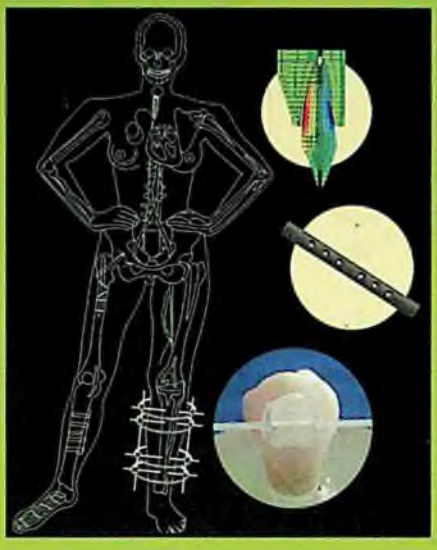

Vol. 1

Series on Biomaterials and Bioengineering

An Introduction to BIOCOMPOSITES

Seeram Ramakrishna
Zheng-Ming Huang
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FOREWORD

Several years of cumulative research has been conducted on the usage of fiber reinforced composites for biomedical application, but no one source exists where this topic has been addressed in a systematic manner in a whole monograph. The focus of this book is on polymer composites applied to bioengineering.

Fiber reinforced composites have highly favorable mechanical and durability related properties. When compared with metals, they offer many advantages such as non-corrosiveness, radio translucency, good bonding properties and ease of repair. Since they also offer the potential for chair-side and laboratory fabrication, it is not surprising that fiber reinforced composites have potential for use in many applications in bioengineering.

For the potential applications to become successful applications, requires taking full advantage of the materials properties together with manufacturing techniques to realize the biomedical application needs. Hence, this book focuses on fiber-based composites applied to bioengineering. The book addresses three main areas: first, a comprehensive survey of biocomposites from the existing literature in various medical applications, primarily focusing on hard tissues related implants is presented. Second, mechanical designs and manufacturing aspects of various fibrous polymer matrix composites are described.

The third major issue addressed in the book is the design and development examples of several medical devices and implants using polymer composites. These devices are supposed to be used for hard tissue applications, including prosthetic socket, dental post, external fixator, bone plate,

orthodontic archwire, orthodontic bracket, total hip replacement, and composite screws and pins. Fabrication and mechanical testing of these have been shown, with comparisons with other clinically used medical devices. In this book, designing procedures for those medical devices using continuous fiber reinforced polymer matrix composites are described in sufficient detail. Based on these, comparable procedures can be followed if other critical designs are to be made.

Professor Eric Wintermantel
2003

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The Authors

CONTENTS

Chapter 1

INTRODUCTION

1.1 Biomaterials

There are different definitions of 'bioengineering' [Berger *et al.*, 1996]. Here, we refer 'bioengineering' to the application of concepts and methods of the physical sciences and mathematics in an engineering approach towards solving problems in repair and reconstructions of lost, damaged or deceased tissues. Any material that is used for this purpose can be regarded as a biomaterial. According to Williams [1987], a biomaterial is a material used in implants or medical device, intended to interact with biological systems. Thus, a biomaterial must always be considered in its final fabricated and sterilized form. Examples of common medical devices are: substitute heart valves and artificial hearts, artificial hip and knee joints, dental implants, internal as well as external fracture fixators, skin repair templates as well as dialysers to support kidney functions or intraocular lenses. A material that can be used for medical application must possess a lot of specific characteristics, of which the most fundamental requirements are related with biocompatibility.

Over the last thirty 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' [Williams, 1988] to indicate the biological

performance of materials. Thus, materials that are biocompatible can be considered as 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 [Black and Hastings, 1998]. Researchers [Wintermantel and Mayer, 1995] extended this definition and distinguished between surface and structural compatibility of an implant. Surface compatibility means 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 deformation characteristics, and optimal load transmission (minimum interfacial strain mismatch) at the implant/tissue interface. Optimal interaction between biomaterial and host tissue is reached when both the surface and the structural compatibilities are met. Furthermore, 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 imposed during implantation, sterilization methods, etc), health condition and activities of the patient. Table 1.1 summarizes several important factors that can be considered in selecting a material for a biomedical application [Ramakrishna *et al.*, 2001].

Until recently, most medical devices are still made from single-phase homogeneous and isotropic materials such as polymers, metals, and ceramics. A large number of polymers are widely used in various medical 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 for load bearing applications, 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, and 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. Most common are stainless steel, cobalt-chromium alloys as well as titanium and titanium base alloys. Major disadvantages of those metals

Table 1.1 Various factors of importance in material selection for biomedical applications [Ramakrishna *et al.*, 2001].

Factors	Description		
	Chemical/Biological Characteristics	Physical Characteristics	Mechanical/Structural Characteristics
1st Level Material Properties	– chemical composition (bulk and surface)	– density	– elastic modulus – shear modulus – Poisson's ratio – yield strength – tensile strength – compressive strength
2nd Level Material Properties	– adhesion	– surface topology – texture – roughness	– hardness – flexural modulus – flexural strength
Specific Functional Requirements (based on application)	– biofunctionality – bioinert – bioactive – biostability – biodegradation behavior	– form & geometry – coefficient of thermal expansion – electrical conductivity – color, aesthetics – 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 & 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			

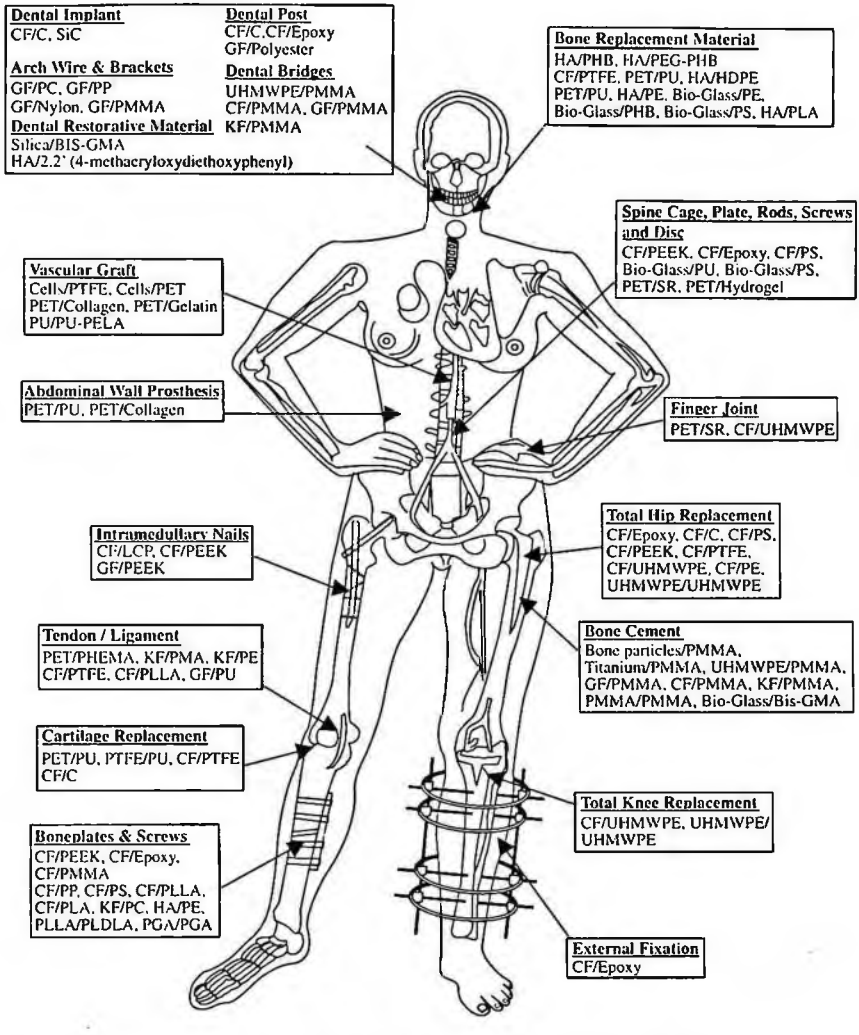
are their high stiffness compared to host tissues as well as their tendency to create severe imaging artifacts in the most advanced diagnostic 3-D imaging procedures i.e. X-ray Computer Tomography (CT) and nuclear Magnetic Resonance Imaging (MRI). Stainless steel and cobalt-chromium alloys are sensitive to corrosion, thus releasing metal ions which may cause allergic

tissue reactions (Nickel and Chromium allergies) [Speidel and Uggowitzer, 1998]. Titanium and its alloys, however, expand their range of applications because of their excellent biocompatibility. Ceramics are known for their good biocompatibility, corrosion resistance, and high compression resistance. Drawbacks of ceramics include brittleness, low fracture strength, difficulty in fabrication, low mechanical reliability and lack of resilience. These drawbacks in the traditional biomaterials have stimulated researchers and engineers to develop composite materials as an alternative choice in bioengineering applications.

1.2 Potential of Biocomposites for Medical Applications

Composites are those materials that contain two or more distinct constituent phases, on a scale larger than the atomic. The term 'biocomposites' specially refers to those composites that can be employed in bioengineering. The constituents retain their identities in the composite. Namely, they do not dissolve or otherwise merge completely into each other although they act in concert. Normally, the constituent components can be physically identified and exhibit an interface between one another. In composites, properties such as the elastic modulus can be significantly different from those of the constituents alone but are considerably altered by the constituent structures and contents. From a structural point of view, composites are anisotropic in nature. Their mechanical properties are different in different directions. Most of the living tissues such as bone, dentin, collagen, cartilage, and skin are essentially composites. Those natural biocomposites are not discussed in this book, but the reader can refer to, e.g. [D. Taylor, 2003]. Synthetic composites are essentially a combination of two constituent phases, i.e. a reinforcing phase such as fiber or particle and a continuous phase called matrix.

The primary motive in the development of biocomposites is that by varying the type and distribution of the reinforcing phases in the composites it is possible to obtain a wide range of mechanical and biological properties, and hence to optimize the structure and performance of the biomedical devices and their interaction with the surrounding tissues. A schematic diagram to show potential use of biocomposites in the repair, reconstruction, and replacement of human hard tissues is given in Fig. 1.1. A number of polymer



CF: Carbon fibers, C: Carbon, GF: Glass fibers, KF: Kevlar fibers, PMMA: Polymethymethacrylate,
PS: Polysulfone, PP: Polypropylene, UHMWPE: Ultra-high-molecular weight polyethylene,
PLDLA: Poly(L-DL-lactide), PLLA: Poly (L-lactic acid), PGA: Polyglycolic acid, PC: Polycarbonate,
PEEK: Polyetheretheketone; 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. 1.1 Various applications of different polymer composite biomaterials.

matrix composite materials were investigated for medical applications over the years. The early composites have been successfully used clinically, e.g. cages for spinal fusion, while the others are still under development. There are a number of factors that led to the development of composite materials. Some specific advantages of polymer composites are highlighted in the following.

In general, tissues are grouped into hard and soft tissues. Bone and tooth are the only examples of hard tissues, whereas skin, blood vessels, cartilage and ligaments are a few examples of soft tissues. As the names suggested, the hard tissues are generally stiffer (with higher elastic modulus) and stronger (with higher tensile strength) than the soft tissues (Tables 1.2 and 1.3). Moreover they are essentially composite materials with anisotropic properties, which depend on the roles and structural arrangements of various components (e.g. collagen, elastin, and hydroxyapatite) of the tissues. For

Table 1.2 Mechanical properties of hard tissues, representative values only, note that tissues show broad variation [Black and Hastings, 1998].

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 1.3 Mechanical properties of soft tissues, representative values only, note that tissues show broad variation [Black and Hastings, 1998].

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

example, the longitudinal mechanical properties of cortical bone are higher than the transverse direction properties (see Table 1.2). The anisotropy of the elastic properties of the biological tissues has to be considered as one an essential design criterion for implants made from composite biomaterials.

From the mechanical point of view, metals or ceramics seem to be more suitable for hard tissue applications (Tables 1.2 and 1.4), while polymers for soft tissue applications (Tables 1.3 and 1.5). However, a closer look at Tables 1.2 and 1.4 reveals that the elastic moduli of metals and ceramics are at least 10 to 20 times higher than those of the hard tissues. Thus, implants made from these materials tend to be much stiffer than the tissue to which

Table 1.4 Mechanical properties of typical metallic and ceramic biomaterials, representative values only [Black and Hastings, 1998].

Material	Modulus (GPa)	Tensile Strength (MPa)
Metal Alloys		
Stainless Steel	190	586
Co-Cr alloy	280	1085
Ti-alloy	116	965
Amalgam	30	58
Ceramics		
Alumina	380	300
Zirconia	220	820
Bioglass	35	42
Hydroxyapatite	95	50

Table 1.5 Mechanical properties of typical polymeric biomaterials, representative values only [Black and Hastings, 1998].

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)	3.3	110
Silicone Rubber (SR)	0.008	7.6
Polysulfone (PS)	2.65	75

they are attached. In orthopedic surgery, this mismatch of stiffness between the bone and the metallic or ceramic implants influences the load sharing between the bone and implant. Since the amount of stress carried by each of them is directly related to their stiffness, bone is insufficiently loaded compared to the implant. According to Wolffs law of stress related bone remodeling [Hayes and Snyder, 1981], this may lead to lower bone density and altered bone architecture. In osteosynthesis, this may affect healing of the fractured bones and may increase the risk of refracture of the bone after removal of the osteosynthesis implant, e.g. bone plate.

It has been recognized that by matching the stiffness of implant with that of the host tissues can reduce such negative effects and support desired bone 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 fiber-reinforced polymers i.e. polymer composite materials offer both low elastic modulus and high strength, they have been proposed for several orthopedic applications. A further merit of composite materials is that by controlling the volume fractions and local and global arrangement of 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.

Composite materials offer several other significant advantages over metal alloys and ceramics, e.g. absence of corrosion and release of allergenic metal ions such as Nickel or Chromium, high fracture toughness and higher resistance against fatigue failure [Hastings, 1983; Tayton, 1983; Tayton and Bradley, 1983]. Polymer composite are basically radiolucent materials, however, their radio transparency can be adjusted by adding contrast medium to the polymer. Moreover the polymer composite materials are highly compatible with the modern diagnostic methods such as computed tomography (CT) and magnetic resonance imaging (MRI) as they show very low X-ray scattering and their magnetic susceptibility is very close to that of human tissue. Considering their light weight and superior mechanical properties, the polymer composites are also used as structural components of these imaging devices. For some applications as in dental implants, polymer composites can offer better aesthetic characteristic. Furthermore,

since implants from polymer composites can be manufactured using high throughput technologies e.g. injection molding, net shape pressing and high speed machining, they become competitive to metal implants from the point of view of cost management too.

1.3 Classification of Composite Materials

There are several means which can be used to classify composites in applications. Figure 1.2 shows main types of bio-composites according to their reinforcement forms. From Fig. 1.2, we can see that there are typically three kinds of reinforcements, i.e. short fibers, continuous fibers, and particulates (powders). All of them have been used in the development of composites for bio-medical applications, such as screws and total hip replacement stems made from short fiber reinforcements (Figs. 1.3 [Tognini, Ph.D. Thesis, ETH Zurich, 2001] and 1.4 [Semadeni, Ph.D. Thesis, ETH Zurich, 1999]), orthopedic bone plates fabricated using unidirectional (UD) laminae or multidirectional tape laminates (Fig. 1.5) [Evans and Gregson, 1998], and powder reinforced dental composites [Nicholson, 1998; Moszner and Salz, 2001]. Another classification for biocomposites

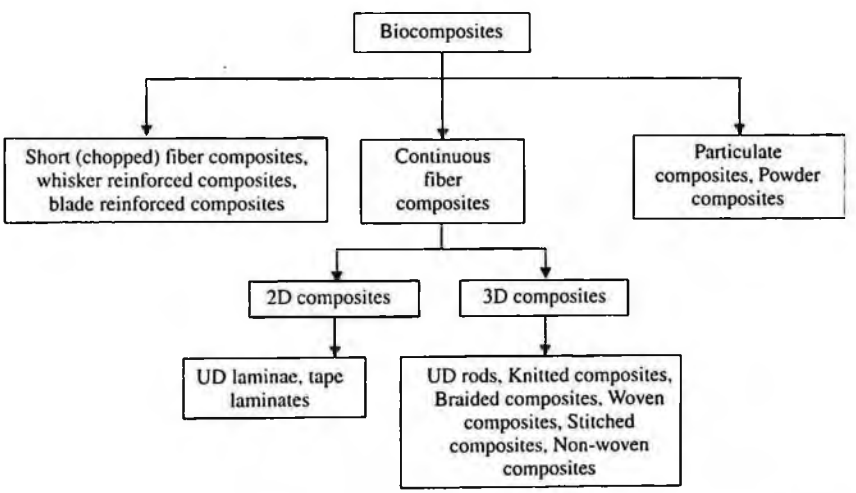


Fig. 1.2 Classification of biocomposites based on their reinforcement form.

Fig. 1.3 Endless carbon fiber reinforced PEEK matrix medical screws made by Composite Flow Molding, carbon fiber volume content 62% (by courtesy of Icotec AG, Switzerland) [R. Tognini, Ph.D. Thesis, ETH Zurich, 2001].

Fig. 1.4 Hip endoprosthesis stem, injection molded, chopped long fiber reinforced PEEK, fiber volume content 50% [M. Semadeni, Ph.D. Thesis, ETH Zurich, 1999].

is based on their biodegradability, i.e. fully resorbable, partially resorbable, and nonresorbable composites, as shown in Fig. 1.6.

Resorbable biocomposites are made from those fibers and matrices both of which are fully absorbable in the body. Those biocomposites are currently and intensively investigated for internal fracture fixation applications. When a metal fixator is used, a second operation generally has to be performed to remove the implants when the fractured bone has healed

Fig. 1.5 Carbon fiber reinforced epoxy composite bone plates (Evans and Gregson, Biomaterials, Vol. 19, No. 15, pp. 1329–1342, 1998).

completely. This would cause the patient additional risk, pain and expenditure. Such an operation can be avoided if a fully resorbable fixator is used. Most work in the literature on fully resorbable biocomposite fracture fixators has been done based on the group of PLA (polylactic acid) polymers. The reason is that PLAs possess two major characteristics that make them an extremely attractive bioabsorbable material [Alexander, 1996]: (1) they can degrade inside the body in a rate that can be controlled, e.g. by varying molecular weight, the share of their enantiomers L and D-lactide or copolymerising it with PGA (polyglycolic acid) polymer, and (2) and, if crystallization of the PLA-polymer is prevented, their degradation products are nontoxic, biocompatible, and easily metabolized. The main problem of those composites is the coordination of the degradation behavior of both phases and, especially, of the interphase between both.

Partially resorbable biocomposites have been fabricated using non-absorbable reinforcing materials and absorbable matrix materials. Historically, they have been the predecessors of the fully degradable composites. However, due to severe inflammatory tissue reactions on the remaining, non-degradable phase, most of the research on these materials has been stopped. They have been investigated for a number of medical applications such as bone replacements, bone cements and also internal fracture fixators. Particulate reinforced materials which have been practiced include PMMA (polymethylmethacrylate) and PBT (poly(butylene terephthalate)) as non-resorbable matrics in combination with HA or PLA's,

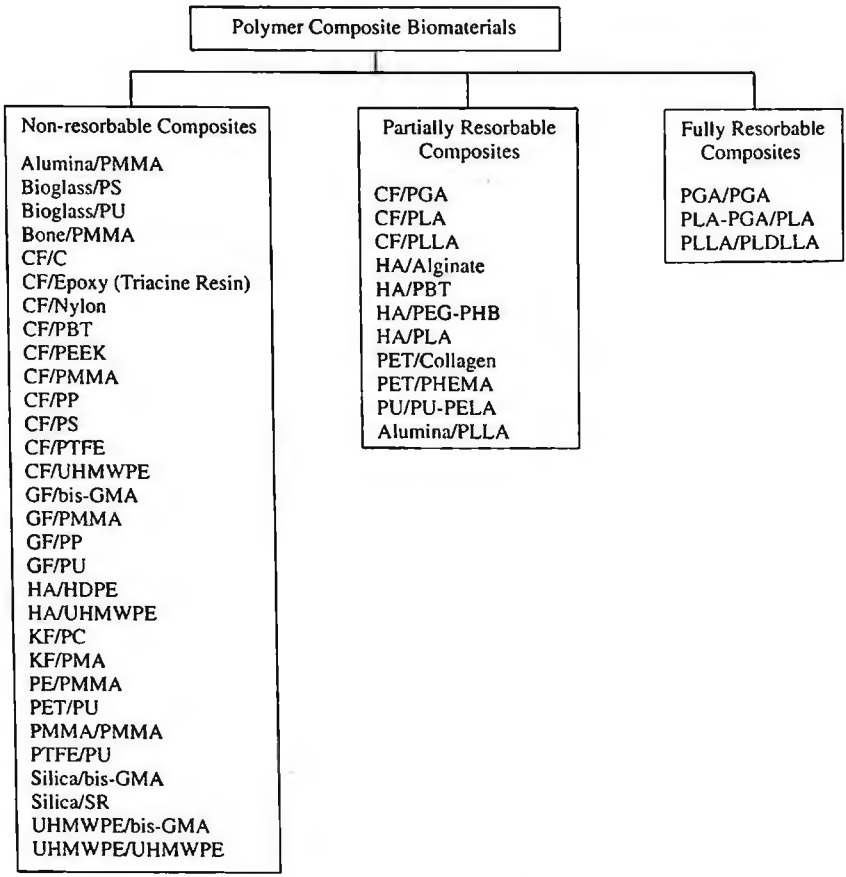


Fig. 1.6 Classification of man-made polymer biocomposites based on biodegradability.

and Polyalkanoates e.g. PHB (polyhydroxybutyrate), combined with non-resorbable filler phases e.g. Alumina or Calcium Carbonate. For internal fixator application, the reinforcing material has been mainly carbon fibers whereas the matrices used have been various PLAs or PLA-PGA copolymers. The only product that raised clinical interest is a composite of polyethylene and HA being applied in some bone graft and replacement applications. In a nonresorbable biocomposite, both the reinforcing phase (fibers or particulate) and the continuous phase are nonresorbable

in the body. There is a large variety of biocomposites which are nonresorbable. They are generally used to provide specific mechanical or clinical properties unattainable with the traditional biomaterials. Currently, the most advanced use of nonresorbable composites is in implants for spinal fusion since they provide superior mechanical stability and allow proper imaging of the reconstructed or stabilized vertebral column. The other potential uses include stems of hip or knee joint prostheses, prosthetic sockets, bone plates, dental posts, external fixators, orthodontic archwires, orthodontic brackets, etc which will be described in Chapter 7 of this book.

1.4 Scope of this Book

This book focuses on polymer composites applied to bioengineering, a topic which has not been systematically addressed in a whole monograph before. There are three purposes for the authors to write the present book. First, a comprehensive survey of biocomposites from the existing literature in various medical applications, primarily focusing on hard tissues related implants, is presented. Second, mechanical (stiffness and strength) designs of various fibrous polymer matrix composites are described only based on their constituent properties. These composites can be tailored to different biomedical applications. For this purpose, a mechanics of composite theory is presented systematically. Finally, a number of typical design and development examples involved with biocomposites are shown in the book.

Although polymer composites have been recognized as potential candidates for medical devices, implants and substitutes, the majority of them are still limited to laboratory investigation level at the present. A great body of studies has accumulated concerning various biomedical applications ranging from the hard tissues to the soft tissues. These study reports have been broadly distributed in many sources of literature publications. There is a need to review and evaluate the contents of these studies so that the non-specialist reader can appreciate the current understanding of polymer biocomposites and that he or she can be stimulated for future investigations in biocomposite science and engineering. Thus, an effort has been made in this book to summarize and survey the various biomedical applications of polymer composites so far achieved.

While a number of issues affect the widespread employment of polymer composites in bioengineering, the technical need for the design and analysis of composite materials and structures remains in place, as an increased use of biocomposites also requires taking full advantages of the material properties together with manufacturing techniques available. For a synthetic composite especially made from continuous or discontinuous fiber reinforcement, its mechanical as well as physiological properties are dependent on a number of variables. The parameters that will influence the composite properties include the mechanical and physiological properties of its constituent materials, constituent contents, reinforcement form, structure, and arrangement pattern in the matrix, interface bonding between the reinforcement and the matrix phases, and so on. Varying these parameters can result in composites with different performances. Thus, a design related problem is to achieve a polymer composite with optimal mechanical as well as physiological performance by choosing suitable values of the design parameters. This is possible only when the composite properties can be quantitatively represented as the functions of those design variables. The micromechanics theory can be applied to accurately estimate the composite properties in terms of its constituent properties and geometrical parameters.

In this book, micromechanics models of the stiffness and strength are presented. Composite elastic behavior, its inelastic and strength properties can be estimated by rigorous application of micromechanics. The detailed development of the model is not shown in the book, but can be found in cited literature [Huang, 2000]. Attention has been focused on its wide applicability. The analysis and designing procedures for various fiber composites including unidirectional lamina, multidirectional tape laminate, woven, braided, and knitted fabric reinforced composites are described in the book. The strength characteristics of any continuous fiber reinforced composite can be simulated, as long as the fiber orientation in the composite can be identified. Prediction of the mechanical properties of a fibrous composite primarily involves an analysis of the geometry of the fibrous structure in the composite. Once an accurate knowledge has been obtained of the relationship between the mechanical characteristics of the composite and the material properties and geometrical structure of its constituents, stiffness and strength designs can be performed. This has been done in the book for composites reinforced with a number of typical fiber preforms and structures.

The third major issue addressed in the book is the design and development examples of several medical devices and implants using polymer composites. These devices are supposed to be used for hard tissue applications, including Prosthetic socket, Dental post, External fixator, Bone plate, Orthodontic archwire, Orthodontic bracket, Total hip replacement, and Composite screws and pins. Fabrication and mechanical testing of them have been shown, with comparisons with other clinically used medical devices if possible. Among them, some devices such as bone plate and archwire are primary load carrying elements. Their ultimate strength behavior must be targeted during the design. It is noted that both of them are mainly subjected to lateral loading (bending) in their clinical application. According to current understanding, the estimation of composite bending strength remains a challenge. In this book, design procedures for those medical devices using continuous fiber reinforced polymer matrix composites are described in sufficient detail. We believe that comparable procedures can be followed if other critical designs are to be made.

References

K. P. Baiday, S. Ramakrishna, and M. Rahman, Quantitative radiographic analysis of fiber reinforced polymer composites, *Journal of Biomaterials Applications*, 2001, **15**(3), 279–289.

S. A. Berger, W. Goldsmith, and E. R. Lewis, eds., *Introduction to Bioengineering*, Oxford University Press, Oxford, 1996.

J. Black and G. W. Hastings, *Handbook of Biomaterials Properties*, Chapman and Hall, London, UK, 1998.

S. L. Evans and P. J. Gregson, Composite technology in load-bearing orthopaedic implants, *Biomaterials*, 1998, **19**, 1329–1342.

G. W. Hastings, Biomedical applications of CFRPs, in *Carbon Fiber and their Composites*, ed. E. Fitzer, Springer-Verlag, Berlin, 1983, pp. 261–271.

W. C. Hayes and B. Snyder, Towards a quantitative formulation of Wolff's law in trabecular bone, in "*Mechanical Properties of Bone*", ed. S. C. Cowin, *The Joint ASME-ASCE Applied Mechanics, Fluids Engineering and Bioengineering Conference, AMD-Vol. 45*, Boulder, Colorado, 1981.

G. O. Hofmann, Biodegradable implants in orthopaedic surgery — A review on the state-of-the-art, *Clinical Materials*, 1992, **10**, 75–80.

Z. M. Huang, S. Ramakrishna, and A. A. O. Tay, Unified micromechanical model for estimating elastic, elasto-plastic, and strength behaviors of knitted fabric reinforced composites, *Journal of Reinforced Plastics and Composites*, 2000, **19**(8), 642–656.

T. W. Lin, A. A. Corvelli, C. G. Frondoza, J. C. Roberts, and D. S. Hungerford, Glass peek composite promotes proliferation and osteocalcin production of human osteoblastic cells, *Journal of Biomedical Materials Research*, 1997, **36**(2), 137–144.

N. Moszner and U. Salz, New developments of polymeric dental composites, *Progress in Polymer Science*, 2001, **26**(4), 535–576.

J. W. Nicholson, Adhesive dental materials — A review, *International Journal of Adhesion and Adhesives*, 1998, **18**(4), 229–236.

S. Ramakrishna, J. Mayer, E. Wintermantel, and K. W. Leong, Biomedical applications of polymer-composite materials: A review, *Comp. Sci. & Tech.*, 2001, **61**(9), 1189–1224.

B. D. Ratner, Biomaterials science: An interdisciplinary endeavor, in Bio-materials science, an introduction to materials in medicine, eds. B. D. Ratner, A. S. Hoffman, F. J. Schoen, and J. E. Lemons, Academic Press, San Diego, 1996, pp. 1–8.

M. O. Speidel and P. J. Uggowitzer, Biocompatible nickel-free stainless steel to avoid nickel allergy. *Materials in Medicine*, eds. M. O. Speidel and Uggowitzer, vdf Hochschulverlag AG an der ETH zurich, Switzerland, 1998, pp. 191–208.

D. Taylor, Hard and soft tissue mechanics, *Comprehensive Structural Integrity, Vol. 9: Bioengineering*, eds. S. H. Teoh and Y.-W. Mai, Elsevier Science Publisher, UK, 2003.

K. Tayton and J. Bradley, How stiff should semi-rigid fixation of the human tibia be? *Journal of Bone and Joint Surgery*, 1983, **65-B**, 312–315.

K. J. J. Tayton, The use of carbon fiber in human implants: The state of the art, *J. Med. Engng. Tech.*, 1983, **7**, 271–272.

P. Tormala, S. Vainionpaa, J. Kilpikari, and P. Rokkanen, The effects of fiber reinforcement and gold plating on the flexural and tensile strength of PGA/PLA copolymer materials *in vitro*, *Biomaterials*, 1987, **8**, 42–45.

D. F. Williams (1987), Definitions in biomaterials, *Proceedings of a Consensus Conference of the European Society for Biomaterials*, Chester, England, March 3–5, 1986, **4**, Elsevier, New York.

D. F. Williams, Consensus and definitions in biomaterials, *Advances in Biomaterials*, eds. C. de Putter, K. de Lange, K. de Groot, and A. J. C. Lee, Amsterdam, Elsevier Science, 1988, pp. 11–16.

E. Wintermantel and J. Mayer, Anisotropic biomaterials strategies and developments for bone implants, in *Encyclopedic Handbook of Biomaterials and Bioengineering*, Part B-1, eds. D. L. Wise, D. J. Trantolo, D. E. Altobelli, J. D. Yaszemiski, J. D. Gresser and E. R. Schwartz, New York, Marcel Dekker, 1995, pp. 3–42.

Chapter 2

BIOCOMPATABILITY

2.1 The Environment Within the Human Body

The human environment within the human body can be crudely defined as warm, aqueous and saline. The mineral composition of most body fluids closely resembles sea water. With the exception of bodily extremities exposed to very cold air, most of the body remains at a constant temperature close to 37 degrees Celsius. Body fluids with the important exceptions of stomach contents, urine, sweat and tears are almost neutral like water itself. The body is intolerant of either strong acidity or severe alkalinity, apart from specialized locations such as the stomach contents. The principle of homeostasis (stability of operating conditions) ensures that there are few, if any, deviations from the norm in the internal environment of the body.

Given these major restrictions on extreme chemical activity in the body, the internal environment of the body is found to be surprisingly demanding on artificial materials. Much of the problems arise from the fact that living tissues are involved where cells routinely secrete large quantities of enzymes, proteins and other chemicals (such as strong oxidants). It is found that proteins, despite their normally mild level of chemical activity, interact strongly with artificial materials. When the interactions between an implant material and the body are beneficial, the material can be said to be biocompatible. When there are hostile interactions, the material is not biocompatible.

2.2 Durability of Artificial Implant Materials in the Body

Metals, polymers and engineering ceramics have been used as implants. Metal implants are subject to corrosion from the saline solution that forms the inorganic component of blood and other body fluids. This corrosion is often accelerated by protein adsorption onto the surface of the metal. The mechanism of how protein adsorption controls corrosion is still not fully understood. A major advantage of ceramics and polymers is that they are largely invulnerable to aqueous corrosion. However, polymers in particular, suffer from other problems such as leaching. It is found that blood is an effective leaching agent since it contains water, inorganic minerals and solubilizing organic compounds such as lipids. Non-degradable ceramics are perhaps the most resistant class of material, but because of their brittleness, the use in the body is limited to very selected application, e.g. articulating bodies in artificial joints.

2.3 Physiological Interactions Between Implant Materials and the Body

From the moment an implant enters the body, there will be some kind of interaction with human tissues at both the biochemical and cellular levels. The initial response after wetting is adsorption of proteins on the surface of the implant. The proteins will in turn attract cells to form an adaptive region around the implant. In some cases, this adaptive region is beneficial and helps generate a bond between the implant and the surrounding tissue. A bond is vitally useful for bone implants where a mechanical strength is required. In most cases, however, a fibrous capsule forms around the implant; the purpose of this is to isolate adjacent tissues from the implant. Biocompatibility usually means that a fibrous capsule does not form; instead, there is a useful degree of integration between the original tissue cells, e.g. bone cells and the implant. A higher degree of biocompatibility known as bioactivity, means that the cells freely integrate and penetrate the implant so that original implant boundary disappears. Instead of a boundary, there is a transition region between the original implant and the adjacent tissue.

With the exception of temporary implants such as 'scaffolds' and drug-releasing implants, the implant should remain intact. However, in many cases, the implant is degraded to progressively leak alien materials into the body. In other cases, for reasons that are not fully understood, the implant may activate the body's immune system to generate local inflammation around the implant. Inflammation involves elevated temperature and swelling. It causes discomfort to the patient with possible impairment of bodily function (e.g. of the limbs). The immune system has evolved to interact and eradicate bacteria and other parasites that are typically minute in size. A major problem is that wear-resistant implant materials may generate wear particles that are comparable in size to bacteria and therefore activate the immune system [Green *et al.*, 1998]. However, the wear debris of biomaterials, unlike bacteria, are not rapidly disposed of. The survival of wear debris causes the inflammation that persists with resultant long-term tissue damage. The body has evolved to tolerate a short episode of tissue damage to combat acute but brief infections. Chronic inflammation that persists over a number of years (the service life of the implant) will at the very least, subject the patient to long periods of pain. A further category of response, which is the most undesirable perhaps, is that the implant is carcinogenic, i.e. initiates cancer. Fortunately, this has only happened in a few exceptional cases. The most common problems associated with implants are chronic inflammation and post-operative infection. The act of placing an implant, i.e. cutting open the body, presents an excellent opportunity for invasion of the body by any bacteria that happen to be present. Once the bacteria are inside the body, they are very likely to cause an infection. The site of an infection is as critical to the patient's survival as is the type of bacteria. If even a common type of bacteria, which lives harmlessly on the skin, is able to gain access to the underlying tissues, then there may be severe consequences for the patient.

Apart from cases where known toxic materials are used, the only instance where implants caused significant numbers of chronic health problems was where large quantities of wear debris was generated. For reasons that are poorly understood, it appears that the minute size and unnatural shape of wear debris may provoke a pathological response in the patient. The early models of orthopaedic implant were fitted with sliding surfaces made of PTFE (Poly-tetra-fluoro-ethylene), since PTFE was known to offer

a low sliding friction coefficient at moderate sliding speeds and temperatures. These implants were found to generate unacceptably large amounts of wear debris, necessitating their urgent replacement. PTFE has among the lowest friction coefficients in dry sliding against metals or ceramics, but it also has amongst the highest wear rates of all known polymers.

The reaction of the body to an implant is illustrated schematically in Fig. 2.1.

When a material is implanted in the body, the body may accept the material and allow cells to achieve close contact with the material and form strong bonds. If a basically biocompatible material is porous or pitted, then the cells may invade the pores and pits to achieve a stronger bond. When a material is not accepted, the living tissues are unable to maintain intimate contact with the material. To prevent such contact, the body forms a casing of dense connective tissue around the implant [Hench]. The fibrous component of this tissue serves to separate the living cells from the implant material. This tissue does not mechanically bond with the implant, which remains loosely held in the capsule of dense connective tissue. In many instances, a heavily loaded implant may loosen and cease to function [Hench]. In extreme cases of incompatibility between the implant and tissue may lead to local tissue necrosis. In those rare cases, a very heavy tissue reaction

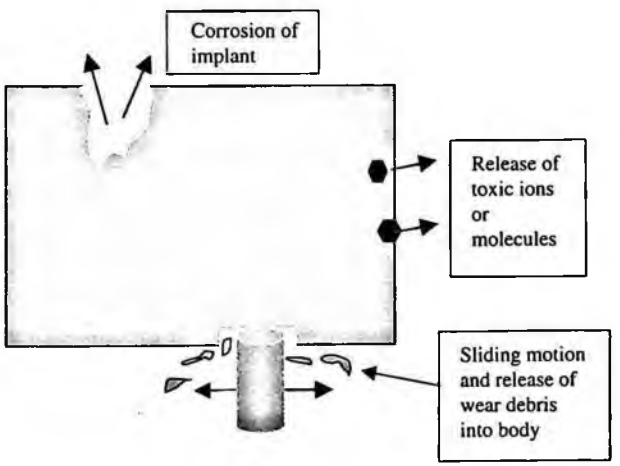


Fig. 2.1 Interaction between the body and implanted material.

follows that normally requires removal of the implant and resection of the necrotic tissue.

The difference between an bioactive implant that fully intrgrates into the tissue and an bioinert implant that becomes encapsulated by a connective tissue membrane is illustrated schematically in Fig. 2.2.

The fibrous component of a biocomposite would be vulnerable to fretting wear if the fibers are not encased in a solid matrix. Fretting wear is caused by microscopic movements between contacting solids. The amplitude of movement is typically only a few micrometers, but the wear rate is high when compared to the total sliding distance. Contacting fibers are

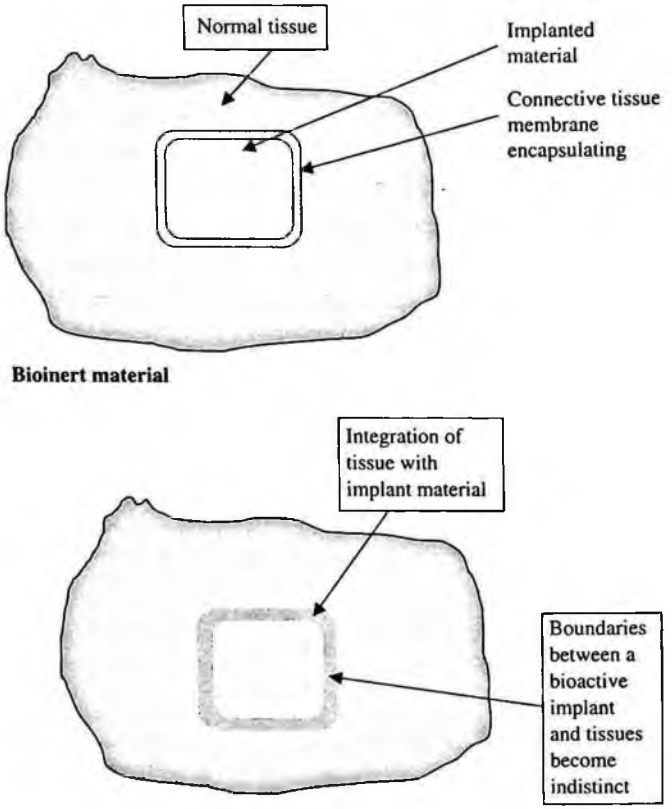


Fig. 2.2 Long-term bonding and non-bonding of implants inside the body.

particularly vulnerable to this form of wear, which may lead to early fracture of the fibers. The fine fretting wear debris might initiate a strong inflammatory response from the body that can lead to bone resorption and implant loosening or the formation of very dense, even calcifying connective tissue membranes. A solid matrix fixes the fibers and prevents escape of any wear debris, so it would be expected to suppress any fretting wear problems inside the body. However, if the implant material is exposed to bone, a hard, wear resistant coating is recommended to protect the composite from the high abrasivity of bone tissue. For hip endoprosthesis stems, combined plasma sprayed hydroxyapatite—titanium have been proposed [Ha, S.-W. *et al.*].

