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Biomedical applications of polymer-composite materials: a review

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Abstract

An overview of various biomedical applications of polymer-composite materials reported in the literature over the last 30 years is presented in this paper. For the benefit of the readers, general information regarding structure and function of tissues, types and purpose of implants/medical devices, and various other materials used, are also briefly presented. Different types of polymer composite that are already in use or are investigated for various biomedical applications are presented. Specific advantages of using polymer-composite biomaterials in selected applications are also highlighted. The paper also examines the critical issues and scientific challenges that require further research and development of polymer composite materials for their increased acceptance in the biomedical industry. © 2001 Elsevier Science Ltd. All rights reserved.

Keywords: Biomaterials; Biocomposites; Polymer composites; Implants; Prosthesis; Medical devices; Biomedical engineering; Bioengineering

1. Introduction

Biomaterials are materials of natural or man-made origin that are used to direct, supplement, or replace the functions of living tissues of the human body [21]. Use of biomaterials dates far back into ancient civilizations. Artificial eyes, ears, teeth, and noses were found on Egyptian mummies [256]. Chinese and Indians used waxes, glues, and tissues in reconstructing missing or defective parts of the body. Over the centuries, advancements in synthetic materials, surgical techniques, and sterilization methods have permitted the use of biomaterials in many ways [178]. Medical practice today utilizes a large number of devices and implants. Biomaterials in the form of implants (sutures, bone plates, joint replacements, ligaments, vascular grafts, heart valves, intraocular lenses, dental implants, etc.) and medical devices (pacemakers, biosensors, artificial hearts, blood tubes, etc.) are widely used to replace and/or restore the function of traumatized or degenerated tissues or organs, to assist in healing, to improve function, to correct abnormalities, and thus improve the quality of life of the patients. According to a report published in 1995 by The Institute of Materials, London, the estimated world market for all medical devices, including diagnostic and therapeutic equipment is in the region of $100 billion per year. Within this industry, the world market for biomaterials is estimated to be around $12 billion per year, with an average global growth of between 7 and 12% per annum. Biomaterials are expected to perform in our body's internal environment, which is very aggressive. For example the pH of body fluids in various tissues varies in the range from 1 to 9. During daily activites bones are subjected to a stress of approximately 4 MPa whereas the tendons and ligaments experience peak stresses in the range 40–80 MPa. The mean load on a hip joint is up to 3 times body weight (3000 N) and peak load during jumping can be as high as 10 times body weight. More importantly, these stresses are repetitive and fluctuating depending on the activities such as standing, sitting, jogging, stretching, and climbing [21]. In a year, the stress cycles of finger joint motion or hip joint motion estimated to be as high as 1×10^6 cycles, and for a typical heart 0.5×10^7–4×10^7 cycles. This information roughly indicates the acute and instantaneous biological environment in which the biomaterials need to

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Nomenclature

BIS-GMA	bis-phenol A glycidyl methacrylate
C	carbon
CF	carbon fibers
GF	glass fibers
HA	hydroxyapatite/hydroxylapatite
HDPE	high density polyethylene
KF	Kevlar fiber
LCP	liquid crystalline polymer
LDPE	low density polyethylene
MMA	methylmethacrylate
PA	polyacetal
PBT	polybutylene terephthalate
PC	polycarbonate
PCL	polycaprolactone
PE	polyethylene
PEA	polyethylacrylate
PEEK	polyetheretherketone
PEG	polyethylene glycol
PELA	block copolymer of lactic acid and polyethylene glycol
PET	polyethylene terepthalate
PGA	poly(glycolic acid)
PHB	polyhydroxybutyrate
PHEMA	poly(HEMA) or poly(2-hydroxyethyl methacrylate)
PLA	poly(lactic acid)
PLDLA	poly(L-DL-lactic acid)
PLLA	poly(L-lactic acid)
PMA	polymethylacrylate
PMMA	polymethylmethacrylate
Polyglactin	copolymer of PLA and PGA
PP	polypropylene
PS	polysulfone
PTFE	polytetrafluroethylene
PU	polyurethane
PVC	polyvinylchloride
SR	silicone rubber
THFM	tetrahydrofurfuryl methacrylate
UHMWPE	ultra high molecular weight polyethylene

survive. Needless to say, the biological environment also depends on the patient's conditions and activities.

In the early days all kinds of natural materials such as wood, glue and rubber, and tissues from living forms, and manufactured materials such as iron, gold, zinc and glass were used as biomaterials based on trial and error. The host responses to these materials were extremely varied. Some materials were tolerated by the body whereas others were not. Under certain conditions (characteristiccs of the host tissues and surgical procedure) some materials were tolerated by the body, whereas the same materials were rejected in another situation. Over the last 30 years considerable progress has been made in understanding the interactions between the tissues and the materials. It has been acknowledged that there are profound differences between non-living (avital) and living (vital) materials. Researchers have coined the words 'biomaterial' and 'biocompatibility' [253] to indicate the biological performance of materials. Materials that are biocompatible are called biomaterials, and the biocompatibility is a descriptive term which indicates the ability of a material to perform with an appropriate host response, in a specific application [22]. In simple terms it implies compatibility or harmony of the biomaterial with the living systems. Wintermantel and Mayer [258] extended this definition and distinguished between surface and structural compatibility of an implant [260]. Surface compatibility meaning the chemical, biological, and physical (including surface morphology) suitability of an implant surface to the host tissues. Structural compatibility is the optimal adaptation to the mechanical behavior of the host tissues. Therefore, structural compatibility refers to the mechanical properties of the implant material, such as elastic modulus (or E, Young's modulus) and strength, implant design (stiffness, which is a product of elastic modulus, E and second moment of area, I), and optimal load transmission (minimum interfacial strain mismatch) at the implant/tissue interface. Optimal interaction between biomaterial and host is reached when both the surface and structural compatibilities are met. Further more it should be noted that the success of a biomaterial in the body also depends on many other factors such as surgical technique (degree of trauma improsed during implantation, sterilization methods, etc), health condition and activities of the patient. Table 1 summarizes various important factors that are considered in selecting a material for a biomedical application.

Clinical experience clearly indicates that not all off-the-shelf materials (commonly used engineering materials) are suitable for biomedical applications. The various materials used in biomedical applications may be grouped into (a) metals, (b) ceramics, (c) polymers, and (d) composites made from various combinations of (a), (b) and (c). Researchers also classfied materials into several types such as bioinert and bioactive, biostable and biodegradable, etc. [90]. As the former classification is known to engineers, it is further followed in this review. Alumina, titania, zirconia, bioglass (or bioactive glasses), carbon, and hydroxyapatite (HA) are widely considered as biocompatible ceramics. Metals and alloys that are successful as biomaterials include: gold, tantalum,

Table 1
Various factors of importance in material selection for biomedical applications

Factors	Description		
1st Level material properties	Chemical/biological characteristics Chemical composition (bulk and surface)	Physical characteristics Density	Mechanical/structural characteristics Elastic modulus Poisson's ratio Yield strength Tensile strength Compressive strength
2nd Level material properties	Adhesion	Surface topology (texture and roughness)	Hardness Shear modulus Shear strength Flexural modulus Flexural strength
Specific functional requirements (based on application)	Biofunctionality (non-thrombogenic, cell adhesion, etc.) Bioinert (non-toxic, non-irritant, non-allergic, non-carcenogenic, etc.) Bioactive Biostability (resistant to corrosion, hydrolysis, oxidation, etc.) Biogradation	Form (solid, porous, coating, film, fiber, mesh, powder) Geometry Coefficeint of thermal expansion Electrical conductivity Color, aessthetics Refractive index Opacity or translucency	Stiffness or rigidity Fracture toughness Fatigue strength Creep resistance Friction and wear resistance Adhesion strength Impact strength Proof stress Abrasion resistance
Processing and fabrication	Reproducibility, quality, sterilizability, packaging, secondary processability		
Characteristics of host: tissue, organ, species, age, sex, race, health condition, activity, systemic response Medical/surgical procedure, period of application/usage Cost			

stainless steel, Co–Cr, NiTi (shape memory alloy), and Ti alloys. A large number of polymers such as polyethylene (PE), polyurethane (PU), polytetrafluoroethylene (PTFE), polyacetal (PA), polymethylmethacrylate (PMMA), polyethylene terepthalate (PET), silicone rubber (SR), polysulfone (PS), polyetheretherketone (PEEK), poly(lactic acid) (PLA), and poly(glycolic acid) (PGA) are also used in various biomedical applications. HA/PE, silica/SR, carbon fiber/ultra high molecular weight polyethylene (CF/UHMWPE), carbon fiber/epoxy (CF/epoxy), and CF/PEEK are few examples of polymer composite biomaterials. Each type of material has its own positve aspects that are particularly suitable for specific application. This paper is intended mainly to provide an overview of various polymer composite biomaterials, and also to stimulate the research in composite biomaterials as this material group has not been explored extensively with regards to the biomedical applications. In this paper, the merits and demerits of polymer composite materials are emphasized by contrasting with the other types of materials. However, it is not the intention of the authors to advocate that polymer composite biomaterials are the only candidates suitable for medical applications.

A large number of polymers are widely used in various applications. This is mainly because they are available in a wide variety of compositions, properties, and forms (solids, fibers, fabrics, films, and gels), and can be fabricated readily into complex shapes and structures. However, they tend to be too flexible and too weak to meet the mechanical demands of certain applications e.g. as implants in orthopedic surgery. Also they may absorb liquids and swell, leach undesirable products (e.g. monomers, fillers, plasticizers, antioxidants), depending on the application and usage. Moreover, the sterilization processes (autoclave, ethylene oxide, and ^{60}Co irradiation) may affect the polymer properties. Metals are known for high strength, ductility, and resistance to wear. Shortcomings of many metals include low biocompatibility, corrosion, too high stiffness compared to tissues, high density, and release of metal ions which may cause allergic tissue reactions [221]. Ceramics are known for their good biocompatibility, corrosion resistance, and high compression resistance. Drawbacks of ceramics include, brittleness, low fracture strength, difficult to fabricate, low mechanical reliability, lack of resilience, and high density. Polymer composite materials provide alternative choice to overcome many shortcomings of homogenous materials mentioned above. The specific advantages of polymer composites are highlighted in the following.

Generally, tissues are grouped into hard and soft tissues. Bone and tooth are examples of hard tissues, and skin, blood vessels, cartilage and ligaments are a few

examples of soft tissues. As the names suggest, in general the hard tissues are stiffer (elastic modulus) and stronger (tensile strength) than the soft tissues (Tables 2 and 3). Considering the structural or mechanical compatibility with tissues, metals or ceramics are chosen for hard tissue applications (Tables 2 and 4), and polymers for the soft tissue applications (Tables 3 and 5). A closer look at Tables 2 and 4 reveals that the elastic moduli of metals and ceramics are at least 10–20 times higher than those of the hard tissues. One of the major problems in orthopedic surgery is the mismatch of stiffness between the bone and metallic or ceramic implants. In the load sharing between the bone and implant, the amount of stress carried by each of them is directly related to their stiffness. Thus, bone is insufficiently loaded compared to the implant, and this phenomenon is called 'stress-shielding' or stress protection. Many investigators [44,168,238], have shown that the degree of stress protection is proportional to the degree of stiffness mismatch. The stress-shielding affects the bone remodeling and healing process leading to increased bone porosity (also known as bone atrophy) [44,103,214,251]. It has been recognised that by matching the stiffness of implant with that of the host tissues limits the stress-shielding effect and produces desired tissue remodeling. In this respect, the use of low-modulus materials such as polymers appears interesting; however, low strength associated with low modulus usually impairs their potential use. Since the fiber reinforced polymers i.e. polymer composite materials exhibit simultanously low elastic modulus and high strength, they are proposed for several orthopedic applications [85,176]. Additional merit of composite materials is that by controlling the volume fractions and local and global arrangement of the reinforcement phase, the properties and design of an implant can be varied and tailored to suit the mechanical and physiological conditions of the host tissues. It is, therefore, suggested that composite materials offer a greater potential of structural biocompatibility than the homogenous monolithic materials. They have reasonably adequate strength [145]. Moreover the human tissues are essentially composite materials with anisotropic properties, which depend on the roles and structural arragements of various components (e.g. collagen, elastin, and hydroxyapatite) of the tissues. For example, the longitudinal mechanical properties of cortical bone are higher than the transverse direction properties (see Table 2). These similarities have led to the development of composite biomaterials. Other reasons for the development of polymer composite biomaterials include: absence of corrosion and fatigue failure of metal alloys and release of metal ions such as Nickel or Chromium which may cause loosening of the implant, patient discomfort, and allergic skin reactions; and low fracture toughness of

Table 2
Mechanical properties of hard tissues [22]

Hard tissue	Modulus (GPa)	Tensile Strength (MPa)
Cortical bone (longitudinal direction)	17.7	133
Cortical bone (transverse direction)	12.8	52
Cancellous bone	0.4	7.4
Enamel	84.3	10
Dentine	11.0	39.3

Table 3
Mechanical properties of soft tissues [22]

Soft tissue	Modulus (MPa)	Tensile strength (MPa)
Articular cartilage	10.5	27.5
Fibrocartilage	159.1	10.4
Ligament	303.0	29.5
Tendon	401.5	46.5
Skin	0.1–0.2	7.6
Arterial tissue (longitudinal direction)		0.1
Arterial tissue (transverse direction)		1.1
Intraocular lens	5.6	2.3

Table 4
Mechanical properties of typical metallic and ceramic biomaterials [22]

Material	Modulus (GPa)	Tensile strength (MPa)
Metal alloys		
Stainless steel	190	586
Co–Cr alloy	210	1085
Ti-alloy	116	965
Amalgam	30	58
Ceramics		
Alumina	380	300
Zirconia	220	820
Bioglass	35	42
Hydroxyapatite	95	50

Table 5
Mechanical properties of typical polymeric biomaterials [22]

Material	Modulus (GPa)	Tensile strength (MPa)
Polyethylene (PE)	0.88	35
Polyurethane (PU)	0.02	35
Polytetrafluoroethylene (PTFE)	0.5	27.5
Polyacetal (PA)	2.1	67
Polymethylmethacrylate (PMMA)	2.55	59
Polyethylene terepthalate (PET)	2.85	61
Polyetheretherketone (PEEK)	8.3	139
Silicone rubber (SR)	0.008	7.6
Polysulfone (PS)	2.65	75

ceramic materials which make them a difficult choice for load bearing applications. Composite materials offer several other significant advantages over metal alloys and ceramics in correcting the above mentioned or perceived deficiencies [88,226,229]. Metals alloys and ceramics are radio opaque and in some cases they result in undesirable artifacts in X-ray radiography [14]. In the case of polymer composite materials the radio transparancy can be adjusted by adding contrast medium to the polymer. Moreover the polymer composite materials are fully compatible with the modern diagnostic methods such as computed tomography (CT) and magnetic resonance imaging (MRI) as they are non-magnetic. Considering their light weight and superior mechanical porperties, the polymer composites are also used as structural components of these imaging devices. Some times, the unreinforced polymers may not have properties sufficient for intended application. For example, fiber reinforced UHMWPE has superior creep and fatigue resistance than the unreinforced UHMWPE. Higher creep and fatigue resistance properties are desirable in total knee joint replacement. As shown in Fig. 1, over the years a wide variety of polymer composite materials have been developed for various biomedical applications [198]. The following sections present details of polymer composite biomaterials in terms of hard tissue and soft tissue applications. In each section, for the benefit of readers, general information regarding sturcture and function of tissues, purpose and type of implants or devices, and various other materials used are also briefly presented. Glossary of medical terms used in this paper is given in Appendix A.

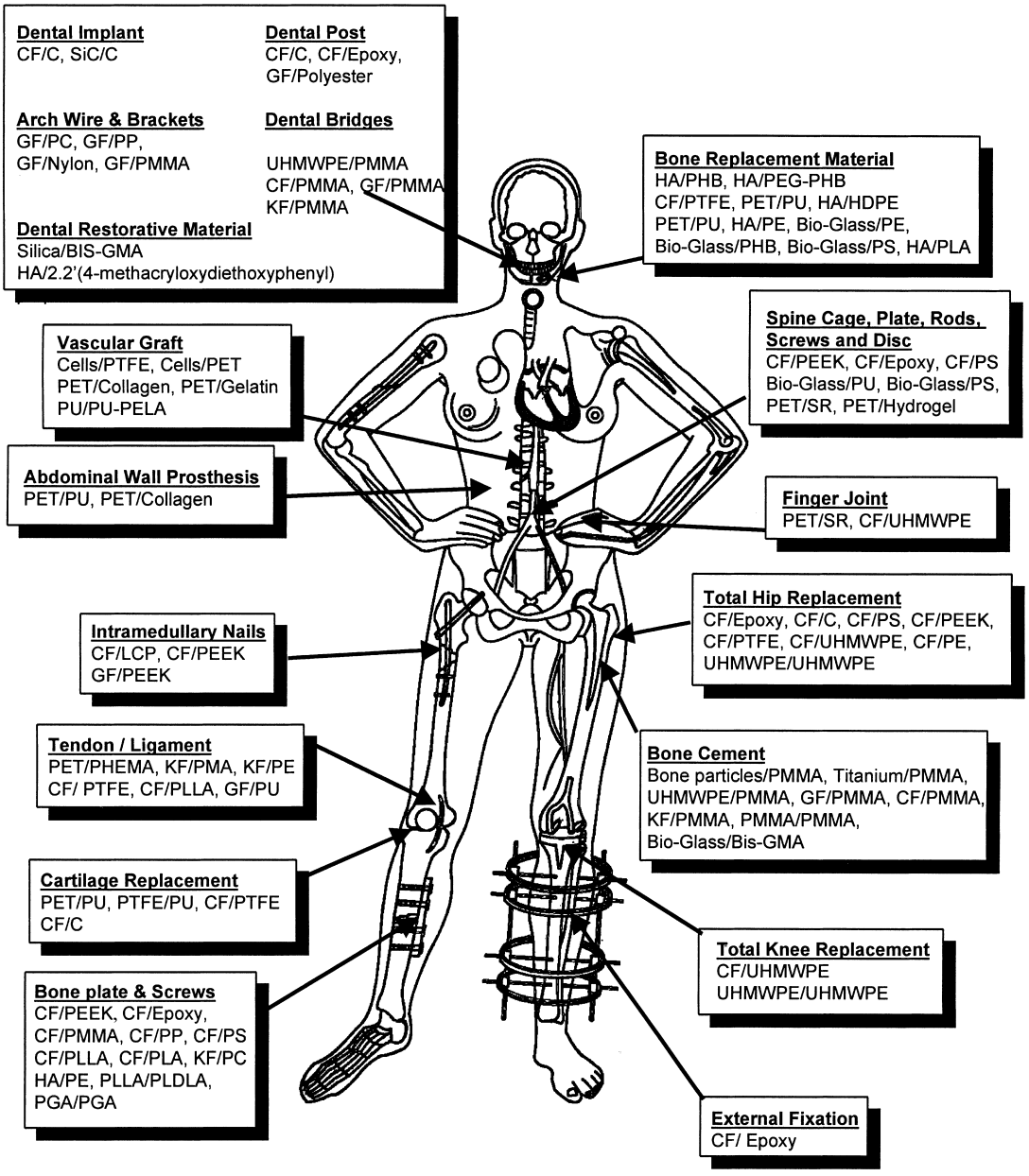
2. Hard tissue applications

2.1. Bone fracture repair

Bones of the skeletal system provide the supporting structure for the body. Bone is a structural composite composed of collagen fibers with hydroxyapatite nanocrystalls precipitated along the collagen fibrils [195]. Bone also contains other constituents such as mucopolysaccharides, blood vessels, and bone cells. The low elastic modulus collagen fibers are aligned in bone along the main stress directions. The high elastic modulus hydroxyapatite mineral comprises approximately 70% of the dry bone mass and contributes significantly to the bone stiffness. Bone can remodel and adapt itself to the applied mechanical environment, which is generally known as Wolff's law (see Appendix A). Density of the living bone is influenced by the stress condition applied to the bone. Higher applied stress leads to denser bone. Conversely, if the applied stress is lower than the normal physiological load, the bone mass decreases and leads to bone weakening. Bone is an anisotropic material because its properties are directionally dependent (Table 2). Bone is generally weak in tension and shear, particularly along the longitudinal plane. Under excessive loading or impact bone fractures, and there are many types of bone fractures depending on the crack size, orientation, morphology, and location. Readers are recommended to refer to AO (Arbeitsgemeinschaft fur Osteosynthesefragen)/ASIF (Association of Surgeons for Internal Fixation) documents for detailed classification of bone fractures. Bone fractures are treated (anatomic reduction) in different ways and they may be grouped into two types namely external fixation and internal fixation. The external fixation does not require opening the fracture site whereas the internal fixation requires opening of the fracture site. In the external fixation approach the bone fragments are held in alignment through various means such as splints, casts, braces, and external fixator systems. Casting materials or plaster bandages are used to form splints, casts or braces [20]. The casting material essentially is a composite material made of woven cotton fabrics (woven gauze) and Plaster of Paris matrix (calcium sulphate). Other reinforcements include fabrics of glass and polyester fibers. Although the plaster bandages have many advantages, they also have many disadvantages such as messy application, heavy, bulky, low specific strength and modulus, low water resistance, low fatigue strength, radiopaque, and long setting time to become load bearing. Recently, casts made of glass or polyester fiber fabrics, and water-activated polyurethanes are gaining popularity. An ideal cast material should be easy to handle, light weight, conformable to anatomical shape, strong, stiff, water proof, radiolucent, and easy to remove. More over it should be permeable to ventilation without which the patient's skin may be scorched or weakened. To address this specific problem, recently Philips [187] developed a new breathable cast material using double wall knitted fabrics as reinforcement.

