Biomaterials – Joints and Problems of Contact Interfaces

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Intensive development of modern materials enables their increasing application in biomedical engineering. To make the material under consideration applicable, it should be made biocompatible with organic material. One of the problems encountered in design of the implant joints in a human body is the problem of the interfaces in contact, caused by significant difference in mechanical properties of the materials applied. Good understanding of these problems is of immense significance in prevention of premature failures of the structures in biomedical engineering.

Keywords: *joints of biomaterials, mechanical continuity, bone, implant, contact interface, failure prevention.*

1. INTRODUCTION

The materials used in biomedical engineering should have adequate mechanical properties. The properties required for the biomaterial under consideration depend on the field of application. Like with structural materials used in other fields of application, there are a few fundamental properties determining mechanical behaviour of a structure in this particular case, too: stiffness, strength, toughness and hardness. Considering the environment in which they are used, the biomaterials should be corrosion- and wear-resistant, assuming that the debris should be minimized. Finally, these structures should have good machining ability, as they should be frequently subjected to reshaping during operation.

The problems arising in design of biomaterial structures assume:

(a) ensuring of so-called mechanical continuity of an implant grafted in the body with surrounding tissues, and

(b) attainment of good mechanical properties most frequently excluding one another (e.g. high hardness and wear resistance accompanied by satisfactory toughness and machinability).

In this article a survey of recent studies aimed at solving the two specified problems is presented, with particular emphasis on the methods of analysis of implant contact zones and organic material in a human body.

2. MECHANICAL PROPERTIES OF MATERIALS USED IN BIOMEDICAL ENGINEERING

In contemporary medicine, a large number of materials are now used whose properties enable the functions of artificial organs or their parts and manufacture of various structures of biomedical accessories as well. In Fig. 1, a survey of one of basic features of material behaviour in elastic region – Young's modulus – is presented for a wide range of materials used in biomedical engineering, so-called Ashby presentation. The authors of this presentation [1] have also given the price in terms of the material mass. The common Ashby's maps show various mechanical properties of the materials in dependence of density, see [2].

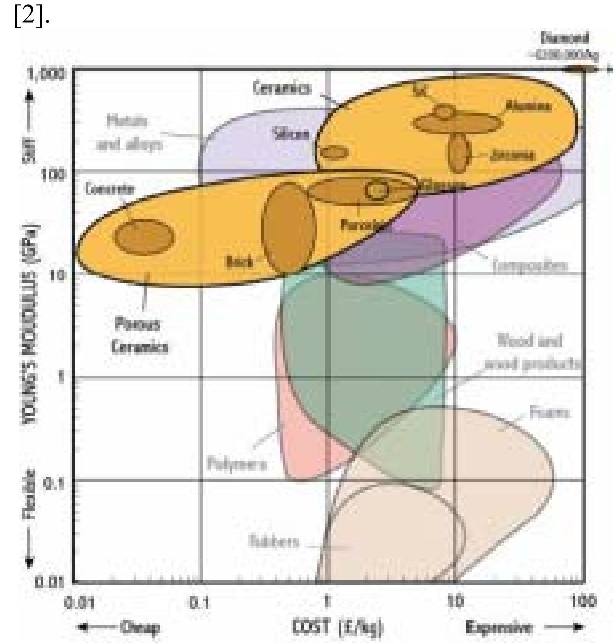

Figure 1. Values of Young's modulus for different materials used in biomedical engineering, [1]

In addition to such presentation, it is also necessary to analyze the differences between the mechanical properties of bones, tissues and materials used to produce implants. In Tab. 1, in addition to the values for Young's modulus, the values of ultimate tensile strength and two properties of the materials describing resistance to fracture (K_{IC} and J_{IC}) are given, too, according to [3,4].

K_{IC} is plane strain toughness, and it represents a material property established according to a procedure

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defined by a standard. J_{IC} (or G_{IC}) is a measure of fracture toughness, and it represents the resistance of the material to the crack growth.

In closing part of the paper, prevention of carrying-capacity failure of the materials in biomedical engineering and parameters with which the resistance to damage and fracture can be expressed will be discussed in more details.

