Biomaterials

F. J. Gil, M.P.Ginebra, J. A. Planell

Department of Materials Science and Metallurgic Engineering.
Escuela Técnica Superior de Ingenieros Industriales de Barcelona
Universidad Politécnica de Catalunya.

Abstract

It is well known that the Science and Technology of Biomaterials is very recent; so much so that there are still no solid regulations concerning the biocompatibility of biomaterials. This article aims to give an introduction to the concept of the Biomaterial, as well as describing the differing types of biomaterial and their medical and surgical applications. Biomaterials is an interdisciplinary subject which should involve mechanical and material engineers, designers and cellular biologists as wellas doctors and surgeons. Any biomaterial must be biocompatible and so the concept of biocompatibility and the most common techniques for evaluating this concept are analysed here. The types of materials used as biomaterials in medical technology or bioengineering are then discussed and finally a description is given of the equipment or systems containing biomaterial applications. These cover a very wide range of equipment from sutures to vascular and orthopaedic prostheses to artificial organs.

Key words

Biomaterials, design, implants, bioengineering, biocompatibility.

Introduction

Research in the area of Biomaterials began many centuries ago; indeed, it is possible to find traces of prostheses implanted in Egyptian mummies. However, it was not until the Second World War when attempts to resolve the daily problems of treating massive numbers of patients led to the making of a vast field of Medical Technology and, in particular, Biomaterials Science. Defining Biomaterial has been very laborious and problematic. In fact, it was not until March 1986 in a conference held in Chester (United Kingdom) by the various International Biomaterials Societies with the aim of finding common definitions, that the definition of Biomaterial was approved as being: a non-living material used in medical equipment and made to interact with biological systems. The second conference regarding definitions in the field of Biomaterials, also held in Chester, produced a wider and more precise definition: "A material designed to act interfacially with biological systems with the aim of evaluating, treating, increasing or replacing a tissue, organ or bodily function".

The implantation of a biomaterial implies a wound in the living tissues, which in turn react to the wound first producing inflammation, followed by a repairing process and finally a scar. Connective tissue often responds to the implantation with a fibrosis. Biocompatibility can be thought of as being biological acceptability and research into the interaction of biomaterials with the tissues that are in contact with them. However, this term was not clearly defined during the Chester conference because biocompatibility is not an intrinsic property of a material. That is, a biomaterial is not biocompatible in every condition. For example, ultra high density polyethylene is a bioinert material when it is used as mass for an acetabulum in joint replacements, but the corrosion particles produced by the friction react to foreign bodies.

Biomaterials Science and Technology was thus created in those countries where there was some kind of industrial interest at the time in the production of medical and surgical equipment. In the industrialized world

the first scientists involved in this area, apart from doctors and surgeons, were Materials Science and Technology specialists. Their work consisted of investigating biocompatible materials which could carry out a particular biomedical function, and it was therefore of vital importance to study how the biological tissue which was to be substituted carried out this function. A biomaterial must carry out mechanical functions, as in orthopaedic prostheses; electric functions, as in pacemakers and biological-chemical functions as in the case of membranes for dialysis. It was later seen that it was only the living biomaterial-tissue interfaces which governed biocompatibility and often, therefore, the physical function demanded of the implant. Work was thus begun in the physical and chemical area of surfaces and this work has continued to evolve. Nowadays, to find a favourable organism-implant interaction, it is cellular biologists who evaluate whether or not a determinate material with a determinate surface type will foster the growth of differing cells such as osteoblasts, condroblasts and so on. With the current demand for inert biomaterials which are not rejected in the long term or for certain biomaterials to be biodegradable, it is of vital importance that Materials Science experts, designers, experts in Surface Physics and Chemistry, Cellular Biologists and specialist doctors and surgeons work closely together.

Biocompatibility.

Tissue reaction to a wound.

The initial focus on the concept of biocompatibility was one which was often given in the search for something unknown and a biomaterial was thus defined negatively as being something which:

- should not produce a response from the immune system,
- should not be toxic, neither in itself nor in the products of its degradation,
- should not be carcinogenic in the short or long term, neither in itself nor in the products of its degradation,
- should not be medically incompatible and
- should not be hemodynamically incompatible.

However, this negative definition can not give rise to experimental methodology which allows the concept of biocompatibility to be characterised. Given that quantitative evaluations to be able to compare and make decisions are of vital importance in Science, it was necessary to establish criteria which could be measured, and the scientific community has thus been working along these lines. By rationalising the processes that take place when a biomaterial comes into contact with the surrounding living tissues, it has been possible to start to establish test protocols which enable measurement, even if it is only the partial grade of compatibility or in certain cases of toxicity. Standardized tests do therefore exist in the literature, but finding complete standards regarding tests which ensure the biocompatibility of a product is still a challenge for the scientific community working in this field.

Given that biocompatibility requires biological acceptability, this can be examined on various levels of interaction:

- The interaction between the material and the tissues.
- The reaction resulting from the degradation of the material.
- Mechanical factors (elasticity, tenacity, etc.) and physical factors.

Furthermore, this interaction must be considered both from the point of view of the implant towards the tissue and vice versa. Any biological risk which may result from the use of biomaterials depends on a series of factors including the use, frequency, duration of the exposure, quantity or identity of substances migrated to the human body, as well as the biological activity of these substances.

Implantable materials.

Metallic materials.

Although in the past a number of metallic materials have been used for implants in the organism, today the range of metals and alloys commonly used is fairly limited. Most metals used alone or as alloys in the manufacture of implants, such as Fe, Cr, Co, Ni, Ti, Ta, Mo and W, can be tolerated in small quantities by the human organism. Some are even essential for human life, such as iron in red blood cells or cobalt in the synthesis of vitamin B, although they are not tolerated in large quantities. Metallic materials corrode in a

hostile environment like the human organism and the material consequently deteriorates, weakening the implant. At the same time, the products of the corrosion, which are released into the surrounding tissues, produce undesirable effects. Metals and alloys used as biomaterials all have good resistance to corrosion. Stainless steel, cobalt-chrome alloys, titanium and alloys are all used, and to a lesser extent, tantalium and noble metals such as platinum and gold.

Stainless steels.

The first stainless steels used for implants were the austentitic types 18% Cr-8% Ni. Mo, which improves resistance to corrosion in saline water, was later added. Finally, nowadays, AISI 316 and 316L type steels with carbon contents lower than 0.08% and 0.03% respectively, are also used.

