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THE EXPERIMENTAL INVESTIGATION AND MODELLING OF THE VISCOELASTIC PROPERTIES OF MANDIBLE BONE TISSUE

Summary of Doctoral Thesis

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The defence of the thesis submitted for doctoral degree of engineering science will take place at an open session in Kaļķu Street 1/119 Riga Technical University on December 02, 2010 at 16.00.

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APPROVAL

I confirm that I have developed this thesis submitted for the doctoral degree at Riga Technical University. This thesis has not been submitted for the doctoral degree in any other university.

Zoja Veide ……………………………….. (signature)

Date:……………………………………..

The doctoral thesis is written in Latvian and includes introduction, 5 sections, conclusions, 63 figures and 16 tables in the main text, 142 pages. The bibliography contains 122 references.

CONTENTS

INTRODUCTION

Due to the continued interest of natural biomaterials, the various specialists of high-tech industry pay a lot of attention to investigation of mechanical properties of biocomposite – compact bone tissue. Rational structure of bone and an effective resistance of bone tissue to mechanical loads can be used as a prototype for development of new composite materials that become common use in the national economy as well as in medicinal practice.

The first investigations of compact bone mechanical properties appear already in the nineteenth century. Despite the fact that mechanical properties of compact bone tissue have been investigated for long time, time-dependent behaviour of this material is not sufficiently explored. Particularly the biomechanical properties of mandibles bone tissue are not adequately investigated.

Motivation of the research

Due to the fact that mandible is subjected to various activities related to fracture healing, malocclusion treatment or tooth replacement, in jaw bone compact part the deformations and stresses arise from the long-term or cyclic loading action. Adequate treatment choices require knowing the consequences of a given stresses.

During normal activities mandibular bone is exposed to cyclic loading, which may compromise the initial fixation of the implant, lead to bone tissue resorption and loss of contact between implant and bone tissue. The evaluation of compact bone tissue behaviour under cyclic loading is necessary for the analysis of the efficiency and reliability of the shape and dimensions of endosseous implants.

The success of biomaterials in the body depends on factors such as the material properties, design, and biocompatibility of the material used. Main biocompatibility characteristic includes adequate mechanical properties such as strength, stiffness, viscoelastic and fatigue properties. It produces the necessity for the investigation of biomechanical properties of bone tissue under constant long-term and cyclic loading, because incorrect choice of implant mechanical properties may induce additional stress concentration in material/bone tissue structure.

For numerical techniques, in particular, for finite element method, which can be used for the investigation of biomechanical problems in the dental area, is necessary experimental data concerning jaw bone mechanical properties. The lack of specific experimental data of the human jaw bone tissue prevents the implementation of these properties into finite element simulation which results in poor quality modelling.

The subject of the thesis is mandibles compact bone tissue. Investigations are related to time-dependent biomechanical properties – active creep and bone tissue behaviour under cyclic load. Used compression load often arise in real life during normal activities. Cyclic deformation occurs under different stress levels at the frequency that arises from physiological activity of organism.

The finite element method is widely accepted as one of the most practical and reliable methods for analyzing mechanical behaviour of biological objects. Modelling obviates the need for experimentation, which inconvenient, costly and difficult to apply in certain situations. In this thesis the finite element model of human fractured mandible was developed with the aim to evaluate the fracture risk of fixation system and bone tissue resorption possibility.

The goal of the thesis

The aim of current thesis was to investigate the active creep and fatigue behaviour of the mandible compact bone material and to develop a three-dimensional finite element model of a human jaw and perform numerical studies.

The tasks of the thesis

In order to achieve the goal of the thesis the following tasks were specified:

- the experimental determination of viscoelastic characteristics of human mandible compact bone material;
- development of mathematical model of compact bone tissue behaviour in creep deformation mode;
- the investigation of jaw compact bone tissue behaviour under cyclic load;
- development of mandible compact bone material fatigue life prediction model;

- development of mandible finite element model.

Scientific novelty of the thesis

In this thesis is investigated mandible – part of bones system that is not adequately explored from the mechanical properties point of view. Research concerning time-dependent biomechanical properties of mandible compact bone tissue – creep and fatigue – for which is no information in the available literature.

In this thesis is developed mathematical model of active creep of human mandible compact bone tissue.

In studies under cyclical load the stress level of each sample is determined using a developed methodology based on the experimentally observed correlation density-yield stress.

In this thesis pig jaw compact bone material behaviour is investigated under cyclical load and developed mathematical model of fatigue strength taking into account specimens stressed volume and probability of failure.

Practical value of the thesis

Obtained results may be used as data base for numerical approach such as finite element method in order to simulate bone tissue response to orthodontic care, surgical manipulation and attachment of implant to bone.

Determined bone tissue properties can be used for development of new biomaterials, because properties of biomaterials should be as close as possible to the biological tissue properties.

Determined bone tissue behaviour characterisations can be applied in orthodontics or maxilla-facial surgery for treatment choices or development of dental corrective devices or jaw fixation plates, because long-term static or cyclic loads act on mandibles bone tissue during normal activities and bone response to it is very important.

Obtained results are necessary for evaluation of efficiency and reliability of the shape and dimensions of dental implants. The assessment of stress and deformation that induce cyclic loading is an essential aspect for prediction of potential risk of implant failure.

Finite element modelling results can be used in facial surgery for evaluation of new fixation system properties and development of individual geometry and shape of fixation plate. These results could provide a good baseline model for modelling time-dependent properties of material.

Structure of the thesis

The thesis includes introduction, 5 chapters, conclusions, bibliography, 63 figures and 16 tables in the main text, 142 pages. The bibliography contains 122 references.

In the introduction the motivation of the thesis, research goals and tasks are defined. Novelty and practical value of the thesis are described as well.

The Chapter 1 includes a literature review about compact bone mechanical properties. The architecture of compact bone and mandible, effect of some factors on the mechanical properties of bone, as well as the viscoelastic properties of bone, bone behaviour under cyclic load and fatigue damage models are described in this chapter. Analysis of different researches results carries inference on necessity of investigations of lower jaw compact bone tissue time-dependent properties.

The Chapter 2 describes the testing methods. First part of second chapter contains motivation of testing material selection and activities related to the preparation of the samples. The second part is devoted to the description of developed methods for testing of bone tissue viscoelastic properties and behaviour under cyclic load, as well as the experimental equipments.

The main attention in Chapter 3 is concentrated on analysis of results of the long-term static compressive load experiments. Creep deformation data were approximated using an exponential function and a mathematical model for creep was developed.

In the Chapter 4 the results related to the behaviour of bone under cyclic loads are analyzed, as well as carried out the prediction of compact bone tissue failure and developed the model of fatigue strength.

In the Chapter 5 the finite element model of mandibles fracture and its fixation system is developed. Model was estimated taking into account bite force values and fracture healing stages and there was given recommendation about fixation system improvement. At the end of the work the main conclusions are presented.

Approbation of the obtained results

The obtained results of research were presented on 7 international scientific conferences.

1. RTU 46st International Scientific Conference. Z. Veide, M. Dobelis, I. Knēts, J. Laizāns, V. Vītiņš. Cilvēka žokļa kompakto kaulaudu šļūde. October 13–15, 2005, Riga.
2. The 9th International Conference „Biomedical engineering”. Z. Veide, M. Dobelis, J. Laizāns, V. Vītiņš. Creep mathematical model development of human jaw compact bone. October 27-28, 2005, Kaunas, Lithuania.
3. The 7th International Symposium on Computer Methods in Biomechanics and Biomechanical Engineering CMBBE 2006. Z. Veide, M. Dobelis, I. Knēts, J. Laizāns, V. Vītiņš, Viscoelastic properties of the human mandible for finite element simulation. March 22–25, 2006, France.
4. The 7th International Symposium on Computer Methods in Biomechanics and Biomechanical Engineering CMBBE 2006. Z. Veide, M. Dobelis, I. Knēts, V. Vītiņš, Biomechanical peculiarities of physiologically wet compact bone in cyclic three point bending. March 22–25, 2006, France.
5. The International Conference „Biomechanics 2006”. Z. Veide, M. Dobelis, I. Knēts, J. Laizāns, V. Vītiņš. Compression creep properties of compact bone tissue. September 6–8, 2006, Zakopane, Poland.
6. The International Conference of the Polish Society of Biomechanics „Biomechanics 2010”. Z. Veide, O. Ozoliņš, M. Dobelis, I. Knēts. The fatigue properties of the pig mandible compact bone tissue. August 25–28, 2010, Warsaw, Poland.
7. RTU 51st International Scientific Conference. Z. Veide, O. Ozoliņš, M. Dobelis, I. Knēts. Žokļa kompakto kaulaudu spiedes noguruma uzvedība. Oktober 14–16, 2010, Riga.
8. RTU 51st International Scientific Conference. Z. Veide, M. Dobelis, I. Knēts, J. Laizāns. Mandibulas lūzuma fiksējošās sistēmas galīgo elementu analīze. Oktober 14–16, 2010, Riga.
9. The 14th International Biomedical Engineering Conference. Z. Veide, O. Ozoliņš, M. Dobelis, I. Knēts. Experimental Investigation of the Pig Mandible Fatigue Behaviour. October 28-29, 2010, Kaunas, Lithuania.