2.4 Structural Biocompatibility

Living tissues are not static structures but are instead continuously renewed. Ageing cells in the tissue die and are quickly decomposed or removed, to be replaced by new cells. A basic example of this process is the human skin where dead skin cells on the exterior of the skin are removed by mechanical abrasion while below the skin surface, new cells gather. The balance between renewal and removal is sensitive to external demands and most tissues are capable of adaptive change. The skin provides another example, when a person, who is unused to manual work, begins to use a shovel or other digging implement, the skin on the hands will become thick and coarse. When the same person ceases to use the shovel, the skin on the hands will soften and revert to its earlier state. A similar process applies to bone tissue. Bone, although a large portion of its volume is filled with an extra-cellular matrix of bone material, is permeated with living cells — the osteoblasts and osteoclasts. Osteoblasts serve to build-up the extra-cellular matrix of bone material, while osteoclasts have the function of resorbing the extra-cellular matrix. Hormones and other control agents enter the bone to establish a balance between formation of bone material by the osteoblasts and bone resorption by the osteoclasts. This was first observed by Wolff who described that the orientation of the trabecular architecture in cancellous bone reacts on changes in the loading regime by rearrangement and adaptation of density. In Wolff's law of bone remodeling, the trabecular architecture was described to align to the main stresses, thus minimizing shear stresses in the material.

During a normal active lifestyle, there is sufficient mechanical load on the bone to ensure that the balance between formation and decomposition of bone material is biased towards retention of bone material. If a bone that is normally subject to heavy loads, such as a leg bone becomes unloaded, then the extra-cellular matrix inside the bone will be partially resorbed. The classic example of this process is bone loss in astronauts after prolonged periods of weightlessness. The dynamic balance between bone strengthening and bone resorption, as defined by Wolff's law is illustrated schematically in Fig. 2.3.

When an orthopaedic implant is placed inside a bone, there are local changes in stress levels at any point in the bone. The loads on the bone, such loads during walking, remain the same, but mechanical stresses in the bone adjacent to the implant will be altered. According to Wollf's Law, this occurs by remodeling of the bone architecture and density. In some cases, the local stress rather than the overall load on the bone appears to be the controlling factor on the retention of bone material. Thus, a significant problem with implants is believed to occur when a normally highly stressed section of bone becomes largely unstressed. This section of bone may progressively atrophy with a reduction in bone mass and increasing mechanical weakness. This phenomenon is termed 'stress shielding'. An implant, which minimizes

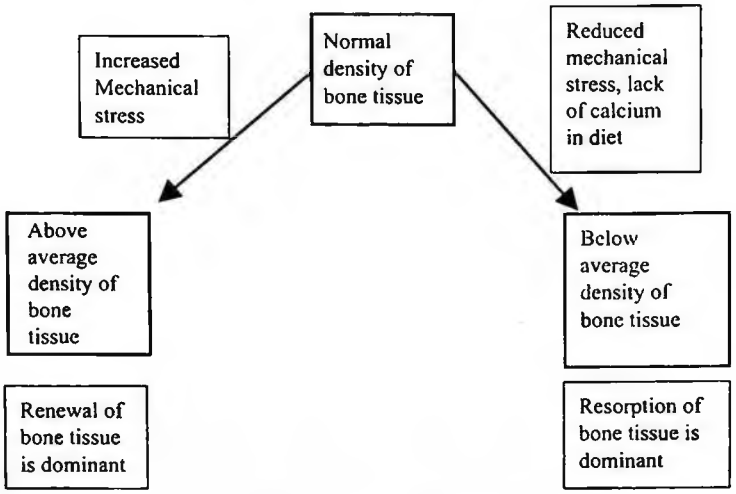


Fig. 2.3 Dynamics of the balance between bone tissue growth and resorption.

this process of bone atrophy (or resorption) can be said to be structurally biocompatible. As is discussed in later chapters, the main cause of structural bio-incompatibility is that most orthopaedic implant materials are much stiffer than bone and secondarily, that because of the isotropy and high stiffness, these implants can cause a high degree of strain mismatch in the interface to bone. This can hinder proper bone bonding of the implant in the long term.

2.5 Example of Biocompatible Implants

A type of implant that is used in very large numbers is the orthopaedic endoprosthesis for the hip or the 'artificial hip' as it is commonly known. This prosthesis has restored mobility to hundreds of thousands of patients and has an average service lifetime that reaches approximately 20 years. This level of treatment was not achieved easily. There were many problems with the initial designs and even the current designs are not without problems. Most of the difficulties relate to biocompatibility as discussed above since the prosthesis permits near normal articulation of the joint and low levels of friction comparable to the original joint. The earliest designs of prosthesis resembled a metal hinge and were relatively stiff and inflexible. A major improvement in prosthesis was achieved with the ball and socket design of prosthesis. This design closely resembled the natural hip where the ball was modeled on the head of the femur (hip bone) and the socket on the acetabulum of the pelvic girdle. A metal-to-metal combination of ball and socket resulted in a significantly high coefficient of friction in the joint. In order to reduce this friction, PTFE was substituted for the socket, resulting in excessive wear debris as discussed above. A successful combination of materials has been found to be an UHMWPE socket and a ceramic or hard metal socket. UHMWPE is the acronym for Ultra High Molecular Weight Polyethylene, a polymer with an average molecular weight in excess of 1 million that is mechanically strong and resistant to wear and corrosion. This combination of materials is found to produce the least amount of wear debris and a long service lifetime. Despite being made of an organic polymer, the UHMWPE socket is found to display only moderate amounts of creep during the service lifetime. There are still problems caused by wear debris and post-operative infection of the implant

site, which is fortunately becoming less common. The release of wear particles from the UHMWPE socket causes localized inflammation around the joint. While inflammation is a vitally important response by the body to infection, the duration of the inflammation is meant to be relatively short. The durability of the UHMWPE particles leads to prolonged inflammation and pain for the patient. In some cases of inflammation induced by wear particles, a phenomenon known as aseptic loosening may occur where the prosthesis loses its original bond to the bone. More recently, high precision metal to metal and ceramic to ceramic pairs have been introduced that show excellent wear properties. However, they cannot be considered as zero-wear systems, consequently, aseptic loosening will remain a major clinical issue.

Another major unresolved problem is the lack of structural biocompatibility since strong high modulus metal alloys are used for the stem that extends from the ball. The loads on the stem, especially bending moments, can be very high during active motion, so a strong material must be used. A major problem with strong metals is that they have a much higher elastic modulus than the bone; this prevents structural biocompatibility. A further problem with the stem is fretting wear between the stem and the cement used to fix the stem in the bone. This results in release of inflammatory cement debris and also some metal particles. Many of the metal alloys used contain elements such as cobalt, vanadium chromium or nickel, which can only be tolerated in minute quantities by the body.

2.6 Sterilization Techniques and the Testing of Biocompatibility

2.6.1 *Sterilization techniques*

Sterilization of any biomaterial is an essential step in the manufacturing process to ensure that bacteria, fungi, viruses or any other parasites do not contaminate the patient during implantation of the biomaterial. It is unfortunately not possible to obtain perfect sterility and a commonly used criterion of the quality of sterilization is the Sterility Assurance Limit or SAL [Ratner *et al.*, 1996], which is based on the probability that an implant remains non-sterile after the sterilization process. A commonly accepted value of SAL is one in a million. There are several sterilization processes

in use and under development which are listed below:

(1) Autoclaving in pressurized steam to temperatures in of approximately 120–140 degrees Celsius at 2–3 bar pressure.
(2) Immersion of the implant in an atmosphere of ethylene oxide (EtO), either pure or with carrier gases such as carbon dioxide or freon.
(3) Exposure of the implant to gamma radiation from a cobalt^{60} isotope source.
(4) Immersion of the implant in supercritical carbon dioxide fluid at pressures of approximately 20 MPa at ambient or slightly elevated temperatures.

Each method has its own particular advantages and disadvantages (Table 2.1). Steam is effective at destroying pathogens (disease causing life-form) but is unfortunately destructive to most engineering polymers

Table 2.1 Comparison of sterilization methods.

Method	Pressurized Steam	EtO	Gamma
Advantages	– simplicity – high penetration power – short processing time – lack of toxic residues	– low cost – widely used for medical devices	– high volume process – proven use for 30 years – non-toxic – no residue & quarantine – complete product penetration – produce minimal heat – dosimetric release
Disadvantages	– high temperature & pressure – cannot be used on products that are moisture sensitive or moisture impermeable	– EtO emissions – Toxic – 100% EtO highly inflammable – physical property changes in polymers due to reactivity of the gas	– discoloration of some products – brittleness of some products – Cobalt-60 is a proprietary material

since the glass transition temperature of most polymers is less than 120 degrees Celsius. Ethylene oxide efficiently kills almost all parasites, the problem is that ethylene oxide is also very toxic to human and animal life! This means that all traces of ethylene oxide have to removed from the sterilized biomaterial by prolonged 'washing' with a sterile non-toxic gas such as nitrogen. Gamma radiation from the cobalt^{60} source is highly effective at destroying pathogens and does not leave any residual radioactivity in the implant. Gamma radiation is widely used to sterilize UHMWPE orthopaedic implants as it is an efficient and rapid form of sterilization. The main difficulties with gamma ray sterilization are the high cost of the equipment and the destructive effect of radiation on polymers. The main effect of radiation is to induce cross-linking and chain scission of the polymer chains. These undesirable changes in molecular structure lead to embrittlement, which necessitates lower service loads and an inferior wear resistance inside the body [Besong *et al.*, 1998].

Supercritical fluid extraction (SFE) with carbon dioxide at elevated pressures is a new form of sterilization based on the unique ability of supercritical gas such as carbon dioxide to act as a very effective solvent and penetrating agent. SFE with carbon dioxide is observed to effectively inactive common infectious bacteria without the toxicity problems or materials degradation associated with the other sterilization methods. It is a relatively recent method and is still not widely used.

Since a biocomposite will contain at least two and possibly three or more materials, the question of the optimum sterilization method may become much more difficult than for a monolithic material. It is conceivable that new sterilization methods may be required.

2.6.2 *Testing of biocompatibility*

The testing of biocompatibility of an implant is a very detailed and time and money consuming process. National and international authorities such as the International Standards Organization, the European Union and American Food and Drug Administration (FDA) rigorously control the methods and protocols of biocompatibility testing. With the delays inherent in formalized testing and obtaining official approvals, it is estimated that the minimum time from the first proposal of the new implant material to its approval for release on the market is about 10 years. There are three levels

of testing. The first level is the *in vitro* testing of e.g. corrosion in saline fluids and strength reduction where *in vitro* means outside of a living body. At this level, basic laboratory tests are performed in a manner that is little different from any other engineering material. This level is probably the cheapest of all three levels and is widely used to exclude unsuitable materials from further testing. Despite being simpler to perform than the higher levels of testing, *in vitro* testing still presents major technical difficulties. An example is wear testing of orthopaedic implants. The required lifetime of the implant is about 20 years before excessive wear occurs. Accelerated testing is performed, but even a 40-fold reduction test period compared to the lifetime still leaves 6 months of continuous wear testing. To provide statistically representative data of wear rates, a multi-channel wear test apparatus is used. A multi-channel wear test apparatus allows many different specimens of the same material to be tested simultaneously for the required period so that sufficient data is generated within a reasonable time. The design and operation of the multi-channel wear test apparatus presents many technical problems [Pellicciari *et al.*, 2000].

The second level is performed *in vivo* where the objective is to determine whether a material is biocompatible or bioactive and can remain serviceable over the required service period. There are two basic forms of *in vivo* testing, cell-culture tests and tests on live animals. Cell-culture tests are a convenient and highly sensitive method of testing for toxicity and carcinogenicity in the candidate material. Unfortunately, cell-culture tests are unable to provide warning of any pathological conditions involving a more complex response by an entire animal, so further testing on live animals is always required [Bollen and Svendsen, 1997].

Animals such as the guinea pig, the rat or the rabbit are used for testing, since these animals are sufficiently closely related to the human species and yet are practical to breed and care for in large numbers. The test animals are known as 'models' since they are expected to provide a reliable model of the human response to the new material. There are many tests, which include the scratch sensitization test and the intra-muscular implantation test [Bollen and Svendsen, 1997]. The scratch sensitization test involves contact between the tested material and the abraded skin of a rabbit. Allergic reaction to the candidate material is manifested by reddening of the rabbit's skin and swelling of the tissue below. In another form of sensitization test, a guinea pig is used instead. For implantation tests, a rabbit is

most often used where a block of the test material is surgically placed in the muscle adjacent to the spine. An alternative method not involving surgery is to inject a small cylinder of the test material into the muscle by using a cannula [Bollen and Svendsen, 1997]. The muscle close to the implantation site is regularly inspected and samples of the muscle may also be examined by microscope. In commercial testing, the emphasis is on rapid testing, usually the shortest period consistent with the requirements of technical standards and regulatory authorities. For research purposes, the period of testing may be considerably longer with a much wider scope of monitoring on the live animal. The health of the test animals after implantation may be assessed by e.g. urine tests to determine whether there is significant release of toxic metals. A more realistic placement site may be chosen instead of the muscle close to the backbone, such as in a limb. The test material may also be formed into prosthesis that performs the same function in the animal model as in the human patient. An example of this could be a miniature orthopaedic implant for placement inside rabbit legs. The mechanical performance of the implant material may also be directly tested during this period. An example of such a test is the pull-test on dental implants placed in the jawbone of an animal. After the specified test period has elapsed, the animals may be sacrificed (killed) to permit inspection of the implant and surrounding tissue. Microscopic analysis of e.g. the level of bonding and integration between the implant material and adjacent tissue is commonly performed. There is always an element of doubt whether an animal model can adequately reproduce the human response and much expert interpretation is required of the data. For instance, how accurately can the hopping and bouncing of a rabbit simulate the walk of an elderly human patient who requires an orthopaedic implant? While animal testing is vital for optimizing the composition of implant materials and the design of implants, some level of uncertainty about the human response cannot be excluded.

The third and final level, involves testing of a finished design of implant, e.g. an orthopaedic implant on volunteers. The purpose of this is to confirm whether the new implant does not present any unforeseen clinical problems and offers an improvement in healthcare over the existing implants and/or treatments. Long-term follow-up of the volunteers is also involved. This becomes a major issue for orthopaedic implants, which are often expected to last for more than 10 years.

2.7 Imaging of Biocomposites after Implantation

When an implant is placed in the body, it is vitally important for the surgeon to be able to confirm that it is correctly located and attached to adjacent tissue. The primary means of confirming location and attachment is to collect a diagnostic image by either radiography or magnetic resonance imaging (MRI). The conventional materials of choice for implants are metals such as titanium alloy or stainless steel. The metallic elements that comprise these metal alloys, i.e. titanium or iron and chromium have sufficiently high atomic number to be effectively opaque to X-rays. Metals such as iron are also magnetic, which leads to distortion of the images obtained by MRI. In some cases, the implant may block the MRI scan not only over the projected area of the implant but also over adjacent tissue. The strong magnetic fields involved in MRI may also cause a metal implant to be heated by induced electric currents. Human proteins are intolerant of even a few degrees in temperature rise. Imaging artifacts and imaging hazards present a major problem for the surgeon, who would prefer an implant material that is amenable to imaging. For the surgeon and other clinical personnel, difficulties in obtaining good images of the implants after surgical placement may be as significant as the long-term problems of e.g. implant wear and bone resorption. The advantages of biocomposites are that they are mostly composed of organic materials, carbon fibre and glasses. The predominant chemical elements in the composites are carbon and hydrogen with silicon and oxygen for glasses. All of these elements have low atomic weight and are non-magnetic. This means that the implants are transparent to X-rays and do not generate blocking artifact in MRI. In some cases, the biocomposites are so transparent to X-ray that a metal is added to the composite in order to generate an X-ray shadow. About 1% by weight of tungsten in the form of short fibres is often added to biocomposites for this purpose.

2.8 Summary

Biocompatibility is a measure of two basic characteristics of a material after implantation in the human body. The first characteristic, which is perhaps the most self-evident, is the durability of the material in the warm saline

and protein-rich environment of the body. The second characteristic, which enables estimation of the long-term consequences of implantation, is the interaction between the material and the body. An incompatible material will release toxic substances into the body and is not fully integrated with the surrounding tissue. A biocompatible material does not release toxic substances and is well integrated with tissue to the extent that there is minimal formation of a fibrous capsule between the implant and the enclosing tissue. The advantage of biocomposites is that a strengthening element of moderate biocompatibility can be enclosed with a matrix of very high biocompatiblity. With sufficiently advanced fabrication methods, the structure of biocomposites can be optimized to reach higher levels of performance than monolithic materials. Biocomposite materials based on carbon and other low atomic weight elements present less problems for medical diagnostic imaging after implantation than conventional metallic materials.

Despite the attractive advantages of biocomposite materials, their widespread use will have to await the completion of a full program of testing for biocompatibility. Such programs consume large amounts of capital and time and are unlikely to be performed without strong commercial interest in biocomposites.

References

Biocompatibility

D. A. Baker, R. S. Hastings and L. Pruitt, Compression and tension fatigue resistance of medical grade ultra high molecular weight polyethylene: the effect of morphology, aging and temperature, Polymer 2000, **41**, 795–808.

A. A. Besong, J. L. Tipper, E. Ingham, M. H. Stone, B. M. Wroblewski and J. Fisher, Quantitative comparison of wear debris from UHMWPE that has and has not been sterilised by gamma irradation, *Journal of Bone Joint Surgery*, 1998, **80-B**, 340–344.

T. R. Green, J. Fisher, M. Stone, B. M. Wroblewski and E. Ingham, Polyethylene particles of a 'critical size' are necessary for the induction of cytokines by macrophages *in vitro*,' *Biomaterials*, December 1998, **19**(24), 2297–2302.

S.-W. Ha, A. Gisep, H. Gruner, M. Wieland, J. Mayer, and E. Wintermantel, Topographical characterization and microstructural interface analysis of vacuum

plasma sprayed titanium and hydroxyapatite coatings on carbon fiber reinforced polyetheretherketone (PEEK), *Journal of Materials Science: Materials in Medicine*, 1997, **8**, 891–896.

R. M. Hall, A. Unsworth and B. M. Wroblewski, Frictional characterisation of explanted artificial hip prostheses, *Wear*, 1994, **175**, 159–166.

Larry L. Hench, The story of Bioglass: from concept to clinic, in *Imperial College Inaugural Lectures in Materials Science and Materials Engineering*, ed. Don W. Pashley, Imperial College Press, 1981, pp. 199–299.

D. F. Williams, Review-tissue biomaterial interactions, *Journal of Materials Science*, 1987, **22**, 3421–3445.

Sterilization Procedures

T. Johnson and D. Devanathan, Nitrogen packaging and gamma radiation sterilization of UHMWPE, Research Laboratories of Zimmer Inc., Warsaw, Indiana, 1996, http://www.zimmer.com/images/nitropkg.pdf

B. D. Ratner, A. S. Hoffman, F. J. Schoen and J. E. Lemons, eds. *Biomaterials Science: An Introduction to Materials in Medicine*, Academic Press, 1996, pp. 415–420.

Testing Methods and Standards

L. S. Bollen and O. Svendsen, Regulatory guidelines for biocompatibility safety testing, *Medical Plastics and Biomaterials*, May 1997, p. 16, http://www.devicelink.com/mpb/archive/97/05/001.html

Biological evaluation of medical devices, ISO 10993 Standard Series, Geneva, International Organization for Standardization.

Council Directive 93/42/EEC of 14 June 1993 concerning Medical Devices, Official Journal of the European Communities, **36**, July 1993.

M. Pewllicciari, G. Barbanti and A. O. Andrisano, Design and development of a multi-axis hip joint simulator, *Proceedings of the. 10th International Conference on Biomedical Engineering*, 6–9 Dec. 2000, Singapore, ed. J.C.H. Goh, pp. 159–160, publ. National University of Singapore, 10 Kent Ridge Crescent, Singapore.

Artifacts during Imaging of Implants

N. Augustiny, G. K. v. Schulthess, D. Meier and P. Bösiger, MR imaging of large nonferromagnetic metallic implants at 1.5 T, *Journal of Computer Assisted Tomography*, July-August 1987, **11**(4), 678–683.

P. L. Davis, L. Crooks and M. Arakawa, Potential hazards in NMR imaging: Heating effects of changing magnetic fields and RF fields on small metallic implants, *American Journal of Radiology AJR*, 1981, **167**, 857–860.

N. J. Kagetsu and A. W. Litt, Important considerations in measurement of attraction force on metallic implants in MR imagers, *Radiology*, 1991, **179**, 505–508.

R. W. Laakman, B. Kaufmann and J. S. Han, MR imaging in patients with metallic implants, *Radiology*, 1985, **157**, 711–714.

M. A. Moscatel, F. G. Shellock and S. M. Morisoli, Biopsy needles and devices: Assessment of ferromagnetism and artefacts during exposure to 1.5 T MR systems, *Society of Magnetic Resonance*, 1995, pp. 43–48.

B. R. Rosen, P. F. New and T. J. Brady, Potential hazards and artefacts of ferromagnetic and nonferromagnetic surgical and dental materials and devices in NMR imaging, *Radiology*, 1983, **147**, 139–148.

R. Rupp, N. A. Ebraheim, E. R. Savolaine and W. T. Jackson, Magnetic resonance imaging evaluation of the spine with metal implants, *Spine*, 1993, **18**, 379–385.

F. G. Shellock, and S. M. Morisoli, *Ex vivo* evaluation of ferromagnetism and artefacts of cardiac occluders exposed to a 1.5 T MR system, *Journal of Magnetic Resonance Imaging*, 1994, **4**, 213–215.

F. G. Shellock, S. M. Morisoli and E. Kanal, MR procedures and biomedical implants, materials and devices, *Radiology*, 1993, **189**, 587–599.

Chapter 3

CONSTITUENT, FABRICATION, AND CHARACTERIZATION

As aforementioned, synthetic composites are generally made from two constituent materials, a reinforcing phase and a continuous (matrix) phase. Presented in this chapter are typical constituent materials suitable for making biocomposites, some commonly used fabrication techniques, and selected testing methods for characterizing mechanical properties of composite materials. As surface treatments are commonly applied to reinforcement fibers, it is generally recommended that biocompatibility tests be performed for the composite biomedical devices to ensure that no toxic/harmful sizing agent has been involved. Furthermore, a composite processing may also give rise to some changes in the material especially biological properties.

3.1 Reinforcement Materials

A variety of fibers and particulates which are biocompatible can be used as reinforcements in the fabrication of biocomposites. Some of them are summarized below:

3.1.1 *Carbon fibers*

Carbon (also referred as graphite) fibers are currently the most widely used fiber material in the development of biocomposites due to their biocompatibility, high inertness, and potential mechanical properties such as high modulus and strength and good resistance to fatigue and moisture absorption. Many primary-load carrying medical devices intended for use inside

and outside the body have been fabricated using carbon fiber reinforced composites, such as total hip (knee) replacement stems, internal fixators (bone plates), screws and nails, spine rods, dental posts, external fixator components, and prosthetic sockets. Carbon fibers are mainly produced from precursors of polyacrylonitrile (PAN), pitch and rayon. Different precursor results in different properties of the carbon fiber. Pitch precursor based carbon fibers possess good stiffness, whereas the PAN based carbon fibers offer very high strength. Some fundamental properties of carbon fibers from these two precursors are listed in Table 3.1. The main drawback of carbon fibers for biocomposites is in their color — black.

To aid bonding of carbon fibers to a matrix material in the composite, a surface treatment is performed. Surface treatment of carbon fibers can be grouped into oxidative and non-oxidative treatments. The principal effects of fiber surface treatments are to enhance the interlaminar shear strength and tensile/flexural strength of the composites, while a loss in the impact fracture toughness is usually experienced depending on the treatment level. These changes in the mechanical properties are attributed to the improved interface bond quality via the following modifications of the fiber interface: (a) increased fiber surface area by promoting mechanical anchoring at the interface region, and (b) removal of weak surface layer from the fiber and functionalizes the fiber surface to enchance the bonding with the matrix. The carbon fibers are supplied in yarns denoted by 3k, 6k, 12k etc., where e.g. 3k indicates that the number of individual fiber filaments in the yarn is 3,000.

3.1.2 *Glass fibers*

Depending on their chemical compositions, glass fibers are provided in different grades, i.e. *A*, *C*, *E*, and *S*. '*A*' refers to alkali glass. '*C*' represents corrosion resistant glass mainly used in chemical industry. The designation '*E*' is for electrical since *E*-glass is a good electrical insulator besides having good strength and a moderate Young's modulus. *S*-glass fiber is of high strength initially developed for military applications, and can withstand higher temperature than other glass-fibers. Its modulus is about 20% greater than that of *E*-glass fiber, and is also stronger and tougher. Its creep rupture resistance is significantly better. *S*-glass fibers have been used in biomedical applications. Their properties are given in Table 3.1. Except for '*S*' grade,

Table 3.1 Properties of reinforcement fibers [Pilato and Michno, 1994; Mark, 1999].

Fiber	Diameter (micron)	Density (kg/m^3)	Longitudinal Tensile Modulus (GPa)	Longitudinal Tensile Strength (GPa)	Longitudinal Compressive Strength (GPa)	Elongation (%)
Aramid (Kevlar 49)	12	1440	60–200	2.6	0.34–0.48	3.8
Carbon, PAN based	7–8	1700–1900	230–830	2.3–7.1	1.05–2.75	1.5–2.4
Carbon, Pitch based	7–8	1600–2200	38–820	0.8–2.3	0.48	0.25–0.50
Glass (S-grade)	3–20	2490	85	4.6	1.1	5.7
UHMWPE (Spectra 1000)	27	970	175	3.0	0.17	2.7
PET (Dacron)	10–12	1310–1380	1.2–6.5	0.510–1.12		8–42
PTFE (Gore-Tex)	10–12	2100–2200	0.3–0.7	1.859		33
PU (Lycra)		1020–1280				
PGA (Dexon)	15–25	1500–1640	7–14	0.76–0.92		15–35
PLLA	15–25	1248–1290	8.5–9.2	0.87–0.9		25

E-glass fibers are also widely used in biocomposite fabrication, which have a typical tensile and compressive strengths of about 2 GPa and 1.5 GPa respectively and a tensile modulus of 74 GPa [Soden *et al.*, 1998]. The diameter of a glass fiber filament is between 3 and 20 μm, and mostly is 12 to 14 μm. The surfaces of glass fibers are normally treated with sizing materials, most commonly silane agents, immediately after processing.

Glass fibers are much cheaper but less suitable than carbon fibers and hence seldom used inside the body. However, glass fibers do have some superiority, i.e. they are transparent. When impregnated with proper matrix materials, they can be best used for applications where esthetic appearance is of importance and stiffness is not of a critical issue, such as used for the development of orthodontic archwires and brackets. Researchers have also developed fully resorbable composites using absorbable polymer matrices and absorbable calcium phosphate glass fibers [Lin, 1986; Zimmerman *et al.*, 1991]. This would provide a possibility to fabricate fully resorbable medical implant devices for the body. However, the absorbable glass fibers have been found to have inferior mechanical properties, with a modulus of about 48 GPa and a tensile strength of approximately 500 MPa [Lin, 1986].

3.1.3 *Aramid fibers*

Aramid fibers are made from polyimide and polyamide which are polymers. Most polymer fibers are not stiff and strong enough to be used as reinforcement in composite fabrication. However, there are some exceptions. One is aramid fibers, which have the general structure of aromatic rings alternating with amide linkages (aromatic amide). The commercial names of aramid fibers include Kevlar (Dupont, USA), Twaron (AKZO, Europe), and Technora (Teijin, Japan). They are produced in several grades (e.g. Kevlar 29, Kevlar 49, and Kevlar 149), by changing the molecular structure. They offer high toughness and stiffness properties, and display superior properties in tension but poor in compression. Some properties of aramid fibers are listed in Table 3.1. The aramid fibers are ductile and able to fracture by splitting into small fibrils. They are susceptible to visible or ultraviolet light, which results in discoloration with accompanying loss of mechanical properties. Water absorption of aramid fibers varies from 1.1 to 7 wt%. These fibers have relatively high usage temperature (i.e. stable up to

300°C in the absence of hydrolysis agents) and very low creep, in addition to the high specific strength. The linear density of aramid fibers is specified using 'denier' system. Denier is equivalent numerically to the number of grams per 9000 meters in length. Aramid fibers, like most other polymer fibers, are generally poor in bonding with polymer matrices. Surface treatments using plasma [Li *et al.*, 1997] or other chemical media are necessary. At present, aramid fiber reinforced biocomposites are mainly investigated in a laboratory research level.

3.1.4 *Other polymer fibers*

Another exception is UHMWPE (ultra-high-molecular-weight-polyethylene) fibers, which are produced via a gel-spinning process in which a low concentration solution of ultra high molecular weight ($M > 2 \times 10^6$) polyethylene is extruded to form a gel precursor fiber. This precursor fiber is subsequently hot drawn to produce a very highly oriented molecular structure. Typical properties of UHMWPE fibers are given in Table 3.1. They are the lightest fibers with a density of only 0.97 g/cm^3. They are chemically inert, are particularly resistant to alkali, moisture, and UV environments, and display good abrasion resistance and radiolucency. The main drawbacks are poor creep resistance and matrix compatibility. Bulk UHMWPE exhibits excellent biocompatibility. However, there are preliminary data demonstrating a less favorable response to UHMWPE fibers [Shieh *et al.*, 1990]. Questions are always raised regarding whether bulk and fiber properties are equated. Although in theory the basic materials should be the same, differences associated with surface characteristics and with different manufacturing and processing can be significant.

Other polymer fibers such as PET, PTFE, PU, PGA, and PLLA are also used for biocomposite reinforcement purposes. However, they are used mainly for their other characteristics such as biostability and bioabsorbability but not for their mechanical superiority. They are produced by spinning processes which are considered in three major categories, i.e. melt, dry, and wet spinnings. Depending on the spinneret shape and the drawing conditions used, fibers with different rugosity and cross-sectional shapes can be obtained. Absorbable polymer (e.g. PGA, PLA, PLLA) fibers have been

Table 3.2 Properties of other types of reinforcements [Black and Hastings, 1998].

Particulate	Density (kg/m^3)	Tensile Modulus (GPa)	Tensile Strength (GPa)	Compressive Strength (GPa)	Fracture Toughness ($MPa\ m^{1/2}$)
Alumina	3980	366–380	0.31	3–3.8	4
Silica	2480	74		1.0	0.7
Bioglass	2660	35	0.042		
Hydroxyapatite (HA)	3160	80–110		0.5–1.0	1.0

used to reinforce absorbable polymer matrices to make fully absorbable fracture fixator system [Vert *et al.*, 1986].

3.1.5 *Other types of reinforcements*

It is noted that all of the above mentioned fibers used in practice can be continuous (long fibers) or discontinuous (short or chopped fibers). In addition to the fiber materials, there is another type of reinforcements, i.e. particulates. Until recently, only ceramic particulates have been used to fabricate composite biomedical devices/components. Commonly used ceramics are alumina, silica, hydroxyapatite, and bioglass, of which typical mechanical properties are summarized in Table 3.2.

3.2 Reinforcement Forms

Continuous fibers can be made into different kinds of fibrous preforms to serve as reinforcement. The simplest preform is arranged unidirectionally. Namely, fibers are gathered together and are placed in the same direction. In such a case, the direction along the fiber axis is called longitudinal. Similarly, we have a transverse direction (transverse to the fiber axis). For clarity, a group (bundle) of fibers is called a yarn, tow, or strand. In some cases, a group of yarns may be used and it is called roving. The mechanical properties of the resulting composite are much higher in the longitudinal than in the transverse direction. In most applications, the unidirectional composites cannot meet the purposes. The composites from different fibrous preform reinforcements will display different properties. Thus, there is another variable which can significantly affect the mechanical properties

of the composites. That is, the fiber (yarn) orientation. In order to make full use of the load carrying ability of the reinforcement phase, composites are fabricated by positioning continuous fibers or yarns in the directions of applied stress. This may be achieved by using techniques such as filament winding or lamination methods. The fibers must be positioned accurately, to obtain the desired mechanical properties. For example, a deviation of 10° lead to a 65% decrease in stiffness of the composite. These composite structures are also vulnerable to splitting and delamination, i.e. poor intra- and inter-laminar properties, due to the lack of entanglement among filament wound or laminated fibers. To overcome these problems, textile preform reinforcements were introduced in 1970s, which are fabricated from continuous yarns by weaving, knitting, or braiding techniques. Typical such preforms are highlighted below. When all of the yarns essentially run in a same plane, the resulting preform is named two-dimensional (2D). If, in additional, there are yarns that go in the direction perpendicular to the plane, it is a 3D predorm. The handling of these fabrics is much easier than that of individual fibers or yarns.

3.2.1 *Woven preform*

Woven fabrics are made by interlacing yarns in mutually orthogonal directions (Fig. 3.1) using weaving looms. The yarns running along the length of the fabric are called 'warp yarns' and those orthogonally interlacing the warp yarns are named 'fill or weft yarns'. The longitudinal direction of the fabric is called warp and the transverse direction is called weft or fill. The frequency of yarns interlacing can be controlled, which results in different weave geometries as illustrated in Fig. 3.1. The waviness of yarn due to interlacing is called 'crimp' or 'fabric crimp'. Lower crimp means straighter fibers, and hence will give better mechanical properties for the composite reinforced by the fabric. Satin fabric has lower crimp compared to plain weave. The drawback of lower crimp is in reduced fabric integrity. Namely, yarns may easily move during handling. The gaps between the adjacent yarns (called inter-yarn gaps) are controlled by the amount of beating (battening) given to the fill yarn after interlacing the warp yarns. Both the weave geometry (including fabric crimp) and the inter-yarn gap influence the mechanical properties of the resulting composite material.

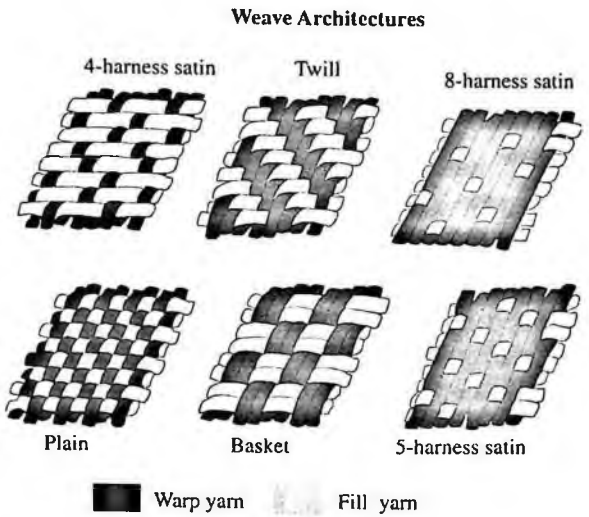


Fig. 3.1 Various 2D weave fabric structures.

Simple weaving looms produce wide and straight (flat) fabrics with two sets of yarns in the warp (0°) and weft (90°) directions. With specialized machinery, it is also possible to produce triaxial weaves with three sets of yarns, which intersect and interlace with one another at angles in the range from 30° to 60°. The woven fabrics can be used to make biocomposites for bandage, socket, bone plate etc., applications.

3.2.2 *Braided preform*

Braided preforms are essentially produced by intertwining warp and fill yarns, altogether along the braiding direction over a cylindrical mandrel (Fig. 3.2). There are two kinds of basic braiding patterns, i.e. the diamond (1×1) and regular (2×2) braids as schematically shown in Figs. 3.3(a) and 3.3(b), which can be made into either a flat (on a flat machine) or a tubular (on a tubular machine) form. The diamond braided fabric has a 1×1 intersection repeat pattern, in which a fiber bundle passes over and under another fiber bundles with a braiding angle (Fig. 3.3(a)). The preform condition of a braided fabric is determined by the combination of a spindle rotation and a taking up movement of the braiding machine, as shown in Fig. 3.2. The spindle moves along the orbit of the braiding machine, and all yarns are

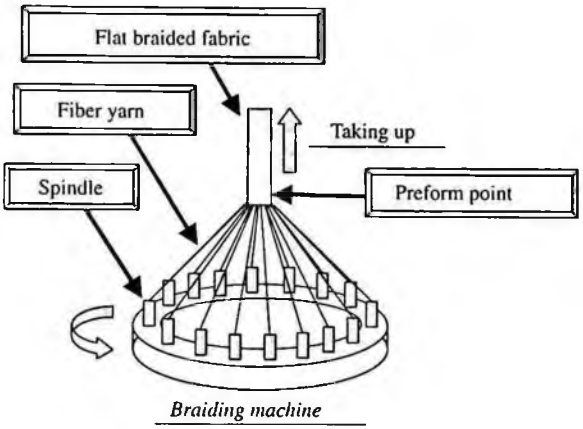


Fig. 3.2 Spindle rotation and taking-up movement of a flat braiding machine.

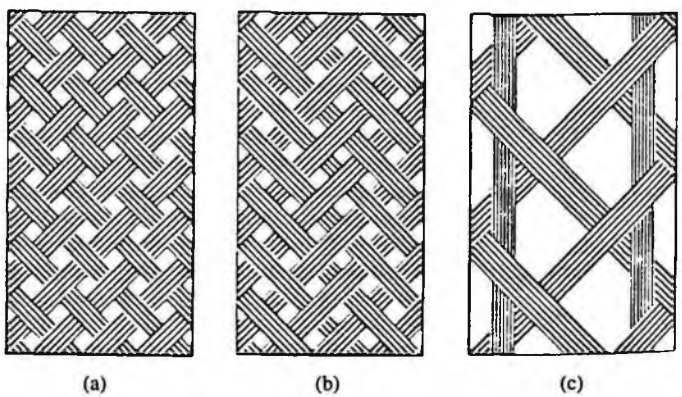


Fig. 3.3 Schematic diagrams of (a) a plain diamond braid, (b) a plain regular braid, and (c) a plain diamond braid with axially interlaid yarns. [After Chou, 1992]

gathered at the preform point, where a braided fabric is formed with a prescribed braiding angle. The regular braided fabric has a 2 × 2 intersection repeat pattern, in which a fiber bundle passes over and under two fiber bundles with a braiding angle (Fig. 3.3(b)). In contrast to the weaving preforms, a third set of yarns, called axial yarns, are also possible to be interlaid in a braided fabric (Fig. 3.3(c)). The interlacing angles between yarns can be varied from 20° to 160°. The bias interlacing nature of the braided fabrics makes them highly conformable and shear resistant. Tubular braids are

normally formed over an axial-symmetric mandrel, which determines the final shape of the braid. Tubular braids of diameters from 1 mm to 300 mm are possible. Separate machines are required to make different dimensions. Flat braiding is a variation of tubular braiding process. It is also possible to introduce 0° yarns to enhance the reinforcement in the 0° direction.

