human and rabbit eyes during hyperthermia treatment, Phys. Med. Biol., Vol. 27.

Mirenzami S. A., Jafarabadi M. R. and Abrishami M. (2013). Temperature distribution simulation of the human eye exposed to Laser radiation, J. Lasers Med. Sci., Vol. 4.

Morchen M., Schelling U., Wuellner C. and Donitzky C. (2009). Effect of time sequences in scanning algorithms on the surface temperature during corneal laser surgery with high repetition rate excimer laser, J. Cataract Refrect Surg., Vol. 35.

Narasimhan A., Jha K. K. and Gopal L. (2009). Transient simulations of heat transfer in human eye undergoing laser surgery, Int. J. Heat Mass Transfer, Vol. 53.

Ng E. Y. K. and Ooi E. H. (2006), FEM simulation of the eye structure with bio-heat analysis, Comp. Methods Programs Biomed., Vol. 82.

Ng E. Y. K. and Ooi E. H. (2011), Effects of natural convection within the anterior chamber on the ocular heat transfer, Int. J. Numer. Meth. Biomed. Engng., Vol. 27.

Ng E. Y. K., Ooi E. H. and Archarya U.R. (2008), A comparative study between the two dimensional and three dimensional human eye models, Mathematical and computer modeling, Vol. 48.

Ooi E. H. and Ng E. Y. K. (2009), Ocular temperature distribution: A mathematical perspective, Journal of mechanics in Medicine and Biology, Vol. 9.

Ooi E. H., Ang W. T. and Ng E. Y. K. (2008). A boundary element model of the human eye undergoing laser thermokeratoplasty, Comp. Biol. Med., Vol. 28.

Ooi E. H., Ang W. T. and Ng E. Y. K. (2009). A boundary element model for investigating the effects of eye tumor on the temperature distribution inside the human eye, Computers in Biology and Medicine, Vol. 39.

Ortueta D. D., Magnago T., Triefenbach N., Mosquera S. A., Sauer U. and Brunsmann U. (2012). In vivo measurement of thermal load during ablation in high speed laser corneal refractive surgery, J. Refract Surgery, Vol. 28.

Pennes H. H. (1998). Analysis of tissue and arterial blood temperatures in the resting human forearm, J. Appl. Physiology, Vol. 85.

Pettit G. H. and Ediger M. N. (1996). Corneal – tissue absorption coefficients for 193nm and 213nm ultraviolet radiation, Applied Optics, Vol. 35.

Pustovalov V. K. and Jean B. (2006). Theoretical investigations of the processes of laser interaction with ocular tissue for laser applications in ophthalmology, Laser Physics, Vol. 16.

Pustovalov V. K. and Jean B. (2004), Investigations and Computer Modeling of the Optical and Thermal Processes of Laser Thermal Keratoplasty, Laser Physics, Vol.14.

Scott J. A. (1988). A finite element model of heat transport in the human eye, Phys. Med. Biol., Vol. 33.

Shibib K. S. (2013). Finite element analysis of cornea thermal damage due to pulse incidental far IR laser, Lasers Med. Sci., Vol. 28.

Silfvast W.T. (2009). *Laser Fundamentals* (2nd editions), Cambridge, New Delhi, India.

Taneri S., Zieske J. D. and Azar D. T. (2004). Evolution, techniques, clinical outcomes and pathophysiology of LASEK: review of literature, Surv. Ophthalmol., Vol. 49.

Voke J. (2008). Occupational Vision Hazards: Infrared Radiation and the Eye, Optometry Today, Vol. 48.

Yu E. Y. W. and Jackson W. B. (1999). Recent advances in refractive surgery. CMAJ, Vol. 160.