wgt% (HAp/matrix)	porosity [%]	bulk density [g/cm³]
0	14.54	1.18
2	18.03	1.12
5	17.84	1.14
10	15.36	1.19
15	14.30	1.28
20	12.09	1.36
25	14.32	1.32

TABLE 2. The open porosity and apparent density of composite samples.

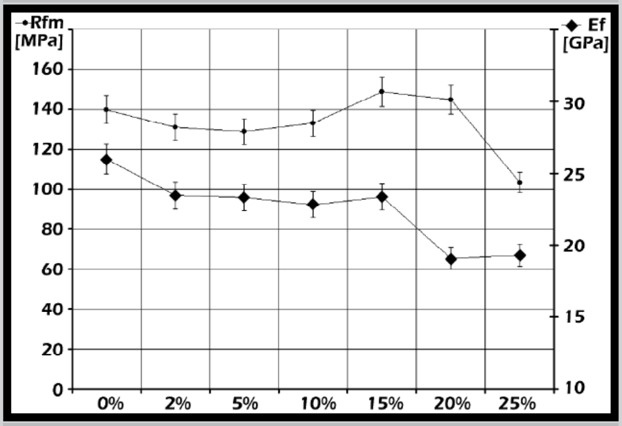

FIG.3. Flexure strength R_{fm} and flexure modulus E_f of composites with HAp nanopartricles.

FIG.4. Pictures of polished cross-section of composite with 5% HAp (left) and 20% HAp (right) illustrate multiple cracking in cured matrix with higher amount of HAp.

of cured matrix (FIG.4). In this case, the matrix seems to be prone to a brittle failure (becoming more ceramic), which could lead to a lower resistance to fatigue failure, which is one of limiting factors for applicability of implant materials.

Conclusions

We obtained flexure strength and flexure modulus of composites based on aramid fabric and polysiloxane resin with various amount of HAp nanoparticles additives. It is important to match the compromise between sufficient mechanical properties and amount of bioactive additives. It seems that higher amount of additives should have a negative influence on mechanical properties. As a further step it will be necessary to define an amount of additives more precisely (step 1-2%) and also to compare different size of particles (nano vs. micro).

Acknowledgements

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ANALYSIS OF A CONTACT STRESS DISTRIBUTION IN NEW SHAPE OF A HIP CUP

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Introduction

We develop a new design of an acetabular component for a total replacement of a hip joint. We indicate on the basis of the comparison our results of mathematical and finite element models of a contact stress distribution, that it is possible to use the finite element method for the modeling of the non-weigh bearing part of the total replacement of the hip joint. The point of this technical solution of the new hip cup is to design such a shape of the joint surface that will be symmetrical towards the hip joint stress. The shape is designed as the basic mathematical models of the distribution of the contact stress [1]. Three basic forms of this shape were designed. The cup with the hole was chose as the most suitable [2].

Materials and methods

We compared two various finite element models of the hip cup, cup with hole (A) and cup with hole and fillet edges (B). The models were loaded by five forces. We got contact stress distribution between the head and the modified hip cup. The each of those forces had different value and different direction of a loading. The forces matched values of a resultant hip joint force in hip joint in the course of different movement of a human body.

Results

The resultant contact stress distributions are on the FIGURES 1-5. The resultant contact stress distributions for each model at loading of a first force are in FIGURE 1, for second force in FIGURE 2, for third force in FIGURE 3, for fourth force in FIGURE 4 and in FIGURE 5 for last force.

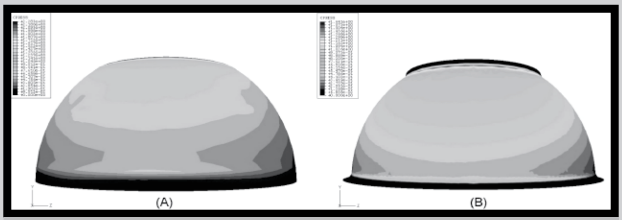

FIG.1.

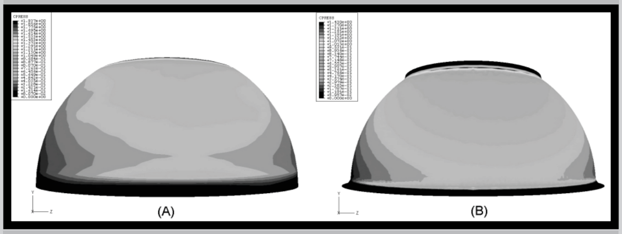

FIG.2.