Publications

The main results of the thesis are reflected in 9 papers (7 of them have been already published and 2 have been accepted for publication):

Z. Veide, M. Dobelis, I. Knēts, J. Laizāns, V. Vītiņš. Cilvēka žokļa kompakto kaulaudu šļūde. RTU zinātniskie raksti. 1. sēr., Materiālzinātne un lietišķā ķīmija, 10. sēj., 2005, 76-84 lpp.

Z. Veide, M. Dobelis, J. Laizāns, V. Vītiņš. Creep mathematical model development of human jaw compact bone. Proceedings of the 9th International Conference BIOMEDICAL ENGINEERING, October 27-28, Kaunas, Lithuania, Technologija, Kaunas, 2005, pp. 17-22.

Z. Veide, M. Dobelis, I. Knēts, J. Laizāns, V. Vītiņš, Viscoelastic properties of the human mandible for finite element simulation. Proceedings of 7th International Symposium on Computer Methods in Biomechanics and Biomechanical Engineering CMBBE 2006, March 22–25, France, 2006, 6 p.

Z. Veide, M. Dobelis, I. Knēts, V. Vītiņš, Biomechanical peculiarities of physiologically wet compact bone in cyclic three point bending. Proceedings of 7th International Symposium on Computer Methods in Biomechanics and Biomechanical Engineering CMBBE 2006, March 22–25, France, 2006, 7 p.

Z. Veide, M. Dobelis, I. Knēts, J. Laizāns, V. Vītiņš. Compression creep properties of compact bone tissue. Zeszyty Naukowe Katedry Mechaniki Stosowanej, nr 26/2006, The International Conference „Biomechanics 2006" September 6–8, Zakopane, Poland, 2006, pp. 371-376.

Z. Veide, O. Ozoliņš, M. Dobelis, I. Knēts. The fatigue properties of the pig mandible compact bone tissue. Book of abstracts of the International Conference „Biomechanics 2010" August 25–28, Warsaw, Poland, 2010, pp. 243-245.

Z. Veide, O. Ozoliņš, M. Dobelis, I. Knēts. Žokļa kompakto kaulaudu spiedes noguruma uzvedība. RTU zinātniskie raksti. Materiālzinātne un lietišķā ķīmija, 2010, (6 p., accepted).

Z. Veide, M. Dobelis, I. Knēts, J. Laizāns. Mandibulas lūzuma fiksējošās sistēmas galīgo elementu analīze. RTU zinātniskie raksti. Materiālzinātne un lietišķā ķīmija, 2010, (6 p., accepted).

Z. Veide, O. Ozoliņš, M. Dobelis, I. Knēts. Experimental Investigation of the Pig Mandible Fatigue Behaviour. Proceedings of the 14th International Biomedical Engineering Conference, October 28-29, Kaunas, Lithuania, Technologija, Kaunas 2010, pp. 202-206..

1. BONE TISSUE BEHAVIOUR UNDER STATIC LONG-TIME AND CYCLIC LOADING (REVIEW OF LITERATURE)

Being a biological tissue, bone has a hierarchical arrangement of its primary constituents, such as carbonated apatite and type I collagen [Valenta J. et all 1993]. The complexity of bone structure gives rise to numerous toughening mechanisms and tissue strength, on the other side, bone architecture is the mechanistic cause for its multifactorial failure involving all bone constituents and each hierarchical level of organization of the tissue. Compact bone tissue is non-linear material and its mechanical properties depend upon different mechanical and biological factors.

The analysis of the results obtained by various researchers showed that data on relationships of mechanical properties of compact bone remains poor and is highly variable between subjects due to combination of the following reasons: anisotropic nature of the bone, samples location in the bone, precision of the dimensions and geometric accuracy of the shape of the test specimens, samples surface condition, drying in process of the experiment, etc.

Bone exhibits viscoelastic behaviour, i.e. the stress depends not only on the strain but also on the time history of the strain [Natali A.N. et all 2003]. Such behaviour can manifest itself as creep, which is a gradual increase in strain under constant stress; stress relaxation, which is a gradual decrease in stress in a specimen held at constant strain; load-rate dependence of the stiffness [Lakes R. S. et. all 1974, 1979]; attenuation of sonic or ultrasonic waves; or energy dissipation in bone loaded dynamically [Saha S. et. all 1977]. Experimental rheology modalities based on each of the above phenomena have been used in the study of bone. The results have been converted to a common representation via the interrelationships inherent in the linear theory of viscoelasticity, to permit a direct comparison of results. In the case of tension / compression, there is very significant disagreement among the published results. This disagreement may result from nonlinear viscoelastic behaviour not accounted for in the transformation process, or from experimental artefacts. In the case of shear deformation, however, there is good agreement between results obtained in different kinds of experiments [Lakes R. S. et. all 1979, 1980; Yang J. et. all 1981].

Behaviour of bone tissue may be considered as a linear viscoelastic up to 0.5 from ultimate stress but after this stress level – as a non-linear viscoelastic solid [Knets I. 2002]. The creep properties depend significantly upon the conditions of the preservation of specimen and testing. The creep strain increased sharply in samples that were kept in the physiological solution and during testing were under moist conditions. The creep properties of human bone tissue depend upon the age: in young bone tissue the viscoelastic behaviour is more expressed than in the old one. The increased creep of young bone tissue may be explained by a low degree of mineralization and, consequently, by a relatively large content of collagen. The increase of crystal size and its distinctive orientation with age causes the decrease of the active surface of mineral component and, therefore, may influence the diffusion rate and the intensity of ion exchange. The bone becomes chemically less active and the processes of resorption prevail over the processes of bone formation. This could be one of the reasons why the heterogeneity of creep properties over the zones of cross-section decreases with age.

Viscoelastic damping in bone exhibits a broad minimum at frequencies 1 to 100 Hz [Lakes R. S. 1982]. These frequencies are associated with normal activities. A shock-absorbing role for bone has been suggested for bone based on its viscoelastic response. However, the observed minimum in damping is inconsistent with such an interpretation. The viscoelasticity in bone may instead be a side effect of constituents in bone. For example, the cement lines give rise to toughness of bone by virtue of their compliance and viscosity and also give rise to viscoelasticity. Substantial damping is observed at low frequency and substantial creep is observed at long time. The physical cause is associated with interfaces such as the cement lines, but the biological significance is not known.

The existence of microcracks in bone tissue is a well-established fact [Cowin C. 2001]. Damage accumulation is a normal response of composite materials to mechanical loading and it is this damageability that contributes to the superior fatigue resistance and toughness of composite materials. In bone, the presence of damage also has a biological consequence. Since microcracks exist at some volume in normal, healthy bone, they may play a role in the "normal" turnover process as well as adaptive behaviour of bone. It is clear that the degradation of any particular property depends on how the damage state interrupts a particular structure–function relationship. Since bone stiffness, strength, and creep exhibit different structure–function relationships, it is not surprising that a particular damage state will affect these properties differently. In this context, expanding the repertoire of damage and degradation measures is crucial to achieving a more thorough understanding of bone as a damaging composite material.

One traditional method for quantitatively characterizing the fatigue behaviour of a bone is by measuring the total cycles to failure, N_f, as a function of the alternating stress amplitude (σ_a) [Kruzic J.J. et. all 2006]. This is termed the stress-life or "S/N" approach. In general, mineralized tissues display S/N curves similar to ductile metals, with N_f increasing with decreasing σ_a. Unfortunately S/N data for all materials tend to be distorted by excessive scatter due to variations in factors such as surface condition or flaw distribution, which can have a large effect on the crack initiation portion of the fatigue life. Test conditions – higher frequencies and room temperature – give higher fatigue-cycle lifetimes [Carter D.K. et. all 1976, Caler W.E. et. all 1989].

Stress fractures that occur in bones as a result of repetitive, cyclic loading are a significant clinical problem [Iwamoto J. 2003, Meurman K.O. 1980]. A better understanding of the mechanisms of failure may lead to improved clinical evaluation and a more advanced design of bone implants. This understanding would be most useful if it leads to an accurate method for predicting fatigue failure, either as a function of load or some other measurable parameter.

Nonetheless, failure mechanisms are still not fully understood, only with suggestions ranging from microcracks formation at the lamellar interface to heterogeneity of local tissue modulus affecting the coalescence of microcracks into macrocracks [Jepsen K.J. et. all 1999; Zioupos P. et. all 2007].

The bone fatigue damage models can be used to predict the inelastic behaviour and the damage accumulation processes under conditions of overload and fatigue. More importantly, they can be used in conjunction with experimental studies to refine descriptions of these behaviours. However, there are numerous specifics about the effects of loading direction, loading mode, loading rate, etc. that remain to be explored[Cowin C. 2001]. Given the complex structure and complex mechanical behaviour of bone, damage model development will require substantial accompanying experimental research in damage behaviour.