In Tab. 1 the values of Young's modulus have been given for two types of bones, cortical and trabecular bones, of which a human skeleton consists. According to numerous references, a bone can be defined as a natural composite material which contains about 45-60 mass % of minerals, 20-30 mass % of matrix and 10-20 mass % of water.

Table 1. Mechanical properties and resistance to fracture of bone and materials used in joint replacements, [3,4]

Material	Young's modulus (GPa)	Ult. tens. strength (MPa)	K_{IC} ($MN/m^{3/2}$)	J_{IC} (kJ/m^2)
Alumina	365			
Cobalt-chromium alloy	230	450 – 1000	~100	~50
Austenitic stainless st.	200	200 - 1100	~100	80 - 200
Ti-6%Al-4%V	105	780 - 1050	~80	20 - 70
Hydroxy-apatite	85	40-100	2 - 12	
Cortical Bone	7-25	50 - 150		0.6 - 5
Trabecular bone	0.05 - 5			
PMMA bone cement	1.5	70		
Polyethylene	1	20-30	4 - 40	~8

3. ARCHITECTURE AND PROPERTIES OF HUMAN BONES

Although the bone is reported to be viscoelastic, it is more commonly characterized by its elastic mechanical properties [4]. However, the elastic mechanical properties of bone vary from bone to bone and within different regions of the same bone as well. This is particularly pronounced in trabecular bones, located at the end of long bones (e.g. femur) and cuboidal bones (e.g. spine). Considering their porosity, these represent a carrier of tissues in a human skeleton.

The matrix is an organic component, which is mainly composed of the protein type I collagen. As a highly aligned form, type I collagen causes extremely anisotropic structure. The mineral, as an inorganic part of bone, is a form of calcium phosphate known as hydroxyapatite (HA). Hydroxyapatite is present in the form of crystallites, approximate dimensions of which are 2 x 20 x 40 nm, according to [4]. The organic matrix ensures flexibility of a bone, while the mineral mainly defines the mechanical properties of a bone. Cortical bone has larger density than trabecular bone, the porosity ranging from 5 to 10% for former one and 30-90% for the latter.

In past few years, intensive research of mechanical properties of trabecular bones has been in progress for better understanding of the mechanism of aging, fracture and effects of medicine treatment (see [5-7]). As the experimental research is highly complex and expensive, numerical calculations using the finite element method (FEM) are more and more carried out. These calculations are extremely intricate because of high porosity of trabecular bone and complex architecture as well. In Fig. 2, the appearance of a trabecular bone modelled using micro finite elements (μFE) in [8] has been shown. The model was prepared using micro-computed tomography on vertebral trabecular bone specimen with 9% volume fraction.

Figure 2. Appearance of a trabecular bone, [8]

The objective of the research activities was to determine mechanical properties of this bone, both on the continuum level and the microstructural level. Non-linear analysis of geometry and material was carried out using ABAQUS programme (www.hks.com). It is interesting that the bone tissue was modelled using the cast-iron plasticity model as procedure providing elastic-plastic behaviour with different yield strengths and hardening in tension and compression. The results obtained were compared with experiments (see [9]) and for the time being a good agreement is attained under compression.

4. THE PROBLEMS OF CONTACT INTERFACES IN IMPLANT JOINTS

Good knowledge of mechanical loading of bones is particularly important in prosthesis of artificial parts in human body. In Fig. 3, the appearance of natural and artificial hip joints has been shown.

In the prosthesis, the ball of the natural hip joint is most commonly replaced by a ball made of metal (stainless steel, cobalt-chrome alloy or Ti-6%Al-4%V alloy) or alumina. Acetular cup is usually made of ultra high molecular weight polyethylene (UHMWPE), and it forms with metal ball articulating surface.