Martensitic stainless steels are used in the manufacture of surgical material. Ferritic stainless steels, despite having excellent resistance to corrosion under stress, are clearly inferior to the austentitics in their mechanical properties and toughening capacity and are thus not found in any application. Even austentitic 316L type stainless steels corrode in the long term within the human body. These materials are hence only used and recommended in temporary implants such as plates, screws and nails like the ones used for osteosynthesis in traumatology.

Austentitic stainless steels toughen through deformation very quickly and consequently in many cases can be worked cold without intermediary treatment by annealing. Needles can be obtained this way with a traction resistance of around 1400 MPa. Austentitic stainless steel implants are made through cold work and are rarely subject to welding. Their surfaces are polished and made passive with nitric acid before being sterilized and packaged.

Co-Cr based alloys.

These materials are primarily used in odontology. There are basically four types: the CoCrMo castable alloy and the forgeable CoCrWNi, CoNiCrMo and CoNiCrMoWFe ones, although only the cast alloy and the forged CoNiCrMo alloy are commonly used today.

The two main elements of these alloys, Co and Cr, form a solid solution of 65% Co-35%Cr. Mo is added to achieve a finer grain size. The alloy with the greatest possibilities is probably CoNiCrMo, which contains 35% of both Co and Ni. This alloy has an excellent resistance to corrosion under stress in salt water. It is very difficult to cold work and is therefore only forged with heat. This alloy is also used in HIP (hot isostatic pressing) synthesis. With the castable CoCrMo alloy, implants are usually moulded using the lost wax method. Controlling the mould temperature enables control of the grain size as well as control of the average free distance between carbides. These alloys behave poorly with friction, as much in contact with itself as with other materials. For this reason, Co-Cr prosthesis joint heads are generally made of ceramic, but never of the same material. Forged alloys have both better ductility and greater traction resistance, although it is unclear whether they have advantages regarding fatigue behaviour. It is important to note that the elastic modulus of these alloys is between 220 and 235 Gpa, that is, 10 to 15% higher than stainless steels and double that of Ti and its alloys. The ways of transferring load from the prosthesis to the bone are therefore different in each case.

Ti and its alloys.

The use of Ti and its alloys in the use of implants was first instigated in Europe, more specifically in the United Kingdom. Later, when the metals which make up Co-Cr alloys became highly strategic, a move was also made to use this metal in the United States. Its low density of 4.7 g/cm^3 compared to 7.9 in stainless steel, 8.3 in the CoCrMo alloy and 9.2 in CoNiCrMo, together with its good mechanical properties and excellent resistance to corrosion, give this metal excellent potential as an implantable material. ASTM and ISO standards give four degrees of unalloyed Ti for surgical implants. The differences between them lie in the contents of gases in solution, N, H and O, and the interstitial C, as well as the residual Fe in solid solution. The most used Ti alloy is Ti-6Al-4V, which is perfectly defined in the norms and widely used in all kinds of structural applications in the aerospace industry. Because Ti undergoes an allotropic change at 882° C, the effect of the differing alloy elements is that of stabilizing one of the two phases. This means that Ti alloys can be α type, ß type, or α + ß type, depending on the alloy elements that it presents as well as their content. The differing degrees of pure Ti present α structure at room temperature, whilst the Ti-6Al-4V alloy is an α + ß type alloy, although its microstructure can vary slightly according to the conformation process and the thermic treatment used.

Titanium reacts highly to oxygen, nitrogen and hydrogen, which all have a high capacity to diffuse interstitially in it. For this reason it must be moulded in an inert atmosphere or vacuum. Ti and its alloys are very difficult to cold work and mechanise as they weaken the tool. Forging is usually carried out under heat, at a temperature of around 950° C.

The Young's modulus of Ti and its alloys is about 110 GPa; around half that of stainless steels and the Co-Cr alloys, which makes this metal more elastically compatible with the bone. The mechanical properties are wide ranging: from elastic limit values and resistance to traction considerably lower than those of stainless steel and Co-Cr alloys for degree 1 Ti, to values of the same order or higher than the materials mentioned for the Ti-6Al-4V alloy, which can reach a tensile resistance of around 1000 MPa. With regard to its weight, Ti alloys are greater than all the others. Furthermore, its excellent resistance to corrosion is due to the formation of a tough layer of oxide which makes the metal passive. It is undoubtedly one of the most inert metals for the manufacture of implants.

Other metals.
Other metals have also been used in the manufacture of implants. Tantalium (Ta) has been proved to be highly biocompatible, but owing to its poor mechanical properties and high density (16.6) g/cm^3, its use is narrowly limited to sutures in plastic surgery and brain surgery.

Platinum and other noble metals in the same group are extremely resistant to corrosion, but also have very poor mechanical properties. They are therefore only used alone or as an alloy to produce electrodes such as the points of a pacemaker, owing to their electrical properties. Gold and silver are similarly resistant to corrosion and likewise have very poor mechanical properties, being hence of little interest as implantable materials.

Ceramic materials

The use of ceramic materials is well known in odontology, but their use in other medical implants is relatively new. The main benefit of ceramics over other materials is that they have very low chemical reactivity which makes them inert and therefore biocompatible with the human body. Carbon implants have been found to be particularly convenient as an interface with blood, as in the case of cardiac valves. Likewise, bone cements based on calcium phosphate are used to fill cavities, such as hydroxyapatite, which has the same mineral content as the bone.

Alumina.
The purity of alumina depends on the system used to obtain it. For the manufacture of implants, the ASTM norm requires a purity of 99.5% with a maximum content of SiO_2 combined with alkaline oxides (mainly Na_2O) lower than 0.1%. The α alumina, with a rhombohedric crystalline structure, has been successfully used in the making of implants.

Porosity plays a very important role in the mechanical properties given that it is usually obtained through synthesis. The relationship that exists between porosity and grain size means that when the porosity falls below 2%, the grain size grows considerably. Alumina is therefore very tough, reaching levels of between 20 and 30 GPa. This high toughness, combined with low friction and low wear properties make this an ideal material for joint replacements, in spite of its fragility and the difficulties of manufacturing it. It has been thought that these materials do not suffer dynamic fatigue because of the lack of plastic deformation, even locally at the point of a crack. Nevertheless, it is well known that if the material is subject to a static load in a determinate medium, there is a possibility that it will break in a given time. The statistic that describes such behaviour is Weibull's statistic, and the phenomenon is qualified as static fatigue. This static fatigue is governed by the medium in which the ceramic is found. Designs for alumina bearings for joints have been made taking this phenomenon into consideration, as well as the load cycles that it can resist before breaking. However, it has recently become clear that under dynamic fatigue, with fluctuating loads and in the same medium, the fracture time is slightly lower than that obtained in static fatigue. Although the existing results are few and fragmented, this may still bring some designs on the market into question.