The analysis of the results from the literature can conclude that it is not resolved theoretically and practically important questions:

- non-linear compact bone viscoelastic behaviour is a insufficiently explored;
- bone tissue time-dependent mechanical properties, especially creep properties, are not adequately investigated;

- there is not enough data on mechanical properties of human jaw compact bone tissue, in particular, on viscoelastic properties;
- the heterogeneity of mandibular compact bone tissue creep properties are unknown;
- there is little information about jaw compact bone tissue behaviour under cyclic loading;
- mandibular compact bone fatigue properties are not sufficiently explored.

2. BONE TISSUE TESTING METHODS

The compact part of human mandible was used for the determination of viscoelastic properties and pig lower jaw was used for testing under cyclic loading. Pig compact bone tissue shows microarchitecture, bone physiology and biomechanical properties similar to humans, especially with regard to intracortical bone remodelling.

2.1 Materials and method for viscoelastic properties determination

Samples for the experiments were prepared from five 49 to 56 years old male lower jaws obtained in autopsy. After the sectioning of the mandible from the premolar and molar regions compact bone specimens of rectangular prismatic shape with approximate dimensions (4 – 12) × (4 – 8) × (4 – 8) mm were prepared. The total number of samples successively tested was 15. The samples were stored in 10% formalin buffer solution at 2 – 4°C temperatures before the experiment.

The viscoelastic properties of jaw compact bones samples were experimentally determined in creep test in specially designed equipment. At each discrete measurement step data of applied loads and displacements were recorder and stored in PC using Lawson Labs. Inc. USA CVI Model 201 A/D (analogue–digital) converter and Lawson Labs. data storage software. The collected load and displacement experimental data in PC were normalized with respect to the dimensions of the specimen to obtain the material properties.

Mandibular bone material time dependence was investigated under constant load compression assuming that at low load levels it corresponds to constant stress. Compression load was applied to the specimen perpendicular to the occlusal plane by means of leverage system and the displacement between loading plates was recorded at the predetermined intervals of time (three measurements per second). Mechanical experiments were conducted in a physiologically wet medium and at 20±2°C temperature. Control experiments determined that the required registration time of creep under the given stresses is approximately 300 minutes. All creep curves were recorded in a stress range from 5 to 40.8 MPa and in a time range from 60 to 268 minutes.

2.2 Materials and method for bone tissue behaviour determination under cyclic load

Pig mandibles compact bone tissue layer quality, thickness and density in different cross-sectional areas of the jaw and its dependence on the cross section location in the bone were assessed using the computerized tomography (CT). It was found that the structure of compact bone tissue of mandible's external – buccal part is most homogeneous compared with jaw lingual and inferior parts. The maximum thickness of compact bone tissue layer was determined in pig mandibles buccal part. The bone density measurements were recorded in Hounsfield units. The maximum density of bone was determined in the upper and middle zones of mandibles anterior buccal part molar and premolar regions. Consequently, for samples preparation the most suitable areas of pig lower jaw – molar and premolar region of anterior buccal part were used.

Three fresh two years old pig mandibles were used for this study. All soft tissue was removed and the mandibles were stored frozen at -18°C in air-tight containers until specimens harvesting. Each lower jaws premolar and molar areas left and right sides were divided using a saw into sections with thickness 10 mm perpendicular to the occlusal plane. After the sectioning of the mandibles harvesting sites have been sliced in order to obtain compact bone specimens. Because of cutting the irregular surface of the jaw, specimens had rectangular prismatic shape with dimensions 3x5x7 mm ± 0.5 mm. All jaw compact bone specimens were visually inspected and samples with machining defects were excluded from the test group. Twenty six specimens were prepared to determine the ultimate compressive stresses, to find regression between density and yield strength and 30 specimens were prepared for fatigue tests.

During the entire process of preparation, harvesting, and testing, the test specimens were kept moist, since purpose of this study was to investigate the properties of wet bone. All tests were conducted at room temperature (approx. 20°C).

Compressive fatigue tests and static experiments were carried out in an INSTRON 8872 servo-hydraulic testing machine equipped with a water bath with transparent acrylic walls. During testing the samples were immersed in water and placed to a specially designed holding tool. Before mechanical experiments for all specimens the density was measured using hydrostatic weighing method.

At the first stage of investigation twenty six specimens were tested under static compression. Specimens were loaded to fracture at a deformation rate of 1 mm/sec. Compression load was applied to the specimen perpendicular to the occlusal plane. For each specimen there was determined the yield strength and recorded ultimate compressive stress. The regression analysis between density and yield strength revealed theoretical relationship:

$$\sigma_y = 228{,}73\rho - 377{,}74 , \qquad (2.1)$$

where σ_y – yield strength, MPa, ρ – density, g/cm^3 , r^2 = 0,922.

The theoretical yield strength was calculated for fatigue test specimens using regression equation obtained in previous step and then the level of loading for 30 specimens was defined.

The fatigue properties of mandibular bone material were investigated under compression cyclic loading. Tests were conducted under stress control. A sinusoidal load waveform was applied to the specimen using the servo-hydraulic testing machine. The frequency of loading was 2 Hz, within a frequency range characteristic for load histories during normal physiologic activities. Load levels that were 60%, 70%, 80%, and 90% of the compressive yield strength were tested with specimens that were randomly allocated to the four test groups. Loading was used in this study with maximum cyclic stress – from 0.5 to 1 MPa, but minimum stress value for each specimen was calculated using selected load level and theoretical yield strength. The cyclic stress range was defined as the difference between maximum and minimum stress.

The number of cycles to failure was recorded: failure was defined in terms of an increase of more than 10% in the cyclic deflection range (i.e. 10% loss of stiffness). In practice this always coincided with the presence of at least one large crack, typically oriented at an angle of 45° to the specimen axis.

3. COMPACT BONE TISSUE LONG TIME STATIC COMPRESSION TESTS

From compression test of jaw compact bone tissue samples typical curves of compression strain changes in time were obtained. It is characteristic that different creep curves for individual test samples were obtained under given compressive stress. This fact substantiates that the material properties depend upon the stress values.

It was determined that compact bone exhibit linear creep behaviour in the range $\sigma_{11} \leq 0.5\sigma_{11}^{*}$ or $\sigma_{11} < 0.28\sigma_{11}^{*}$, if the loading was carried out along the bone longitudinal axis, where σ_{11}^{*} is ultimate stress along the longitudinal axis of bone. The ultimate compressive strength of the human mandible compact bone tissue is reported 127 MPa. The material exhibited linearly elastic creep behaviour within the stress range (5 – 40.8 MPa) used in these studies. The linearly elastic creep of the human jaw compact bone material under constant stress was described by exponential function. The creep strain of the human jaw bone tissue in the direction of the occlusal plane may be presented in the following form (3.1):

$$\varepsilon = C\sigma \frac{1}{n}\sum_{i=1}^{n}\left[1-\exp(-\alpha_i t)\right], \qquad (3.1)$$

where C is material compliance – the ratio of the creep strain at $t \to \infty$ to stress σ ;

α_i is inverse of relaxation time; n is a number of relaxation times, i. e. a number of exponents. In this study experimental data were approximated by two and three exponents.

The approximation of experimental data was carried out by determination of the minimum of aim function (3.2)

$$\Phi = \sum_{j=1}^{m}\left[\frac{\varepsilon_j^{(t)} - \varepsilon_j^{(e)}}{\varepsilon_j^{(e)}}\right]^2, \qquad (3.2)$$

where m is a number of common points along which the approximation was realized;

$\varepsilon_j^{(t)}$ and $\varepsilon_j^{(e)}$ theoretical and experimental values of creep strain at time t_j , respectively.

The approximation results with two and three exponents in a time range from 60 to 268 minutes and in a stress range from 5 to 40.8 MPa for each sample separately are shown in Tables 3.1 and 3.2.

The analysis of approximations results showed that S_y decreases on an average by 30 per cent in case of using three exponents. From 15 separately approximated specimens the samples with creep strain ranging from 0.05 to 0.3 % were selected and an attempt to approximate experimental data for 9 specimens group was made. For the estimation of the approximation results a Romanovsky's criterion was used which provides one of the most efficient ways of homogeneity analysis of experimental distribution for small number of samples. In accordance with this analysis of S_y from 9 specimens group were excluded another 4 specimens.

After repeated approximation of the whole group of 5 specimens with two and three exponents, good average results were achieved. Because of this we can conclude that obtained general results represent creep properties of human jaw bone tissue.

Table 3.1

The creep properties for individual specimens of human jaw compact bone tissue. Approximation with two exponents. S_y – standard deviation of arithmetic mean between theoretical and experimental curves.

Sample No.	C, [MPa^{-1}]	$\alpha_{1,}$ [min^{-1}]	α_2 [min^{-1}]	S_y
1	0.00060	0.09149	0.01556	0.00501
2	0.02911	0.47505	0.03366	0.00512
3	0.02739	0.01101	0.20536	0.00242
4	0.00741	0.20455	0.01850	0.00146
5	0.02306	0.63904	0.00823	0.00083
6	0.00956	0.91439	0.00661	0.00151
7	0.02997	0.32206	0.01505	0.00127
8	0.05097	0.01117	0.39552	0.00164
9	0.02111	0.63713	0.02298	0.00215
10	0.00662	1.22314	0.23036	0.00168
11	0.00929	0.15662	0.01259	0.00264
12	0.00772	0.06993	0.00731	0.00204
13	0.01358	0.32849	0.01085	0.00096
14	0.03613	0.93289	0.01979	0.00110
15	0.01631	0.46661	0.01163	0.00072

Table 3.2

The creep properties qualifications of human jaw compact bone tissue from approximation by three exponents.