A typical external fixation system [16] comprises of Kirschner wires or pins that are pierced through the bone and held under high tension by screws to the external frame (Fig. 1). The wires can be oriented at different angles across the bone, and their tension is adjusted to provide necessary fixation rigidity. To ensure stability, the external fixators are designed with high rigidity and strength. Traditional designs are made of stainless steel, which is heavy and causes discomfort to the patients as they carry the system for several months. External fixators constructed from CF/epoxy composite materials are gaining acceptance owing to their lightweight yet sufficient strength and stiffness [15]. Moreover, the evaluation of the bone union by radiography becomes easy, as the radiolucency of polymer composites is good and they do not cause artifacts in the radiographs. The external fixation is also used for bone lengthening purposes.

In the internal fixation approach the bone fragments are held together by different ways using implants such



CF: carbon fibers, **C**: carbon, **GF**: glass fibers, **KF**: kevlar fibers, **PMMA**: Polymethylmethacrylate, **PS**: polysulfone, **PP**: Polypropylene, **UHMWPE**: ultra-high-molecular weight polyethylene, **PLDLA**: poly(L-DL-lactide), **PLLA**: poly (L-lactic acid), **PGA**: polglycolic acid, **PC**: polycarbonate, **PEEK**: polyetheretherketone; **HA**: hydroxyapatite, **PMA**: polymethylacrylate, **BIS-GMA**: bis-phenol A glycidyl methacrylate, **PU**: polyurethane, **PTFE**: polytetrafluoroethylene, **PET**: polyethyleneterephthalate, **PEA**: poltethylacrylate, **SR**: silicone rubber, **PELA**: Block co-polymer of lactic acid and polyethylene glycol, **LCP**: liquid crystalline polymer, **PHB**: polyhydroxybutyrate, **PEG**: polyethyleneglycol, **PHEMA**: poly(20hydroxyethyl methacrylate)

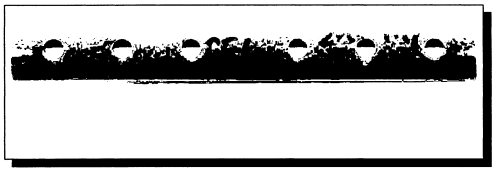
Fig. 1. Various applications of different polymer composite biomaterials.

as wires, pins, screws, plates, and intramedullary nails. The conventional implants are made of stainless steel, Co–Cr, or Ti alloys. The surgeon based on his experience and the type of fracture judges the bone fracture treatment method. Surgical wires and pins are the simplest implants used to hold the small fragments of bones together. For example wires are used to reattach the greater trochanter, which is often detached during total hip joint replacement. They are also used to provide additional stability in long oblique or spiral fractures of long bones (femur, humerus, radius, ulna, tibia, and fibula). Most widely used bone screws are two types, cortical bone screws (with smaller threads), and cancellous screws (with larger threads). They are used either to

directly fasten bone fragments together or to attach a plate to the fractured bone. However proper implant design and surgical technique must be utilized to ensure the desired biomechanical outcome of the fixation and to avoid additional tissue trauma and devascularization at the fracture site [41]. Fracture healing also would depend on the patient activities, as they determine the stable or unstable mechanical conditions at the fracture site. It may be noted that all these implants are temporarily placed inside the body. After satisfactory healing of the bone fracture, the implants may be removed based on the discretion of the surgeon.

2.1.1. Bone plates

Plate and screw fixation as shown in Fig. 1 is the most popular method for rigid internal fixation of the fractured bone. The bone plates are also called osteosynthesis plates. They are made of stainless steel, Cr–Co and Ti alloys. The rigid fixation is designed to provide high axial pressures (also known as dynamic compression) in the fragments of the bone, which facilitate primary bone healing without the formation of external callus. This method allows the exercise of joints near the fracture site just after the operation. After a complete bone healing has been obtained by the plate fixation, normally it takes from 1 year to 2 years after the operation, the plate and screws are removed. However, the rigid fixation is not free from complications and reported that it results in bone atrophy beneath the plate. There is a possibility of refracture of bone after the removal of the plates due to bone atrophy [60,95,264]. This is attributed to the stress shielding effect explained earlier. It may be noted that the modulus of stainless steel (210–230 GPa) is much higher than 10–18 GPa modulus of the bone. The stiffness mismatch results in a situation that the plate transmits the majority of the stress, and the bone directly beneath the plate experience less stress even after the fracture has been repaired [233]. The bone underneath the plates adapts to the low stress and becomes less dense and weak. Therefore, the strength of the healed bone is low. Consequently, there is a possibility of bone refracture upon removal of the fixation plate [44]. The stress shielding effect is more pronounced with the stainless steel plates than the Ti alloy plates. Moyen et al. [168] and Uhthoff and Finnegan [238] reported that the magnitude of bone atrophy under a Ti alloy plate is significantly lower than that under a stainless steel plate. It may be noted that the modulus of stainless steel (230 GPa) is higher than that of the Ti alloy (110 GPa). This example suggests that 'less rigid fixation plates' diminish the stress-shielding problem and it is desirable to use plates whose mechanical properties are close to those of the bone. In other words reduced stiffness mismatch between the implant and the host tissues. The adaptation of stiffness also changes the fracture healing mechanisms. Due to the higher strains at the fracture site, primary healing is no longer possible and is replaced by a more physiological bone healing process, which is characterized by the formation of an external callus bridging the fracture. Thereby, the callus increases the cross-section of the newly formed bone and, thus, prevents refracture. In early studies, researchers tried using polymers such as PA, PTFE, and polyester for bone plate applications, and found them to be not suitable because of their too low stiffness. They overlooked the fact that the materials proposed for bone plate application must also posses sufficiently high fatigue strength (comparable to stainless steel), as the orthopedic devices are subjected to extremely high cyclic loads, and must not lead to large strains at the fracture site, which may affect the bone union. It is now clearly established that any new material proposed for bone plate application must have sufficiently high fatigue strength and appropriate stiffness. Polymer composite materials offer desired high strength and bone like elastic properties [28]. Hence, several investigators proposed a variety of polymer composite materials for bone plate applications (Fig. 2) [86,227]. They may be grouped into non-resorbable, partially resorbable, and fully resorbable bone plates [47,133]. The non-resorbable composite plates are made of either thermoset polymer composites or thermoplastic composite materials. CF/epoxy, GF/epoxy are few examples of non-resorbable thermoset composites [5,29,30,159,223]. Some researchers expressed concern over the toxic effects of monomers in partially cured epoxy composite materials [167,184] and hence research activity on these materials gradually decreased. As the technology for making good quality thermoplastic composites made available, researchers developed CF/PMMA [263], CF/PP [43], CF/PS [48,105,107,159], CF/PE [209], CF/nylon, CF/PBT [77], and CF/PEEK [118,135,157,185, 200,249,253] non-resorbable thermoplastic composite



(a)

(b)

Fig. 2. Bone plates made of (a) CF/epoxy and (b) CF/PEEK composite materials.

bone plates. Unlike the thermoset composites, the thermoplastic composites are considered free from the complications associated with unused monomers. More over, similar to metal alloy plates, thermoplastic composite plates can be bent or contoured (under some conditions) to the shape of the bone at the time of surgery. At the moment there is insufficient data on the long-term in vivo behavior of non-resorbable thermoplastic composite materials. Among various materials investigated, the CF/PEEK is reportedly biocompatible [167] and has good resistance to hydrolysis and radiation (a sterilization method) degradation. The other promising properties include high strength, fatigue resistance [51,157], and biological inertness with no mutagenicity or carcinogenicity [44]. The tissue response to carbon fibers and composite debris has been described as minimal. Initially, researchers used short carbon fiber reinforced PEEK composites, as the technologies for fabricating continuous fiber reinforced PEEK composites were not available at that time. As can be expected from the composite reinforcement principles, the short fiber composites posses low modulus and strength compared to continuous fiber reinforced composite materials [77]. This means that plates made of short fiber composites must have greater bulk to approximate the mechanical stiffness required for a bone plate. The bulk limitation of short fiber composites may be increased considering their susceptibility to *in vivo* degradation. Hence, there is a need to develop suitable technologies to fabricate good quality continuous carbon fiber reinforced PEEK composites. Mayer [155,156] developed knitted CF/PEEK composite bone plates using commingled yarns of carbon and PEEK fibers. Recently, Ramakrishna et al. [200] developed braided CF/PEEK composite bone plates using a new technique [276]. They initially made micro-braided yarns by combining carbon and PEEK fibers. Micro-braided yarns were again braided into flat fabrics of desired dimensions. Compression molding above the melting point of PEEK matrix resulted in continuous CF/PEEK composites bone plates. Considering the superior mechanical properties of continuous carbon fiber reinforced PEEK composites, it is possible to produce relatively less bulky bone plates with out compromising the mechanical requirements of the plate. Researchers also developed CF/carbon [23] and CF/PEEK [147,185] composite screws (Fig 3), for osteosynthethesis. The squeeze casting method developed by Peter et al. [185] uses a new net shape flow process, which allows fabrication of complex shaped components with fiber contents as high as 62% by volume. The fatigue properties of the implants made by this process surpass those of the titanium implants by up to 100%. Combination of polymer composite plates and screws overcomes the corrosion problem faced by the metal plates and screws. The non-resorbable polymer composite materials are designed to be stable in in vivo conditions with no change in the plate stiffness with implantation time.

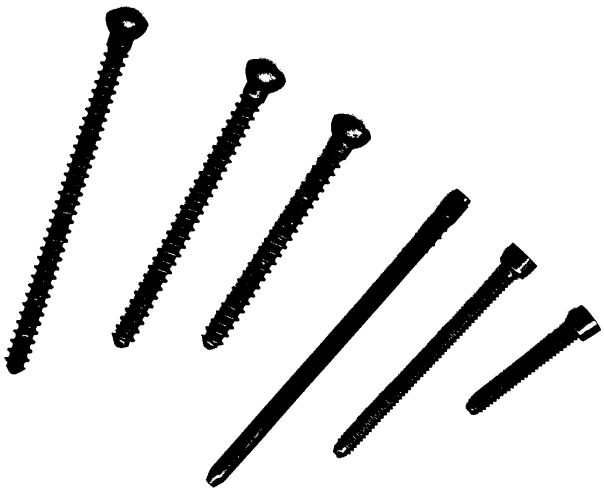


Fig. 3. CF/PEEK composite screws.

As the bone healing progresses, it is desirable that the bone is subjected to gradual increase of stress, thus reducing the stress-shielding effect. In other words, the stress on the plate should decrease with time whereas the stress on the bone should increase. This is possible only if the plate looses rigidity in in vivo environment. The non-resorbable polymer composites do not display this desired characteristic. To meet this need, researchers introduced resorbable polymers for bone plate applications [75]. The polymers such as poly(lactic acid) (PLA) and poly(glycolic acid) (PGA), resorb or degrade upon implantation into the body [150,177]. As such these polymers are either brittle or too weak and flexible for safe clinical use in load bearing applications. Many bioresorbable polymers found to loose most of their mechanical properties in few weeks. Tormala et al. [236] and Choueka et al. [42] proposed fully resorbable composites by reinforcing resorbable matrices with resorbable fibers (poly(L-lactic acid) (PLLA) fibers and calcium phosphate based glass fibers). One of the advantages often sighted for resorbable composite prostheses is that they need not be removed with a second operative procedure, as is recommended with metallic or non-resorbable composite implants. The maximum mechanical property of resorbable materials is continues to be a limitation and hence they are limited to only applications where the loads are moderate [215]. In order to improve mechanical properties, the resorbable polymers are reinforced with variety of non-resorbable materials including carbon fibers [55,170,180,235,272] and polyamide fibers [206,208]. Because of the non-resorbable nature of reinforcements used these composites are called partially resorbable composites. According to Zimmerman et al. [272], CF/PLA composite possessed superior mechanical properties before the implantation. However, they lost mechanical properties too rapidly in

in vivo environments because of delamination. Further work is necessary to tailor the composite material such that the resorption of the plate and the healing rate of the bone are synchronized [65]. The long-term effects of resorbed products, and biostable or slowly eroding fibers in the living tissues are not known, and these are the concerns yet to be resolved [27].

2.1.2. Intramedullary nails

Intramedullary nails or rods are mainly used to fix the long bone fractures such as fracture of femoral neck or intertrochanteric bone fracture. It is inserted into the intramedullary cavity of the bone and fixed in position using screws or friction fit approach (Fig. 1). From the surgery point of view they can be inserted through a small skin incision without opening the fracture site which is not the case with the bone plates. However, the insertion of nail often requires reaming of the medullary canal, which affects intramedullary blood vessels and nutrient arteries. As opposed to the plate system mentioned above, the intramedullary nail fixation method places the neutral-axis of the nail-bone structure at the center of the bone itself. This also allows early mobilization and load bearing of the limb without the plaster support. In the case of plate fixation system, the neutral axis of the plate-bone structure is along the plate, and dynamic forces may cause fatigue failure of plate or screws. The nail must be of sufficient strength to carry the weight of the patient without bending in either flexure or torsion, yet not completely disrupt the blood supply. In order to achieve these objectives intramedullary rods with a number of cross-sectional areas and end designs have been employed. Stainless steel is one of the widely used materials in intramedullary nails. Recently, Lin et al, [145] proposed short GF/PEEK composite material for intramedullary application. The rationale behind this proposal is the claimed biocompatibility of the composite material and its matching mechanical properties compared to the cortical bone. Kettunen et al. [122] developed unidirectional carbon fiber reinforced liquid crystalline (Vectra A950) polymer composite intramedullary rod. The material is biologically inert, with flexural strength higher than the yield strength of stainless steel and elastic modulus close to the bone. Compared to the plate fixation, the intramedullary nail fixation is better positioned to resist bending since it is located in the center of the bone. However, its torsional resistance is much less than that of the plate, which may be physiologically critical.

2.2. Spine instrumentation

The spine serves two distinct and apparently conflicting roles. First, it must provide a strong, yet mobile, central axis onto which the appendicular skeleton is applied. Second, it must protect the spinal cord and the roots of delicate nerves connecting the brain to the periphery. The proper blending of mobility, stability, and structural integrity is essential to fulfill these goals simultaneously. The dual function is realized by a linked structure consisting of 33 vertebrae superimposed on one another. The vertebrae are separated by fibrocartilaginous intervertebral discs (IVD) and are united by articular capsules and ligaments. The IVD is a composite structure made up of a core, nucleus pulposus, surrounded by multilayered fibers (90 concentric layers) of the annulus fibrosis. The orientation of annulus fibers vary from 62° at the periphery to 45° in the vicinity of the nucleus, thus imparting structurally graded architecture to the disc [10]. The disc is covered on the upper and lower surfaces by a thin layer of cartilaginous endplates, which contain perforations that allow the exchange of water, nutrients and products of metabolism. The main role of the disc is to act as a shock absorber for the spine, to cushion adjacent vertebral segments. A number of spine related disorders is identified over the years. Often reported spine disorders include metastasis of vertebral body and disc, disc herniation, facet degeneration, stenosis, and structural abnormalities such as kyphosis, scoliosis, and spondylolistheses. Often one disorder has cascading effect on the other, and primary causes of many spinal disorders remain largely speculative. A variety of reasons including birth deformities, aging, tumorous lesions (metastasis), and mechanical loads caused by sports and work, lead to spine disorders.

In the case the defect is limited to few vertebrae alternative approaches such as: (a) spinal fusion and (b) disc replacement are used. These methods are used alone or in combination depending on the patient condition and prognosis. In broader sense, spinal fusion means surgical immobilization of joint between two vertebrae. Various methods are employed in spinal fusion. One such approach is the surgical removal of the affected (portions of) vertebrae and restore the defect using synthetic bone graft, as the autologous or homologous bone grafts are limited by risk of infection, shortage of donor bone sites (with risk of AIDS and hepatitis in the case of autologous donors), and postoperative resorption and collapse of the graft. Synthetic bone graft material must have adequate strength and stiffness, also capable of bonding to the residual vertebrae. Ignatius et al. [109] and Claes et al. [50] developed Bioglass/PU composite material for vertebral body replacement. Similarly Marcolongo et al. [151] developed Bioglass/PS composite material for bone grafting purposes. In vivo studies indicated that these materials are bioactive and facilitate direct bone bonding (osseous integration). Another approach is to use special vertebral prostheses such as baskets, cages, and threaded inserts, which are made of metals or bioceramics [240,259]. They are designed such that tissues grow into the prostheses there by ensuring rigid anchoring of protheses to the bone. Sometimes stainless steel or titanium rods, plates, and screws are

used in conjunction with these prostheses to provide necessary stabilization. Several problems have arisen with these devices. Due to the poor form fit of these implants, local stress concentrations are considered as a possible reason for bone resorption and implant loosening. Additionally the metallic implant systems complicate postoperative assessment with X-rays, computed tomography (CT), and magnetic resonance imaging (MRI) through reflection and artifacts. Inadequate biomechanical capabilities of bioceramic prostheses may lead to the collapse of instrumented spine and injury of neurological structures and blood vessels. To over come disadvantages of conventional materials, Brantigan et al. [32] and Ciappetta et al. [46] developed CF/PEEK and CF/PS composite cages for lumbar interbody fusion. The composite cage has an elastic modulus similar to that of the bone, thus eliciting maximum bone growth into the cage. The composite cages are radiolucent and therefore do not hinder radiographic evaluation of bone fusion. Moreover they produce fewer artifacts on CT images than other implants constructed of metal alloys. Researchers also developed CF/PEEK and CF/PS [44,185] composite plates and screws for stabilizing the replacement body and spine. Flexural and fatigue properties of the CF/PEEK composites are comparable to those of the stainless steel, which is normally used for spine plates and screws. The success rate of spinal fusion is poorly defined in the literature and varies in a very wide range between 32% and 98%. Biomechanical study also shows that fusion alters the biomechanics of the spine and causes increased stresses to be experienced at the junction between fused and unfused segments. This promotes further disc degeneration. This seems to contradict a primary purpose of the patient seeking treatment and that is to improve the mobility of his back, in addition to alleviating the pain. Such arguments have given rise to intervertebral disc prostheses.

Problems related to intervertebral discs are treated by replacing affected nucleus with a substitute material or by replacement of the total disc (nucleus and annulus) using an artificial disc [17]. Both methods require duplication of the natural structure, significant durability to last longer than 40 years, and ease and safety during implant placement or removal. Some researchers used metal balls to replace the nucleus after discectomy. These nucleus substitutes did not restore the natural flexibility of the disc. Problems included migration and subsidence of the balls into the vertebral bodies as pressure was not evenly distributed, and no pressure modulation was possible with position change. Concurrent to the development of metals balls, other researchers proposed injectable silicone elastomers or hydrogels as nucleus substitutes. Several artificial disc designs are proposed over the years [17]. A variety of materials such as stainless steel, Co–Cr alloy, PE, SR, PU, PET/SR [202,203,239], and PET/hydrogel [8] composites are proposed for disc prostheses either alone or in combinations. However, their performance is not yet been acceptable for long-term applications. To date, there has been no artificial disc that is able to reproduce the unique mechanical and transport behavior of a natural disc satisfactorily. This may be as a result of the difficulty in finding a suitable non-human experimental model to test devices in vivo. For total disc replacement, it is important to select materials and create designs, which possess the required biocompatibility and endurance, while providing kinematic and dynamic properties similar to the natural disc.

Structural abnormalities or curvatures (lordosis, kyphosis, and spondylolistheses) of spine are corrected using either external or internal fixations. Splints and casts form the external fixation devices. The internal fixations require surgery and there are many types of instrumentation (screws, plates, rods, and expanding jacks) available [33]. In some cases, an adjustable stainless steel rod, also known as a Harrington spinal distraction rod, is used to stabilize or straighten the curvature. The rod is attached to the spinous process at two points and by adjusting the rod length between the attachment points, the spine is straightened. Schmitt-Thomas et al. [213] made initial attempts to develop a polymer composite rod using unidirectional and braided carbon fibers and biocompatible epoxy resin. The main motivation for this work is to over come the problems of metal alloys such as corrosion and interference with the diagnostic techniques.

It may be noted that efficient fixation of spinal deformities is difficult. This is attributed to the irregular shape of the vertebrae, and complex and large forces the prostheses need to withstand. Most of the designs used in various spine instrumentation, and the criteria that have evolved are primarily based on general biologic and engineering principles. Unfortunately, the specific mechanical and physical properties required for ideal spine instrumentation have not yet been defined. Until controlled clinical investigations provide these guidelines, many materials and designs must be evaluated in the laboratory.

2.3. Joint replacements

Joints enable the movement of the body and its parts. Many joints in the body are synovial types, which permit free movement. Hence, we are able to do various physical activities such as walk, jog, run, jump, turn, bend, bow, stand, and sit in our daily life. Hip, knee, shoulder, and elbow are a few common examples of synovial joints. They all posses two opposing articular surfaces, which are protected by a thin layer of articular cartilage and lubricated by elastic-viscous synovial fluid. The fluid is made of water, hyaluronic acid, and high molecular weight mucopolysaccharides. The synovial fluid adheres to the cartilage and upon loading can be permeated out onto the surface to reduce friction. The

coefficient of friction in a synovial joint is less than 0.01, better than that of a skate blade on ice. Coordinating the ligaments, tendons, and muscles performs the actual articulation of the joint. Osteoarthritis is one of the common causes for joint degeneration and sometimes hypertrophic changes in the bone and cartilage of joints in middle aged people. This is associated with progressive wearing down of opposing joint surfaces with consequent distortion of joint position. Joints also become damaged upon exposure to severe mechanical or metabolic injury. Over the years a number of artificial joints have been designed to replace or augment many joints in the body. Unlike those used to treat bone fractures, the artificial joints are placed permanently in the body. The extensive bone and cartilage removed during implantation makes this procedure irreversible. Considering the extent of loading, complexity of joint function, and severity of the physiological environment, joint replacement is one of the most demanding of all the implant applications in the body. The most commonly used artificial joints are total hip replacement (THR) and total knee replacement (TKR) (see Fig. 1).