Hydroxyapatite.
Hydroxyapatite is widely used as artificial bone because it in fact makes up the mineral part of natural bone and can be obtained from it removing the organic constituents such as collagen and mucopolysaccharides. Industrial processes which enable the synthesis of hydroxyapatite in various forms now exist.

As the mineral part of bone and the teeth, hydroxyapatite contains phosphorus and calcium, and its

formula is $Ca_{10}(PO_4)_6(OH)_2$. The apatite minerals crystallize in the shape of a hexagonal rhombic prism. The ideal Ca/P relation in hydroxyapatite is 10/6 and the calculated density is 3.2 g/cm^3. It is worth pointing out that substituting the OH for F gives greater structural stability owing to the superior coordination of F to the nearest Ca. This explains why fluoride provides greater resistance to dental decay.

There are a number of somewhat complicated methods to produce hydroxyapatite precipitates from an aqueous solution of Ca $(No_3)_2$ and NaH_2PO_4. The precipitates obtained are filtered and dried to form a fine white powder. Following calcination, the powder is pressed and synthesized at different temperatures (always above 100° C) and for different times according to the desired results.

Literature concerning the mechanical properties of hydroxyapatite is not unanimous. The elastic modulus thus varies between 40 and 150 GPa, but in any case these values are higher than those obtained in mineralized tissue tests such as dentine or cortical bone, where the Young's modulus varies between 10 and 175 GPa. The values mechanical resistance to compression are greater than those for traction and flex. It appears that resistance to compression is lower than 500 MPa and that the toughness is no greater than 5 GPa. All of these factors indicate that the mechanical behaviour of hydroxyapatite is clearly less valuable than that of alumina and the other ceramics.

The most interesting property of hydroxyapatite is its excellent biocompatibility. This is apparently owing to the fact that it is chemically very similar to the mineral constituent of living bone, which in turn means that it can establish chemical links with the growing tissues surrounding it. It has also been proved to adhere well to Ti and vitroceramic materials.

Vitroceramics.

Vitroceramics are polycrystalline ceramics obtained through controlled crystallization of glass. The technique was developed in the early 60s using photosensitive glass in which small quantities of Cu, Ag and Au were precipitated through ultraviolet radiation. These metallic precipitates help the glass to nucleate and precipitate in the form of a fine grain ceramic with excellent mechanical and thermic properties. Pt, TiO_2, ZrO_2, and P_2O_5 groups are used for the nucleation, whilst suitable thermic treatment provides crystallization higher than 90% with grain sizes between 0.1 and 1 m; much smaller than those which can be obtained with any other conventional ceramic. Two types of vitroceramics have been developed for implants: SiO_2 – CaO-Na_2O - P_2O_5 and LiO_2 – ZnO – SiO_2, although the former is probably the most used.

Vitroceramics possess a range of desirable qualities: their thermic expansion coefficient is very low and may even be negative, their resistance to traction is quite high and in resistance to abrasion they are very similar to sapphire. Furthermore, their capacity to dissolve their surface ions in an aqueous medium seems to be the basis for their superb reaction and link with the hard living tissues which they can be implanted in. It has been shown that living bone can grow in close contact with vitroceramics without fibrous encapsulation. The main drawback is possibly their fragility and there are also restrictions regarding the chemical composition which enables tenacity to be increased, as this improvement is made at the cost of biocompatibility. This means that they can not be used in structural applications such as joint implants, but they are used as fillings in dental composites, bone cements and coating material.

Carbons.

Carbons can be obtained in very different forms: allotropic, crystalline, diamond and graphite, vitreous, almost crystalline and pyrolytic. Among these it is only pyrolytic carbon that is widely used in the manufacture of implants. The crystalline structure of carbon is very similar to that of graphite – planar hexagonal structures formed by strong covalent links in which one valency electron per atom is free to move, resulting in electric conductivity which is very high, but anisotropic. The small crystals have very anisotropic properties owing to the strong planar links and the weak interplanar ones. If there are a lot of small crystals randomly dispersed, the properties of the material will be isotropic.

Carbons can deposit themselves on implant surfaces with a hydrocarbon gas in a fluid bed and at a controlled temperature. The anisotrophy, density, crystalline size and carbon structure deposited can all be controlled by means of the temperature, composition of the fluid gas, geometry of the bed and the permanency of the gas molecules in the bed. All of these factors are important as they control the deposit structure and therefore its mechanical behaviour. Recently, pyrolytic carbon has successfully been deposited on the surfaces of vascular implants. Such a carbon is thin enough not to impede the flexibility of the grafts,

whilst at the same time exhibiting excellent compatibility with the blood.

Vitreous carbons are produced through the controlled pyrolysis of polymers like phenolformaldehyde, and polyacrylonitrile at high temperatures and in a controlled environment. This process is particularly useful in making carbon fibres and tissues which at the same time can be used to make composites. The mechanical properties of carbons, especially the pyrolytes, depends strongly on the density, in such a way that the properties increase with density. Graphite and vitreous carbon have much lower mechanical properties than those of pyrolytic carbon, although they all have more or less the same elastic modulus. The production of carbon composites with carbon fibres in has also been considered. It is well worth mentioning the fact that the excellent compatibility of carbon has made it ideal, particularly in certain vascular applications and heart valves.

Polymeric materials.

The possibilities offered by polymers to be implanted in the human body are very large owing to the fact that they can be easily manufactured in very diverse forms such as fibres, tissues, films and blocks of various shapes and sizes. Indeed, the constituents of natural tissues are actually polymeric structures, which means that synthetic polymers are largely very similar. Among other things, they are used for acrylic bone cements, vein and artery substitutes, organ holding fibres and suture threads. The composite materials are combinations of the other three types of materials which have already been discussed: metallics, ceramics and/or polymers. These materials can combine the most suitable properties of all the differing materials. Of the carbon fibres family, the most notable are ceramic-polymer composites for fillings to be used in odontology and glass fibre composites for plates in osteosynthesis.