Sample No.	C, [MPa^{-1}]	$\alpha_{1,}$ [min^{-1}]	$\alpha_{2,}$ [min^{-1}]	$\alpha_{3,}$ [min^{-1}]	S_y
1	0.00066	0.13120	0.01617	0.01617	0.00419
2	0.02932	1.09966	0.04522	0.04522	0.00212
3	0.03094	0.01028	0.43689	0.01027	0.00076
4	0.00729	0.43058	0.01930	0.03961	0.00097
5	0.02297	0.00515	1.54210	0.05320	0.00095
6	0.00809	0.16762	2.03817	0.00911	0.00203
7	0.03627	0.49539	0.00407	0.02484	0.00048
8	0.05684	0.00446	0.90680	0.02619	0.00128
9	0.03175	0.65433	0.00879	0.00878	0.00151
10	0.00662	0.90730	0.16656	0.90674	0.00168
11	0.00970	0.34627	0.01513	0.01513	0.00129
12	0.00803	0.12650	0.00886	0.00886	0.00125
13	0.01878	0.38260	0.00067	0.01382	0.00091
14	0.04086	1.37974	0.00487	0.09416	0.00055
15	0.02141	0.52901	0.00050	0.01418	0.00068

The dispersion of approximation results of excluded specimens can be due to combination of the following reasons: anisotropic nature of the bone, condition of the jaw preservation, sample's location in the bone, precision of the dimensions and geometric accuracy of the shape of the test specimens, samples surface condition, drying in process of the experiment, etc.

The approximations results at general creep equation constants by two and three exponents in a stress range from 10.4 to 20.48 MPa for group of specimens number 4, 6, 10, 11, 12 are shown in Table 3.3 and in Figure 3.1.

Table 3.3

Average values of the creep properties of human jaw compact bone tissue for 5 specimens group from approximation by two and three exponents.

Number of exponents	C, [MPa^{-1}]	$\alpha_{1,}$ [min^{-1}]	$\alpha_{2,}$ [min^{-1}]	$\alpha_{3,}$ [min^{-1}]	S_y
2	0.007624	0.151668	0.014158	-	0.008017
3	0.007605	0.258850	0.011936	0.035335	0.007965

From the results in table 1 follows that S_y varies negligibly, e.d. the value decreases only by 0.65 % in case of using three exponents in the approximation.

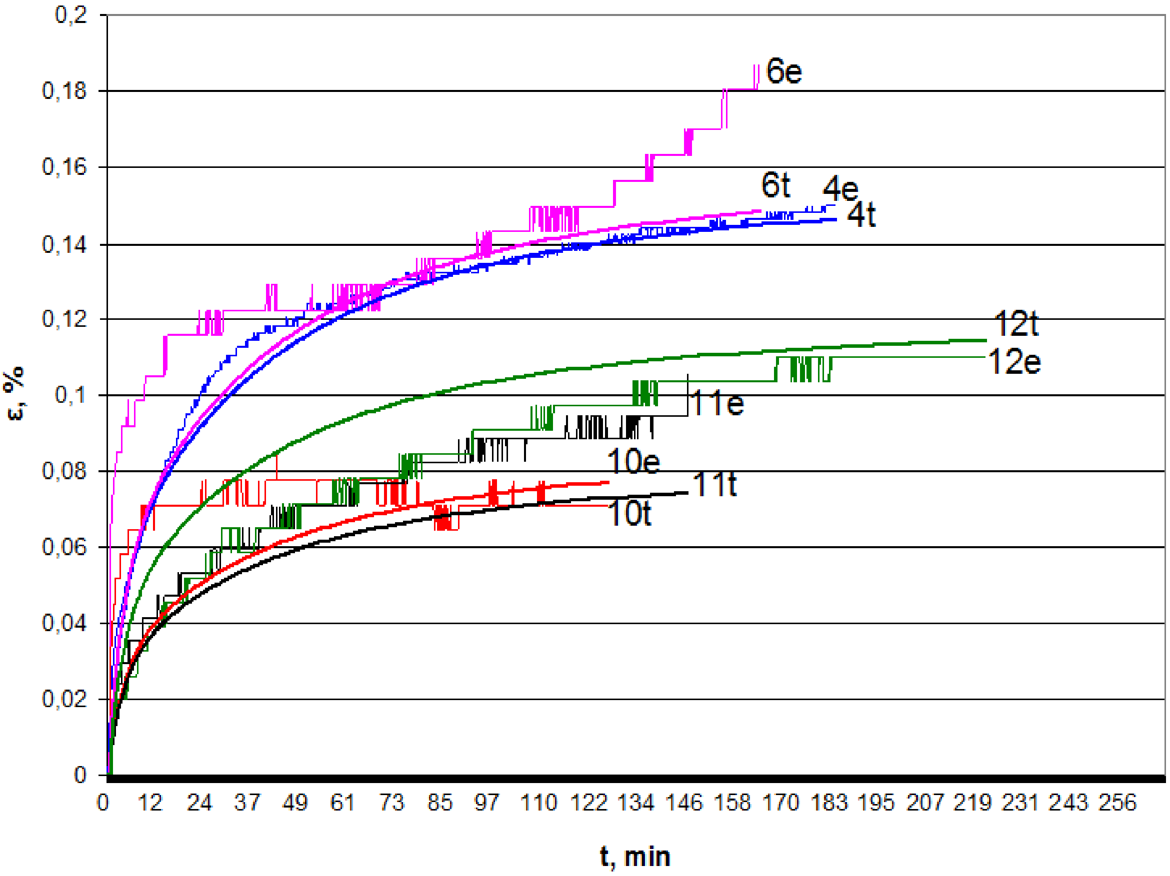

Figure 3.1. Theoretical and experimental creep curves of human jaw compact bone tissue for 5 specimens group from approximation by three exponents. 4, 6, 10, 11, 12 – the specimens No.; e – experimental creep curves; t – theoretical creep curves.

After analysis of presented results we can conclude:

From compression tests of jaw compact bone tissue samples the active creep strains were calculated which characterise the material average creep properties in a stress range from 10.4 to 20.48 MPa.

The experimental creep strains of human mandible compact bone material were approximated with exponential function by two and three exponents for 15 individual specimens. The precision of the approximation increases on an average by 30 % in case of using three exponents.

After estimation of obtained results and repeated approximation for whole group of 5 specimens with two and three exponents, average values of the creep properties of human jaw compact bone tissue were achieved. The precision of the approximation increases for the group only by 0.65 % in case of using three exponents.

Obtained results should be taken into account in finite element studies to simulate e. g. fracture repair and bone remodelling.

4. COMPACT BONE TISSUE EXPERIMENTAL INVESTIGATION UNDER CYCLIC LOAD

4.1 Cycles to failure dependence on stress, localization of samples in bone and its density

Of the 30 specimens prepared for fatigue testing, 29 were tested successfully. One specimen was not used due to premature shutdown of the testing machine during the fatigue test. Results from fatigue tests of pig jaw cortical bone in compression are summarized as stress range versus cycles (S-N curve). The effect of applied cyclic loading on the number of cycles to failure has the form of an inverse power law:

$$N = F\Delta\sigma^{G} \tag{4.1}$$

where N – the number of cycles to failure,

$\Delta\sigma$ – stress range (MPa), $\Delta\sigma = \sigma_{max} - \sigma_{min}$,

F and G – empirical constants.

Equation constants were determined from curve fits of fatigue data of pig jaw compact bone tissue (Fig. 4.1.): $F = 1.47 \cdot 10^{20}$; $G = -10.05$; $r^2 = 0.804$.

The fatigue properties of pig cortical bone determined in this study were compared with previous studies from the literature. For comparison were used data determined in compression and shear for compact bone tissue specimens from human femur or bovine tibia. The S-N curves were normalized by cortical bone strength. The fatigue data taken from current study were normalized by ultimate compressive stress (mean value is 68±1.8 MPa); shear fatigue data of human compact bone were normalized by longitudinal shear strength (52 MPa); bovine tibia compressive fatigue data – by 272 MPa, human femur fatigue data – by longitudinal compressive strength (182 MPa). In Figure 4.1 are shown uniaxial compressive fatigue data from current investigation and from other investigations after normalization. Analysis of covariance showed that there were no significant differences between the normalized shear fatigue S-N curve and the normalized compressive S-N curve, suggesting that fatigue behaviour is similar for human, bovine and pig cortical bone.

