New inorganic materials for medicine

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New inorganic materials (metals and their alloys, carbon materials, ceramics, glass, and glass-ceramics) that are used in one of the branches of medicine, *viz.*, bone endoprosthetics and implantology, are considered. Attention is concentrated on calcium- and phosphorus-containing glassy and glass-ceramic materials that exhibit bioactivity with respect to a living bone. The main statements of the modern theory of bioactivity of these materials are outlined; physicochemical aspects of this theory are considered in detail. Extensive possibilities for controlling the structures, medico-biological, physicochemical, and mechanical properties of bioactive glasses, glass-ceramics, ceramics, and composites based on them are demonstrated.

Key words: bioglass, bioglass-ceramics, bioactivity, bone endoprosthetics; hydroxylapatite.

A new promising branch has emerged in medical materials science: the use of inorganic materials for production of bone implants and endoprostheses. Organic materials, most of all polymers, have been used in clinical practice for quite a long time (artificial lens and cornea in ophthalmology, heart valves, elements of the aorta and vessels in cardiology, suture materials in surgery, *etc.*[1]). However, the problem of the production of materials compatible with living bone tissue, which could be used in the future for the development of artificial bones, has not been fully solved. The development of technology of new biocompatible and bioactive inorganic materials, physicochemical methods for the investigation of these materials, and methods for their clinical testing, as well as improvement of the design of endoprostheses based on the materials in question makes it possible to expect that this urgent problem will be successfully solved.[2–6]

Classification of inorganic materials for bone endoprosthetics

Inorganic materials to be used for bone prosthetics should possess a number of properties and meet certain requirements. Therefore, the primary task was to summarize them:

— *medical and biological requirements:* non-toxicity and biological compatibility, *i.e.*, the materials should not cause immunological, carcinogenic, or bacteriological effects, inordinate or inflammatory responses in the organism, and malignant degeneration or injury to tissues;

— *chemical properties:* controlled chemical stability in the physiological medium of the organism, which would ensure a required "life time" of the implant or prosthesis (ranging from its complete biodegradability over a certain period of time to long-term bioinertness), resistance to oxidation, which would preclude accumulation of harmful transformation products in organs, and the absence of galvanic electrical phenomena, leading to metallosis of surrounding tissues;

— *mechanical properties:* proximity of the strength and elastic characteristics of the materials to those of a living bone tissue, high long-term strength (durability) under the action of the physiological medium and under static and dynamic loads;

— *technological requirements:* the possibility of producing easily articles of diverse shapes and sizes and controlling the structure including porosity; machinability when needed; and economical efficiency.

Inorganic materials that conform to these requirements to one degree or another can be divided into several groups:

I. **Metals and alloys:** first of all, titanium and its alloys (VT-0, VT6-S, LTS 314) and cobalt-chromium-molybdenum alloys ("Komokhrom", "Endokast").

II. **Carbon-based materials:** glassy carbon, graphite, carbon-containing syntactic foams ("Intost"), carbon felt and nonwoven materials ("Karbotekhtim"), graphitized fabric (TGN-2M) and materials based on it ("Ostek"), carbon-fiber reinforced plastic composites, *etc.*

III. **Bioinert ceramics:** polycrystalline corundum ceramics, an alumina-based ceramic material ("Bioloks"), sapphire single crystals.

IV. **Bioactive materials:** calcium phosphate and calcium phosphorsilicate glasses ("Bioglass"), glass-ceramics ("Ceravital", "Bioverit", "Cerabone", *etc.*), and hydroxylapatite ceramics.

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Stainless-steel endoprostheses, which do not possess sufficient corrosion resistance in an organism, have given way to **titanium and cobalt-chromium-molybdenum** bone endoprostheses, mostly for the hip joints and ankles. Due to the formation of an insoluble protective titanium dioxide layer, titanium prostheses undergo almost no corrosion and remain inert with respect to surrounding tissues. Titanium is characterized by high mechanical strength (Table 1) and, owing to its low density, it surpasses other metals in specific strength.

Development of industrial methods for titanium casting, which allow production of various founded models and for the formation of three-dimensional porous structures of the "coral—metal—standard" or "coral—metal—forte" type or granular coatings on the surface of articles, solves the problem of manufacturing endoprostheses having complex shapes and modified porous surfaces, which make it possible for bone cells (osteoblasts) to penetrate into them.[7,8] The above properties were beneficial for the wide use of titanium and its alloys in endoprosthetic practice.

