Biomaterials

L. L. Hench

As living beings get older, they begin to wear out. Although many factors responsible for aging are not understood, the consequences are quite clear. Our teeth become painful and must be removed, joints become arthritic, bones become fragile and break, the powers of vision and hearing diminish and may be lost, the circulatory system shows signs of blockage, and the heart loses control of its vital pumping rhythm or its valves become leaky. Tumors appear almost randomly in bones, breast, skin, and vital organs. And, as if these natural processes did not occur fast enough, we have achieved an enormous capacity for maiming, crushing, breaking, and disfiguring the human body with motor vehicles, weapons, and power tools or as a result of our participation in sports.

A consequence of these natural and unnatural causes of deterioration of the human body is that some 2 million to 3 million artificial or prosthetic parts are implanted into individuals in the United States each year. A list of some of the devices and their function is given in Table 1. More than 50 implanted devices made from more than 40 different materials are included alone and in various combinations. Although many materials appear several times in this table there is no apparent commonality of microstructure, atomic structure, composition, or surface features.

The challenge of the field of biomaterials is that all implant devices replace living tissues whose physical properties are a result of millions of years of evolutionary optimization, and which have the capability of growth, regeneration, and repair. Thus, all man-made biomaterials used for repair or restoration of the body represent a compromise. The relative success or failure of a biomaterial reflects the scientific and engineering judgment used in achieving this compromise. The interaction of many complex physical, biological, clinical, and technological factors must be considered.

The author is professor and head, Ceramics Division, Department of Materials Science and Engineering, University of Florida, Gainesville 32611.

For example, consider the following characteristics of a natural tooth that must be satisfied in some measure to achieve a successful tooth implant: a tensile strength of 15,000 to 20,000 pounds per square inch in flexure; a biologically bonded interface with epithelial skin cells, gingival tissues, and bone, which results in a difference of more than 10^3 in elastic moduli across the various interfaces in contact with a tooth; and an attachment structure (the periodontal ligament) that converts compressive stresses applied to the tooth to tensile stress within the jawbone. Although many materials have the requisite flexural strength, no material known today can reliably achieve the stable interfacial attachments required to mimic a natural tooth. Is it any wonder that few extensive clinical studies show more than 50 percent success rates for long-term (> 5 years) dental implants?

Control over the biomaterials-tissue interface is the paramount problem in this field of materials science (*1*). The physical properties of most tissues can be matched within engineering limits by careful selection of metals, ceramics, or polymer materials singly or in specially designed combinations (Table 1) (*1*, *2*). Even the requirements that the biomaterial be nontoxic to the host tissues can be achieved relatively easily by screening of the materials with tissue culture tests or short-term implants. But, achieving the necessary match or gradient in physical properties across the interface between living and nonliving matter is a formidable scientific challenge. Part of the difficulty is that the science of adhesion of biological interfaces is still being developed. Until cell biologists and biochemists discover which molecular species control the bonding of cells to each other, the understanding of adherence or lack of adherence of tissues to implant devices will remain incomplete.

The field of biomaterials developed historically so as to achieve a suitable combination of physical properties to match those of the replaced tissue with a minimal toxic response to the host. This approach has led to a reasonably large catalog of "nearly inert" biomaterials (most of those in Table 1) that comprise the bulk of the 2 million to 3 million devices implanted yearly. A common feature of these materials is that they initiate the growth of a thin, fibrous capsule which separates the normal tissue from the implant.

Figure 1A shows an example of the thin fibrous capsule formed between a nearly inert biomaterial, a copolymer of methylmethacrylate and hydroxyethyl methacrylate, and the subcutaneous tissue of a rat 8 weeks after the material was implanted. The muscle and subcutaneous connective tissue are normal; the thin fibrous capsule is the only evidence that the implant has been present in the host tissue.

In contrast, a reactive material such as an acrylic acid and methylmethacrylate copolymer (Fig. 1B) produces a very thick fibrous capsule which extends throughout the subcutaneous region 8 weeks after implantation in a rat. Even the muscle tissue shows extensive inflammation resulting from the reactivity of the implant.