The braided fabrics are widely used as reinforcements to make biomedical components such as socket (tubular braid), dental post (tubular braid), bone plate (flat braid), blood vessel (tubular braid), and so on.

3.2.3 *Knitted preform*

Knitted fabrics (Fig. 3.4) are created by intermeshing loops of yarns using knitting needles [Gohl and Vilensky, 1991]. Depending on the direction in which the loops are formed, knitted fabrics may be broadly categorized into two types—warp knitted fabrics (Fig. 3.4(a)) and weft knitted fabrics (Fig. 3.4(b)). By controlling the loop (stitch) geometry and density, a wide variety of knitted fabrics can be produced. Because of the looped structure, knitted fabrics generally possess more hole (porous) areas than woven or braided fabrics, which translate to lower composite mechanical properties. On the other hand, this looped structure makes knitted fabrics to be more flexible than woven or braided fabrics. In general, weft knitted fabrics are less stable and hence, stretch and distort more easily than warp knitted fabrics so that they are also more formable. In order to enhance the mechanical properties, straight yarns can be integrated into the knit loops (Fig. 3.5).

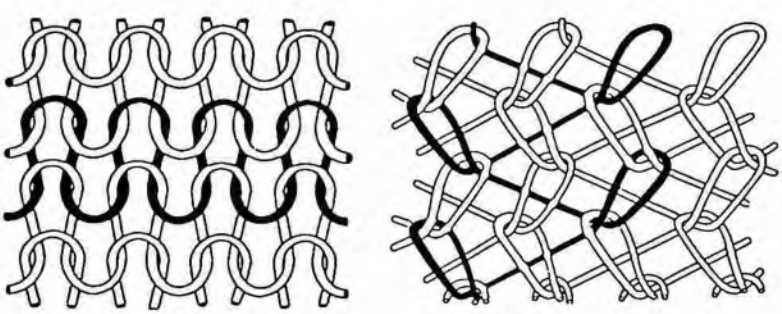


Fig. 3.4 Schematic diagrams of (a) weft knit structure (b) warp knit structure.

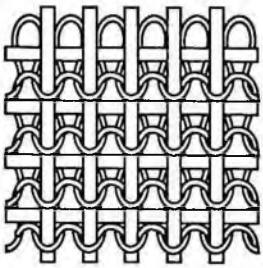


Fig. 3.5 Schematic diagram of a weft knit with interlaid yarns in two directions. [After Ko, 1989]

In this way, fabric can be tailored for stability in certain directions and conformability in the other directions. Knitted fabrics can be used for making composite bone plate, external fixator ring, and other medical devices.

3.2.4 *Non-woven preform*

Non-woven fabrics are processed from fibers by chemical (binder), electronic (spinning), thermal or mechanical (needle punching and stitching) bonding or a combination of the above. In these structures, the orientation of the fibers may range from highly regular to completely random. The maximum fiber volume fraction of composites reinforced with these fabrics is limited to about 30%.

3.3 Matrix Materials

There are very many polymers which can be used as matrix materials in fabricating various biocomposites. Polymers are very large molecules (macromolecules) that consist of a number of small repeating units called 'mers'. Variation of the 'mers' in terms of chemistry, arrangement, and combination result in polymers with a large variety of properties. The best examples of polymers are within our body. The deoxyribonucleic acid (DNA), the genetic material of all living organisms is a polymer. The important components of tissues, collagen, elastin, and fibrin are also polymers. The majority of polymers used in composite fabrication is synthetic polymers. Polymers can be easily processed into various forms including films, solids, fibers, foams, and coatings. Their biocompatibility and interesting range of

properties led to the widespread use of polymers in various medical applications including sutures, joint articulating surfaces, blood pumps, total artificial hearts, cardiac assist devices, dialysis equipment, catheters, and biosensors. As a polymer matrix plays an important role in a resulting composite, knowledge of its properties is required. Some typically used polymer matrices are briefly summarized in Table 3.4. These polymers can all be used as matrix materials to develop composites for bioengineering applications. The polymer properties are determined by its chemistry, molecular structure, and processing history. Table 3.5 summarizes physical and thermal properties of a number of commonly used polymers, whereas Table 3.6 lists typical mechanical properties.

Essentially, polymers can be divided into two classes according to their thermal processing behavior, i.e. thermoset and thermoplastic polymers. Once processed, a thermoset polymer cannot be softened again at a high temperature. On the other hand, a thermoplastic polymer can be softened and processed to required forms many times upon being heated.

3.4 Fabrication

A number of techniques have been developed to make polymer based composite materials over the years. Some of them are suited for thermoplastic-based composites, some for thermoset-based composites, and a few for the both. Some methods are limited to particulate and short fiber reinforcements, whereas others are best suitable for handling continuous fiber reinforcements. Similarly, some processes make use of dry reinforcements, whereas others use fibers already combined with the matrix polymer as the raw materials (called prepregs). Presented in this section are brief descriptions for some typical methods which can be used for biocomposite fabrication or laboratory development. Additional knowledge can be found in the literature, e.g. Astrom [1997] and Gutowski [1997].

3.4.1 *Filament winding*

Filament winding (Fig. 3.6) is a process in which continuous fiber yarns are passed through a low viscosity resin bath for impregnation and then precisely wound over a rotating or stationary mandrel. Successive layers

Table 3.3 Methods of making various fabrics and integrated fiber performs.

Textile Process	Preform Geometry	Fiber Orientation	Properties	Comment/Applications
2D Weaving	Flat fabrics	Limited to inplane warp (0°) and weft (90°) directions. In triaxial fabrics the in-plane orientation of fibers range from ±30° to ±60°.	High stiffness and strength in 0° and 90° directions. Poor in-plane shear properties in the case of bi-directional fabrics. Poor interlaminar or through- the-thickness properties.	Bi-directional woven fabrics are commonly used as reinforcements. Applications include bone plates, artificial limbs, hip joint stem, bone replacement, tibial component of knee joint prosthesis, cartilage prosthesis.
3D Weaving	Flat solid fabrics, Intergral structures, Sandwich structures.	Range of in-plane & out-of-plane (z-direction or through the thickness) fiber orientations.	Very good damage tolerance. Good in-plane and out-of-plane mechanical properties.	Bone plates, Artificial limbs, Spine instrumentation.
2D Braiding	Narrow tapes, flat fabrics Tubular fabrics	Fiber orientations in the range ±10°–80° with respect to the braiding direction. Integration of straight 0° fibers also possible.	Good tensile and shear properties. Poor compressive properties.	Tendons/ligaments, bone plates, dental post, orthodontic wires, pins, hip joint stem, spine instrumentation, catheters.

Table 3.3 (continued)

Textile Process	Preform Geometry	Fiber Orientation	Properties	Comment/Applications
3D Braiding	Thick flat fabrics Rods, thick tubular structures, and beams of various cross-sections, I, L, Z, C, and U.	Fiber orientations in the range ±10°–45° with respect to the machine direction. Integration of straight 0° fibers possible.	High structural integrity. Good tensile, in-plane shear, and flexural properties.	Artificial limbs, Bone plates, Hip joint stem.
2D Knitting	Flat fabrics	Looped fiber architecture (or stitch yarn orientation). Inserted straight fiber orientation.	Good flexibility and conformability to complex shapes. High energy absorption.	Prosthetic sockets, artificial limbs, and bone plates.
3D Knitting	Flat, thick fabrics (also called multi-axial warp knitted fabrics) Complex shapes Integral sandwich structures.	In-plane orientations range from ±30° to ±60°, and 0° and 90°.	High stiffness and strength in the direction of straight fibers. Good in-plane shear properties. Good drapability and structural integrity.	Bone plates, Artificial Limbs, Prosthetic sockets.
Non-woven	Flat fabrics or mats Thick fabrics	Random fiber orientation	Low mechanical properties	

Table 3.4 List of biomedical polymers.

Non-degradable Polymers	Biomedical Applications/Comments
Polyethylene (PE)	[-CH2CH2-] Artificial kidney blood ports, pacemaker lead insulation, catheters, dentures, and syringes. High-density PE (HDPE) is used as reinforcement in hernia repair (commercial name Marlex). Ultra-high molecular weigh PE (UHMWPE) is used in hip, knee, and shoulder joints. Low-density PE absorbs some lipids and looses strength upon implantation.
Polypropylene (PP)	Sutures, sewing ring of heart valves, ligaments, skin patches, abdominal patches, hinges of finger joint, blood oxygenation membranes, sterile bags and vessels, and syringes.
Polyurethane (PU)	Blood pumps, total artificial hearts, wound dressings, small diameter vascular grafts, heart valves, cardiac assist devices, breast implants, insulation for pacemaker leads, tympanic membranes, intervertebral discs, and penile prostheses; adhesives and dentures. Certain types of PU are susceptible to biodegradation by hydrolysis and oxidation.
Polytetrafluoro-ethylene (PTFE)	Expanded-PTFE (Gore-Tex) is used for medium size vascular grafts and sutures. Also used for auditive prostheses, and blood oxygenation membranes. Proplast (porous composite of PTFE and carbon or alumina fibers) used in reconstructive surgery. Highly biostable material. Not suitable for load bearing applications owing to low mechanical properties.
Polyvinylchloride (PVC)	Flexible blood tubing, shunts, sacs, containers, dispensers, and X-ray impermeable aprons. The plasticizers present in PVC may leach out into the body, thus leading to polymer embrittlement and provoking tissue reaction. Not recommended for implant applications.
Polyamides (PA)	Commercial grades nylon 6, nylon 6,6, and Kevlar 49. Sutures, drug delivery systems, membranes for hemodialysis, and specialist catheters. Packaging films, syringes, and nozzles. Polyamides are hygroscopic and not suitable for long term implantation.
Polymethylmetha-crylate (PMMA)	Commercial names, Perspex, and Plexiglas. Also called 'acrylics'. Widely used as bone cement (combination of PMMA powder and monomer methylmethacrylate liquid) for retention of hip prostheses. Heat generated during polymerization cause tissue damage. Intraocular lens, contact lens, and artificial teeth. Susceptible to crazing.
Polyacetal (POM)	Commercial name Derlin. Discs of mechanical heart valves.
Polycarbonate (PC)	Mainly used for general medical applications such as syringes, feeding bottles, plasma and drainage vials, and perfusion tubules. Also used in applications such as contact lens, skull implants, and haemodialysis membranes.
Polyethyleneteréph-thalate (PET)	Commercial name for PET fabrics is 'Dacron'. Vascular grafts, sutures, sewing ring for prosthetic heart valves, and tendon, ligament, laryngeal, and esophageal prostheses. PET fabrics are used for bridging large cavities in the maxillofacial area, chest wall and abdomen. Also used as medical fabrics. Susceptible to hydrolysis and loss of strength.

Table 3.4 (continued)

Non-degradable Polymers	Biomedical Applications/Comments
Polyetheretherketone (PEEK)	Used in the form of composite (reinforced with carbon fibers). Femoral stems of total hip joint replacement, internal fixation plates (bone plates) and screws for bone fractures, and spinal instrumentation (cages, plates, and screws). Excellent resistance to creep, environmental stress cracking and hydrolysis.
Polysulphone (PS)	Hollow fibers and membranes for hemodialysis. Also used as internal fracture fixation plates in the composite form (reinforced with carbon fibers).
Polysiloxane (Silicon Rubber or silicones)	Finger and toes joints, heart valve prostheses, rteriovenous shunts, blood oxygenation membranes, breast implants, artificial ventricles, wound dressings, plastic surgery, penile prostheses, intraocular lens, and vitreous humour. Lubrication of hypodermic needles and syringes, cannulas, catheters, drainage tubes, dental molds, and contact lens.
Degradable Polymers	
Poly (glycolic acid) (PGA) $[-CH_2-COO-]$	Sutures, drug delivery systems, and tissue engineering scaffolds.
PGA/PLLA Copolymer $[CH_2-COO]_m-[CH(CH_3)-COO]_n$	Sutures, staples, orthopedic fixation devices (pins, screws, and plates), vascular grafts, stents, wound dressing, periodontal membranes (dental application), drug delivery systems, and tissue engineering scaffolds.
Polydioxanone (PDS) $[-O-(CH_2)_2-OCH_2CO-]$	Sutures
Collagen	Blood oxygenation membranes, prosthetic heart valves, inner surface linings of artificial hearts.

Table 3.5 Physical and thermal properties of polymers.*

Polymer	Density (kg/m^3)	Water Absorption (%)	Glass Transition Temperature (Tg) (°C)	Softening Temperature (°C)	Melting (T_m)/ Decomposing Range (Td*) (°C)
PE	954–965	0.001–0.02	−113 to −103	40–50	125–135
PP	900–915	0.01–0.035	−30 to −3	70–100	160–180
PU	1002–1280	0.1–0.9	−73 to −23	100	180–250*
PTFE	2100–2200	0.01–0.05	20 to 22	-	322–327
PVC	1160–1700	0.04–0.75	−23 to 90	40–110	150*
PA	1020–1150	0.25–3.5	20 to 92	80–200	220–267
PMMA	1120–1200	0.1–0.4	106 to 115	70–115	~ 170*
POM	1400–1420	0.2–0.4	−13 to 75	110–163	164–175
PC	1200–1260	0.15–0.7	145	138–148	225–250
PET	1310–1380	0.06–0.3	67 to 127	70–185	245–255
PEEK	1290–1490	0.15–0.51	144	140–315	335
PS	1130–1600	0.14–0.43	167 to 230	150–216	>500*
Polysiloxane	1050–1220				

* Adopted from Black and Hastings (1998), pp. 288–323.

are laid on at a constant or varying angle until the desired thickness is attained. After curing of the part, the mandrel is removed if necessary. Sometimes (e.g. thermoplastic polymers), a hot-melt or solvent-dip process is used to impregnate the fibers. In another approach (called tow winding), thermoplastic prepreg tape is heated to the melting point of the polymer just before winding on to the mandrel. To avoid uneven cooling across the laminate thickness which may generate residual stresses, mandrel is normally heated to above the glass transition temperature of the polymer.

The process is best suited to parts of rotational symmetry (tubes and cylinders). It can generate good control of fiber orientation and higher fiber contents up to 65% by volume. Care should be exercised to avoid void formation at yarn cross-over and regions between layers with different fiber orientations. As the process uses only one-sided tooling, depending on the process control, some cases may lead to poor surface finish.

3.4.2 *Pultrusion*

Many biocomposite devices can be made through pultrusion (Fig. 3.7). It is a process that involves pulling the reinforcement through a bath of

Table 3.6 Mechanical properties of polymers*.

	Young's Modulus (GPa)	Elastic limit (MPa)	Tensile strength (MPa)	Elonation at break (%)	Compressive strength (MPa)	Poisson's ratio	Bulk Modulus (GPa)	Shear Modulus (GPa)	Hardness (MPa)	Endurance limit (MPa)	Fracture toughness (MPa $m^{1/2}$)
PP	1–1.6	20–33	21–40	100–300	30–45	0.4–0.45	1.6–2.5	0.4–0.6	60–100	11–18.2	1.7–2.1
PU	0.002–0.03	20–50	20–50	180–1050	33–50	0.47–0.49	1.5–2	0.0008^{-4}–0.003	50–120	21–30	0.1–0.4
PTFE	0.3–0.7	15–30	15–40	250–550	30–60	0.44–0.47	1–2	0.11–0.24	27–90	9–18	2.5–3
PVC	1.0–3.8	23–52	10–75	10–400	32–80	0.37–0.43	3–4	0.7–1.1	70–155	13.8–31.2	1–4
PA	1.4–2.8	40–58	44–90	40–250	60–100	0.38–0.42	2.4–3.3	0.52–0.9	100–160	22–31.9	1.8–2.6
PMMA	1.8–3.3	35–70	38–80	2.5–6	45–107	0.4–0.43	3–4.8	0.6–1.2	100–220	19.3–38.5	0.8–1.3
POM	2.55–3.5	65–72	70–75	15–80	70–80	0.38–0.43	4–5.6	0.79–1	110–220	28–42	1–1.5
PC	2–2.9	53–75	56–75	80–130	100–120	0.39–0.44	2.8–4.6	0.95–1.05	110–180	29.2–41.3	2.5–3.2
PET	2.2–3.5	50–72	42–80	50–300	65–90	0.38–0.43	3–4.9	0.83–1.1	97–210	30–43.2	1.2–2
PEEK	3.6–13	12–60	70–208	1.3–50	80–120	0.38–0.43	4–4.5	1.2–1.4	100–120	33–36	2.3–2.5
PS	2.4–2.9	58–70	50–100	25–80	72–100	0.38–0.42	3.8–4.6	0.8–1	180–240	34.8–42	1.3–2
PE	0.45–1.3	20–30	30–40	130–500	30–40	0.4–0.42	0.8–2.2	0.18–0.46	60–90	13–19.6	2.2–4
UHMWPE#	0.9–2.7		40–66	140–500							
Polysiloxane	Upto 0.01	2.8–12.4	2.8–12.4	100–1200					20–80		

* Adopted from Black and Hastings (1998), pp. 288–323. # UHMWPE-ultra high molecular weight polyethylene.

Table 3.7 Comparisons between different process techniques for composite fabrication.

Process	Advantage	Drawback	Comment
Filament Winding	Best suited to parts of rotational symmetry (tubes and cylinders); good control of fiber orientation; higher fiber contents up to 65% by volume.	As the process uses only one-sided tooling, depending on the process control, some cases may lead to poor surface finish.	Care should be exercised to avoid void formation at yarn cross-over and regions between layers with different fiber orientations.
Pultrusion	Best suited for making parts with constant cross-section (rods, tubes, pipes, beams, angles, and sheets) in large quantities; very good fiber alignment; fiber volume fractions as high as 60% can be obtained.		Reinforcements in different forms such as unidirectional fibers (rovings), continuous strand mat, braided, woven, and stitched fabrics can be used as feedstock.
Extrusion		Limited to particulate and short fiber reinforcements with sections of uniform cross-section. The capital cost is high.	Typical reinforcement contents are in the range of 10 to 30% by volume.
Injection Molding	Capable of mass producing complicated parts to very accurately controlled dimensions.	Expensive equipment; limited to particulate or short fiber reinforcement.	Typical reinforcement content in the range of 10 to 30% by volume.
Compression Molding	Suitable for making both thermoset and thermoplastic composites.	Extremely intricate parts containing undercuts, side draws, small holes, and delicate inserts with close tolerances are difficult to produce.	Quality determined by the mold design, molding temperature cycle, and the application of pressure in the correct sequence. It is essential that mold be adequately vented to allow water vapor and other volatiles to escape during the curing/consolidation process.
Thermoforming	Low tooling costs; possible to produce thin and large components; typical fiber volume fraction in the range of 35% to 60%.	Mainly suitable for producing thermoplastic composite components. Components of simple shapes.	Different configurations can be made readily. Improvements in the technique allow producing deeply drawn and geometrically complex components.

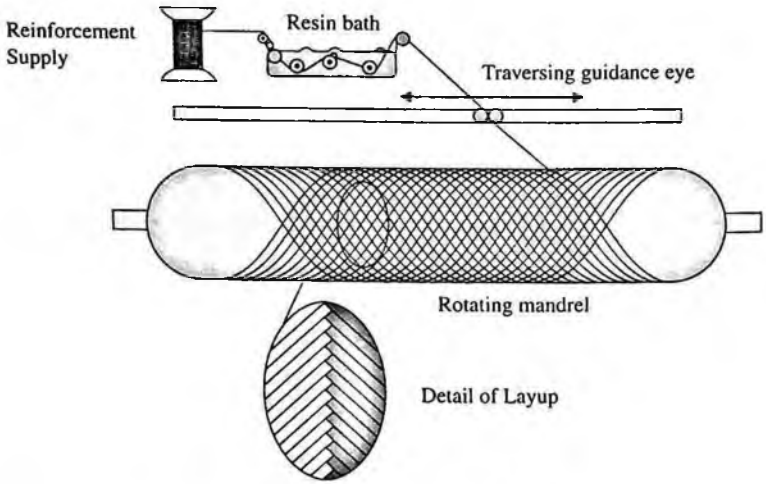


Fig. 3.6 A schematic of filament winding.

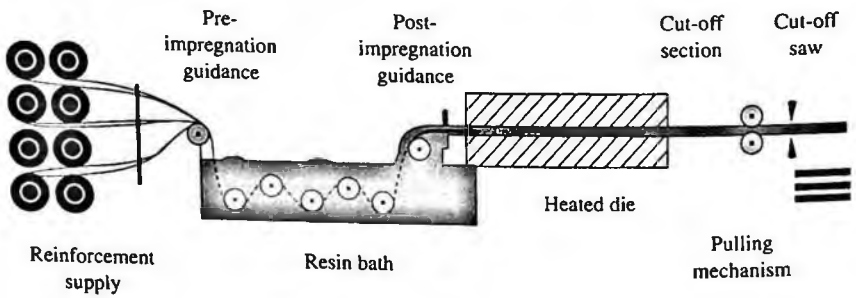


Fig. 3.7 A schematic show of pultrusion.

liquid thermosetting resin and then directly and continuously through a heated die to produce a continuous section. While passing through the bath, the reinforcement is properly impregnated with the resin. The die has a constant cross-section cavity throughout most of its length, except at the tapered entrance, which is designed to squeeze out the excess resin from the reinforcement. The heated die permits curing of thermosetting resin and determines the cross-sectional shape. Subsequently, the hot solid is cooled and cut to the required lengths. In some special cases, prepregs are also pultruded to make good quality components.

The process can also be used to produce thermoplastic composites. In such a case, the feedstock (prepreg) possesses both the reinforcement fibers and matrix polymer in pre-consolidated or non-preconsolidated form (e.g. a commingled yarn containing both reinforcement and thermoplastic polymer matrix fibers). A preheating system is used to heat the feedstock to a temperature near or in excess of the softening point of the matrix. The feedstock then enters a heated die, which melts the matrix polymer and determines the cross-sectional shape of the composite. Subsequently, the composite is consolidated in a cooling die, pulled out, and cut to the desired lengths.

The pultrusion is best suited for making parts with constant cross-section (rods, tubes, pipes, beams, angles, and sheets) in large quantities. Very good fiber alignment and fiber volume fractions as high as 60% can be obtained. However, there are some limitations on the fiber orientations that can be achieved. The reinforcements in different forms such as unidirectional fibers (rovings), continuous strand mat, braided, woven, and stitched fabrics can be used as feedstock.

3.4.3 *Extrusion*

An extrusion machine (Fig. 3.8) mainly consists of rotating screws in a heated barrel. At one end of the barrel, a die (die cavity designed based on the desired cross-sectional geometry of the component) is attached. The feedstock (pellets, combined form of polymer matrix and reinforcement) is

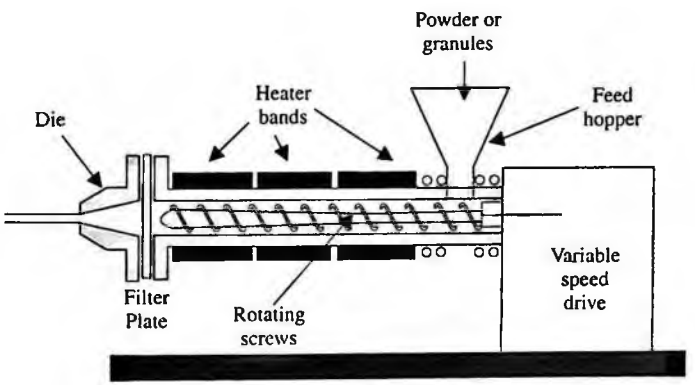


Fig. 3.8 A schematic show of an extrusion.

fed from a hopper at the other end of the barrel. The feedstock is mixed and heated to plasticity and then passed through the die. The extruded product is cooled and cut to the desired dimensions.

This process is limited to particulate and short fiber reinforcements with sections of uniform cross-section. The capital cost is high. Typical reinforcement contents are in the range of 10 to 30% by volume.

3.4.4 *Injection molding*

In an injection molding (IM) (Fig. 3.9), the feedstock containing polymer matrix and reinforcement in a combined form is heated to plasticity in a cylindrical barrel at controlled temperature. By means of a rotating screw inside the barrel, the material is forced through a nozzle into spruces, runners, gates, and cavities of the mold. Upon solidification or crosslinking of the polymer, the mold is opened and the part ejected. This process is widely used for making thermo-plastic composites and to a lesser extent for thermoset composites.

Injection molding equipment is expensive. The process is limited to particulate or short fiber reinforcement. The severe shearing action of the

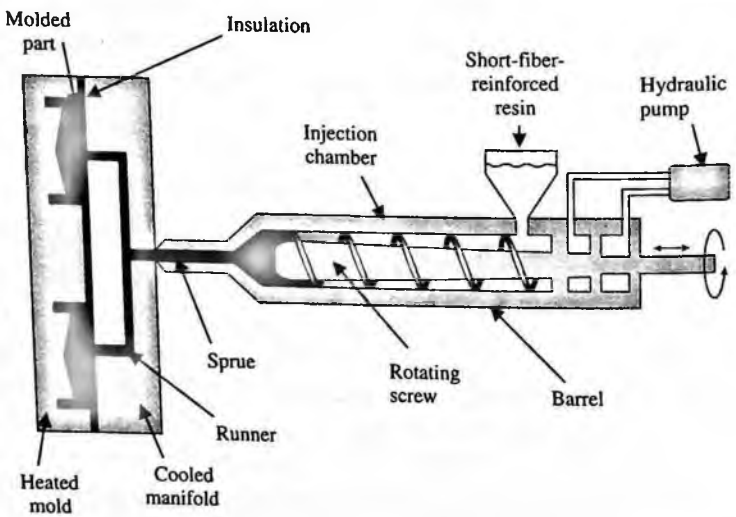


Fig. 3.9 A schematic diagram of injection molding (IM).

screws may also reduce the length of reinforcement. Typical reinforcement contents are in the range of 10 to 30% by volume. The process is capable of mass producing complicated parts to very accurately controlled dimensions.

3.4.5 *Compression molding*

Compression molding, as shown in Fig. 3.10, is widely used for making composites from prepregs such as sheet molding compounds (SMC), bulk molding compound (BMC), or glass mat reinforced thermoplastic (GMT). This process uses matching male and female mold halves. A pre-weighed charge cut to the size is placed inside the mold, which is then closed, and suitable pressure and temperature are applied using a hot press. The applied temperature and pressure force the material to fill the mold cavity and facilitate polymerization (or cross-linking) and consolidation of composite material. The above mentioned prepregs contain short and randomly distributed fibers, and they readily flow to fill the mold. The mold filling ability is limited in the case of prepregs with aligned and continuous fibers (e.g. fabric prepregs). However, the technique is used widely to make flat laminates or simple shapes from fabric prepregs.

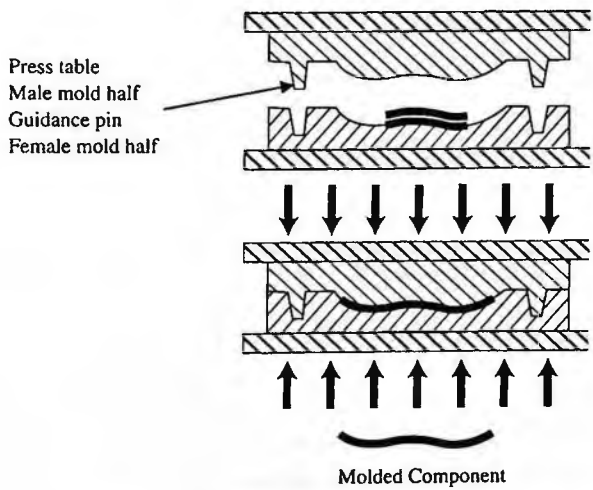


Fig. 3.10 A schematic diagram of compression molding.

The technique is suitable for making both thermoset and thermoplastic composites. Extremely intricate parts containing undercuts, side draws, small holes, and delicate inserts with close tolerances are difficult to produce. The quality of the composite is determined by the mold design, molding temperature cycle, and the application of pressure in the correct sequence. Ideally, the pressure is applied slowly as the charge softens before it starts to gel. It is also essential that mold be adequately vented to allow water vapor and other volatiles to escape during the curing/consolidation process.

3.4.6 *Thermoforming*

Thermoforming (Fig. 3.11) is a technique to transform a flat sheet of composite into a three-dimensional shape. 'Press forming or sheet forming' is the simplest such method. The composite sheet, heated to a temperature excessive of the softening point of the polymer, is squeezed into shape between two tools. Both the tools may be made of metals as in the case of compression molding or one metal and another rubber (the latter is called rubber-block molding). The rubber tool generates an even pressure and reduces the risk of wrinkles in the part. A related method 'hydroforming' uses hydraulic fluid pressure and membrane to form the composite sheet into shapes.

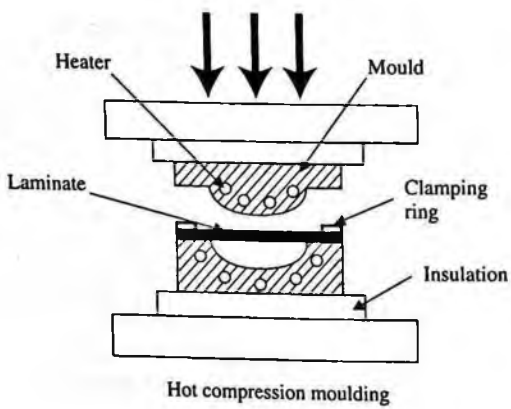


Fig. 3.11 A schematic diagram of thermoforming for composite fabrication.

A variant is the 'diaphragm forming' method, in which the stacked composite sheets are sandwiched between two diaphragms (super-plastic aluminium sheets or polyamide films). The edges of the diaphragms (not the composite sheets) are clamped to a frame and heated to a temperature above the melting point of the polymer. Pressure is applied to one side, which conforms the diaphragms to the shape of a one-sided tool. To aid the forming process, air is evacuated from between the assembly and the tool. The component is resulted after cooling of the assembly and removal of the diaphragms. This process is also carried out in purpose-built autoclaves.

3.5 Characterization

The usefulness of biocomposites is characterized by their properties. The relevant properties can be physical, mechanical, chemical, optical, electrical, etc. These properties must be quantitatively determined through experiments following some standards. Perhaps the most well known standardization organizations are ISO (International Standards Organization) and ASTM (American Society for Testing and Materials), which have documented and never stop updating many standards for the characterization of various composite properties. Summarized in this section are only some testing methods for composite mechanical properties. For other properties, the reader can refer to the relevant standards. In addition, tests for constituent properties are also highlighted.

3.5.1 *Constituent properties*

The mechanical properties of composite materials depend heavily on the relative amounts and mechanical properties of its constituents. Measurements of these quantities are described as follows:

3.5.1.1 *Fiber tests*

Due to its small diameter (from 3 μm to 20 μm), mechanical testing of single fiber is difficult and only good success has been reported with regard to tensile properties. ASTM D 3379 standard describes the testing procedure for a single filament, whereas ASTM D 4018 specifies the tensile testing of resin-impregnated fibers. In the latter case, the impregnating resin is used

to produce a rigid specimen, which is easier to handle and test than a loose bundle of fibers, and which may ensure uniform loading of the fibers in the bundle.

The majority of reported mechanical properties of reinforcement fibers has been obtained indirectly: the composite properties are measured and employed to retrieve the fiber properties using some micromechanics formulae. For example, a direct measurement of fiber properties under longitudinal compression or transverse loading does not appear to be feasible. Such properties may be inferred from matrix and composite test data.

3.5.1.2 *Neat matrix tests*

Neat matrix polymers are generally isotropic. Test procedures for these materials are well documented. Density, strength, modulus, toughness, Poisson's ratio, failure strain, and creep are some of the most important bulk properties to be evaluated for polymer matrices. These properties are determined following methods described in ASTM specifications D 792 (density), D 638 (tension), D 695 (compression), and D 2990 (creep). Test methods for measurement of other properties such as coefficient of thermal expansion, impact, creep, and fatigue response can be found in other ASTM standards published in 1995.

3.5.1.3 *Tests of fiber/matrix interface properties*

Experimental techniques for characterizing the fiber/matrix interface properties may be grouped into two major means [Mai and Kim, 1998] depending on the nature of specimens used and the scale of testing: one involves testing of single fiber microcomposites wherein individual fibers are embedded in specially made matrix blocks; and the other uses bulk composite laminates to measure the interlaminar/intralaminar properties.

Testing with bulk composite materials always has a limitation in which the exact location and modes of failure may possibly not be consistent with the underlying principles of the test. Validity of the test must be based on the actual examination of the onset of failure. In addition, both the interlaminar and intralaminar properties measured in these tests depend largely on the fiber volume content as well as on the strength of the matrix relative to the bond strength at the fiber-matrix interface. In fact, the failure may occur at

the fiber-matrix interface, in the matrix, or in a combination of the both. Therefore, they cannot inform the true values of the interface bond quality. The significance of these tests is that they provide some measure of the relative bond quality of different fiber, matrix and interface combinations.

To assess the fiber/matrix interface bond quality and strength, the test methods using microcomposites include the single fiber compression test, the fiber fragmentation test, the fiber pull-out test, the fiber push-out (or indentation) test, and the slice compression test, with a variety of specimen geometries and scales involved. In these tests, the interface bond quality is measured in terms of the interface fracture toughness, interface shear bond strength at the bonded interface, and the interface frictional shear strength which is a function of the coefficient of friction and the radial residual fiber clamping stress at the bonded or frictionally bonded interface. The fiber pull-out test is one of the most reliable and direct test methods [Mai and Kim, 1998]. It allows determination of the properties both at the bonded and debonded interfaces from the forces required to break the interface bond as well as to pull-out the fiber against the frictional resistance after complete debonding. All these micromechanical tests require a suitable theoretical model for proper evaluation of the experimental data.

3.5.1.4 *Constituent contents*

The relative amounts of the reinforcement and the matrix in a composite are determined essentially using matrix burn-off (combustion) method (ASTM D 2584), matrix digestion method (ASTM D 3171), or some microscopy method. In the microscopy method, sections are taken from the composite and polished using standard metallographic techniques. The polished surface of the cross-section is viewed under optical microscope to determine the area fraction of the fiber ends, which is assumed equal to the volume fraction. To make the measurement reliable, a number of areas of the specimen should be observed and the fiber volume fraction is obtained by averaging the respective values.

3.5.2 *Composite properties*

Over the years, a number of methods for characterizing the mechanical properties of composite materials have been made into standardizations,

mainly by ASTM. Some of the most widely used techniques are described below:

3.5.2.1 *Tensile properties*

Static tensile properties, such as tensile strength, tensile modulus, failure strain, and Poisson's ratio of flat composite materials, are determined according to ASTM D 638 and D 3039 standards, whereas tensile fatigue behaviors are obtained following D 3479. D 638 specifies a dog-bone-shaped, un-tabbed specimen which is particularly suited for evaluating the properties of discontinuous, fiber-reinforced polymer composites, but should not be used for highly oriented composite materials because of their tendency to split longitudinally at the neck-down region of the specimen. D 3039 should be utilized for highly oriented specimens and specifies a straight-sided, rectangular test specimen geometry (Fig. 3.12), with the standard primarily focusing upon the appropriate procedures for applying cyclic versus quasi-static loading. A compliant and strain-compatible material (usually aluminum or glass fiber reinforced epoxy materials) is used for making the end tabs to reduce the stress concentrations in the gripped area and thereby promote tensile failure in the gage section.

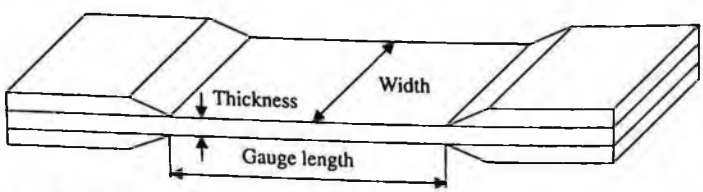


Fig. 3.12 Straight-sided composite specimen with end-tabs used for tensile test.

3.5.2.2 *Compressive properties*

Unlike metals, properties of a composite in tension and compression are not the same in general. Compression testing is one of the most difficult tests which can be performed on composites. A number of test methods together with specimen designs have been proposed, with a primary focus on the avoidance of the specimen buckling or global instability under a compressive load. In-plane static compressive properties can be determined using either ASTM D 695 or D 3410 test methods. D 3410 is recommended only

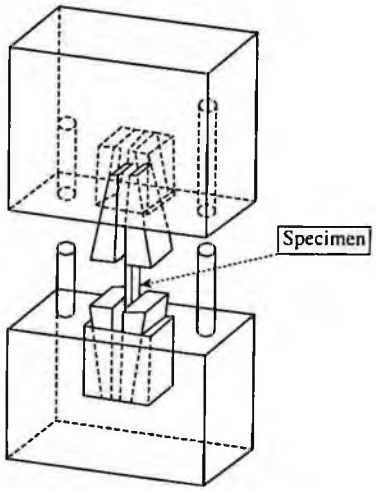


Fig. 3.13 IITRI fixture for composite compression test.

for unidirectional, or cross-plaid fiber-reinforced polymer composites (i.e. [0], [90], or [0/90] laminates), whereas D 695 can be used with any other fiber-reinforced polymer composite material. One of the most commonly used is the **IITRI** fixture (named after researchers at the Illinois Institute of Technology Research Institute), as shown in Fig. 3.13. During the test, care must be taken to ensure the specimen actually fails under uniform compression rather than bending or buckling. The best method for determining the proper state of loading that is achieved in the test is to do preliminary testing with representative samples which are strain gauged on both the front and back. These samples should then be tested to failure and the percent difference between the front and back strain gauge signals compared [Zhang *et al.*, 1996]. If the percent difference in strain ($\varepsilon_f - \varepsilon_b$) between these gauges is 10% or less, where ε_f is the strain on the front face of the coupon and ε_b is the strain on the back face, then the fixture and sample combination can be considered to be appropriate for compressive strength determination; if the percent difference is greater than 10%, excessive bending will occur during the test. This problem can usually be corrected by realigning the test fixture, using a thicker test specimen, or shortening the gauge length of the sample.

