A COMPOSITE BASED ON GLASS FIBERS AND SILOXANE MATRIX AS A BONE REPLACEMENT

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A composite material, based on glass fibers and a siloxane matrix, was designed for application in bone surgery. On the composite surface, open pores were produced, the sizes of which were in the range: i) 0.2-0.4 mm, ii) 0.4-0.6 mm, and iii) >0.6 mm, and the matrix was enriched with hydroxyapatite (HAp) powder of 5 μm grain size. Samples of the composite were tested mechanically and for wettability, and their biological properties were tested in vitro and in vivo, complemented by tests of the composite-bone tissue interfacial bonding strength. This paper suggests a simple method for preparing a composite based on inexpensive biologically-neutral components, which displays mechanical properties similar to those of human bone and can be used in the form of elements where no adherence to the human bone is desirable, e.g., bone plates for internal osteosynthesis of long bones. When the composite structure is modified by adding HAp and creating optimal open pores (0.4-0.6 mm) on the composite surface, stimulation of the adhesion of the bone cells to the composite increases and this can be successfully used in applications of fill-in and replacement elements, e.g., in osteoporous bones.

INTRODUCTION

Human bone is a perfect biological composite, consisting mostly of nano components, which provide it with unique properties. Replacing this with a natural material, i.e., the patient's own bone requires an additional operation, or, in the case of a donor, there is danger of infection. Mankind has been trying to produce artificial bone replacements for thousands of years. Scientists have always tried to develop materials that will display mechanical properties as close as possible to those of human cortical bone (Young's modulus of 12-20 GPa, tensile strength of 50-150 MPa, compressive strength 170-193 MPa [1-3]) while being biologically neutral (for use as joining elements - bone plates). In recent years, however, efforts have focused on designing materials showing biological activity, which involves stimulating the in-growth of osteocytes into implants. Moreover, bone, as an ideal composite, is not only a mechanical support for the organism, but also serves as a reservoir for minerals, particularly calcium and phosphorus [2].

Monolithic materials already used in clinical practice are metals, polymers, ceramics and glass. Metal replacements, which have most frequently been used until now, based on titanium, titanium alloys, stainless steel, chromium alloys and cobalt, display Young's modulus values of 110-210 GPa and flexural strengths of 300-1200 MPa. Their stiffness is too high, and they may sometimes cause sores and subsequent spongilisation of the bone, and they are frequently subject to corrosion in the body fluid. Polymer replacements, e.g., polylactides, polycaprolactans, polyethylenes and polyurethanes are biologically compatible, but display very low Young's modulus values (0.02-3 GPa) and relatively low tensile strength (16-60 MPa). Ceramic materials, e.g., HAp, zirconium oxide or aluminium oxide, bioactive glasses or bioactive glass ceramics display high mechanical strength, particularly under pressure (900-3900 MPa), with Young's modulus values of 80-390 GPa and flexural strength of 50-1300 MPa, while also being fairly bioactive, but their brittleness is a disadvantage [3, 4]. Due to this fact, various reinforcements have been used to improve their applicability. A large group of HAp biocomposites has been used, e.g. HAp particles/PE [5, 6], or composites based on HAp particles with reinforcing materials PLLA [7, 8] or PEG, PBT [9]. Many reinforcements, including fibers, whiskers and nano-particles have been used [4]. The application of HAp in the form of a composite material

provides an increase in the toughness and strength of HAp ceramics. Moreover, they can be fabricated to control the biological properties of the implant. Unfortunately, there are also some disadvantageous effects due to the increase in the elastic modulus and, therefore, insufficient loading of the bone around the implant [4]. Sufficient bioactivity is exhibited by bioactive glasses (most of them are silicate glasses containing 40-55 mol.% SiO_2, a significant amount of CaO, often some Na_2O and less than 10 mol% P_2O_5). Bioactive glass ceramics are other progressive materials; their constituents are the same as those for bioactive glasses, but with a higher P_2O_5 content and a lower Na_2O content. The application of bioactive glasses and bioactive glass ceramics is limited by higher Young's modulus in comparison with that of human bone (e.g., 3-4 times higher in the case of silicate-based glasses), but they have been widely used to produce bioactive bone cement and bioactive bone filler [10].