Table 1. Properties of natural bone and materials used for endoprosthetics

Characteristic	Bone	Titanium alloy	Dense sintered corundum ceramics	Bioglass-ceramics
Density	—	4500	3900	2500—2700
Mohs hardness	—	5	9	5—7
TLEC* /$K^{-1} \cdot 10^{-7}$	—	99	60—80	80—100
Strength /MPa:				
compression	90—170	860—970	3900—4900	1200—2600
bending	120—180	390—440	490—590	120—260
Modulus of elasticity /GPa	15.5—18.0	110—130	300—400	70—100
Encapsulation with connective tissue	No	Yes	Yes	No
Probability of intergrowth with bone	Yes	No	Yes	Yes
Rupture strength of the "implant—bone" connection/MPa	—	—	0.14	1.5—40
Time needed for intergrowth with bone /a week	—	—	25	4—8

* TLEC is temperature linear expansion coefficient.

Zirconium is another non-traditional metal that should be mentioned among the most promising metals for bone endoprosthetics. Its alloys with niobium possess good mechanical properties and high plasticity, and, what is more important, they cause minimum tissue response after implantation, among the metals and alloys studied.

The second group of new materials for endoprosthetics includes a number of **carbon materials and composites** based on them.[9] Carbon meets the requirement of inertness with respect to an organism, and the products of its wear exert no harmful effect on the adjacent tissues. The surface energies and electrochemical potentials characterizing graphite, glassy carbon, and materials based on them are close to those of living tissues, which confirms the high biocompatibility of carbon implants and endoprostheses. The surfaces of the manufactured articles are readily polished; this ensures the high antifriction properties of these materials and their low wear during operation. This is especially important in the construction of prostheses for hip joints. The diversity of the material structure (foams, fabrics, felts, *etc.*) is also important.

Materials based on alumina of various sorts have been used in bone endoprosthetics for a fairly long period. This is due to their high bioinertness. This group comprises dense sintered alumina ceramics, which surpasses many materials in its strength characteristics (see Table 1), porous alumina ceramics, which is capable of forming a biomechanical connection with a bone upon prolonged contact with it, and single crystal alumina (sapphire), which is used for manufacturing stomatological implants. Apparently, the group of bioinert ceramic materials would be supplemented in the nearest future by a ceramic material made of stabilized zirconia and by a number of alumina- and zirconia-based composite materials.

Representatives of the three groups of materials listed above are widely used in clinical practice for the manufacture of a variety of bone endoprostheses and implants both in Russia and abroad. However, these materials differ substantially in chemical nature from living bone tissue and form no biochemical bonds with it. Therefore, various tissues (granulation, connective, or osteoid) arise around the implant with time, their type and predominant development being determined by the chemical nature of the implant material.[10] These tissues form a so-called capsule around the implant, which prevents the bone from regeneration and from growing together with the implant. In the best case, no encapsulation of the implant takes place, and its surface occurs in intimate contact with the bone; however, cohesion is ensured only by mechanical intergrowth of bone cells into the implant pores, if the latter exist. When there is no biochemical mechanism of binding, the strength of the "implant—bone" contact is low (see Table 1).

Numerous studies carried out in recent years in various scientific centers[11—13] have shown that bio-

chemical binding of a living bone with an implant or endoprosthetic appliance and stimulation of osteogenesis (growth of a new bone) can be achieved only by using so-called **bioactive materials**, which form the fourth group in the above-presented classification. The synthesis of materials of this group is based on several fundamental statements and theories: the theory of biochemical activity of calcium-phosphate materials, the theory of sintering of ceramic materials with controlled porosity, the ion-exchange theory of chemical stability of glassy materials, and the theory of oriented crystallization of glasses. The main principle used in the manufacture of bioactive materials is that the chemical and phase composition of the mineral part of a natural bone are reproduced by artificial means.

The bone tissue of a living organism is known to consist of the mineral phase (70%), organic phase (22%), and water (8%). The ash resulting from burning a bone contains calcium phosphates (85.0%), calcium carbonate (10.0%), magnesium phosphates (1.5%), and calcium fluoride (0.3%). The phase composition of the mineral part of a living bone consists of crystals of hydroxylapatite, whose composition differs somewhat from the stoichiometric composition, which is $Ca_5(PO_4)_3OH$. Therefore, bioactive materials are synthesized using calcium- and phosphorus-containing systems in which formation of both glassy and polycrystalline structures is possible.