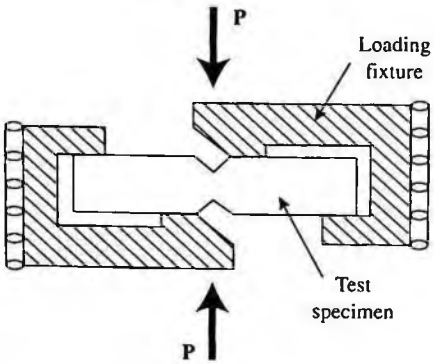


Fig. 3.14 A schematic diagram of the Iosipescu shear test.

3.5.2.3 *Shear properties*

Due to anisotropic behavior, the in-plane shear properties of a composite material are not necessarily equal to its through-thickness shear properties. A number of techniques have been developed to determine both types of shear properties of composite materials (ASTM Standards D 2344, D 3518, D 3846, and D 5379). Amongst, the Iosipescu test method (ASTM D 5379) provides a near-perfect shear stress state in the gage section and allows determination of shear properties in all three planes of the composite material. A schematic diagram of the Iosipescu shear test is shown in Fig. 3.14. The action of Iosipescu fixture is to produce pure shear loading with no bending at the mid-span section of the specimen between the notches. Any thickness specimen may be used, but a thickness of 3–4 mm is preferred. The average shear stress generated in the notched section is given by

$$\tau = P/ct$$

where P is the applied load, c is the distance between roots of notches and t is the specimen thickness. Normal strains (α°_{45}) at 45° to the specimen axis are measured by using a strain gage and shear strains are computed using $\gamma_{xy} = 2\alpha^{\circ}_{45}$

3.5.2.4 *Flexural properties*

Static and fatigue flexural properties, such as flexural strength and modulus, are determined by ASTM D 790 test method. In this test, a composite beam

specimen of rectangular cross-section is loaded in either a three-point or a four-point bending mode. In either mode, a large span (L) to thickness (t) ratio of 16, 32, 40, or 60 is usually recommended to minimize interlaminar shear deformation.

3.5.2.5 *Interlaminar fracture properties*

Delamination or interlaminar fracture represents one of the weakest failure modes in laminated composites, and is considered to be the most prevalent life-limiting crack growth mode in most composite structures. Consequently, ever-increasing attention has been devoted to the understanding and characterization of this failure mode. Delamination in composite structures seldom leads to immediate catastrophic failure. Instead, delamination occurring under in-plane loading normally induces local damages resulting in the loss of stiffness, local stress concentration and local instability, and often leads to a redistribution of stresses which could eventually promote gross failure. In general, delamination is induced from a crack driving force with a mixture of mode I (opening), mode II (forward shear) and mode III (anti-plane shear) stress intensities (Fig. 3.15). Because delamination is constrained to grow between individual plies, both interlaminar tension and shear stresses are commonly present at the delamination front. Therefore, delamination is often a mixed-mode fracture process.

Extensive research efforts [Gillespie Jr *et al.*, 1987; Smiley and Pipe, 1987; Davies *et al.*, 1992] have been devoted towards establishing a standard method using various interlaminar fracture tests. However, no widely

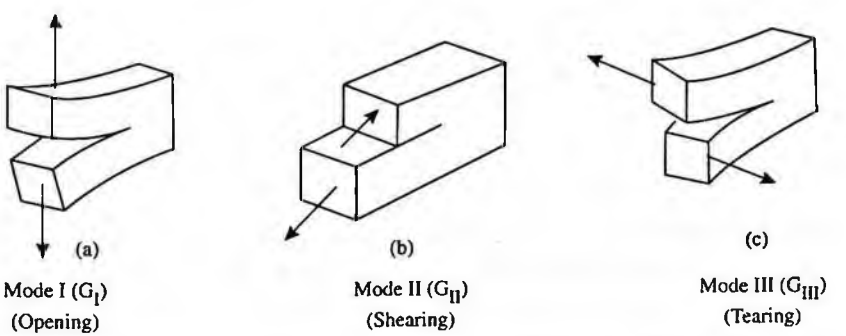


Fig. 3.15 Representative of interlaminar fracture modes.

accepted testing standard has been achieved yet [Davies *et al.*, 1992]. This may be attributed to the fact that the 'fracture mechanics approach well suited to characterizing the linear elastic fracture behavior of carbon epoxy composites, has been less amenable' when applied to thermoplastic-matrix composites.

3.5.2.6 *Test methods for creep*

Like bulk polymers, polymer matrix composite materials, depending upon their design, may undergo significant amount of permanent deformation following long-term exposure to even relatively low levels of load. This behavior may be strongly influenced by exposure to moisture and/or organic molecules such as lipids. Thus, creep behavior characterization in environments simulating *in vivo* conditions is important when designing composite-material-based medical implant devices. Tensile, compressive, and flexural creep and creep-rupture of composite materials may be determined by test methods presented in ASTM D 2290.

3.5.3 *Nondestructive evaluation*

Unlike homogeneous materials which typically fail by the initiation of distinct surface cracks which can then propagate through the material, composite materials, due to their complex heterogeneous nature, typically fail by the development of dispersed internal damage zones. These damage zones can occur within the material during processing, or can be induced by post-processing treatment. They may not be evident by visual inspection of their external surfaces. Several techniques have been developed for the nondestructive evaluation (NDE) of composite materials which allow the extent of internal damage to be monitored and evaluated, such as stiffness monitoring, ultrasonic imaging, dye-penetrate/radiography, and acoustic-emission [Henneke, 1987].

3.5.4 *Environmental considerations*

An important part of composite material evaluation for medical applications is the response of the material to the *in vivo* environment. In particular, fiber/matrix interfacial bond strength in some composite materials has been found to be significantly reduced by physiologic saline exposure [Henneke, 1987], whereas the creep resistance of some polymer matrices, such as

polysulfone, has been found to be greatly reduced by exposure to lipids [Asgian *et al.*, 1989].

The rate of diffusion and the amount of swelling in environments simulating *in vivo* conditions are relevant environmental response properties which can be evaluated in composite materials. ASTM D 5229 provides a test method for measuring through-the-thickness diffusion coefficient for moisture diffusion through thin composite plates based upon a weight-gain-versus-time method. Important parameters obtained from this test method are the percent maximum weight-gain and the out-of-place coefficient of diffusion. In this test method, plate width and length are specified to be very large compared to plate thickness to minimize errors caused by in-place moisture diffusion via exposed edges of the plate. For specimens with thickness-to-width ratios greater than that specified in ASTM D 5229, weight-gain studies are still meaningful for determining saturation weight-gain levels, and specimen edges can be sealed with protective coatings (such as those used for moisture protection in strain gauge applications) to minimize sample edge diffusion.

Because moisture diffusion can have a significant effect upon composite behavior, composite based medical implants should be fully saturated prior to conducting performance tests of implant function. The ASTM provides a composite material environmental conditioning test method developed by the F04.14 subcommittee on composites (F 1643) which describes a weight-gain-versus-time method that should be followed to properly condition composite samples prior to mechanical performance evaluation.

References

C. M. Asgian, L. N. Gilbertson, E. E. Blessing and R. D. Crowninshield, Environmentally induced fracture of polysul-fone in lipids, *Trans Society for Biomaterials*, 1989, **12**, 17.

B. T. Astrom, *Manufacturing of Polymer Composites*, Chapman & Hall, London, UK, 1997.

J. Black and G. W. Hastings (eds.), *Handbook of Biomaterials Properties*, Chapman and Hall, London, UK, 1998.

T. W. Chou, *Microstructural Design of Fiber Composites*, Cambridge University Press, Cambridge, 1992.

P. Davies *et al.*, Round-robin interlaminar fracture testing of carbon-fiber-reinforced epoxy and PEEK composites, *Composites Science and Technology*, 1992, **43**, 129–136.

J. W. Gillespie Jr, L. A. Carlsson and A. J. Smiley, Rate-dependent mode I interlaminar crack growth mechanisms in graphite/epoxy and graphite/PEEK, *Composites Science and Technology*, 1987, **28**, 1–15.

A. J. Smiley and R. B. Pipes, Rate sensitivity of mode II interlaminar fracture toughness in graphite/epoxy and graphite/PEEK composite materials, *Composites Science and Technology*, 1987, **29**, 1–15.

E. P. G. Gohl and L. D. Vilensky, *Knitting, Textile for Modern Living*, 4th Edition, Longman Cheshire, 1991.

T. G. Gutowski, *Advanced Composites Manufacturing*, John Wiley & Sons, Inc., New York, USA, 1997.

E. G. Henneke, Destructive and nondestructive tests, Composites, *ASM Engineered Materials Handbook*, Vol. 1: Metals Park, OH, ASM International, 1987, pp. 774–778.

R. Kamiya, B. A. Cheeseman, P. Popper and T. W. Chou, Some recent advances in the fabrication and design of three dimensional textile preforms: A review, *Composites Science and Technology*, 2000, **60**(1), 33–47.

F. K. Ko, Preform fiber architecture for ceramic-matrix composites, *Ceramic Bull.*, 1989, **68**, 401–414.

R. Li, L. Ye and Y.-W. Mai, Application of plasma technologies in fiber-reinforced polymer composites: A review of recent developments, *Composites Part A*, 1997, **28**, 73–86.

T. C. Lin, Totally absorbable fiber reinforced composite from internal fracture fixation devices, *Trans. Soc. Biomater.*, 1986, **9**, 166.

Y. W. Mai and J. K. Kim, *Composite Interfaces*, Elsevier Science Publishers, UK, 1998.

J. E. Mark (ed.), *Polymer Data Handbook*, Oxford University Press, UK, 1999.

L. A. Pilato and M. J. Michno, *Advanced Composite Materials*, Spinger-Verlag, New York, USA, 1994.

S. Ramakrishna, Textile scaffolds in tissue engineering, in Smart fibers, fabric and clothing, Xiaoming Tao ed., CRC, Cambridge, England, 2001, pp. 291–313.

S. J. Shieh, M. C. Zimmerman and J. R. Parson, Preliminary characterization of bioresorbable and nonresorbable synthetic fibers for the repair of soft tissue injuries, *Journal of Biomedical materials Research*, 1990, **24**(7), 789–808.

P. D. Soden, M. J. Hinton and A. S. Kaddour, Lamina properties, lay-up configurations and loading conditions for a range of fiber-reinforced composite laminates, *Composites Science and Technology*, 1998, **58**, 1011–1022.

M. Vert, P. Christel, H. Garreau, M. Audion, M. Chanavax and F. Chabot, Totally bioresorbable composites system for internal fixation of bone fractures, *Polymers in Medicine*, C. Migliaresi and L. Nicolais eds., Plenum Publ., New York, 1986, **2**, 263–275.

G. Zhang, R. A. Latour Jr. J. M. Kennedy, H. D. Schutte Jr. and R. J. Friedman, Long-term compressive strength durability of CF/PEEK composite in physiologic saline, *Biomaterials*, 1996, **17**, 781–789.

M. C. Zimmerman, H. Alexander, J. R. Parsons, and P. K. Bajpai, The design and analysis of laminated degradable composite bone plates for fracture fixation, *Hi-Tech Textiles*, T. Vigo ed., *American Chemical Society*, Washington, D. C., 1991.

Chapter 4

MECHANICS OF COMPOSITE MATERIALS

4.1 Introduction

A composite material primarily consists of two distinct materials, a fiber (reinforcement) and a matrix (binder) each with its own mechanical properties such as stiffness and strength. When the two materials are combined to make a composite, the mechanical properties depend not just on those of the two materials being mixed, but on the relative amount of each material, the shape and size of the reinforcement and its orientation with respect to the loads that are to be applied to the composite.

The study of composite materials can be divided into two levels. The first level, micromechanics, is based on the consideration of the interaction of the constituent materials at the microscopic level, i.e. determination of properties from those of the fiber and matrix. At the second level, macromechanics, the materials are assumed to be homogeneous with effective composite properties being used that are not specifically related to those of individual constituents.

The simplest of all composite materials is the one made up from identical continuous fibers, which, are all aligned in the same direction and are subjected only to loads applied parallel to the fiber direction. There are mainly two properties that are of interest, the stiffness (elastic modulus) and the strength (tensile strength).

4.2 Fiber Volume Fraction

The relative amount of each constituent is indicated through volume fraction, and is normally expressed as the ratio of volume of reinforcement and the total volume of composites. This describes the relative volume occupied by the fiber or reinforcing material in the composite. Fiber volume fraction is one of the key inputs for the calculation of all mechanical properties.

Fiber volume fraction primarily depends on the packing pattern of the fibers within the composite. The higher the fiber volume fraction, the higher the fiber controlled mechanical properties of composite materials. Based on idealized packing models, the upper limit of achievable fiber volume fraction can be established. There are two simple packing models, which can be used to establish an upper bound for the volume fraction — a square array (Fig. 4.1) and a hexagonal array (Fig. 4.2) with circular section reinforcement.

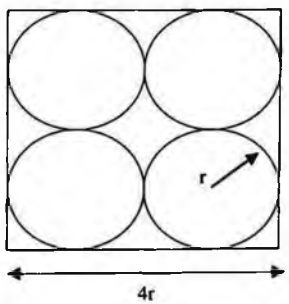


Fig. 4.1 Fibers in square array.

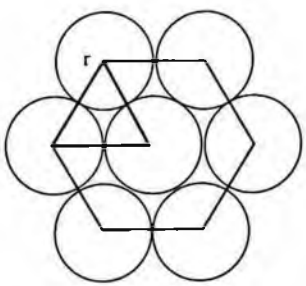


Fig. 4.2 Fibers in hexagonal array.

For a square array

$$\text{Maximum } V_f = \frac{\pi r^2}{4r^2} = 0.78 \tag{4.1}$$

For a hexagonal array

$$\text{Area of triangle cell} = r \cdot 2r \sin\frac{\pi}{6} = r^2\sqrt{3} \tag{4.2}$$

$$\text{Area of fiber section in cell} = 3 \cdot \frac{\pi r^2}{6} = \frac{\pi r^2}{2} \tag{4.3}$$

$$\text{Maximum } V_f = \frac{\pi}{2\sqrt{3}} = 0.9 \tag{4.4}$$

From the two figures, it is readily apparent that volume fractions higher than 90% are impossible and that even 78% fiber loading would be very difficult to achieve. In practice, the maximum volume fraction is around 70% in unidirectional aligned fiber composites. In woven materials, the total volume fraction rarely exceeds 50% in a given layer of cloth and so the effective fiber fraction in either the warp or weft directions is unlikely to exceed 40% for a plain weave, satin or harness weave fabric. For loosely packed fabrics such as chopped strand mat, the total volume fraction of fibers is unlikely to exceed 10% and are normally used to provide filler layers between the outer load bearing layers in a multilayer laminate.

4.3 Elastic Properties of Composite Materials

The stiffness properties of composites are strongly dependent on many parameters such as fiber volume fraction, fiber length, packing arrangement and fiber orientation. For high performance applications the reinforcement consists of continuous aligned or woven fibers in discrete layers, with typical fiber volume fractions being in the range of 50–70%. To design effectively with these systems, it is essential that the material can be characterized in terms of its thermal and mechanical properties. In structural composites, the reinforcement of high stiffness and strength provides the load-bearing constituent, whilst the matrix is generally of low modulus and modest strength.

Its contribution, however, is significant in that it acts as the medium of load transfer in the fibers as well as offering a degree of protection. In deriving expressions to predict mechanical behavior, the contribution of each constituent needs to be taken into account.

Simple determination of some of the mechanical parameters are described here. A comprehensive description can be found in the references at the end of the chapter.

4.3.1 *Unidirectional (UD) composites*

The analysis of the mechanics of composite material response can take place on a number of levels. On a micromechanical level, basic assumptions can be made regarding the nature of the interaction between constituents and expressions derived to relate the behavior of fiber and matrix directly to that of the composite. With a larger scale an assessment based on macroscopic homogeneity can be made to relate properties of a structural form to be compiled from those of individual lamina layers. Each level of analysis has its own strengths and weaknesses, the relative magnitude of which depends largely on the extent and quality of property data available for the design exercise of concern.

For a material such as a unidirectional lamina, the conditions of transverse anisotropy apply. Defining the principal axes of the composite as, '1' along the fiber direction and '2' transverse to the fiber direction as shown in Fig. 4.3. The corresponding elastic modulus along fiber direction would be (E_{11}) and transverse to fiber direction (E_{22}) and G_{21} is the shear modulus.

4.3.1.1 *Modulus of elasticity* E_{11}

When a tensile or compressive load is applied parallel to the fibers in a unidirectional laminae, it is assumed that the strains on the fiber, matrix and composite in direction 1 are the same. This is known as the iso-strain condition. Considering the composite material is in equilibrium with the application of the force, F_{11}, this force must be balanced by an equal and opposite force in the fiber, F_f and the matrix F_m (Fig. 4.4).

The force on the fibers is the stress on the fibers, σ_f, multiplied by the cross-sectional area of the fibers lying perpendicular to the stress. Similarly,

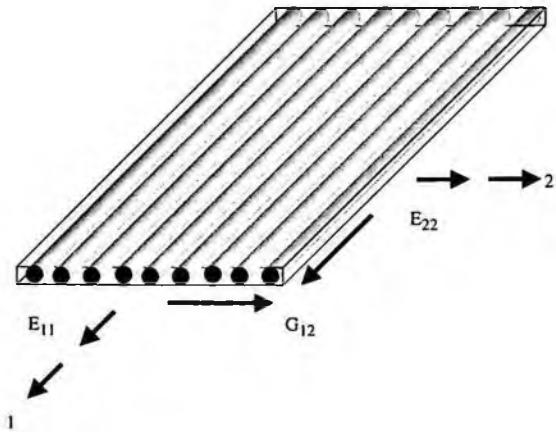


Fig. 4.3 Principal axes of UD lamina.

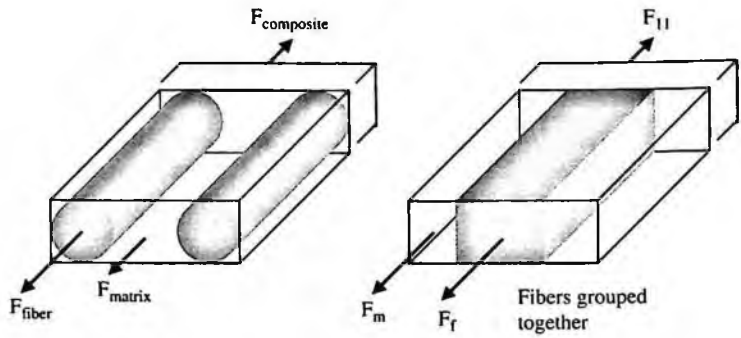


Fig. 4.4 Forces acting on composite along fiber direction.

the force on the matrix is the stress in the matrix multiplied by the cross-sectional area of the matrix in the composite. So the stress in the composite is the sum of the stresses in the fiber and the matrix multiplied by their relative cross-sectional areas (volume fractions).

$$F_{\text{composite}} = F_{\text{matrix}} + F_{\text{fiber}} \tag{4.5}$$

$$F_{11} = F_m + F_f \tag{4.6}$$

$$\sigma_{11} = \sigma_m A_m + \sigma_f A_f \tag{4.7}$$

Or

$$\sigma_{11} = \sigma_m V_m + \sigma_f V_f \tag{4.8}$$

where A is the area of the constituent, V is the volume fraction of the constituent and σ is the stress in the constituent. The subscripts m and f refer to the constituents matrix and fiber respectively.

$$\sigma_{11}/\varepsilon_{11} = E_m V_m + E_f V_f \tag{4.9}$$

where, E is the modulus of elasticity of the constituent indicated by its subscript and $\sigma_{11}/\varepsilon_{11}$ is the modulus of elasticity of composite in direction '1'.

$$E_{11} = E_m V_m + E_f V_f \tag{4.10}$$

This is the "rule of mixtures" equation.

$$E_{11} = E_m(1 - V_f) + E_f V_f \tag{4.11}$$

where E_{11} is the elastic modulus of the composite in '1' direction — the bigger this number, the stiffer the material.

Since the fiber and matrix often have different elastic moduli, the stress in each must be different. The stress is higher in the material with the higher elastic modulus (usually the fiber). In fiberglass, the elastic modulus of the glass (~75 GPa) is much greater than that of the epoxy matrix (~6 GPa). Hence when the volume fraction of fibers is increased, the elastic modulus of the composite (measured parallel to the fibers) increases linearly.

In practice, it is very difficult to get more than 70% volume of fibers. This puts a practical limit on the maximum stiffness of the composite at $0.7 \times E_f$. While the rule of mixtures has proved adequate for tensile modulus (E_{11}) in the axial direction, the isostrain rule of mixtures does not work for either the transverse modulus (E_{22}) or shear modulus (G_{12}).

4.3.1.2 *Modulus of elasticity* E_{22}

When a tensile or compressive load is applied perpendicular to the fibers in an unidirectional laminae, the composite would respond in a very different way.

In a fibrous composite with the applied stress perpendicular to the fibers, the stress is transferred to the fibers through the fiber matrix interface and both the fiber and the matrix experience the same stress. If the matrix and

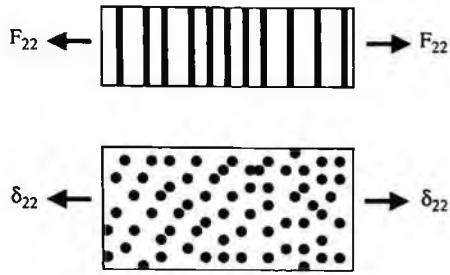


Fig. 4.5 Force acting on composite transverse to fiber direction.

fiber have different elastic properties then each will experience a different strain and the strain in the composite will be the volume average of the strain in each material. Since the stress is the same in each phase this is known as the iso-stress condition.

If a force is applied perpendicular to the fibers, the fibers and matrix will stretch in the same direction. The total deflection (δ_{22}) is just the sum of the deflections in the fiber (δ_f) and the matrix (δ_m).

$$\delta_{22} = \delta_m + \delta_f \tag{4.12}$$

$$\varepsilon_{22} = \varepsilon_m(1 - V_f) + \varepsilon_f V_f \tag{4.13}$$

$$\sigma_{22}/E_{22} = \sigma_{22}(1 - V_f)/E_m + \sigma_{22}V_f/E_f \tag{4.14}$$

Using Hooke's law to introduce the elastic modulus and since the stress is the same in both the matrix and fiber, we can get the elastic modulus (E_{22}) perpendicular to the fibers

$$E_{22} = (E_f E_m)/(E_f[1 - V_f] + E_m V_f) \tag{4.15}$$

The stiffness of the composite, measured perpendicular to the fibers increases much more slowly than stiffness measured parallel to the fibers as the volume fraction of fibers is increased. Since the properties of the composite are different in different directions, the composite is anisotropic.

4.4 Strength of Composite Materials

In simple unidirectional composites when loaded along the fibers direction, both the matrix and the fiber experience the same strain (iso-strain

condition). It is therefore expected for the composite to break at the lower of the matrix fracture strain or the fiber fracture strain. There are two cases to consider, firstly, where the matrix fails first and secondly, where the fiber fails first. The former situation is common in polymer matrix composites with low strength brittle matrices such as polyesters, epoxies or bis-melamides. The latter case is observed in metal matrix composites or thermoplastic polymer composites where, because of plastic deformation in the matrix, the failure strain of the fiber is the smaller value.

4.4.1 *Matrix triggred failure*

At low volume fractions of fibers, the matrix constitutes the major load bearing section and the addition of fibers gradually increases the strength as the applied load is divided between the fibers and the matrix. However, when the strain in the composite reaches the fracture strain of the matrix, the matrix will fail. All of the load will then transfer instantly to the fibers, which occupying a small fraction of the composite area will see a large jump in stress and they too will fail. At the strain at which the matrix is about to fracture, ε_m, the stress in the composite can be determined using Hookes' Law since both the fiber and the matrix are still behaving elastically, i.e.

$$\sigma_{11} = E_{11}\varepsilon_m \tag{4.16}$$

$$\sigma_{11} = E_f V_f \varepsilon_m + \sigma_m(1 - V_f) \tag{4.17}$$

The stress in the matrix, σ_m, is now equal to the matrix fracture stress, but the stress in the fiber is still much less that the fiber fracture stress. Before the matrix breaks, the load on the composite is

$$F_{11} = \sigma_{11}A = [E_f V_f \varepsilon_m + \sigma_m(1 - V_f)]A \tag{4.18}$$

$$\sigma_{11} = F_{11}/V_f = [E_f V_f \varepsilon_m + \sigma_m(1 - V_f)]/V_f \tag{4.19}$$

After the matrix breaks, only the fibers remain to carry the load and the stress in the fiber jumps by $(1/V_f - 1)\sigma_m$. If this increase takes the stress in the fiber above its fracture strength then the fibers too will snap. This is most likely to happen when V_f, the volume fraction of fibers is small and when the strength of the matrix is large. This is called matrix controlled fracture.

However, if the increase in stress is not sufficient to break the fibers then the load can be increased until the fibers break, i.e.

$$\sigma_{11} = \sigma_f V_f \tag{4.20}$$

This is known as the fiber controlled fracture.

Figure 4.6 shows an increase in strength of a glass fiber reinforces composite with the increase in the volume fraction of fibers. At low fiber fractions, the strength is controlled by the fracture of the matrix; the strength increases as the fibers are added. Matrix fracture strength is ~50 MPa and the failure strain is 0.010. Fiber fracture strength is ~1,200 MPa and the failure strain is 0.016. Above a fiber content of 10%, the fibers begin to dominate the fracture process and the composite can sustain high stresses. Below this, the structural integrity would be lost when the matrix fractures as the composite would be full of cracks. Then effective strength of the composite is given by the matrix controlled strength.

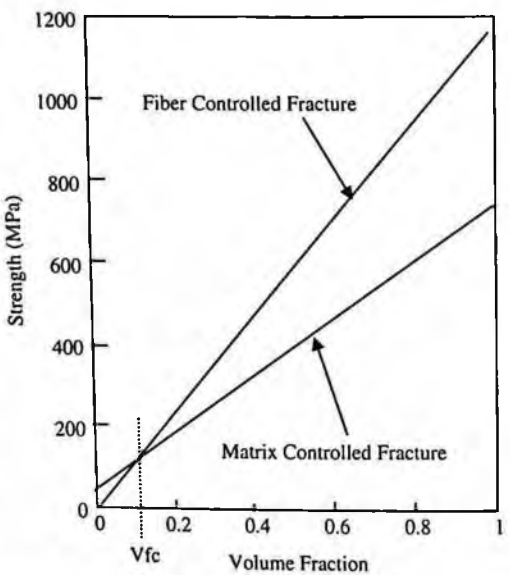


Fig. 4.6 Typical strength controlled failure.

4.4.2 *Fiber triggered failure*

Considering the case where the matrix is ductile and the elastic strain to fracture in the fibers is less than the elastic/plastic extension of the matrix. At low volume fractions of fibers, the fiber fails first and break. The load carried by fibers will transfer to the matrix which, having a reduced cross-section, will see a sudden jump in stress. The subsequent failure process would depend on the magnitude of an increase in stress in the matrix.

The stress on the composite at the point of fiber fracture (ε_f) is

$$\sigma_{11} = E_f \varepsilon_f V_f + E_m \varepsilon_f (1 - V_f) \tag{4.21}$$

The force on the composite is just the product of the stress and the cross-sectional area, so the stress on the matrix after the fibers break is

$$\sigma_m = E_m \varepsilon_f + (V_f / 1 - V_f) \sigma_f \tag{4.22}$$

So the stress on the matrix increases by $(V_f/1 - V_f)\sigma_f$. If the rise in stress is not sufficient to fracture the matrix then it will continue to support the applied load. Then, the fracture strength of the composite will be given by

$$\sigma_{11} = (1 - V_f)\sigma_m \tag{4.23}$$

where σ_m is the ultimate tensile strength of the matrix; i.e. the addition of fibers leads to a reduction in the strength of the composite to levels below that of the unreinforced matrix. As the fiber volume fraction increases, the fibers carry more of the applied load. When the fibers break, the load transferred to the matrix is large and the much reduced cross-sectional area of the matrix will be unable to support the load and the matrix too will fail. The strength of the composite is determined by the strength of the fibers i.e.

$$\sigma_{11} = E_f V_f \varepsilon_f + (1 - V_f) E_m \varepsilon_f \tag{4.24}$$

In Fig. 4.7, for glass-polyamide composite, we can see that the tensile strength of a composite in which the fibers fail at a lower strain than that of matrix initially decreases below that of neat matrix, reaching a minimum and thereafter increase. There is, therefore, a minimum volume fraction, $V_{f\min}$, of fibers that must be added in order for the composite to have strength at least equal to that of the matrix alone.

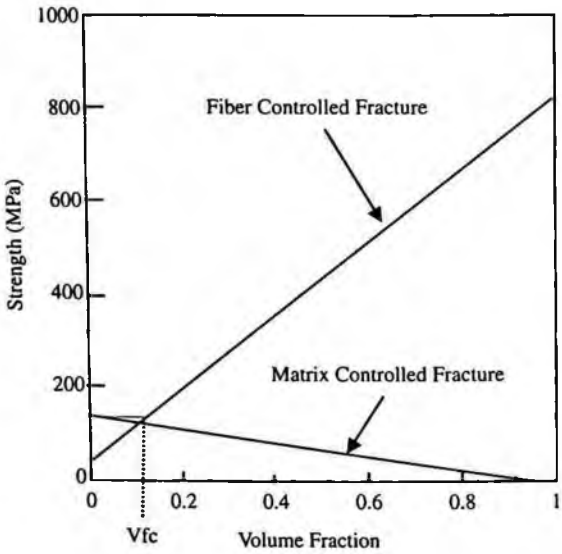


Fig. 4.7 Typical strain controlled failure.

In the example shown above, where glass fibers are used to reinforce a polyamide matrix, $V_{f\min}$ is around 9%.

4.4.3 *Transverse strength*

The behavior of the composite when loaded in a direction perpendicular to the fibers is very different compared to the loading along the fiber direction. The additional complexity is introduced due to the presence of fiber-matrix interface.

When loaded in the transverse direction, both the fibers and the matrix experience the same stress — isostress condition. Hence, the strength in transverse direction is determined by the weakest link in the composite. Of the two materials that make up the composite, the matrix is invariably the weaker material and so fracture will occur when the stress reaches the matrix fracture stress. This applies when the matrix and the fiber is perfectly bonded and will transmit the entire load applied to it. The presence of the fiber-matrix interface either makes it stronger or weakens it, depending on the interface properties. Hence, the transverse strength of unidirectional composites can be lower than the strength value of matrix alone.

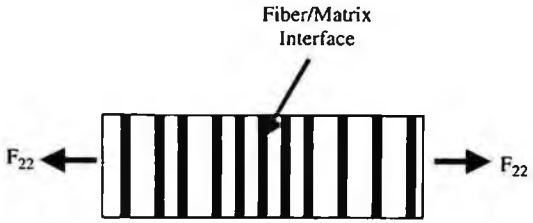


Fig. 4.8 Transverse failure in UD composites.

4.5 Effect of Fiber Orientation on Elastic Properties

The elastic and strength properties discussed earlier relate to the properties in the primary materials directions, i.e. parallel and perpendicular to the fiber direction. For practical application and to derive the best of advantages from the composite materials, it is necessary to compute the values in other orientations.

The Hooke's law relationships for the lamina are:

$$\sigma = Q\varepsilon \tag{4.25}$$

$$\begin{bmatrix} \sigma_1 \\ \sigma_2 \\ \tau_{12} \end{bmatrix} = \begin{bmatrix} Q_{11} & Q_{12} & 0 \\ Q_{21} & Q_{22} & 0 \\ 0 & 0 & Q_{66} \end{bmatrix} \cdot \begin{bmatrix} \varepsilon_1 \\ \varepsilon_2 \\ \gamma_{12} \end{bmatrix} \tag{4.26}$$

where, Q is the stiffness matrix, and,

$$Q_{11} = \frac{E_{11}}{(1 - \nu_{12}\nu_{21})}$$

$$Q_{22} = \frac{E_{22}}{(1 - \nu_{12}\nu_{21})}$$

$$Q_{12} = \frac{\nu_{21}E_{11}}{(1 - \nu_{12}\nu_{21})} = \frac{\nu_{12}E_{22}}{(1 - \nu_{12}\nu_{21})} = Q_{21}$$

$$Q_{66} = G_{12}$$

$$\nu_{21} = \frac{\nu_{12} E_{22}}{E_{11}}$$

here, ν_{12} is Poisson's ratio representing a strain in the '2' direction resulting from a load applied in the '1' direction, i.e. $(-\varepsilon_2/\varepsilon_1)$; similarly $\nu_{21} = (-\varepsilon_1/\varepsilon_2)$.

In this case, plane stress is assumed and therefore there are no through thickness stresses, i.e. $\sigma_3 = \tau_{23} = \tau_{31} = 0$. However we must remember that the composite is not isotropic and thus E_{11} and E_{22} are not the same.

The compliance matrix **S**, which is the inverse of the stiffness matrix **Q** is as follows,

$$S = Q^{-1} \tag{4.27}$$

$$\varepsilon = S\sigma \tag{4.28}$$

$$\begin{bmatrix} \varepsilon_1 \\ \varepsilon_2 \\ \gamma_{12} \end{bmatrix} = S \cdot \begin{bmatrix} \sigma_1 \\ \sigma_2 \\ \tau_{12} \end{bmatrix} \tag{4.29}$$

$$S = \begin{bmatrix} \dfrac{1}{E_{11}} & \dfrac{-\nu_{12}}{E_{11}} & 0 \\ \dfrac{-\nu_{12}}{E_{11}} & \dfrac{1}{E_{22}} & 0 \\ 0 & 0 & \dfrac{1}{G_{12}} \end{bmatrix}$$

Defining the rotation from the special '1-2' coordinate system that is aligned with the fibers to a general '$x - y$' coordinate system that is aligned with the direction of loading, the angle between the two is θ.

To determine lamina properties with respect to other coordinate axes, the stiffness and compliance matrices must be transformed. For rotation by

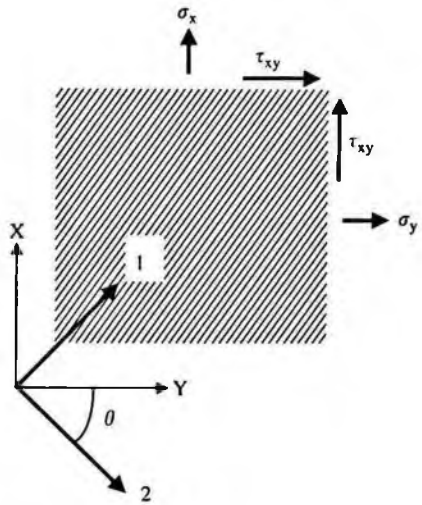


Fig. 4.9 Rotation of fibers from the principal axes.

angle θ to axes x, y, it can be shown that:

$$\begin{Bmatrix} \sigma_x \\ \sigma_y \\ \tau_{xy} \end{Bmatrix} = [\bar{Q}] \begin{Bmatrix} \varepsilon_x \\ \varepsilon_y \\ \gamma_{xy} \end{Bmatrix} \tag{4.30}$$

where $[\bar{Q}]$ is transformed stiffness matrix given by,

$$[\bar{Q}] = [T]^{-1}[Q][T]^{-T} \tag{4.31}$$

where $[T]$ is defined as,

$$T = \begin{bmatrix} \cos^2\theta & \sin^2\theta & 2\sin\theta\cos\theta \\ \sin^2\theta & \cos^2\theta & -2\sin\theta\cos\theta \\ -\sin\theta\cos\theta & \sin\theta\cos\theta & \cos^2\theta - \sin^2\theta \end{bmatrix}$$

From the above equations, the engineering constants can be expressed as functions of the fiber orientation angle as follows;

$$\frac{1}{E_x} = \frac{1}{E_1}\cos^4\theta + \left(\frac{1}{G_{12}} - \frac{2\nu_{12}}{E_1}\right)\sin^2\theta\cos^2\theta + \frac{1}{E_2}\sin^4\theta$$

$$\nu_{xy} = E_x\left[\frac{\nu_{12}}{E_1}(\sin^4\theta + \cos^4\theta) - \left(\frac{1}{E_1} + \frac{1}{E_2} - \frac{1}{G_{12}}\right)\sin^2\theta\cos^2\theta\right]$$

$$\frac{1}{E_y} = \frac{1}{E_1}\sin^4\theta + \left(\frac{1}{G_{12}} - \frac{2\nu_{12}}{E_1}\right)\sin^2\cos^2\theta + \frac{1}{E_2}\cos^4\theta$$

$$\frac{1}{G_{xy}} = 2\left(\frac{2}{E_1} + \frac{2}{E_2} + \frac{4\nu_{12}}{E_1} - \frac{1}{G_{12}}\right)\sin^2\theta\cos^2\theta + \frac{1}{G_{12}}(\sin^4\theta + \cos^4\theta) \tag{4.32}$$

4.6 Elastic Properties of Multi-Ply Laminates

A laminate is a collection of laminae stacked one over the other in a specified manner. Adjacent lamina may be of the same or different materials and their fiber orientation with respect to the laminate reference axis may be arbitrary. Although the laminate is made up of multiple laminae, it is assumed that the individual laminae are perfectly bonded together so as to behave as a single, non-homogeneous, anisotropic plate. This assumption means that classical plate theory can be used for the laminated plates. The x-y plane is the plane of the laminate, the z-direction is perpendicular to the plane of the laminate (see Fig. 4.10).

The displacements in the x, y and z directions are u, v and w respectively. It is assumed that laminate displacements in the z-direction only arise from bending. There is no variation in thickness in the z-direction (i.e. no through thickness strain). The centerline is a line through the thickness of the laminate that divides the laminate vertically into two regions of equal thickness.

When the laminate bends, as shown in Fig. 4.11, there is no extension at the centerline, but on the inside of the bend, above the center line there is an increasing amount of compression as we move away from the center line. On the outside of the bend, below the center-line, there is an increasing

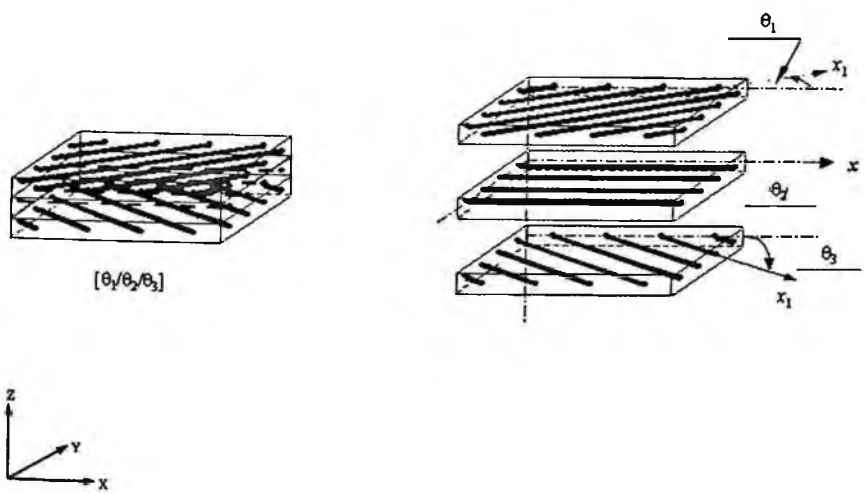


Fig. 4.10 Typical laminate construction.