2.3.1. Total hip replacement

THR is the most common artificial joint in human beings [63]. For example, over 150,000 total hip replacements are performed every year in USA alone. Over the years the design of total hip replacement evolved completely from a simple intuitive design to biomechanics based functional design. A typical THR consists of a cup type acetabular component, and a femoral component whose head is designed to fit into the acetabular cup, thus enabling joint articulations. The shaft of the femoral component (also called femoral stem) is tapered such that it can be fixed into a reamed medullary canal of the femur. Several types of THRs are designed by changing the material and geometry of acetabular cups and femoral stems, and fixation methods. Conventional THRs use stainless steel, Co–Cr and Ti alloys for the femoral shaft and neck, and Co–Cr alloy or ceramics such as alumina and zirconia materials for the head or ball. Earlier designs of acetabular cups were made of Co–Cr alloy. An effort to minimize friction and eliminate metallic wear on particles led Charnley in the early 1960s to use polymers for the acetabular component. He first implanted the stainless steel femoral component with a mating acetabular component made of PTFE. The PTFE was selected for a number of reasons. It has a high thermal stability, it is hydrophobic, stable in most types of chemical environments, and generally considered to be inert in the body. It does not adhere to other materials. It has the lowest coefficient of friction of all solids. However, clinical studies involving PTFE acetabular cups in the total hip replacement prostheses showed unacceptably high wear and distortion. The wear debris resulted in extensive tissue reaction and even formation of granuloma. This is attributed to its low compressive stiffness and strength, and increased wear under high stresses during sliding. PTFE is no longer used in such load bearing applications. Subsequently acetabular cups made of UHMWPE were developed and found to be successful. The UHMWPE cups are usually supported with a metal backing. Some reported data suggest that creep deformation, plastic distortion, and high wear or erosion of UHMWPE is possible. Although the short-term function of UHMWPE acetabular cups is satisfactory, their long-term performance has been a concern for many years. To improve the creep resistance, stiffness and strength, researchers proposed reinforcing UHMWPE with carbon fibers [209,216,222] or UHMWPE fibers [61]. Deng and Shalaby [61] found no appreciable difference in wear properties of reinforced and unreinforced UHMWPE. With opposite results reported in the literature, the effect of carbon fibers on the wear characteristics of the UHMWPE is a controversial subject. In recent years, certain designs use dense alumina or zirconia ball and matching acetabular cup made of similar materials mainly because of potential advantages of ceramic materials in terms of high hardness and compressive strength, low coefficient of friction, low wear rate, and good biological acceptance of wear particles.

Although THRs are used widely, one of the major unsolved problems in this important application has been the mismatch of stiffness of the femur bone and the prosthesis. As mentioned above the commercial hip joint stems are made from metal alloys, which are isotropic and at least five to six times stiffer than the bone. It has been acknowledged that the metallic stems due to stiffness mismatch induce unphysiological stresses in the bone, thereby affecting its remodeling process. It is discussed that this leads to bone resoprtion and eventual aseptic loosening of the prosthesis (it may be noted that the aseptic loosening is also linked to wear particles/debris) [9,24,37,214,244,247,251]. This is particularly a problem with young and more active patients. This may cause severe pain and clinical failure necessitating repeat surgery. About 10–15% fail within 5–7 years. Gese et al. [74] demonstrated that Ti alloy stems result in a 50% reduction in the femur peak stress compared to the Co–Cr alloy stem. It has been acknowledged that the implant loosening and eventual failure could be reduced through improvements in the prosthesis design and using a less stiff material with mechanical properties close to the properties of bone (i.e. isoelastic materials). However, because of the high strength requirement for hip prosthesis design, materials suitable for these implants are very limited. Fortunately, the advanced polymer composites can offer strength comparable to metals, and also more flexibility than metals. Strength of composite stems can be changed without affecting stiffness and vice versa. More over they also offer the potential to tailor

implant properties by selecting material ingredients and spatially controlling ingradient composition and configuration, which is useful in reducing the development of high stress regions. This allows one to control engineering properties such as strength and modulus according to the performance requirements of the prosthesis. A prosthesis made of polymer composite with spatially or locally varying mechanical properties along the boundary of the prosthesis, results in a more uniform and efficient transfer of stress from the stem to the bone. This may lead to better bone remodeling and longer implant service life. Researchers introduced CF/PS [222] and CF/C [45] composite stems. They reported faster bone bonding in the case of composite implants compared to the high stiffness conventional implants. The quicker bone bonding or bone contact was attributed to the lower stiffness of the implant. The composite stems were found to be stable with no release of soluble compounds, and high static and fatigue strength. Chang et al. [40] made CF/epoxy stems by laminating 120 layers of unidirectional plies in a pre-determined orientation and stacking sequence. Simoes et al. [220] made composite stems using braided hybrid carbon–glass fiber preforms and epoxy resin. Some researchers [4,185,259,261] designed and injection molded CF/PEEK composite stems (Fig. 4), which possess a mechanical behavior similar to that of the femur. Animal studies indicated that CF/PEEK composite elicits minimal response from muscular tissue. Both the in vivo and in vitro aging studies confirmed mechanical stability of CF/PEEK up to 6 months (it may be noted that this period is short and further long term testing is needed). Finite element analyses and in vitro measurements [4,268,269] indicated that compared to conventional metallic stems more favorable stresses and deformations could be generated in the femur using composite stems. Due to complexities in the geometry of hip prostheses, hip loads, and material properties of composites, design of composite implants require greater attention in order to achieve the desired in vivo performance of the implants. It is in order to mention here that if one tries to reduce stress shielding by using a less stiff implant it leads to increased implant deformation and relative movement (also called micromotion) between the implant and bone tissue during loading. The micromotion also influences bone remodeling [214,244] and often leads to residual pain. The stress shielding and micro motion are conflicting phenomena [104,134]. In other words, for appropriate structural compatibility the implant design should reduce stress shielding and micromotion simultaneously.

In addition to the prosthesis design and material, the fixation method is also important for the success of THRs [207]. Various methods for fixing THRs to the bones can be grouped into four generic types namely mechanical means, cemented, ingrown, and adhered.

Fig. 4. An injection molded CF/PEEK composite stem for total hip joint replacement.

Currently the cemented and ingrown approaches are widely used. As the name suggests in the first approach, the implant is secured in the bone by press fitting and/or using a wide range of pegs, posts, and screws. In the last method, fixation is achieved by direct adhesion of stem to the bone. In the 'cemented' approach, the PMMA or PMMA variant bone cements are used to fix the total hip replacement. More details of bone cements are described in Section 2.3.4. The quality of cemented prostheses fixation depends on various factors such as cement thickness, voids in cement or blood and tissues in contact with the cement bed during operation [31,53,119,143,165]. Problems cited include thermal damage to the bone due to cement curing, cytotoxic effects of methacrylate monomers, migration of cement and other wear particles in the cement–bone interface or the physiological process of bone resorption and intramedullary canal widening [64,103,191,252,266]. The best way to overcome these problems is not use the bone cement. An alternative approach, known as 'cementless approach', promotes fixation by encouragement of tissue growth into porous surface of the stem. Porous surface coatings have been

fabricated from various materials including Bioglass, bioactive glass-ceramics, hydroxyapatite, and bioactive polymers. In other designs the prosthesis surfaces are sintered with metal wire meshes or beads. The surface bioactiveness and/or porosity facilitate in growth of bone tissues and thus good anchoring of the prosthesis to the bone. The main shortcoming of these cementless approaches is that the long time required for achieving rigid fixation. On the other hand, in the case of cemented implants, the firm fixation is immediate.

2.3.2. Total knee replacement

The knee joint has a more complicated geometry and biomechanics of movements than the hip joint. The incidence of knee injuries and degeneration is higher than most other joints. Similar to most other joint replacements, the knee joint replacement development has been an evolutionary process, relying on intuitive design, empirical data, and laboratory studies. A typical TKR mainly consists of femoral and tibial components (Fig. 1). The femoral component articulates on the tibial component. The materials used for femoral components are predominantly Co–Cr and Ti alloys [245]. The tibial component is made of UHMWPE polymer supported by a metallic tibial tray. Clinical data indicated that the UHMWPE undergoes cold deformation, which leads to sinking of prosthesis. Inoue et al. [111] simulated and compared the performance of metal alloy femoral component articulating on a UHMWPE tibial component, and metal alloy femoral component articulating on a fiber reinforced UHMWPE composite tibial component. It is reported that the former material combination resulted in a high stress concentration in the vicinity of tibial stem, whereas the later material combination resulted in minimal stress concentration. This also explains the reasons for sinking of knee prostheses. Carbon fibers were used to reinforce UHMWPE to reduce its cold flow (creep) deformation [219]. The reinforcement enhances the stiffness, tensile yield strength, creep resistance, and fatigue strength of UHMWPE [265]. However, the results describing the effect of carbon fibers on the wear characteristics of UHMWPE are contradictory. Early studies reported that wear is reduced because of carbon fibers. But the later studies reported that the composite wear rates were 2.6–10.3 larger than those of unreinforced UHMWPE. This was attributed to the poor bonding between the carbon fibers and UHMWPE. The addition of carbon fibers does not improve the resistance of the material to surface damage. It should be emphasized that the composite by itself may not be suitable for low friction bearing but a combination of a UHMWPE surface and a composite substrate appears to offer some advantages. Recently, Deng and Shalaby [61] reinforced UHMWPE polymer with UHMWPE fibers. They reported no difference in the wear characteristics of unreinforced and reinforced UHMWPE. However, the improved stiffness, strength and creep resistance properties of reinforced UHMWPE are desirable for the joint replacement application.

2.3.3. Other joint replacements

Other joint replacements include ankle, toe, shoulder, elbow, wrist, and finger joints. The success rate of these joint replacements is limited due to loosening of prostheses and hence they are used less commonly compared with THR and TKR. The prostheses failures are attributed to limited bone stock available for fixation, minimal ligamentous support, and high stresses on the prostheses. More details on these joints can be found in references [178,179]. Materials such as Co–Cr and Ti alloys, HDPE, and UHMWPE remain to be the candidate materials for these joint replacements. Some designs use CF/UHMWPE instead of UHMWPE to provide higher strength and creep resistance. In certain types (space filler design) of finger joint replacements, silicone rubber (SR) is considered. Tearing of SR at the junction of prosthesis and roughened arthritic bone is a major concern. In order to improve the tear strength and flexural properties of SR, it is reinforced with PET fabrics. Goldner and Urbaniak [79] reported that the composite prosthesis also successful in decreasing pain, improving stability, increasing hand function, and in providing an adequate range of motion.

2.3.4. Bone cement

Proper fixation to the bones is as important as the design of joint replacement itself. Several different methods are adopted for fixing the artificial joints to the bones. One of the earliest methods, is to press-fit the joint prosthesis into the bone using a grouting material called bone cement. The most widely used bone cement is based on PMMA, also called acrylic bone cement [210]. It is self-polymerizing and contains solid PMMA powder and liquid MMA monomer. It has minimal adhesive properties, because of which attachment requires undercuts, holes, or furrows in the prosthesis. Therefore, when the bone cement sets or hardens, it mechanically interlocks with the roughened bone surface and the prosthesis. Cement must endure considerable stresses in in vivo applications, thus strength characteristics are important for its clinical success. The main function of the bone cement is to transfer load from the prosthesis to the bone or increase the load carrying capacity of the surgical construct. Researchers expressed concern over the release of monomers into the blood stream. Concerns were also expressed about the exothermic reaction associated with polymerization process, which produces elevated temperatures in the tissues that may induce locally bone necrosis [64]. The polymerization process is also associated with undesirable shrinkage of acrylic polymer. Another issue is the deterioration of cement/implant or cement/bone inter-

face with time, leading to problems of mechanical failure and instability [31]. Fatigue failure has been found to be a predominant in vivo failure mode of bone cement [114,131]. Researchers have tried to improve bone cement mechanical properties by reinforcing with stainless steel and Ti alloy wires, and polymer fibers such as UHMWPE [192,231,243,267], Kevlar, carbon [189], and PMMA [76]. Use of such fiber reinforcement also reduces the peak temperature during polymerization of the cement, and thus reducing the tissue necrosis [231]. The reinforced cement posses higher fracture toughness, fatigue resistance and damage energy absorption capabilities than the unreinforced cement. In another approach, bone particles or surface-reactive glass powders are mixed with PMMA bone cement in order to combine immediate mechanical fixation of PMMA with chemical bonding of bone particles [137,175,179] or surface-active glasses (Bioglass) with the bone [225]. Formation of this chemical bond makes it possible for mechanical stresses to be transferred across the cement/bone interface in a manner that prevents the fracture of the interface even when the implant or the bone is loaded to failure. Despite the experimental evidence of superior mechanical performance, reinforced cements have not yet been accepted in current clinical practice, primarily because of limitations such as the addition of fibers increases the apparent viscosity of bone cement thereby severely decreasing its workability and deliverability. Furthermore, uniform distribution of fibers in the bone cement is difficult, if not impossible, to obtain.

Gerhart et al. [71] proposed partially resorbable bone cement, which is a composite of tricalcium phosphate particles and a gelatin matrix. It is intended to provide immediate structural support and subsequent resorption of resorbable component of the composite cement facilitates bone ingrowth and direct bonding by the host bone. In contrast, the standard PMMA bone cement does not permit direct bonding by the host bone even though it provides the immediate structural support. PMMA is vulnerable to the accumulation of fatigue damage, as repetitive mechanical stresses lead to loosening at the cement–bone interface. It is in order to mention that the usefulness of the partially resorbable bone cement may be limited by a tendency for particle migration away from the implant site. Moreover the strength of partially resorbable bone cement is considerably lower than that of the PMMA bone cement.

Optimum use of bone cement is very important, otherwise, cement failure leads to loosening of the implant, which in turn causes pain to the patient. As the implant loosens, greater loads are experienced by the implant. Excessive loosening necessities removal of the implant and also some times leads to implant failure. The guiding principles for developing new bone cements include, the cement can be shaped, molded or injected to conform to complex internal cavities in bone, it must harden in situ and develop mechanical properties sufficient to permit functional loading of the implant site, it should maintain adequate mechanical integrity long enough to provide useful stabilization of the implant, and it should not be a barrier to bone remodeling.

It is in order to mention that wear of articulating surfaces is the major concern of many joint replacements [21]. Particulate debris that is formed becomes incorporated into the surrounding tissues, and even though the material may be quite inert in the bulk form, the fine particles are much more reactive and thus cause tissue irritation and inflammation. This process if repeated excessively, leads to bone resorption, bone loss, implant loosening, and fracture of bone. Hence, wear rate and wear products are of great importance in the design of joint replacements. Many efforts have been made to measure the rate of wear debris production in the laboratory. In general, the results depend on the geometry of the test, on the lubricant selected to simulate synovial fluid, and to some degree, on the experimenter. There have been great difficulties encountered in reproducing in vitro experimental results. Due to the inherent complexity of conducting a wear test, the exact mechanisms of wear and wear rate, and isolated effects of wear debris on the body are not clear. It is believed that more than one mechanism may take place simultaneously. Many studies are being conducted to understand the local and systemic effects of wear particles or debris.

2.4. Bone replacement (synthetic bone graft) materials

Synthetic bone grafts are necessary to fill bone defects or to replace fractured bones [128]. The bone graft material must be sufficiently strong and stiff, and also capable of bonding to the residual bones. PE is considered biocompatible from its satisfactory usage in hip and knee joint replacements for many years. Stiffness and strength of PE are much lower than those of the bone. For load bearing applications, properties of PE need to be enhanced. In order to improve the mechanical properties some researchers [25,26,58,91,223,255] reinforced PE using HA particles, which are bioactive. The resulting composite has an elastic modulus of 1–8 GPa and strain to failure value of over 90–3% as the volume fraction of HA increases to 50%. It was reported that for HA particulate volume fractions above 40% the composite is brittle. More over the bioactivity of the composite is less than optimal because the surface area of HA available is low and the rate of bone bonding of HA is slow. Further work requires consideration of using more bioactive materials such as Bioglass as reinforcements in PE [91,92]. A typical composition of Bioglass is 45% SiO_2, 6% P_2O_5, 24.5% Cao, 24.5% Na_2O by weight. The Bioglass reacts with physiological fluids and forms tenacious bond to hard or soft tissues

through cellular activity. To increase the interface between HA particles and the bone tissues, some researchers developed partially resorbable composites. They reinforced resorbable polymers such as PEG, PBT [146], PLLA [96,205,241,242], PHB [25,126], alginate and gelatin [124] with bioactive particles. Upon implantation, as the matrix polymer resorbs, more and more bioactive particles come in contact with the growing tissues, thus achieving good integration of the biomaterial into the bone. The wide range of material combinations offers the possibility of making composites with various desired properties such as stiffness, strength, biodegradation, and bioactivity.

2.5. Dental applications

All teeth are made of two portions, crown and root, which are demarcated by the gingiva (gum). The root is placed in a socket called alveolus in the maxillary (upper) or mandibular (lower) bones. Teeth possess a thin (<1 mm) surface layer of highly mineralized (90%) dental enamel (the hardest substance found in the body). The calcium salts of enamel are arranged as fine prisms running perpendicular to the surface. Underlying and supporting this is dentine, a less mineralized (70%) tissue that contains fine liquid-filled tubules running through to the pulp chamber. The pulp chamber carries the nerve and extends up through the root to the center of the tooth. In place of enamel, the surface of the root portion of tooth is covered by cementum, a mineralized tissue similar to bone. Teeth are non-homogenous, anisotropic, and unsymmetrical. Teeth experience a varied amounts and types (compression, flexural, torsion, and their mixed versions) of forces during mastication or chewing. Masticatory and traumatic forces vary from 100N to 450N [54,99,112].

Dental treatment is one of the most frequent medical treatments performed upon human beings. Dental treatment ranges from filling cavities (also called 'dental caries') to replacing fractured or decayed teeth. A large variety of materials are used in the dental treatments such as cavity lining, cavity filling, luting, endodontic, crown and bridge, prosthetic, preventive, orthodontic, and periodontal treatment of teeth. These materials are also generally described as biomaterials. The choice of material is dependent on its ability to resemble the physical, mechanical and esthetic properties of natural tooth structure. Here we only consider the applications in which composite materials are used, or the potential of using composite materials, is considerably high.

Dental restorative materials as the name suggests are used to fill the tooth cavities (caries) and some times to mask discoloration (veneering) or to correct contour and alignment deficiencies. Amalgam, gold, alumina, zirconia, acrylic resins and silicate cements are commonly used for restoring decayed teeth. Amalgam and gold are mainly used in the restoration of posterior teeth, and not preferred for anterior teeth for cosmetic reasons. Moreover there is concern over the long-term toxicity of silver-mercury amalgam fillings. Acrylic resins and silicate cements have been used for anterior teeth. However, they exhibit poor mechanical properties, which lead to short service life and clinical failures. Dental composite resins, which are translucent with a refractive index matching that of the enamel, have virtually replaced these materials and are very commonly used to restore posterior teeth as well as anterior teeth. The dental composite resin comprises of BIS-GMA as the matrix polymer and quartz, barium glass, and colloidal silica as fillers. The BIS-GMA is derived from the reaction of bis (4-hydroxyphenol) and glycidylmethacrylate. Low viscosity liquids such as triethylene glycol dimethacrylate are used to lower the viscosity and inhibitors such as BHT (butylated trioxytoluene, or 2,4,6-tri-tert-butylphenol) are used to prevent premature polymerization. Polymerization can be initiated by a thermochemical initiator such as benzoyl peroxide, or by a photochemical initiator (benzoin alkyl ether) that generates free radicals when subjected to ultraviolet light from a lamp used by the dentist. In other types of composites a urethane dimethacryate resin is used rather than the BIS-GMA. The filler particle concentration varies from 33 to 78% by weight and size varies from 0.05 to 50 μm. The glass fillers reduce the shrinkage upon polymerization of the resin, and also the coefficient of thermal expansion mismatch between the composite resin and the teeth. They impart high stiffness and strength, and good wear resistance to the dental composite resins [121]. Strong bonding between the fillers and resin is achieved using silane-coupling agents [132]. Key requirements for a successful restorative material include: sufficiently low viscosity so as to enable it to fill the cavity completely; controllable polymerization; coefficient of thermal expansion similar to the dentine/enamel, otherwise the stresses due to the mismatch is thought to contribute to leakage of saliva and bacteria at the interface margins; low shrinkage; and good resistance to creep, wear and water absorption. When the dental composites are used as a posterior restorative material, their radio-opacity is very important. The detection of caries under a non-radio-opaque composite is virtually impossible, and would allow the caries process to continue undetected for far too long. It is not clear what the optimum radio-opacity for composite is, since excessive radio-opacity can potentially mask out caries lying behind the restoration. Nevertheless, the composite should at least be as radio-opaque as the enamel. Active research is being pursued to develop dental composite resins with improved performance.

