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Biomechanical modelling of periodontal ligament behaviour under various mechanical loads

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In this review paper, some results of biomechanical modelling of periodontal ligament behaviour, which were obtained by the authors, are discussed. The periodontal ligament behaviour is emphasized to result significantly from a kind of mechanical load acting on teeth. It is primarily determined by features of the periodontal ligament structure which were experimentally revealed by the authors. Moreover, from unified biomechanical grounds some problems were formulated and solved in which the teeth were considered to be loaded with functional, traumatic and orthodontic loads.

Key words: dentofacial biomechanics, periodontal ligament, permeable deformable tissues, poroelastic analysis, periodontal fluid, tooth trauma, orthodontic treatment

1. Introduction

The periodontal ligament (PDL) is a dense connective tissue that surrounds the root of the tooth and attaches it to the alveolar bone. The PDL is one of the components of the masticatory apparatus and is of considerable importance for the functioning of adapto-compensating mechanisms of the masticatory system [11].

The PDL fulfils a broad spectrum of functions. The primary functions are to provide tooth support and blood supply. In literature, there is no disagreement as regards the role of the PDL in providing support for the teeth, but there is much controversy about the mechanism through which this function is accomplished. The precise sequence of events occurring in the PDL when force was applied to a tooth remained unknown [2].

Such a multifunction of the PDL depends upon its complex structure. The PDL consists mainly of collagen fibres, most of them being arranged into fibre bundles. Collagen fibres comprise 50% of the PDL by weight. Furthermore, the PDL also contains considerable amounts of cells, nerves, vasculature and interstitial fluid.

To explain the mechanism of tooth support by the PDL we have conducted experiments with guinea-pigs [5]. It was experimentally shown that under a short-term load the PDL may be considered as a porous solid (collagen fibres, walls of blood and lymphatic vessels) containing a fluid (interstitial fluid, blood, lymph). Moreover, the permeability coefficient k turned out to be of the order of 10^{-8} m^2/(Pa·sec).

One of the theories suitable for description of behaviour of such media is poroelastic theory proposed by BIOT [1]. The equations describing behaviour of elastic matrix filled with mobile fluid may be written in the following form:

- equilibrium of effective stresses

$$\vec{\nabla} \cdot \tilde{\tau} = 0 \qquad \forall \vec{r} \in V,$$

- Darcy's law

$$\frac{\partial \vec{w}}{\partial t} = -k\vec{\nabla} p \qquad \forall \vec{r} \in V,$$

- constitutive relation

$$\tilde{\tau} = 2\mu \tilde{e} + \lambda e \tilde{g} - \alpha p \tilde{g} \qquad \forall \vec{r} \in \overline{V},$$

- constitutive relation

$$\theta = \alpha e + \frac{p}{M} \qquad \forall \vec{r} \in \overline{V},$$

- geometric relation

$$\theta = -\vec{\nabla} \cdot \vec{w} \qquad \forall \vec{r} \in \overline{V},$$

- geometric relation

$$e = \vec{\nabla} \cdot \vec{u} \qquad \forall \vec{r} \in \overline{V},$$

- geometric relation

$$\tilde{e} = \frac{1}{2}\left(\vec{\nabla}\vec{u} + \vec{\nabla}\vec{u}^T\right) \qquad \forall \vec{r} \in \overline{V},$$

- boundary condition

$$\vec{n} \cdot \tilde{\tau} = \hat{\vec{t}} \qquad \forall \vec{r} \in S_t,$$

- boundary condition

$$\vec{u} = \hat{\vec{u}} \qquad \forall \vec{r} \in S_u,$$

- boundary condition

$$p = \hat{p} \qquad \forall\, \vec{r} \in S_p,$$

- boundary condition

$$\vec{n} \cdot \frac{\partial \vec{w}}{\partial t} = 0 \qquad \forall\, \vec{r} \in S_i,$$