Polyolephines.
Thermoplastic lineal polymers such as polythene and polypropylene fall into this category. Polythene is on the market in three basic grades: high and low density and ultra-high molecular weight. Its structure consists of a repeated unit of the type: $-(CH_2 - CH_2)_n-$. Its crystalline index can reach 80%. Ultra-high molecular weight polythene is very commonly used in orthopaedic implants, particularly in surfaces which are subject to high stress, for example; the cotilo of hip replacements or the tibial plate in knee replacements. There is no known dissolvent at room temperature for this material. Extrusion systems or conventional moulds cannot be used to produce it. The basic unit of polypropylene has a structure in which, depending on the position of the methyl group, there are three types of polypropylene: atactic, isotactic and sindiotactic. Isotactic materials can have up to 95% of crystallinity. Hence, the properties of polypropylene will depend on the percentage of isotactic material present, the crystallinity and the molecular weight, which on increasing will increase the density, the melt temperature and the resistance to chemical attack. Owing to the excellent flex life of polypropylene, it is used as a moulded hinge for finger joint prostheses. It also has excellent cracking behaviour under stress in aggressive environments.

Polyamids.
Polyamids are also known as nylons. They have a very good capacity to be transformed into fibres due to the hydrogen link between chains, as well as a high degree of crystallinity which increases their resistance in the direction of the fibre. Nevertheless, their hydrogen links can be destroyed in *in vivo* conditions, which means that these materials can be used in biodegradable applications such as absorbable sutures. The attraction of chains via hydrogen links is controlled by the presence of –CONH- shape groups. The quantity and distribution of these groups are what govern its properties.

Acrylic polymers.
Structure and properties of the acrylics and hydrogels. These polymers are very widely used in medical applications such as contact lenses (soft), implantable ocular lenses and bone cement for the fixation of joint prostheses. Owing to their good physical properties, acrylic materials are also used in dentures and maxillofacial prostheses as they can be very well coloured and they are easy to manufacture. The basic structure of the acrylics can be represented as: $-(CH_2 -CR_1COOR_2)_n-$. The only difference between polymethylacrylate (PMA) and polymethyl methacrylate (PMMA) is in the R groups, with R_1 and R_2 being respectively H and CH_3 in PMA and CH_3 in PMMA. Their polymerization takes place through free radical reactions. Because of the largeness of the lateral groups, these polymers are amorphous. It is for the same reason that PMMA has greater resistance to traction (60 MPa) and a higher melt

temperature (125° C) than PMA (7 MPa and 33° C respectively). PMMA has excellent transparency (92% of transmission), a high refraction index (1.49), and excellent resistance properties to degradation. In addition, it has particularly good chemical resistance and is very biocompatible in its pure state. Its drawback is its fragility compared to other polymers.

The first hydrogel polymer (water absorption higher than 30% of its weight) was polyHEMA, which was used for soft contact lenses. Its formula is $-(CH_2CCH_3COOCH_2OH)_n-$, where the OH group is hydrophilic and responsible for the hydration of the polymer. However, the hydrogels have a relatively low permeability for oxygen if they are compared to silicon.

Bone cement (PMMA).

Bone cement has a number of clinical applications in fixing joint prostheses to the bone. It is prepared by mixing two components; one powder and the other liquid. In general, for most commercial brands, the powder constituent contains polymethyl methacrylate, barium sulphate as a radiopaque agent and benziol peroxide as an activator. The liquid component contains methyl methacrylate monomer N,N-dimethyl-p-toluidine as the accelerant and hydroquinone as the monomer stabiliser.

The contents of each constituent can vary slightly from one commercial brand to another. The two components of the liquid monomer polymerise upon mixing, setting the cement. However, the mechanical properties of bone cements are undoubtedly inferior to those of commercial acrylic resins. A number of factors, both intrinsic and extrinsic, explain this deficient behaviour. Among the intrinsic factors the most important are the composition of the monomer and the powder, the distribution of shapes and sizes of powder and with it the degree of polymerisation, and the liquid to powder relation. The most outstanding of the extrinsic factors is the medium in which the mixing takes place, particularly the temperature in the operating theatre, the mixing technique used and the medium in which the curing takes place (temperature, pressure, surface contacts and so on).

Fluorocarbonate polymers.

The most well-known fluorocarbonate polymer is polytetrafluoroethylene (PTFE), more commonly called teflon. Among other polymers containing fluoride are polytrifluorochloro-ethylene (PTFCE) and polyvinyl fluoride (PVF). I shall only discuss PTFE here as the others are very rarely used in the making of implants.

The repeated unit in this polymer is similar to that of polythene; the only difference being that the hydrogen atoms have been substituted for fluoride atoms: $-(CF_2-CF_2)n-$. This is a highly crystalline polymer (over 94% of crystallinity), with considerable molecular weight. Its density is very high (2.2), resistance to traction low (17-28 MPa) and its elastic modulus is also low (0.5 GPa). It is ideal for contact surfaces due to its very low friction coefficient (0.1). PTFE can not be moulded by injection, nor can it be extruded because of its high viscosity and the fact that it can not plasticize. For this reason, to produce implants its powder must be synthesised above 327° C and under pressure.

Rubbers

Three types of rubbers have been used in the manufacture of implants: silicon, natural rubber and synthetic rubbers. It is generally understood that a rubber can be repeatedly stretched at room temperature to at least double its original length and can forcefully recover the original length when the applied stress is removed. The stretching capacity of rubber is due to the curled cis structure of the polyisoprene, and the fact that this can be done repeatedly is because of the reticulated links between chains. The number of reticulating links governs the flexibility of the rubber.

Natural and synthetic rubbers.

Natural rubber is obtained from latex from the Hevea brasiliensis tree. In its pure form it is compatible with blood. When it is reticulated through x-rays of organic peroxides the compatibility with blood is higher than when it is reticulated through the conventional vulcanization with sulphur. Synthetic rubbers were developed to substitute natural rubber and there are three fundamental types: SBR (copolymer of butadiene and styrene), butyl (copolymer of isobutylene and isoprene) and neoprene (polychloroprene).

Silicon rubbers.

Silicon rubber is one of the few polymers which has been developed for medical use. It is obtained with low molecular weight and low viscosity and it can be reticulated to produce a material with rubber-like characteristics. Two different types of vulcanization (reticulation) can be used: heated or at room temperature. In each case different kinds of reticulating agents are used.

These rubbers can use silicon powder (SiO_2) as a load to improve the mechanical properties.