There is little, if any, adhesion between the implants and the fibrous capsules. Consequently, movement of the implant within the capsule can occur when stress is applied with the following possible results. (i) The capsule may increase in thickness. A thick capsule may interfere with the local blood supply to tissues (*3*) or provide a site for accumulation of biochemical by-products perhaps associated with formation of tumors (*3*, *4*). (ii) The capsule may calcify and harden. Progressively stiffening capsules around devices such as silicone breast implants produce pain and deterioration of underlying tissues because of the mismatch of mechanical properties (*5*). (iii) Localized concentrations of stress may result. Mechanical damage to the host tissue, such as the microfracture of boney spicules adjacent to the stem of a hip or knee implant, can cause pain around and progressive loosening of the implant (*6*). More motion results, and even larger stress concentrations occur until either the bone or the implant fractures. (iv) The effects of infection may be magnified (*7*). An infection may occur or persist at an implant site because there is not a normal blood supply to the capsule. The lack of blood prevents the invasion of white cells necessary to attack the infection and retards the transport of cell debris away from the site of infection. (v) The capsule may separate from the device. Spalling of fragments of a poorly adherent layer from the surface of

0036-8075/80/0523-0826$01.50/0

cardiovascular implants such as heart valves, arterial or venous grafts, or from the walls of an artificial heart can result in fatal emboli (*8*). (vi) Corrosion products may accumulate. Again because of the lack of circulation, products from the corrosion of metals or deterioration of polymers can accumulate within the capsule or at the capsule-implant interface (*9*).

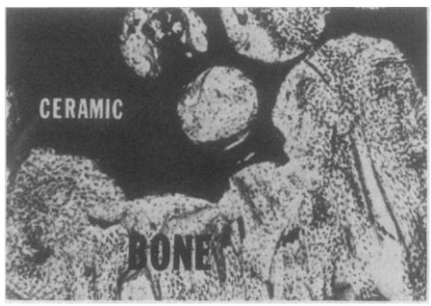

Fig. 2. Thin section of Al_2O_3 with 200-μm pores with rabbit femoral bone ingrowth in 8 weeks. [Photo courtesy of J. J. Klawitter and S. Hulbert]

Methods of Controlling Biomaterials Interfaces

Because of the interfacial problems associated with nearly inert biomaterials, much research during the last decade has been directed toward stabilizing the tissue-biomaterial interface by controlling either the chemical reactions or the microstructure of biomaterials. That the microstructure can be controlled is based on the hypothesis that tissue can grow into pores or surface depressions if the pores are big enough and if the tissue can maintain a vascular supply and tissue vitality (*10, 11*). A fibrous membrane will still interpose itself between the surface and the pores and the infiltrating tissue. However, the mechanical keying caused by the interdigitation of the living and inanimate material serves to inhibit growth of a fibrotic capsule by retarding motion and by distributing stress over a large interfacial area. Because of the large interfacial area exposed to tissue and tissue fluids it is important that porous biomaterials be especially resistant to corrosion and deterioration in the body. Bioceramics, especially aluminum oxide, have potential for microstructural control of the interface without formation of potentially toxic corrosion products. Only a few metals, that is, titanium and cobalt-chromium alloys, exhibit sufficient corrosion resistance to be considered for use in porous implants (*12*).

Quantitative analyses of the growth of tissues into pores of different sizes show that soft connective tissue will grow into pores of greater than 50 micrometers in diameter and remain healthy over periods of at least several years; bone will grow into pores bigger than 100 μm (*13, 14*). There is debate about whether bone will develop as fully mineralized tissue within a porous structure under the continual stress and micromovement associated with a functional load-bearing implant (*15*). Most studies of hard tissue ingrowth into porous materials have been done with the use of nonfunctional plugs in the long bones of dogs, and it is not known how much stress is transmitted to the plug. When stress transfer to bone via porous interfaces does occur a high modulus of elasticity by the implant can cause extensive bone resorption (*16*). Design of porous or porous surface implants must take these factors into consideration.

Another factor still open to debate is the mechanical fatigue resistance of an interface composed of a large number of thin webs of tissue and material. Figure 2 shows the microstructure of bone ingrowth into 200-μm (average) pores of an aluminum oxide ceramic implanted in a rabbit femur for 8 weeks (*17*). Since there is a distribution of the pore area filled with dense bone there must also be distribution of the localized strength, elastic moduli, and fatigue resistance of such an interface. Current studies involving modeling of microstructurally controlled interfaces with finite element analyses coupled with postmortem stress-strain testing of functional implants put into animals, should help determine the long-term utility of this type of interface control (*18*).

Microporous structures have been successfully used as vascular prosthetic devices (*19*). To ensure that blood will continue to flow without clotting through such a replacement artery it is essential that the flow surface, originally of "biomaterial," be lined by a natural lining—the so called "neointima." Once laid down this lining must be maintained and one way to do this is to establish transmural tissue growth from the outer scar tissue that includes blood vessels to nourish the inner lining. Alternative methods of maintaining this neointima, by simple diffusion of oxygen from the blood passing through and by growth of thin blood vessels into the neointima from the ends where the biomaterial joins the natural artery, are sufficient only over relatively short distances. Transmural growth seems to be desirable when longer lengths of vessel need replacement.