With some exceptions, composite materials are still in the stage of pre-clinical tests. In contrast to particulate composites, fibrous composites (i.e., uni-, bi-, or multi-directional) can be tailored, by selecting their reinforcement and matrix, to display mechanical properties close to those of human bone. Composites based on polymers are mostly composed of carbon fibers in the form of short or infinitely long fibers, or fabrics embedded in a thermoset or thermoplastic matrix. Materials composed of carbon or glass fibers and epoxy resin are typical representatives of thermoset composites. However, when these composites are introduced into tissue they frequently degrade to monomers, which may be toxic. Thermoplastic composites, whose reinforcement is formed of carbon fibers and a matrix of polymers such as PA, PMMA, PP, PS, PE and PEEK, are not prone to producing a monomer and can be more easily adapted to the shape of the bone [11]. Composites based on glass reinforcement, e.g., S-glass, whether unidirectional, woven or mantled, and polymers, e.g., epoxy, PETG, PMMA, have been tested as materials suitable for dentistry [12]. Carbon-carbon composites, composed of carbon fibers or fabrics placed in a carbon matrix, formed of glass-like carbon or carbonized black-coal pitch, have been tested as biological materials for more than thirty years. They display relatively high flexural strength values (250 MPa), Young's modulus of about 80 MPa, and are biotolerant. When impregnated and covered with pyrolytic carbon by the CVD method, their disadvantage, i.e. release of carbon particles, is reduced. Another disadvantage of these composites is their high price, due to multi-stage preparation [13, 14].

A successful replacing material has to display not only sufficient mechanical characteristics but has to be biocompatible and has to have specific properties leading to sufficient osseointegration. In general, biocompatibility is controlled mainly by the interface between biomaterial surface and living tissue cells. This interaction is controlled by several factors, mainly by surface morphology, surface wettability, free chemical bonds and present chemical groups or the stability and degree of micromotion between the implant and bone as well as the presence of gaps between the implant and the bone surface [15, 16, 17]. Intra- and intercellular proteins play an important role in normal tissue physiology. The integrity of connective tissues is maintained by the equilibrium between anabolic and catabolic processes which are controlled by cytokines. Changes in the equilibrium of one of these mechanisms may be reflected in the development of a destructive process. TNF-α and IL-8 cytokines were assigned autocrine as well as paracrine functions, which, in the first case, have the ability to amplify their basic effect. Cytokine IL-10 is an anti-inflammatory cytokine, and is capable of reducing the expression of other proteins even at the level of transcription. IL-8 was identified as the principal mediator of chemotaxis and is also effective in releasing enzymes from the granules of certain cells. Its production is initiated by the action of some cellular metabolites, e.g., lipopolysaccharides, and other cytokines, especially TNF and IL-1. TNF acts as the principal mediator of phlogistic processes and the development of septic shock. Its multilateral effect leads to the expression of various antigenic structures on the cellular surface, and also stimulates the production of further cytokines. TNF-α is produced mainly by macrophages. BAP is the sialoglycoprotein found in the surface of osteoblast cells. The serum level of this enzyme can be considered as a measure of the activity of osteoblasts. There have been various approaches to the improvement of the bone integration of the implant and its stability, e.g. by increasing the surface area with meshes or beads, changes in surface roughness changing, and coating the surface with some bioactive materials [18, 19].

The aim of our study was to overcome the disadvantages discussed above and to design a composite biomaterial for use as a bone replacement. We have suggested the preparation of a composite based on inexpensive biologically neutral components and the use of a sufficiently inexpensive production technology. We suggest a composite biomaterial, which displays not only mechanical properties similar to those of human cortical bone but, after treatment of the composite surface and matrix composition, also bioactivity. All these presumptions were verified by mechanical properties tests and both in vitro and in vivo biocompatibility tests with emphasis on the influence of different pore sizes on osseointegration.

EXPERIMENTAL

A basic material was produced, based on glass fabric of R-glass (21055 satin-woven fabric, R-glass, VETROTEX, Saint Gobain, France, number of fibers weft:warp ratio 1:5), and on polysiloxane matrix M130 (Lučební závody Kolín, Czech Republic). The fabric was impregnated by the matrix precursor and then cut into square-shaped pieces with dimensions of 118×118 mm. Eleven impregnated layers were placed into the curing form respecting the axis of the fibers (each layer has the same orientation of the warp with ply direction of 0°, and the fill with ply direction of 90°). This prepreg was cured under pressure of 1 MPa at a temperature of 160°C in an air atmosphere for 2 hours and finally under a pressure of 0.5 MPa at a temperature of 220°C for 4 hours. This pressing cycle corresponds to the observed temperature viscosity rise of the matrix used. Finally, the curing form with the formed piece was placed inside a laboratory oven for 4 hours at a temperature of 250°C in air at atmospheric pressure. The percentage of fiber volume (V_f) in the composite with R-glass was 65 % and the porosity was 16 %. After curing, cured plates were cut with a diamond saw to an appropriate size for further tests (see below).