In terms of the type of their interaction with the physiological medium of the organism, these materials can be divided into *biodegradable* (bioglass, tricalcium phosphate ceramics) and *surface active* (hydroxylapatite ceramics, bioglass, bioglass-ceramics, biocomposites) materials. Biodegradable materials are gradually dissolved in an organism and intensify processes of osteogenesis (growth of a new bone). These materials are used for short periods as matrices for the regeneration of bone tissue. Surface-active materials are characterized by a controlled surface solubility, accompanied by chemical reactions and physicochemical and biochemical processes resulting in the formation of calcium phosphates and, subsequently, new bone tissue on the implant surface. After some period, the interface between the implant and the living bone completely disappears. The strength of the "implant—bone" connection reaches 40 MPa (see Table 1). At present, materials of this class are of the greatest interest.

The main statements of the theory of bioactivity of inorganic materials

Numerous studies carried out in various scientific and clinical centers in various countries have been devoted to the theoretical grounds of the behavior of bioactive materials in a living organism.[14—16] As a result of these studies, two main conditions were formulated, necessary for inorganic materials to exhibit bioactivity:

first, a transient amorphous-crystalline zone incorporating a polycrystalline layer with an apatite-like structure, close to that of bone hydroxylapatite, must be formed on the surface of the material;

second, biochemical processes that involve collagen, proteins, pectins, and macrophages and cause osteogenesis in the transient zone must occur on the surface of the material.

If we take these conditions into account, it becomes clear that a bioactive material should contain calcium and phosphorus oxides, which constitute the basis of the mineral part of a bone (recently, Japanese researchers have found that in the case where the implant material contains no phosphorus oxide, phosphate ions can be supplied from a physiological solution or artificial plasma in which implant is placed for some period);[17] it should possess a specified level of solubility, which ensures the diffusion of calcium and phosphorus-containing ions into a living organism; and local areas of nucleation of the apatite crystals should exist on the material surface.

In terms of physical chemistry, the mechanism of binding of a biomaterial with a bone involves the following main surface phenomena and properties:[18]

ion-exchange mutual diffusion of components of the cation sublattice of the material and components of the surrounding environment (leaching);

hydrolysis and condensation of the silicon- and oxygen-containing structural network of the material;

dissolution, *i.e.*, transfer of components of the material to the surrounding medium;

precipitation of components of the surrounding medium, mostly calcium and phosphorus, on the surface of the material, following supersaturation of the environment caused by the material dissolution;

heterogeneous (involving the surface of the material) nucleation of the apatite crystals;

structural transformations in the near-surface layers of the material: formation of pores, filling of the pores with the physiological medium, growth of the apatite crystals, *etc.*

These processes result in the formation of a transient zone at the "implant—bone" interface; this zone is characterized by clear-cut gradients of concentrations of individual components (Fig. 1).[3,15] This zone is nonuniform over its thickness: an amorphous silica layer depleted in other components forms directly on the implant surface; then a calcium phosphate layer having an amorphous structure is located; this layer crystallizes with time (7—10 days) thus giving an apatite-like polycrystalline structure. The formation of the apatite-like layer on the surface of bioactive glassy or glass-ceramic materials is shown schematically in Fig. 2.

These physicochemical processes, which can also be observed in *in vitro* experiments, are accompanied in a living organism by biochemical processes occurring on the surface of a biomaterial and involving organic compounds and living cells. They have an effect on the

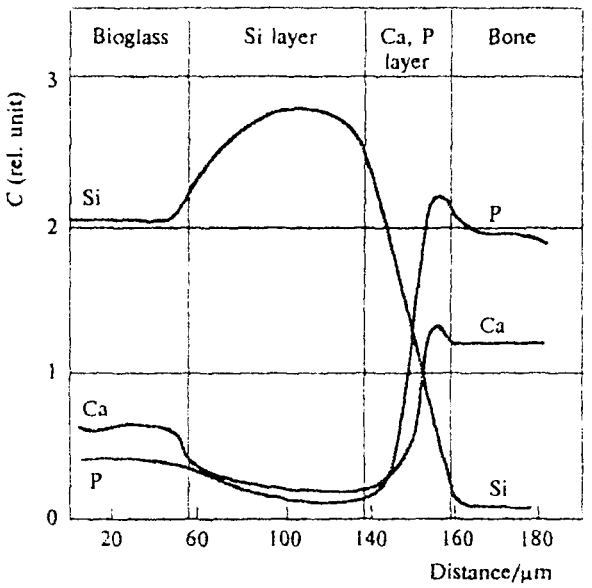

Fig. 1. Distribution of the components at the "bioglass—bone" interface.