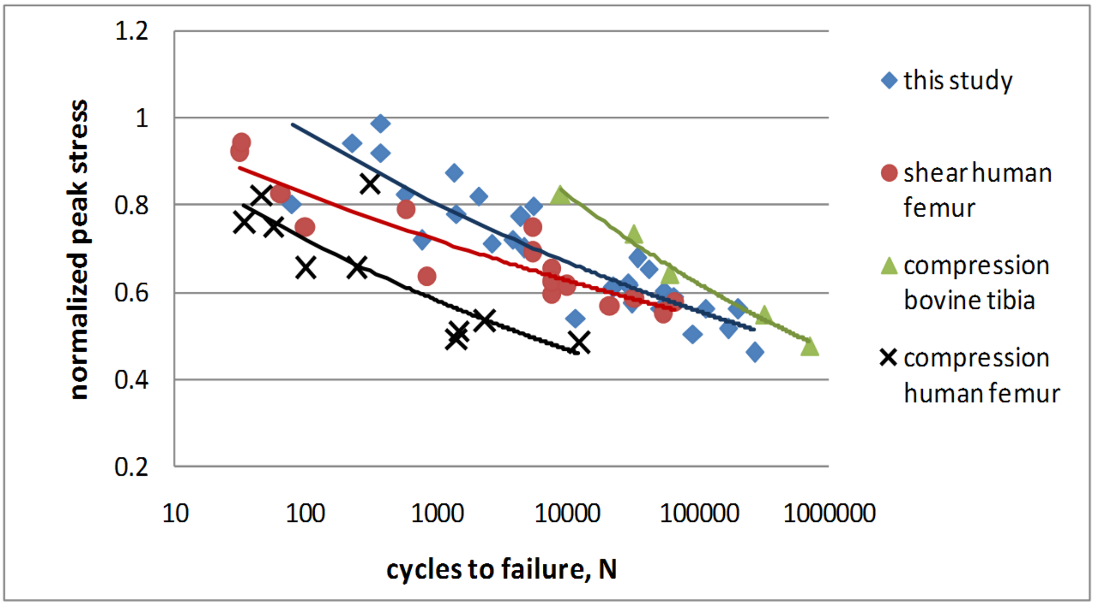

Figure 4.1. S-N curves comparing fatigue properties after peak stress was normalized by the ultimate strength.

Using the experimental data the dependence of cycles to failure on the sample location in the mandible was stated. The number of cycles to failure for samples from premolar area is greater than for samples from the molar area. This relationship is characterized for both upper and middle part of buccal area of the mandible for all applied cyclic stress levels. On average, the number of cycles to failure for samples from premolar zone upper part of buccal area is greater about of 31% than in the molar area; while number of cycles for samples from the middle part premolar zone of the buccal area is greater about of 18% than in the molar area.

Based on a study data, the cycles to failure dependence on compact bone density was determined. After analysis of the obtained curves, there was stated that the pig mandibular compact bone fatigue life increases with bone density decrease. This trend was observed both at the low and high cyclic stress levels. Improvement in fatigue properties with the density decrease can be explained by the nonlinear behaviour of compact bone.

Different authors studied and experimentally demonstrated the influence of moisture on the mechanical properties of bone tissue. In the general case, with the increase of the moisture content of bone tissue compression strength, the modulus of elasticity and stiffness decrease. At the same time, some structural alterations are taking place in the material with the increase of its moisture. The water that had penetrated into bone tissue causes the hydrostatic tension in bone matrix. The increase of moisture content causes the transfer of fracture mode from the brittle to the ductile one: the bone tissue behaves like a viscoelastic material. In a moist bone the hydroxyapatite crystals deform elastically, while the behaviour of collagen matrix is viscoelastic. The moisture content of samples with a lower density t. i. with higher porosity at our tests conditions may be higher than for samples with higher density. Based on the above speculations, we can say that the viscoelastic behaviour and resistance to failure under cyclic loads in samples with the lowest density is more expressed than in samples with the highest density, provided that the loading stress is calculated depending on the density of bone tissue and samples are immersed in water during tests.

4.2 Failure prediction

The ability to predict the effects of damage on functional load-bearing behaviour and the ability to predict the effects of changes in bone composition and organization on the accumulation of damage are relevant to both clinical medicine and biomedical research. Pig jaw compact bone tissue time to failure was predicted using a general model for mechanical and biological damage accumulation in a self-healing, living structure:

$$D_F(t) = \omega t / F\Delta\sigma^{-G} \tag{4.2}$$

where ω is the loading frequency (Hz) and t is time (s).

If the specimen fractures due to the accumulation of fatigue damage only, one could solve for the time to fracture, t_b, by setting $D_F(t_b) = 1$ in equation (4.2). This component is shown as the line in Figure 4.2. For this particular loading history, this model prediction shows good correlation with the data, suggesting that 0-C cyclic loading of pig jaw cortical bone produces primarily fatigue damage.

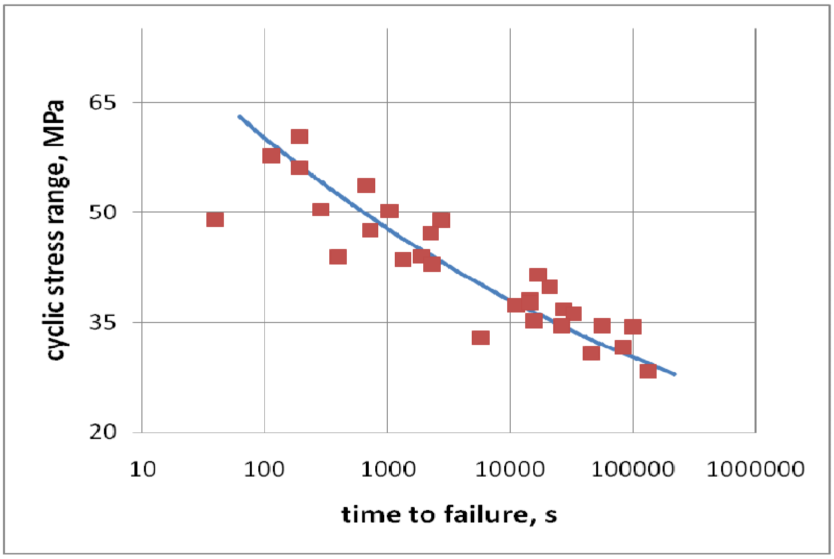

Figure 4.2. The effect of stress range on time to failure for zero-compression fatigue specimens of the pig mandible compact bone tissue is shown in relation to the failure times predicted by equation (4.2).

4.3 Failure prediction

For the development of fatigue life model Weibull analysis was used, which is well known in the treatment of scatter in fatigue fracture. It is assumed that, if a series of nearly identical specimens are tested, their measured fatigue strength will vary according to:

$$P = 1 - exp\left(-\left(\frac{\Delta\sigma_0}{\Delta\sigma_0^*}\right)^m\right), \tag{4.3}$$

where P is the probability that the fatigue strength of a given specimen will be less than or equal to $\Delta\sigma_0$; m and $\Delta\sigma_0^*$ are constants.

If the stressed volume of the specimen is V_0, it can be shown that the probability of failure in a specimen of a different volume, V_S is:

$$P = 1 - exp\left(-\left(\frac{V_S}{V_0}\right)\left(\frac{\Delta\sigma_0}{\Delta\sigma_0^*}\right)^m\right). \tag{4.4}$$

A larger specimen can be considered to be made up of several smaller specimens, each of which obeys Eq. 4.3. If P is set to some value, then $\Delta\sigma_0$, will decrease with increasing volume:

$$\Delta\sigma_0 = \Delta\sigma_0^* \left(-\left(\frac{V_0}{V_s}\right) ln(1-P) \right)^{1/m} . \qquad (4.5)$$

To use this equation we need to determine the constants $\Delta\sigma_0^*$ and *m*. This can be done independently by considering the scatter in properties measured from a large data set of specimens of identical geometry. In Figure 4.3 are shown pig jaw compact bone tissue fatigue data, Weibull distribution, used to obtain the constants in Eq. 4.3 and data from other investigation. In this investigation authors [Carter D. R. et. all, 1981] tested 74 specimens of human compact bone tissue at various levels of stress under uniaxial tension-compression loading at temperature 37 °C and frequency of 1 Hz. As shown in the figure, the data obtained in this study are little different from the data of human compact bone.

Experimental probability of compact bone tissue failure is defined by the number of data points that occur below a given stress level, divided by the total number of data points. The stress axis has been normalized by the mean stress level $\Delta\sigma_{min}$ (i.e., the centre of the scatter band) to allow comparison of data at different absolute stress levels.

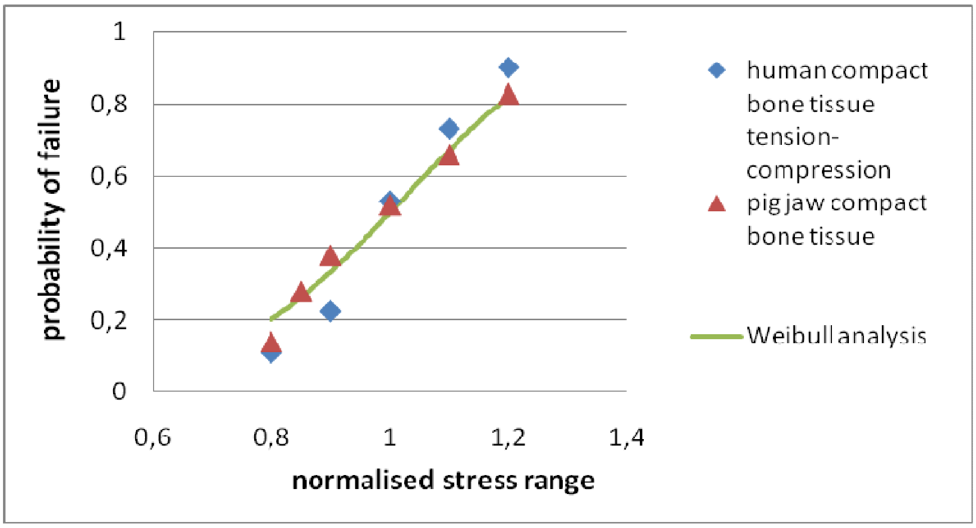

Figure 4.3. Comparison of experimental data of pig mandible compact bone tissue, human compact bone tissue and best-fit Weibull distribution.