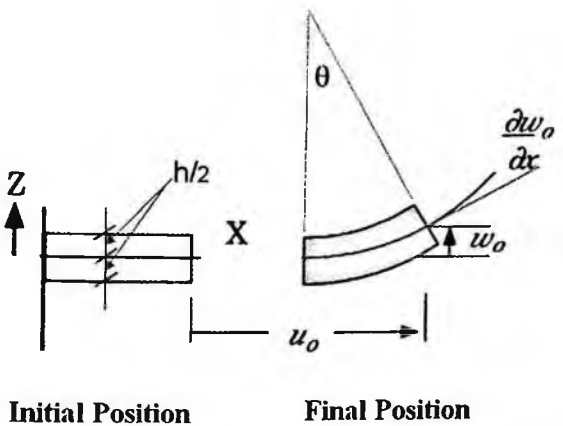


Fig. 4.11 Strain-displacement relationship in a laminate.

amount of tension as we move away from the center-line to the outer surface of the laminate. For small angles $\theta = \tan(\theta) = \partial w/\partial x$.

The in-plane displacements (u and v), which are functions of position (x, y, z) within the laminate can be related to the center-line displacements,

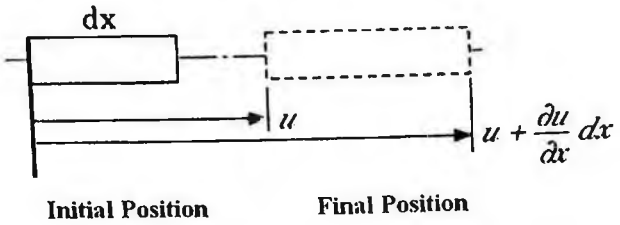


Fig. 4.12 Definition of normal strain.

u_0 and v_0 and the slopes by

$$u(x, y, z) = u_0(x, y) - z\frac{\partial w_0(x, y)}{\partial x} \tag{4.33}$$

$$v(x, y, z) = v_0(x, y) - z\frac{\partial w_0(x, y)}{\partial y} \tag{4.34}$$

Now that we have the displacements, we can get the normal strain.

$$\varepsilon_x = \frac{\partial u}{\partial x} = \frac{\left(u + \frac{\partial u}{\partial x} \cdot dx\right) - u}{dx} \tag{4.35}$$

Substituting for u, from eqn. 4.33,

$$\varepsilon_x = \frac{\partial u_0}{\partial x} - z\frac{\partial^2 w_0}{\partial x^2} \tag{4.36}$$

The strain term ε_y is obtained in the same way. The engineering shear strain is just the change in the angle between two initially perpendicular sides. For small strains, $\gamma = \tan(\gamma)$.

$$\gamma_{xy} = \frac{\partial u}{\partial y} + \frac{\partial v}{\partial x} = \frac{\left(\frac{\partial u}{\partial y}\right) dy}{dy} + \frac{\left(\frac{\partial u}{\partial x}\right) dx}{dx} \tag{4.37}$$

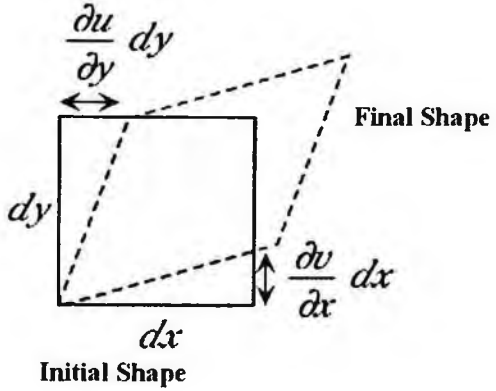


Fig. 4.13 Definition of shear strain.

Again substituting for u and v then differentiate with respect to x and y. The resulting strain matrix may be written as:

$$\begin{bmatrix} \varepsilon_x \\ \varepsilon_y \\ \gamma_{xy} \end{bmatrix} = \begin{Bmatrix} \dfrac{\partial u_0}{\partial x} \\ \dfrac{\partial v_0}{\partial y} \\ \dfrac{\partial u_0}{\partial y} + \dfrac{\partial v_0}{\partial x} \end{Bmatrix} + z \cdot \begin{Bmatrix} -\dfrac{\partial^2 w_0}{\partial x^2} \\ -\dfrac{\partial^2 w_0}{\partial y^2} \\ -2\dfrac{\partial^2 w_0}{\partial x \cdot \partial y} \end{Bmatrix} \tag{4.38}$$

The above equation can be written more simply as

$$\{\varepsilon\} = \{\varepsilon^0\} + z\{\kappa\} \tag{4.39}$$

where $\{\varepsilon^0\}$ is the center-line strains and $\{\kappa\}$ the curvatures:

$$\{\varepsilon^0\} = \begin{Bmatrix} \varepsilon_x^0 \\ \varepsilon_y^0 \\ \gamma_{xy}^0 \end{Bmatrix} \quad \text{and} \quad \{\kappa\} = \begin{Bmatrix} -\dfrac{\partial^2 w}{\partial x^2} \\ -\dfrac{\partial^2 w}{\partial y^2} \\ -2\dfrac{\partial^2 w}{\partial x \cdot \partial y} \end{Bmatrix} \tag{4.40}$$

When a force is applied to the edge of a laminate, all the laminae of the laminate will stretch the same amount, i.e. they will experience the same

strain. However, the elastic properties of each lamina in the laminate is different and depend on:

(1) The fibre and matrix materials.
(2) The volume fraction of fibers and its packing.
(3) The orientation of the fibers and form of fiber placement.

For a specific lamina (termed "kth" lamina), the stress-strain relationship will be $\{\sigma\}_k = [\bar{Q}]_k\{\varepsilon\}$.

Since the stress in a given lamina is the product of stiffness and strain of that lamina, the stress in each lamina will also be different. Force is the product of the stress and the cross-sectional area of the lamina (thickness × width), the force acting on each lamina can be determined. Within a laminate, the sum of the forces in the individual laminae must add up to the applied force for equilibrium,

$$F = \sum_{k=1}^{n} F_k = \sum_{k=1}^{n} \sigma_k \cdot (t_k \cdot W) \tag{4.41}$$

where F_k is the force in the kth lamina of the laminate, σ_κ, is the stress, t_k, is the thickness of the kth lamina and W, the width of the laminate. By convention, N is described as force per unit width of the laminate or the force resultant. Mathematically, the force resultant is defined as,

$$N_x = \int_{-h/2}^{+h/2} \sigma_x dz \tag{4.42}$$

for the force resultant in the x-direction. The term h is the total thickness of the laminate. We can write down both the force resultants and moment resultants (force per unit width of laminate × distance) as,

$$\begin{Bmatrix} N_x \\ N_y \\ N_{xy} \end{Bmatrix} = \int_{-h/2}^{+h/2} \begin{Bmatrix} \sigma_x \\ \sigma_y \\ \sigma_{xy} \end{Bmatrix} dz \tag{4.43}$$

$$\begin{Bmatrix} M_x \\ M_y \\ M_{xy} \end{Bmatrix} = \int_{-h/2}^{+h/2} \begin{Bmatrix} \sigma_x \\ \sigma_y \\ \sigma_{xy} \end{Bmatrix} \cdot z dz \tag{4.44}$$

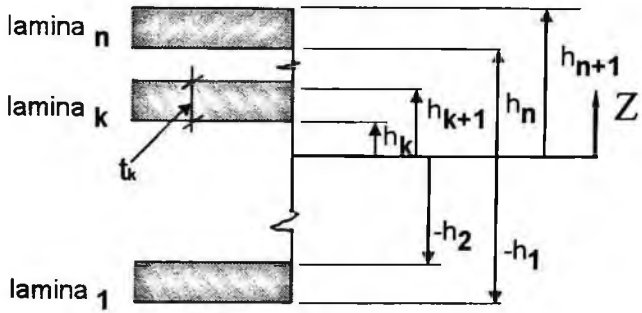


Fig. 4.14 Lamina position in a laminate.

The above equations (4.43) and (4.44) can be expressed as,

$$\begin{Bmatrix} N_x \\ N_y \\ N_{xy} \end{Bmatrix} = \sum_{k=1}^{n} \int_{h_k}^{h_{k+1}} \begin{Bmatrix} \sigma_x \\ \sigma_y \\ \sigma_{xy} \end{Bmatrix} dz \tag{4.45}$$

$$\begin{Bmatrix} M_x \\ M_y \\ M_{xy} \end{Bmatrix} = \sum_{k=1}^{n} \int_{h_k}^{h_{k+1}} \begin{Bmatrix} \sigma_x \\ \sigma_y \\ \sigma_{xy} \end{Bmatrix} \cdot z dz \tag{4.46}$$

where h_k is the position of the bottom of the k_{th} lamina with respect to the center-line of the laminate and h_{k+1}, the position of the top of the k_{th} lamina with respect to the center line of the laminate as shown in Fig. 4.14.

The lamina stress is the product of the specific lamina stiffness and the strain experienced by this lamina,

$$\begin{Bmatrix} \sigma_x \\ \sigma_y \\ \tau_{xy} \end{Bmatrix}_k = [\bar{Q}]_k \begin{Bmatrix} \varepsilon_x \\ \varepsilon_y \\ \gamma_{xy} \end{Bmatrix} \tag{4.47}$$

where the stiffness matrix $[\bar{Q}]_k$ is a function of orientation, fiber fraction and fiber and matrix materials of the kth lamina. In a given ply, $[\bar{Q}]$ is

constant hence,

$$\begin{Bmatrix} N_x \\ N_y \\ N_{xy} \end{Bmatrix} = \sum_{k=1}^{n} \int_{h_k}^{h_{k+1}} [\bar{Q}]_k \begin{Bmatrix} \varepsilon_x \\ \varepsilon_y \\ \gamma_{xy} \end{Bmatrix} dz$$

$$= \sum_{k=1}^{n} \int_{h_k}^{h_{k+1}} [\bar{Q}]_k \begin{Bmatrix} \varepsilon_x^0 \\ \varepsilon_y^0 \\ \gamma_{xy}^0 \end{Bmatrix} dz + \sum_{k=1}^{n} \int_{h_k}^{h_{k+1}} [\bar{Q}]_k \begin{Bmatrix} \kappa_x \\ \kappa_y \\ \kappa_{xy} \end{Bmatrix} \cdot z dz \tag{4.48}$$

similarly, for the moment resultants

$$\begin{Bmatrix} M_x \\ M_y \\ M_{xy} \end{Bmatrix} = \sum_{k=1}^{n} \int_{h_k}^{h_{k+1}} [\bar{Q}]_k \begin{Bmatrix} \varepsilon_x \\ \varepsilon_y \\ \gamma_{xy} \end{Bmatrix} \cdot z dz$$

$$= \sum_{k=1}^{n} \int_{h_k}^{h_{k+1}} [\bar{Q}]_k \begin{Bmatrix} \varepsilon_x^o \\ \varepsilon_y^o \\ \gamma_{xy}^o \end{Bmatrix} \cdot z dz + \sum_{k=1}^{n} \int_{h_k}^{h_{k+1}} [\bar{Q}]_k \begin{Bmatrix} \kappa_x \\ \kappa_y \\ \kappa_{xy} \end{Bmatrix} \cdot z^2 dz \tag{4.49}$$

Since the stiffness matrices $[\bar{Q}]_k$, the center-line strains $\{\varepsilon^0\}$ and curvatures $\{\kappa\}$ are constant in each ply. The only variable is z, the vertical position within each ply. Integrating equation (4.48) and (4.49), we get,

$$\int_{h_k}^{h_{k+1}} dz = h_{k+1} - h_k$$

$$\int_{h_k}^{h_{k+1}} z dz = \tfrac{1}{2}\left(h_{k+1}^2 - h_k^2\right)$$

$$\int_{h_k}^{h_{k+1}} z^2 dz = \tfrac{1}{3}\left(h_{k+1}^3 - h_k^3\right)$$

and

$$\begin{Bmatrix} N \\ M \end{Bmatrix} = \begin{bmatrix} A & B \\ B & D \end{bmatrix} \begin{Bmatrix} \varepsilon^0 \\ \kappa \end{Bmatrix} \tag{4.50}$$

$$\begin{Bmatrix} N \\ M \end{Bmatrix} = \begin{bmatrix} \sum_{k=1}^{n} [\bar{Q}]_k \cdot (h_{k+1} - h_k) & \frac{1}{2}\sum_{k=1}^{n} [\bar{Q}]_k \cdot \left(h_{k+1}^2 - h_k^2\right) \\ \frac{1}{2}\sum_{k=1}^{n} [\bar{Q}]_k \cdot \left(h_{k+1}^2 - h_k^2\right) & \frac{1}{3}\sum_{k=1}^{n} [\bar{Q}]_k \cdot \left(h_{k+1}^3 - h_k^3\right) \end{bmatrix} \begin{Bmatrix} \varepsilon^0 \\ \kappa \end{Bmatrix} \tag{4.51}$$

Each of the 3 × 3 matrices, [A], [B], [D] in equation (4.50) has a distinct function and are termed;

[A] — Extensional stiffness matrix,
[B] — Extensional-bending coupling matrix,
[D] — Bending stiffness matrix.

In symmetric laminates, $[B] = 0$.

4.7 Textile Composites

Textile reinforced composites are fiber reinforced composites whose unit structures are characterized by more than one fiber orientation. Figure 4.15 shows a schematic illustration of the hierarchical nature of textile materials.

The majority of structures made from composites, are made from textile composites rather than the simple uniaxial fibers. A rigorous analysis of the stiffness of a composite made from textile composites shown above is much more complex compared to the unidirectional composite described earlier.

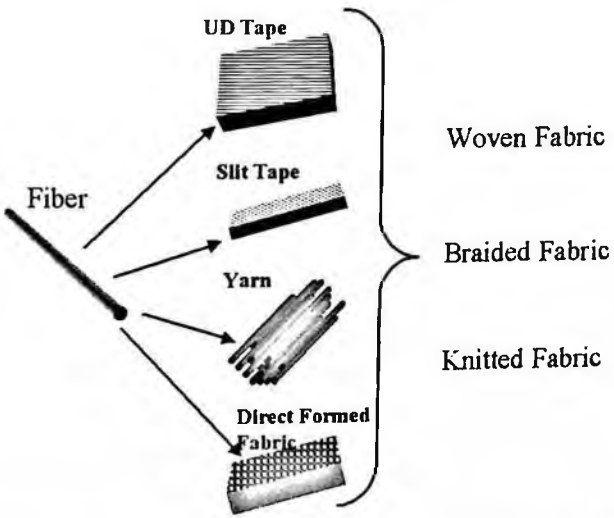


Fig. 4.15 Schematic illustration of the hierarchy of fibers, yarns, and fabrics in textile processes.

For a simple plain woven fabric in which the yarns are not excessively crimped and having half of the yarns in the warp (0°) direction and the other half in the weft (90°) direction, the stiffness in each of these directions is then simply calculated using the isostrain rule of mixtures, but assuming that the volume fraction of fibers, V_f, is only half the total fiber content.

For other cases, the analysis of the textile composite essentially consists of three steps, somewhat similar to the analysis of a multidirectional tape laminate. In the first step, the composite is divided into a number of UD laminae. For a multidirectional tape laminate, such division is straightforward: each layer is a UD lamina in its local coordinate system. For a textile composite, however, a geometric description for the textile fabric is necessary before doing this. In the second step, the UD laminae are analyzed in their local coordinate systems. The third step is an assemblage of the contribution of all the UD laminae to obtain the properties of the original textile composite.

It should be noted that a single layer of textile composite could also be a multilayer laminate, with each lamina being a single layer braided/woven/knit fabric reinforced composite. In such a case, the laminate analysis is exactly the same as that expressed in the preceding section. The only difference is that in the analysis of a single lamina layer, the lamina instantaneous stiffness matrix expressed in the global coordinate system should be used.

4.8 Behavior of Composite Properties

The overall performance of a composite depends primarily on the properties of its constituent fiber and matrix materials, the fiber content, and the fiber arrangement. By using techniques described earlier, it is possible to quantitatively determine this primary dependence. Furthermore, the composite performance also depends secondarily on the "binding condition" involved, such as fiber/matrix interface bonding, fabrication defect (e.g. misalignment of fibers), void content, etc. Evidently, such a "binding condition" is indeterminate *a priori*. It is difficult to model exactly the secondary dependence of the composite properties on the "binding condition". Fortunately, as with the advancement of the composite fabrication technology, this secondary dependence can be reduced to a minimum. At the first stage

of a composite design, this kind of effect can be neglected. A preliminary design will only be based on the primary effect consideration. Given the working conditions of a structural component and several candidate fiber (i.e. reinforcement) and matrix materials together with their properties, suitable fiber and matrix materials, their contents, and their arrangements to develop an optimal composite that can meet all the working conditions can be determined.

For example, in order to develop more suitable artificial limbs for amputated patients, measurements from trans-femoral amputee subjects have been performed under their walking conditions. Taking Y as upward (normal to the ground) coordinate direction, X as the forward direction of the objects' progression, and Z to the right, the measured maximum dynamic forces and moments on the amputees are given as [Peter, 1996].

$$F_Y = 819N, \quad F_X = 330N, \quad F_Z = 265N,$$
$$M_X = 77Nm, \quad M_Y = 35Nm, \quad \text{and} \quad M_Z = 93Nm.$$

Thus, a developed composite socket should be able to sustain the above load conditions. In practice, another kind of designing procedure is also followed. Namely, a new composite material can also be developed by comparing its performance with that of an existing material. This is especially true in some biomedical applications for which designing standards have not yet been established. For the below knee prosthesis application, a currently used socket material exhibits the properties as shown in Tables 4.1 and 4.2. Thus, our objective was to develop a new composite socket material that can meet the stiffness and strength requirements of Tables 4.1 and 4.2. In general, there are sufficient degrees of freedom to enable the developed composite materials to satisfy the given requirements. An optimal design

Table 4.1 Tensile properties of an existing below knee socket material [Huang and Ramakrishna, 1999].

Young's Modulus (GPa)		Ultimate Strength (MPa)		Ultimate Strain (%)	
Longitudinal	Transverse	Longitudinal	Transverse	Longitudinal	Transverse
3.68	2.12	42.5	32.4	2.15	3.65

Longitudinal = socket axial; transverse = transverse to the socket axis.

Table 4.2 Bending properties of an existing below knee socket material [Huang and Ramakrishna, 1999].

Young's Modulus (GPa)		Ultimate Strength (MPa)		Ultimate Strain (%)	
Longitudinal	Transverse	Longitudinal	Transverse	Longitudinal	Transverse
3.12	2.65	55.1	43.7	4.25	6.45

is then possible by requiring that the developed materials also satisfy some additional conditions such as minimum weight or minimum cost.

Here, we only deal with stiffness and strength design of the composites. Possible influence of different designing variables on these two properties is considered. The designs will be shown in figures and charts.

4.8.1 *Unidirectional composites*

The in-plane engineering elastic constants of eight UD laminae made from four different fibers, given in Table 4.3, and the R57 epoxy (Table 4.4) and the PEEK matrix (elastic modulus E = 3.6 GPa and Poisson's ratio $\nu = 0.4$) materials are calculated. Results are graphed in Figs. 4.16–4.19. Here and in the following, "Gev" refers to "E-glass 21(K43 Gevetex" fibers

Table 4.3 Mechanical properties of four fibers.

Type	*AS4*	*T*300	*E*-glass 21 × K43 Gevetex	Silenka *E*-Glass 1200tex
E_{11}^f (GPa)	225	230	80	74
E_{22}^f (GPa)	15	15	80	74
G_{12}^f (GPa)	15	15	33.33	30.8
V_{12}^f	0.2	0.2	0.2	0.2
G_{23}^f (GPa)	7	7	33.33	30.8
σ_u^f (MPa)	3206	2468	1804	2093
$\sigma_{u,c}^f$ (MPa)	2459	1470	910	1312

Table 4.4 Properties of R57 epoxy resin. (Tensile strength: $\sigma_u = 45$ MPa, compressive strength: $\sigma_{u,c} = 93$ MPa, Poisson's ratio $\nu = 0.414$)

	$(\sigma_Y)_1$ (MPa)	$(\sigma_Y)_2$ (MPa)	$(\sigma_Y)_3$ (MPa)	$(\sigma_Y)_4$ (MPa)	$(E_T)_1$ (GPa)	$(E_T)_2$ (GPa)	$(E_T)_3$ (GPa)	$(E_T)_4$ (GPa)
Uniaxial tension	15.8	33.3	42.5	44.7	2.28	1.54	0.94	0.22
Uniaxial compression	35.3	62.2	83.8	92.2	5.42	3.86	2.23	0.63

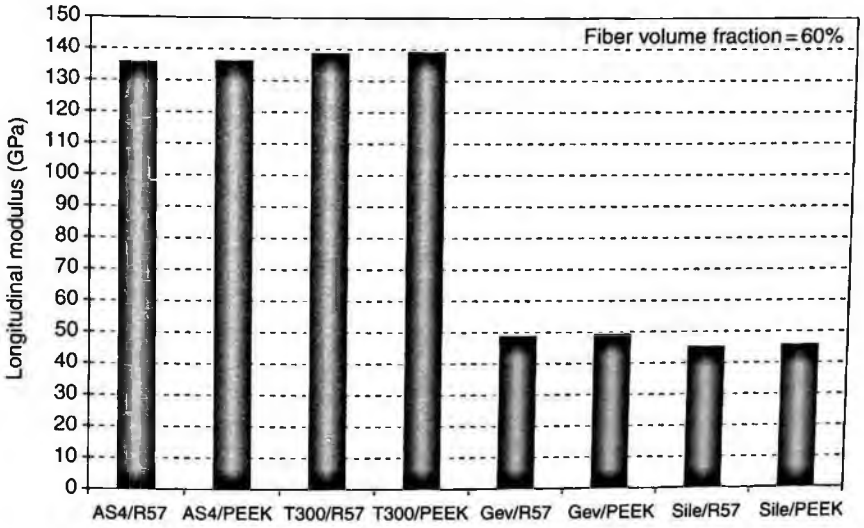


Fig. 4.16 Longitudinal tensile moduli of UD composites made from different material systems, with $V_f = 0.6$.

and "Sile" designates "Silenka E-Glass 1200tex" fibers. From Fig. 4.16, one can see that the matrix property has negligibly small effect on the composite longitudinal modulus, as the longitudinal stiffness of the fibers used is significantly higher than the matrix stiffness. On the other hand, the matrix property considerably affects the overall transverse and in-plane shear moduli of the resulting UD composites, as shown in Figs. 4.17 and 4.18 respectively. This influence is more distinct for the glass fibers based composites. Note that the transverse and in-plane shear moduli of the carbon fibers are lower than those of the glass fibers, as a result of lower transverse

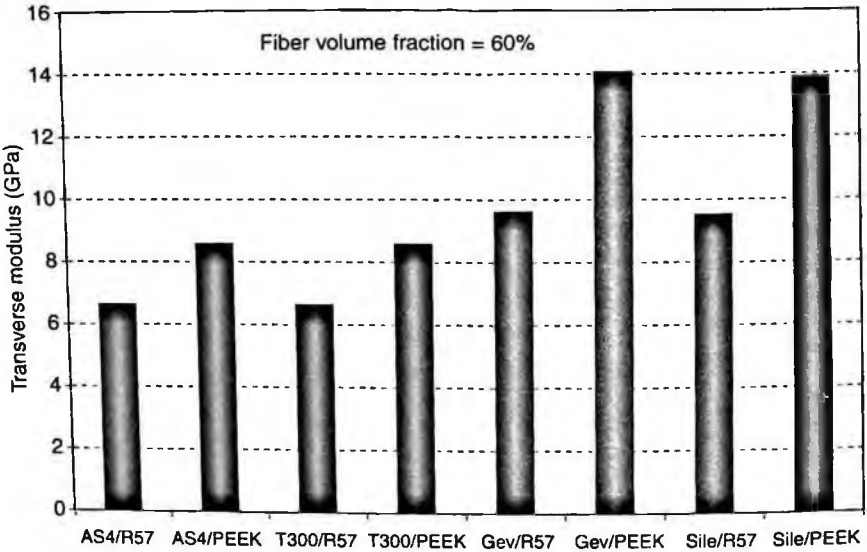


Fig. 4.17 Transverse tensile moduli of UD composites made from different material systems, with $V_f = 0.6$.

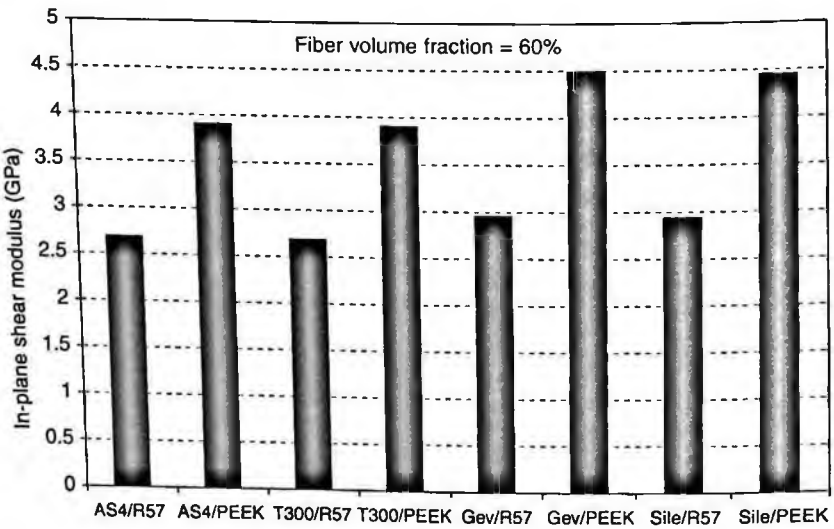


Fig. 4.18 In-plane shear moduli of UD composites made from different material systems, with $V_f = 0.6$.

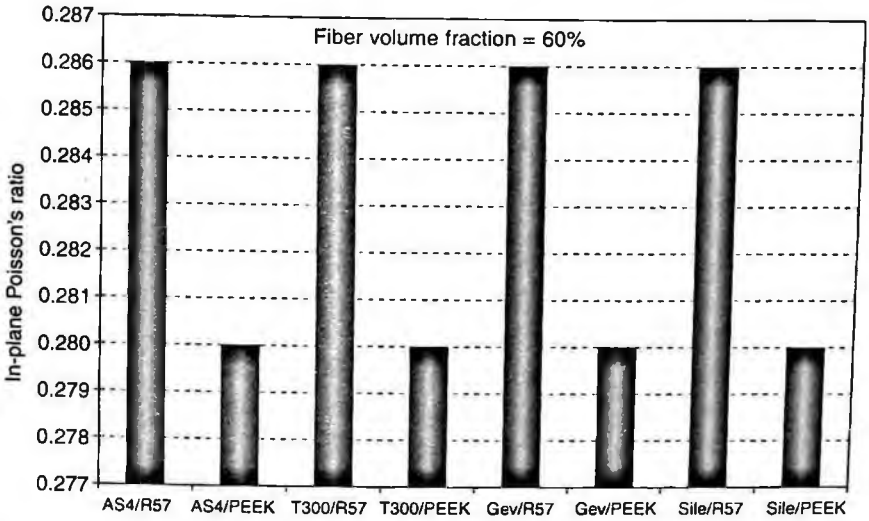


Fig. 4.19 Poisson's ratios of UD composites made from different material systems, with $V_f = 0.6$.

and in-plane moduli for the carbon fibers. This means that the influence of the matrix stiffness on the overall transverse or in-plane shear modulus of the composite is more if the modular ratio between the fiber and the matrix in that direction is higher. Off-axial tensile stiffnesses of the AS4/R57 (carbon/epoxy) and the Silenka/R57 (glass/epoxy) varied with different fiber volume fractions are plotted in Figs. 4.20 and 4.21 respectively. The figures indicate that the modulus variation with the increase of the fiber volume fraction is apparent for the carbon/epoxy composite only when an off-axial angle is smaller than 40°. In the contrast, this variation is more distinct for the glass/epoxy composite for the whole off-axial loading range. This is attributed to two reasons. The first reason is that the carbon fibers have much larger modular ratio in the fiber axial direction with respect to the R57 resin than the glass fibers do. The second reason is that the glass fibers are isotropic and have higher stiffness values in the transverse and the in-plane shear moduli, relative to the carbon fibers, giving more increase to the overall transverse and in-plane shear moduli of the resulting composite (see Figs. 4.17 and 4.18).

The ultimate strengths of four UD composites based on four different fibers (Table 4.3) and the R57 epoxy matrix (Table 4.4) subjected to uniaxial

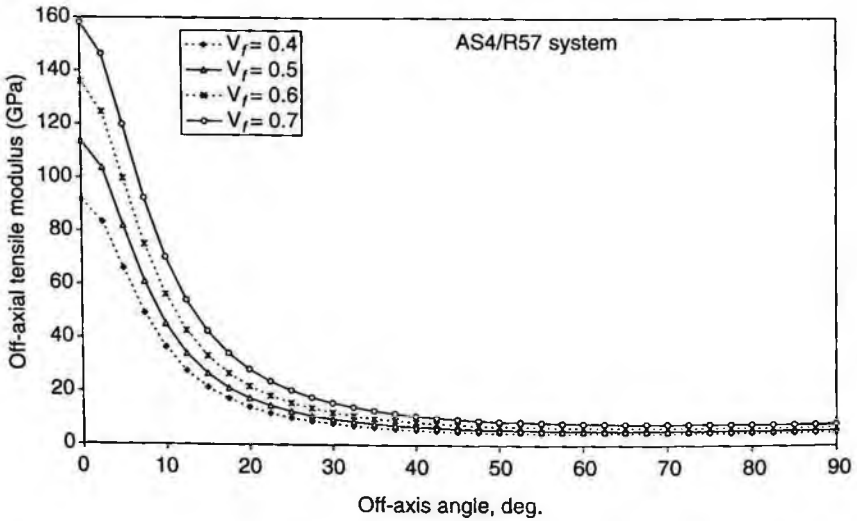


Fig. 4.20 Off-axial tensile moduli of UD composites made from AS4/R57 material systems, with $V_f = 0.4$, 0.5, 0.6, and 0.7.

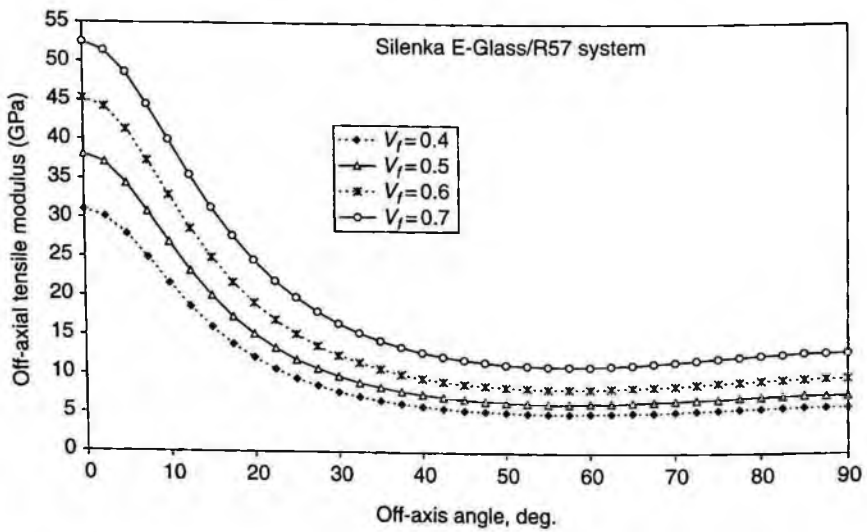


Fig. 4.21 Off-axial tensile moduli of UD composites made from Silenka E-Glass/R57 material systems, with $V_f = 0.4$, 0.5, 0.6, and 0.7.

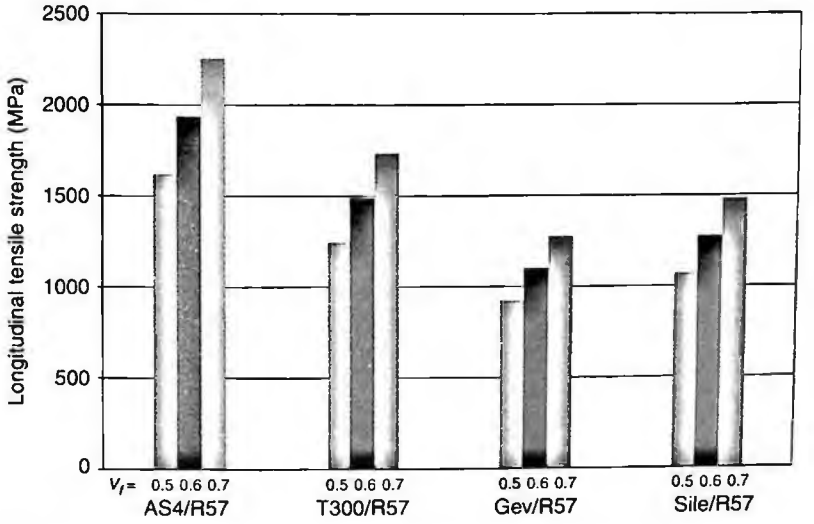


Fig. 4.22 Longitudinal tensile strengths of UD composites made from different material systems, with $V_f = 0.5$, 0.6, and 0.7.

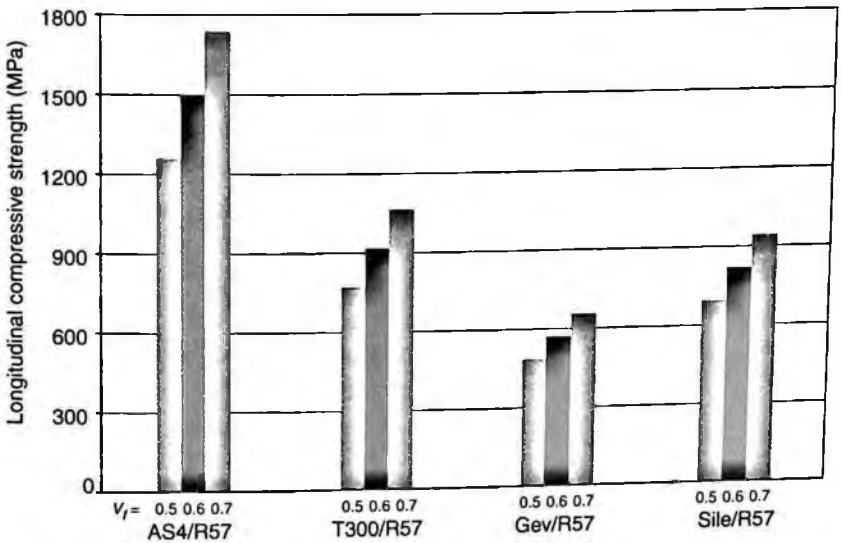


Fig. 4.23 Longitudinal compressive strengths of UD composites made from different material systems, with $V_f = 0.5$, 0.6, and 0.7.

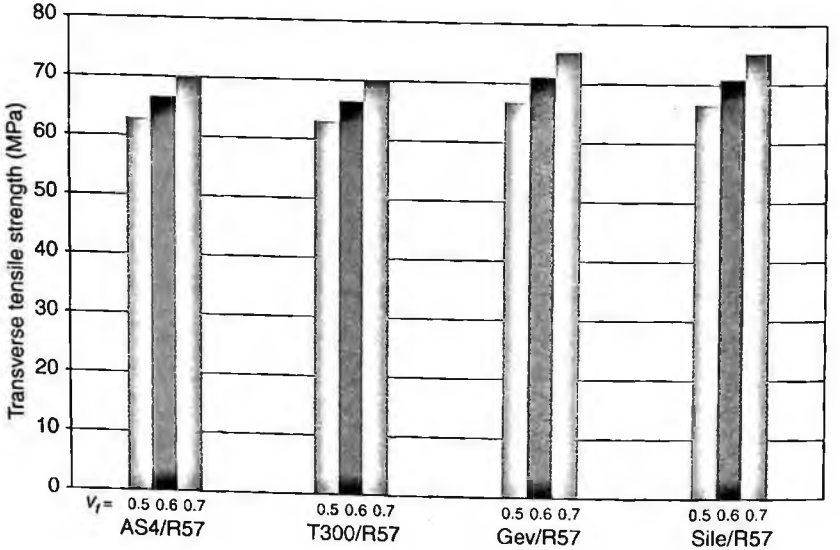


Fig. 4.24 Transverse tensile strengths of UD composites made from different material systems, with $V_f = 0.5$, 0.6, and 0.7.

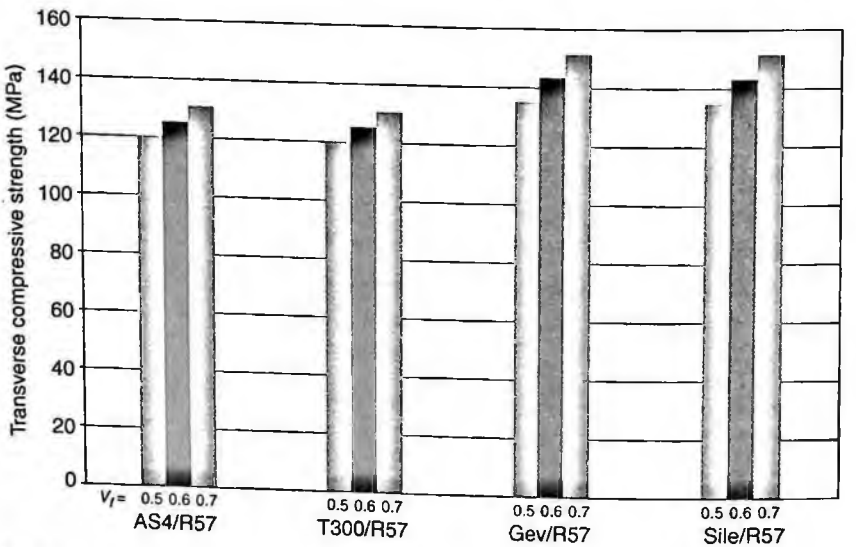


Fig. 4.25 Transverse compressive strengths of UD composites made from different material systems, with $V_f = 0.5$, 0.6, and 0.7.

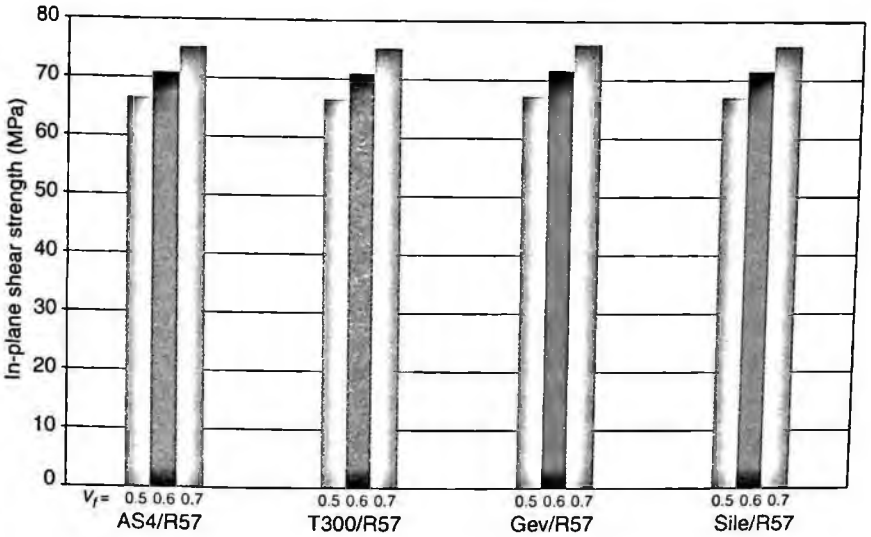


Fig. 4.26 In-plane shear strengths of UD composites made from different material systems, with $V_f = 0.5$, 0.6, and 0.7.