In cases when the severely damaged tooth lacks the structure to adequately retain a filling or restoration, often pins are used. In situations where the amount of

coronal tooth structure remaining is small (also referred as pulpless tooth), a dental post or a cast dowel is used to reinforce the remaining tooth structure [100,149], on which the core and crown are built (Fig. 5a). The post is normally inserted in the root canal and fixed in position using dental cement. It provides a retentive support to the core and crown assembly, and also distributes the forces of mastication to the supporting structures: the root, periodontal ligament, and surrounding bone. Sometimes pins are used either alone or in combination with the post to provide retention to the core material. The core replaces the coronal tooth structure that has been lost because of caries and previous restorations. It provides a base that has sufficient bulk and retention for the final restoration, the crown. Cores are usually formed from dental composite resins or amalgam or may be cast in precious or nonprecious alloys in combination with the metal post [237]. Traditionally posts made of stainless steel, Ni–Cr, Au–Pt or Ti alloys are used based on the assumption that the post should be rigid. Failures reported include corrosion of posts, bending or fracture of posts, loss of retention, core fracture and root fracture. In recent years this old basic tenet has been strongly questioned and it has been suggested that the modulus mismatch between the post and the dentine should be reduced so as to minimize the occurrence of root fractures (root fracture frequency is 2–4%) and failure of restorations. Newer posts made of zirconia, short glass fiber reinforced polyester, and unidirectional carbon fiber reinforced epoxy composite posts [113,234] are introduced. These new posts are adequately rigid, resistant to corrosion and fatigue [196]. In the frame of an ongoing project at the National University of Singapore, one of the authors looked at the function of a dental post in order to fully understand its mechanical requirements. In addition to providing support to the core, the dental post also helps to direct occlusal and excursive forces more apically along the length of the root. A finite element study by Cailleteau et al. [38] indicated that a post-restored model results in a decreased level of stress along the coronal facial portion of the root surface which peaked abruptly near the apical end of the post (labels 1 and 13 in Fig. 5a indicate coronal and apical ends respectively). These findings contradict the belief that the conventional posts strengthen the tooth by evenly distributing the external forces acting on the tooth. An ideal post should have varying stiffness along its length. Specifically, the coronal end of the post should have higher stiffness for better retention and rigidity of the core, and the apical end of the post should have lower stiffness matching that of the dentine so as to over come the root fractures due to stress concentration. In other words, it is desirable to have a post with varying stiffness. A post with varying stiffness but no change in the cross-sectional geometry along its length is only possible by using functionally graded composite materials. Ramakrishna et al. [201] designed and developed a functionally graded dental post using braided CF/epoxy composite technology [70,277]. It has a high stiffness in the coronal region and this stiffness gradually reduces to a value comparable to the stiffness of dentine at the apical end. In addition to overcoming the root fracture, the graded stiffness post decreases the chances of the post loosening from the dentine by means of eliminating stress concentration in the dentine, and reduction of post/dentine interfacial shear stresses (Fig. 5b). This clearly suggests that innovations in composites design and fabrication lead to better prostheses with improved performance.

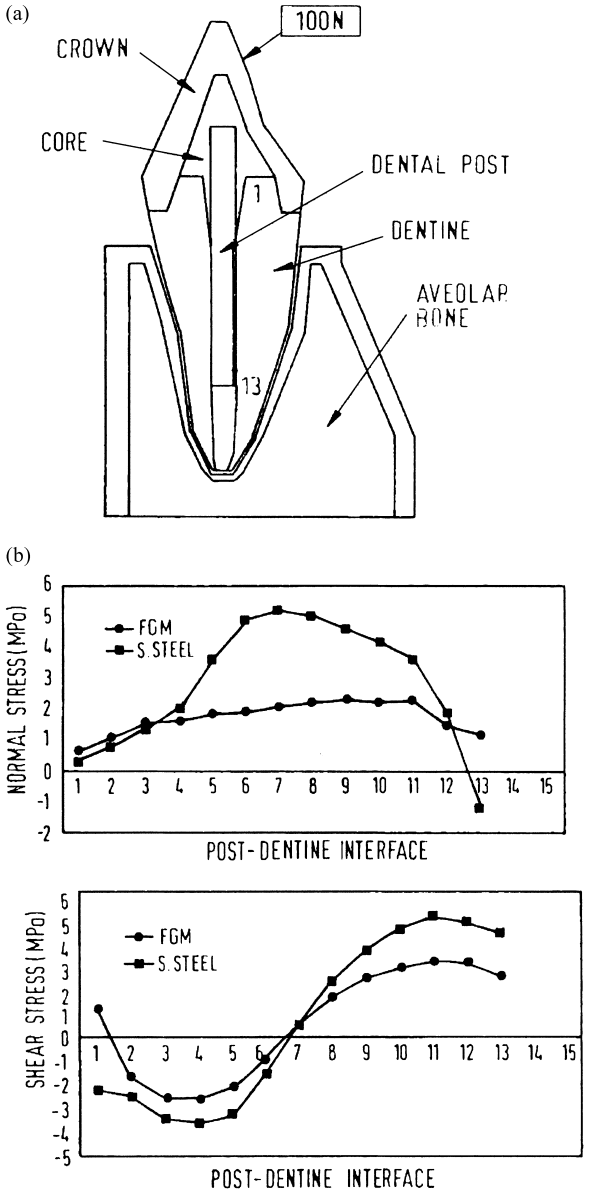


Fig. 5. (a) Post restored tooth, and (b) normal and shear stress distributions along the post-dentine interface. S.Steel indicates stainless steel post and FGM indicates functionally graded polymer composite post. Numbers 1 and 13 on the x-axis correspond to the coronal and apical ends, respectively.

In the extreme case, the damaged or condemned tooth is extracted and replaced with a dental implant. Dental implants are an artificial tooth roots that permanently replace missing teeth, and they are an alternative to bridges or false teeth. The dental implant may be designed to enter the jawbone or to fit on to the bone surface. The types of dental implants available are numerous. For example, certain root forms have threads, which facilitate to secure the root form into the jaw bone, whereas in some other designs, the surface is coated with porous bioactive materials, which allow bone growth and osseointegration. They are made of a wide range of materials [36,274] such as metals (Co–Cr–Mo alloys, Ti alloys, stainless steel, platinum, silver,), ceramics (zirconia, alumina, glass, and carbon), polymers (UHMPE, PMMA, PTFE, PS, and PET), and composites (SiC/carbon and CF/carbon) [1,35,148,166]. Compared to ceramic and metal alloys, the outstanding properties of composites are high or sufficient strength combined with low modulus. Such composite materials may offer protection against the alveolar bone resorption. Moreover fatigue properties of composites are far superior to the metal alloys and ceramics. The dental implants need to be designed to withstand extremely large and varying forces applied during mastication.

A bridge is a partial denture (false teeth) used to replace one or more tooth completely. In an extreme case removable dentures are used to overcome the loss of all the teeth. A large percentage of adults over the age of 50 years have full or upper or lower dentures. The root form mentioned previously is also used to anchor dentures and bridges to the jawbone. The high cost and time consuming preparation of current gold bridges has led to the development of relatively inexpensive and easy to use CF/PMMA [19], KF/PMMA [93], UHMWPE/PMMA [56] and GF/PMMA [164] composite bridges and dentures [67].

Orthodontic arch wires (approximately 0.5 mm diameter) are used to correct the alignment of teeth. This is facilitated by bonding orthodontic brackets on to the teeth. An arch wire is placed through the brackets and retained in position using a ligature, a small plastic piece. By changing the tension in the arch wire the alignment of the teeth is adjusted. The bracket acts as a focal point for the delivery of forces to the tooth generated from wire. It is important for a bracket to have high strength and stiffness to prevent distortion during tooth movement. This technique is also used to splint the traumatized teeth. Traditionally, the arch wires were made of stainless steel and Ni–Ti (beta titanium) alloys. Jancar and Dibenedetto [115], Jancar et al. [116] and Imai et al. [110] proposed GF/PC, GF/Nylon, GF/PP, and GF/PMMA composite materials for arch wires. The stated advantages of using composite arch wires include aesthetics, easy forming in the clinic, and the possibility of varying stiffness without changing component dimensions [273]. Commonly used materials in the manufacture of brackets are stainless steel, polycrystalline alumina, and single crystal alumina. Brackets made from metal alloys show high strength and stiffness but suffer from poor aesthetics. The ceramic brackets have improved aesthetics, however, ceramic brackets are bulkier than the metal alloy brackets. Furthermore, ceramic is abrasive to tooth enamel, and this has, therefore, limited the use of ceramic brackets to upper teeth. Some patients are hypersensitive to metals (Ni, Cr, and Co). It has been reported that these patients' immune system responds with vigorous foreign body allergic reactions causing dermatitis. Use of metallic restorations or braces is not recommended for metal sensitive patients. There is a need to develop suitable polymer composite orthodontic brackets. For any material combination to succeed in orthodontics, it is also important to consider the friction and abrasive wear characteristics of arch wires and brackets.

3. Soft tissue applications

Many different types of implants are used in the surgery to correct soft tissue deformities or defects which can be congenital, developmental, or acquired defects, the last category usually being secondary to trauma or tumor excision. Depending on the intended application, the soft tissue implants perform various functions: fill the space from some defect; enclose, store, isolate, or transport something in the body; and mechanical support or serve as a scaffold for tissue growth.

3.1. Bulk space fillers

Bulk space fillers are used to restore cosmetic defects, atrophy, or hypoplasty to an aesthetically satisfactory condition [158]. They are mostly used in the head and neck [39]. The materials used in these applications include SR, PE and PTFE [59,84]. The space-fillers are also investigated for the replacement of articular cartilage in the case of its deterioration by osteoarthritis. Articular cartilage, 1–2 mm thick, covers the opposing bony surfaces of typical synovial joints. The cartilage provides a means of absorbing force and provides low-friction bearing surfaces for joints. The cartilage replacement material must be hydrophilic with controlled water content, must have sufficient strength, and should be very smooth. Polymers such as SR and PTFE [178] are proposed to fill the defects in the articular surfaces or to replace meniscus or fibrous tissues following the condylar shave or high condylectomy in the treatment of painful arthritis and to restore normal joint function. Messner and Gillquist [163] reported that composites comprising PET or PTFE fabrics and PU are more suitable for this purpose, as they were found to reduce the cartilage

degeneration following the meniscectomy. At the same time Pongor et al. [190] clinically used woven carbon fiber fabrics and their composites for the treatment of cartilage defects. No inflammatory change or deterioration in joint damage was reported, indicating the usefulness of the prostheses. Further improvements in the composite materials in terms of retaining the shape of the implant could further improve the joint biomechanics.

3.2. Encapsulants and carriers

3.2.1. Wound dressing

Burn victims are often treated with skin dressings. In order to conform to irregular surfaces, the skin dressing must be elastic and flexible. There are two opposite requirements for skin dressing to meet: it should prevent loss of fluids, electrolytes and other biomolecules from the wound and obstruct bacterial entry, but it should also be permeable enough to allow the passage of discharge through pores or cuts. In addition it should be able to adhere to the wound surface, and be easy to peel from the skin without disturbing new tissue growth. Woven fabrics or porous layers of resorbable polymers such as collagen, chitin, and PLLA are used in many skin dressings. In hybrid skin dressings, synthetic polymers and cultured cells are combined to form vital/avital composites. They are designed to initiate, accelerate and control the natural skin repair process. Until now there is no synthetic material that can meet all the requirements of a skin substitute exactly.

3.2.2. Ureter prosthesis

Ureter prostheses made of PVC, PE, nylon, PTFE, and SR were used without much long-term success. They were not very successful because of the difficulty of joining a fluid-tight prosthesis to the living system. In addition, constant danger of microbial infection and blockage of passage by calcification deposits from urine have proven to be difficult to overcome. Polyester fiber reinforced glycol methacrylate gel prostheses with a fabric backing was reported to be successful [92,130]. The fabric backing facilitated easy attachment of a prosthesis firmly on to the mucous membrane without irritation, and the hydrophilic nature of the gel helped to maintain a clear inner space. A similar solution was proposed for the replacement of portions of intestinal wall. There is a need to develop new materials with improved surface properties of minimal microbial adhesion, low friction, and control of cell and protein adsorption.

3.2.3. Catheters

Catheters (tubes) are increasingly used to access remote regions of the human body to administer fluids (e.g. nutrients, isotonic saline, glucose, medications, blood and blood products) as well as to obtain data (e.g. artery pressure, gases, collecting blood samples for analysis). PU and SR are widely used materials for making catheters because of their flexibility and ease of fabrication into a variety of sizes and lengths in order to accommodate the wide range of vessels to be cannulated. SR is reinforced with silica particles to improve its tear strength and to decrease wettability. Andreopoulos, et al. [11] reported that with increasing the volume fraction of silica particles up to 35%, the tensile strength and elongation at break increased, whereas the elastic modulus only changed marginally. Since the catheter interfaces with blood, it is important that its design and material properties ensure blood compatibility, nonthrombogenicity, and inhibit infection. An ideal vascular catheter also must be flexible enough to allow vein and patient movement without becoming extravascular and damaging both the vessel and the surrounding structures. Catheters that are initially supple may become brittle over time, resulting in vascular wall damage. Newer designs consist of polymers (PU, LDPE, and PVC) reinforced with braided Nitinol (Ni–Ti alloy) ribbons with the purpose of making a catheter having an exceptionally thin wall, controlled stiffness, high resistance to kinking, and complete recovery in vivo from kinking situations.

3.3. Functional load-carrying and supporting implants

3.3.1. Tendons and ligaments

Artificial tendons and ligaments are the best examples of load-bearing soft tissue implants. A tendon is a strong fibrous band of tissue that extends from a muscle to the periosteum of the bone. A ligament is a connective tissue band that links bones in the vicinity of every synovial joint. Tendons and ligaments hold the bones of a joint thus facilitating their stability and movement. They also transmit force between muscle and bone. They are essentially composite materials comprising undulated collagen fiber bundles aligned along the length and immersed in a ground substance, which is a complex made of elastine and mucopolysaccharide hydrogel [193]. The unique mechanical feature of tendons/ligaments is their non-linear J-shaped convex stress–strain curve as opposed to the concave stress–strain relationships of common engineering materials. For example, the static tensile curve of ligaments characteristically exhibit a 'toe' region (low modulus) at low strain, a linear region at intermediate strain, and eventually a failure at high strain. Tissue structural parameters such as fiber composition and structure, hydration, fiber–matrix interaction, and fiber–fiber interaction determine its the mechanical behavior. Tendons have little regenerative capacity and require very long times to regenerate fully.

The use of biomaterials in tendon/ligament repair is one of the most demanding applications of prostheses in soft tissues. A ligament or tendon prosthesis should: (a)

possess the same flexibility as the natural tissue in order to bend around articulations and assure the transmission of the force to the muscle always in the mode of a traction (Seedhom, [218] reported that estimated forces in the anterior cruciate ligament of the knee joint are 196 N for level walking, 72 N for ascending stairs, 93 N for descending stairs, 67 N for ascending a ramp, and 445 N for descending a ramp), (b) reproduce similar mechanical properties including J-shaped stress-strain behavior, large extensibility without permanent deformation, and damping properties, and (c) assure time invariance of the mechanical properties. Biomaterials are used in a number of ways in tendon healing. They may be used to replace the tendon, they may be used to hold a damaged tendon in proper alignment, or they may be used to form a new sheath. In the last approach, a two-stage surgical procedure is followed. In the first operation, the tendon is replaced by a gliding implant that facilitates the formation of a new tendon sheath. In the second operation, a tendon graft replaces the gliding implant inside the newly formed sheath.

Synthetic biomaterials used thus far include UHMWPE, PP, PET, PTFE, PU, Kevlar 49, carbon, and reconstituted collagen fibers in the multifilament form or braided form [13,18,62,66,117,123,152,161,183,186,228,248]. Permanent fixation of the implant assumed to be provided by tissue ingrowth into the spaces between the filaments. The clinical experience with synthetic prostheses has so far been disappointing. The problems with synthetic prostheses include difficulty of anchorage to the bone, and abrasion and wear of the prostheses, which deteriorate in strength in the long term and lead to mechanical failure (such as fatigue). Further, the particulate matter generated by abrasion against rough bony surfaces may cause synovitis, as well as inflammation of the lymph nodes should the size of the particulate matter produced allow its migration to the nodes [218]. To reduce particle migration and improve handling properties, prostheses are coated with polymers such as SR, poly(2-hydroxyethyl methacrylate) (PHEMA), and PLA. Pradas and Calleja [193] reported that by combining flexible polymer such as PMA or PEA with crimped Kevlar-49 fibers, the stress-strain behavior of natural ligaments can be reproduced to a certain extent. Iannace et al. [108] and Ambrosio et al. [6,7] developed a ligament prosthesis by reinforcing a hydrogel matrix (PHEMA) with helically wound rigid PET fibers, and demonstrated that both static and dynamic mechanical behavior of natural ligaments can be reproduced. This has been achieved by controlling the structural arrangement of reinforcing fibers and the properties of the components. It may be noted that PET is sensitive to hydrolytic, stress induced degradation. Surgeons are still looking for suitable synthetic materials that adequately reproduce the mechanical behavior of natural tissue for long-term application, while they are currently using prostheses of natural tissues (homografts, allografts, and xenografts). Many consider that a combination of autegenous tissue and synthetic materials is an ideal choice for tendon/ligament prostheses. These materials reportedly possess the desired biomechanical properties such as low coefficient of friction, and improved compliance, strength, creep, and fatigue resistance.

3.3.2. Vascular grafts

Arterial blood vessels are complex, multilayered structures comprising collagen and elastin fibers, smooth muscle, ground substance and endothelium. The blood vessel is anisotropic because of the orientation of inherent fibrous components. Like other soft tissues, the blood vessel also behaves in a non-Hookean way when subjected to physiological loads, and displays J-shaped stress–strain behavior. Vascular grafts are used to replace segments of the natural cardiovascular system (mainly successful in the case of blood vessels with lumen diameter of over 5 mm) that are diseased or blocked (atherosclerosis, deposits on the inner surface of the vessels restricting the flow of blood and increasing blood pressure). A typical example is to replace a section of aorta where an aneurysm has occurred. Another example is the arteries in the legs of diabetic patients that have a tendency to be blocked. Grafts, essentially tubular structures, are inserted to bypass the blockages and restore circulation. Most widely used vascular grafts are woven or knitted fabric tubes of PET material or extruded porous wall tubes of PTFE and PU materials. The most important property of a graft is its porosity. Certain porosity is desirable as it promotes tissue growth and acceptance of the graft by the host tissues. However, excessive porosity leads to leakage of blood. Most synthetic grafts are preclotted prior to transplantation to minimize blood leakage. In another approach, vascular grafts are impregnated with collagen or gelatin to seal the pores and also to improve the dimensional stability of grafts. These are known as composite grafts. The seal degrades in approximately 2–12 weeks after the implantation. In addition to porosity, good handling and suturing characteristics, satisfactory healing (rapid tissue growth), mechanical and chemical stability (good tensile strength and resistance to deterioration) are major requirements of vascular grafts. Since vascular grafts are subjected to static pressure and repeated stress of pulsation in application, they should have good dilation and creep resistance. The fabric tubes are crimped to make them bulky, resilient, and soft. Moreover, crimping facilitates extensibility, and bending of fabric tubes without kinks and stress concentrations, which are very important in blood transporting vascular grafts.

PET (Dacron) vascular grafts (woven or knitted fabric tubes) are mainly successful in the replacement of large diameter blood vessels (12–38 mm diameter). A major issue for a vascular grafts is the reaction between

the surface of the material and blood that can cause destruction of blood cells and thromboembolism. Biocompatibility of PET fibers and fabrics is generally considered to be acceptable. Protein and platelet absorption of PET is minimal, however it is thrombogenic. PET vascular grafts are seeded with endothelial cells to reduce the thrombogenic character and to improve patency. These grafts are essentially composites of PET fabrics and cells (see Section 5.2 for further details).

Expanded PTFE (e-PTFE or Gore Tex) is widely used for medium diameter (6–12 mm) vascular grafts. A modified extrusion process produces the porous e-PTFE. The porous non-woven microstructure of e-PTFE provides vascular grafts with a mechanical behavior matching to that of the host blood vessels compared to the vascular grafts made of non-porous (solid) materials. Moreover, the inner (luminal) surface of e-PTFE graft facilitates formation of neointima (newly formed endothelial tissue lining) that avoid the complications such as formation of thrombi (blood clot) and emboli (dislodged blood clot). However, the exact mechanisms of neointima formation are not clear.

It is widely accepted now that a major requirement for optimal healing and patency of a vascular graft is matching of its mechanical properties to those of the anastomosed natural tissues. Lack of compliance matching with the host artery is detrimental to the acceptance of synthetic vascular grafts, when used in the reconstruction of arteries. A compliance mismatch results in a mechanical incongruity, and in a blood flow of high shear stress and turbulence, with local stagnation. These factors may lead to local thrombosis, and may damage the arterial wall. Hence, there is a greater need to match compliance of both the vascular graft and the attached blood vessel. The conventional vascular prostheses are predominantly rigid structures, lacking anisotropy and non-linear compliance. Gershon et al. [72,73] and Klein et al. [125] developed composite grafts comprised of polyurethane (Lycra trade name) fibers in a matrix of polyurethane (Pellathane trade name) and PELA (block copolymer of lactic acid and polyethylene glycol) mixture. The non-linear stress strain behavior and compliance of the composite graft are varied by controlling the fiber orientation [197]. The composite graft is anisotropic, and isocompliant with the natural artery. The matrix material is designed to resorb in *in vivo* condition. At the time of implantation the impervious graft prevents any loss of blood. The resorption of matrix material during healing process will result in pores. The ingrowth of granulation tissue into pores provides a stable anchorage for the development of a viable cellular lining. The optimum pore size of the outer and inner layers of the graft can be designed to meet the exact needs of ingrowth and anchorage. The composite grafts are in the clinical research phase and yet to be used clinically.

3.4. Others

Hernia is an irregular protrusion of tissue, organ, or a portion of an organ through an abnormal break in the surrounding cavity's muscular or connective tissue wall. A number of materials such as nylon, PP, PTFE, PET, carbon, stainless steel, and tantalum in the form of fabrics or meshes are used to repair hernias [246]. The fabrics or meshes facilitate tissue ingrowth thus providing stability to the prosthesis. Recently, Werkmeister et al. [250] developed PET fabrics coated with collagen and PU materials suitable for repairing hernia and abdominal wall (abdominal wall lines the abdominal cavity that contains liver, gallbladder, spleen, stomach, pancreas, intestine, and kidney) defects. The composite is designed to display adequate mechanical properties as well as facilitate tissue ingrowth. The composite material is reportedly superior to uncoated fabrics in terms of biocompatibility. Other suitable applications being currently investigated include tracheal prostheses (combined with stainless steel mesh or SR), prosthetic sphincters for gastrointestinal tracts, and urethral prostheses.

