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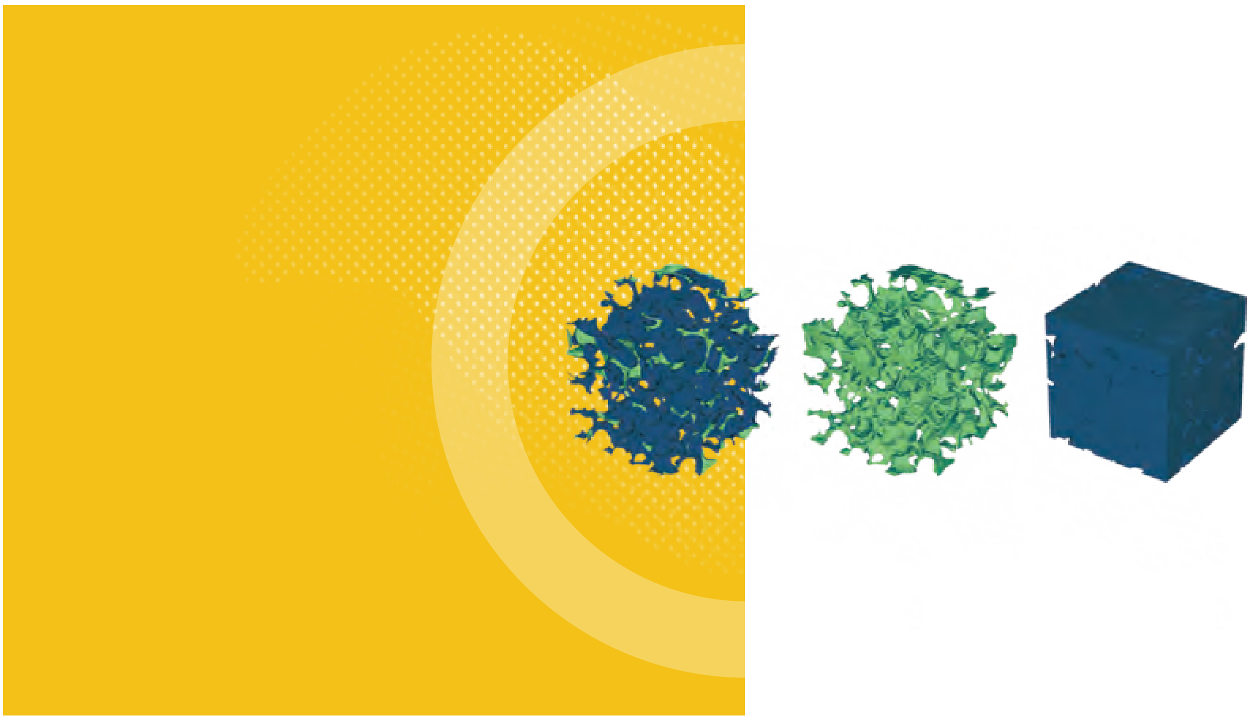

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Biomechanical Simulations of Corneal Refractive Surgery

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Summary: This short paper reports some preliminary results on the biomechanical simulation of corneal refractive surgery techniques such as Radial Keratotomy (RK) and PhotoRefractive Keratectomy (PRK). These surgical techniques are used to reshape the human cornea and thus modify its refractive power as needed to resolve most common refractive disfunctions such as myopia, hyperopia and astigmatism.

Introduction

Refractive surgery of the cornea has become in recent years a diffuse technique to cope with most common refractive defects in human vision [1, 3, 6]. After the pioneering attempts of RK in the second half of 1900, the availability of excimer-based laser techniques such as PRK and flap-based LASIK (LAser in SItu Keratomileusis) have blown-up the number of patients that have been treated all over the world. The development and tuning of such techniques on the single, individual cornea, seems to have been approached mainly by an experience-gaining and trial-and-error approach. Despite the massive use of refractive surgery, an appropriate biomechanical model of the cornea under physiological conditions and under the effect of surgical treatments seems still to be lacking. Such an approach appears to be necessary for a correct tuning of the techniques on both a general and an individual basis. This would help in reducing the risk of complete or partial failure that are still connected to the implementation of these removal techniques (that weaken the cornea from a structural point of view).