dissolution of the implant and on the formation of the apatite-like layer and also ensure osteogenesis. In the presence of macrophages, the corroding effect of the physiological medium on the surface of biomaterial is enhanced, since macrophages scavenge some elements of the implant. The collagen fibers participate in the formation of the apatite-like layer by interacting structurally with apatite agglomerates. The 80—100 nm-thick zone thus arising, which is enriched in organic compounds, is mineralized with time and forms the basis of the new growing bone.[19—21]

The above-considered physicochemical and biochemical processes that occur on the surface of a biomaterial lead to quick replacement of the soft tissues adjacent to the material by a bone tissue (Fig. 3) and ensure high strength of the "implant—bone" connection. For example, apatite-wollastonite bioglass-ceramics and a bone grow together between the second and fourth weeks after implantation. After 25 weeks, the rupture strength of the connection reaches 1.5±0.3 MPa, and its compression strength is 1.7±0.2 MPa, which far exceeds the strength of the connection between a bone and bioinert corundum ceramics (which is no more than 0.5 MPa).[22] The rupture strength of the connection between a bone and the "Ceravital" bioglass-ceramics can be as high as 2.2—3.9 MPa, while its shear strength reaches 17 MPa. A year after implantation, the contact formed between the implant and a bone can withstand a load of up to 100 MPa.[10]

Physicochemical aspects of bioactivity of inorganic materials

Let us consider briefly the role of the physicochemical processes mentioned above in the formation of the apatite-like layer and the possibilities of controlling these processes.

Solubility of biomaterials. An extremely significant feature of glassy and glass-ceramic biomaterials is that their solubility can be controlled by varying their chemical composition and structure. The relationship between these parameters and the behavior of the material in a physiological medium has been elucidated. For example, it was shown by Hench,[3,11,23] whose studies cited have become classical, that the $Na_2O—CaO—SiO_2—P_2O_5$ sys-

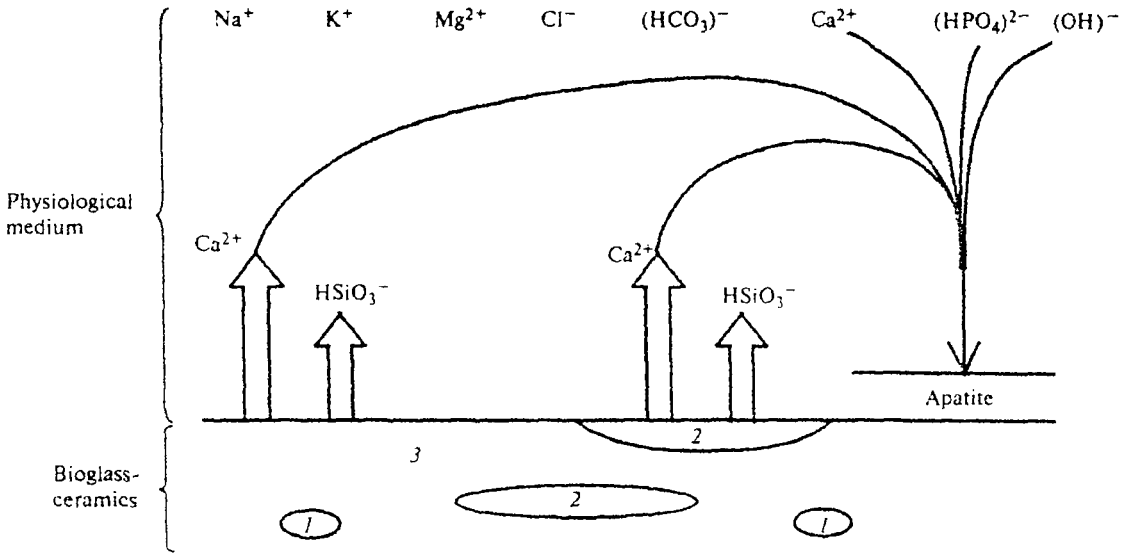

Fig. 2. Scheme of the formation of the apatite-like layer on the surface of the A—W—GC bioglass-ceramic material in a physiological medium. Bioglass-ceramics: apatite (*1*); wollastonite (*2*); glass phase (*3*): MgO, 16.6%; CaO, 24.2%; SiO_2, 59.2%.