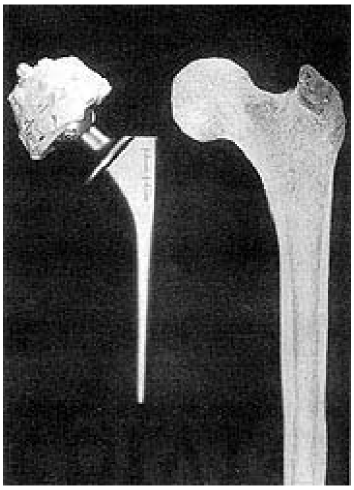


Figure 3. Natural and artificial hip joints, [3]

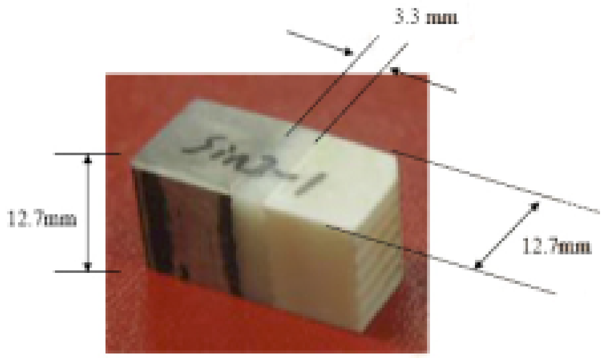
The two components were cemented into prepared bone using poly(methylmethacrylate) PMMA, known as bone cement. Having in mind the mechanical properties of this material (see Tab. 1), the efforts have recently been made to avoid it by implantation of the femoral implants as a press fit, for details, see [3].

However, satisfactory solution for grafting of this implant is possible only with application of synthetic hydroxyapatite coatings. The reason for application of the HA is to enable the bone to grow up to the implant and thus make a joint by direct bone bonding. Second solution assumes the application of porous metal coatings, and the objective is the same - encouraging bone growth into the prosthesis.

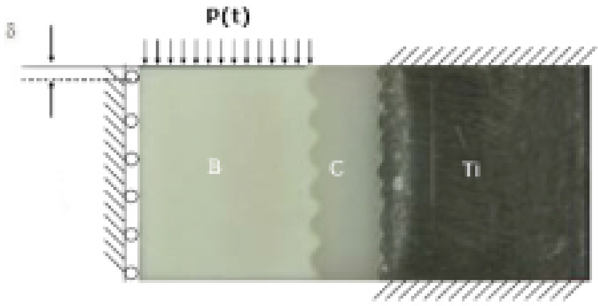
In case of a disease spread over larger part of bones that should be removed from the body, porous alumina has proved to be a suitable material. The porous architecture of these implants encourages the bone to growth into hollow parts, where extremely stiff carrying structure of alumina becomes a carrier of a new bone. The application of this material in formation of articulating surfaces in artificial hip joints has also been reported. It has been shown [10] that wear rates for alumina on UHMWPE have been as much as 20 times less than that for metal on UHMWPE, making this combination far superior and producing less tribological debris. It is well-known that debris could lead to complications in surrounding tissues and, as it has already been mentioned, it should be reduced to minimum value. The disadvantage of application of alumina as biomaterial is that the body can recognize it as a foreign material and attempt to isolate it by forming a layer of non adherent fibrous tissue around the implant where possible.

In a few last mentioned examples the joints are made by bonding the materials of extremely different mechanical properties. For instance, PMMA-containing structures have very poor properties to shearing. That may lead to the problems related to the interface failure of such joints, see [11-13]. In [13] are the samples of the bone/cement/titanium joints tested by fatigue loading. The samples of bovine femur bone and commercially pure titanium (cpTi) were used. In Fig. 4, the specimen dimensions (Fig. 4a) and boundary conditions (Fig. 4b) used for fatigue testing have been shown. During the testing minimum value of δ was 0.5 mm. The maximum load was 642 N, which corresponds to 30% of ultimate shear strength obtained in static analysis. Total number of loading cycles was 10 million, at stress ratio $R = 0.1$ and frequency of 10 Hz. It was found that surface topography of bone significantly affected the occurrence of fatigue fracture at bone/cement interface. However, that interface is not the only problem.