With regard to the intended application of the material as a connecting, filling or replacement element in bone surgery, the basic composite was further modified. To design a successful biomaterial surface treatment, the surface chemistry and structure of the implant have to be taken into account [20]. Several approaches to the improvement of bone integration of the implant and its long-term stability were found in the literature, e.g. surface improvement (fiber-based, machined, micromachined or microetched, plasma-sprayed or coated surfaces) [21, 22]. The relation between several porous surfaces and bone has been investigated [22]. A wide range of sizes of open pores, e.g. 50-200 μm, larger than 50 or 100 μm, 200-600 μm etc., enhancing bone formation, is cited in the literature (for a review see [22]). Surface waviness and sharpness have also been discussed [23]. It should be noted that differences in these estimated values should result from variations in materials, pore shape and thickness or precise topography [23]. In our case, to increase the stimulation of the bone cell ingrowth, fractions of common salt were introduced under pressure into the uncured surface of the composite and after curing were eluted. By this change in the composite surface structure, composite specimens were produced with pores of different sizes, i.e. 0.2-0.4 mm, 0.4-0.6 mm and over 0.6 mm (verified by MahrSurf TS 50, Mahr GmbH equipment measurement). The density of the pores depending on their distribution ranged from 350 to 450 pores per cm^2. For further improvement of osseointegration, an HAp powder with particle size of 5 μm was added to the matrix before impregnation of samples intended for further surface treatment, in the amount of 10 wt.% (LASAK, Ltd., Prague, Czech Republic). For this purpose, a DI 18 Basic homogenizer (IKA Werke GMbH) was used. A weighted amount of HAp was gradually inserted into a weighted amount of polysiloxane matrix M130, so that a uniform dispersion of the HAp filler in the matrix (running speed of the homogenizer 17 500 rpm, dispersion time 5 min) was achieved. After this procedure, the glass fabric was impregnated and composite samples with a treated surface were prepared (the same procedure was used as in the case of surface-treated composite samples). This procedure aimed at increasing the stimulation of osseointegration. In this way, three different kinds of samples were prepared: basic samples without any modifications, samples with various surface porosity, and finally samples with various surface porosity with the matrix modified by the addition of HAp, see Figure 1. The effect of these modifications was studied using several methods, both in vitro and in vivo.

Tests of mechanical properties

Young's modulus (E) and the shear modulus in elasticity (G) in the direction of the fiber axis were measured with an Erudite electrodynamic resonant frequency tester (Erudite, London, UK). For both fiber directions (warp and weft), 6 samples with dimensions of 50×2.5×7 mm (length × thickness × width) were measured. Flexural modulus (E_f) and flexural strength (σ_{fM}) in the direction of the fiber axis were determined by a four-point bending setup with an Inspekt 100 HT material tester (Hagewald & Peschke, Germany) with respect to ISO 14125, and for both fiber directions 6 samples with dimensions of 60×2.5×15 mm, crosshead speed of 0.5 mm/min, load cell HT Beige 2kN, Maytec Germany, were measured. Compression modulus (E_c) in the weft fiber direction was measured with a MTS 858.2 Mini Bionix material tester (MTS Systems Corporation, USA) with respect to ISO 14126, and 6 samples with dimensions of 60×2.5×10 mm, crosshead speed of 1 mm/min, load cell MTS 662.200-05 Force Transducer (2.5kN) (MTS, USA), were measured.

Wettability tests

To assess the effect of various sizes of open pores on the wettability of the composites due to a body fluid solution, the samples were tested using the Wilhelmy plate method in a tissue culture medium (TCM) D-MEM (Sevapharma, a.s., Prague, Czech Republic).