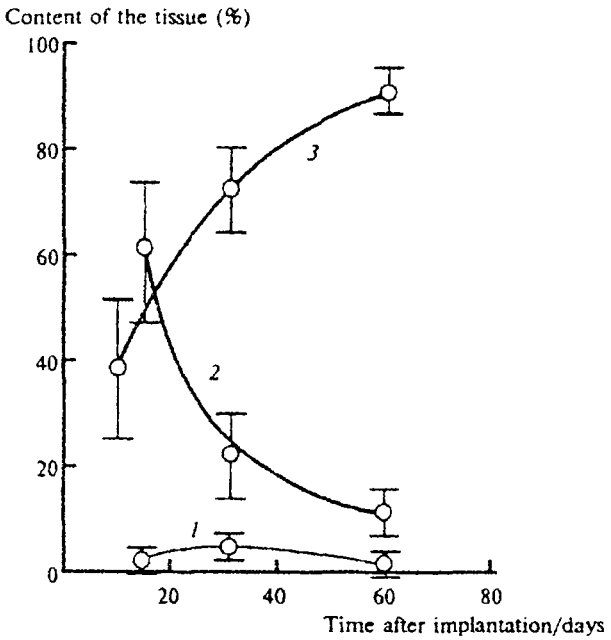

Fig. 3. Contents of various tissues on the surface of a bioglass-ceramic material after implantation: connective (*1*); soft (*2*); bone (*3*).

tem, which is the basis for many bioglasses and bioglass-ceramics, includes soluble glasses that are completely biodegraded in a physiological medium and glasses with surface-controlled solubility that form the group of bioresistive glasses. Similar glasses have also been found in other silicate and calcium phosphate systems. The introduction of Al_2O_3, B_2O_3, TiO_2, CaF_2, MgO, and other components into a bioglass affects substantially not only the level but also the mechanism and kinetics of its solubility. For example, the presence of Al_2O_3 in a sodium silicate glass leads to dissolution of the protective leached layer on its surface (when the Al : Si molar ratio ranges from 0.15 to 0.25) up to its complete dissapearance (when the Al : Si ratio is larger than 0.25), so that the solubility of the glass increases. Conversely, the introduction of 5 mol.% Al_2O_3 into calcium phosphate glasses and glass-ceramics, according to our results, increases substantially their resistance with respect to aqueous and physiological media, which is due to "cross-linking" of the phosphate structural units in these materials with the modifying Al^{3+} ions.

This fact made it possible to synthesize bioresistive glass-ceramics in the metaphosphate region of the $CaO—P_2O_5$ system, which generally possesses relatively low chemical stability.[24]

The processes of phase separation (liquation, crystallization), which determine the microstructure of glasses and glass-ceramics and the phase composition of the latter, also have an effect on their solubility and bioactivity. For example, we found[18,25] that variation of the concentration of phosphorus oxide in sodium calcium silicate bioglass in the 1—9 mol.% range makes it possible to obtain homogeneous single-phase glasses, liquating glasses, and glasses that crystallize during manufacturing. Besides, both the level of solubility of a material and the mechanism and kinetics of its interaction with an aqueous medium vary as functions of its microstructure. Correspondingly, quantitative changes occur in the formation of the apatite-like layer on the surface of the material after its treatment in physiological media. The above parameters are correlated with one another (Figs. 4 and 5).

The solubility and bioactivity can also be controlled by varying the specified macrostructure of glasses and glass-ceramic and ceramic materials, which is achieved by selecting the technological expedients that ensure the preparation of porous structures. The manufacture of