Due to large difference in mechanical properties, a stress and strain jump occurs at cement/titanium interface. Considering that in [13] experimental research was carried out, we included numerical calculation, too, using the finite elements method [14]. The stress and strain state under exterior static loading was monitored on a simple model, for boundary conditions given in Fig. 4b. The calculation was made under conditions of plane strain state, using the ABAQUS programme.



a)



b)

Figure 4. Specimen dimensions (a) and boundary conditions (b), [13]

In Fig. 5, the distribution of shear stress, in upper region of the specimen, at prescribed displacement $\delta = 0.5$ mm has been shown. The highest values of shear stress are obtained in narrow region of the interfaces bone/cement and cement/titanium. Unfavourable distribution of shear stress is particularly exhibited at the interface to the left (bone/cement) where the change in sign of the shear stress occurred.

In Fig. 6, the variation of shear stress (Fig 6a) and shear strain (Fig 6b) below the upper specimen interface has been shown (0.2 mm from the top edge). The distance of 3 mm represents the middle part of cement zone at both diagrams (the width of that zone is

3.3 mm). At interfaces jumps of shear components are significant. It has already been mentioned that synthetic hydroxyapatite is more and more used for coating of metal implants, to lower the discontinuity between the mechanical properties of bones and used biomedical material.

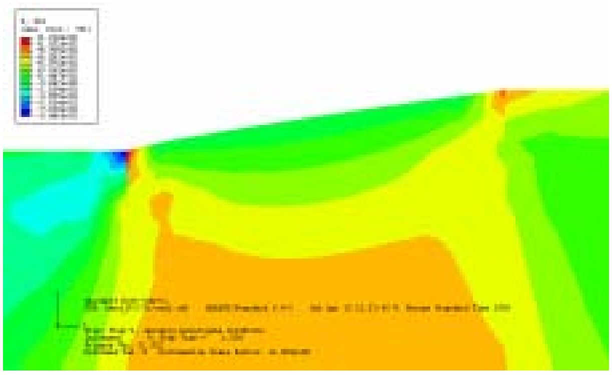
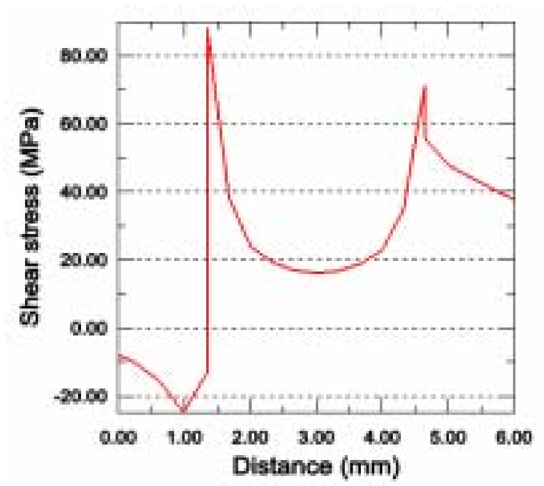
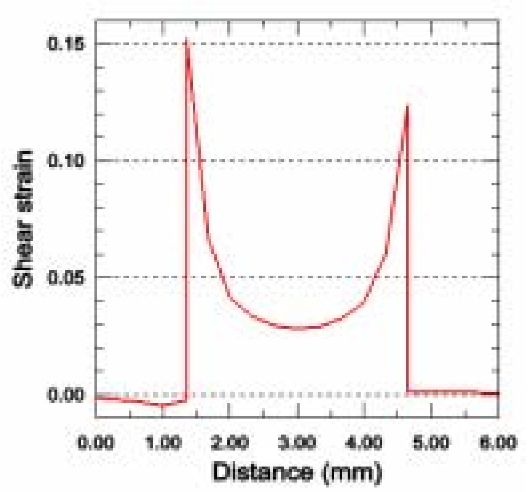


Figure 5. The distribution of shear stress at the interfaces in upper region of the specimen

These are thin coatings (50-120 μm), recently made as multi-layer coatings so as to achieve continuous variation of mechanical properties from implant to bone.



a)



b)

Figure 6. The variation of shear stress (a) and shear strain (b) in the zone of joints - 0.2 mm from the top edge

From Tab. 1 one can see that HA has the properties somewhere between ceramics, metal and bone. Mechanical properties of HA are, in fact, least disproportionate to mechanical properties of bones. In Fig. 7 [15] a gradual variation of applied coating from pure HA to Ti-6Al-4V alloy for attainment of so-called functionally graded material (FGM) has been shown