A second method of manipulating the biomaterials-tissue interface is controlled chemical breakdown, that is, resorption, of the material. Resorption of biomaterials appears a perfect solution to the interfacial problem because the foreign material is ultimately replaced by regenerating tissues (*20*). Ideally, there is eventually no discernible difference between the implant site and the host tissue. This method comes closest to the grafting of a patient's own tissue when

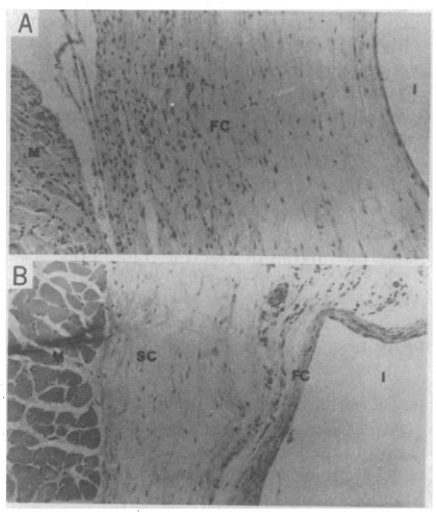

Fig. 1. Fibrous capsule (*FC*) developed between an implant (*I*) and the subcutaneous tissue (*SC*) and muscle (*M*) of a rat 8 weeks after implantation. (A) Nearly inert biomaterial: hydroxylpolymethylmethacrylate; (B) reactive biomaterial: acrylic acid-methylmethacrylate copolymer (× 100). [Photo courtesy of J. Wilson]

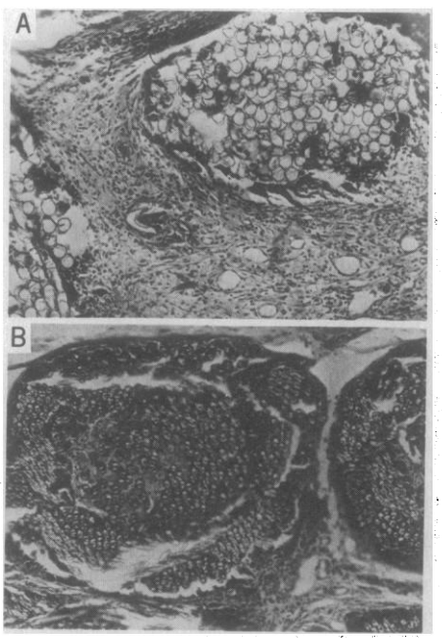

Fig. 3. Effect of time on resorption of a Dexon (PGA) suture implanted subcutaneously in rat (× 250). (A) After 2 weeks; (B) after 8 weeks. [Photo courtesy of A. Reed, J. Wilson, and K. Gilding]

Table 1. Implant devices in use or test today, their function, and the biomaterials used.