Prostheses are also used for restoring the conductive hearing loss from otosclerosis (a hereditary defect which involves a change in the bones of the middle ear). Otology prostheses made of polymers namely PMMA, PTFE, PE, and SR, and CF/PTFE composites have been tried to replace defective ossicles (three tiny bones of middle ear, malleus, incus, and stapes) (it may be noted that the clinically established prostheses are made from titanium, gold, stainless steel, hydroxyapatite, alumina und glassceramics). Migration of prostheses is the main problem reported and it is essential to apply suitable surgical method. Researchers [202,230] are also developing PE/PU flexible composite materials as tympanic membrane replacements. Tympanic membrane transmits sound vibrations to the inner ear through three auditory ossicles.

4. Other biomedical applications

4.1. Prosthetic limbs

Initial artificial legs are designed primarily to restore walking of the amputees. They were made of wood or metallic materials. These materials are limited by their weight, and poor durability due to corrosion and moisture induced swelling. As a result the user is often restricted to slow and non-strenuous activities. Strenuous activities, such as playing ball games and running are not possible due to the weight of these devices and their poor elastic response during stance. The lightweight, corrosion resistance, fatigue resistance, aesthetics, and ease of fabrication of polymer composite materials made them ideal choice for modern limbs systems [204]. Several designs of artificial limbs with different commercial names

are available. Thermoset polymer composites reinforced with glass, carbon, or Kevlar fibers are widely used in these systems [52]. A typical artificial leg system consists of three parts namely socket, shaft, and foot (Fig. 6). The most highly customized and important part of the prosthesis is the socket, which has to be fabricated individually to the satisfaction of each amputee. Sockets can be divided into two categories, namely, direct and indirect sockets. A widely used indirect socket is fabricated by wrapping several layers of knitted or woven fabrics [224] on a customized plastic mold, vacuuming the fabrics enclosed in a plastic bag, and impregnating the vacuumed fabrics with polyester resin. The socket is formed after the resin is cured under the vacuum pressuring condition. It is reported that the performance of an indirect socket depends mainly on the quality of the mold. Moreover, the fabrication process is time-consuming and greatly influenced by the prosthesist skills. A direct socket, as the name suggests, is fabricated directly on the stump of a patient, without using any kind of mold. Compared with indirect sockets, the benefit of direct socket fabrication is that it can reduce the amount of skill dependency in the creation of a socket and lead to reduction of fitting errors between the stump and the socket. In addition, the direct socket fabrication also reduces the number of patient visits and improves service to the physically disabled people. The direct sockets appeared in the market in recent years, are made using a combination of knitted or braided carbon or glass fiber fabrics and water-curable (water-activated) resins. As expected the braided fabric reinforced sockets are stiff and strong, whereas the knitted fabric reinforced sockets are flexible and more conformable to the patient's stump [102].

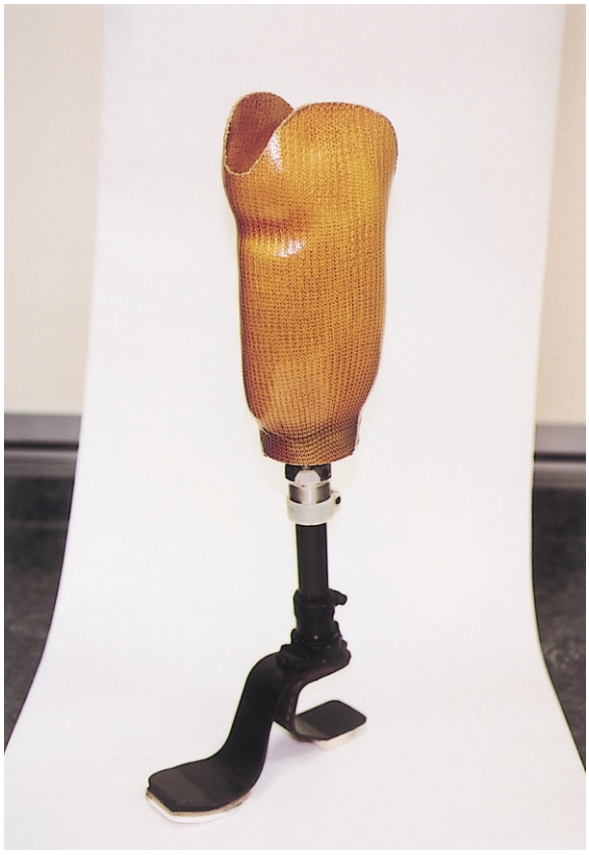


Fig. 6. Photograph of a typical prosthetic leg showing socket, shaft, and foot.

The shaft or stem is often made of filament wound or laminated woven/braided fabric carbon fiber reinforced epoxy composites. It provides structural support and force trasmittance to mimic the skeleton [69]. In some designs, the foot unit consists of heel and forefoot components, which are made of laminated CF/epoxy composites and are designed to serve as flat spring-like leaves so that the foot provides strong cushioning and energy storing effect [232]. They are designed to store energy during stance and release energy as body weight progresses forward, thus helping to propel the body and to achieve smooth ambulation. This gives the user a higher degree of mobility with a more natural feel compared with conventional wood prosthetic feet [78]. Delamination of plies is a major concern and need to be addressed for longer life of the foot. Polymer composites are also used in knee braces.

4.2. Medical instrumentation

High technology machines such as CT and MRI scanners are gaining wider usage for medical diagnostic purposes. These machines have larger bodies fitted with moving tables for the patients. The moving table needs to strong and stiff, at the same time lightweight, radiolucent and non-magnetic to obtain clear sliced images of the patient. As expected the moving tables are made of carbon fiber reinforced polymer composites [129]. These materials are also used in making surgical clamps, head rests frames, X-ray film cassettes and CT scan couches.

5. Critical issues

From the previous sections it is apparent that a wide variety of polymer composite materials were investigated or developed for possible biomedical applications. For the purpose of clarity, the various man-made polymer composite materials are classified into several subgroups as shown in Fig. 7. A composite material made of avital (non-living) matrix and reinforcement phases, is called 'avital/avital composite'. Alternatively, a composite material comprising of vital (living) and avital (non-living) materials is called 'vital/avital composite'. These composites are further discussed in Section 5.2. The avital/avital composites are analogous to polymer composites

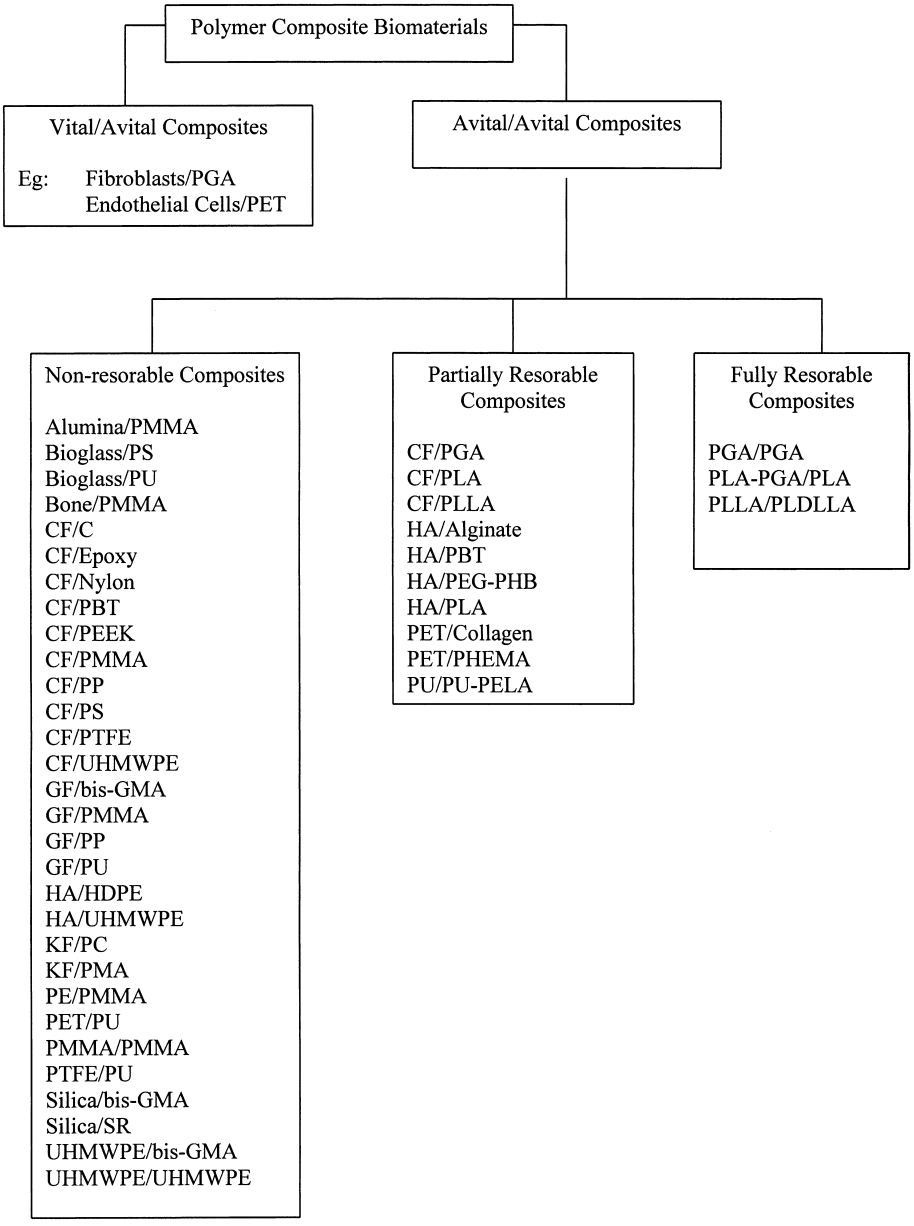


Fig. 7. Classification of man-made polymer composite biomaterials.

known to engineers. The avital/avital composites are further divided into non-resorbable, partially resorbable and fully resorbable composite biomaterials. The non-resorbable composites are designed not to degrade in the in vivo (inside the body) environment. They are particularly promising for long-term implants such as total joint replacements, bone cement, spine rods, fusion cages, discs, plates, dental posts, and hernia patches. They are also proposed for short term applications such as bone plates, rods, screws, ligaments, and catheters. On the other hand the resorbable composites are intended to loose their mechanical integrity in in vivo conditions. They are particularly promising as short-term or transient implants namely bone plates, screws, pins, rods, ligaments, tendons, bone replacement, vascular grafts, and artificial skin. The need and usefulness of non-resorbable and resorbable composites are highlighted in the previous sections. The specific issues common to various avital/avital composite materials are discussed further in the following sections.

5.1. Avital/avital composites

5.1.1. Effects of in-vivo environment, and new failure criteria

As mentioned earlier, the non-resorbable composites are intended not to degrade in in vivo conditions. However, some researchers pointed out that the in vivo conditions might introduce profound changes in the physical, chemical, and mechanical properties of composite bio-

materials. Hence, knowledge of the effects of the in vivo environment on the composite properties is very important [249]. McKenna et al [159] investigated the stability of GF/epoxy and CF/PS composites in simulated in vivo conditions (i.e. in vitro testing in saline solution). They reported only a small change in stiffness and strength of GF/epoxy composite whereas a significant reduction in the properties of CF/PS composite material. This difference was attributed to the variations in the fiber/matrix interfacial bond strengths of both the composite materials. Latour and Black [140] investigated the effect of simulated in vivo environments such as saline and exudate (it is acellular biologic fluid similar to interstitial fluid) on the fiber/matrix interfacial bond strength of CF/PC, CF/PS, KF/PC, and KF/PS composites, which are candidates for orthopedic applications. They adopted a single fiber pull-out test to measure the interfacial bond strength. The bond strength of each material combination was significantly degraded by exposure to either saline or exudate. The water and/or salt ions were found to be responsible for the deterioration of interfacial bond strength. Later, Latour and Black [141] also conducted fatigue studies on the CF/PS and KF/PS composites in simulated in vivo environments. They found that the fiber/matrix interface failed at approximately 10^5 load cycles at a maximum applied load level of only 15% of its ultimate dry bond strength without indication of an endurance limit being reached. They expressed serious concern about the durability of polymer composites in load bearing orthopedic applications. In another study, Brown et al. [34] investigated the effect of exposure to saline solution (0.9% NaCl) on the flexural and fracture toughness properties of short carbon fiber reinforced PS, PBT, and PEEK composites. CF/PS and CF/PBT composites showed significant degradation of mechanical properties following exposure to saline solution. However, no such reduction in mechanical properties was reported for the CF/PEEK composites. This was attributed to good bond between the carbon fibers and PEEK matrix [254].

Suwanprateeb et al. [223] conducted in vitro tests on HDPE and HA/HDPE in a simulated body environment, Ringer's solution. They reported that unreinforced HDPE properties were unaffected by the solution, whereas the composite creep resistance and stiffness decreased. The effect increased with increasing volume fraction of HA and time of immersion. The decrease in properties was attributed to penetration of solution into the material through the interface. Various methods have been developed to improve the interface of HA with a polymer matrix. Silane coupling agents [58], zinconyl salts, polyacids and isocyanates [146] were used to form direct chemical linkage between the HA particles and the polymer matrix. By optimizing the surface treatment, a further improvement of in vivo behavior of composites can be expected. However, Jancar and Dibenedetto [115] found opposite results. They used single fiber pull-out and flexural tests to investigate the effect of silane treatment on the interfacial bond strength of GF/PC and GF/PP composites. They reported best results for composites with untreated fibers compared to the composites reinforced with silane treated fibers. The silane treatment reportedly led to the problems of hydrolytic instability under extreme conditions of stress and moisture. The best results in the case of untreated fiber composite were attributed to the annealing treatment given to the composite, which resulted in a strongly bonded, highly water resistant interface through nucleation of highly ordered polycarbonate at the fiber/matrix interface. The above studies clearly indicate that the quality of fiber and matrix interface is of principal importance in determining the response of polymer composite materials to in vivo environments. The effect of in vivo exposure upon the fiber/matrix interface, and the subsequent effect upon the implant's mechanical properties must be considered in the design and selection of polymer composites to ensure satisfactory long term durability/performance in vivo. The review of present knowledge on the polymer composite biomaterials leads to the recognition that there is lack of accumulated experience and knowledge about the long-term stability of these materials in physiological environment. The studies reported in the literature only illustrate the effect of diffusion of environment on the mechanical properties of composite materials. It is to be noted that the in vivo conditions depending on the purpose and the site of implantation include different tissue fluids and dynamic mechanical loads. Hence, knowledge of combined effects of diffusion of environment and mechanical stresses (static and dynamic) on the long-term behavior of composite materials is important. More importantly, for the same implant, the results obtained from one composite material system cannot be extrapolated to another system, even though there may be one common phase in both the systems. Similarly, one composite system evaluated and found suitable for one biomedical application cannot be used in another application without systematic studies and design. This also calls for thorough experimental evaluation of durability of different polymer composite biomaterials in in vivo conditions. This knowledge is very important in making proper judgements for their clinical use.

The readers are reminded that the above discussion is limited to non-resorbable composites, which are designed to remain stable in vivo environment. In contrast, the resorbable composites are designed to be influenced by the in vivo environment. The components of resorbable composites are selected such that the water absorption (hydration) and/or enzymatic degradation leads to controlled degradation of mechanical integrity of the composite material. This involves simple intentional delamination to loss of total mass of the composite material. Current constitutive models and failure criteria used for engineering polymer composites may not be applicable to

the resorbable composites, as they are developed assuming no change in the material geometry and total mass. Hence, there is a need to develop new constitutive models as well as failure criteria to understand or simulate the in vivo behavior of resorbable composite materials. With regard to resorbable composite materials, the goal that remains to be achieved is how to tailor the composite material such that it would loose its mechanical properties at approximately the same rate as required by the intended application (related to tissue healing). Furthermore, after loss of the mechanical functionality the implant should disappear as fast as possible. Otherwise, the long residual time of the implant may lead to formation of a thick fibrous capsule, which subsequently results in undesirable calcification. An important aspect of bioresorbable biomaterials is that not only the original material but also the degradation products have to be non-toxic and removed from the body without side effects. Moreover they need to have adequate initial strength and stiffness at the time of implantation. Currently, this is an area of intensive research.

5.1.2. Improved test methods and new design criteria

Among biomedical researchers, there has been a considerable variability in the method of testing or evaluating implants. It is very important to standardize the test methodology so as to obtain a meaningful comparison of various results and also to reproduce results with confidence. The problem has been compounded with the introduction of polymer composite biomaterials, which are anisotropic and inhomogenous. Testing methods that have been used to evaluate implants made of homogenous isotropic materials may not work for testing composite material implants. This aspect has been illustrated by Heiner et al. [89] with regards to the testing of metallic and polymer composite femoral stems. Further improvements and standardization of evaluation methods could contribute to the design of better implants.

A major flaw in the majority of the literature dealing with implants made of polymer composite biomaterials is the lack of proper understanding of composite behavior and theories. Many researchers used directly the implant geometry/design originally meant for isotropic materials in producing the polymer composite implants. As the composite materials are distinctly different from the homogenous materials in terms of anisotropy, fracture behavior, and environmental sensitivity, the polymer composite implants must be designed using criteria separate from those, which have been used for isotropic material-based implants. This may even lead to design of superior performance implants. Innovations such as spatially varying fiber volume fraction and/or fiber orientation are leading to new types of functionally graded composite materials. New design criteria need to be developed to harness the potential of this new class of materials and to design implants with improved performance.

5.1.3. Wear debris, and leached or resorbed products

Wear of materials is particularly important for articulating joint applications. Research reports published on the wear characteristics of polymer composites from the viewpoint of biomedical applications are very few. The effect of reinforcement on the wear characteristics of polymers is a controversial subject, and further systematic investigations are necessary to clearly understand the in vivo wear mechanisms of polymer composite materials. Also the long-term systemic effects of polymer composite wear debris are still unclear, and hence, accumulation of clinical data and its careful analysis is needed [171].

In the case of thermoset polymer composites, there are concerns about possible harmful effects of residual monomers, catalysts, and additives that may leach into the tissues. Further efforts are necessary to develop newer thermoset polymers, which are biocompatible.

In the case of resorbable polymers, concerns are expressed over the long-term effects of resorbed products. Efforts are needed to design these materials such that they are removed from the body without side effects.

5.1.4. Improved manufacturing methods, and effect of sterilization

The success of polymer composite biomaterials also relies greatly on the quality of the implant, which is determined by the reproducibility of the fabrication process, sterilization treatment, material storage and handling. Many of the polymer composite biomaterials investigated so far were produced in biomedical research laboratories with limited success. This is because of the trial-and-error approach followed in making the composites without proper understanding and implementation of finer aspects of polymer composite fabrication processes. More over, the composite fabrication methods used for engineering applications have been used directly for producing implants. It is important to acknowledge that the requirements for both the applications are different, and the composite fabrication methods need to be tailored to suit the biomedical applications. For example, for a hip joint replacement application, the composite material surface should be completely covered with a continuous matrix layer in order to prevent a potential release of fiber particle debris during implantation. More over the fabrication method need to be optimized such that it enables desired local and global arrangement of reinforcement phase so as to make the composite implant structurally compatible with the host tissues. Review of existing literature suggests that the various flexibilities of composites in terms of material combinations, fiber/matrix interface control, fiber volume fractions, and fiber and matrix distributions are yet to be fully exploited in fabricating functionally superior implants. Thus far, polymer composite biomaterials are mainly reinforced with

particulates, short fibers and unidirectional fibers, and very few works reported on woven fabric composites. The many advantages offered by textile composite materials have not been exploited in the biomedical field. Efforts should be made to harness the potential of textile composite materials in designing implants with improved performance. It is also important to consider the cost of composite implant. Efforts must be made to develop suitable manufacturing methods for composite implants so as to compete with the current commercial implants.

Like any other material, polymer composite biomaterials are also sterilized prior to implantation. It is known that the polymer properties are sensitive to the sterilization procedure used [181]. For example, the gamma sterilization reportedly causes long-term embrittlement of UHMWPE (used in hip joint cups) due to radiation-induced oxidation. Hence, some effects of sterilization on the mechanical propitioes of polymer composites can also be expected. McKenna et al. [160] investigated the effect of autoclave sterilization on a number of candidate composite materials. They reported that CF/PP composites did not undergo significant degradation even at long autoclave times. The CF/PS composites degraded at even the shortest autoclave times. This study highlights the need for evaluating the degradation resistance of composite materials under sterilization conditions. A suitable sterilization procedure for composite of interest needs to be established through careful experiments.

5.1.5. Surface coatings

As mentioned earlier, the success of an implant also depends on its surface chemistry, which determines the interactions at the implant material–tissue interface. To elicit desirable material–host tissue interactions, the polymer composite implants may need to be coated with suitable coatings. Another important reason for a suitable surface coating is the wear of the implant surface being in contact with the host tissues. For example, the hardness of bone leads to very heavy abrasion by fretting or direct wearing as soon as the interfacial strains between the implant and hard tissue occur. Thus, there is a need for developing suitable coating methods for polymer composite implants. For example, Ha [81] and Ha et al. [80,82] developed a method of coating CF/PEEK composite hip stem surface with bioactive coating. They first vacuum plasma sprayed the composite surface with titanium. Subsequently, the surface is treated with NaOH and immersed in simulated body fluid (SBF), containing ions in concentrations similar to those of human blood plasma. Formation of biocompatible and bioactive calcium phosphate layer similar to hydroxyapatite on the composite surface was reported. To date very limited knowledge is available with regards to surface coating of polymer composite implants, and this warrants further research and development.