This work attempts a biomechanical modelling of refractive surgery. Efforts in this direction have been already produced by different authors, mainly with respect to well-established RK, but also to PRK [7, 2, 9, 4, 5]. Analytical and numerical (FEM) approaches are employed to estimate the change in dioptric power of the myopic cornea following RK and PRK under the assumption of linear elastic behaviour. The following specific aspects as described in the sequel have been considered. A comprehensive account of the study is given in [8].

Membrane/flexure behaviour of the cornea based on Shells Theory

The basic equations of Shells Theory [10] are solved to analyze the stress/deformation response of the cornea in physiological conditions under internal IOP (IntraOcular Pressure). The pure membrane regime is investigated for a vanishing constraint at the limbus interface between cornea and sclera (rollers). Flexure behaviour is instead taken into account by assuming perfect built-in constraints at the limbus. Results are presented in terms of: i) the analytical solution with 8 terms of the hypergeometric series; ii) Geckeler approximate analytical solution I; iii) Hetény approximate analytical solution II. An average human cornea is considered as a constant thickness spherical elastic shell with medium radius R_m=7.35 mm, thickness s=0.59 mm, half opening apex-to-limbus angle θ_c=48°, Young's modulus E=1 MPa, Poisson's ratio ν=0.49, IOP p=15 mmHg=2 kPa. Output is obtained in terms of both stress resultants and deformation. Fig. 1 reports the vertical displacement of the shell η as a function of the anomaly angle θ from the apex. Notice that the truncated exact solution remains valid to represent the apical zone near θ=0°. These results have been compared to FEM simulations of an average physiological cornea with variable thickness, with order-of-magnitude agreement on both static and kinematic output.

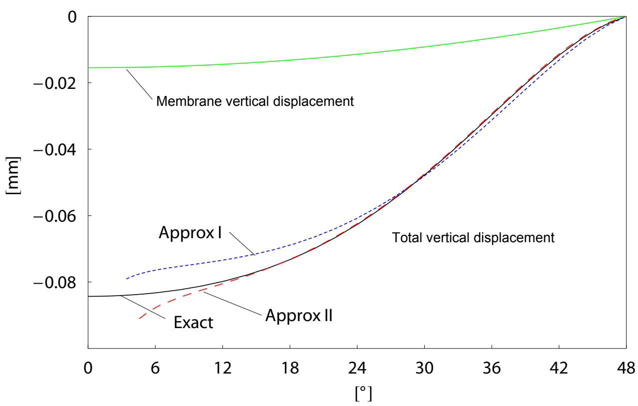

Figure 1: Vertical displacement of the corneal linear elastic shell under IOP according to Shells Theory.

FEM modelling of RK

A 3D model of revolution is assembled, with built-in conditions at the limbus. Four incisions are considered, by constraints removal, at 85% thickness depth of the cornea, preserved free optical zone of 4 mm diameter, length of 2.5 mm from that. Flattening of the central cornea and bulging of the peripheral regions are observed, trends in agreement with previous simulations [7, 4]. A FEM/CAD procedure has been developed to evaluate the local radii of curvature of the central cornea in pre- and post-operative conditions. This allows to evaluate the change in dioptric power of the cornea according to the formula

$$\Delta D = 337.5\ (1/R_f - 1/R_i)\ , \tag{1}$$

where R_i and R_f are the estimated initial and final radii of curvature in mm [3]. Values of -3.85 D and -3.81 D have been evaluated along the incision meridians and at 45° between them.