Device	Function	Biomaterial
	Sensory and neural systems	
Artificial vitreous humor	Fill the vitreous cavity of the eye	Silicone Teflon sponge; polyglycerylmethacrylate (PGMA)
Corneal prosthesis	Provide an optical pathway to the retina	Polymethylmethacrylate (PMMA); hydrogels
Intraocular lens	Correct problems caused by cataracts	PMMA (lens); nylon, polypropylene, Pt, Ti, Au loops
Artificial tear duct	Correct chronic blockage	PMMA
Artificial eustachian tube	Provide clear ventilation passage	Silicone rubber, Teflon
Nerve tubulation	Align severed nerves	Silicone membrane, porous surgical metals
Middle ear prostheses	Replace diseased bones of the middle ear	PMMA; metallic wire, Proplast (PTFE + carbon fiber); Bioglass
Percutaneous leads	Conduct power to electrical sensory devices	Nylon or Dacron velour, PMMA
Auditory prostheses, visual prostheses	Restoration of hearing and vision	Pt and Pt-Ir wires and electrodes; Ta-Ta_2O_5 electrodes, stainless steel, Elgiloy wires; silicone rubber; PMMA
Electrical analgesia	Eliminate chronic pain	Pt and Pt-Ir wires and electrodes, Ta-Ta_2O_5 electrodes, stainless steel, Elgiloy wires, silicone rubber, PMMA
Electrical control of epileptic seizure	Conduct electrical signals to brain	Same
Electrophrenic stimulation	Control breathing electrically	Same
Bladder control	Stimulate bladder release	Same
	Heart and cardiovascular system	
Myocardial and endocardial stimulation (heart pacer)	Maintain heart rhythm	Stainless steel, Ti cans; silicone rubber, wax epoxy encapsulants; Pt or Pt-Ir alloy-electrode, Elgiloy wire
Chronic shunts and catheters	Assist hemodialysis	Polyethylene, hydrophilic coatings
Cardiac heart valves	Replace diseased valves	Co-Cr alloys; low-temperature isotropic carbon; porcine grafts; Ti alloy with Silastic or pyrolytic carbon disks or balls
Arterial and vascular prostheses; artificial heart components; heart assist devices	Replace diseased arteries and blood vessels; replace the heart; augment diseased heart	Segmented polyurethanes; silicone rubber or pyrolytic carbon mandrels with Dacron mesh sheaths; heparin + GBH or TGBH coupled coatings on Teflon or silicone rubber; poly-HEMA-coated polymers; Dacron velours, felts, and knits; textured polyolefin (TP), TP with cross-linked gelatin surface; Teflon (PTFE) alone
	Skeletal system repair and replacement	
Artificial total hip, knee, shoulder, elbow, wrist	Reconstruct arthritic or fractured joints	Stems: 316L stainless steel; Co-Cr alloys; Ti and Ti-Al-V alloy; Co-Cr-Mo-Ni alloy cups: high-density, high-molecular-weight polyethylene; high-density alumina; "cement" PMMA; low-density alumina; polyacetal polymer; metal-pyrolytic carbon coating; metal-Bioglass coating; porous polytetrafluroethylene (PTFE); and PTFE-carbon coatings on metal; PMMA-carbon fibers, PMMA-Ceravital powder composite; porous stainless steel; Co-Cr; Ti and Ti alloys
Bone plates, screws, wires	Repair fractures	316L stainless steel; Co-Cr alloys; Ti and Ti alloys; polysulfone-carbon fiber composite; Bioglass-metal fiber composite; polylactic acid-polyglycolic acid composite
Intramedullary nails	Align fractures	Same
Harrington rods	Correct chronic spinal curvature	Same
Permanently implanted artificial limbs	Replace missing extremities	Same plus nylon or Dacron velours on Silastic for soft tissue ingrowth
Vertebrae spacers and extensors	Correct congenital deformity	Al_2O_3
Spinal fusion	Immobilize vertebrae to protect spinal cord	Bioglass
Functional neuromuscular stimulation	Control muscles electrically	Pt, Pt-Ir electrodes; silicone; Teflon insulation
	Dental	
Alveolar bone replacements, mandibular reconstruction	Restore the alveolar ridge to improve denture fit	PTFE carbon composite (Proplast); porous Al_2O_3; Ceravital: hema hydrogel-filled porous apatite; tricalcium phosphate; PLA/PGA copolymer; Bioglass
Endosseous tooth replacement implants (blades, anchors, spirals, cylinders—natural or modified root form)	Replace diseased, damaged, or loosened teeth	Stainless steel, Co-Cr-Mo alloys, Ti and Ti alloys, Al_2O_3, Bioglass, LTI carbon, PMMA, Proplast, porous calcium-aluminate, $MgAl_2O_4$ spinel, vitreous carbon, dense hydroxyapatite
Subperiosteal tooth replacement implants	Support bridge work or teeth directly on alveolar bone	Stainless steel, Co-Cr-Mo alloy, LTI carbon coatings
Orthodontic anchors	Provide posts for stress application required to change deformities	Bioglass-coated Al_2O_3; Bioglass-coated Vitallium

possible. Practically, there are severe limitations on achieving the combination of the requisite physical properties and resorption into chemical constituents that can be processed by the metabolic system. Polymer systems based on polyglycolic (PGA) or polylactic acids (PLA), or both, decompose into CO_2 and H_2O and are used clinically as resorbable sutures and implantable drug delivery systems (*21*). Copolymer systems composed of PGA and PLA also show some promise for use as resorbable fracture fixation devices. Resorbable calcium phosphate or apatite ceramics break down to soluble calcium and phosphate salts which are able to be metabolized within hard tissues and are being tested for use in dental reconstruction (*22*).

A problem with the chemical breakdown method is that the strength of the resorbable biomaterial decreases as resorption occurs. Unless there is close matching of the reduction in implant strength with the increase in strength of the healing tissues the implant-tissue system will fail. The physicochemical reactions associated with resorption are complex and this makes it difficult to match time-dependent physical properties. Figures 3 and 4 show that the strands of a PGA suture, Dexon (American Cyanamid), become progressively infiltrated with tissue during an 8-week period in rat (*23*). Although one can still see the suture after 8 weeks, its tensile strength has diminished to only a few percent of the original value. A progressive breakdown of the higher molecular-weight polymerized structure occurs during this period. In contrast, experimental PLA implants show no detectable degradation within the 8-week period (*23*). Thus, it might seem that copolymers of PGA and PLA could be used to carefully control time-dependent changes of strength. However, recent studies show that PGA/PLA copolymers produce wide variations in rates of resorption that depend on both the degradation of the end-members and the degree of crystallinity in the copolymer (*23*). Prediction of the time dependence of properties is still not possible without knowing more about the mechanisms of chemical breakdown of copolymers in the body and how molecular structure is related to strength.