- initial condition

$$\vec{u}(\vec{r}, t = 0) = 0 \qquad \forall \vec{r} \in \overline{V},$$

- initial condition

$$\theta(\vec{r}, t = 0) = 0 \qquad \forall \vec{r} \in \overline{V},$$

where p is the pore fluid pressure; θ is the fluid content; $\tilde{\varepsilon}$ is the strain tensor of the solid; $\vec{u}$ is the displacement vector of the solid; $\vec{w}$ is the relative fluid displacement vector; μ, e, α, M are material parameters; $\hat{\vec{t}}$, $\hat{\vec{u}}$, $\hat{p}$ are given functions of coordinates and time (surface forces on boundary S_t; solid displacements on S_u; fluid pressure on S_p, respectively); boundary S_i is impermeable; $S = S_t \cup S_u = S_p \cup S_i$ is the full surface; V is the interior volume; $\overline{V} = V \cup S$.

The poroelastic behaviour of the PDL was investigated in [7]. Consider a short-term load which leads to the translational movement of the tooth. In this case, a part of the PDL is subjected to compressive loads. The width of human PDL ranges from 0.1 to 0.3 mm. These values are far less than the length of the tooth root (the order of 15 mm) [8]. Because of that the porous medium was considered to be sandwiched between two approached parallel rigid impervious plates. An approximate analytical solution to the problem of stationary flow of the interstitial fluid completely saturating the incompressible porous medium was found. The fields of displacements, strains and stresses of the porous medium as well as fields of fluid pressure and fluid flow were obtained. Figure 1 shows the streamlines of the filtered fluid. The solution provides a qualitative description of some processes taking place in such porous tissues, and the PDL fluid movement in particular.

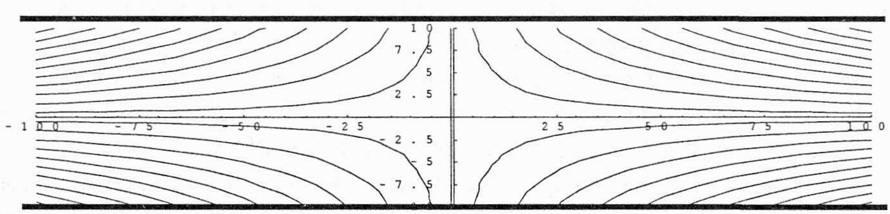

Fig. 1. Streamlines of the filtered fluid

2. Influence of different mechanical factors on the kind of tooth trauma

The PDL behaviour is very diverse and depends strongly on the kind of mechanical load acting on the tooth. Such a load may lead to the different tooth traumas, their kind being mainly determined by the degree of PDL fluid redistribution [6].

The traumas of teeth and parodontium are common in different age groups. In the wake of caries a tooth trauma is the second reason of its loss. In this connection, from the point of view of both theory and practice, it is important not only to study the clinical alterations occurring in traumatized teeth and parodontium under the influence of traumatic mechanical loads, but also to investigate biomechanical causes of these alterations. It should be emphasized that the upper central incisors are the teeth most often (95%) traumatized. This is due to the fact of the maxilla being motionless, as well as the anterior locations of those teeth and their not being protected by soft tissues. This turns out to be true for traumas of both deciduous and permanent teeth.