High resistance thermoplastics.
New polymeric materials have recently been developed with the aim of being able to compete with the light metals. These polymers have excellent mechanical, thermic and chemical properties because the vertebral chain of their principal links is stiffer. The materials which come under this category are: polyacetales, which have been evaluated for use in the acetabulum of the hip joint; polysulphones, which have been tested as the porous coating in orthopaedic implants, and polycarbonates, which have been used in food packaging applications and in cardiopulmonary assistance systems.

Polymer deterioration.
Polymers deteriorate because of chemical, thermic and physical factors. These factors can act synergetically, speeding up the deterioration process, which can affect the vertebral column of the chain, the lateral groups, the reticulating links and the original molecular structure. I do not wish to analyse specific processes here, but it is nevertheless worth mentioning deterioration caused by sterilisation. All sterilisation processes contribute to the deterioration of the polymer in question. Dry heat sterilisation takes place between 160 and 190° C, which is well above the melt and fusion temperatures of most thermoplastics. Both autoclave and steam sterilisation are carried out at temperatures of between 120 and 135° C. Here it is not only the temperature that is important, but also the fact that the water value can attack links in certain polymers. It is possible to use chemical agents at low temperatures, for example: ethylene oxides, gas propylene oxides and phenolic and hypochlorite solutions. However, these methods are very time-consuming, more costly, the chemical agents can at times be highly toxic and difficult to eliminate, and they can also cause the polymer to deteriorate. Lastly, radiation sterilisation above a certain dose deteriorates all kinds of polymers, breaking the chains and recombining links.

Implant types.

Substitution of soft tissues.

Development of the synthetic polymers has been the main contributing factor to the success with which soft tissues have been substituted. This is due to the capacity of polymers to be made with fairly similar chemical and physical properties to natural soft tissues, and they can be manufactured in very different physical forms: fibres, tissues, films, solids and even liquids.

Sutures, surgical tape and adhesives.
The most commonly used and economically significant implants are sutures. In recent years surgical tapes and adhesives have been added to these as complementary or alternative systems.

There are two types of suture, depending on their *in vivo* physical integrity: absorbable and non-absorbable. They can also be distinguished according to the raw material they come from: natural sutures (catgut, silk and cotton) and synthetic sutures (nylon, polythene, polypropylene, stainless steel and tantalium). Another way of classifying them is by their physical shape: monofilament and multifilament.

The absorbable suture known as catgut consists of collagen and is made from the intestinal submucous of sheep, which can live for up to 40 days when treated correctly. However, the knots which are made to fix the suture point reduce its mechanical resistance.

Along with other absorbable sutures (nylon, lactic and polyglycolic acids), catgut produces tissue reactions, although their effect decreases upon being absorbed. This is also the case with other non-absorbable natural sutures such as silk or cotton, which cause a greater reaction than synthetic sutures such as polyester, nylon and polyacrylonitrile. In the event of an infection, the suture's chemical structure and geometric configuration do not appear to have any influence. Polypropylene, nylon and polyglycolic acid cause fewer infections than other sutures such as stainless steel, catgut and polyester.

It is thought that surgical tape can avoid problems that may be caused by sutures: abscesses, weakness of the scar tissues, necrosis caused by pressure, and so on. Nevertheless, it does cause problems, and these are similar to those produced by sticking plasters: unaligned wound edges, poor adhesion because the wound is damp or dirty and separation of the tape in draining the wound or because of a haematoma. These problems have been the reason why surgical tape has not been as successful as it was first hoped to be, although a use has been found for it in skin grafts and in connecting herve tissues. It has not been easy to develop an ideal adhesive because of the kind of environment of tissues and

their capacity to regenerate. Existing experience has led us to believe that an adhesive must: be capable of being wet and sticking to the tissues, be capable of polymerising quickly without producing an excessive amount of heat or freeing toxic products, be reabsorbable as the wounds scar without interfering in the normal scarring process, be easy to prepare in the operating theatre, be sterilizable, have a sufficiently long life to be stored and be possible to manufacture on a large scale. There are currently several types of adhesives on the market but the most well-known ones are the derive from cyanoacrilate. At the end of ten days the resistance of tissue which has been treated with adhesive is half that of sutured tissue. Adhesives are therefore only used in fragile tissues, such as the spleen, liver, kidney and lung. Their use in plastic surgery and in hard tissues like fractured teeth, has been of little success.

Percutaneous and skin implants.

The need for percutaneous implants has increased with the advent of artificial hearts and kidneys, and because of the prolonged injection of medication and nutrients. Artificial skin is of vital importance in maintaining the body's temperature when it has undergone large burns. The problem of obtaining a viable interface between the tissue (skin) and the implant, is mainly due to the following factors:

- even if the tissue is initially fixed well to the interstices of the implant surface, it can not be maintained for long periods of time because the dermal tissue cells alter it and the epithelium tends to grow around the implant thereby isolating it;
- any opening which is big enough for bacteria to penetrate will end up causing an infection.

There are a plethora of factors to be taken into consideration in the development of a percutaneous system:

- the function of the implant, which may be from energy transmission to electric stimulation to the transmission of matter as in the blood cannulas;
- engineering factors such as the material selection, the geometric design of the implant and the mechanical stress which it will be subject to;
- biological variables – an implant for a human being is not the same as one for a laboratory rat, and an abdominal implant differs from a back position implant;
- and lastly, human factors, which refer to post-operation treatment, aesthetic factors and the implantation technique.

In the case of artificial skin the problems are very similar. Here what is needed is a material that can adhere to a large burned surface in order to avoid the loss of fluids, electrolytes and other biomolecules until the wound forms a scab. The aim is to find a permanent skin implant, but this has not been possible to date and it seems that we are still a long way off something definitive. Autografts and homografts with permanent solution are currently being used. Composite reticulated membranes of collagen-mucopolysaccharide, reconstructed collagen and vinyl chloride and acetate methyl-2-cyanoacrylate copolymers have also been used until now, although with little success in the case of the last ones.

Maxillofacial implants and implants increasing other soft tissues.

In this section I shall discuss cosmetic implants and reconstructions. Soft tissues can be divided into the following categories: 1) those which fill a space, 2) those which make up a mechanical support, and 3) those which store or transport a liquid, although most soft tissues carry out more than one of these functions.

There are two types of maxillofacial implants: extraoral and intraoral. A large number of polymeric materials are available for the former. The general requirements are: 1) that the colour and texture can be matched to the patient, 2) that it is mechanically and chemically stable, and 3) that it can be easily manufactured. The most commonly used materials are vinyl and acetate polychloride copolymers, PMMA, silicone and polyurethane rubbers. The requisites for intraoral implants are the same as for other implants given that they are in fact implanted. Metallic materials such as tantalium or the Co-Cr alloy are used to correct maxillary, jaw or facial bone defects. To increase soft tissues such as gum and chin, silicon rubber or PMMA is used.