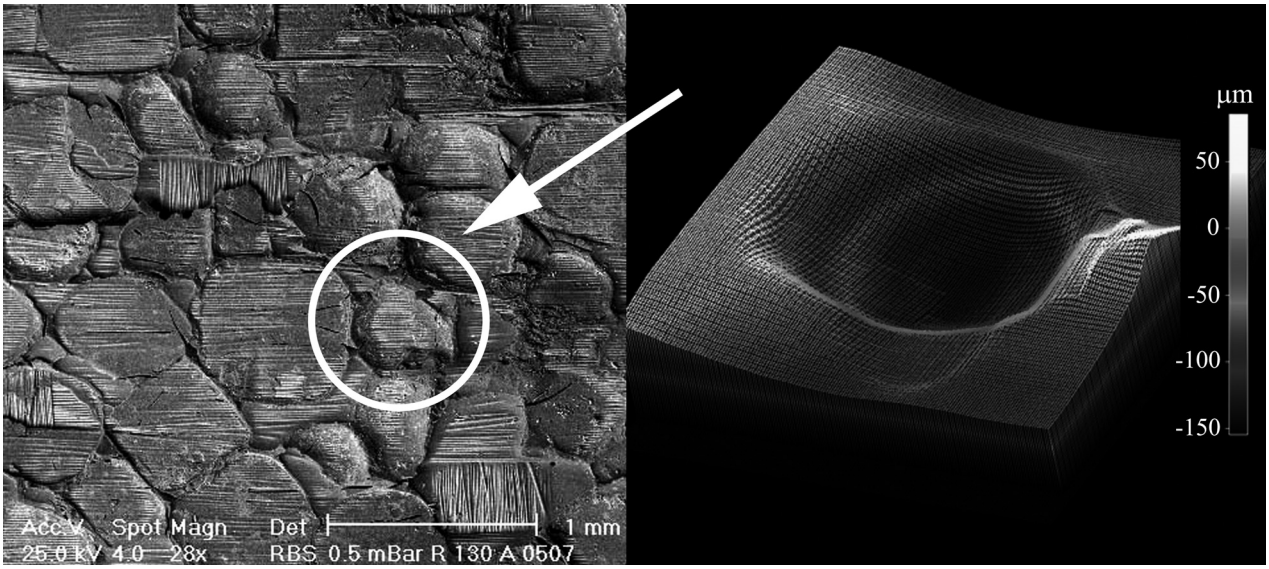

Figure 1. Surface and pore (right) structure of a composite with a modified surface (open pores 0.4-0.6 mm).

In vitro tests

The viability of osteoblasts and the synthetic activities of some cytokines were determined. Normal human osteoblasts NHOst (Cambrex Bio Science, Walkersville, MD, USA) were used. The osteoblasts were cultivated up to the 3rd passage to obtain the number of cells necessary to test all types of material simultaneously. After three days of cultivating (OGM Bullet Kit medium, Cambrex Bio Science), the cultivation medium was drained from each sample and frozen for subsequent determination of some extruded cellular mediators. The viability of cells growing directly on the materials was determined from the mitochondrial oxidation activity of the cellular mono-layer by the MTT test [24]. The cultivation medium of the osteoblasts was used to detect cytokines (TNF-α, IL-8, IL10) and the levels of bone alkaline phosphatase (BAP) were determined with the aid of ELISA kits and an Immulite analyzer (DPC, Los Angeles, USA). Briefly: the concentration of cytokines IL-8 and IL-10 were determined in a cellular medium using the ELISA method (Biosource Europe S.A., Belgium). The absorbance was measured with an ELISA SUNRISE reader (Tecan, Salzburg, Austria) at the primary wavelength of 450 nm. The concentration of cytokine TNF-α was measured immunochemically with the aid of the IMMULITE system (DPC, Los Angeles, CA, USA). The enzyme activity of bone BAP isoenzyme was detected with the aid of pNPP (p-nitrophenylphosphatase), and the intensity of the coloring was measured at 405 nm.

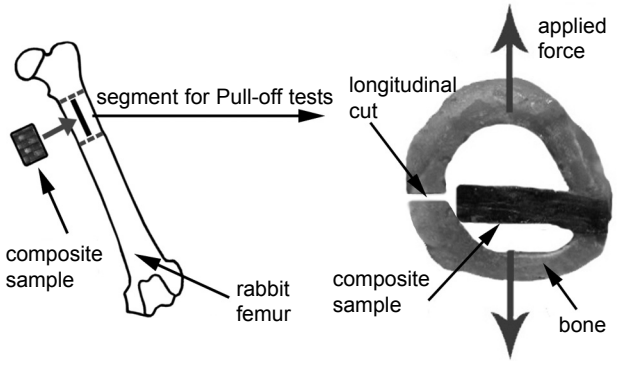

Figure 2. The measured sample for pull-off tests (based on Nakamura et al. [25, 27]).