5.2. Vital/avital composites

Current trend in biomaterials development is to grow tissues in the laboratory using cells (patient's cells, auto or xenologous cells, human stem cells or genetically engineered cells) of the target tissue (i.e. tissue to replaced or augmented) and porous scaffolds. The combination of polymers (avital or non-living) in the form of foams or fabrics (woven, braids, knits, and non-wovens) and cells (vital or living) results in special type of composite materials, namely vital/avital composites [57]. If the patient's own cells can be used, the vital/avital composites are readily biocompatible and well accepted by the host tissues. Many consider the vital/avital composites are ideal for implant applications. The vital/avital composites are in their infant stage of development, however, it is an area of intensive research worldwide and called by different names including 'tissue engineering' and 'cellular engineering'. Researchers are developing vital/avital composites for a number of applications including vascular grafts [162], tendon/ligament prostheses [13,18,21,275], artificial skin [137], dural substitutes [188], hernia patches, artificial bladder wall, and regenerated cartilage. A wide variety of non-resorbable polymers such as PET, PU, and PTFE, and resorbable polymers such as PGA, PLA, and their blends are used as porous scaffolds. In order to introduce time dependent porosity, some researchers [68,270,271] used bicomponent scaffolds containing both resorbable and non-resorbable polymers. To facilitate the attachment of cells to the avital scaffolds, they have been coated with different systems including pyrolitic carbon [3,212], collagen, albumin, gelatin, and antibiotic drug-releasing gels. It may be noted that the scaffold surfaces are functionalized for a variety of other reasons. Different kinds of cells are seeded onto the porous scaffolds depending on the intended application (target tissue) of the composite material. The cell attachment to the avital scaffolds, and the differentiation and maturation of the ingrown or in situ newly formed tissue depend on a number of variables including pore size and geometry, porosity, pore distribution, nature (two dimensional or three dimensional), inter-connectivity of pores, scaffold thickness, surface topography and biochemical functionalization, types of cells, external stimuli (mechanical, electrical or chemical), etc. Specific details are outside the scope of the present paper. Interested readers may consult the references cited appropriately [83,97,98,120,136,138,139,142,144,154,172–174, 182,199,262,270,271]. The majority of the information reported in the literature on the vital/avital composites is chemistry, biochemistry, and biology related. Little is known about the mechanical characteristics of this new class of polymer composite materials. The vital/avital composite materials require relooking into the traditional composite principles and theories originally developed

for mostly linear and small deformation composite materials. Further work illustrating the principles of deformation behavior of these materials would be very useful to innovatively design new implants, and also would be useful to understand the behavior of natural tissues itself. Ultimately, this knowledge may give insights to unravel the mysteries of many natural tissues.

6. Conclusions

With increased understanding of function and interaction of implants with the human body, it is clear now that for greater success, the implants should be surface compatible as well as structurally compatible with the host tissues. In this respect, the polymer composite biomaterials are particularly attractive because of their tailorable manufacturing processes, and properties comparable to those of the host tissues. Innovations in the composite material design and fabrication processes are raising the possibility of realizing implants with improved performance. However, for successful application, surgeons must be convinced with the long term durability and reliability of polymer composite biomaterials. Monolithic materials have long been used and there is considerable experimental and clinical data supporting their continued usage. Such data with respect to polymer composite biomaterials is relatively small. This requires further research efforts to elucidate the long-term durability of composite biomaterials in the human body conditions.

Appendix A

Apical	Near the apex or extremity of a conical structure, such as the tip of the root of a tooth
Acetabulum	The socket potion of the hip joint
Allograft	Transplanted tissue or organ between unrelated individuals of the same species. Also called 'homograft'
Alveolar bone	The bone structure that supports and surrounds the roots of teeth
Amalgam	An alloy of two or more metals, one of which is mercury
Anastomosis	Interconnection between two blood vessels
Aneurysm	Abnormal dilatation of bulging of a segment of a blood vessel
Ankylosis	Fixation of a joint; in dentistry, the rigid fixation of the tooth to the aveolar bone and ossification of the periodontal membrane
Anterior	Direction referring to the front side of the body
Arthritis	Inflammation of joints
Arthrodesis	Fusion or fixation of a joint
Arthroplasty	Surgical repair of a joint
Articular cartilage	The cartilage at the ends of bones in joints which serve as the articulating, bearing surface.
Artificial organ	A medical device that replaces, in part or in whole, the function of one of the organs of the body.
Atrophy	Wasting away of tissues or organs
Autograft	A transplanted tissue or organ transferred from one part of a body to another part of the same body
Biocompatibility	Acceptance of an implant by surrounding tissues and by the body as a whole. The implant should be compatible with tissues in terms of mechanical, chemical, surface, and pharmacological properties. Simply it is the ability of the implant material to perform with an appropriate host response in a specific application.
Bioglass	Surface-active glass compositions that have been shown to bond to tissue
Biomaterial	The term usually applied to living or processed tissues or to materials used to reproduce the function of living tissue in conjunction with them. Simply it is a material intended to interact with biological systems.
Bone cement	A biomaterial used to secure a firm fixation of joint prostheses, such as hip and knee joints. It is primarily made of polymethyl methacrylate powder and monomer methyl methacrylate liquid
Callus	The hard substance that is formed around a bone fracture during healing. It is usually replaced with compact bone.

Cancellous bone	The reticular or spongy tissue of bone where spicules or trabeculae form the interconnecting latticework that is surrounded by connective tissue or bone morrow
Catheter	An instrument (tube) for gaining access to and draining or sampling fluids in the body
Celestin tube	A nylon reinforced latex tube used to bypass esophgeal tumors
Cochlear implant	A type of surgically implanted hearing aid used to treat sesorineural hearing loss
Collagen	The supporting protein from which the fibers of connective tissues are formed
Compression plate	Bone plate designed to give compression on the fracture site of a broken bone for fast healing.
Condylar prostheses	Artificial knee joints
Congenital	A physical defect existing since birth
Cortical bone	The compact hard bone with osteons
Crown	The part of tooth that is exposed above the gum line or covered with enamel. Largely made of hydroxyapatite mineral.
CT	Computed tomography or computed axial tomography (CAT), an X-ray technique for producing cross-sectional image of the body.
Dacron	Polyethylene terephthalate polyester that is made into fibers, a product of Dupont Co, USA. If the same polymer is made into a film, it is called Mylar.
Dental caries	Tooth decay caused by acid-forming micro-organisms
Dental restoration	Another name for dental fillings
Dentine	The main substance of the tooth, with properties and composition similar to bone.
Dermatitis	Inflammation of skin
Dura mater	The dense, tough connective tissue over the surface of the brain
Elastin	The elastic fibrous mucoprotein in connective tissue
Enamel	A hard, white substance that covers the dentine of the crown of a tooth; enamel is the hardest substance in the body
Endosseous	In the bone, referring to dental implants fixed to the jaw bone
Endosteal	Related to the membrane lining the inside of the bone cavities
Extracorporeal	Outside the body
Femur	The thigh bone, the bone of the upper leg
Fixation devices	Implants used during bone-fracture repair to immobilize the fracture
Fracture plate	Plate used to fix broken bones by open (surgical) reduction. It is fixed to the bone by using screws.
Gingiva	The gum tissue; the dense fibrous tissue overlying the alveolar bone in the mouth and surrounding the necks of teeth
Graft	A transplant
Ground substance	The amorphous polysaccharide material in which cells and fibers are embedded
Hard tissue	The general term for calcified structures in the body, such as bone
Heparin	A substance (mucopolysaccharide acid) found in various body tissues; that prevents the clotting of blood
Herniated disk	A herniated disk is the rupture of the central portion, or nucleus, of the disk through the disk wall and into spinal canal. It is also called slipped disk.
Heterograft	A graft from one species to another. Also called xenograft
Hyaline cartilage	Cartilage with a frosted glassy appearance
Hydrogel	Highly hydrated (over 30% by weight) polymer gel. Acrylamide and poly-HEMA (hydroxyethy-methacrylate) are two common hydrogels.
Hydroxyapatite (HA)	Mineral component of bone and teeth. It is a type of calcium phosphate, with composition $Ca_{10}(PO_4)_6(OH)_2$.
Ilizarov technique	A technique used most often in reconstructive settings to lengthen limbs, transport bone segments, and correct angular deformities
Implant	Any medical device made from one or more materials that is intentionally placed within the body, either totally or partially buried beneath an epithelial surface.

Intervertebral disc	A flat, circular platelike structure of cartilage that serves as a cushion, or shock absorber, between the vertebrae
Intima	Inner lining of a blood vessel
Intramedullary rod or nail	An orthopedic rod or nail inserted into the intramedullary marrow cavity of the bone to promote healing of long bone fractures
Intraosseous implant	An implant inserted into the bone
In vivo condition	Inside the living body
In vitro condition	Simulated in vivo condition in the laboratory
Kirschener wire	Metal surgical wires
Kyphosis	Abnormally increased convexity in the curvature of the lumbar spine
Ligament	A sheet or band of fibrous connective tissue that join bone to bone, offering support to the joint
Long bones	Bones that are longer than they are wide and with distinctive shaped ends, such as femur
Lordosis	Abnormally increased concavity in the curvature of the lumbar spine
LTI carbon	Low-temperature istropic carbon
Lumen	The space within a tubular structure
Mandibular bone	Lower jaw of the mouth
Maxillary bone	Upper jaw of the mouth
Medullary cavity	The marrow cavity inside the long bones
Metastasis	Transfer of disease producing cancer cells or bacteria from an original site of disease to another part of the body with development of a similar lesion in the new location
Myocardium	The muscular tissue of the heart
Necrosis	Death of tissues
NMR	Nuclear magnetic resonance
Nonunion	A bone fracture that does not join
Occlusion	Becoming close together; in dentistry, bringing the teeth together as during biting and chewing
Orthopedics	The medical field concerned with the skeletal system
Orthotics	The science and engineering of making and fitting orthopedic appliances used externally to the body.
Ossicles	The small bones of the middle ear which transmit sound from ear drum to the body
Osteoarthritis	A degenerative joint disease, characterized by softening of the articular ends of bones and thickening of joints, sometimes resulting in partial ankylosis
Osteopenia	Loss of bone mass due to failure of osteoid synthesis
Osteoporosis	The abnormal reduction of the density and increase in porosity of bone due to demineralization, commonly seen in the elderly
Osteotomy	Cutting of bone to correct a deformity
Percutaneous	Transcutaneous, of having to do with passing through the epidermis or skin
Periodontal ligament	Periodontium; the connective tissue (ligament) joining the tooth to the alveolar bone
Polysaccharides	Major constitutents of the ground substance; carbohydrates containing saccharide groups
Posterior	Direction referring to the back side of the body
Proplast	A composite material made of fibrous PTFE and carbon. It is usually porous and has low modulus and low strength.
Prosthesis	A device that replaces a limb, organ or tissues of the body
Proximal	Nearest the trunk or point of origin; opposed to distal
Pyrolitic carbon	Isotropic carbon coated onto a substrate in a fluidized bed
Resorption	Dissolution or removal of a substance
Rheumatoid arthritis	Chronic and progressive inflammation of the connective tissue of joints, leading to deformation and disability
Scoliosis	An abnormal lateral (sideward) curvature of a portion of the spine
Silastic	Medical grade silicone rubber, Dow Corning Corporation
Silica	The ceramic SiO_2
Spondylosis	Any of various degenerative diseases of the spine
Spondylolisthesis	Forward bending of the body at one of the lower vertebrae
Stapes	One of the ossicles of the middle ear

Stenosis	A narrowing or constriction of the diameter of a bodily passage or orifice.
Stress-shield effect	Prolonged reduction of stress on a bone may result in porotic bone (osteoporosis), which may weaken it. This process can be reversed if the natural state of stress can be restored to its original state
Subcutaneous	Beneath the skin
Subperiosteal	Underneath the periosteum
Synovial fluid	The clear viscous fluid that lubricates the surfaces of joints and tendons, secreted by the synovial membrane
Tendon	A band or cord of fibrous tissue connecting muscle to bone
THR	Total hip replacement
Thromboembolism	An obstruction in the vascular (blood circulating) system caused by a dislodged thrombus
Thrombosis	Formation of a thrombus, blood clot
Thromus	A fibrinous blood clot attached at the site of thromsosis
TKR	Total knee replacement
Transplantation	Transfer of a tissue or organ from one body to another, or from one location in a body to another
Trachea	A cylinder-shaped tube lined with rings of cartilage that is 115 mm long, from the larynx to the bronchial tubes; the windpipe
Ureter	The tube that conducts urine from the kidney to the bladder
Urethra	The canal leading from the bladder to the outside for discharging urine
Vascular	Blood vessels
Vitallium	A Co-Cr alloy, Howmedica Inc.
Vitreaous carbon	A term generally applied to isotropic carbon with very small crystallites
Wolff's law	The principle relating the internal structure and architecture of bone to external mechanical stimuli. Remodeling of bone takes place in response to mechanical stimulation so that the new structure becomes suitably adapted to the load.
Xenograft	A transplanted tissue or organ transferred from an individual of another species

References

[1] Adams D, Williams DF, Hill J. Carbon fiber-reinforced carbon as a potential implant material. J Biomedical Materials Research 1978;12:35–42.

[3] Aebischer P, Goddard MB. Tissue reaction to fabrics coated with turbostratic carbon: subcutaneous versus vascular implants. Biomaterials 1988;9(1):80–5.

[4] Akay M, Aslan N. Numerical and experimental stress analysis of a polymeric composite hip joint prostheses. J Biomedical Materials Research 1996;31:167–82.

[5] Ali MS, Hastings GW, Rae T, Rushton N, Ross ERS, Wynn-Jones CH. Carbon fiber composite bone plates. J Bone and Joint Surgery 1990;72-B(4):586–91.

[6] Ambrosio L, Carotenuto, Nicolais L. Composite materials. In: Black J, Hastings G, editors. Handbook of Biomaterial Properties, Editors. London, UK: Chapman and Hall, 1998. p. 214–69.

[7] Ambrosio L, De Santis R, Iannace S, Netti PA, Nicolais L. Viscoelastic behavior of composite ligament prostheses. J Biomedical Materials Research 1998;42(1):6–12.

[8] Ambrosio L, Netti PA, Iannace S, Huang SJ, Nicolais L. Composite hydrogels for intervertebral disc prostheses. J Materials Science: Materials in Medicine 1996;7:251–4.

[9] Amstutz HC, et al. Mechanism and clinical significance of wear debris-induced osteolysis. Clinical Orthopedics and Related Research 1992;276:7–18.

[10] Andersson GBJ. Intervertebral disk. In: Wright V, Radin EL, editors. Mechanics of human joints: physiology, pathophysiology, and treatment. New York: Marcel Dekker Inc, 1993. p. 293–311.

[11] Andreopoulos AG, Evangelatou M, Tarantili PA. Properties of maxillofacial silicone elastomers reinforced with silica powder. J Biomaterials Applications 1998;13:66–73.

[13] Arnoczky SP, Torzilli PA, et al. Biologic fixation of ligament prostheses and augmentations. An evaluation of bone in growth in the dog. Am J Sports Med 1988;16(2):106–12.

[14] Baidya KP, Ramakrishna, S, Rahman M, Ritchie A. Quantitative radiographic analysis of fiber reinforced polymer composites. J Biomaterials Applications 2001;15(3):279–89.

[15] Baidya KP, Ramakrishna S, Rahman M, Ritchie A. Advanced textile composite ring for ilizarov external fixator system. J Engineering in Medicine, IME UK 2001;215(H1):11–23.

[16] Baidya KP, Ramakrishna S, Ritchie A, Rahman M. Design and evaluation of composite external fracture fixation device. Proc. ICCM-12, Paris, France, p. 496.

[17] Bao QB, McCullen GM, Higham PA, Dumbleton JH, Yuan HA. The artificial disc: theory, design and materials. Biomaterials 1996;17:1157–67.

[18] Berry JL, Berg WS, et al. Evaluation of Dacron-covered and plain bovine xenografts as replacements for the anterior cruciate ligament. Clin Orthop 1988;236:270–8.

[19] Bjork N, Ekstrand K, Ruyter IE. Implant-fixed, dental bridges from carbon/graphite fiber reinforced poly(methyl methacrylate). Biomaterials 1986;7:73–5.

[20] Black J. Orthopaedic Biomaterials in Research and Practice. New York: Churchill Livingstone, 1988 (p. 197–210).

[21] Black J. Biological Performance of Materials: Fundamentals of Biocompatibility. New York: Marcel Dekker, 1992.

[22] Black J, Hastings GW. Handbook of Biomaterials Properties. London, UK: Chapman and Hall, 1998.

[23] Blazewicz S, Chlopek J, Litak A, Wajler C, Staszkow E. Experimental study of mechanical properties of composite carbon screws. Biomaterials 1997;18:437–9.

[24] Blumenthal NC, et al. A new technique for quantitation of metal particulates and metal reaction products in tissues near implants. J Applied Biomaterials 1994;5:191–3.

[25] Bonfield W. Composites for bone replacement. J Biomedical Engineering 1988;10:522–6.
[26] Bonfield W, Grynpas MD, Tully AE, Bowman J, Abram J. Hydroxyapatite reinforced polyethylene — a mechanically compatible implant material for bone replacement. Biomaterials 1981;2:185–6.
[27] Bostman O, Hirvensalo E, Makinen J, Rokkanen P. Foreign-body reactions to fracture fixation implants of biodegradable synthetic polymers. J Bone and Joint Surgery 1990;72-B: 592–6.
[28] Bradley JS. Cautionary note on the strength of materials for internal fixation devices. Biomaterials 1981;2:124.
[29] Bradley JS, Hastings GW. Carbon fiber reinforced plastics for orthopedic implants Mechanical properties of biomaterials. Hastings GW, Williams DF, editors. 379–386
[30] Bradley JS, Hastings GW, Johnson-Nurse C. Carbon fiber reinforced epoxy as a high strength, low modulus material for internal fixation plates. Biomaterials 1980;1:38–40.
[31] Bragdon CR, et al. Defects in the cement mantle: a fatal flaw in cemented femoral stems for THR. San Francisco: Society for Biomaterials, 1995.
[32] Brantigan JW, Steffee AD, Geiger JM. A carbon fiber implant to aid interbody lumbar fusion Mechanical testing. Spine 1991;16(6S):S277–0S282.
[33] Bridwell KH, DeWald RL, Lubicky JP, Spencer DL, Hammerberg KW, Benson DR, Neuwirth MG. The Textbook of Spinal Surgery. Philadelphia, (PA): J.B. Lippincott Company, 1991.
[34] Brown SA, Hastings RS, Mason JJ, Moet A. Characterization of short-fiber reinforced thermoplastics for fracture fixation devices. Biomaterials 1990;11:541–7.
[35] Bruckmann H, Huttinger KJ. Carbon, a promising material in endoprosthetics. Part 1: the carbon materials and their mechanical properties. Biomaterials 1980;1:67–72.
[36] Bruckmann H, Keuscher G, Huttinger KJ. Carbon, a promising material in endoprosthetics. Part 2: tribological properties. Biomaterials 1980;1:73–81.
[37] Buchert PK, et al. Excessive Metal Relase due to Loosening and Fretting of Sintered Particles on Porous-Coated Hip Prostheses. J Bone and Joint Surgery 1986;68-A:606–9.
[38] Cailleteau JG, Rieger MR, Ed Akin J. A comparison of intracanal stresses in a post-restored tooth utilizing the finite element method. J Endodontics 1992;18(11):540–4.
[39] Chalian VA, Phillips RW. Materials in maxillofacial prosthtics. J Biomedical Materials Research Symposium 1974;5(2):349–63.
[40] Chang FK, Perez JL, Davidson JA. Stiffness and strength tailoring of a hip prostheses made of advanced composite materials. J Biomedical Materials Research 1990;24:873–99.
[41] Chao EYS, Aro HT. Biomechanics of fracture fixation. In: Mow VC, Hayes WC, editors. Basic Orthopaedic Biomechanics. Philadelphia: Lippincott-Raven, 1997. p. 317–50.
[42] Choueka J, Charvet JL, Alexander H, Oh YH, Joseph G, Blumenthal NC, LaCourse WC. Effect of annealing temperature on the degradation of reinforcing fibers for absorbable implants. J Biomedical Materials Research 1995;29:1309–15.
[43] Christel PS, Leray JL, Sedel L, Morel E. (1980). Mechanical evaluation and tissue compatibility of materials for composite bone plates. In: Mechanical Properties of Biomaterials. Hastings GW, Williams DF, editors. p. 367–377.
[44] Christel P, Claes L, Brown SA. Carbon reinforced composites in orthopedic surgery. In: Szycher M, editor. High Performance Biomaterials: A Comprehensive Guide to Medical and Pharmaceutical Applications. Lancaster, (USA): Technomic, 1991. p. 499–518.
[45] Christel P, Meunier A, Leclercq S, Bouquet Ph, Buttazzoni B. Development of carbon-carbon hip prostheses. J Biomedical Materials Research: Applied Biomaterials 1987;21(A2):191–218.
[46] Ciappetta P, Boriani S, Fava GP. A carbon fiber reinforced polymer cage for vertebral body replacement: a technical note. Neurosurgery 1997;41(5):1203–6.
[47] Claes LE. Mechanical characterization of biodegradable implants. Clinical Materials 1992;10:41–6.
[48] Claes L, Hutter W, Weiss R. Mechanical properties of carbon fiber reinforced polysulfone plates for internal fixation. In: Christel P, Meunier A, Lee AJC, editors. Biological and Biomechanical Performance of Biomaterials. Amsterdam, The Netherlands: Elsevier Science Publishers, 1997. p. 81–6.
[50] Claes L, Schultheiss M, Wolf S, Wilke HJ, Arand M, Kinzl L. A new radiolucent system for vertebral body replacement: its stability in comparison to other systems. J Biomedical Materials Research, Applied Biomaterials 1999;48(1):82–9.
[51] Cogswell FN. Thermoplastic Aromatic Polymer Composites. UK: Butterworth Heinemann, 1992.
[52] Coombes AGA, Greenwood CD, Shorter JJ. Plastic materials for external prostheses and orthoses. In: Wise DL, Trantolo DJ, Altobelli DE, Yaszemski MJ, Gresser JD, editors. Human Biomaterials Applications. (NJ): Humana PressTotowa, 1996. p. 215–55.
[53] Culleton TP, et al. Fatigue failure in the cement mantle of an artificial hip joint. Clinical Materials 1993;12:95–102.
[54] Darendeliler S, Darendeliler H, Kinoglu T. Analysis of a central maxillary incisor by using a three-dimensional finite element method. J Oral Rehabilitation 1992;19:371–83.
[55] Dauner M, Caramaro L, Missirlis Y, Panagiotopoulos E. Resorbable continuous-fiber reinforced polymers for osteosynthesis. J Materials Science: Materials in Medicine 1998; 9:173–9.
[56] Davy KWM, Parker S, Braden M, Ward IM, Ladizesky H. Reinforcement of polymers of 2,2 bis-4(2-hydroxy-3-methacryloyloxy propoxy) phenyl propane by ultra-high modulus polyethylene fibers. Biomaterials 1998;13(1):17–19.
[57] de Haan J, Peijs T, Mayer J, Wintermantel E. Modeling of the deformation behavior of flexible knitted composites. In: Proc. ECCM-7, London: Woodhead Publishing Limited, 1996, p. 421–7.
[58] Deb S, Wang M, Tanner KE, Bonfield W. Hydroxyapatite-polyethylene composites: effect of grafting and surface treatment of hydroxyapatite. J Materials Science: Materials in Medicine 1996;7:191–3.
[59] Dehl RE. On the characterization of porosity in PTFE-carbon composite implant materials by mercury porosimetry. J Biomedical Materials Research 1982;16:715–9.
[60] Deluca PA, Lindsey RW, Ruwe PA. Refracture of bones of the forearm after the removal of compression plates. J Bone and Joint Surgery 1988;70-A(9):1372–6.
[61] Deng M, Shalaby SW. Properties of self-reinforced ultra-high-molecular weight polyethylene composites. Biomaterials 1997;18:645–55.
[62] Durselen L, Claes L, Ignatius A, Rubenacker S. Comparative animal study of three ligament prostheses for the replacement of the anterior cruciate and medial collateral ligament. Biomaterials 1996;17:977–82.
[63] Esslinger JO, Rutkowski EJ. Studies on the skeletal attachment of experimental hip prostheses in the pygmy goat and the dog. J Biomedical Materials Research Symposium 1973;4: 187–93.
[64] Feith R. Side effects of acrylic cement, implanted to bone. Acta Orthop Scand 1975; (Suppl.):161.
[65] Flahiff CM, Blackwell AS, Hollis JM, Feldman DS. Analysis of a biodegradable composite for bone healing. J Biomedical Materials Research 1996;32:419–24.
[66] Forster IW, Ralis ZA, Jenkins DH. Filamentous carbon fiber induction of new tendon: tissue reactions and environmental conditions. In: Winter GD, Leray JL, de Groot K, editors.