Equation 4.5 was optimized with respect to the data when the values of the constants were m = 5 and $\Delta\sigma_0^* / \Delta\sigma_{min}$ =1.077. The error in the fit to these points, expressed in terms of the value of P, was 5 %. Thus, if we know the fatigue strength $\Delta\sigma_0{}'$ for some specimen volume, $V_S{}'$, then we can find $\Delta\sigma_0$ for any other V_S by

$$\Delta\sigma_0 = 1.077\Delta\sigma_0'\left(-\left(\frac{V_S'}{V_S}\right)ln(1-P)\right)^{1/5}. \tag{4.6}$$

For fatigue strength prediction experimental data obtained under mean stress range 30±2 MPa were used. At this level, the applied stress is sufficiently low to eliminate any effects of creep, so results will be relevant to physiological conditions. The mean life obtained under this stress was N_{min}=134775 cycles. In Figure 4.4 are shown pig mandible compact bone tissue theoretical fatigue strength as a function of stressed volume and experimental data. The fatigue strength was calculated using equation (4.6) at probability of failure of 0.5, 0.8 and 0.05. Agreement between prediction and data for the present test series was very good, with an error of only 2 % in the fatigue strength, which is within experimental error for tests of this kind.

For the purpose of failure prediction as closely as possible physiological conditions, the fatigue strength was determined at body temperature (37 °C). As is known the difference between these two temperatures was a factor of 1.16 on stress, so this factor was used with equation (4.6) to predict fatigue strength at body temperature. Figure 4.5 is a plot of fatigue strength as a function of stressed volume, showing the prediction for physiological frequency and body temperature at probability of failure of 0.5, and 0.05, and also showing the experimental point.

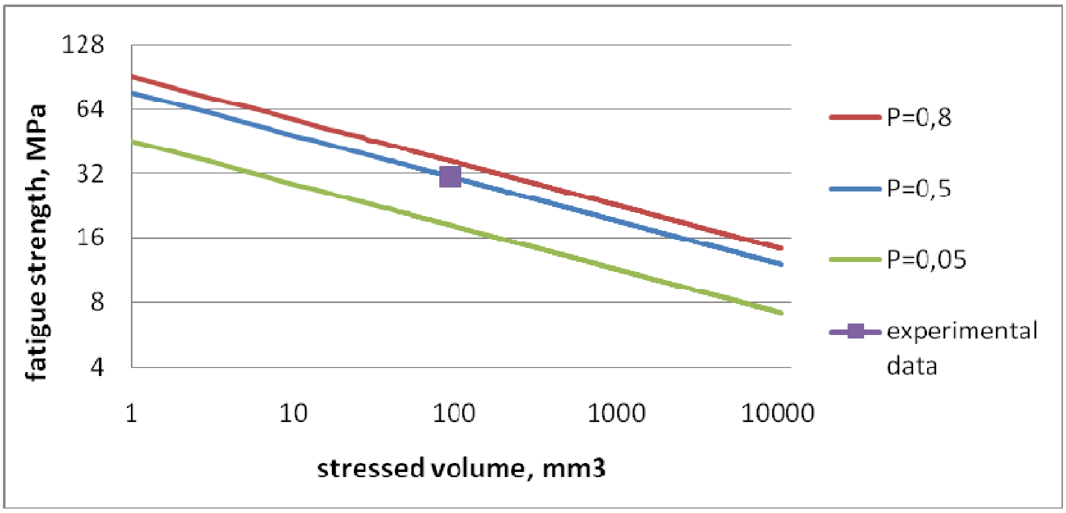

Figure 4.4. Pig mandible compact bone tissue fatigue strength at 134775 cycles as a function of stressed volume for physiological frequency loading at room temperature.

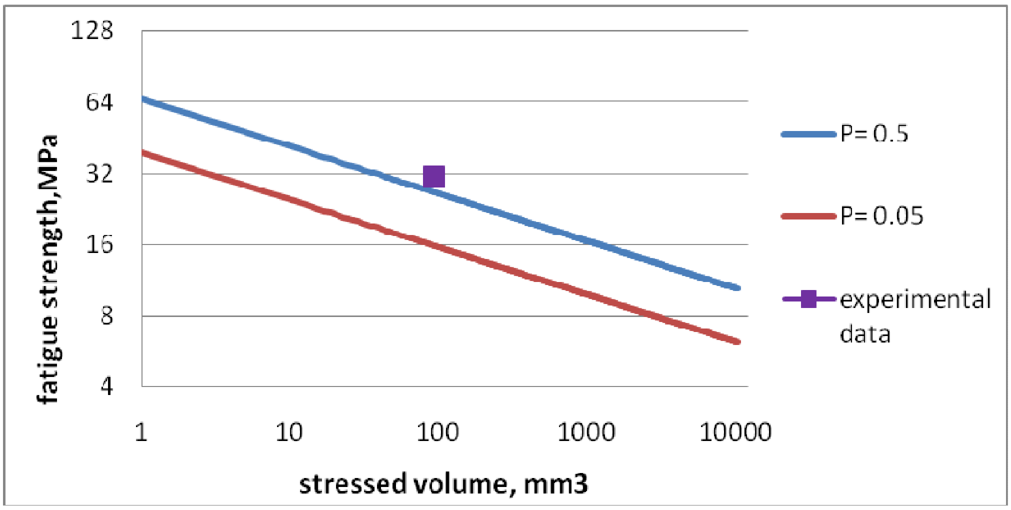

Figure 4.5. Pig mandible compact bone tissue fatigue strength at 134775 cycles as a function of stressed volume for physiological frequency loading at body temperature.

The results of Weibull analysis use have been very successful: the model was able to combine the effects of specimen size, temperature and loading stress in such a way as to predict the strength of pig jaw compact bone in compression with high accuracy.

Using this model one can predict the fatigue strength for both the very small samples (1 mm^3 and less) and the very large. Perhaps the most interesting use of this method is in the prediction of fatigue strength for whole mandible. But there must be considered that repair processes are active and that they are able to respond to the emergence of fatigue damage within a period of months. The probability of failure of a bone *in vivo* has been estimated as 0.02-0.07. Analysis of presented curves (Fig.4.4, 4.5) indicates that mandible theoretical fatigue strength *in vivo* is sufficiently high for used number of cycles to failure. However, some individuals in the population will be at risk of failure even at very low cyclic stress levels.

5. HUMAN MANDIBLE FINITE ELEMENT ANALYSIS

5.1 The modelling of mandible fracture fixation system

Miniplate osteosynthesis has become a standard treatment of mandibular fracture treatment and reconstruction in oral and maxillofacial surgery. To avoid postoperative mechanical loosening of screws caused by excessive loads, it is important to determine the nature of stress distribution in the bone surrounding the biomaterials. Loosening of screws and bone resorption is associated with high peak stresses at the interface in the immediate postoperative stage. To analyze such incongruences with respect to the complex biomechanical behaviour of the mandible in the living subject, a mathematical method – finite-element analysis – was used.

5.1.1 Geometric model

A 3D finite element model of mandible fracture fixation system developed in previous studies was modified and improved: the orthotropic material properties was assigned to the mandible, a new type of contact between the mandible and the fixation plate was created, the number of muscle pair in loading conditions was increased by 5. For modelling of screw connection of the mandible and fixing system the holes with a diameter of 1.8 mm and a length of 4.5 mm were created in jaw, and fixing the system was enhanced with special screws. Due to irregular shape of the mandibular surface the gap was assumed between the plate and the bone. Modifications of plate and mandible geometric models were made in the SolidWorks 2008 software and then exported to finite element analysis software ANSYS v.11, using a IGES data file format.

5.1.2 Material properties

The bone was modeled as an orthotropic, heterogenous, composite material made of cancellous and compact elements. In Table 5.1 are given mandible properties – Young's modulus of elasticity (E), Poisson's ratio (μ) and shear modulus (G) – depending on regions and bone tissue type.

The material properties were assigned to the FEM according to mandibular geometry with a curve representing the long axis of the jaw. This curve was drawn along the entire mandibular centre from the right to left posterior condylar surfaces, and its longitudinal direction represented the material X axis of each bone element (Fig. 1). The Y-axis was aligned perpendicularly to the X axis but tangentially to each related jaw cross section. The material Z-axis was derived from the X and Y axes and lay perpendicularly to both.

Table 5.1

Material properties assigned to the finite element models of the human jaw

[Sukhun J. et. all 2007].