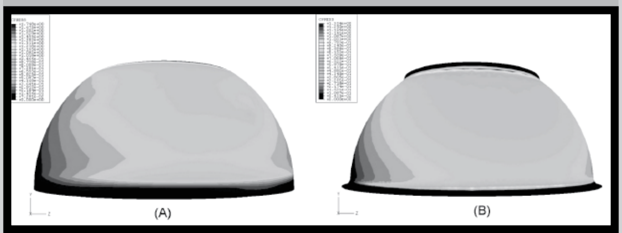

FIG.3.

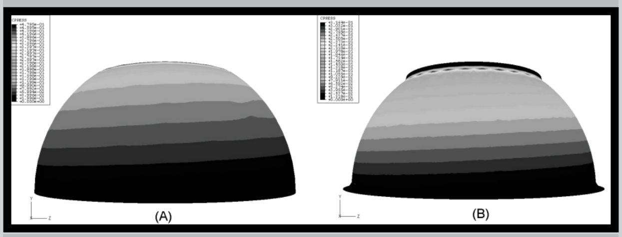

FIG.4.

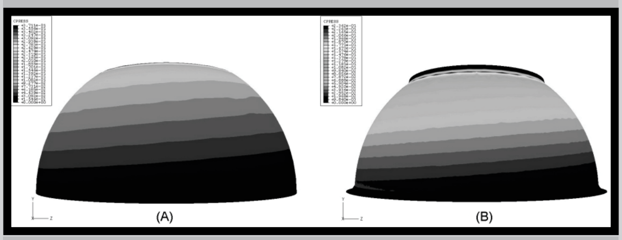

FIG.5.

Discussion and conclusion

Decrease of the contact stress gradient by way of modification of the non-weigh bearing area was succeeded. The maximal value of the contact stress was increased by the hole in the non-weigh bearing area. This value was on the edge of the hole. We substituted this stress concentrator and a cup margin by fillet edge. We tried to decrease maximal value of the contact stress by this trim. We tried to decrease enhancement of the contact stress by fillet edge. The maximal value of the contact stress was decreased by this fillet. The resultant contact stress distribution was not negatively affected by this trim. The results are in the FIGURES 1-5. The fillet inside edge of the cup and edge of the hole in the non-weigh bearing area was acquired more homogenous contact stress distribution. The contact stress distribution and maximal value of the contact stress is influenced by the different geometry of a contact surface. This fillet decreased maximal value of one third. We found out from comparison of results of the distribution of the classic cup that fillet edge did not influence gradient of the resultant contact stress distribution. The change geometry influence maximal value of the contact stress distribution only.

Acknowledgement

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DEGRADABLE SCAFFOLD MATERIALS FOR CARTILAGE REGENERATION

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Abstract

Two scaffolding materials for cartilage regeneration were produced from poly(L-lactide-co-glycolide) (PLG) and PLG modified with sodium hyaluronate (PLG-Hyal). The scaffolds were characterized in terms of their microstructure and surface chemistry. Biological properties of the scaffolds were also evaluated by implantation of the scaffolds into auricular cartilage of the rabbits for 1 and 4 weeks. Histological and histochemical examinations show that both scaffolds promote regeneration of the cartilage, although the quickest regeneration was found after implantation of PLG-Hyal.

[Engineering of Biomaterials, 69-72, (2007), 3-5]

Introduction

Defects and diseases of cartilage (articular, nasal, aural) have been often treated surgically with the use of autografts, allografts or artificial biomaterials. For example in the treatment of auricular reconstruction silicone implants were the most frequently utilised, but they have gradually lost favour due to unacceptably high rates of infection and extrusion [1]. Recently porous polyethylene is the most widely used, because of its lower tissue reaction and better resistance to collapse [2]. It was shown that resorbable polymers such as polyglycolide and polylactide support chondrocyte adhesion and formation of cartilaginouslike tissue in dynamic conditions in vitro [3]. In order to better mimic the structure and composition of the cartilage extracellular matrix hyaluronic acid or hyaluronyan / polymeric scaffolds are also produced [4].