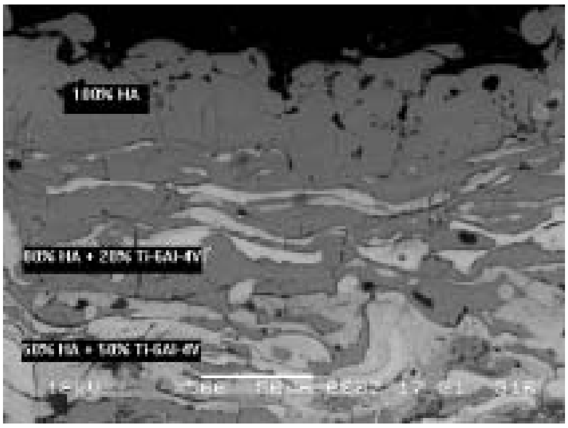


Figure 7. Multi-layer coating of hydroxyapatite on Ti-6Al-4V alloy, [15]

The development of FGM structures is intensive in various fields. In Fig. 8 a continuous variation of hardness and impact toughness in cast-in-carbide wear-resistant steel castings has been shown. Such alloys are used in manufacture of the structural parts exposed to wear with severe impact loading, for details see [16].

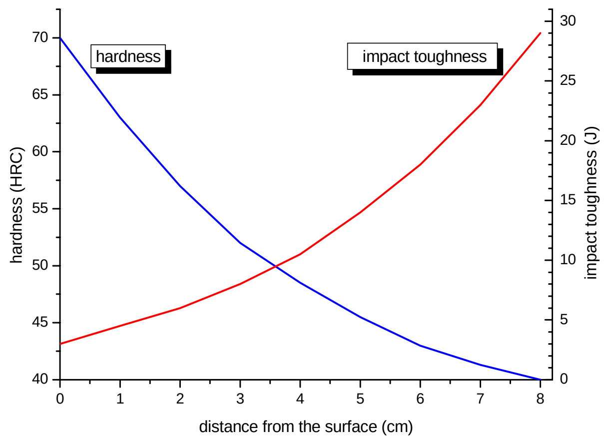


Figure 8. Continuous variation of hardness and impact toughness in cast-in-carbide wear-resistant material

The same requirements are in effect in design of the structures in biomedicine. Typical example is a composite material HAPEX^{TM}, developed at the Queen Mary and Westfield College. This composite material is composed of hydroxyapatite and high-density polyethylene meant as a synthetic substitute for bone (hydroxyapatite in collagen). The objective is to obtain adequate mechanical properties and machinability. Unlike the sintered HA as an extremely brittle material, HAPEX could be easily machined and even reshaped during the operation, see [3].

At the beginning, this composite was applied in manufacture of the bone supporting the eye in the second half of the eighties of the last century. Later on, HAPEX was used for manufacture of the middle ear implants (Fig. 9). Namely, the head of this implant is made of hydroxyapatite, while the shaft is made of the HAPEX composite. Commercial use of this structure was approved by the American Academy of Otology, in the year 1995. A good body response to these implants, which was not the case with previous solutions for middle ear (see [3]), and the possibility of machining of the HAPEX shaft during operation, to fit the patient, contributed to it.

Figure 9. Middle ear implants with hydroxyapatite heads and HAPEX shafts, [3]

5. FAILURE PREVENTION OF BIMATERIAL STRUCTURE WITH FLAW IN CONTACT ZONE

Of course, the oldest implants are the dental implants. As before, today tooth implantation is most frequent procedure of implantation of a synthetic-material structure in the human body. Attainment of biointegrity is essential here, as the implantation is based on direct contact between the bone and implant material. Attainment of adequate contact between a bone and an implant provides stability and long life. Detailed presentation of the possibility of application of modern materials in dental medicine is given in [17].

Here we shall stop for a while to analyze the region of contact between bones and implants. In Fig. 10 a schematic presentation of this region is given for a dental implant, where the joint is made using a thread on an implant structure.