There are also implants to totally or partially substitute the bones in the ear. The materials used here include PMMA, PTFE, polythene, silicon rubber, stainless steel and tantalium. However, recent experiments have shown that the most suitable materials for otic implants seem to be PTFE and carbon based composites, porous polythene and pyrolytic carbon.

Ocular implants are used to restore the functionality of both the cornea and the pupil when they are damaged or ill. These transplants are generally transparent acrylics, especially PMMA. In recent times spectacular

growth has been seen in intraocular lenses. They are currently only being used after cataract operations but their use will most likely be extended in the very near future to patients needing thick glasses.

Implants for fluid circulation have also been necessary, in cases such as hydrocephalus and incontinence. Hydrocephalus, which is caused by abnormally high pressure in the cerebrospinal fluid in the brain, can be treated by draining off this fluid through a special shunt. The use of implants for the urinary system on the other hand has not been as successful, due to the difficulties in fitting the prosthesis to the living system. In addition, infection is a constant threat because the solid deposits in urine can block the orifice. A number of materials have been tested without significant long term success, including glass, rubber, silver, tantalium, Co-Cr, polythene, PTFE, polyvinylalcohol and so on.

Within the group of implants which must fill a space, the most representative is probably breast implants. Initially, breast size was increased using materials like paraffin, wax or silicon fluids which were either injected directly or implanted in the form of a silicon globe. By the end of the 1960s, injectable implants were banned in the United States due to problems relating to progressive instability and final loss of shape and texture, as well as infection and pain. Although increasing or substituting breasts for cosmetic reasons is not recommended, it does appear to be psychologically beneficial for patients who have undergone dramatic mastectomies or who have asymmetric deformations. The prosthesis is generally a bag of silicon rubber which is filled with silicon gel and covered in a polyester mesh to enable the living tissue to grow and attach itself to it. Artificial vagina, testicle and penis prostheses also fall into this category.

Implants in contact with the blood.

Implants in contact with the blood can be divided into two large categories: extracorporeal implants for short term use, such as artificial organ membranes (artificial lungs or kidneys), tubes and catheters to carry blood, and *in situ* implants for long term use, as in implantable artificial organs and vascular implants. Although pacemakers are not in direct contact with the blood, they are also included in this category because they help the blood to circulate around the body.

The most important requisite for implants in contact with the blood is that they be compatible with it. Blood clotting is the most important factor in blood compatibility, but at the same time the implant must not damage proteins, enzymes and the elements which make up blood (red cells, white cells and platelets). Implants are used to substitute or repair large arteries and veins, including the heart and its valves. Most of the materials used in these applications are polymers owing to their flexibility and ease of production.

Some difficulties have been encountered with vein implants owing to the adjacent vein walls collapsing or because of low pressure and almost static flow. Nevertheless, these difficulties have been overcome in most cases through the use of autografts. Nylon, PTFE and polyester have been used for these applications.

The first designs for vascular implants consisted of solid tubes made of glass, aluminium, gold, silver and PMMA. However, all of these designs produced blood clots. At the start of the 1950s porous implants were introduced which allowed tissue to grow in its interstices. These are the tissues which are in contact with the blood and they therefore minimise clots. Ironically, the best materials found for this application were thrombogenic materials. Another advantage can be seen in the anchorage that the growth of the new tissue produces. The drawback is the initial loss through the pores, although this can be avoided by pre-coagulating the exterior surface of the implant before fitting it. Nylon, PTFE, polyester, polypropylene and polyacrylonitrile have all been used to manufacture this type of implants. However, PTFE, polyester and polypropylene seem to be the most favourable owing to the very low level of deterioration of their physical properties *in vivo*. The most widely used among these is polyester (polyethyleneteraftalate). An arterial graft coated in pyrolytic carbon has recently been developed which may improve the non-obstruction of the graft and lessen the need for post-operation anticoagulants because of its non-thrombogenic properties. With reference to the porosity of the tissue, it is generally considered that in one minute between 5000 and 10,000 ml of water pass through one cm^2 of tissue, at 120 mm Hg.

There are four valves in the heart's ventricles. Usually the valves in the left ventricle (mitral and aortic) deteriorate before the valves in the right ventricle because of the higher ventricular pressure in the left. The aortic valve is the most important and often the most critical, as it is the last door through which the blood must pass before circulating around the body. Since they were first used in the 1960s there have been many different types of valve implants. The first ones

imitated natural valves and used thin sheets, but later designs consisted of a ball or disc inside a cage. The requisites are the same as those for vascular implants, but they must also maintain blood flow and blood pressure with the minimum production of sound. It is important to note that animal valves and collagen have also been used in human implants. In general all artificial valves have a ring made of polymeric material which can be sewn onto the living tissue. This helps the implant to attach itself initially, until the true living tissue growth explained above takes place.

The cage is usually metallic with a pyrolytic carbon coating, the same as the ball or disc so as to produce a non-thrombogenic surface. The ball or disc can be polymeric (polypropylene, polyoxymethylene, polychlorotrifluoroethylene), metallic (titanium, Co-Cr), or pyrolytic carbon deposited on a graphite substrate. The possible use of ceramic materials such as aluminium oxide or circonium oxide is currently being researched.

The aim of cardiac aid systems is to sustain blood circulation when the heart can not function normally, or during cardiac surgery. The blood is pushed by a pump which can be oxygenated by the patient's own lung or by an artificial oxygenator. In all of the different kinds of oxygenators (membrane, bubble and film), the oxygen comes into contact with the blood and the waste gas CO_2 is simultaneously eliminated. The membranes used are usually made from silicon rubber or PTFE, although silicon rubber appears to be better in transferring the O_2 and CO_2.

Although most implants are designed to carry out mechanical functions, among those which carry out electrical functions the most outstanding is possibly the pacemaker, whilst in the chemical functions group it may be said to be implants like the artificial kidney and lung. Artificial hearts and cardiac aid systems use chambers, valves and blood pumping systems. Independent artificial hearts are as yet impossible, owing to the need to supply such a heart with electric al energy.