Evaluation of Biomaterials. New York: John Wiley & Sons Ltd, 1980. p. 367–71.
[67] Freilich MA, Meiers JC, Duncan JP, Jon Goldberg A. Fiber-Reinforced Composites in Clinical Dentistry. IL: Qintessence Publishing, 2000.
[68] Galletti PM, Aebischer Petal. Experience with fully bioresorbable aortic grafts in the dog. Surgery 1988;103(2):231–41.
[69] Ganesh VK, Ramakrishna S. Prothesis: another concept. The European periodical for technical textile users. Quarterly Magazine 1998;29(3):56–60.
[70] Ganesh VK, Ramakrishna S, Leck HJ. Fiber reinforced composite based functionally gradient materials. J Advanced Composite Letters 1998;7:111–5.
[71] Gerhart TN, Miller RL, Kleshinski SJ, Hayes WC. In vitro characterization and biomechanical optimization of a biodegradable particulate composite bone cement. J Biomedical Materials Research 1988;22:1071–82.
[72] Gershon B, Cohn D, Marom G. Utilization of composite laminate theory in the design of synthetic soft tissues for biomedical prostheses. Biomaterials 1990;11:548–52.
[73] Gershon B, Cohn D, Marom G. Compliance and ultimate strength of composite arterial prostheses. Biomaterials 1992;13:38–43.
[74] Gese H, et al. Relativbewegungen und stress-shielding bei zementfreien Hüftendoprothesen — eine Analyse mit der Methode der Finiten Elemente. In: Die zementlose Hüftprothese. Gräfelfing, Demeter Verlag: p. 75–80.
[75] Giardino R, Glannini S, Fini M, Giavaresi G, Martini L, Orienti L, Experimental in vivo model to evaluate resorbable implants into bone. In: Kim HL, Chung LP editors. Biodegradable Implants in Fracture Fixation. 1993. p. 143–51.
[76] Gilbert JL, Ney DS, Lautenschlager EP. Self-reinforced composite poly(methyl methacrylate): static and fatigue properties. Biomaterials 1995;16(14):1043–55.
[77] Gillett N, Brown SA, Dumbleton JH, Pool RP. The use of short carbon fiber reinforced thermoplastic plates for fracture fixation. Biomaterials 1986;6:113–21.
[78] Goh JCH, Tan PH, Toh SL, Tay TE. Gait analysis study of an energy-storing prosthetic foot-a preliminary study. Gait and Posture 1994;2(2):95–101.
[79] Goldner JL, Urbaniak JR. The clinical experience with silicone-Dacron metacarpophalangeal and interphalangeal joint prostheses. J Biomedical Materials Research Symposium 1973;4:137–63.
[80] Ha SW, Eckert KL, Wintermantel E, Gruner H, Guecheva M, Vonmont H. NaOH treatment of vacuum-plasma-sprayed titanium on carbon fiber-reinforced poly(etheretherketone). J Materials Science: Materials in Medicine 1997;8:881–6.
[81] Ha S-W. Bioactive calcium phosphate coatings on poly (etheretherketone) and its composites, PhD. thesis, ETH, Zurich, Switzerland, 1997.
[82] Ha S-W, Mayer J, Koch B, Wintermantel E. Plasma-sprayed hydroxylapatite coating on carbon fiber reinforced thermoplastic composite materials. J Materials Science: Materials in Medicine 1994;5:481–4.
[83] Hake U, Gabbert H, Iversen S, Jakob H, Schmiedt W, Oelert H. Evaluation of the healing of precoated vascular dacron prostheses. Langenbecks Arch Chir 1991;376(6):323–9.
[84] Halstead A, Jones CW, Rawlings RD. A study of the reaction of human tissue to proplast. J Biomedical Materials Research 1979;13:121–34.
[85] Harris B. The mechanical behavior of composite materials. The Mechanical Properties of Biological Materials. Cambridge, UK: Cambridge University Press, 1980 (p. 37–74).
[86] Hastings GW. Is there an ideal biomaterial for use as an implant for fracture fixation? In Hastings GW editor. Biodegradable Implants in Fracture Fixation, 1993. p. 19–34.
[88] Hastings GW. Biomedical applications of CFRPs. In: Fitzer E, editor. Carbon Fibre and their Composites, Springer-Verlag, Berlin, p. 261–271.
[89] Heiner AD, Brown SA, Davy DT. Effects of mechanical testing device variables on polymer composite femoral stem strains. Biomaterials 1996;17(23):2211–7.
[90] Hench LL. Bioceramics: from concept to clinic. J American Ceramic Society 1991;74:1487–510.
[91] Hench LL. Inorganic biomaterials Materials Chemistry: An Emerging Discipline. In: Interrante LV, Caspar LA, Ellis AB, editors. Advances in Chemistry Series 245. Washington, (DC): American Chemical Society, 1995. p. 523–47.
[92] Hench LL, Ethridge EC. Biomaterials An Interfacial Approach. New York: Academic Press, 1982.
[93] Henderson JD, Mullarky RH, Ryan DE. Tissue biocompatibility of kevlar aramid fibers and polymethylmethacrylate, composites in rabbits. J Biomedical Materials Research 1987;21:59–64.
[95] Hidaka S, Gustilo RB. Refracture of bones of the forearm after plate removal. J Bone and Joint Surgery 1984;64-A(8):1241–3.
[96] Higashi S, Yamamuro T, Nakamura T, Ikada Y, Hyon SH, Jamshidi K. Polymer-hydroxyapatite composites for biodegradable bone fillers. Biomaterials 1986;7:183–7.
[97] Hirt SW, Aoki M, Demertzis S, Siclari F, Haverich A, Borst HG. Comparative in vivo study on the healing qualities of four different presealed vascular prostheses. J Vasc Surg 1993;17(3):538–45.
[98] Hirt SW, Dosis D, Siclari F, Rohde R, Haverich A. Collagen-presealed or uncoated aortic bifurcation Dacron prostheses: a 5-year clinical follow-up study. Thorac Cardiovasc Surg 1991;39(6):365–70.
[99] Ho MH, Lee SY, Chen HH, Lee MC. Three-dimensional finite element analysis of the effects of posts on stress distribution in dentin. The J of Prosthhetic Dentistry 1994;72:367–72.
[100] Hoag EP, Dwyer TG. A comparative evaluation of three post and core techniques. The Journal of Prosthetic Dentistry 1982;47:177–81.
[102] Huang ZM, Ramakrishna S. Development of knitted fabric reinforced composite material for prosthetic application. Advanced Composites Letters 1999;8(6):289–94.
[103] Huiskes R. Some fundamental aspects of human-joint replacement. Acta Orthop Scand 1980; (Suppl.), 185.
[104] Huiskes R, et al. The relationship between stress shielding and bone resorption around total hip stems and the effects of flexible materials. Clinical Orthopaedics and Related Research 1992; 274:124–34.
[105] Hunt MS. Development of carbon fiber/polysulfone orthopaedic implants. Materials and Design 1987;8(2):113–9.
[107] Huttner W, Keuscher G, Nietert M. Carbon fiber reinforced polysulfone thermoplastic composites. In: Ducheyne P, Van der Perre G, Aubert AE, editors. Biomaterials and Biomechanics. Amsterdam, The Netherlands: Elsevier Science Publishers, 1984. p. 167–72.
[108] Iannace S, Sabatini G, Ambrosio L, Nicolais L. Mechanical behavior of composite artificial tendons and ligaments. Biomaterials 1995;16(9):675–80.
[109] Ignatius A, Unterricker K, Wenger K, Richter M, Claes L. A new composite made of polyurethane and glass ceramic in a loaded implant model: a biomechanical and histological analysis. J Materials Science: Materials in Medicine 1997;8:753–6.
[110] Imai T, Watari F, Yamagata S, Kobayashi M, Nagayama K, Nakamura S. Effects of water immersion on mechanical properties of new esthetic orthodontic wire. Americal Journal of Orthodontics and Dentofacial Orthopedics 1999:533–538.
[111] Inoue N, Hirasawa Y, Hirai T, Katayama T. Composite materials in bio-medical engineering. Materiaux and Techniques 1994;4-5:23–6.

[112] Issidor F, Brendum K. Intermittent loading of teeth with tapered individually cast or prefabricated parallel-sided posts. Int J Prosthodontics 1992;5:257–61.
[113] Issidor F, Odman P, Brondum K. Intermitttent loading of teeth restored using prefabricated carbon fiber posts. Int J Prosthodontics 1996;9(2):131–6.
[114] James SP, Jasty M, Davies J, Piehler H, Harris WH. A fractographic investigation of PMMA bone cement focusing on the relationship between porosity reduction and increased fatigue life. J Biomedicl Materials Research 1992;26:651–62.
[115] Jancar J, Dibenedetto AT. Fiber reinforced thermoplastic composites for dentistry, Part I hydrolytic stability of the interface. J Materials Science: Materials in Medicine 1993;4:555–61.
[116] Jancar J, Dibenedetto AT, Goldberg AJ. Thermoplastic fiber-reinforced composites for dentistry Part II Effect of moisture on flexural properties of unidirectional composites. J Materials Science: Materials in Medicine 1993;4:562–8.
[117] Jenkins DHR. Experimental and clinical application of carbon fiber as an implant in orthopedics. J Bone and Joint Surgery 1977;59-B(4):501.
[118] Jockisch KA, Brown SA, Bauer TW, Merritt K. Biological response to chopped-carbon-fiber-reinforced peek. J Biomedical Materials Research 1992;26:133–46.
[119] Kahl S, et al. Einfluss von Zementierungsfehlern auf die Knochenzement-Beanspruchung. Biomedizinische Technik 1993;38:298–302.
[120] Kajiwara H. A new cardiac wall substitute made of Dacron fabric prosthesis sealed with autologous tissue fragments: comparative studies of neointima formation and bleeding resistant property against fibrinolysis phenomenon with those of a preclotted sealing graft. Artif Organs 1995;19(1):64–71.
[121] Kennedy KC, Chen T, Kusy RP. Behavior of photopolymerized silicate-glass-fiber-reinforced dimethacrylate composites subjected to hydrothermal ageing. J Materials Science: Materials in Medicine 1998;9:243–8.
[122] Kettunen J, Makela A, Miettinen H, Nevalainen T, Heikkila M, Tormala P, Rokkanen P. Fixation of femoral shaft osteotomy with an intramedullary composite rod: An experimental study on dogs with a two-year follow-up. J Biomaterials Science Polymer Edition 1999;10(1):33–45.
[123] King MW, Poddevin N, Guidoin R, Marois Y, et al. Designing textile structures to repair and replace knee ligaments. Canadian Textile Journal 1996;113(3):53–7.
[124] Klein CPAT, van der Lubbe HBM, de Groot K. A plastic composite of alginate with calcium phosphate granules as implant material: an in vivo study. Biomaterials 1987;8:308–10.
[125] Klein N, Carciente ML, Cohn D, Marom G, Uretzky G, Peleg H. Filament-wound composite soft tissue prostheses: controlling compliance and strength by water absorption and degradation. J Materials Science: Materials in Medicine 1993;4:285–91.
[126] Knowles JC, Hastings GW. In vitro and in vivo investigation of a range of phosphate glass-reinforced polyhydroxybutyrate-based degradable composites. J Materials Science: Materials in Medicine 1993;4:102–6.
[128] Knowles JC, Hastings GW, Ohta H, Niwa S, Boeree N. Development of a degradable composite for orthopedic use: *in vivo* biomechanical and histological evaluation of two bioactive degradable composites based on the polyhydroxybutyrate polymer. Biomaterials 1992;13(8):491–6.
[129] Ko FK. Fiber reinforced composites in medical applications. In: Lectures on textile structural composites. Taipei, Taiwan: China Textile Institute 1999. p. 3B3-3B6.
[130] Kocvara S, Kliment CH, Kubat J, Stol M, Ott Z, Dvorak J. Gel-fabric prostheses of the ureter. J Biomedical Materials Research 1967;1:325–36.
[131] Krause W, Mathis RS. Fatigue properties of acrylic bone cements: Review of the literature. J Biomedical Materials Research: Applied Biomaterials 1988;22(A1):37–53.
[132] Krause WR, Park SH, Straup RA. Mechanical properties of BIS-GMA resin short glass fiber composites. J Biomedical Materials Research 1989;23:1195–211.
[133] Krebs J, Friedrich K, Heuwinkel R, Rieger P, Kosack P. Fiber-reinforced thermoplastic composite osteosynthesis implants with variable mechanical properties for better treatment of bone fractures. Proc. ICCM-11, 1997. p. 509–517.
[134] Kuiper JH, Huiskes R. Mathematical optimization of elastic properties: application to cementless hip stem design. J Biomechanical Engineering 1997;119:166–74.
[135] Kwarteng KB, Stark C. Carbon fiber reinforced PEEK (APC-2/AS4) composites for orthopedics implants. SAMPE Quarterly 1990;22(1):10–14.
[136] Lacroix H, Boel K, Nevelsteen A, Suy R. Early inflammatory response to gelatin- and collagen-sealed Dacron prostheses. AnnVascSurg 1995;9(2):152–4.
[137] Lakes R. Composite biomaterials. In: Bronzino JD, editor. The Biomedical Engineering Handbook. Bota Racon, Florida, USA: CRC Press, 1995. p. 598–610.
[138] Langer R, Vacanti J. Tissue engineering. Science 1993;260:920–6.
[139] Langer R, Vacanti J. Artificial organs. Scientific American 1995; September 100–3.
[140] Latour RA, Black J. Development of FRP composite structural biomaterials: Ultimate strength of the fiber/matrix interfacial bond in *in vivo* simulated environments. J Biomedical Materials Research 1992;26:593–606.
[141] Latour Jr. RA, Black J. Development of FRP composite structural biomaterials: fatigue strength of the fiber/matrix interfacial bond in simulated in *vivo* environments. J Biomedical Materials Research 1993;27:1281–91.
[142] Laurencin CT, Norman ME, Elgendy HM, el-Almin SF, Allcock HR, Pucher SR, Ambrosio AA. Use of polyphosphazenes for skeletal tissue regeneration. J Biomed Mater Res 1993;27:963–73.
[143] Lee IY, et al. Effects of variation of prosthesis size on cement stress at the tip of a femoral implant. J Biomedical Materials Research 1994;28:1055–60.
[144] Leong KW, Ramakrishna S. Scaffolds engineering. Annual Reviews of Biomedical Engineering (in press).
[145] Lin TW, Corvelli AA, Frondoza CG, Roberts JC, Hungerford DS. Glass peek composite promotes proliferation and osteocalcin production of human osteoblastic cells. J Biomedical Materials Research 1997;36(2):37–144.
[146] Liu Q, de Wijn JR, van Blitterwijk CA. Composite biomaterials with chemical bonding between hydroxyapatite filler particles and PEG/PBT copolymer matrix. J Biomedical Materials Research 1998;40(3):490–7.
[147] Loher U, Tognini R, Mayer J, Koch B, Wintermantel E. Development of a cortical bone screw made with endless carbon fiber reinforced Polyetheretherketone (CF-PEEK) by extrusion. A new method. In: 7th international conference on polymers in medicine and surgery, Amsterdam, The Netherlands, The Institute of Materials, 1993. pp 88–97.
[148] Louis JP, Dabadie M. Fibrous carbon implants for the maintenance of bone volume after tooth avulsion: first clinical results. Biomaterials 1990;11:525–8.
[149] Lovdahl PE, Nicholls JI. Pin-retained amalgam cores Vs cast-gold dowel-cores. J Prosthetic Dentistry 1977;38(5):507–14.
[150] Majola A, Vainionpaa S, Vihtonen K, Mero M, Vasenius J, Tormala P, Rokkanen P. Absorption, biocompatibility, and fixation properties of polylactic acid in bone tissue: an experimental study in rats. Clinical Orthopedics 1991;268:260.
[151] Marcolongo M, Ducheyne P, Garino J, Schepers E. Bioactive glass fiber/polymeric composites bond to bone tissue. J Biomedical Materials Research 1998;39(1):161–70.