In vivo tests

Three types of in vivo tests were applied on rabbits, Belgian Giant species (age 1 year) and minipigs, Goettingen species (age 1 year). Samples of the composite materials, size of 10×2.5×8 mm (length × thickness × width), were implanted under the proximal condyle of the rabbit's femur and into the same place of the pig's femur under sterile conditions and full anaesthesia of the animals. First a hole of corresponding size was made in one side of the bone, this being parallel to the longitudinal axis of the femoral metaphysis with the proximal end located approx. 10 mm distally to the point into which the sample of the implant was inserted (see Figure 2). The hole was made by means of a diamond-impregnated grinding wheel (of the thickness of the samples). After cleaning, the composite sample was inserted into the hole so that it protruded minimally from the bone and would not later irritate the surrounding tissue. All the above mentioned composite types were thus gradually implanted, only samples with pores of 0.4-0.6 mm with and without HAp added to the

matrix were implanted into the pigs. The animals had been bred under standard conditions according to rules applicable for the breeding of laboratory animals in the Czech Republic. The pig tests focused on histological testing: ten weeks after implanting, the animals were put down, and the implants were removed together with the surrounding tissue. The samples obtained in this way were fixed in Baker's solution and were then cut into two parts, which were tested using one of the two following independent methods:

1) the implant was removed from the bone tissue, decalcified in formic acid and sodium nitrate, and routinely deposited in paraffin. The 7μm paraffin sections obtained in this way were colored with hematoxylineosine (Sigma-Aldrich, Prague, Czech Republic).
2) the implant with the surrounding tissue was deposited in acrylate resin. After cutting and grinding into 50 μm thick sections, these were colored with hematoxylineosine.

The rabbit tests focused on pull-off tests: the rabbit samples were taken out seven weeks after implanting and were used in tests of interfacial bonding strength, derived from the Nakamura method [25]. First, the whole femur with the surrounding tissue was removed and cleaned. Using a grinding wheel, the segment containing the implant was separated from the bone by parallel cuts. This segment was treated so that the union of the two parts of the segment was effected only by the adhesion of the bone and implant. The sample treated in this way was fixed in the fixture in such a way that the composite sample was placed horizontally. An MTS 858.2 Mini Bionix testing system was used to load the sample. The force applied for the purpose of separation acted perpendicular to the composite sample at a crosshead speed of 0.5 mm/min. The maximum force required for primary separation of the composite sample from the bone was recorded, see Figure 2.

RESULTS

Mechanical properties

The purpose of the mechanical tests was to test the behavior of the composite and, with regard to the intended application, to compare it with the properties of the human cortical bone. First, we concentrated on the glass composite stiffness characteristics and the flexural strength. The results (see Table 1) indicate that the stiffness of the glass composite is comparable to that of human cortical bone and that its flexural strength can be sufficient. We assume that these properties will provide a glass composite, when used in the form of replacement or fill-in elements, with a suitable load transfer, as well as partial loading of the treated bone, which may stimulate the proliferation and suitable remodeling of the bone tissue.

Wettability properties

It is generally considered that the adhesive capacity of cells to the implant is better if its surface is hydrophilic [26]. Figure 3 shows the advancing angles that express the wettability of composites in dependence on the different pore sizes in TCM. The values of the advancing angles, ranging from 30 to 70 degrees, indicate that the surfaces of the composites are wettable, i.e. hydrophilic, suitable for bone cells in-growth.

In vitro properties

The lowest levels of the three monitored cytokines were found in samples whose surface had the smallest pores, size 0.2-0.4 mm. In this case, the strongly inflammatory TNF-α displays a lower value and chemokine IL-8 is only very slightly higher as compared with the standard reference test piece (SRTP). The anti-inflammatory cytokine IL-10, however, displays a lower value than SRTP. The most negative elements of synthetic

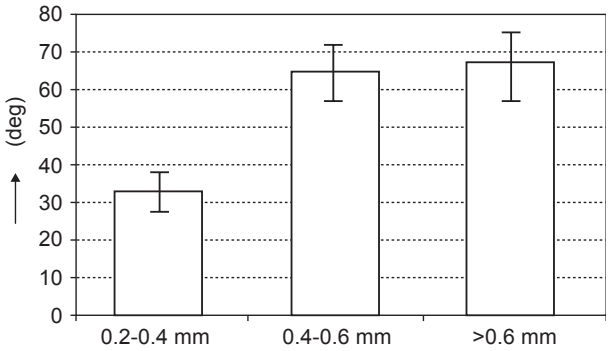

Figure 3. Advancing contact angles of composite samples (R-glass + M130) with different open pores (sizes: 0.2-0.4 mm, 0.4-0.6 mm, >0.6 mm).