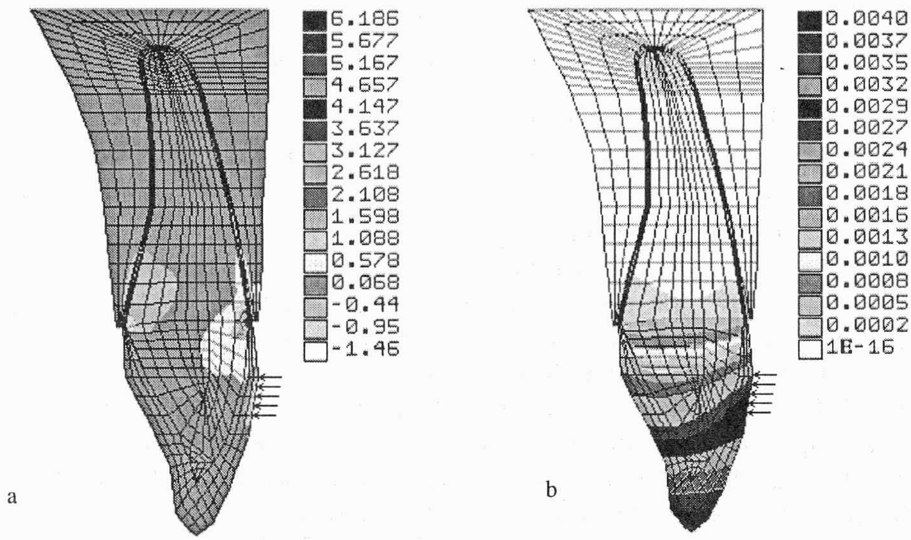

Fig. 2. A case of crack (fracture): a) maximum principal stresses, b) displacement magnitudes

It is known from clinical practice that a kind of tooth trauma depends on magnitude and direction of the force applied and point of its application, age of traumatized person, features of the structure and properties of their tooth and bone. We have investigated one more factor, namely, the duration of force application to the crown of an upper central incisor. It is the authors' opinion that duration influences the degree of redistribution of PDL fluid and thereby the degree of load amortization by the PDL.

If the load applied is decidedly a short-acting one, then the PDL fluid practically has no time to move and thus offers an additional resistance to tooth movement. An analysis of the foregoing poroelastic equations written for uniaxial loading shows that during this period the PDL behaves like an elastic medium characterized by a sharply increased Young's modulus. It followed from the two-dimensional finite element simulation of the three-dimensional problem that the maximum principal stresses were localized within the region of force application (figure 2a), whereas displacements of the points belonging to the PDL were slight (figure 2b). As a consequence, the probability of crack or even fracture occurring in the hard dental tissues (enamel, dentin) dramatically increased and yet injury of the soft tissues (pulp and PDL) was unlikely.

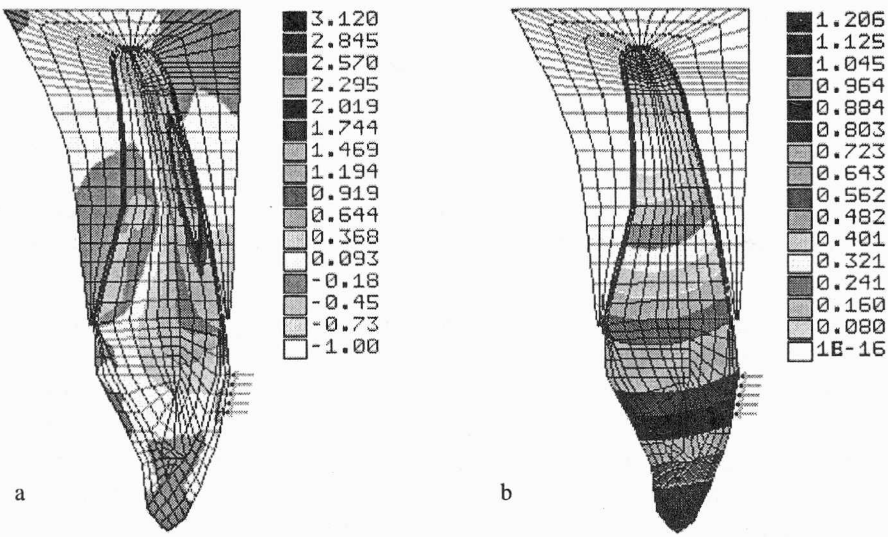

Fig. 3. A case of dislocation: a) maximum principal stresses, b) displacement magnitudes

If load acts over an extended period, then the PDL fluid redistributes inside and outside the PDL space, i.e., according to poroelastic theory, the PDL Young's modulus decreases. In this case the stresses were distributed more uniformly in the hard dental tissues and thus the probability of cracks arising in them was reduced (figure 3a). However displacements of the points belonging to the PDL were significant (figure 3b). This may cause injury of the soft tissues and rupture of the part of collagen fibres, blood and lymphatic vessels as well as neurovascular bundle.