[152] Marois Y, Roy R, et al. Histopathological and immunological investigations of synthetic fibres and structures used in three prosthetic anterior cruciate ligaments: in vivo study in the rat. Biomaterials 1993;14(4):255–62.
[154] Matsumoto A, Noishiki Y, Ichikawa Y, Soma T, Kondo J, Kosuge T. Sealing of a fabric vascular prosthesis with autologous adipose tissue: a preliminary report of its clinical application. Artificial Organs 1995;19(1):51–6.
[155] Mayer J. Gestricke aus Kohlenstoffasern fur biokompatible Verbundwerkstoffe, dargestellt an einer homoelastischen Osteosyntheseplatte. PhD thesis, ETH Zurich, Switzerland, 1994.
[156] Mayer J, Wintermantel E. Thermoforming process for knitted fabric reinforced thermoplastics. New manufacturing techniques for load bearing anisotropic implants. In: Bhattacharyya D, editor. Composite Sheet Forming. Amsterdam: Elsevier Science, 1997. p. 403–40.
[157] Mayer J, Ruffieux K, Koch B, Wintermantel E, Schulten T, Hatebur A. The double die technique (DDT): biomaterials processing for adaptable high fatigue resistance thermoplastic carbon fiber osteosynthesis plates. J Biomedical Engineering, Application, Basis, Communication 1993;5:778–83.
[158] McGrath MH. Biomaterials in plastic surgery. In: Geogiade GS, Georgiade NG, Riefkohl R, Barwick WJ, editors. Textbook of Plastic, Maxillofacial, and Reconstructive Surgery. Baltimore, USA: Williams and Wilkins, 1992. p. 93–101.
[159] McKenna GB, Bradley GW, Dunn HK, Statton WO. Mechanical properties of some fiber reinforced polymer composites after implantation as fracture fixation plates. Biomaterials 1980;1:189–92.
[160] McKenna GB, Bradley GW, Dunn HK, Statton WO. Degradation resistance of some candidate composite biomaterials. J Biomedical Materials Research 1979;13:783–98.
[161] Mendes DG, Iusim M, Angel D, Rotem A, Mordehovich D, Roffman M, Lieberson S, Boss J. Ligament and tendon substitution with composite carbon fiber strands. J Biomedical Materials Research 1986;20:699–708.
[162] Menger MD, Hammersen Fetal. In vivo assessment of neovascularization and incorporation of prosthetic vascular biografts. Thorac Cardiovasc Surg 1992;40(1):19–25.
[163] Messner K, Gillquist J. Prosthetic replacement of the rabbit medical menicus. J Biomedical Materials Research 1993;27: 1165–1173.
[164] Miettinen VM, Vallittu PK. Release of residual methyl methacrylate into water from glass fiber-poly (methyl methacrylate) composite used in dentures. Biomaterials 1997;18(2): 181–5.
[165] Mjöberg B. Fixation and loosening of hip prostheses. Acta Orthopedica Scandinavia 1991;62:500–8.
[166] More N, Baquey C, Barthe X, Rouais F, Rivel J, Trinquecoste M, Marchand A. Biocompatibility of carbon-carbon materials: in vivo study of their erosion using ^{14}carbon labeled samples. Biomaterials 1988;9:328–34.
[167] Morrison C, Macnair R, MacDonald C, Wykman A, Goldie I, Grant MH. In vitro biocompatibility testing of polymers for orthopedic implants using cultured fibroblasts and osteoblasts. Biomaterials 1995;16(13):987–92.
[168] Moyen BJ-L, Lahey PJ, Weinberg EH, Harris WH. Effects on intact femora of dogs of the application and removal of metal plates. J Bone and Joint Surgery 1978;60A(7):940–7.
[170] Nazre A, Lin S. Theoretical strength comparison of bioabsorbable (PLLA) plates and conventional stainless steel and titanium plates used in internal fracture fixation. In: Harvey Jr. JP, Games F. editors. Clinical and Laboratory Performance of Bone Plates. 1994. p. 53–64.
[171] Neugebauer R, Helbing G, Wolter D, Mohr W, Gistinger G. The body reaction to carbon fiber particles implanted into the medullary space of rabbits. Biomaterials 1981;2:182–4.
[172] Noishiki Y, Yamane Y, Tomizawa Y. Sealing of highly porous fabric vascular prostheses by adipose connective tissue fragments instead of preclotting with fresh blood. J Invest Surg 1993;6(3):231–40.
[173] Noishiki Y, Yamane Y, Tomizawa Y, Okoshi T, Satoh S, Takahashi K, Yamamoto K, Ichikawa Y, Imoto K, Tobe M. Rapid neointima formation with elastic laminae similar to the natural arterial wall on an adipose tissue fragmented vascular prosthesis. Asaio J 1994;40(3):M267–0M272.
[174] Noishiki Y, Yamane Y, Tomizawa Y, Okoshi T, Satoh S, Wildevuur CR, Suzuki K. Rapid endothelialization of vascular prostheses by seeding autologous venous tissue fragments. J Thorac Cardiovasc Surg 1992;104(3):770–8.
[175] Okazaki M, Ohmae H. Mechanical and biological properties of apatite composite resins. Biomaterials 1988;9:345–9.
[176] Orthopedic Composites (1991). Stuart M Lee, editor. International Encyclopedia of Composites. VCH Publishers, New York, 1991. Vol. 4, p. 74–87.
[177] Paganetto G, Mazzullo S, De Lolliis A, Buscaroli S, Rocca M, Fini M, Giardino R. Poly-L-Lactid acid: biointeraction and processing variable relationship. Biomaterials 1991;5(2):179.
[178] Park JB. Biomaterials Science and Engineering. New York: Plenum Press, 1984.
[179] Park JB, Lakes RS. Biomaterials An Introduction. New York: Plenum Press, 1992 (p. 169-83).
[180] Parsons JR, Alexander A, Weiss AB. Absorbable polymer-filamentous carbon composites: a new concept in orthopedic biomaterials. In: Szycher M, editor. Biocompatible Polymers, Metals, and Composites. Lancaster: Technomic Publishing Co. Inc, 1983. p. 873–905.
[181] Pascaud RS, Evans WT, McCullagh PJJ, FitzPatrick DP. Influence of gamma-irradiation sterilization and temperature on the fracture toughness of ultra-high-molecular-weight polyethylene. Biomaterials 1997;18:727–35.
[182] Patel M, Arnell RE, Sauvage LR, Wu HD, Shi Q, Wechezak AR, Mungin D, Walker M. Experimental evaluation of ten clinically used arterial prostheses. Ann Vasc Surg 1992;6(3):244–51.
[183] Pedro Kato Y, Christiansen DL, Hahn RA, Shieh SJ, Goldstein JD, Silver FH. Mechanical properties of collagen fibers: a comparison of reconstituted and rat tail tendon fibers. Biomaterials 1989;10:38–42.
[184] Peluso G, Ambrosio L, Cinquegrani M, Nicolis L, Saiello S, Tajana G. Rat peritoneal immune response to carbon fiber reinforced epoxy composite implants. Biomaterials 1991;12:231–5.
[185] Peter T, Tognini R, Mayer J, Wintermantel E. Homoelastic, anisotropic osteosynthesis system by net-shape processing of endless carbon fiber reinforced polyetheretherketone (PEEK). In: Goh JCH, Nather A editors. Proc. of 9th Conference on Biomedical Engineering, Singapore, National University of Singapore, 1997. p. 317–9.
[186] Peterson CJ, Donachy JH, Kalenak A. A segmented polyurethane composite prosthetic anterior cruciate ligament *in vivo* study. J Biomedical Materials Research 1985;19:589–94.
[187] Philips D. Characterization and development of 3D-knitted composites. PhD thesis, Katholieke University Leuven, Belgium, 1999.
[188] Pietrucha K. New collagen implant as dural substitute. Biomaterials 1991;12(3):320–3.
[189] Pilliar RM, Blackwell R, Macnab I, Cameron HU. Carbon fiber-reinforced bone cement in orthopedic surgery. J Biomedical Materials Research 1976;10:893–906.
[190] Pongor P, Betts J, Muckle DS, Bentley G. Woven carbon surface replacement in the knee: independent clinical review. Biomaterials 1992;13(15):1070–6.
[191] Poss R, et al. Femoral expansion in the presence of total hip arthroplasty. J Arthroplasty 1987;2:259–64.

[192] Pourdeyhimi B, Wagner HD. Elastic and ultimate properties of acrylic bone cement reinforced with ultra-high-molecular-weight polyethylene fibers. J Biomedical Materials Research 1989;23:63–80.
[193] Pradas MN, Calleja RD. Reproduction in a polymer composite of some mechanical features of tendons and ligaments. In: Szycher M, editor. High Performance Biomaterials: A Comprehensive Guide to Medical and Pharmaceutical Applications. Lancaster, USA: Technomic Publishing, 1991. p. 519–23.
[195] Profio AE. Biomedical Engineering. New York: John Wiley & Sons, 1993.
[196] Purton DG, Payne JA. Comparison of carbon fiber and stainless steel root canal posts. Quintessence International 1996;27(2):93–7.
[197] Ramakrishna S. Developoment of flexible knitted fabric reinforced composites for soft tissue replacements. The Japanese Journal of Artificial Organs 1997;26(4):S–61.
[198] Ramakrishna S. Biomedical applications of polymer composite materials. In: Hong CS, Kim CS, Kim CS, editors. Proc. of the second Asian-Australasian Conference on Composite Materials (ACCM-2000), Korea. Advanced Institute of Science and Technology: Korea, 2000. p. A431–0A436.
[199] Ramakrishna, S. (2000B). Textile scaffolds in tissue engineering. Smart fibers, fabrics, and clothing: fundamentals and applications. Editor, X. Tao, Woodhead Publishing Limited, UK.
[200] Ramakrishna S, Fujihara K, Yoshida E, Hamada H. Design and development of braided carbon fiber reinforced polymer composites for dental posts and bone plates. In: 29th FRP symposium, Japan Society for Materials Science, Japan, 2000. p. 11–4.
[201] Ramakrishna S, Ganesh VK, Teoh SH, Loh PL, Chew CL. Fiber reinforced composite product with graded stiffness. Singapore Patent Application No. 9800874-1, 1998.
[202] Ramakrishna S, Ramaswamy S, Teoh SH, Tan CT. Development of a knitted fabric reinforced elastomeric Composite intervertebral disc prosthesis. Proc ICCM-11, Conard Jupiters-Golad oast, Australia 1997;1:458–66.
[203] Ramakrishna S, Ramaswamy S, Teoh SH, Hastings GW, Tan CT. Application of textiles and textile composites concepts for biomaterials development. In: International conference on New Textiles for Composites, Aachen, Germany, 1996. p. 27/1–27/27.
[204] Robin GC. Below-knee drop-foot braces: stresses during use and evaluation of design. In: Ghista DN editor Biomechanics of Medical Devices, Marcel Dekker Inc. 1981. p. 535–567.
[205] Rodriguez-Lorenzo LM, Salinas AJ, Vallet-Regi M, San Roman J. Composite biomaterials based on ceramic polymers. I. Reinforced systems based on Al_2O_3/PMMA/PLLA. J Biomedical Materials Research 1996;30:515–22.
[206] Roman JS, Garcia PG. Partially biodegradable polyacrylic-polyester composites for internal bone fracture fixation. Biomaterials 1991;12:236–41.
[207] Rubin P. Hip joint replacement and morphometric aspects of the bone-implant system. Ecole Polytechnique Federale De Lausanne, Lausanne, 1992. p. 189.
[208] Rubin LR. Biomaterials in maxillo-facial surgery. In: Szycher M, editor. Biocompatible Polymers, Metals, and Composites. Lancaster, USA: Technomic Publishing, 1983. p. 941–51.
[209] Rushton N, Rae T. The intra-articular response to particulate carbon fiber reinforced high density polyethylene and its constituents: an experimental study in mice. Biomaterials 1984;5:352–6.
[210] Saha S, Pal S. Mechanical properties of bone cement: a review. J Biomedical Materials Research 1984;18:435–62.
[212] Sbarbati R, Giannessi D, et al. Pyrolytic carbon coating enhances Teflon and Dacron fabric compatibility with endothelial cell growth. International J Artificial Organs 491-;14(8).
[213] Schmitt-Thomas KhG, Yang ZG, Hiermer T. Performace characterization of polymeric composite implant rod subjected to torsion. In: Proceedings of ICCM-11, Gold Coast, Australia, 1997. p. V277–V286
[214] Schneider E, et al. A comparative study of the initial stability of cementless hip prostheses. Clin Orthop 1989;248:200–9.
[215] Scholz C, Huimacher D, Bahr W, Cales L. The development and testing of three biodegradable screw plate-systems for maxillofacial surgery. In: Kim HL, Chung LP. editors. Biodegradable Implants in Fracture Fixation; 1993. p. 67–73.
[216] Sclippa E, Piekarski K. Carbon fiber reinforced polyethylene for possible orthopedic uses. J Biomedical Materials Research 1973;7:59 –570.
[218] Seedham BB. Ligament reconstruction with reference to the anterior cruciate ligament of the knee. Mechanics of Human Joints. In: Wright V, Radin EL, editors. Physiology, Pathophysiology, and Treatment. New York, USA: Marcel Dekker, 1993. p. 163–201.
[219] Silverton C, Rosenberg AO, Barden RM, Sheinkop MB, Galante JO. The prosthesis-bone interface adjacent to tibial components inserted without cement. The J Bone and Joint Surgery 1996;78-A(3):340–7.
[220] Simoes JA, Marques AT, Jeronimidis G. Design of a controlled-stiffness composite proximal femoral prosthesis. Composites Science and Technology 1999;60:559–67.
[221] Speidel MO, Uggowitzer PJ. Biocompatible nickel-free stainless steel to avoid nickel allergy. In: Speidel MO, Uggowitzer, editors. Materials in Medicine. Switzerland: vdfHochschulverlag AG an der ETH zurich, 1998. p. 191–208.
[222] St.John KR. Applications of Advanced Composites in Orthopedic Implants. In: Szycher M., editor. Biocompatible Polymers, Metals, and Composites. Lancaster: Technomic Publishing, 1983. p. 861–71.
[223] Suwanprateeb J, Tanner KE, Turner S, Bonfield W. Influence of Ringer's solution on creep resistance of hydroxyapatite reinforced polyethylene composites. J Materials Science: Materials in Medicine 1997;8:469–72.
[224] Tallent MA, Cordova CW, Cordova DS, Donnelly DS. Thermoplastic fibers for composite reinforcement. In: Lee SM, editor. International Encyclopedia of Composites, Volume 2. New York: VCH Publishers, 1990. p. 466–80.
[225] Tamura J, Kawanabe K, Yamamuro T, Nakamura T, Kokubo T, Yoshihara S, Shibuya T. Bioactive bone cement: The effect of amounts of glass powder and histologic changes with time. J Biomedical Materials Research 1995;29:551–9.
[226] Tayton K, Bradley J. How stiff should semi-rigid fixation of the human tibia be? J Bone and Joint Surgery 1983;65-B:312–5.
[227] Tayton K, Johnson-Nurse C, Mckibbin B, Bradleym J, Hastings GW. The use of semi-rigid carbon fiber reinforced plastic plates for fixation of human fractures. J Bone and Joint Surgery 1982;64B(1):105–11.
[228] Tayton K, Phillips G, Ralis Z. Long-term effects of carbon fiber on soft tissues. J Bone and Joint Surgery 1982;64-B(1):112–4.
[229] Tayton KJJ. The use of carbon fiber in human implants: the state of the art. J Med Engng Tech 1983;7:271–2.
[230] Teoh SH, Tang ZG, Ramakrishna S. Development of thin elastomeric composite membranes for biomedical applications. J Materials Science: Materials in Medicine 1999;10:343–52.
[231] Timmie Topoleski LD, Ducheyne P, Cuckler JM. The fracture toughness of titanium-fiber-reinforced bone cement. J Biomedical Materials Research 1992;26:1599–617.
[232] Toh SL, Goh JCH, Tan PH, Tay TE. Fatigue testing of energy storing prosthetic feet. Prosthetics and Orthotics International 1993;17:180–8.
[233] Tonino AJ, Davidson CL, Klopper PJ, Linclau LA. Protection from stress in bone and its effect. J Bone and Joint Surgery 1976;58-B(1):107–13.

[234] Torbjorner A, Karlsson A, Syverud M, Hensten-Pattersen A. Carbon fiber reinforced root canal posts, mechanical properties and cytotoxic properties. Eur J Oral Sci 1996;104:605–11.
[235] Tormala P, Vainionpaa S, Kilpikari J, Rokkanen P. The effects of fiber reinforcement and gold plating on the flexural and tensile strength of PGA/PLA copolymer materials in vitro. Biomaterials 1987;8:42–5.
[236] Tormala P, Vasenius J, Vainionpaa S, Laiho J, Pohjonen T, Rokkanen P. Ultra-high-strength absorbable self-reinforced polyglycolide (SR-PGA) composite rods for internal fixation of bone fractures: In vitro and in vivo study. J Biomedical Materials Research 1991;25:1–22.
[237] Trabert KC, Cooney JP. The endodontically treated tooth, restorative concepts and techniques. Dental Clinics of North America 1984;28(4):923–51.
[238] Uhthoff HK, Finnegan M. The effects of metal plates on post-traumatic remodeling and bone mass. J Bone and Joint Surgery 1983;65B(1):66–71.
[239] Urbaniak JR, Bright DS, Hopkins JE. Replacement of intervertebral discs in chimpanzees by silicone-Dacron implants: a preliminary report. J Biomedical Materials Research Symposium 1973;4:165–86.
[240] Valdevit ADC, Inoue N, MacWilliams BA, Anderson LL. Methods for mechanical testing of spinal constructs. Spine 1996;10(2):231–48.
[241] Verheyen CCPM, de Wijin JR, van Blitterwijk CA, de Groot K. Evaluation of hydroxyapatite/poly(L-lactide) composites: mechanical behavior. J Biomedical Materials Research 1992;26:1277–96.
[242] Verheyen CCPM, de Wijn JR, van Blitterwijk CA, de Groot K, Rozing PM. Hydroxyapatite/poly(L-lactide) composites: an animal study on push-out strengths and interface histology. J Biomedical Materials Research 1993;27:433–44.
[243] Wagner HD, Cohn D. Use of high-performance polyethylene fibers as a reinforcing phase in poly(methylmethacrylate) bone cement. Biomaterials 1989;10:139–41.
[244] Walker PS, et al. Strains and micromotions of press-fit femoral stem prostheses. J Biomechanics 1987;20:693–702.
[245] Walker PS, Blunn GW. Biomechanical principles of total knee replacement design. In: Mow VC, Hayes WC, editors. Basic Orthopaedic Biomechanics. Philadelphia, USA: Lippincott-Raven Publishers, 1997. p. 461–93.
[246] Ward R, Minns RJ. Woven carbon fiber mesh patch versus Dacron mesh in the repair of experimental defects in the lumbar fascia of rabbits. Biomaterials 1989;20:425–8.
[247] Weinans H. Mechanically induced bone adaptations around hip implants. PhD thesis. University of Nijmegen, Netherlands, 1991.
[248] Wening JV, Marquardt H, Katzer A, Jungbluth KH. Cytotoxicity and mutagenicity of Kelvar®: an in vitro evaluation. Biomaterials 1995;16(4):337–40.
[249] Wenz LM, Merritt K, Brown SA, Moet A, Steffee AD. In vitro biocompatibility of polyetheretherketone and polysulfone composites. J Biomedical Material Research 1990;24:207–15.
[250] Werkmeister JA, Edwards GA, Casagranda F, White JF, Ramshaw JAM. Evaluation of a collagen-based biosynthetic material for the repair of abdominal wall defects. J Biomedical Materials Research 1998;39(3):429–36.
[251] Whiteside E. The effect of stem fit on bone hypertrophy and pain relief in cementless total hip arthroplasty. Clinical Orthopedics 1989;247:138–47.
[252] Willert HG, Ludwig J, Semlitsch M. Reaction of bone to methacrylate after hip arthroplasty. J Bone and Joint Surgery 1974;56A:1368–82.
[253] Williams DF. Consensus and definitions in biomaterials. In: de Putter C, de Lange K, de Groot K, Lee AJC, editors. Advances in Biomaterials. Elsevier Science: Amsterdam, 1988. p. 11–16.
[254] Williams DF, McNamara A, Turner RM. Potential of Polyetheretherketone (PEEK) and carbon-fiber-reinforced PEEK in medical applications. J Materials Science Letters 1987;6:188–90.
[255] Williams D. Concise Encyclopedia of Medical and Dental Materials. Oxford, UK: Pergamon Press, 1990.
[256] Williams DF, Cunningham J. Materials in Clinical Dentistry. Oxford, UK: Oxford University Press, 1979.
[258] Wintermantel E, Mayer J. Anisotropic biomaterials strategies and developments for bone implants. In: Wise DL, Trantolo DJ, Altobelli DE, Yaszemiski JD, Gresser JD, Schwartz ER, editors. Encyclopedic Handbook of Biomaterials and Bioengineering, Part B-1. Marcel Dekker: New York, 1995. p. 3–42.
[259] Wintermantel E, Bruinink A, Eckert K, Ruffiex, K Petitmermet M, Mayer J. Tissue engineering supported with structured biocompatible materials: goals and achievements. In: Speidel MO, Uggowitzer, editors. Materials in Medicine. Switzerland: vdf Hochschulverlag AG an der ETH zurich, 1998. p. 1–136.
[260] Wintermantel E, Ha SW. Biokompatible Werkstoffe und Bauweisen: Implantate fur Medizin und Umwelt. Berlin, Germany: Springer-Verlag, 1998.
[261] Wintermantel E, Koch B, Mayer J, Snobl B, Scheidegger A. CAD, CAE and CAM for a new anisotropic carbon fiber reinforced hip endoprosthesis stem, technology, test procedures and qualifying preclinical results. J Biomedical Engineering, Application, Basis, Coomunication 1994;6:18–21.
[262] Wintermantel E, Mayer J, Blum J, Eckert KL, Luscher P, Mathey M. Tissue engineering scaffolds using superstructures. Biomaterials 1996;17:83–91.
[263] Woo SLY, Akeson WH, Levenetz B, Coutts RD, Matthews JV, Amiel D. Potential application of graphite fiber and methyl methacrylate resin composites as internal fixation plates. J Biomedical Materials Research 1974;8:321–38.
[264] Woo SL-Y, Akeson WH, Coutts RD, Rutherford L, Doty D, Jemmott GF, Amiel D. A comparison of cortical bone atrophy secondary to fixation with plates with large differences in bending stiffness. J Bone and Joint Surgery 1976;58A(2):190–5.
[265] Wright TM, Fukubayashi T, Burstein AH. The effect of carbon fiber reinforcement on contact area, contact pressure, and time-dependent deformation in polyethylene tibial components. J Biomedical Materials Research 1981;15:719–30.
[266] Wroblewski BM. Wear and loosening of the socket in the Charnley low-friction arthroplasty. Orthop Clin North Am 1988;19:627–30.
[267] Yang JM, Huang PY, Yang MC, Lo SK. Effect of MMA-g-UHMWPE grafted fiber on mechanical properties of acrylic bone cement. J Biomedical Materials Research 1997;38(4):361–9.
[268] Yildiz H, Chang FK, Goodman S. Composite hip prostheses design. II. Simulation. J Biomedical Materials Research 1998A;39(1):102–19.
[269] Yildiz H, Ha SK, Chang FK. Composite hip prostheses design. I. Analysis. J Biomedical Materials Research 1993;39(1):92–101.
[270] Yu TJ, Chu CC. Bicomponent vascular grafts consisting of synthetic absorbable fibers. I. in vitro study. J Biomedical Materials Research 1993;27(10):1329–39.
[271] Yu TJ, Ho DM, Chu CC. Bicomponent vascular grafts consisting of synthetic absorbable fibers: Part II: in vivo healing response. J Invest Surg 1994;7(3):195–211.
[272] Zimmerman M, Parsons JR, Alexander H. The design and analysis of a laminated partially degradable composite bone plate for fracture fixation. J Biomedical Materials Research: Applied Biomaterials 1987;21(A3):345–61.
[273] Zufall SW, Kennedy KC, Kusy RP. Frictional characteristics of composite orthodontic arch wires against stainless steel and ceramic brackets in the passive and active configurations. J Materials Science: Materials in Medicine 1998;9:611–20.

[274] Ratnev, BD, Hoffman, AS, Schoen, FJ, Lemons, JE. An introduction to materials in medicine, Academic Press, New York, USA, 1996;308–18

[275] Cao, Y, Vacanti, JP, Ma, PX, Ibarra, C, Paige, KT, Upton, J, Langer, R, Vacanti, CA. Tissue engineering of tendon. In: Proceedings of Materials Research Society Symposium, 1995;(394):83–9

[276] Fujihara, K, Huang, ZM, Ramakrishna, S, Satkunanathan, S, Hamada, H. Development of braided carbon/PEEK Composite bone plates. Advanced Composites Letters 2001;10(1):13–20.

[277] Loh, PL, Ravi, K, Ganesh, UK, Ramakrishna, S, Chew, CL. Moisture absorption of carbon fiber reinforced posts. J. Dental Research 2000;79(5):1317.