Table 1. Mechanical properties of the glass composite (σ_{fM} - flexural strength, E_f - flexural modulus, E_c - compression modulus, E - Young's modulus, G - shear modulus in elasticity) and human cortical bone [1, 2, 3]

material	V_f (%)	σ_{fM} (MPa)	E_f (GPa)	E_c (GPa)	E (GPa)	G (GPa)
R-glass + M130 (warp direction)	65	390.30 (±25.3)	55.94 (±2.5)	–	53.4 (±5.8)	6.69 (±0.6)
R-glass+M130 (weft direction)		97.9 (±9.2)	12.22 (±0.8)	15.54 (±2.7)	16.2 (±2.0)	5.79 (±0.6)
human cortical bone	-	133-295	5-23	14.7-34.3	14-20	3.1-3.7

activity of the monitored tissue mediators were found in materials with no pores: a significantly increased level of inflammatory chemokine IL-8, higher content of TNF-α, and a lower level of anti- inflammatory IL-10. The level of chemokine IL-8 is higher in samples enriched with HAp as compared with the SRTP, as well as with the other tested samples, with the exception of untreated samples. The levels of the two other monitored tissue mediators also increased in this case, specifically in samples with lower surface porosity. It appears that HAp boosted the synthetic activity of osteoblasts. For a review see Figure 4.

In vivo properties

The biological behavior of replacements made of glass composite materials, monitored in vivo on laboratory animals, indicated that the studied system had good tissue tolerance. The material used was encapsulated by fibrous tissue after 10 weeks with minimal phlogistic infiltration. No separate glass fibers were found on the acrylate sections, which proves that the material is not prone to releasing fibers or to abrasion. Enriching the implant material with HAp led to thinning, and in parts even to disappearance, of the fibrous encapsulation and, subsequently, also to closer contact of the newly forming bone spongiosis with the material surface. Stronger erosion of the original bone spongiosis also results in faster remodeling of the bone with regard to the shape of the inserted material. In this case the results of the histological monitoring correspond to the in vitro finding, which proved faster metabolic activity of osteoblasts in samples with HAp, see Figure 5.

In vivo pull-off tests

Pull-off tests are a relatively simple method, easy to repeat, for measuring the interfacial bonding strength of the implant-bone interface. Nevertheless, this has its difficulties. The operation is, in its final consequence, invasive. The bones of some experimental rabbits were found to be brittle: after extracting the femur, cracks several tens of mm in length could be seen in the direction of the section. These samples had to be eliminated from the present analysis. Unlike with the method published by Nakamura, we attempted to decrease the number of these secondary cracks by creating one hole for the implant only on one side of the bone, as described above. As published by Nakamura, it was impossible to determine with sufficient accuracy the adhesion area used for calculating the stress (image analysis, software LUCIA, ver. 4.8, Laboratory Imaging, Inc., Czech Republic, was used for this purpose). Therefore, the bonding strength was not calculated, and the quality of the adhesion or osseointegration was assessed only on the basis of the failure load. The results of the pull-off tests clearly indicated an increase in the strength of the bone/implant concrescence with the composite materials with surface pores in the range 0.4-0.6 mm. A further increase could be observed in samples where HAp was added, see Figure 6.

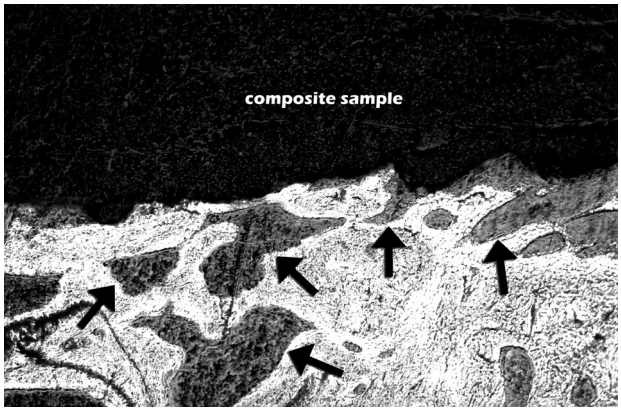

Figure 5. Photomicrograph of the interface of a pig bone and a composite implant (R-glass + M130 + HAp) with open pores (0.4-0.6 mm); the composite sample is surrounded with only a small amount of the fibrous encapsulation (arrows). Sections were stained with hematoxylin-eosin, original magnification ×150.