Cardiac pacemakers are used to regulate the rhythm of the heart's muscle contractions. A pacemaker must basically supply the heart with an exact amount of electric stimulation, for all its differing speeds. In fact, pacemakers consist of conducting electrodes which are connected to a stimulator. The electrodes are well isolated, usually with silicon rubber, except their tips which are sutured or attached directly to the cardiac network. The tip is usually made from a noble metal with high resistance to corrosion and reasonable mechanical resistance, for example the Pt – 10 % Ir alloy. Generally, the most significant problems caused are that the electrodes break because of fatigue and the formation of collaginous scar tissue on the electrode tip which increases the electrical resistance of the contact. The battery and electronic components are isolated with a coating of polymeric resin. Pacemakers are usually replaced every 2 to 5 years because the electric battery runs out. Longer lasting batteries will not be of any use until the problems of electrode tip fatigue and conductivity decrease are first resolved. To this end, porous electrodes are currently being developed which enable anchorage by means of the growth of cardiac muscle tissue.

The kidney's main function is to eliminate metabolic waste products. The waste products it filters are mainly: urea, sodium, chloride, bicarbonate, potassium, glucose, creatinine and uric acid. In an artificial kidney therefore, the key component is the membrane which can filter all of these; a process known as dialysis. Apart from the membrane, dialysis equipment consists of a bath; a pump which allows blood to circulate from the artery to the vein once it is clean. The material used in most membranes is cellophane, a derivative of cellulose. Ideally, the membrane must eliminate all of the waste products that a healthy kidney eliminates, it must be compatible with the blood and it must have enough mechanical resistance when wet to enable ultrafiltering without significant dimensional changes.

Attempts have been made to improve the cellophane membranes by means of reticulations, copolymerisation and reinforcements through fibres of other polymers such as nylon. Likewise, the surface has been coated with heparin to prevent coagulation. Other types of membranes have also been used, such as a polymer of polyethyleneglycol and polyethyleneteraftalate, which can filter selectively owing to the alternated hydrophilic and hydrophobic segments.

Besides attempts to improve the membrane for better dialysis, emphasis is being placed on research into obtaining a much more compact machine (portable kidney) and one which is more convenient (being able to do the dialysis at home or having reusable filters). Another of the drawbacks is the use of a shunt which must be connected to the blood vessels. In chronically ill patients this can be traumatic to their blood vessels and in the long term a percutaneous implant is necessary.

Substitution of hard tissues.

The only implants that come under this category are dental implants and skeletal implants. Before going further it is important to highlight that hard tissues, particularly bone, have a large capacity to remodel themselves, thus enabling the osteosynthesis of, for example, a fractured bone. However, this remodelling can go further and make the bone grow or disappear according to its mechanical needs. How exactly this osteogenetic activity is regulated is as yet unknown, meaning that something so common as a fractured bone is still not a fully resolved issue.

The design principles and production criteria of orthopaedic implants are the same as those used in other engineering applications in which a part is required to fulfil a mechanical function. The remodelling capability of the living tissues eliminates the relevance of fatal phenomena in synthetic materials, such as fatigue and corrosion. For this reason, implants which substitute bone can not be similar to it in either shape or resistance because self-repair help will not occur.

Internal fracture fixation systems.

Fracture fixation systems were in fact the first implants to be developed. The most simple and versatile are probably the various metal wires (such as Kirschner's needles), which are used to fix different bone fragments. Problems regarding corrosion and fatigue worsen in the subjection points because of the concentration of stresses produced there. Needle and nail tips are usually cut specially so that they can penetrate the bone relatively easily.

Screws are another of the most widely used implants in bone fixation, either on their own or with osteosynthesis plates (Figure 1). There are essentially two screw types (Figure 2): self-tapping and non self-tapping – the latter of which do not require a needle to be previously put into the bone with a drill bit. Resistance to being pulled out is almost the same in both types of screw. This resistance is governed by the screw's cross-section, although the geometry of the thread also has a certain influence on it. If the screw is firmly fixed, the tissues around the screw die immediately and are reabsorbed.

The dead tissues are later replaced with living tissues. However, if there is any kind of macro or micro-movement between the screw and the bone, collaginous fibre tissues forms and encapsulates it, thereby progressively loosening the screw.

Osteosynthesis plates are available on the market in many different shapes and sizes. Given the considerable force which limb muscles generate, these plates must be resistant, particularly in the femur and tibia. The plates are fixed with screws, which although they should not be loose, neither should they be tightened excessively as this can produce bone necrosis, and the screws can be deformed and consequently corrode under stress. Discussion as to the convenience of having the bone subject to compression or not has led to the availability of self-compression plates on the market which protect the bone. It appears that once the metallic plate is removed, the bone is mechanically more fragile at that point. Moreover, bone is reabsorbed at the points where there is a resistant element which exercises the mechanical function instead of the bone. Having a plate for a long time can therefore be counterproductive. Finally, it is also important to bear in mind that when the plate is removed, the needles that held the screws stay in place. These needles concentrate stress and may be the cause of a new fracture so certain precautions must be taken, even when the fracture is healed and the plate removed.

Another way of fixing long bone fractures is by tightly inserting what is known as an intramedullar system or nail, into the intramedullar cavity.

This type of implant must exercise some kind of elastic force in the cavity interior so as to avoid rotations and firmly fix the fracture. Compared to fixation with a plate, the intramedullar system is in a better position to resist flex stress because it is in the middle of the bone. Its weakness is that torsion resistance is very low. There is more than one type of intramedullar nail and these are basically differentiated by their profile. The most well-known ones are probably Kuntscher's nails, whose cross-section is in the shape of a clover leaf, and Schneider's, whose nails take the shape of a four point star.

Aside from long bones, fixation systems are very important in the spine. These are used in correcting excessive curvature of the vertebral column and also in carrying out arthrodesis which entails making a group of vertebral segments rigid. These systems can be very complex, and combine plates, nails, screws and a whole range of smaller parts which are fixed to the vertebra.

The best material to carry out all of these osteosynthetic systems is austentitic stainless steel. Nonetheless, that is not to say that Co-Cr plates or Ti alloy nail-plates have not been manufactured, or that PMMA with carbon fibre plates have not been tested.

As I said earlier, the osteogenetic process is not fully understood and it appears that not only mechanical factors but also electrical and magnetic factors play a part in it. In fact, the bone's piezoelectric properties are well-known and it is clear that osteogenetic activity can be improved with electromagnetic stimulus. The stimulation mechanism is not known either, although it is thought to be linked to the naturally high negativity of the fractured region due to the increase in ionic and metabolic activity in the region.

Joint substitution.