(longitudinal, transverse, or in-plane shear) load are indicated in Figs. 4.22–4.26. The longitudinal strength of the composite is determined by the fiber strength, whereas the transverse and in-plane shear strengths mainly depend on the matrix strength. It is seen that the longitudinal strengths of the carbon/epoxy composites are considerably higher than the corresponding strengths of the glass/epoxy composites. This is because the carbon fibers are stronger than the glass fibers. However, the overall transverse and in-plane shear strengths of the glass/epoxy composites are slightly higher than the counterparts of the carbon/epoxy composites. This means that the material stiffness in the corresponding direction also has an influence, although not very significant in the transverse or in-plane shear case, on the composite ultimate strength in that direction.

The influence of the fiber content is different in different fiber direction. This is more clearly indicated in Figs. 4.27 and 4.28. The fiber content can significantly affect the composite strength only in or very close to the longitudinal direction.

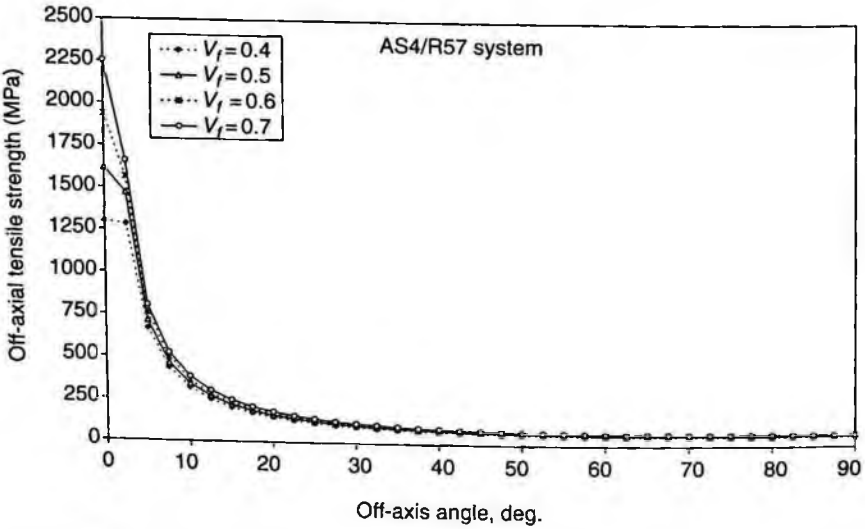


Fig. 4.27 Off-axial tensile strengths of UD composites made from AS4/R57 material systems, with $V_f = 0.4, 0.5, 0.6$, and 0.7.

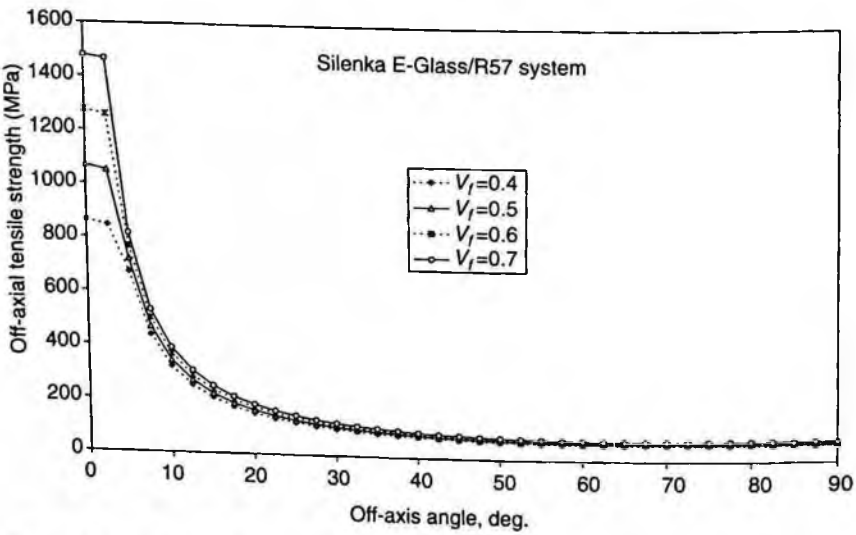


Fig. 4.28 Off-axial tensile strengths of UD composites made from Silenka E-Glass/R57 material systems, with $V_f = 0.4, 0.5, 0.6$, and 0.7.

4.8.2 *Multi-ply laminates*

There are more design variables for a multidirectional tape laminate, relative to a UD composite. Here, the main focus is on the effect of laminate lay-ups on the in-plane stiffness and strength of the resulting composites. Two material systems, i.e. the AS4/R57 (carbon/epoxy) and the Silenka E-Glass/R57 (glass/epoxy) systems are employed to bring out the effects.

Three sets of laminates with symmetric lay-ups of $[\pm\theta]_{2s}$, $[90°/\pm\theta/)0°]_s$, and $[\pm(90°-\theta)/\pm\theta]_s$ have been considered. The full arrangements of these laminates are designated as (from the top layer till the bottom layer), $[\theta/-\theta/-\theta/\theta/\theta/-\theta/-\theta/\theta]$, $[90°/\theta/-\theta/0/0/-\theta/\theta/90°]$, and $[90°-\theta/-90°+\theta/\theta/-\theta/-\theta/\theta/-90°+\theta/90°-\theta]$, where the angle θ can vary from 0° to 90°. Each lamina ply in a respective laminate is assumed to have the same thickness and the same fiber volume fraction. Further, every laminate consists of the same constituent materials. The composites are loaded in the global x direction, which is taken to be of 0°-direction, and the composite stiffness (modulus) and strength are calculated along this direction.

Designed moduli of the angle plied laminates, $[\pm\theta]_{2s}$, made from the carbon/epoxy and the glass/epoxy systems are shown in Figs. 4.29 and 4.31

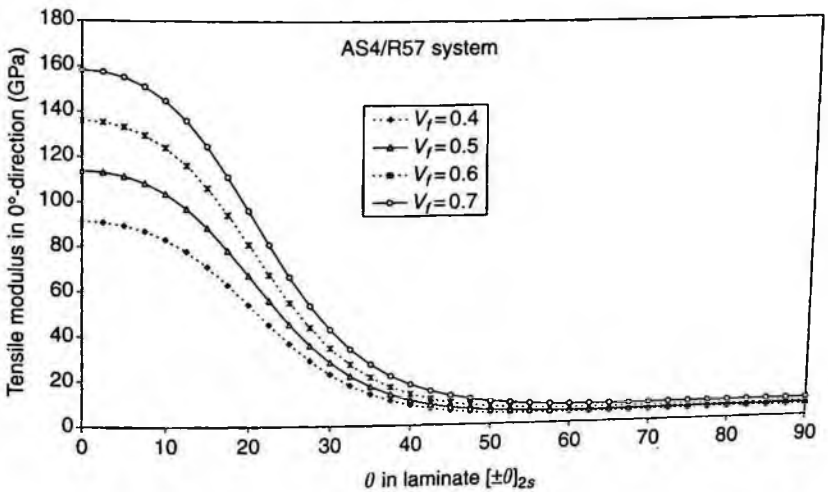


Fig. 4.29 Uniaxial tensile moduli, loaded in 0°-direction, of angle-plied laminates, $[\pm\theta]_{2s}$, made from AS4/R57 material systems, with $V_f = 0.4, 0.5, 0.6$, and 0.7.

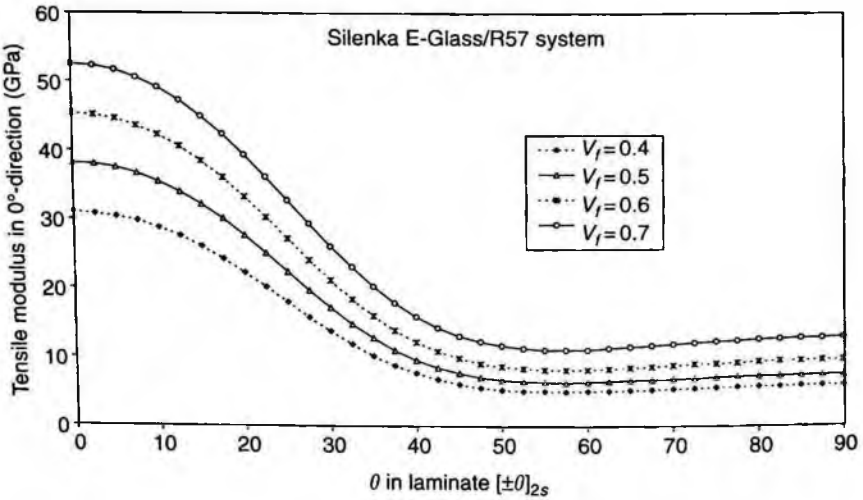


Fig. 4.30 Uniaxial tensile moduli, loaded in 0°-direction, of angle-plied laminates, $[\pm\theta]_{2s}$, made from Silenka E-Glass/R57 material systems, with $V_f = 0.4$, 0.5, 0.6, and 0.7.

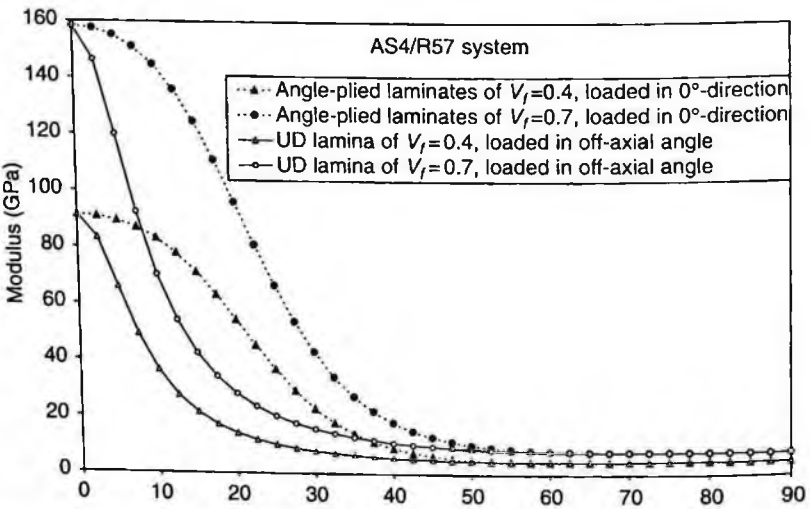


Fig. 4.31 Comparison between the tensile moduli of angle-plied laminates, $[\pm\theta]_{2s}$, loaded in uniaxial (0°-)direction and UD lamina loaded in off-axial direction. The composites are made of AS4/R57 system.

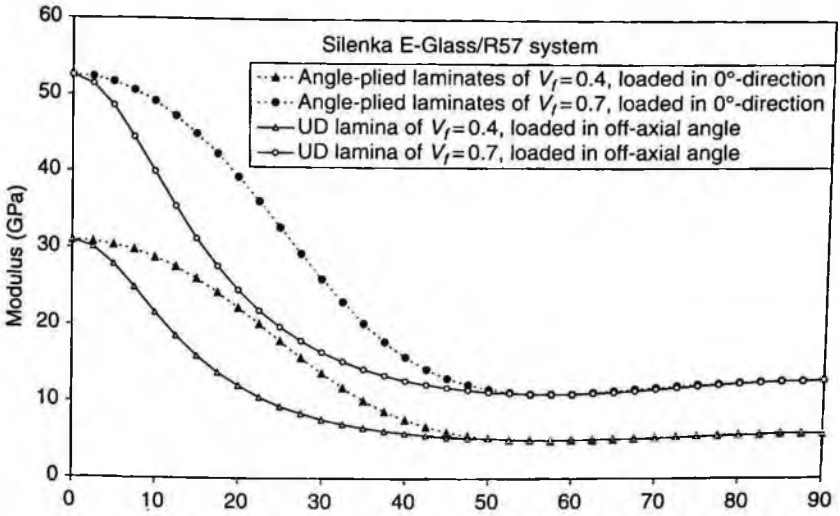


Fig. 4.32 Comparison between the tensile moduli of angle-plied laminates, $[\pm\theta]_{2s}$, loaded in uniaxial (0°-)direction and UD lamina loaded in off-axial direction. The composites are made of Silenka E-Glass/R57 system.

respectively. At the first glance, they have the same variation trend as the corresponding UD laminae loaded in off-axial angles (Figs. 4.20 and 4.21). More detailed comparisons as indicated in Figs. 4.31 and 4.32 do show that the uniaxial (0°-directional) stiffness of an angle-plied laminate coincides with the off-axial stiffness of a UD lamina loaded at the same off-axial angle when this angle is greater than 45°. However, when the angle is less than 40°, apparent difference exists between the uniaxial moduli of angle-plied laminates and the off-axial moduli of the UD laminae. The angle-plied laminates "elevate" the stiffness considerably.

Uniaxial strengths of the angle-plied laminates versus ply angles are graphed in Fig. 4.33 for the carbon/epoxy system and in Fig. 4.34 for the glass/epoxy system. In the case of a UD composite, the fiber failure dominated region is very small, see Figs. 4.27 and 4.28. The composite failure is caused by fiber fracture only when the external load is applied in a very small neighbourhood (less than 5 degree) to the longitudinal direction. Out-of this region, the composite failure is initiated from the matrix fracture. Accordingly, the off-axial strength of the UD composite

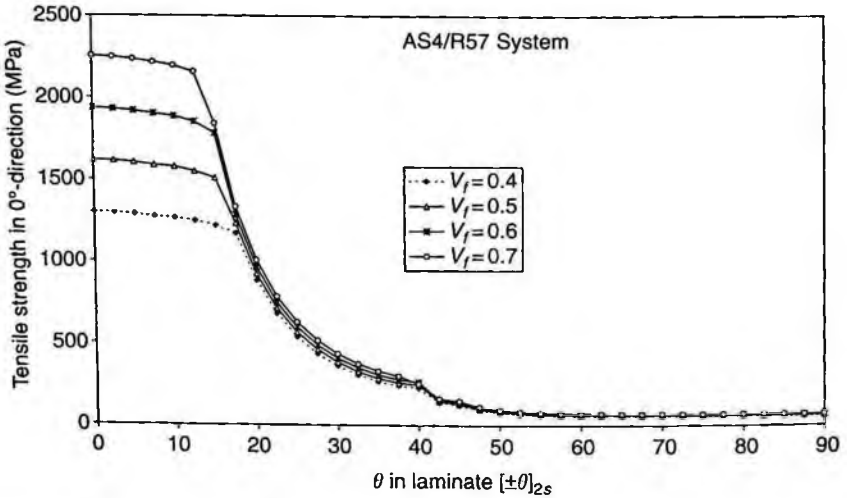


Fig. 4.33 Uniaxial tensile strengths, loaded in 0°-direction, of angle-plied laminates, $[\pm\theta]_{2s}$, made from AS4/R57 material systems, with $V_f = 0.4$, 0.5, 0.6, and 0.7.

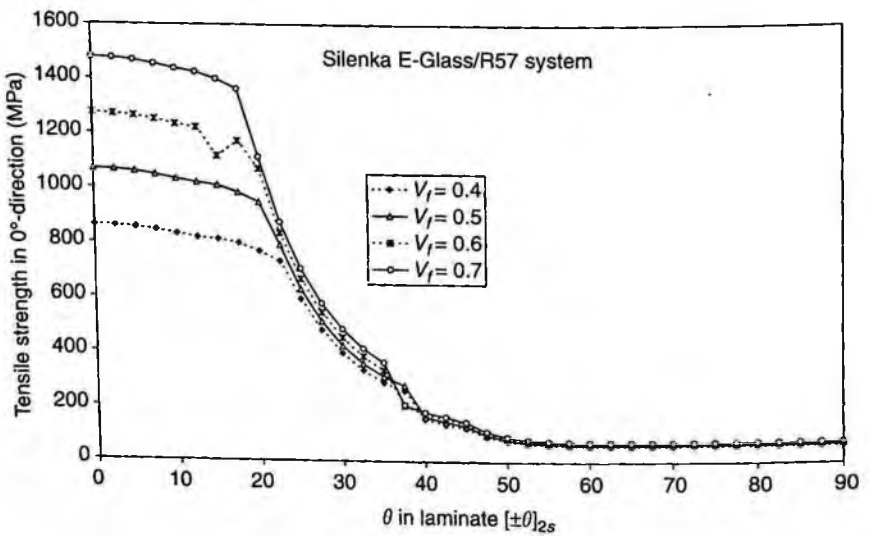


Fig. 4.34 Uniaxial tensile strengths, loaded in 0°-direction, of angle-plied laminates, $[\pm\theta]_{2s}$, made from Silenka E-Glass/R57 material systems, with $V_f = 0.4$, 0.5, 0.6, and 0.7.

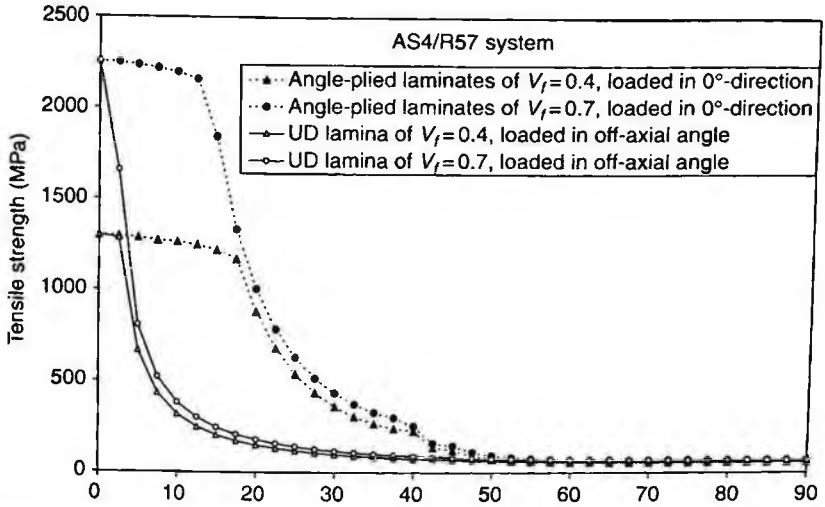


Fig. 4.35 Comparison between the tensile strengths of angle-plied laminates, $[\pm\theta]_{2s}$, loaded in uniaxial (0°-)direction and UD lamina loaded in off-axial direction. The composites are made of AS4/R57 system.

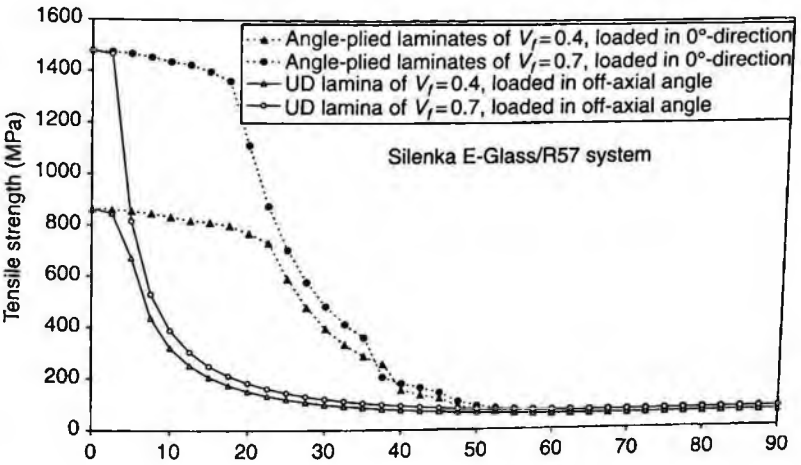


Fig. 4.36 Comparison between the tensile strengths of angle-plied laminates, $[\pm\theta]_{2s}$, loaded in uniaxial (0°-)direction and UD lamina loaded in off-axial direction. The composites are made of Silenka E-Glass/R57 system.

reduces significantly when the off-axial angle is greater than 5°. For angle-plied laminates, however, the fiber failure dominated region can be as large as more than 12.5°. In this region, the composite load carrying capacity is the highest. Comparison shown in Figs. 4.35 and 4.36 clearly indicates that the laminated composites have superior load carrying ability over the UD composites. When the off-axial angle is even larger, equal to or greater than 40°, no difference has been found between the off-axial strength of the UD lamina and the uniaxial strength of the angle-plied laminate. This means that the design of a composite as an angle-plied laminate, such as through a filament winding fabrication, is efficient only when the external load is applied in a relative small region (within $\pm 12.4° \sim \pm 22.4°$, depending on fiber content and on the fiber and matrix modular ratio) with respect to the unidirectional direction.

Designed stiffnesses and strengths of the other laminate arrangements, i.e., laminates $[90°/\theta/-\theta/0°/0°/-\theta/\theta/90°]$ and $[90°-\theta/-90°+\theta/\theta/-\theta/-\theta/\theta/-90°+\theta/90°-\theta]$ with varied angles θ, are shown in Fig. 4.37 through Fig. 4.44. In general, if loaded in all possible directions, the arrangement $[90°/\theta/-\theta/0°/0°/-\theta/\theta/90°]$ shows some superiority. However, different choice of the angle θ makes the resulting composite have different load carrying ability.

In order to further investigate the load carrying capacity of a resulting laminated composite per laminate lay-up, several typical composites from each set of arrangements, $[\pm\theta]_{2s}$, $[90°/\pm\theta/0°]_s$, and $[\pm(90°-\theta)/\pm\theta]_s$, have been designed to sustain varied biaxial loads. The results of the AS4/R57 material system are plotted in Figs. 4.45 to 4.47. It appears that the arrangement $[\pm\theta]_{2s}$ is the worst for carrying varied biaxial loads. However, these laminates are suitable to carry on loads arranged in near close to the longitudinal direction, especially for a low angle arrangement (i.e. θ is small). Of the three sets of arrangements, $[\pm\theta]_{2s}$, $[90°/\pm\theta/0°]_s$, and $[\pm(90°-\theta)/\pm\theta]_s$, only one set always has an optimal lay-up. Figure 4.48 clearly indicates that as long as the composite is subjected to an in-plane load, the lay-up $[\pm 90°/\pm 0°]_s$ is the best among all the possible arrangements of $[\pm(90°-\theta)/\pm\theta]_s$.

It should be realized that while the composite modulus (stiffness) generally increases with the increase of fiber content, this is not always true for the composite strength. The reason is that the composite stiffness is linearly

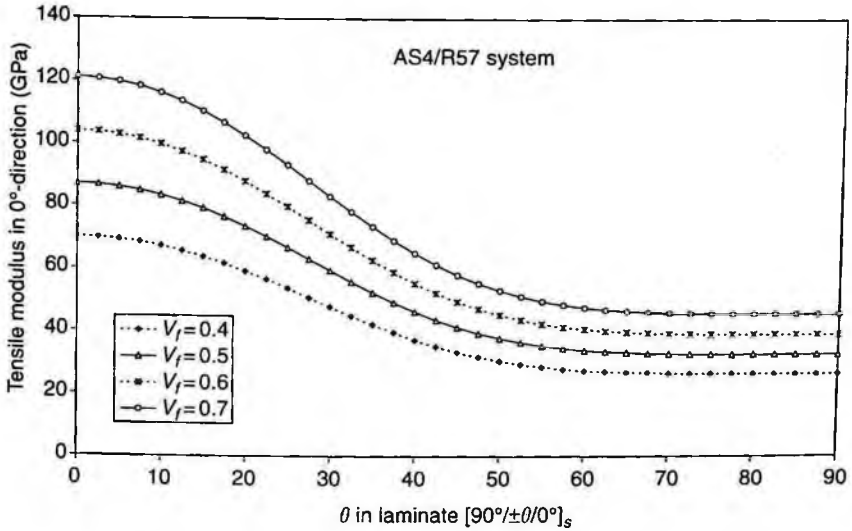


Fig. 4.37 Uniaxial tensile moduli, loaded in 0°-direction, of angle-plied laminates, $[90°/\pm\theta/0°]_s$, made from AS4/R57 material systems, with $V_f = 0.4$, 0.5, 0.6, and 0.7.

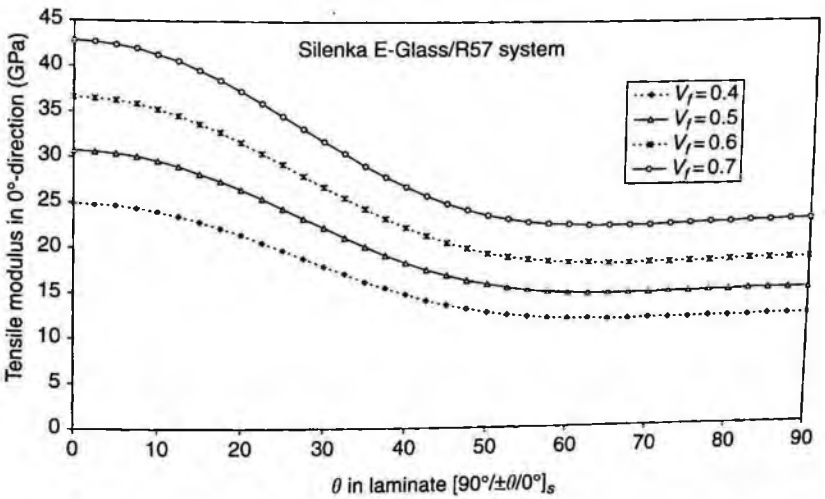


Fig. 4.38 Uniaxial tensile moduli, loaded in 0°-direction, of angle-plied laminates, $[90°/\pm\theta/0°]_s$, made from Silenka E-Glass/R57 material systems, with $V_f = 0.4$, 0.5, 0.6, and 0.7.

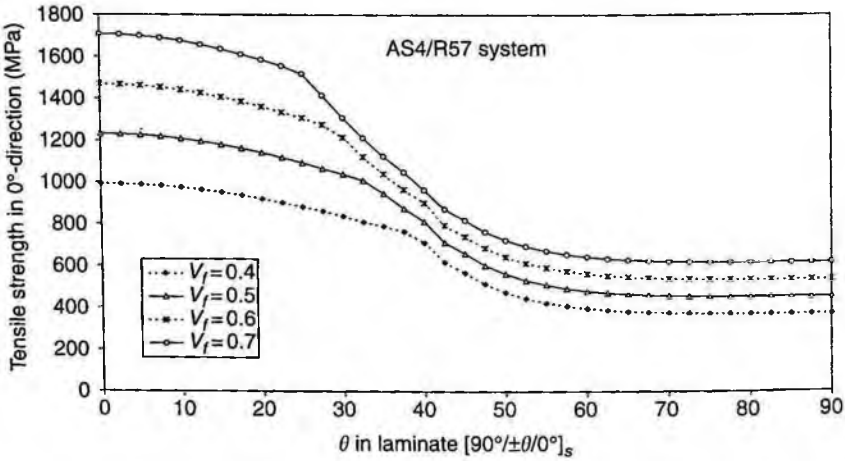


Fig. 4.39 Uniaxial tensile strengths, loaded in 0°-direction, of angle-plied laminates, $[90°/\pm\theta/0°]_s$, made from AS4/R57 material systems, with $V_f = 0.4$, 0.5, 0.6, and 0.7.

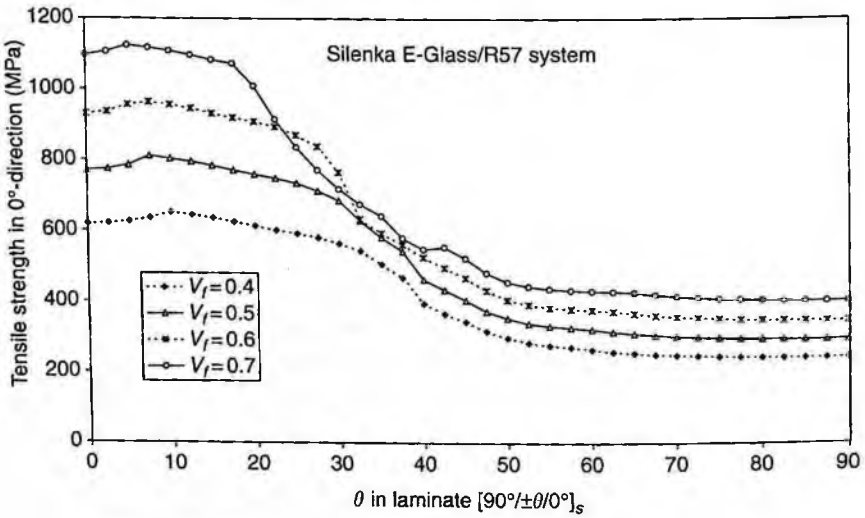


Fig. 4.40 Uniaxial tensile strengths, loaded in 0°-direction, of angle-plied laminates, $[90°/\pm\theta/0°]_s$, made from Silenka E-Glass/R57 material systems, with $V_f = 0.4$, 0.5, 0.6, and 0.7.

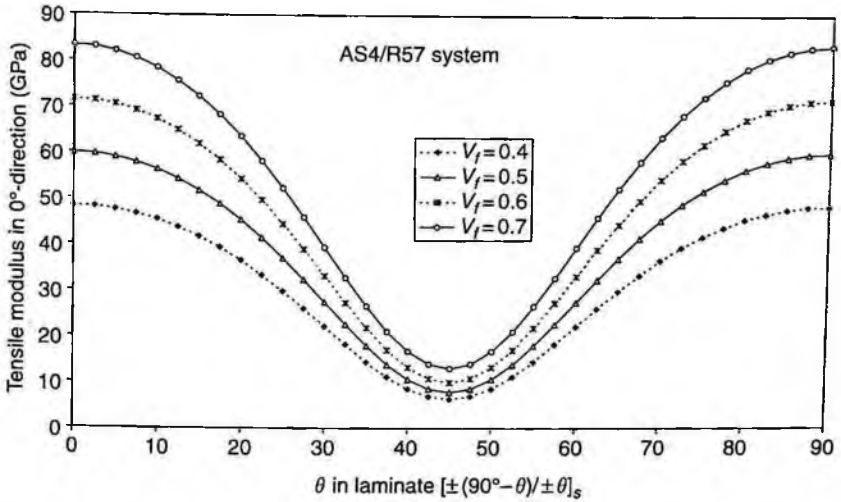


Fig. 4.41 Uniaxial tensile moduli, loaded in 0°-direction, of angle-plied laminates, $[\pm(90°-\theta)/\pm\theta]_s$, made from AS4/R57 material systems, with $V_f = 0.4, 0.5, 0.6$, and 0.7.

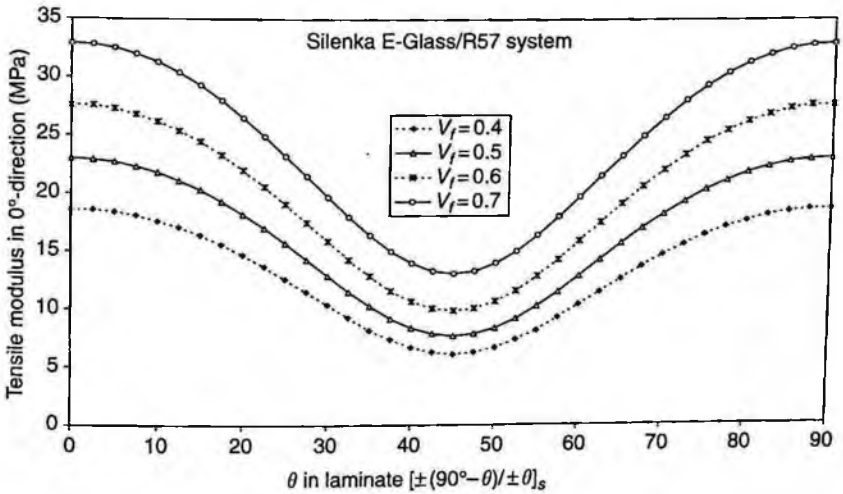


Fig. 4.42 Uniaxial tensile moduli, loaded in 0°-direction, of angle-plied laminates, $[\pm(90°-\theta)/\pm\theta]_s$, made from Silenka E-Glass/R57 material systems, with $V_f = 0.4, 0.5, 0.6$, and 0.7.

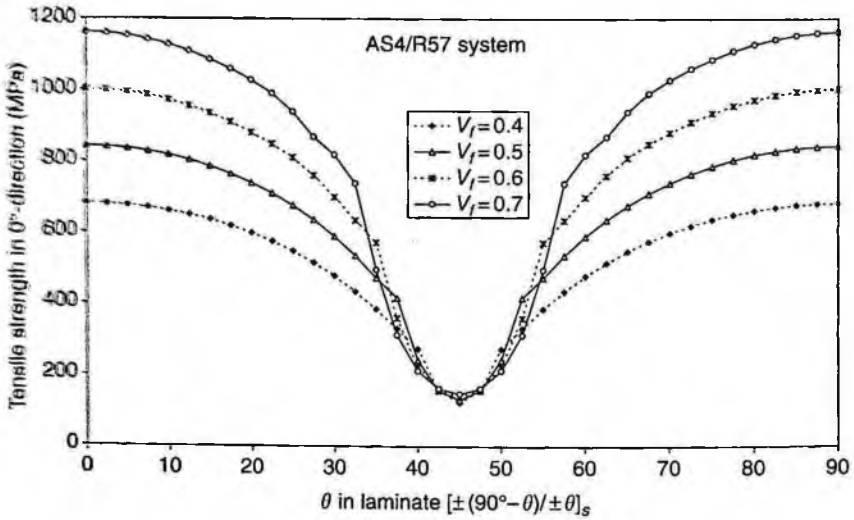


Fig. 4.43 Uniaxial tensile strengths, loaded in 0°-direction, of angle-plied laminates, $[\pm(90^\circ - \theta)/\pm\theta]_s$, made from AS4/R57 material systems, with $V_f = 0.4$, 0.5, 0.6, and 0.7.

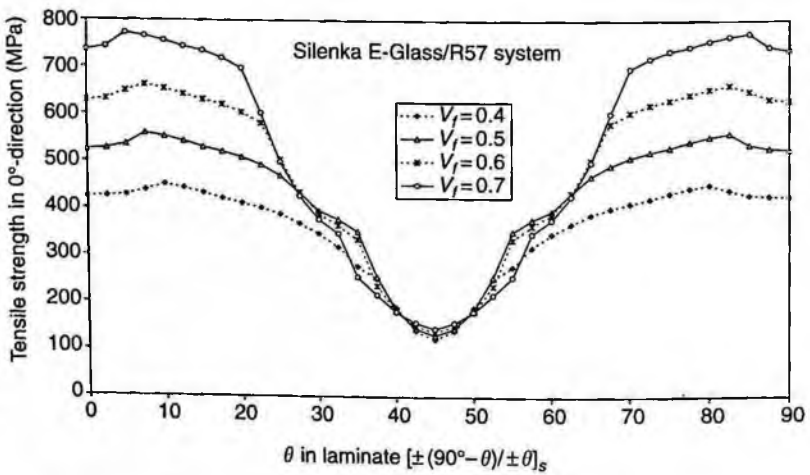


Fig. 4.44 Uniaxial tensile strengths, loaded in 0°-direction, of angle plied laminates, $[\pm(90^\circ - \theta)/\pm\theta]_s$, made from Silenka E-Glass/R57 material systems, with $V_f = 0.4$, 0.5, 0.6, and 0.7.

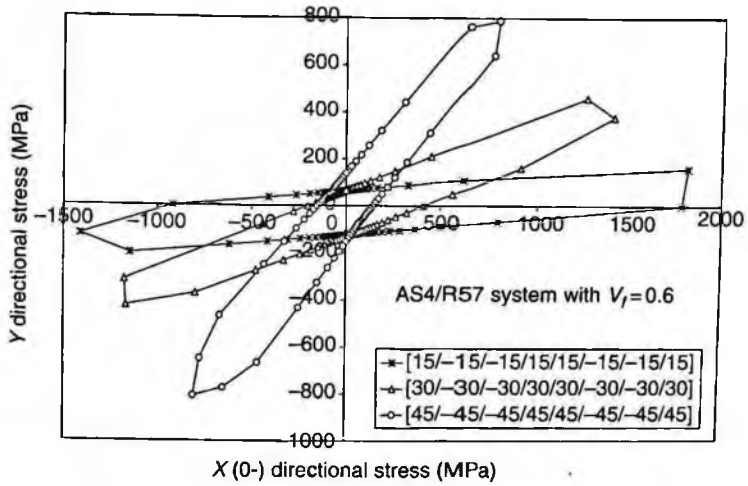


Fig. 4.45 Failure envelops of three typical angle-plied laminates, $[\pm 15°]_{2s}$, $[\pm 30°]_{2s}$, and $[\pm 45°]_{2s}$, made from AS4/R57 system subjected to biaxial load conditions. The composite is safe when loaded within the envelop.

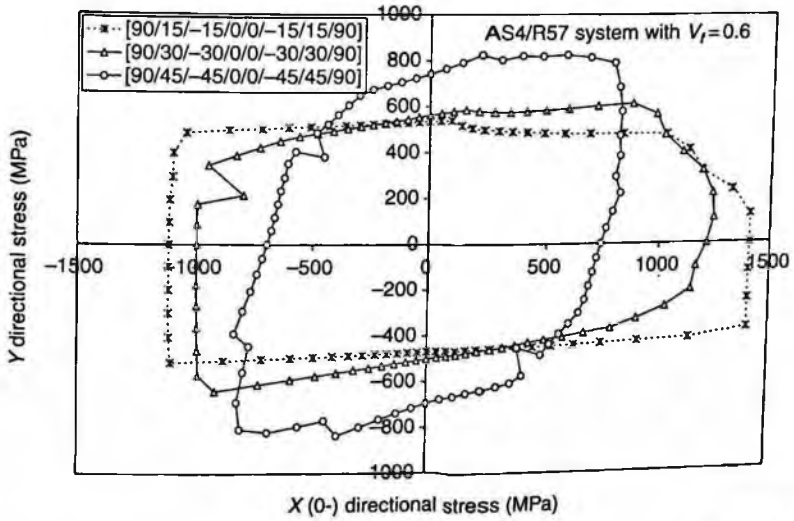


Fig. 4.46 Failure envelops of three typical angle-plied laminates, $[90°/\pm 15°/0°]_s$, $[90°/\pm 30°/0°]_s$, and $[90°/\pm 45°/0°]_s$, made from AS4/R57 system subjected to biaxial load conditions. The composite is safe when loaded within the envelop.