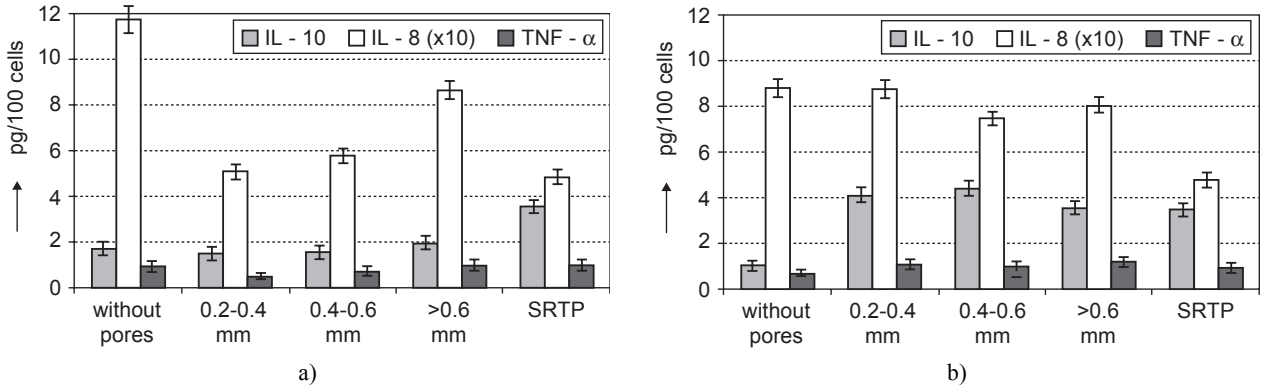

Figure 4. Results of in vitro tests, concentration of cytokines IL-10, IL-8, TNF-α calculated as weight [pg] per 100 cells; a) R-glass+M130, b) R-glass + M130 + HAp, (the values of the chemokine IL-8 levels are multiplied by ten to make them easier to read)

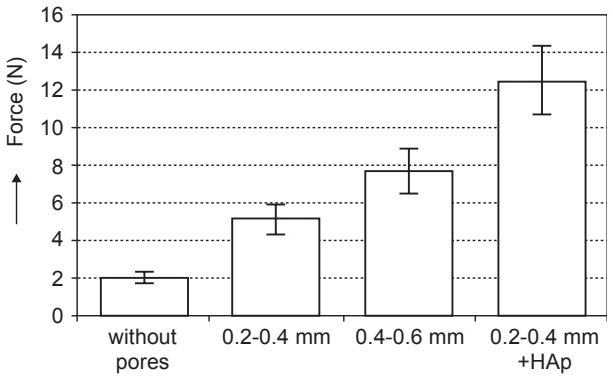

Figure 6. Results of R-glass + M130 and R-glass + M130 + HAp pull-off tests.

DISCUSSION

In vitro determination of the viability and synthetic activity of osteoblasts growing on composite materials indicates that increasing the area and roughening the surface of the materials by making a number of orifices of various sizes on its surface stimulates the cellular activity. This increase in the porosity of the surface has a favorable effect not only on the proliferation of osteoblasts but also on the stimulation of the production of anti-inflammatory cytokine IL-10, and on the inhibition of the expression of phlogistic tissue mediators. It holds that cells accept orifices 0.2-0.6 mm in size better than pores larger than 0.6 mm. Enriching the matrix with HAp leads to a moderate increase in the expression of tissue mediators, i.e. to an increase in the synthetic activity of osteoblasts, but under a simultaneous decrease in cellular proliferation.