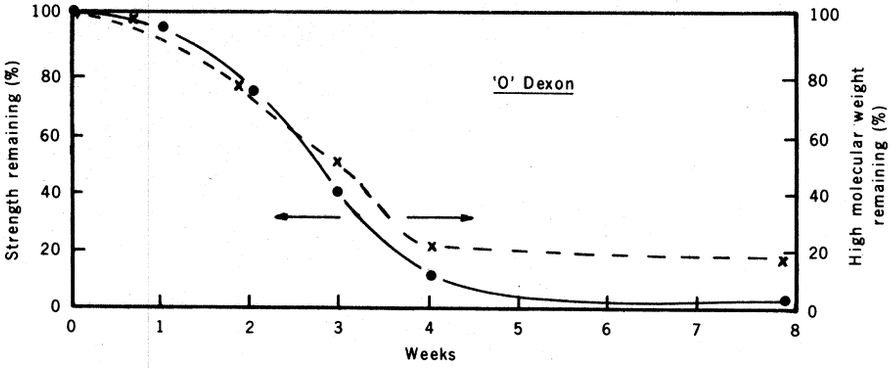

Fig. 4. Effect of implantation time on tensile strength (●) and molecular breakdown (X) of Dexon (PGA) suture in rat subcutaneous tissue in vivo. [Based on Reed (*23*)]

Resorbable polymers are too weak to be used as replacements for bones and joints. Tricalcium phosphate and calcium-aluminate-phosphate bioceramics show promise for use as hard tissue replacement (*22*), but more research is required to get the right rate of loss of strength (*24*), ensure the nontoxic metabolism of the large concentration of mineral salts, and relate the effects of microstructure to the rates of resorption for this class of biomaterials (*22*).

The third approach to control of the

Table 1 (continued).

Device	Function	Biomaterial
	Space-filling soft tissue prostheses	
Facial contouring and filling prostheses (nose, ear, cheek)	Replace diseased, tumorous, or traumatized tissue	Silicone rubber (Silastic), polyethylene, PTFE, silicone fluid, dissolved collagen fluid, polyrane mesh
Mammary prosthesis	Replace or augment breast	Silicone gel and rubber, Dacron fabric; hydron sponge
Cranial boney defects and maxillofacial reconstruction prostheses	Fill defects	Self-curing acrylic resin; stainless steel, Co-Cr alloy, Ta plates; polyethylene and polyether urethane-coated polyethylene terephthalate-coated cloth mesh
Artificial articular cartilage	Replace arthritis deterioration cartilage	Crystallized hydrogel-PVA and polyurethane polymers; PFTE plus graphite fibers (Proplast)
	Miscellaneous soft tissues	
Artificial ureter, bladder, intestinal wall	Replace diseased tissue	Teflon, nylon-polyurethane composite; treated bovine pericardium; silicone rubber
Artificial skin	Treat severe burns	Processed collagen; ultrathin silicone membrane polycaprolactone (PCA) foam-PCA film composite
Hydrocephalus shunt	Provide drainage and reduce pressure	Silicone rubber
Tissue patches	Repair hernias	Stainless steel, Marlex, Silastic, Dacron mesh
Internal shunt	Provide routine access to dialysis units	Modified collagen; Silastic
External shunt	Provide routine access to dialysis	Silastic-Teflon or Dacron
Sutures	Maintain tissue contact to aid healing	Stainless steel, silk, nylon PGA, Dacron, catgut, polypropylene
Drug delivery systems	Release drugs progressively; immobilize enzymes	Silicone rubber, hydrogels ethylene-vinyl acetate copolymer, PLA/PGA polysaccharides-vinyl polymers
Artificial trachea	Reconstruct trachea	Porous Dacron-polyether urethane mesh, Ta mesh, Ivalon sponge and polypropylene mesh