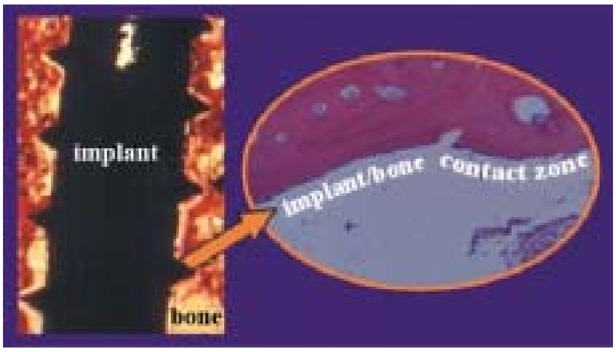


Figure 10. Schematic presentation of implant/bone contact zone in dental application, [17]

In spite of successfully realized joint from the macroscopic point of view, which in present dental practice is a routine procedure, the analysis on a microscopic level shows the possibility of damage. These damages lead to reduction of the implant life, and sometimes can cause the complications in case of some patients.

Failure prevention in heterogeneous structures represents one of the most demanding fields of present applied fracture mechanics. Lately, significant money resources have been invested in this field, with the aim to provide structural integrity of inhomogeneous structures. Detailed survey of recent research activities is given [18]. Most of the papers are based on defining the so-called debonding criteria, for the cases where the flaw is on an interface between two materials in a composite structure. However, the flaws occur not only along the interface, but in its vicinity as well.

To express quantitatively the effects of structural inhomogeneity on the possibility of occurrence of carrying-capacity failure, in recent investigations the concept of material forces is used. Presented in detail in [19], the concept provides the possibility to analyze all kinds of inhomogeneities in materials (phase boundaries, interfaces, dislocations, voids, cracks, etc.). The application of this theory in determination of structural integrity of bimaterial structures is enabled by the research of Simha et al., for details see [20], while the investigation of the fracture phenomenon in elastically inhomogeneous bimaterials is given in [21].

These cases are typical for the analysis of contact between implant and bone in human body, as well. The problem with these joints is a significant difference in mechanical properties, particularly pronounced in presence of so-called sharp interfaces, where their discontinuous variation exists. The analysis of the bimaterials with cracks, composed of the materials of different mechanical properties, has been given in [21], while the case studies of bimaterials divided by sharp interface in the field of residual stresses have been discussed in [22].

The aim of these investigations was to determine so-called effective crack driving force with which the resistance of a bimaterial joint to the onset of a crack growth could be quantitatively determined. The advantage of the developed concept is the possibility of analysis of the effects of a wide range of values for the properties of the materials making an inhomogeneous joint on the phenomenon of fracture in contact area.

6. CONCLUSION

In this article a survey of recent investigations in the field of application of modern materials in biomedicine has been presented. Different implant joints in a human body have been considered and the problems of their contact interfaces with organic material have been analysed.

Using a structure consisting of bone, PMMA bone cement and titanium, exposed to shear loading, discontinuous stress and strain variation in the area of material joints has been analysed. The efforts have been made to reduce this unfavourable effect through development of functionally graded materials, where the material properties are continuously varied in the so-called transition zone. More and more intensive application of synthetic hydroxyapatite and composites based on this material has been presented in particular.

In the closing part of the article, a theory is presented that has been developed in order to predict the failure resistance of the structures composed of the materials of different properties. A review of recent investigations in this field provides the foundations for application of the concept of material forces in design of the joints in biomedical engineering.

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БИОМАТЕРИЈАЛИ – СПОЈЕВИ И ПРОБЛЕМИ ДОДИРНИХ ПОВРШИНА

Марко Ракин, Александар Седмак

Интензиван развој савремених материјала омогућава њихову све ширу примену у биомедицинском инжењерству. Да би разматрани материјал могао да се примени као бикматеријал, потребно је да се обезбеди биокомпатибилност са органским материјалом. Један од проблема који се јавља при пројектовању спојева импланта у људском организму је проблем додирних површина изазван великом разликом у механичким особинама материјала који чине спој. Стога је добро разумевање ове проблематике од изузетног значаја у спречавању превремених отказа конструкција у биомедицинском инжењерству.