Histological examinations of the developed composite material indicate good quality of the composite-bone adhesion, which is not connected with increased formation of fibrous encapsulation in the vicinity of the implant. Moreover, indications of stronger erosion of the original cortical bone occurred in the vicinity of the adhesion, which also led to faster remodeling of the bone according to the shape of the introduced material. A possible and limiting complication in using composite materials in general is abrasion. Histological examinations indicate that a composite based on glass - polysiloxane is not prone to abrasion. This finding, however, results primarily from in vivo tests, in which the material was subjected to the mechanical effects of the surrounding tissue only under particular conditions. Abrasion may, of course, occur under extreme conditions, e.g., if a splint (plate) breaks. This possibility must be taken into account. Apart from suitable dimensioning of the elements used, a possible solution is to use surface treatment, e.g., the application of hydrogel HEMA (poly(-2 hydroxyethyl methacrylate)), the biological properties of which are known, causing a significant reduction in material wear [14]. In this case, the solubility of R-glass, mainly due to the CaO content, has to be considered, because it could be advantageous if abrasion is released.

CONCLUSIONS

A composite material based on glass fibers and a polysiloxane matrix was developed, and displayed relatively good properties for applications in orthopedics. It appeared to be a material with good biological tolerance. Having an inert surface, it can be used in the form of elements where no adherence to the human bone is desirable, e.g., bone plates for the internal osteosynthesis of long bones. Due to its relatively low stiffness (close to that of human bone), gradual partial loading of a fractured bone and the possibility of micro-motions in the place of adhesion will lead to better formation of new bone, to faster remodeling of the bone tissue and, finally, also to cutting down the time required for the patient's rehabilitation. When the composite structure is modified by adding HAp and creating optimal open pores on the composite surface, stimulation of the adhesion of the bone cells to the composite can be increased. This modification can be used successfully in applications of fill-in and replacement elements, e.g., in osteoporous bones. It has been found that, apart from other factors such as surface energy and chemical composition [28], surface characteristics play an important role in fixing the implant in the patient's body [29]. Osseointegration, leading to a good union of the implant with the surrounding bone tissue, is thus greatly affected by the surface porosity (larger adhesion area) and also by the mechanical stability of the implant [22].

Acronyms

BAP . . . . . .Bone Isoenzyme Alkaline Phosphatase
CVD . . . . . .Chemical Vapour Deposition
HAp . . . . . .Hydroxyapatite
HEMA . . . . . .Poly(-2 Hydroxyethyl Methacrylate)
PA . . . . . .Polyamide
PBT . . . . . .Polybutylene Terephthalate
PE . . . . . .Polyethylene
PEEK . . . . . .Polyetheretherketone
PEG . . . . . .Polyethylene Glycol
PETG . . . . . .Polyethylene-Terephtalate-Glycol
PLLA . . . . . .Poly(L-lactic acid)
PMMA . . . . . .Polymethylmethacrylate
PP . . . . . .Polypropylene
PS . . . . . .Polysulfone
SRTP . . . . . .Standard Reference Test Piece
TCM . . . . . .Tissue Culture Medium

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KOMPOZITNÍ MATERIÁL NA BÁZI SKELNÝCH VLÁKEN A SILOXANOVÉ MATRICE PRO APLIKACE VE FORMĚ KOSTNÍCH NÁHRAD

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Byl navržen kompozitní materiál na bázi skelných vláken a siloxanové matrice pro aplikace v kostní chirurgii. Jeho povrch byl opatřen otevřenými póry o velikostech 0,2-0,4 mm, 0,4-0,6 mm, > 0,6 mm a matrice byla obohacena hydroxyapatitem o velikosti částic 5μm. Takto připravené vzorky byly mechanicky testovány, byla měřena smáčivost připravených povrchů. Jejich biologické vlastnosti byly sledovány jak in vitro, tak in vivo a současně byly realizovány zkoušky pevnosti spojení implantovaných vzorků s kostí. Tato práce pojednává o návrhu kompozitního materiálu na bázi nenákladných biologicky neutrálních komponent. Jejím cílem je návrh kompozitu s mechanickými vlastnostmi podobnými lidské kosti a současně s inertním povrchem, nesrůstajícím s kostí, např. pro aplikace ve formě vnitřních dlah pro léčbu konkrétních zlomenin dlouhých kostí. Modifikací matrice přidáním hydroxyapatitu a současně změnou struktury povrchu vytvořením otevřených pórů (0,4-0,6 mm) vhodné velikosti lze naopak vhodně stimulovat přilnavost kostních buněk a aplikovat kompozit ve formě prvků s kostí srůstajících, např. ve formě výplňových nebo náhradních elementů (pro osteoporotické kosti apod.).