Location	Material properties								
	E, GPa			μ			*G*, GPa		
	X	*Y*	*Z*	*XY*	*YZ*	*XZ*	*XY*	*YZ*	*XZ*
Compact bone tissue (symphysis)	22.9	14.2	10.5	0.19	0.31	0.29	6.0	3.7	4.8
Compact bone tissue (premolar area)	25.5	14.5	10.2	0.15	0.30	0.25	6.2	3.4	6.23
Compact bone tissue (molar area)	19.5	13.6	10.2	0.39	0.20	0.55	6.2	4.1	5.9
Compact bone tissue (ramus)	17.0	6.9	8.2	0.31	0.33	0.31	4.6	2.9	2.8
Cancellous bone	0.38	0.38	0.38	0.47	0.47	0.47			

The fixation plate and screws were considered as one mechanical system. It was assumed that the monocortical fixation plate and screws were made from titanium material. The mechanical properties for titanium were used: modulus of elasticity – 100 MPa, Poisson's ratio – 0.3. The fracture was modelled as a thin layer (0.8-1.2 mm) of material with varying properties during bone repair and healing period. The layer material properties behaviour was isotropic, homogeneous and linear elastic.

5.1.3 Boundary and loading conditions

Three-dimensional constraints were placed bilaterally at the endosteal surfaces of the temporal bones. These constraints imitated static mandibular opening with fixation of the mandibular apparatus at the cranium. In this way both condyles were assumed to be centred in their glenoid fossae. The model was also constrained from vertical movement with the goal modelling of standard closure occlusal contact. For modelling of the biting point the restraints were located on molar regions superiorly.

The FEM was loaded with multiple force vectors to simulate muscle forces over wide areas of attachment. Groups of parallel vectors simulated eight pairs of masticatory muscles (superficial and deep masseter; anterior, middle and posterior temporalis; medial, inferior lateral and superior lateral pterygoid) assumed to be directly attached to bone (Table 5.2). The *X*, *Y*, and

Z coordinates represent the muscle loads and their three-dimensional directions. All coordinates are referenced to a global Cartesian coordinate system where the *X–Y* plane is the frontal plane, *X–Z* represents the horizontal plane and the *Y–Z* indicates the mid-sagittal plane.

Table 5.2

Number and magnitudes of muscle loads used in the finite element model [Sukhun J. et. all 2007].

Muscle group	Muscle weighing factor, N	Muscle orthogonal components (N)					
		Right			Left		
		X	*Y*	*Z*	*X*	*Y*	*Z*
Superior masseter	190.4	17	72	43.5	-17	72	34.5
Profundus masseter	81.4	5.85	8.19	-3.9	-5.7	7.98	-3.8
Medial pterygoid	174.8	-28.98	47.04	22.26	29.04	47.08	22
Anterior temporalis	158.0	34.2	122.4	0.2	-34.2	122.4	0.2
Medial temporalis	95.6	3.5	12.74	-7.7	-4.2	15.26	-9.1
Posterior temporalis	75.6	3.36	7.56	-13.56	-3.36	7.56	-13.56
Inferior lateral pterygoid	66.9	-37.8	-10.5	45.6	37.8	-10.5	45.6
Superior lateral pterygoid	28.7	17.04	1.64	14.44	17.04	1.64	14.44

Contact between the mandible and fracture fixation plate was modelled with the nonlinear finite element method as the contact friction. There was used orthotropic friction model, which was determined by the coefficient of friction. Contact was modelled between the surfaces of screw and the lower jaw. The friction coefficient value in a direction parallel to the longitudinal axis of the screw was defined as 0.9, but in the tangential direction – as 0.4.

5.1.4 Analysis of results

The evaluation of the effectiveness of jaw fracture fixation was performed with respect to stresses and strains in the titanium fixation system and in the mandible. For the titanium fixation plates, the Mises equivalent stresses and total deformations were determined, for mandible – the stress intensity and total mechanical strain intensity. Results were obtained at 2 load cases: at masticatory muscles value which corresponds to normal bite force and at heavy bite force with load increasing by 25 %. Moreover analysis was performed at healing periods of 0.1 %, 25 % and 50 % which correspond to following values of modulus of elasticity for fracture layer: 0.01 GPa, 5.5 GPa and 11 GPa respectively. The calculations were conducted using finite element analysis software – ANSYS Ver.11.

For the titanium fixation system, the maximum Mises equivalent stresses and total deformations resulting from 2 load cases and at 0.1 %, 25 % and 50 % fracture healing are given in Table 5.3. For the same object the stresses distribution at healing periods of 0.1 % and 50 % and at normal bite force are shown in Figures 5.1 and 5.2.

The results show that the greatest stress values were obtained at 0.1 % fracture healing. In addition the stress value increases by 31 % in case of use of heavy bite force. The stress values at 25 % fracture healing are little different from ones at 50 %.

The variations of total strain correspond to stresses behaviour and the maximum of strain is less than 1.98×10^{-3}. Maximum equivalent stress localization in fixation system is different during the bone repair. At the healing period of 0.1 % the maximum stress was localized at the point where the second screw is attached to the plate. At 25 and 50% fracture healing the maximum stress localization was detected on the first screw point (Fig.5.1 and 5.2). Both in the first and in the second instance, the high values of the stress were observed in fixation plate near the second screw. In this place in case of adverse conditions, such as the load increases, which are caused by muscles force or the bite point changes, we can expect the cracks initiation and subsequent fracture site formation.

Table 5.3

The maximum Mises equivalent stresses and total deformations of titanium fixation system under load corresponds to normal bite force (norm.) and heavy bite force (+25 %) and at bone healing periods of 0.1 %, 25 % and 50 %

Parameters	Load mode	Healing period		
		0.1%	25%	50%
σ, MPa	norm.	116.7	27.5	25.3
	+ 25 %	152.8	33.2	32.6
ε	norm.	1.51×10^{-3}	0.34×10^{-3}	0.33×10^{-3}
	+ 25 %	1.98×10^{-3}	0.48×10^{-3}	0.42×10^{-3}

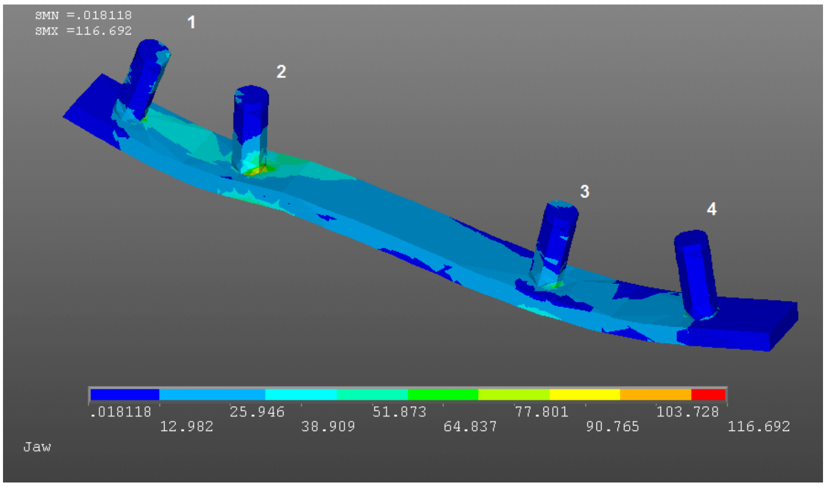

Figure 5.1. The Mises equivalent stresses of titanium fixation system at 0.1 % fracture healing and at normal bite force

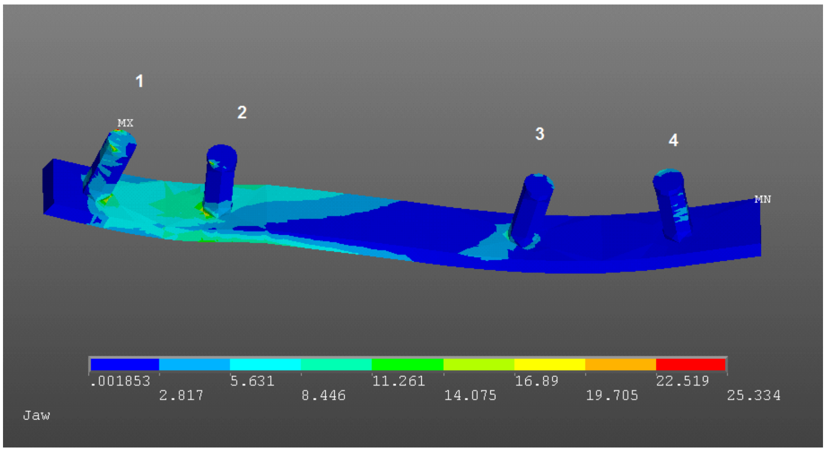

Figure 5.2. The Mises equivalent stresses of titanium fixation system at 50 % fracture healing and at normal bite force

Titanium yield limit (0.2% strain limit) under static compression is approximately 900 to 1000 MPa. The value of 900 MPa is not exceeded by any loading mode and healing period of bone tissue. Moreover, the fatigue limit for bending of titanium, which is approximately 450 to 500 MPa, is significantly higher than Mises equivalent stresses values. We can conclude that in fixing system with chosen shape and size at given loading conditions there is no fracture risk of plate or screw as a result of plastic deformation or fatigue.