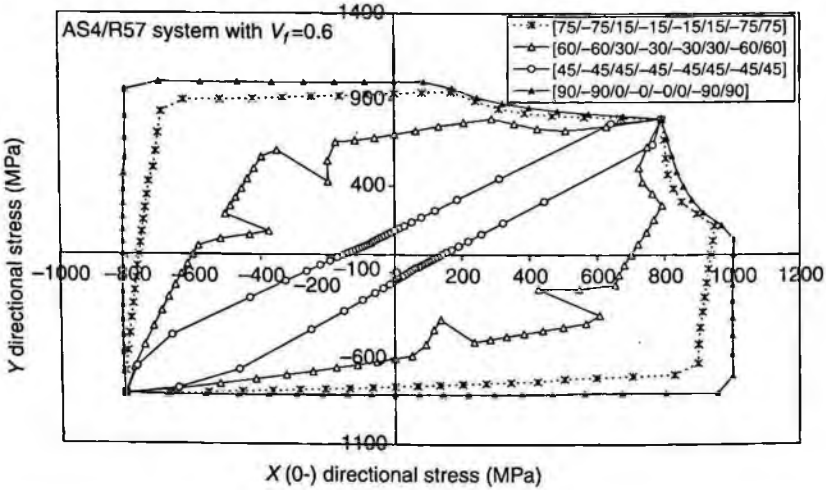


Fig. 4.47 Failure envelops of four typical angle-plied laminates, $[\pm 75°/\pm 15°]_s$, $[\pm 60°/\pm 30°]_s$, $[\pm 45°/\pm 45°]_s$, and $[\pm 90°/\pm 0°]_s$, made from AS4/R57 system subjected to biaxial load conditions. The composite is safe when loaded within the envelop.

proportional to the constituent moduli, whereas the composite strength is not. The matrix plastic deformation makes a laminated composite have a more complicated load distribution to each lamina to laminate.

4.8.3 *Woven and braided fabric composites*

Compared with UD lamina and multidirectional tape laminates, woven and braided fabric composites have better out-of-plane stiffness, strength, and toughness properties, lower fabrication costs, and easier handling for good production quality. Design of these composites is more difficult, and can be influenced by far more parameters. In this section, we only consider the basic weave and braid pattern, i.e. a plain weave or a diamond braid preform. Further, the fabric is considered to consist of two sets of yarns, the fill yarns and the warp yarns. The yarn architectures in the fabric can be described using equations. Thus, only four fabric geometric parameters are required. They are the yarn spacing (width), the yarn thickness, the inter-yarn gap, and the braiding angle of the fabric. A braided fabric composite can be regarded as a special type of the woven composite with a braiding angle of 45°, from a theoretical analysis point of view. It is noted that the first three geometric

parameters, i.e. the yarn spacing, the yarn thickness, and the inter-yarn gap, are related with each set of yarns. Therefore, the fill yarn may have different values of them from the warp yarn. Furthermore, the overall fiber volume fraction of the composite, V_f, and the composite thickness, h_m, also have effects on the mechanical performance of the woven and braided composites. Their effects are reflected by the fiber packing density, which determines the impregnated yarn (lamina) properties in the local coordinate system, and the relative amount of pure matrix in the representative volume element.

For illustration purpose, we choose the following yarn parameters in our design:

$$a_1 = a_2 = 2.0\,\text{mm}, \quad t_1 = t_2 = 0.6\,\text{mm}, \quad \text{and} \quad g_1 = g_2 = 0$$

Further, let $h_m = 0$. These conditions imply that the composites considered will have the same thickness, although different fiber volume fractions can be assumed.

Two kinds of material systems are employed: one is the AS4/R57 carbon/epoxy and the other is the Silenka/R57 glass/epoxy systems. Designed results are shown in Fig. 4.48 through Fig. 4.49.

In Figs. 4.48–4.51, the in-plane moduli and tensile strengths of the two kinds of woven composites are plotted. It is seen that the stiffness and strength properties of the woven composites are the highest along the fiber yarn (fill or warp) direction, and decrease rapidly when the load direction is off the yarn axis aligned. This decrease is more significant when the modular ratio between the two constituent materials is higher. Furthermore, the decrease in strength is more distinct than that in stiffness. The figures also indicate that increasing the fiber volume fraction only have apparent effect on the composite stiffness and strength in the yarn (fill or warp yarn) axial direction only. In most other off-aligned directions, the ultimate strength of the woven composite does not increase much when the fiber content assumes larger values. This means that design of a moderate volume fraction is enough for a woven fabric composite. Fabrication of woven fabric composites with higher fiber volume fractions, which may costs more, does not bring much benefit into composite load carrying capacity if this load is not applied in a yarn axial direction.

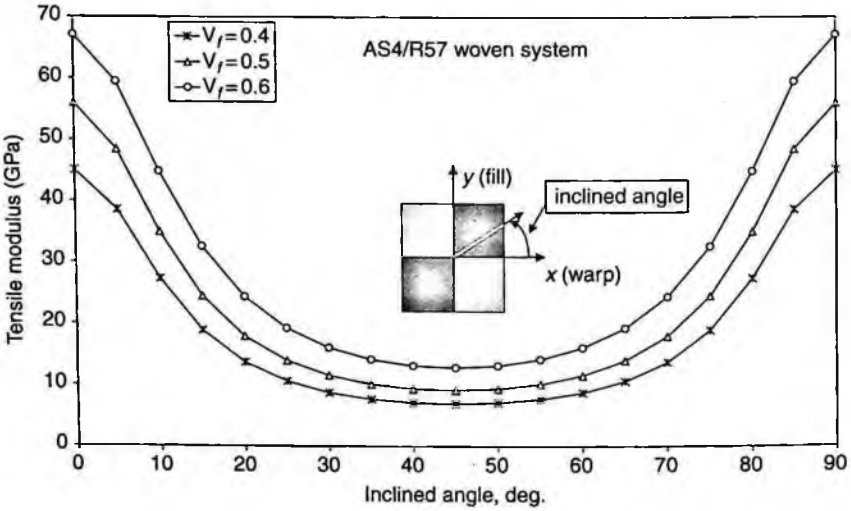


Fig. 4.48 Uniaxial tensile moduli of AS4/R57 woven composites versus inclined angles.

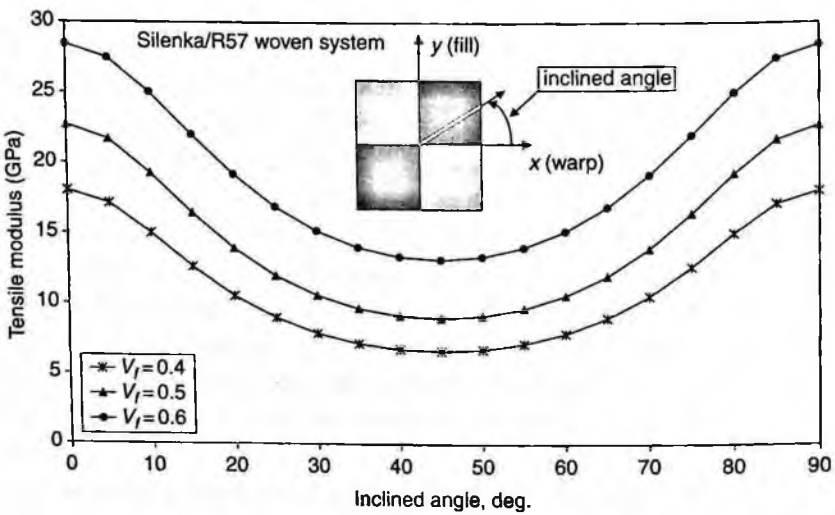


Fig. 4.49 Uniaxial tensile moduli of Silenka/R57 woven composites versus inclined angles.

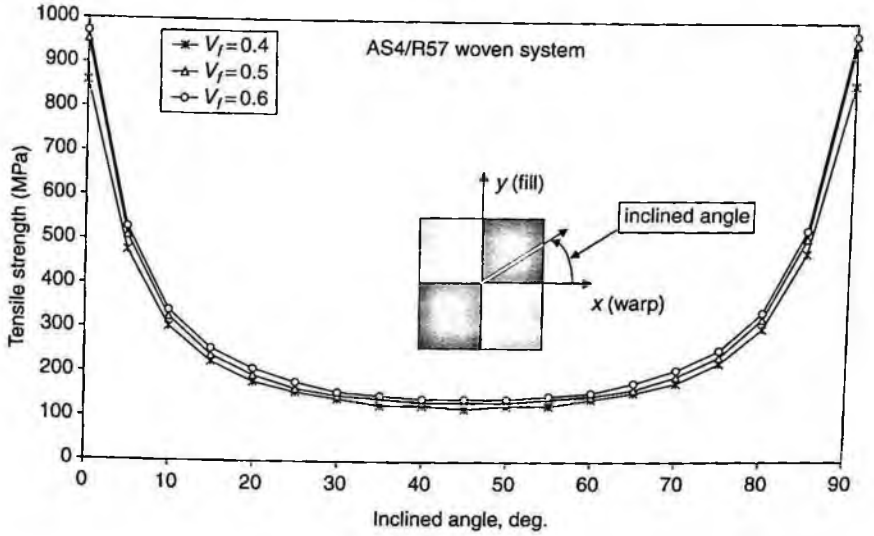


Fig. 4.50 Uniaxial tensile strengths of AS4/R57 woven composites versus inclined angles.

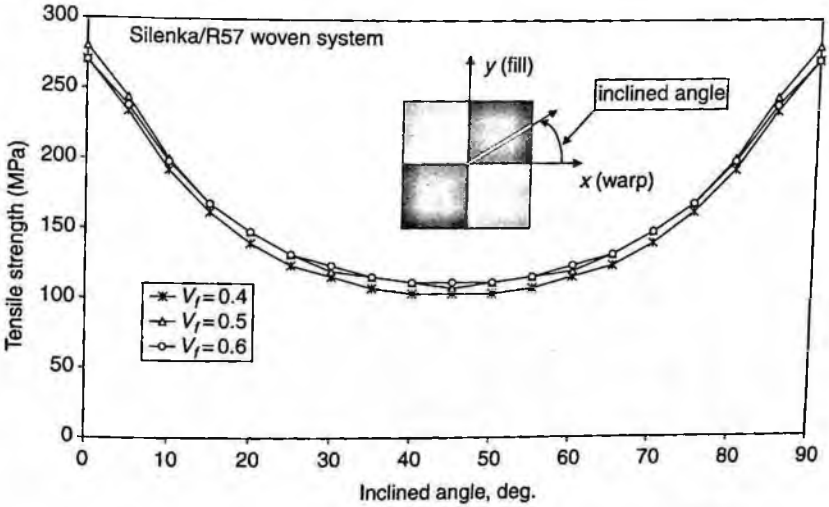


Fig. 4.51 Uniaxial tensile strengths of Silenka/R57 woven composites versus inclined angles.

Strength envelopes of the carbon/epoxy and the glass/epoxy composites with several fiber volume fractions when subjected to biaxial in-plane loads are plotted in Figs. 4.52 and 4.53 respectively. In general, the carbon/epoxy composite has a very much higher load carrying capability than the glass/epoxy composite. The reason is attributed more to the high modular ratio rather than to the high strength ratio between the carbon/epoxy system, relative to the glass/epoxy system. This can be understood from Fig. 4.53, in which only limited load combinations have attained their highest level. This means that for the majority of the biaxial load combinations, the fiber material in the glass/epoxy woven composite has not sustained an enough load share before the composite failure. The load sharing capacity of a constituent material is directly related with its stiffness. One more interesting phenomenon is shown in Fig. 4.52, in which when a compression load in the x-direction is equal to the tensile load in the y-direction, the composite ultimate strength is significantly lower than the other neighboring load combinations. This means that under the given load combination, the composite failure is due to the matrix fracture, whereas the other neighboring load combinations force the fibers to carry more load share. Furthermore, comparisons between Figs. 4.52 and 4.53 with

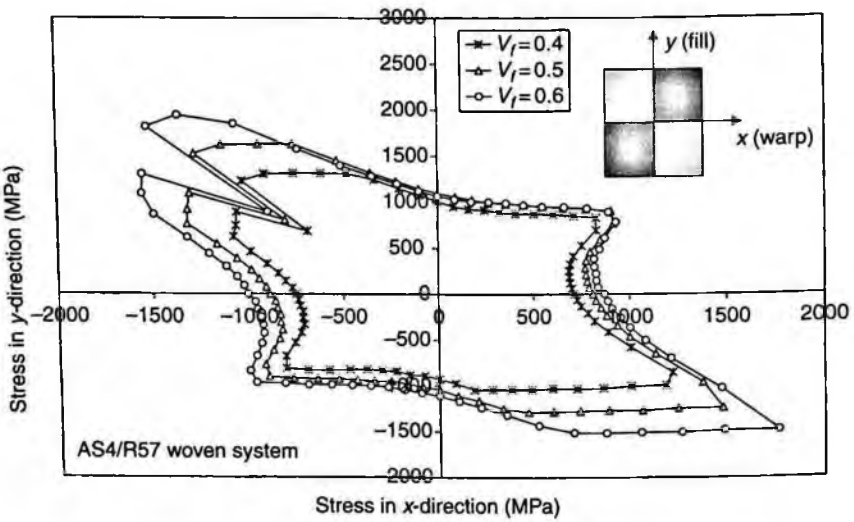


Fig. 4.52 Strength envelops of AS4/R57 woven composites subjected to biaxial in-plane loads.

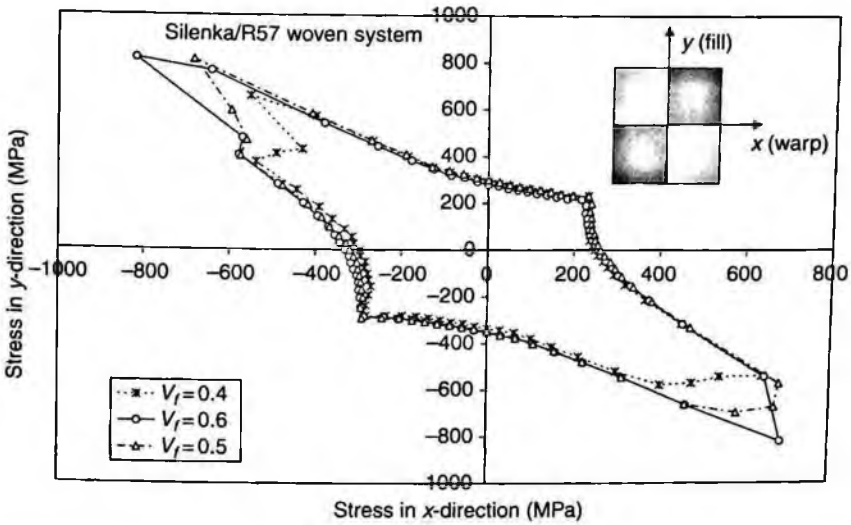


Fig. 4.53 Strength envelops of Silenka/R57 woven composites subjected to biaxial in-plane loads.

Figs. 4.50 and 4.51 clearly indicate that the composite load carrying capacity when subjected to arbitrary biaxial load combinations, as long as these loads are along the weaving yarn directions, is significantly higher than that when subjected to an off-yarn axis aligned load condition.

Longitudinal tensile moduli and strengths of braided fabric composites made of the two material systems versus braiding angles are graphed in Fig. 4.54 through Fig. 4.57 respectively. As can be understood, the stiffness and strength decrease significantly as the braiding angle becomes larger and larger, as long as this angle does not exceed 45°. When the braiding angle is larger than 45°, the stiffness and strength in the longitudinal direction almost keep unchanged. As a whole, the composite longitudinal stiffness is proportional to the increase of the fiber volume fraction, and inversely proportional to the braiding angle. However, there are some exceptions, as indicated in Fig. 4.55. The tensile modulus of the glass/epoxy braid composite with a fiber volume fraction of 0.4 or 0.5 and a braiding angle of 25° is equal to or even higher than that of the braid composite with a braiding angle of 20°. The reason is that when a longitudinal tensile load is applied, the resulting matrix material in the braid composite of 20°

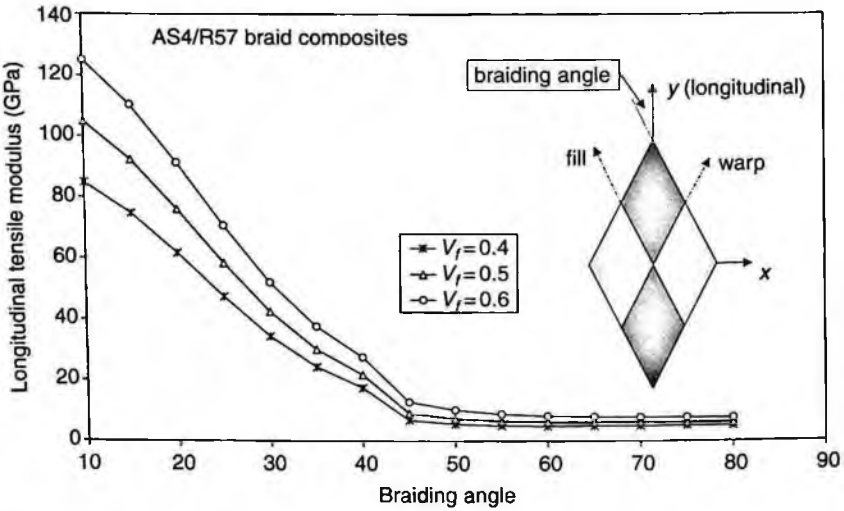


Fig. 4.54 Longitudinal tensile moduli of AS4/R57 braided composites versus braiding angles.

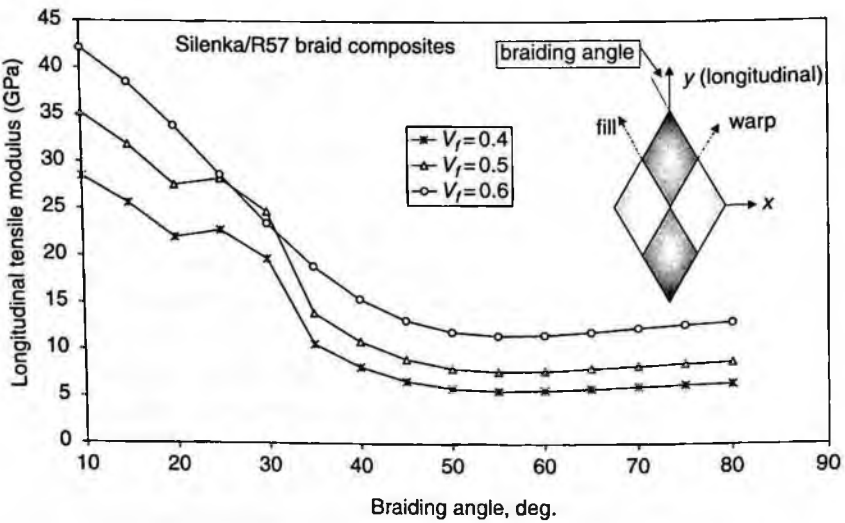


Fig. 4.55 Longitudinal tensile moduli of Silenka/R57 braided composites versus braiding angles.

braiding angle is subjected to an essential tension. On the other hand, the matrix material in the 25° braid composite is under an internal compression when the composite is subjected to a longitudinal tensile load. As the epoxy matrix is stiffer under compression than under tension, the overall modulus of the composite follows the same trend.

Comparing Figs. 4.56–4.57 with Figs. 4.27–4.28 and Figs. 4.33–4.34, it can be seen that the in-plane load carrying ability of braid composites is superior to UD composites, but worse than angle-plied laminates. However, if loaded in only the longitudinal direction, the UD composite can sustain a larger amount of load than the braid composite does. On the other hand, the braid composite can generally prevent the composite from delimination, whereas an angle-plied laminate is prone to delimination which may significantly reduce the composite load carrying capacity.

4.8.4 *Knitted fabric composites*

In comparison with woven and braided preforms, knitted fabrics have looped structures. These looped structures can make a resulting composite deform easily. On the other hand, the reinforcing efficiency of the fibers

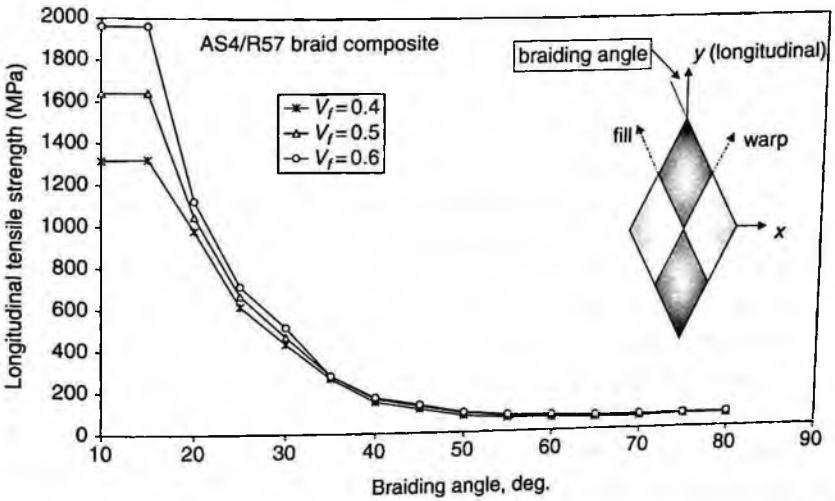


Fig. 4.56 Longitudinal tensile strengths of AS4/R57 braided composites versus braiding angles.

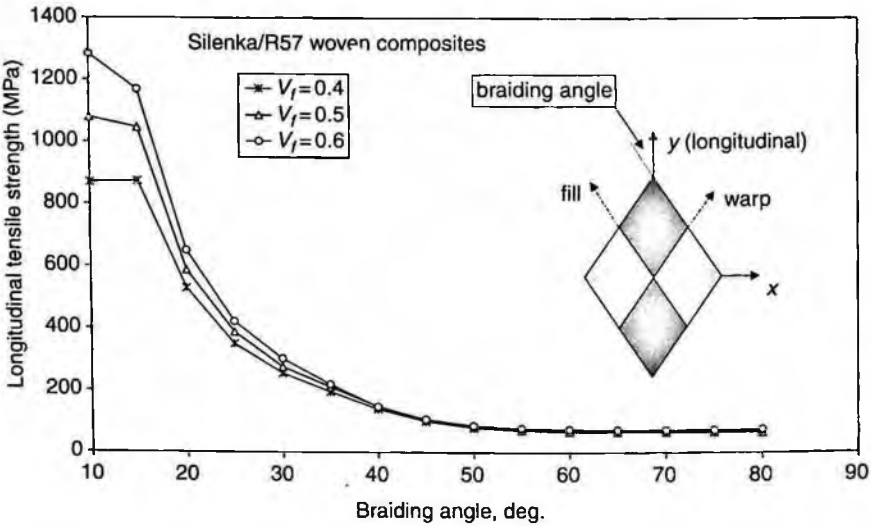


Fig. 4.57 Longitudinal tensile strengths of Silenka/R57 braided composites versus braiding angles.

in the composite can be significantly reduced due to the looped structures. As we understand, a composite can sustain the maximum amount of load if it is applied in the fiber axial direction. A UD composite exhibits the highest stiffness and strength in its longitudinal direction. Since no straight fibers are arranged in a common knitted structure, it can be expected that the stiffness and ultimate strength of the resulting knitted fabric reinforced composite cannot gain very high values along any direction.

Two kinds of composites made by reinforcing plain weft knitted AS4 carbon fibers and Silenka glass fibers in R57 epoxy matrix have been designed. The mechanical properties of these composites can be pre-estimated as long as the fabric geometric parameters have been given. Plain knitted fabrics have two principal directions, the wale (0°) and the course (90°) directions. Four geometric parameters are enough. They are the wale number (W), the course number (C), the fiber yarn diameter (d), and the overall fiber volume fraction (V_f) in the composite. Suppose the first three parameters remain unchanged, namely,

$$W = 3.9 \text{ cycles/cm}, \quad C = 3.4 \text{ cycles/cm}, \quad \text{and} \quad d = 0.058 \text{ cm}.$$

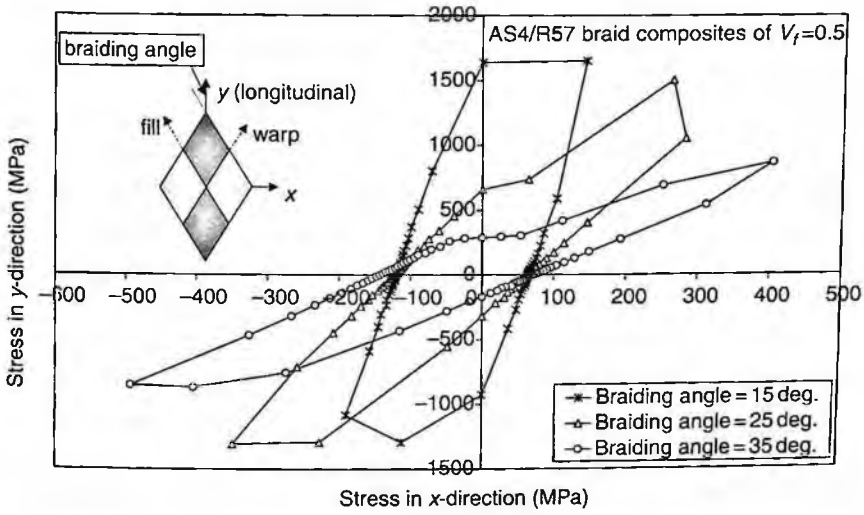


Fig. 4.58 Strength envelops of AS4/R57 braid composites subjected to biaxial in-plane loads.

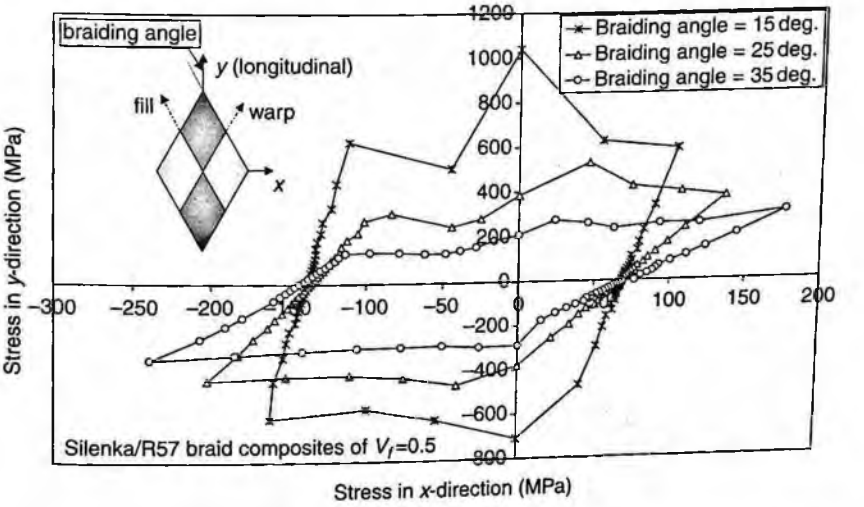


Fig. 4.59 Strength envelops of Silenka/R57 braid composites subjected to biaxial in-plane loads.

The stiffness and strength of the composites having several fiber volume fractions ($V_f = 0.1$, 0.15, 0.2, and 0.25) at different load conditions are plotted in Figs. 4.60–4.65 respectively.

Figures 4.60 and 4.61 indicate that the stiffness of a knitted fabric composite is the highest in the wale (0°) direction, but gradually decreases to a minimum at the course (90°) direction. The same is true for the composite strength, as shown in Figs. 4.48 and 4.49. Although the knitted AS4/R57 composites have higher stiffness and strength in the wale direction than the Silenka/R57 composites, the difference between them is not significant: only a slight increase has been found. This means that the matrix material plays a dominant role for the overall stiffness and strength of the knitted fabric reinforced composites. The high stiffness and strength properties of the fiber materials seem to be not well reflected in the knitted fabric composites. In fact, the ultimate failure of the knitted composite is initiated by the matrix fracture, giving relatively low ultimate strength to the composite.

Figures 4.64 and 4.65 show bi-axial failure envelopes of the above composites subjected to in-plane loads that are applied in the wale and course directions. The figures show that the composite ultimate strength increases

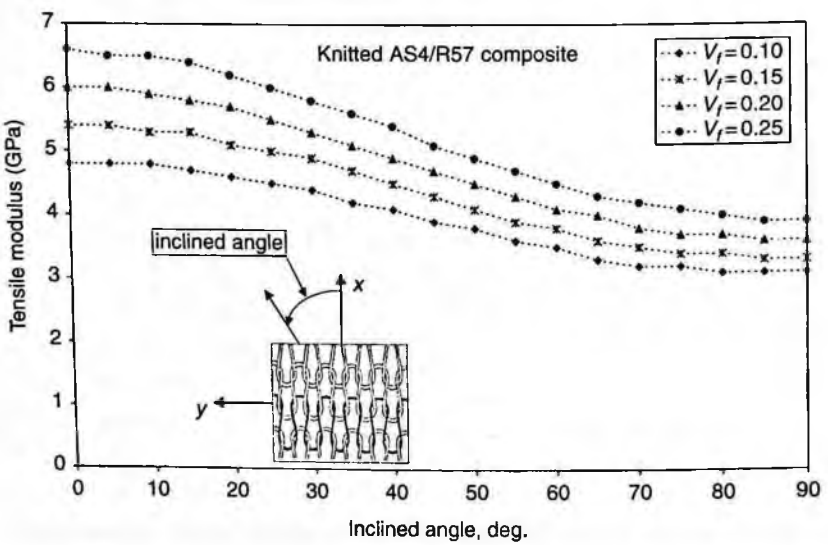


Fig. 4.60 Off-axial tensile moduli of knitted AS4/R57 composites of different fiber volume fractions versus inclined angle.

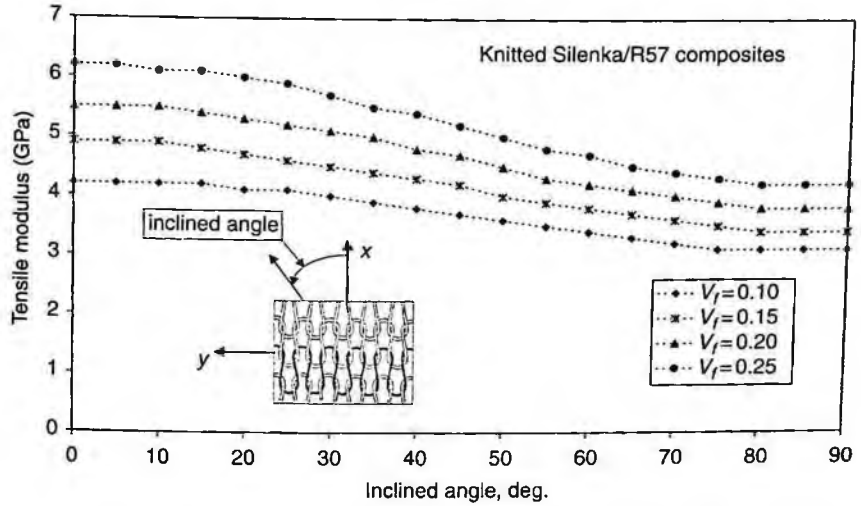


Fig. 4.61 Off-axial tensile moduli of knitted Silenka/R57 composites of different fiber volume fractions versus inclined angle.

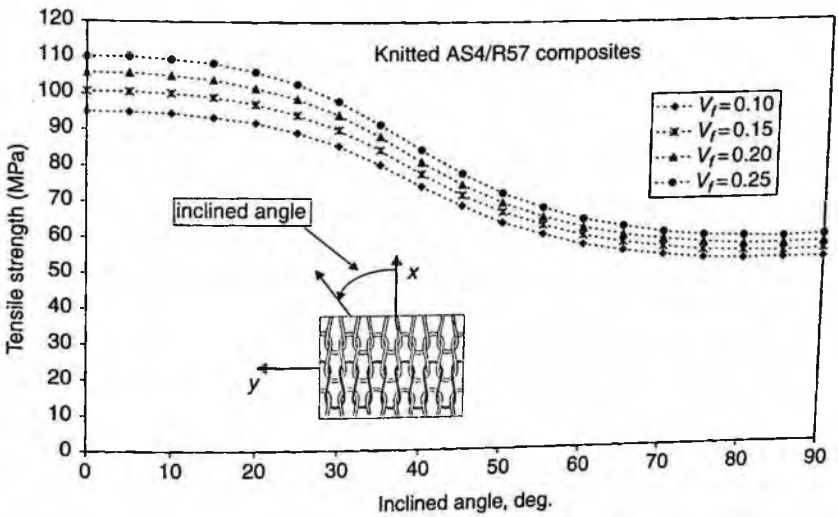


Fig. 4.62 Off-axial tensile strengths of knitted AS4/R57 composites of different fiber volume fractions versus inclined angle.

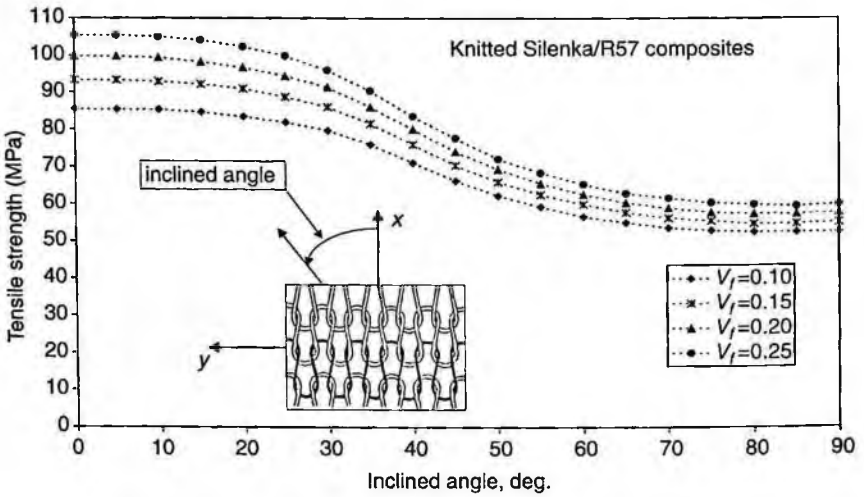


Fig. 4.63 Off-axial tensile strengths of knitted Silenka/R57 composites of different fiber volume fractions versus inclined angle.

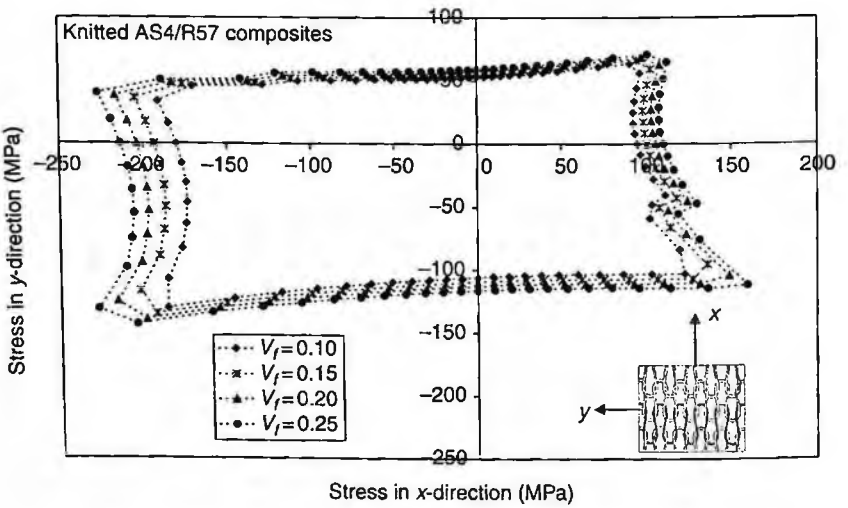


Fig. 4.64 Strength envelops of knitted AS4/R57 composites of different fiber volume fractions subjected to in-plane biaxial loads.

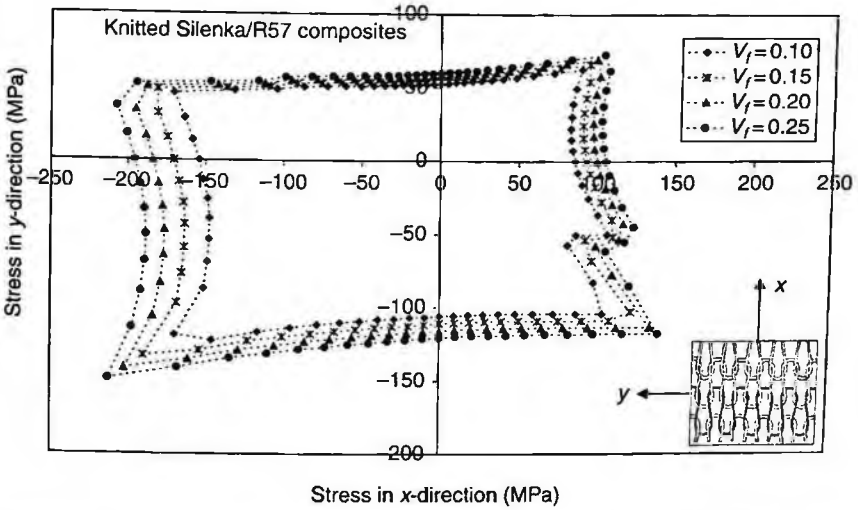


Fig. 4.65 Strength envelops of knitted Silenka/R57 composites of different fiber volume fractions subjected to in-plane biaxial loads.

with the increase of fiber volume fraction, but not very significantly. The effect is more distinct when the composite is subjected to its largest possible compressive load in the wale direction. It should be noted that the present compressive analysis does not consider any buckling effect on the composite ultimate strength.

4.8.5 *Comparisons among different composites*

In the previous sections, composites with various fiber preform reinforcements have been designed in terms of their stiffness and strength properties with respect to the same two material systems, i.e. the AS4/R57 carbon/epoxy and the Silenka/R57 glass/epoxy systems. It would be useful to make a comparison on the mechanical performances of those different preform composites. Only the in-plane stiffness and strength in their respective principal materials directions will be compared. For example, the main direction of a UD composite or a braid composite is along the composite longitudinal direction, whereas the main direction of a knitted fabric composite is along its wale direction. As different fiber preform can achieve different fiber volume fraction, a reasonable comparison should

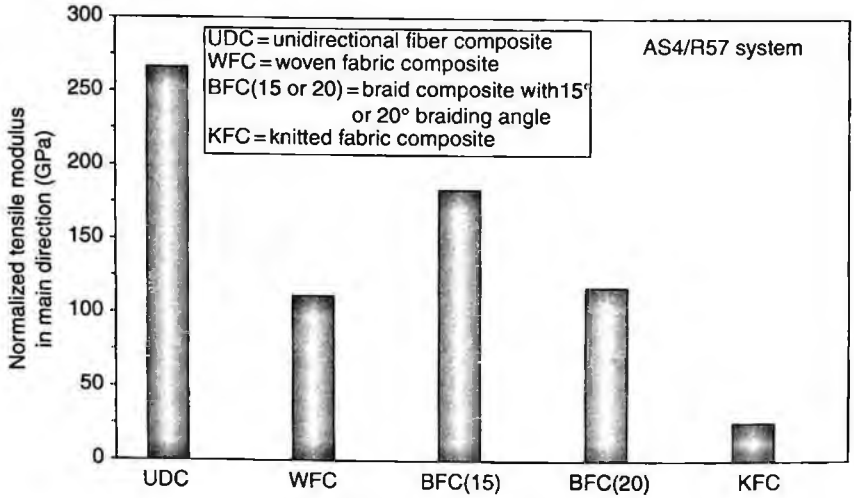


Fig. 4.66 Normalized (with respect to fiber volume fraction) tensile moduli of AS4/R57 composites with different fiber preforms.