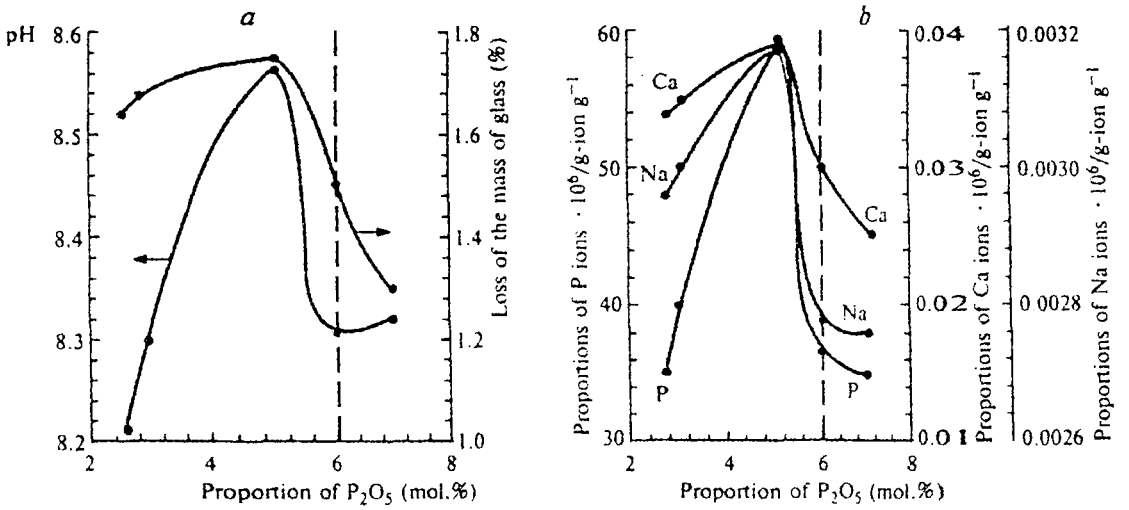

Fig. 4. Solubility of single-phase (2—6 mol.%), liquating, and crystallizable (≥6 mol.%) sodium calcium silicate bioglasses as a function of the content of P_2O_5: *a*, pH of a solution and weight loss of glass; *b*, elution of P, Ca, Na ions.

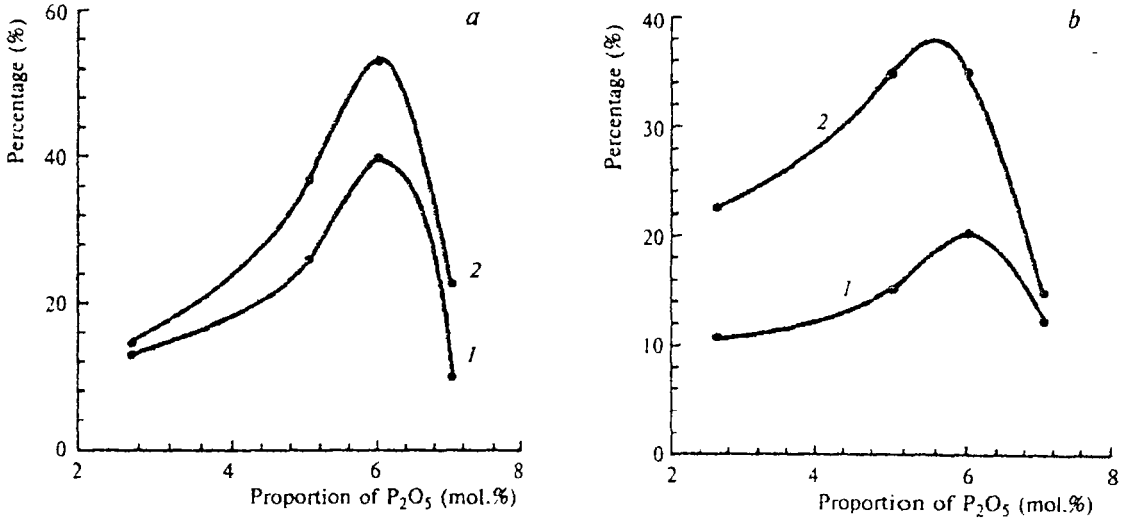

Fig. 5. Content of hydroxylapatite on the surface of sodium calcium silicate bioglasses as a function of the proportion of P_2O_5 and their microstructure (single-phase, 2—6 mol.%, liquating and crystallizable, ≥6 mol.%): *a*, treatment in a physiological solution for 14 (*1*) and 30 (*2*) days; *b*, treatment in artificial plasma for 7 (*1*) and 14(*2*) days.

porous structures with opened porosity and controllable pore sizes is based on the ceramic technology for sintering of materials with or without fillers. Due to the extensive phase interface, porous materials are more soluble, which makes it possible to increase their bioactivity. The relationship between the bulk mass (M) of a material (which is inversely proportional to the porosity), dispersity of the initial raw materials (S_{sp}), and mass losses (L_M) after treatment in a physiological solution can be illustrated in relation to a composite biomaterial that we synthesized by sintering an artificial hydroxylapatite with glass (Fig. 6).