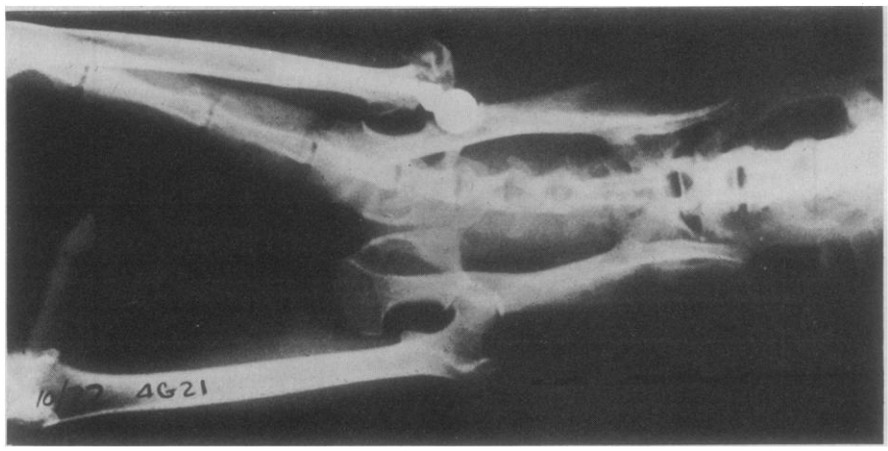

Fig. 5. X-radiograph of a Bioglass-coated stainless steel femoral head replacement in monkey after 40 weeks without use of a polymer "bone cement." [Photo courtesy of W. Petty and G. Piotrowski]

materials interface is to use biomaterials with controlled surface reactivity. In this class of biomaterials the composition is designed such that the surface undergoes a selected chemical reactivity with the physiological system and thereby establishes a chemical bond between tissues and the implant surface (*25, 26*). The desired bonded interface protects the implant material from further deterioration with time, that is, it self-passivates. Thus the potential of this approach is to combine the high strength or flexibility of nearly inert biomaterials with the surface chemical reactivity needed for tissue adherence and bonding. Ideally, interface stabilization by surface reactivity produces more flexibility in device design and fabrication than does mechanical interlocking or resorption. Practically, it is difficult to get the requisite mechanical and surface chemical properties in the same material. Certain compositional ranges of soda-calcia-phospho-silicate glasses (*27*), the glass-ceramics (Bioglass, Bioglass-Ceramic, Ceravital) (*28*) and dense biologically reactive hydroxylapatite (Durapatite) (*29*) develop a chemical bond with living bone. The mechanical limitations of these compositions have required the development of a number of means of using the controlled surface-active materials as a component in a composite system, such as coatings on dense high-strength alumina (*30*) or surgical alloys (*31*), as an active filler in bone cement (*32*), or with metal fiber reinforcement (*33*).

Figure 5 shows an x-radiograph of a Bioglass coated 316L stainless steel partial hip replacement in a monkey 40 weeks after implantation. Mechanical testing of the implant in tension showed that it resisted a fracture load nearly equivalent to that of the bone of the opposite leg. Thus, a viable chemical bond had been established at the interface between the bone and the implant via the reactive glass coating.

A scanning electron micrograph of the bonded interface (Fig. 6) shows continuity between the metal, the reactive glass coating, and the bone. Energy dispersive x-ray analysis at various points across the interface shows the characteristic secondary x-ray peaks for calcium and phosphorus in the bone and silicon and calcium in the reactive glass, and intermediate concentrations in the 100-μm-thick bonding zone. The mechanism of formation of the bond has been shown to be the development of a biologically reactive hydroxylapatite and silica-rich layers on the glass surface (*34*). This active surface incorporates metabolic constituents such as collagen within itself as polymerization and crystallization of the inorganic phases proceeds. A scanning electron micrograph of collagen bonded within such a surface after just 2 weeks in vitro (Fig. 7), shows that mechanically and chemically graded interfaces between physiologically derived substances and prosthetic devices can be achieved (*35*).

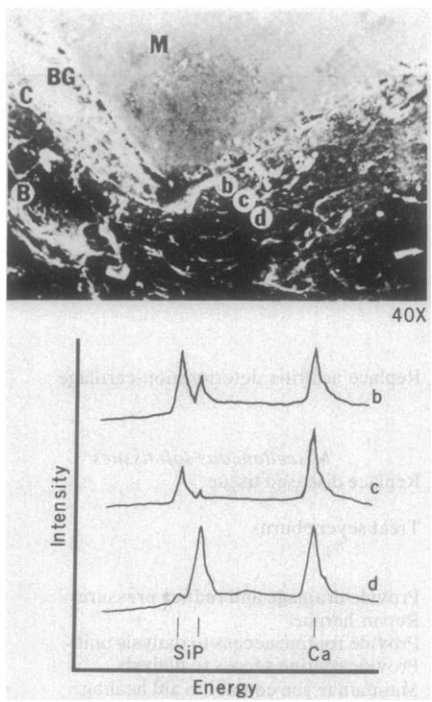

Fig. 6. (Top) Scanning electron micrograph of an interface (*C*) of monkey femoral bone (*B*) bonded to Bioglass (*BG*) coated to a stainless steel metal (*M*) prostheses. (Bottom) Compositional analysis to show tissue at spot (*d*), bonding interface (*c*), and Bioglass coating (*b*) (×20).