3. Determination of optimal orthodontic forces

Under a long-term load the PDL may be viewed as an elastic material [8]. When a tooth is subjected to an action of orthodontic force it begins to move in its socket. Compressive areas and tensile areas appear in the PDL. It is well known that tensile stresses in the PDL stimulate apposition of the alveolus and compressive stresses cause alveolar resorption. This results in the remodelling of tooth socket and tooth movement (figure 4).

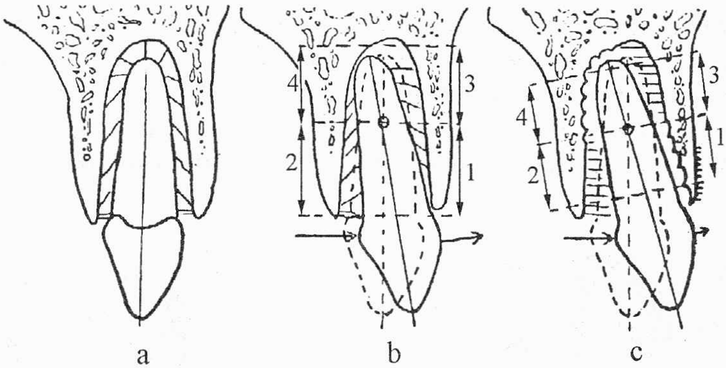

Fig. 4. Schematic representation of the tooth movement biomechanics [3]:
a) normal location of the tooth; b) tooth movement when force is applied, 1, 4 – PDL compressive areas, 2, 3 – PDL tensile areas, c) bone remodelling, 1, 4 – bone resorption areas, 2, 3 – bone apposition areas

Adequate magnitudes of orthodontic force are one of the decisive factors of a successful tooth movement [3].

It is believed that success of orthodontic treatment is mainly dependent on the magnitude of the PDL squeeze [4]. If stresses in compressive areas of the PDL are too high, the PDL is squeezed greatly. Therefore the blood supply stops and no resorption occurs in that area. If the compressive stresses are very small, the PDL squeezing is insignificant and bone remodelling proceeds very slowly. Thus certain force magnitudes are required to cause necessary tissue remodelling.

As SCHWARZ [5] stated: ... *biologically the most favourable treatment is that which works with forces not greater than the pressure in the blood capillaries. This pressure in humans ... is about 20 to 26 g for sq. cm surface* (2.0–2.6 kPa).

The behaviour of tissues around the PDL tensile areas does not depend very much on stress magnitude in the PDL. In any case there is alveolar apposition around the tensile region of the PDL.

Thus, favourable stress distributions in the PDL, especially in compressive areas, are the necessary condition during orthodontic treatment.

The purpose of this part of the study was to investigate the stress distributions in the PDL and then to determine optimal forces which should cause predesigned (controlled) orthodontic tooth movement and satisfy some biological restraints [4].

One of the stages in the process of solving this problem was the development of a three-dimensional finite element model of the upper right central incisor and its environment for calculation of stresses, especially within the PDL.

To model predesigned orthodontic tooth movement, exact mathematical definitions of the concepts of a "centre of resistance" and a "centre of rotation of a tooth" were introduced [9]. These concepts are widely used in the literature on orthodontic biomechanics, however they are not strictly defined. A centre of resistance and a centre of rotation are shown to exist only in particular cases. The theorems of their existence and uniqueness as well as the formulas for determining their positions were obtained. Among other things, these concepts make it possible to obtain something of the tooth movement even without solving the boundary-value problem.

4. Conclusions

The PDL behaviour results significantly from a kind of mechanical load acting on teeth. Since literature data relative to mechanisms of reaction of the PDL to mechanical loads were contradictory the experiments on guinea-pig's mandible samples have been performed.