Hydration of the surface of biomaterial and nucleation of apatite. On the surface of a biomaterial placed in a physiological medium, hydrolysis and condensation processes occur. Silicate materials are hydrolyzed to give a gel-like high-silica layer saturated with OH^- groups. This layer plays an important role in the mechanism of

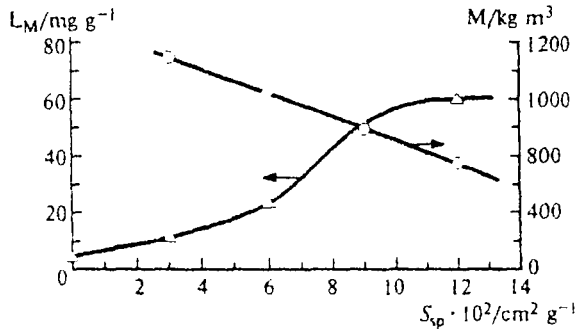

Fig. 6. Solubility (L_M) and bulk weight (M) of a porous bioglass-ceramic material based on hydroxylapatite as a function of dispersity of the initial raw materials (S_{sp}).

apatite formation. It is believed that the Si—OH groups present in a structure act as sites for preferable heterogeneous nucleation of apatite crystals.[10,15,26] This is supported by the fact that when various materials are treated in an artificial plasma (which contains ions needed for the apatite formation), apatite crystals are mostly formed in the case where the material contains silica in the bound state (glass, glass-ceramics) or in the free state (silica gel prepared by sol—gel technology). Heat treatment of a hydrated siliceous gel at 900 °C (dehydration) hampers the formation of apatite on its surface.[26,27] Regarding bioactivity of a material, the most preferable content of SiO_2 in it is 46—55 mol.%. In this case, a hydrated silica layer is formed initially on the surface of an implant, and after that, an apatite-like layer is formed. At low contents of SiO_2 (less than 40 mol.%), these two layers are formed simultaneously, while when the content of silica is high (more 60 mol.%), only the silica layer appears.[15]

The latest studies have shown that the formation of apatite crystals on a hydrated silica surface is only a special case of more general regularities of nucleation of apatite. In fact, development of an apatite-like layer has been observed on hydrated surfaces of titanium-containing materials (a titanate gel prepared by sol—gel technology, titanium-containing glasses) enriched in Ti—OH groups.[28] Thus, apparently, various structural units of the M—OH type can initiate the nucleation of apatite. These assumptions provided the basis for the synthesis of bioactive materials from metallic titanium and organic polymers whose surface has been specially treated in aqueous solutions.[26]

Crystal growth in an apatite-like structure. The chemical composition of bone hydroxylapatite, which forms the basis of the mineral part of a bone, is known

to differ somewhat from the compositions of its natural and artificial analogs in the Ca : P ratio, which can vary in the 1.5—1.67 range and also by the presence of foreign ions, Na^+, K^+, Mg^{2+}, $(CO_3)^{2-}$, $(SO_4)^{2-}$, and F^-, whose overall content can exceed 5%. The apatite crystals formed on the surface of a biomaterial in a living organism or following treatment in an artificial plasma capture the same ions during growth. Therefore, in this case, one should speak of the formation of an apatite-like structure similar to a bone. Moreover, in our experiments, we found that during growth of the crystals, crystal lattice parameters vary as functions of the time that the implant has spent in the physiological medium. The composition and acid-base characteristics of the medium exert a crucial effect not only on the structure of crystals but also on the kinetics of their growth. In water or in a buffer solution (pH 7), the rate of formation of the apatite-like layer is relatively low: this layer has been detected on the surface of a phosphorus-containing sodium calcium silicate glass only after 14 days of treatment; in an artificial plasma, its formation has been observed already on the seventh day.[25]

Types of bioactive materials

At this time, a lot of bioactive glassy, ceramic, and glass-ceramic materials have been developed, and their number is permanently increasing. These materials include:

— bioglass and materials based on it;

— bioceramics based on crystalline calcium phosphates;

— bioglass-ceramics;

— bioactive composite materials;

— bioactive coatings on metals, alloys, and corundum ceramics.