Importance of Reliability

Regardless of whether a biomaterial responds to the body as: (i) nearly inert, (ii) porous microsurface, (iii) resorbable, or (iv) surface-reactive material, the central issue today is the reliability of the biomaterial and the devices made from it. The use of even more prostheses is likely in the future, for several reasons. The high (85 to 98 percent) success rates reported for many short-term (< 5 years) implants encourages patients to seek physicians who will use prostheses and, at the same time, encourages surgeons to use devices in patients with a wide range of symptoms and in younger patients. Since more surgeons are gaining confidence in the use of implants, and since prosthetic devices are being used in new clinical applications, longer and more severe service is imposed on the implants, which they do not always withstand. For this and other reasons (*36*) the number of reoperative cases involving implants is steadily increasing. Most reparative operations are more difficult technically than the initial operation, less amenable to the use of generalized procedures and devices, and are complicated by the extensive tissue damage resulting from the implant failure. These factors, the increased age of the patients, and the negative psychological consequences of a previous implant failure all reduce the probability of subsequent success. Reoperative cases generally require the services of the more specialized and capable implant surgery teams, and can consume a progressively larger fraction of their time and facilities. Obviously, the reliability of prostheses must be improved.

I recommend that reliability oriented biomaterials research be directed toward three areas.

1) Composite biomaterials systems.

Many of the devices listed in Table 1 are constructed of more than one material or include modifications of the surface of a material. However, the unique combinations of biological and physical properties that can be achieved in this way have only begun to be exploited.

2) Mechanisms of interfacial reactions. The interface between tissue and implant surface and that between phases in a composite, such as a coating and substrate, are potential weak links in long-term reliability. Only by a thorough study of the mechanisms and kinetics of interfacial reactions will it be possible to determine why failure occurs. Knowledge of failure mechanisms is essential in designing better devices. Likewise, predicting the reliability of a biomaterial or device in service requires understanding the modes of failure of the tissue-implant system.

3) Performance prediction for long-term service (*36*). To develop endurance tables for prostheses will require (i) an understanding of the mechanisms and kinetics of interfacial reactions; (ii) use of mathematical techniques, such as finite element stress analysis, and biomechanics to describe the anticipated stress-time cycles to be applied to prostheses and to define reasonable safety margins of stress; (iii) expansion and application of appropriate theories of the fracture mechanics of brittle materials to predict the expected lifetimes of implants loaded at given stress levels (*37*); (iv) collaboration with the fracture mechanics R & D community to extend life prediction theories to include viscoelastic materials such as bone and polymeric materials, plastically deforming metallic components and combinations thereof, with variable degrees of interfacial attachment between the materials and between materials and tissue; (v) developing accelerated fatigue tests of simple sample configurations and devices that are representative of in vivo conditions; and (vi) establishing methods to correlate predictive relationships with data on implants that have been removed from patients.

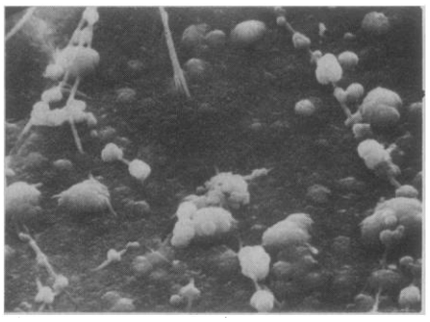

Fig. 7. Scanning electron micrograph of collagen fibers attached to a Bioglass surface after exposure in vitro at 37°C for 10 days (×5000). [Photo courtesy of C. G. Pantano]

An effort has been made to predict the long-term reliability of a Bioglass coated-alumina system by using some of these approaches (*38*). A potential problem was identified in the use of this composite for load-bearing orthopedic implants, and research has been redirected to try to solve it.

Conclusions

The impact of advanced materials technology on the biomaterials field is threefold. New types of composite materials can be created with previously unobtainable combinations of biological and physical properties. Some are already in clinical trials. New techniques for characterizing biomaterials and their interfaces are now available. Methods are becoming available for predicting service lives of materials and prostheses. Use of these new capabilities in concert should produce better prostheses in the decade ahead.