It was experimentally shown that the PDL structure allowed the PDL fluid to move under pressure gradient within the PDL. From the results obtained it may be concluded that a part of fluid constituent of the PDL must be free under certain conditions.

Moreover, the experiment provided for estimation of the permeability coefficient of the PDL. It proved to be of the order of 10^{-8} $m^2/(Pa{\cdot}sec)$.

We think that consideration of the PDL as a porous solid containing a free fluid might help to furnish insights into the mechanism and nature of providing tooth support by the PDL under load as well as other functions of the PDL.

That is why the authors use the Biot's poroelastic theory to model the PDL. As an example the behaviour of the interstitial fluid saturating and flowing through a deformable porous matrix was considered for the case of unconfined compression of such a porous medium.

Further, as a result of analyzing the Biot's poroelastic equations and finite element modelling, it was shown that reaction of the PDL was considerably influenced by the kind of tooth trauma.

Moreover, the optimization problem of finding appropriate orthodontic forces was formulated and solved. The appropriate forces were assumed to be those which cause predesigned orthodontic tooth movement and satisfy some biological restraints.

To achieve predesigned orthodontic movement of a tooth, exact mathematical definitions of the concepts of a "centre of resistance" and a "centre of rotation of a tooth" were introduced and several corresponding theorems were proved.

The present biomechanical model of the PDL realistically described familiar clinical effects occurring in the PDL and around it under different mechanical loads acting on the teeth.

Improvements of the model with respect to a more realistic representation of the PDL are intended for the future.

Acknowledgements

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References

[1] BIOT M.A., *Mechanics of deformation and acoustic propagation in porous media*, Journal of Applied Physics, 1962, 33, 1482–1498.

[2] CAPUTO A.A., STANDLEE J.P., *Biomechanics in clinical dentistry. Chapter 3. The periodontium*, Quintessence Publishing Co., 1987.

[3] KALVELIS D.A., *Biomorphologic fundamentals of the orthodontic treatment* (in Russian), 1961, Riga.

[4] NYASHIN M.Y., PECHENOV V.S., RAMMERSTORFER F.G., *Determination of optimal orthodontic forces*, Russian Journal of Biomechanics, 1997, 1 (1–2), 84–96.

[5] NYASHIN M.Y., OSIPOV A.P., BOLOTOVA M.Ph., NYASHIN Y.I., SIMANOVSKAYA E.Y., *Periodontal ligament may be viewed as a porous material filled by free fluid, experimental proof*, Russian Journal of Biomechanics, 1999, 3 (1), 89–95.

[6] NYASHIN M.Y., BAGAUTDINOVA I.V., SIMANOVSKAYA E.Y., CHERNOPAZOV S.A., *Biomechanical investigation of a trauma of the upper central incisor*, Russian Journal of Biomechanics, 1999, 3 (1), 96–100.

[7] NYASHIN M.Y., *Unconfined compression of the periodontal ligament, intervertebral disc, articular cartilage and other permeable deformable tissues, a poroelastic analysis*, Russian Journal of Biomechanics, 1999, 3 (3), 23–31.

[8] NYASHIN M.Y., *Mathematical model of the periodontal ligament* (in Russian), Ph.D. Thesis, Perm State Technical University, 1999.

[9] OSIPENKO M.A., NYASHIN M.Y., NYASHIN Y.I., *Center of resistance and center of rotation of a tooth, the definitions, conditions of existence, properties*, Russian Journal of Biomechanics, 1999, 3 (1), 5–15.

[10] SCHWARZ A.M., *Tissue changes incident to orthodontic tooth movement*, International Journal of Orthodontics, 1932, 18, 331–352.

[11] SIMANOVSKAYA E.Y., BOLOTOVA M.Ph., NYASHIN Y.I., NYASHIN M.Y., *Functional adapto-compensating mechanisms of the masticatory apparatus as a special biomechanical system*, Russian Journal of Biomechanics, 1999, 3 (3), 3–11.