In Table 5.4 are given the stress intensity and total mechanical strain intensity values that were calculated inside the mandible in surroundings of the fixation device. The greatest stress intensities were obtained at 0.1 % of fracture healing. In this case the load rise by 25 % causes increase of stress value by 16.5 % and the strain value by 60 %. At healing periods of 25 % and 50 % with improving of bone tissue strength at fracture site the values of stresses and strains differ slightly. The stress value increases on an average by 5 % during bone repair (by comparison 25 % and 50 % fracture healing) and by 18 % in case of use of heavy bite force. The strain values vary by 8 % and by 20 % respectively.

Table 5.4

The maximum stress intensity and total mechanical strain intensity of mandible bone tissue under load corresponds to normal bite force (norm.) and heavy bite force (+25 %) and at bone healing periods of 0.1 %, 25 % and 50 %

Parameters	Load mode	Healing period		
		0.1 %	25 %	50 %
σ, MPa	norm.	31.04	19.5	18.5
	+ 25 %	36.5	23.1	22.6
ε	norm.	0.1353	0.0114	0.0105
	+ 25 %	0.2158	0.0143	0.0132

The stress distribution inside the mandible bone tissue at healing periods of 0.1 % and 50 % and at heavy bite force are shown in Figures 5.3 and 5.4. The stress distribution in the direct surroundings of the fixation system differs significantly. If in the first case (0.1 % fracture healing) the maximum stress is concentrated at the third screw and a high stress values were also observed at the first screw, while in the second case (50 % fracture healing) the maximum stress location was found on the internal side of mandible surface near the second screw. In this case, high stress concentration was determined in the bone between the second and third as well as between the third and fourth screws.

Analysis of the strain distribution inside the mandible bone tissue indicates that the greatest strain values are located in different places. If in initial healing phase the maximum strain values are obtained directly in the fracture site then at 50 % fracture healing most deformation is concentrated at the second screw. Analysis of the maximum strain localizations and a resulting displacement distribution in the initial healing phase gave the possibility to conclude that modelled fracture fixation type is not sufficiently good for bone healing in the first stage. In the first healing phase this type of fractures should be fixed preferable by two plates. Next, at healing periods of 25 % the fracture can be fixed by a single plate.

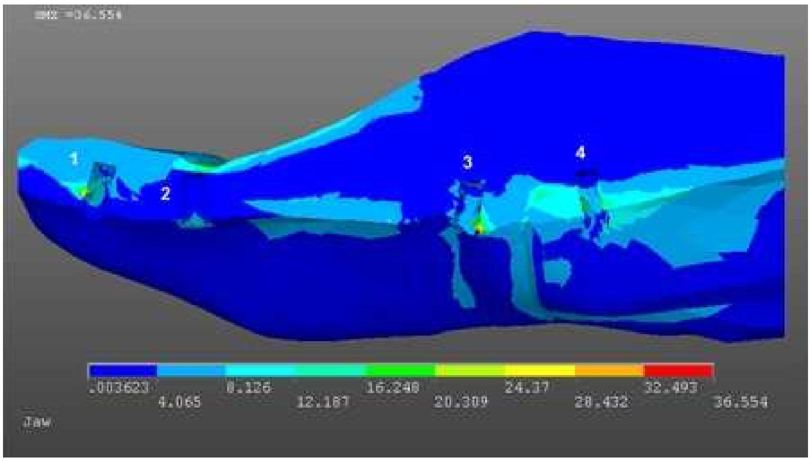

Figure 5.3. The stress intensity inside the mandible bone tissue at healing periods of 0.1 % and at heavy bite force

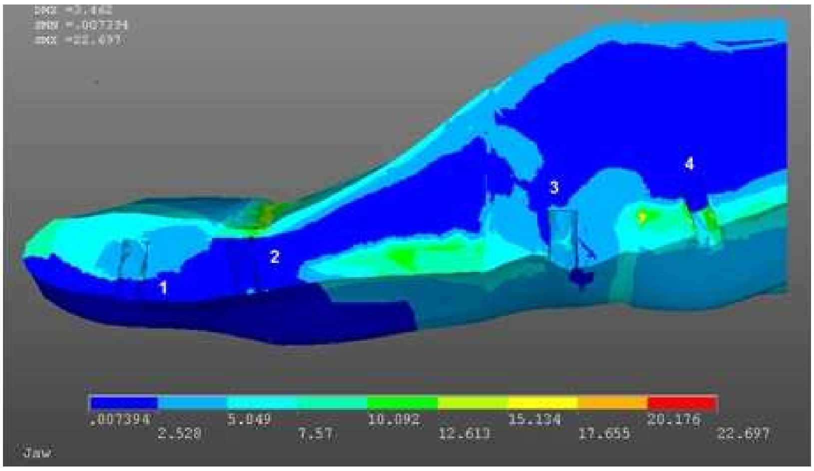

Figure 5.4. The stress intensity inside the mandible bone tissue at healing periods of 50 % and at heavy bite force

Nowadays, the problem of bone damage has not been completely understood yet. Nonetheless, it constitutes a fundamental issue for prosthetic dimensioning. Since bone architecture is very complex and mechanical properties of bone are highly heterogeneous, it is difficult to define a reliable critical value of stress or strain to be used, which triggers the phenomena of bone remodelling and further bone loss. In the literature the value of threshold stress causing bone resorption varies from 50 to 25 MPa. In this investigation the critical threshold of 50 MPa has not been exceeded, but stress of 25 MPa was exceeded at healing period of 0.1 % in case of use loading which corresponds to normal and heavy bite force. It can be concluded that in fracture fixation system the bone resorption may occur around the third screw and as a consequence of this – the loosening of screws.

This local stress peak normally will not lead to spontaneous damage of the plate. If it was assumed that the patient performs several thousand masticator movements within a week, a dynamic strain would be present due to the large number of changes in loading, so that there was a danger of a fatigue fracture at these points, as is also confirmed by clinical practice. In this computer simulation for given model the potentially dangerous place was found in the fixation plate between the second and third screw.

Occlusal relationships also have to be mentioned in terms of both a protective and a destructive effect on osteosynthesis. In cases in which full dentition exists with an adequate occlusal relationship, bite forces will be evenly distributed in the dentition, and critical stresses might not be exerted inside the mandible or fixation device. However, a patient with full dentition might exert higher masticatory forces and, therefore, more load on the fracture and plates could be applied.

CONCLUSIONS

1. From the literature review is concluded that the lower jaw compact bone viscoelastic properties and behaviour under cyclical loads are insufficiently explored, despite the fact that these research topics are necessary for orthodontics and maxilla-facial surgery.
2. Experimentally determined active creep strains of human jaw compact bone tissue samples characterise the material viscoelastic behaviour in the stress range from 3.3 to 27 % from ultimate compressive stress.
3. The approximated experimental creep strains give the possibility to conclude that the use of exponential function with three exponents increases the precision of the approximation on an average by 30 % compared to the use of two exponents. Obtained creep mathematical model characterises mandible compact bone tissue viscoelastic properties in a stress range from10.4 to 20.48 MPa and can be taken into account in finite element studies.
4. From compression tests of jaw compact bone tissue samples the fatigue stress-life data were obtained which characterise the material behaviour in a cyclic stress range from 30.82 to 57.56 MPa at frequency of 2 Hz. The S/N curves obtained in present study and the results of previous studies on fatigue data were normalized by ultimate strength. Analysis of the results

showed that human, bovine and pig cortical bone fatigue behaviour is similar in compression and shear regardless that the specimens were taken from different bones.

5. The determined dependence of cycles to failure on sample location in the mandible characterize both upper and middle part of premolar zone of buccal area as the more fatigue-resistant compared to the molar zone. A negative correlation between density of mandible compact bone tissue and the number of cycles to failure was found, which makes it possible to conclude that the samples with the lowest density exhibit better viscoelastic and fatigue properties than the samples with the highest density at the above-described experimental methodology.
6. The predicted time to failure for pig mandible compact bone tissue, according to the cumulative damage model, allows to conclude that bone failure occurs under compressive cyclic loading by the accumulation of cycle-dependent damage only. The developed fatigue strength statistical model, in which stressed volume is accounted for, makes it possible to predict the pig mandibular compact bone fatigue strength at 134775 cycles to failure and at different probability of failure.
7. Obtained results may aid in understanding of mechanism of cortical bone failure and should be taken into account in finite element analysis to simulate e. g. fracture repair and bone remodelling, in treatment programs designing and in improvement of mechanical parameters of endosseous implants, as well as in development of new biomaterials.
8. The developed finite element model of human mandibles fracture fixation system allows to improve the accuracy of investigations of stresses and strains arising in the plate and into bone during mastication. Obtained by numerical simulation stress and strain values leads to the conclusion that in chosen fixation system at given loading conditions is not a fracture risk of plate or screw as a result of plastic deformation or fatigue. However, used device cannot provide fracture fixation in the initial period of bone healing. Analysis of finite element modelling data indicates that it is necessary to improve the fixation system and there was recommended the use of two plates.