Bioglass, which was developed by Hench in the 1970s,[3,11,23] have become the first object used to demonstrate the unique ability of calcium- and phosphorus-containing materials to grow together with a living bone tissue. This glass was comprehensively studied by physicochemical, structural, morphological, medical, and biological methods and introduced in the practice of bone endoprosthetics. Due to the high bioactivity of this material, the strength of its connection with a bone can be as high as 40 MPa.

The aims of the subsequent studies in the field of bioglasses were to modify their chemical compositions, to develop methods for their strengthening, and to synthesize porous bioglasses and compositions with other bioactive materials (collagen, hydroxylapatite). These studies resulted in the development of biomaterials that can be used as powders, granules, blocks, or bone elements for manufacture of medical preparations or implants for endoprostheses operating under conditions of relatively small static loads.

Extensive studies have been made in the field of production and clinical application of **bioactive ceramics**, most of all, those based on hydroxylapatite.[23,29—33] Hydroxylapatite ceramic materials are obtained by methods of ceramic technology that make it possible to control their porosity and the size of pores and to produce open "channel" structures, which favor penetration of bone cells into the implant material. The relatively low strength characteristics of porous hydroxylapatite ceramics (bending strength is about 35 MPa) can be enhanced by reinforcement of the material with zirconium, magnesium, and yttrium oxides. High-purity hydroxylapatite ceramics grow well together with a living bone, and the strength of the connection reaches 23 MPa.

Bioactive glass-ceramics are multi-phase glass-ceramic materials in which crystals of calcium phosphates, silicates, and other crystal phases are joined by a cementing glassy interlayer, the residual glass phase. These materials are widely used abroad in bone endoprosthetics, because their mechanical strengths are much higher than those of bioglasses or hydroxylapatite ceramics. In fact, the bending strength of the bioglass-ceramic material "Ceravital" is 100—150 MPa and that of "Cerabone" is 220 MPa. An important feature of bioglass-ceramics regarding their use as endoprostheses that experience dynamic loads is their relatively high resistance to cracks, which reaches 3.0 MPa m^{-2} (and is especially high for materials reinforced with ZrO_2 or Y_2O_3 particles). These strength characteristics of bioglass-ceramic materials are due to their finely crystalline structures (the size of crystals varies in the 0.5—3 mm range). The long-term strength of bioglass-ceramics under static and dynamic loads ensures a long useful life of the prosthetic appliance. For example, apatite-wollastonite bioglass-ceramics can operate in a physiological medium under a load of 65 MPa for at least 10 years, whereas glass is destroyed under these conditions after 1 minute.

Bioglass-ceramics are synthesized by methods of glass and ceramic technology; this makes it possible to obtain nonporous (cast) and porous (sintered) materials with controlled microstructures (the size and the number of crystals and their orientation) and macrostructures (porosity). The reinforcement of porous bioglass-ceramic materials with threadlike "whiskers," fibers, and carbide or oxide particles allows their strength and crack-resistance to be brought to those of corundum ceramics. The compositions, the methods of production, and properties of bioglass-ceramic materials developed both abroad and in Russia have been reported in previous papers.[3,4,6,11,18,34]

At present, bioglass-ceramics are used successfully in maxillofacial surgery, stomatology, and otolaryngology.

A promising line in the development of bioactive inorganic materials is elaboration of **biocomposites and biocoatings** based on hydroxylapatite, bioglasses, and bioglass-ceramics. In biocomposites, a bioactive phase can serve as a matrix in which the second, non-bioactive, phase (metallic, carbon, or polymeric) is dispersed or as a filler distributed in a polymeric or organic matrix.

When biomaterials are used as coatings, they are applied on the surface of endoprostheses made of stainless steel, titanium, or corundum ceramics using conventional enameling technology (slip or powder method) or highly effective methods for deposition of technological coatings (plasma method, solution method, *etc.*). Biocomposites and implants covered with biocoatings combine the properties of both phases, *i.e.*, ability to grow together with a living bone (bioactive phase) on the one hand, and high mechanical strength (metals, corundum ceramics), elasticity (high-density polyethylene and other polymeric materials), and osteoplasticity (collagen), on the other hand.

Thus, the brief characteristics of inorganic materials for medical purposes presented in this review makes it possible to conclude that there is a diversity of materials with a broad spectrum of functions for use in bone endoprosthetics and implantology.

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