FEM modelling of PRK

The physiological cornea is considered with both a Katsube-type [5], step-shaped constant thickness and a Munnerlyn-type variable thickness ablation profile [1]. The two main parameters in the latter PRK procedure are the maximum ablation depth h at the apex and the diameter d of the ablation zone. Parametric axisymmetric FEM analyses have been performed for various h and d. Figs. 2–3 resume the prediction of the degree of correction in diopters. Fig. 4 compares predictions that can be made in various ways, with reference to the so-called Munnerlyn's formula [1], $h = -\Delta D d^2/3$, that accounts just for the geometrical reshaping of the external surface of the cornea after laser ablation. This formula is normally employed in defining the input parameters h, d of the surgical PRK treatment for a given desired ΔD. Our results turn out under-corrective. However, it has to be noted that no attempts where made in the study towards a quantitative prediction, with ad-hoc calibration of the parameters (for example to the Young's modulus E is given the nominal value 1 MPa). On the other hand, the analysis shows the importance of considering not only the pure geometrical reshaping of the cornea but also the consequent change of the biomechanical response of the weakened cornea, which is reflected by the discrepancies between curves 1-2 and 3-4 in Fig. 4.

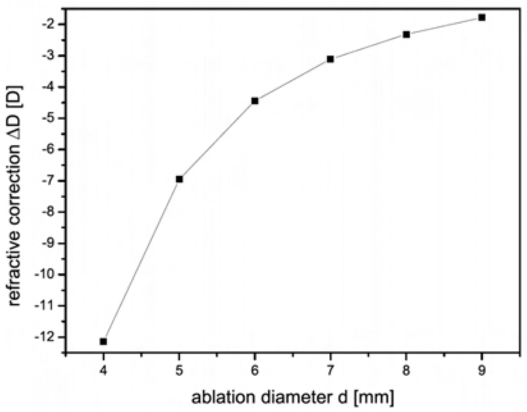

Figure 2: Degree of refractive correction ΔD after PRK as a function of ablation diameter d at constant maximum ablation depth (h=0.1 mm).

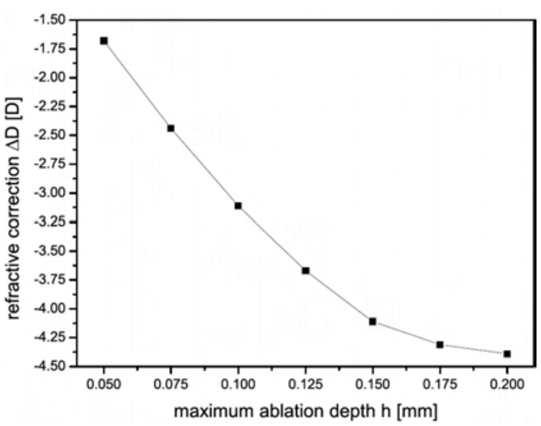

Figure 3: Degree of refractive correction ΔD after PRK as a function of maximum ablation depth h at constant ablation diameter (d=7 mm).

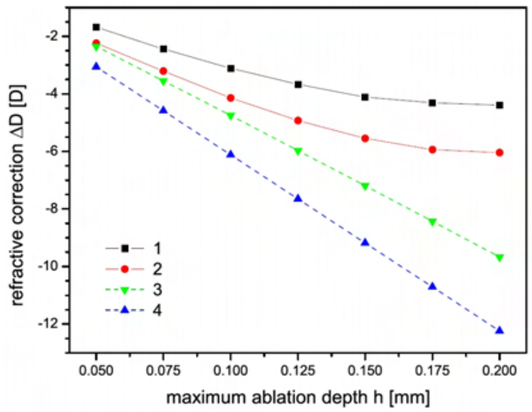

Figure 4: Idem as Fig. 3. Comparison of different estimates of pre- and post-operative shapes from: 1) deformed profiles; 2) same but through Munnerlyn's formula; 3) undeformed profiles; 4) pure Munnerlyn's formula on target ablation profile.

Conclusions

These preliminary results further support the need of a quantitative biomechanical modelling of refractive surgery, possibly even on an individual basis. This should complement the ophtalmologist's experience in defining the most appropriate treatment parameters and thus help in reducing the risk of failure.

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