References and Notes

1. L. L. Hench and E. C. Ethridge, *Biomaterials: An Interfacial Approach* (Academic Press, New York, 1980).
2. J. B. Park, *Biomaterials: An Introduction* (Plenum, New York, 1979).
3. I. Elsebai, *Cancer J. Clin.* **27**, 100 (1977).
4. C. B. Ripstein, D. M. Spain, I. Bluth, *J. Thorac. Cardiovasc. Surg.* **45**, 362 (1968).
5. J. Smahel, *Plast. Reconstr. Surg.* **61** (No. 1), 80 (1978).
6. P. Ducheyne, A. Kagan, J. A. Lacey, *J. Bone J. Surg.* **60-A** (No. 3), 384 (1978).
7. J. Calnan, *Br. J. Plast. Surg.* **16**, 1 (1963).
8. S. D. Bruck, *Ann. N.Y. Acad. Sci.* **283**, 332 (1977).
9. L. L. Hench, H. A. Paschall, W. C. Allen, G. Piotrowski, *Natl. Bur. Stand. U.S. Spec. Publ.* **415**, 19 (1975).
10. S. F. Hulbert, S. J. Morrison, J. J. Klawitter, *J. Biomed. Mater. Res.* **6**, 347 (1972).
11. P. Predecki, B. A. Auslaender, J. E. Stephan, V. L. Mooney, C. Stanitski, *ibid.*, p. 401.
12. J. Galante, in *Mechanical Failure of Total Joint Replacement* (Document 916-78, Steering Committee, American Academy of Orthopedic Surgeons, Chicago, 1978), p. 107.
13. S. F. Hulbert and J. J. Klawitter, *J. Mater. Res. Bull.* **7**, 1239 (1972).
14. J. J. Klawitter and A. M. Weinstein, *Acta Orthop. Belg.* **40**, 755 (1974).
15. P. Ducheyne, P. DeMeester, E. Aernoudt, M. Martens, J. C. Mulier, *J. Biomed. Mater. Res.* **11**, 811 (1977).
16. R. M. Pilliar, H. U. Cameron, A. G. Binnington, J. Szinek, I. Macnab, *ibid.* **13**, 799 (1979).
17. J. J. Klawitter and S. F. Hulbert, *J. Biomed. Mater. Res. Symp.* **2**, 231 (1971).
18. R. A. Brand, in *Mechanical Failure of Total Joint Replacement* (Document 916-78, Steering Committee, American Academy of Orthopedic Surgeons, Chicago, 1978), p. 81.
19. C. D. Campbell, D. H. Brooks, M. W. Webster, H. T. Bahmson, *Surgery* **79**, 485 (1976).
20. S. N. Bhaskar, J. M. Brady, L. Getter, M. F. Cromer, T. Driskell, M. O'Hara, *Oral Surg.* **32**, 336 (1971).
21. T. M. Jocknicz, H. A. Nash, D. L. Wise, J. B. Gregory, *Contraception* **8** (No. 3), 227 (1973).
22. K. de Groot, *Biomaterials* **1**, 47 (1980).
23. A. M. Reed, thesis, University of Liverpool (1978).
24. J. T. Frakes, S. D. Brown, G. H. Kenner, *Am. Ceram. Soc. Bull.* **53**, 183 (1974).
25. L. L. Hench, R. J. Splinter, W. C. Allen, T. K. Greenlee, Jr., *J. Biomed. Mater. Res. Symp.* **2**, 117 (1972).
26. L. L. Hench and H. A. Paschall, *ibid.* **4**, 25 (1973).
27. L. L. Hench, R. W. Petty, G. Piotrowski, *An Investigation of Bonding Mechanisms at the Interface of a Prosthetic Material* (Summary Report to U.S. Army Medical R & D Command, Contract DAMD17-76-C-6033, 1979).
28. B.-A. Blencke, H. Bromer, E. Pfeil, H. H. Kas, *Langenbecks Arch. Klin. Chir.* **116**, 119 (1973).
29. M. Jarcho, J. F. Kay, K. I. Gumaer, R. H. Doremus, H. P. Drobeck, *J. Bioeng.* **1**, 79 (1970).
30. D. C. Greenspan and L. L. Hench, *J. Biomed. Mater. Res.* **10**, 503 (1976).
31. D. E. Clark, M. C. Madden, L. L. Hench, in *An Investigation of Bonding Mechanisms at the Interface of a Prosthetic Material* (Annual Report No. 8 to U.S. Army Medical R & D Command, Contract DAMD17-76-C-6033, 1977). pp. 67-77.
32. H. Bromer, K. Deutscher, B. Blencke, H. Pfeil, *Sci. Ceram.* **9**, 94 (1978).
33. P. Ducheyne and L. L. Hench, in preparation.
34. L. L. Hench, in *Proceedings of Surfaces and Interfaces of Glass and Ceramics*, V. D. Frechette, W. C. LaCourse, V. Burdick, Eds. (Plenum, New York, 1974), pp. 265-283.
35. C. G. Pantano and L. L. Hench, in preparation.
36. L. L. Hench, *Biomater. Med. Devices Artif. Organs* **7**, 339 (1979).
37. J. E. Ritter, Jr., in *Fracture Mechanics of Ceramics*, R. C. Bradt, D. P. H. Hasselman, F. F. Lange, Eds. (Plenum, New York, 1978), vol. 4.
38. J. E. Ritter, Jr., D. C. Greenspan, R. A. Palmer, L. L. Hench, *J. Biomed. Mater. Res.* **13**, 251 (1979).