Joints pose new problems with regard to bone repair as research must also be done into corrosion and wear phenomena as well as the dynamics of load transfer by the joint. Infections must also be taken into account, and perhaps most importantly, the fact that second time replacements are much more complicated than first time ones owing to the amount of natural tissue that is destroyed. The first joint replacements to be carried out successfully were in the hip (figure 3), and these were later followed by knee replacements (figure 4). These were both implanted in patients who would otherwise have totally lost the ability to walk.

The hip and shoulder joints consist of a ball and socket, whereas the others, such as the knee and elbow, are hinge joints. They all have smooth cartilaginous joint surfaces which are lubricated with sinovial fluid. Joints with large areas minimise the effect of stress concentrations. Instant impacts are absorbed by the cartilage and the spongy subchondral bone underneath, which owing to its viscoelastic properties gradually transfers the impact onto the cortical bone. The joint in itself is made up of the ligaments, tendons and muscles, but analysing the forces which act upon the differing tendons and ligaments is far from straightforward. The kinematics of the joints is also very difficult to study, and it is not easy to say exactly how the position of the instant centre of rotation varies, even in such a well-known joint as the knee. The forces applied during a determined activity have been measured, as is shown in table 1. As can be seen from this, the maximum forces exerted on the joints in such simple activities as walking or going up a flight of stairs, can be very high.

Table 1. Maximum force values exerted on the hip and knee in various activities. Maximum force on the joint (in multiples of the body weight (see page 52).

Historically, hip replacements have undergone enormous change. Initially, the only aim was to correct the healthy tissue. However, it soon became evident that this was not a solution but rather led to new and constant problems. Tests were then carried out which entailed coating the femoral head and/or acetabulum with metal, or using a threaded nail finishing in a ball, and nowadays there are two main types of hip replacement: total and partial. In both cases the neck of the femur is sectioned and the replacement rod is inserted into the interior of the medullar cavity. However, whilst the acetabular cartilage is maintained in the partial replacement, in the total replacement this cartilage is removed. A socket is put in its place which will hold the replacement ball. Partial replacements are usually used in considerably older patients as the acetabular cartilage deteriorates very quickly in patients who are very active. It is worth pointing out that there is currently a huge variety of hip replacement models and designs on the market. This highlights the fact that knowledge of the working of the joints is still very limited, and it also demonstrates the organism's capacity to withstand and survive the possible traumas that these implants can cause.

The most important issue in hip replacements, as with other joint replacements, is their fixing. Studies of the stress distribution in healthy femurs and in those which have hip replacements show that the situation changes dramatically because of the geometry change in the resisting element. Thus, whilst the stress in the distal region in the furthest point of the rod increases significantly in relation to a healthy femur, in the calcar region (contact region of the replacement with the bone underneath the ball position), the stress decreases considerably, causing bone reabsorption and consequent weakening. This led to the geometry of replacements being redesigned, albeit with the addition of new problems. Until recently replacements have been fixed to the bone using acrylic bone cement (PMMA). The cement is inserted into the bone cavity when it is still soft, and the replacement is then fixed on. The cement does not only act as the initial fixation of the replacement to the bone - its viscoelastic properties enable it to better distribute loads from the ball to the femur wall, preventing this load from being at times repeated. Using cement does not stop the calcar region receiving as many solicitations as a healthy femur, and this produces bone reabsorption and ultimate weakening of the implant or fracture of the rod, which means that the

replacement needs to be changed in the long run. It is important to note that in designing an implant, the surgical technique for fitting it is as important as for removing it in the event of an infection, weakening or breakage. Current replacement designs emphasise modular replacements in which partial elements can be substituted: the ball, rod, socket etc. Non-cemented replacements have also become the norm nowadays, particularly in young patients. Moreover, replacements which have a porous coating or those coated with hydroxyapatite seem to be more highly favoured with surgeons due to the fact that they induce bone growth on their surface. Another important point to underline is the friction between the ball and the socket. Tests have shown that friction and therefore wear is less using a ceramic ball and polythene socket than with a metallic ball and polythene socket, and that the latter combination is in turn better than if both surfaces are metallic.

Even though knee joints degenerate more than any other kind of joint, knee replacements have been developed and accepted more slowly than hip replacements owing to the fact that the biomechanics of their movements are more complicated. The knee replacements currently used can be split into two main groups: hinged and non-hinged. Although I generalised earlier that the knee is a hinged joint, the reality is in fact otherwise – there is no fixed centre of rotation for stretching and bending movements, but rather an instant centre of rotation for each position. The replacements which probably best replicate the anatomical movement are biocompartmental (non-hinged). Here the upper part which is fixed to the femur is totally metallic, whilst the lower part which is fixed to the tibia contains the tibial plate made of ultra-high molecular weight polythene. The porous rods of these implants are currently designed to induce bone growth and with it better fixation.

Other, less important joint replacements on the market are ankle, shoulder, elbow, wrist and finger. I shall not go into any detail of these as the philosophy is very similar to that already discussed.

Dental implants.

Although most people are familiar with some kind of dental implant, particularly amalgam implants to cover dental decay, I shall now describe the total substitution of a tooth or socket. Totally substituting teeth has proved to be something of a challenge due to the percutaneous nature in a hostile environment which is constantly changing its chemical composition, pH and temperature. Furthermore, teeth undergo the highest compressive stress in the entire organism (up to 850 N) and as yet a satisfactory material or technique has not been found which sustains not only compressive stress but also torsion and shear stress upon chewing. There are two main kinds of dental implants: subperiosteal/transosteal and endosseous implants. The former give support to dentures and the latter restore the function of the teeth.

The endosseous implant is inserted in the place of the missing tooth in order to restore its original function. There are many differing types of endosseous implants. The underlying principal of all of them is to fix a metallic part to the bone for the long term. When the implant is firmly fixed, between one and four months later, this part will be covered by a suitable crown. These parts can be made of stainless steel, Co-Cr or Ti-6Al-4V, although the vast majority are commercially pure titanium owing to its capacity of osteointegration. Good hydroxyapatite coatings have recently been achieved which enable excellent fixation to the bone in a relatively short period of time.

Subperiosteal and transosteal implants have successfully been used to fix false teeth in patients with no teeth. The materials here are very similar to those of the previous case.

Figure 1. Fracture plates (see page 53).

Figure 2. Osteosynthesis nails (see page 53).

Figure 3. Hip replacements (see page 54).

Figure 4. Knee replacement (see page 54).

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