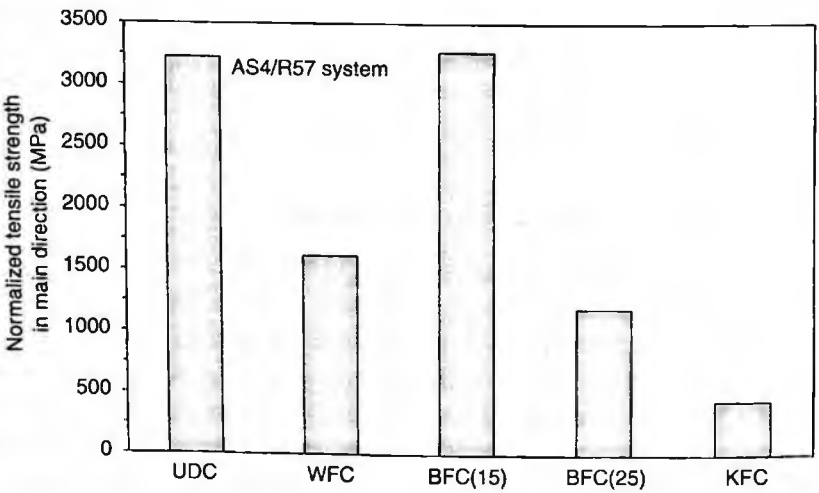


Fig. 4.67 Normalized (with respect to fiber volume fraction) tensile strengths of AS4/R57 composites with different fiber preforms.

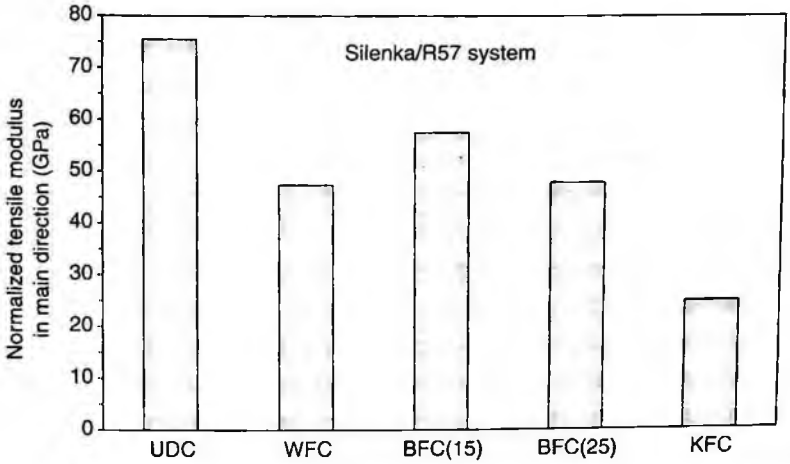


Fig. 4.68 Normalized (with respect to fiber volume fraction) tensile moduli of Silenka/R57 composites with different fiber preforms.

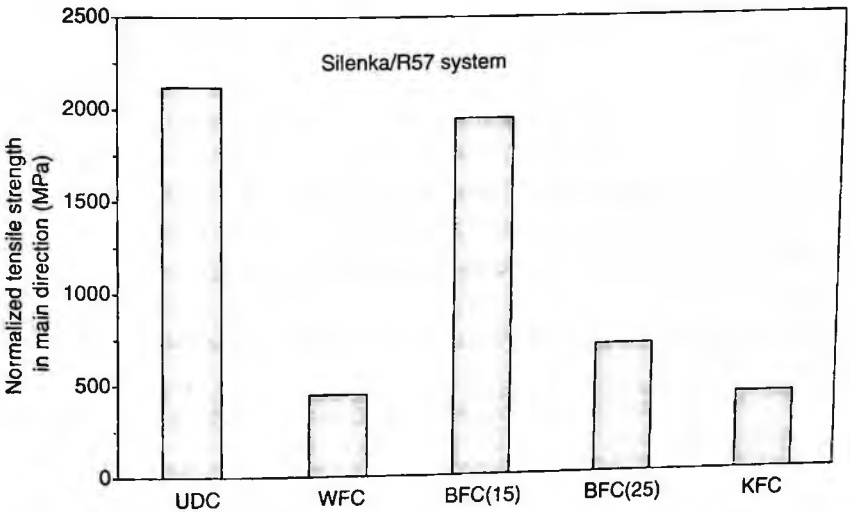


Fig. 4.69 Normalized (with respect to fiber volume fraction) tensile strengths of Silenka/R57 composites with different fiber preforms.

be performed without the effect of this fraction. Thus, the moduli and the strengths are normalized with respect to the specific fiber volume fraction. Results are shown in Figs. 4.66 through 4.69. These figures demonstrate that the composite superiority order in stiffness and strength is arranged as follows: UD composite, braid composite with 15° braiding angle, woven composite, braid composite with 25° braiding angle, and knitted composite. This comparison gives an overall idea for choosing a suitable fibrous preform as reinforcement in biocomposite development. It should be, however, realized that the present comparison only focused on stiffness and strength performances of the composites along their main directions. In a practical design, other issues such as fabrication cost, fabrication defect induced reduction in mechanical properties, the performances along other directions, etc should also be taken into account.

References

T. W. Chou, *Microstructural Design of Fiber Composites*, Cambridge University Press, 1992.

R. F. Gibson, *Principles of Composite Materials Mechanics*, McGraw-Hill Inc., 1994.

R. Hill, *The Mathematical Theory of Plasticity*, Oxford University Press, London, 1950.

L. Hollaway, *Polymer Composites for Civil and Structural Engineering*, Blackie Academic & Professional, 1993.

D. Hull, *An Introduction to Composite Materials*, Cambridge University Press, 1981.

R. M. Jones, *Mechanics of Composite Materials*, Hemisphere Publishing Corporation, 1975.

S. W. Tsai, Structural behavior of composite materials, NASA Contract Report CR-71, 1974.

Chapter 5

DESIGNING WITH COMPOSITE MATERIALS

5.1 Design Considerations

The usage of fibrous composite materials has been increasing in volume and applications. Initially, fibrous composites found application in weight critical aero-space components. Later, the domain enlarged to infrastructure applications with additional performance requirements like, environmental stability, moldability, damage resistance, etc. Today, with biocompatible fibers and matrix systems, fibrous composites are finding application as biomaterial. A number of fibrous composite implants and devices for orthopedic and dental applications have been developed recently.

5.1.1 *Properties*

The properties that a designer generally finds attractive in composite materials are:

(a) Low density: Arises because the density of both the reinforcing fibers and resin matrix are low compared to the metal. This translates directly to patient's convenience in application like external fixators.
(b) High stiffness and strength: This is particularly attractive when combined with low density to give high specific stiffness and strength.

(c) Tailor-ability: With the availability of various performing technologies the reinforcing fibers can be placed at the desired quantity and angle to perform in the most optimal manner.
(d) Aesthetics: With a variety of ceramic fiber and transparent matrix systems, almost all type of color shades can be obtained. This property of composite materials is particularly attractive for dental applications.
(e) Non-corrosive: The polymer matrix composites are less susceptible to corrosion compared to the metal alloys. This inherent property of the matrix translates directly to the composite materials as the fibers are enclosed within the matrix.
(f) Toughness: The polymer matrix composites are not brittle. Hence, the problem of abrasion and fracture is reduced.
(g) Metal allergy: An increasing number of people exhibit an allergic reaction to the presence of metallic devices in the body. The devices made of polymer composites eliminate such allergic reaction.
(h) Chair-side handling: Many of the restoration and fabrication of medical devices need to be done *in situ*. The excellent handling properties of the fibers and matrix systems, capability of the composite to adapt to any shape while they are still green and availability of light curable matrix systems have imparted ease of chair-side handling to fibrous composites.

5.1.2 *Processability*

The composite materials in addition to having many attractive properties also exhibit advantages in terms of economics, manufacture and materials usage. Some of the important advantages in relation to the biomedical device applications are:

(a) Economy of scale: This benefit can occur both in small and large volumes. With small volumes of biomedical products, relatively simple and low cost tooling can be used.
(b) Integration of parts: With the ability to cast or mold complex shapes and devices with a lesser number of components can be fabricated. This translates into reduced fabrication time and wastages.
(c) Surface texture: Textured surfaces can be directly achieved without any additional finishing process.

(d) Near net manufacture: This advantage eliminates additional operation like machining and finishing.

5.1.3 *Functionality*

Product that are made of composite materials can accept wider design brief and performance specifications to achieve innovative functionalities. In most cases, the functional, ergonomic and aesthetic requirements can be considered simultaneously for each of the components. Later, incremental design changes are easier to make with composite materials because for the lower prototyping costs.

5.2 Design Process

It is necessary to have a framework in which each design can function. In the present case, the composite material as a primary material enhances this necessity. From concept to the final design requires a number of steps, where the materials characteristics and manufacturing processes are allowed for in a systematic manner. The time spent on each step depends on the nature of the product and its intended use. Figure 5.1 describes the general design process. Many at times it is desirable to consider an alternative design with another material, if the advantages of composite materials are not obvious.

5.2.1 *Conceptual design*

The conceptual design emerges from the design brief, regulations and guidance elements such as the existing case studied, standards and research literature. A well-defined design brief is essential for the success of the design. The effect of the design brief on composite materials is important particularly with respect to the multiple functionality requirements. Many of the parameters can range by a factor of 100. The design brief therefore must contain comprehensive information on performance requirement, initial cost and operational costs. The regulation and guidance elements help the designer to uncover reasons for rejecting certain design configurations, materials or processes. The design options derived from the elimination process with the addition of a number of other components that

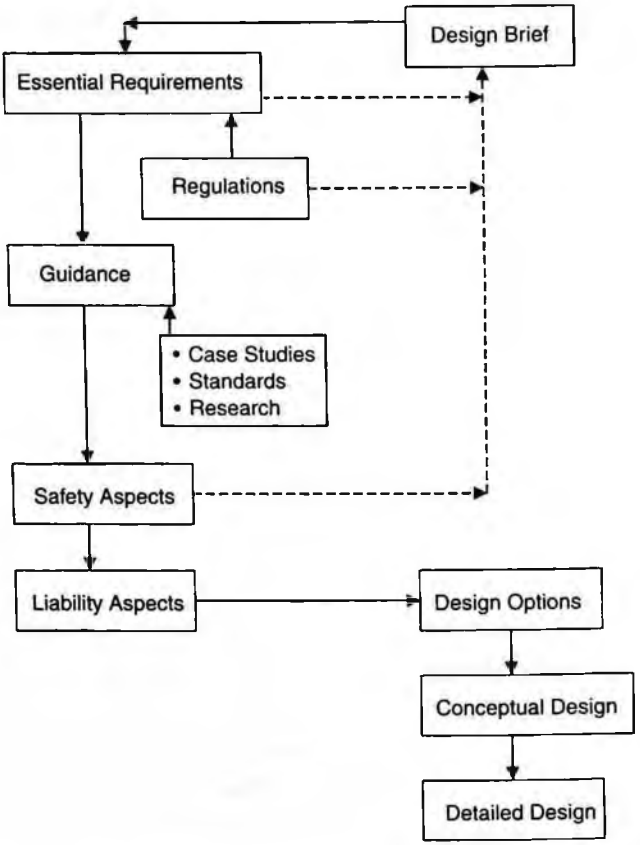


Fig. 5.1 Design process chart.

would constitute the initial input for the conceptual design. The various components will depend on biocompatibility requirements, biomechanical requirements, clinical requirements, materials requirements, fabrication processes and cost factors. If any of the functional, ergonomic and the aesthetic requirements cannot be met, then both the components and/or the design options need to be reconsidered. This is the stage in the design process that makes the greatest demands and allows for the maximum scope for striking improvements. Theory and practical knowledge are brought together during this phase. All conflicting arguments at this stage must be resolved before proceeding further. Figure 5.2 depicts a general conceptual design process.

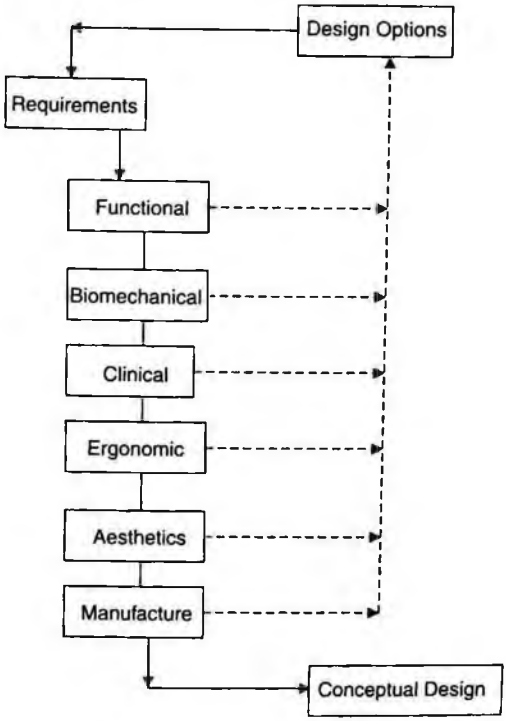


Fig. 5.2 Conceptual design process.

The cost of design is another aspect that needs to be addressed in the first phase. The design cost should form an asset rather than an overhead. Consequently, it should be costed and included in the overall costs. A design team working to a properly costed program is better than a team working to an open budget.

5.2.2 *Detailed design*

This step involves both the embodiment design and alterations required as a result of the validation of the embodiment design and any other changes in the design brief (Fig. 5.3). The outcome from the conceptual design is checked for its fitness of purpose. A weighed analysis is made at this stage and based on the deviation from the purpose requirements, a reassessment of the design is done. During the weighed analysis, higher weightage is

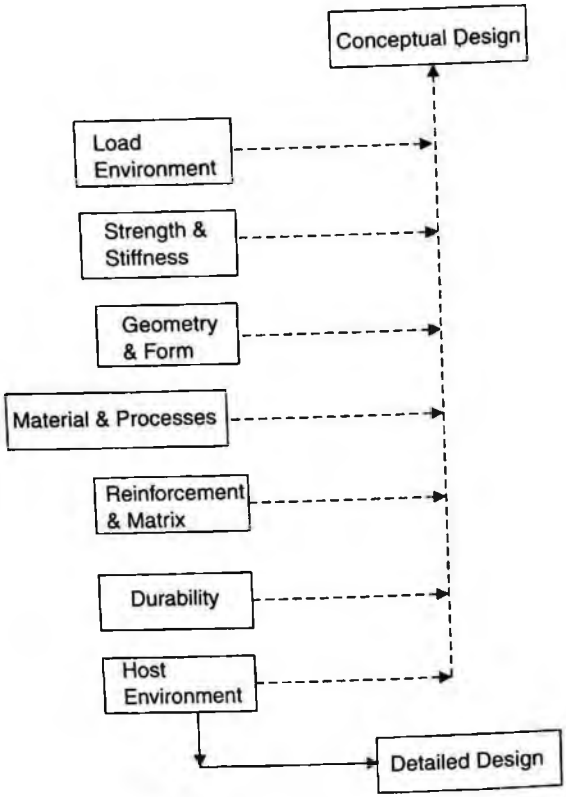


Fig. 5.3 Detailed design process.

given to the more crucial components and lower weightage to less important components.

The nature of fiber reinforced composite is such that the selection of processes, materials and properties are closely interrelated, and the choice of any one may define the other two. Though this may restrict the designer's choice, it is beneficial in the sense that it tends to reduce the potentially very large number of possible combinations to a manageable number of options.

The design loop is always iterated a number of times, at least once for each of the major stages of conceptual and detailed design, and in successively greater depth in order to ensure that there is no fundamental problem which would prevent the design being realized.

The embodiment design should be validated by model, mock-up or prototype, depending on which aspects of the design need to be checked. This will ensure that the requirements of the design brief can be met. In the next step of the detailed design process, the embodiment design process is repeated in more depth based on the results of the prototype validation.

5.3 Materials Design

The unique aspect of fiber reinforced composite is the availability of a number of material options to achieve the design objectives. Broadly, the options fall in two domains — material selection and material properties.

5.3.1 *Materials selection*

Materials selections can come from three different groups:

(1) Finished products: Finished products are in the form of sub-components. It can be in bars, rod, panels, tubes, fixtures and fittings, etc. A design can also be derived by assembling suitable combinations of items. Finished products have benefits such as consistent property value, better mechanical properties and low maintenance. The limitations are the availability of limited specific shapes, sizes and materials.
(2) Compounds and intermediate materials: The combination of reinforcement, resin and additives is preselected by the supplier or compounder and production process route is prescribed to produce the optimum properties. These can be in the form of injection molding compounds, dough molding compounds, sheet molding compounds, prepregs and thermoplastic sheet compounds.
(3) Raw materials: Raw materials provide the largest range of options and materials can be tailored to a wide range of properties and fabrication processes. The major disadvantage is that property levels must be checked to insure that the reinforcement, resin and additive chosen are all compatible and that the correct process parameters are used.

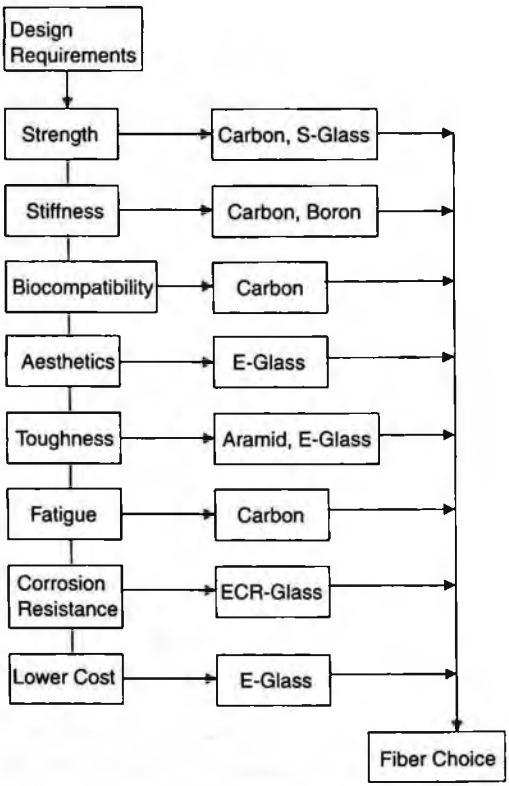


Fig. 5.4 Fiber selection process.

5.3.2 *Material properties*

Critical design requirement guide the type of resin and fiber selection that will give the highest property level. The level for each of the other properties then needs to be checked to insure that they are satisfactory.

(1) Fiber-controlled properties: Fiber-controlled properties depend upon both the amount of fiber and its orientation. The degree of anisotropy of the fibers depend upon the fiber length. For short fibers the properties are essentially isotropic, while for longer fibers the properties become increasingly anisotropic. Figure 5.4 briefly describes the fiber selection process.

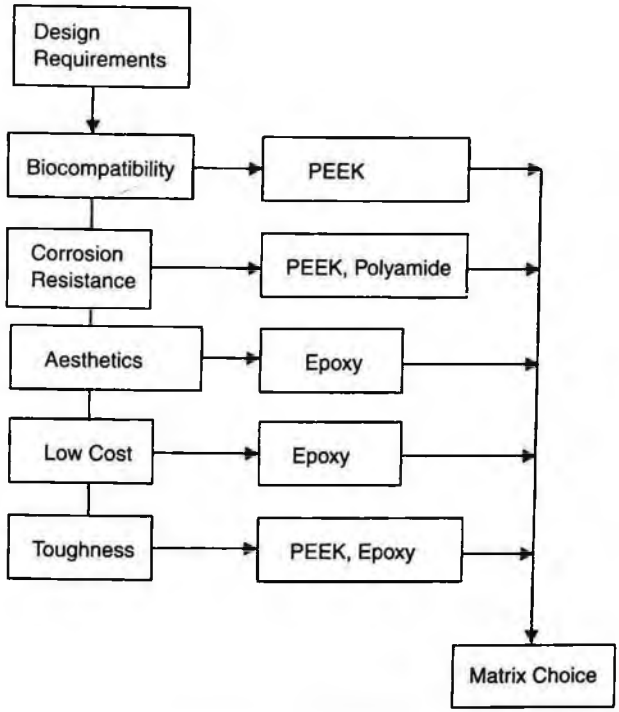


Fig. 5.5 Matrix selection process.

(2) Resin-controlled properties: Surface finish and corrosion resistance characteristics of the product are mostly controlled by the resin properties. Unlike, fiber, many functional composite properties largely depends on the amount of resin. The resin properties also decide the processing method. A brief resin selection method is presented in Fig. 5.5.

5.4 Component Design

At the materials design level, the properties of the two basic constituents i.e. reinforcement and matrix properties are exploited to provide additional qualities, which they are unable to provide individually. During the component design stage, other requirement such as functionality, environmental considerations and manufacturability are considered.

5.4.1 *Design of laminate*

Many structural materials generally have isotropic properties and they are homogeneous having uniform properties in all directions. A composite materials can take a number of different forms. They can be orthotropic, where the stiffness and strength in the fiber direction is considerably larger than that to transverse fiber direction. It can be planar-isotropic as in the case of textile-reinforced composites and approach isotropy as in the case of randomly placed chopped fiber reinforced composites. The extent of anisotropy is primarily controlled by the type and form of reinforcement.

The anisotropic properties of composite materials are the key to developing highly efficient components. Reinforcement fibers can be strategically placed so that they locally provide the required strength and stiffness. Furthermore, by combining different types of fibers, the particular property of the fiber can be exploited.

5.4.2 *Design for environment*

The environmental factors of temperature, aggressive fluids and sterilization environment can have an adverse effect on the performance of the composite materials over time. The extent of the effect is a function of the fiber and resin system used and the severity of environmental condition. In comparison with other engineering materials, composite materials perform well in many arduous environmental conditions. The acceptance of this material in marine and chemical industry is a pointer towards this. Care is needed particularly in hot and wet conditions during which the performance can fall by as much as half from those at normal ambient conditions. Adequate testing of materials in the expected environment is the best way to ensure a satisfactory integrity.

5.4.3 *Design for joints*

Part integration during manufacture is a prime advantage of composite materials. Consequently, a carefully manufactured component need not have any joint. However, it is not always practical to make single unit and more often than not a complete assembly is obtained from a number of components. The designer needs to have a good understanding of

the performance of mechanical joints, which are often governed by the strength of the matrix as opposed to the fastener. Consequently, long-term loads are affected by creep properties of the matrix. Bonded joints show better performance compared to fastened joints. Here care needs to be taken to avoid stress concentrations caused by abrupt changes in section.

5.4.4 *Design for robustness*

Robustness is a difficult parameter to quantify, as it is a measure of the ability of component to survive shocks and rough handling. Composite materials have good resilience and toughness, particularly aramid and glass reinforced polymers. The ability of a component to withstand through-life impacts can be enhanced by good manufacturing quality, eliminating production flaws such as voids, excessive air inclusions and shrinkage cracks. In addition, a good design plays an equally important role. The correct selection of materials, the provision of adequate load paths and the avoidance of stress concentrations are the key aspects to be considered.

5.4.5 *Design for manufacture*

An important aspect of the composite design is the consideration of the manufacturing process during the design stage. This is one element, which significantly improves performance and at the same time reduces the component cost. The designer needs to have a good understanding of how the many processes work and how the design will be affected by the process. Different processes have varied level of achievable quality control and hence this aspect needs consideration during the design. The factors of safety used in design needs to reflect these differences and the selection of design stress needs to reflect the variation in properties that is expected to occur.

5.4.6 *Design for cost*

Expensive materials do not necessarily lead to expensive components. Reduced handling costs reduce product cost, increased environmental resistance, prolonged life, etc. Composite materials have many unique characteristics which the designer can exploit to advantage. For any dynamic

components, cost must include a through-life cost assessment. In the case of biocomposites, additional benefits accrued by the reduction in hospitalization cost and clinical procedure cost needs to be accounted for.

References

Proceedings of International Conference on *Designing Cost-Effective Composites*. Published by Professional Engineering Publishing for the Institutional Mechanical Engineers, 1998.

R. M. Mayer, Design with reinforced plastics, The Design Council, London, UK, 1993.

G. Eckold, *Design & Manufacture of Composite Structures*, Woodhead Publishing Ltd. Cambridge, England, 1994.

Handbook of Polymer Composites for Engineers, ed. Leonard Hollaway, Woodhead Publishing Ltd., Cambridge, England, 1994.

Chapter 6

BIOMEDICAL APPLICATIONS OF POLYMER COMPOSITES

Over the years, polymer composites have been proposed for various aspects of biomedical applications. These applications have been reported in different sources of the literature from time to time. There have however been very few attempts to combine the wide variety of literature sources into a single article of book chapter, which would allow the reader to fully appreciate the development of biocomposites so far. One exception is a recent survey by Ramakrishna *et al.* [2001], which forms the basis of this chapter.

6.1 Hard Tissue Applications

6.1.1 *Bone fracture repair*

Bones constitute the skeletal system and provide the supporting structure for the body. Bone can remodel and adapt itself to the external environment, with its density influenced by the stress condition applied on it. This is referred to as Wolff's law [Hayes and Snyder, 1981]. Higher applied stress leads to denser bone, whereas lower than the normal physiological load stress can decrease the bone mass and result in bone weakening.

There are many types of bone fractures depending on the crack size, orientation, morphology, and location. Bone fractures are treated in different ways, which can 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 does. 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 [Black, 1988]. The casting material essentially is a composite made of woven cotton fabrics (woven gauze) and Plaster of Paris matrix (calcium sulfate). Other reinforcements include fabrics of glass and polyester fibers. Although the plaster bandages have many advantages, they also show disadvantages such as untidiness during preparation, high density and bulkiness, 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. Moreover, it should be permeable to ventilation without which the patient's skin may be scorched or weakened.

In the internal fixation approach the bone fragments are held together by different ways using implants such as wires, pins, screws, plates, and intramedullary nails. All these implants are temporarily placed inside the body.

6.1.1.1 *Bone plates*

Plate and screw fixation as shown in Fig. 1.1 is the most popular method for rigid internal fixation of the fractured bone. The bone plates are also called osteosynthesis plates. The rigid fixation is designed to provide high axial pressures (also known as dynamic compression) in the fragments of the bone, which facilitates 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 complete bone healing has been obtained by the plate fixation which normally takes from half a year to two years since 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 re-fracture of bone after the removal of the plates due to the bone atrophy. It may be noted that the modulus of a metal (210–230 GPa for stainless steel, and 110 GPa for Ti alloy) plate is much higher than 10–18 GPa modulus of the bone. In the load sharing between the bone and the implant, the amount of stress carried by each of them is directly related to their stiffness. Thus, bone is

insufficiently loaded compared to the metal implant, and this phenomenon is called 'stress-shielding' or stress protection. Many investigators [Moyen *et al.*, 1978; Uhthoff and Finnegan, 1983] have shown that the degree of stress protection is proportional to the degree of stiffness mismatch. The stress protection affects the bone remodeling and healing process, leading to increased bone porosity (also known as bone atrophy). The bone underneath the plates adapts to the low stress and becomes less dense and weak. The stress shielding effect is much more pronounced with the stainless steel plates than with the Ti alloy plates. 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. As polymer composite materials can offer desired high strength and bone like elastic modulus, many investigators proposed a variety of polymer composite bone plates (Fig. 6.1), which can be grouped into nonresorbable, partially resorbable, and fully resorbable. 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 external callus bridging the fracture. Thereby, the callus increases the cross section of the newly formed bone and, thus, prevents re-fracture.

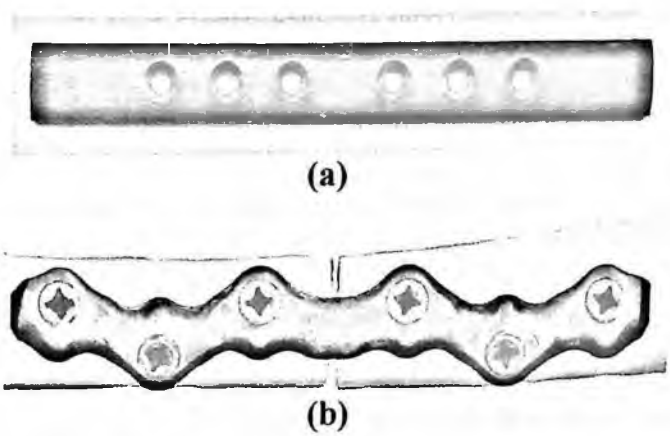


Fig. 6.1 (a) A typical composite bone plate made using braided carbon fiber fabric and epoxy matrix. (b) A composite bone plate made of Carbon/PEEK material system.

Earlier researches on the nonresorbable composite plates used CF/epoxy and GF/epoxy material systems [Bradley *et al.*, 1980]. However, concerns over the toxic effects of monomers in partially cured epoxy composite materials were later raised [Peluso *et al.*, 1991] and research activity on these materials gradually decreased. As the technology for making good quality thermoplastic composites became available, researchers developed CF/PMMA [Woo *et al.*, 1974], CF/PP, CF/PS [Hunt, 1987], CF/PE [Rushton and Rae, 1984], CF/nylon, CF/PBT [Gillett *et al.*, 1985], and CF/PEEK [Williams *et al.*, 1987; Wenz *et al.*, 1990; Fujihara *et al.*, 2001] non-resorbable thermoplastic composite bone plates. Unlike the thermoset composites, the thermoplastic composites are considered free from the complications associated with unused monomers. Moreover, similar to metal alloy plates, thermoplastic composite plates could potentially be bent or contoured with heating 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 [Wenz *et al.*, 1990] and has good resistance to hydrolysis and radiation (a sterilization method) degradation. The other promising properties include high strength, fatigue resistance, and biological inertness with no mutagenicity or carcinogenicity. The tissue response to carbon fibers and composite debris has been described as minimal. Researchers also developed CF/carbon [Blazewicz *et al.*, 1997] and CF/PEEK composite screws (Fig. 1.3) for osteosynthesis. 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.

As the bone healing progresses, it is desirable that the bone is subjected to gradual increase of stress, thus reducing the stress protection effect. In other words, the stress on the plate should decrease while stress on the bone should increase with time. This is possible only if the plate looses rigidity in *in vivo* environment. The non-resorbable polymer composites do not display this characteristic. To meet this need, researchers introduced resorbable polymers for bone plate applications. The polymers such as poly(lactic acid) (PLA) and poly(glycolic acid) (PGA) resorb or degrade upon implantation into the body. Many bioresorbable polymers are either brittle or too weak

and flexible, and loose most of their mechanical properties in few weeks. Choueka *et al.* [1995] proposed fully resorbable composites by reinforcing resorbable matrices with resorbable fibers (poly(L-lactic acid) (PLLA) fibers and calcium phosphate based glass fibers). One major advantage for resorbable composite prostheses is that they need not be removed with a second operation, as is required with metallic or non-resorbable composite implants. The low strength and modulus of resorbable materials remains a limitation and hence they are only used for applications where the loads are moderate.

In order to improve mechanical properties, the resorbable polymers are reinforced with variety of non-resorbable materials including carbon fibers [Zimmerman *et al.*, 1987] and polyamide fibers. The resulting bone plates are called partially resorbable composite plates. According to Zimmerman *et al.* [1987], CF/PLA composites possessed superior mechanical properties before the implantation. However, they lose strength and stiffness too rapidly in *in vivo* environment 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. 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 that stopped almost any research in this direction.

6.1.1.2 *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. 1.1). Up to date, the most successful applications of fully bio-resorbable implants are in the forms of pins, rods, and screws [Elst *et al.*, 2000]. Most implants are manufactured from hydroxy fatty acids such as PGA, PLA, and copolymers. The self-reinforcing (SR) technique in which oriented filaments are used as a scaffold for a matrix with the same chemical structure produces strong implants. Tormala *et al.* [1988] introduced the sintering SR technique, by which implants can be built from strong fibrous polymer units. The more advanced SR technique by partial fibrillation by orientational drawing has further strengthened the self-reinforced implants [Tormala *et al.*, 1990].

In view of the modest mechanical properties of the first generation implants, their use was restricted to low-stress applications in cancellous bone, mainly in the small fracture regions of Ankle and Foot [Rokkanen *et al.*, 1985], Knee [Friden and Rydholm, 1992], Elbow [Hirvensalo *et al.*, 1990], and Wrist and Hand [Kumta *et al.*, 1992]. However, with the introduction of the bio-resorbable fracture fixation devices, a new type of medical complication, the sterile sinus formation, has been observed. Santavirta *et al.* [1990] reported an incidence of this aseptic inflammatory response that varies from 3% in Chevron osteotomies to 22% in distal radius fractures. In another survey [Bostman *et al.*, 1992], out of 216 patients with malleolar fractures, 24 developed a transient local inflammatory reaction with a painful erythematous fluctuant swelling. Average postoperative period was three months. Santavirta *et al.* [1990] designed an immunological survey to elicit the immune response on the alpha polyesters. PGA seemed to be an inert material but it does induce inflammatory mononuclear cell migration and adhesion, leading to slightly nonspecific lymphocyte activation.

Recently, Lin *et al.* [1997] proposed short GF/PEEK composite material for intramedullary application. The rationale behind this proposal is in the claimed biocompatibility of the composite material and its matching mechanical properties compared to the cortical bone. Kettunen *et al.* [1999] 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.

6.1.2 *Spine instrumentation*

Spine is a linked structure consisting of 33 vertebrae superimposed on one another. A number of spine related disorders have been 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.

In the case the defect is limited to few, vertebrae alternative approaches, such as (a) spinal fusion and (b) disc replacement, are used. In a 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. Synthetic bone graft material must have adequate strength and stiffness, also capable of bonding to the residual vertebrae. Ignatius *et al.* [1997] developed Bioglass/PU composite material for vertebral body replacement, whereas Marcolongo *et al.* [1998] 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 [Valdevit *et al.*, 1996]. 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 post-operative assessment with X-rays, computed tomography (CT), and magnetic resonance imaging (MRI) through reflection and artifacts (see Chapter 2). Inadequate biomechanical capabilities of bioceramic prostheses may lead to the collapse of instrumented spine and injury of neurological structures and blood vessels. To overcome disadvantages of the conventional materials, Brantigan *et al.* [1991] and Ciappetta *et al.* [1997] 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 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 with 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 risen to intervertebral disc prostheses.

Problems related to intervertebral disc are treated by replacing affected nucleus with a substitute material or by replacement of total disc (nucleus and annulus) using an artificial disc [Bao *et al.*, 1996]. Both methods require duplication of the natural structure, significant durability to last longer than forty 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 [Bao *et al.*, 1996]. A variety of materials such as stainless steel, Co–Cr alloy, PE, SR, PU, PET/SR, and PET/ hydrogel composites are proposed for disc prostheses either alone or in combinations [Urbaniak *et al.*, 1973; Ambrosio *et al.*, 1996]. However, their performance is not yet been accepted for long-term applications. To date, there has been no artificial disc that is able to reproduce the unique mechanical and transport behavior of natural disc satisfactorily. This may be a result from 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 [Bridwell *et al.*, 1991].

Schmitt-Thomas *et al.* [1997] 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 overcome the problems of metal alloys such as corrosion and interference with the diagnostic techniques.

6.1.3 *Joint replacements*

Joints enable the movement of body and its parts. Many joints in the body are of synovial types, which permit free movement. Hip, knee, shoulder, and elbow are few common examples of synovial joints. They all possess two opposing articular surfaces, which are protected by a thin layer of articular cartilage and lubricated by elastic-viscous synovial fluid. 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 are also damaged upon exposure to severe mechanical or metabolic injury.

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. 1.1).

6.1.3.1 *Total hip replacement (THR)*

THR is the most common artificial joint in human beings. For example, over 150,000 total hip replacements are performed every year in USA alone. A typical THR consists of a cup type acetabular component, and a femoral component whose head is designed to fit into the acetabular cup and thus to enable 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. 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 materials like Co–Cr alloy and PTFE polymer. Subsequently acetabular cups made of UHMWPE were developed and found to be successful. The UHMWPE cups are usually supported with metal backing. Some reported data suggests 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 [Sclippa and Piekarski, 1973] or UHMWPE fibers [Deng and Shalaby, 1997]. Deng and Shalaby [1997] 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.

Although THRs are used widely, one of the major unsolved problems 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, which may lead to bone resorption and eventual aseptic loosening of the prosthesis (note that the aseptic loosening is also linked to wear particles/debris) [Whiteside, 1989]. This is particularly a problem with young and more active patients, and may cause severe pain and clinical failure necessitating repeat surgery. It has been demonstrated that Ti alloy stems result in a 50% reduction in the femur peak stress compared to the Co–Cr alloy stems. Thus, 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. The advanced polymer composites can offer strength comparable to metals, but more flexibility than metals. They also offer the potential to tailor implant properties by selecting material ingredients and spatially controlling ingredient composition and configuration, which is useful in reducing the development of high stress regions. 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. Christel and co-workers [Christel *et al.*, 1987] introduced CF/PS and CF/C composite stems. They reported faster bone bonding in the case of composite implants compared to the high stiffness conventional implants. The composite stems were found to be stable with no release of soluble compounds, and high static and fatigue strength. Chang *et al.* [1990] made CF/epoxy stems by laminating 120 layers of unidirectional plies in a pre-determined orientation and stacking sequence. Simoes *et al.* [1999] made composite stems using braided hybrid carbon-glass fiber performs and epoxy resin. Some researchers [Akay and Aslan, 1996] designed an injection molded CF/PEEK composite stems (Fig. 1.4), which posses 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 [Akay and Aslan, 1996] 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 will lead to increased implant deformation and relative movement (also called micromotion) between implant and bone tissue during loading. The micromotion also influences bone remodeling and often leads to residual pain. The stress shielding and micromotion are conflicting phenomena [Huiskes *et al.*, 1992]. In other words, for appropriate structural compatibility, the implant design should reduce stress shielding and micromotion simultaneously.

6.1.3.2 *Total knee replacement (TKR)*

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. A typical TKR mainly consists of femoral and tibial components (Fig. 1.1). The femoral component articulates on the tibial component. The materials used for femoral component are predominantly Co–Cr and Ti alloys [Walker and Blunn, 1997]. 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 the sinking of prosthesis. Inoue *et al.* [1994] 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 latter material combination resulted in a lower stress concentration than the former in the vicinity of tibial stem. This probably explain the reason for the sinking of knee prostheses. Carbon fibers were used to reinforce UHMWPE to reduce its cold flow (creep) deformation [Silverton *et al.*, 1996]. The reinforcement enhances the stiffness, tensile yield strength, creep resistance, and fatigue strength of UHMWPE [Wright *et al.*, 1981]. 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 to 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 a low friction bearing but a combination of a UHMWPE surface and a composite substrate appears to offer some advantages. Recently, Deng and Shalaby [1997] 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.

6.1.3.3 *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 to 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 Park [1984], with Co–Cr and Ti alloys, HDPE, UHMWPE, etc as candidate materials. 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 is considered. Tearing of the silicone rubber at the junction of prosthesis and roughened arthritic bone is a major concern. In order to improve its tear strength and flexural properties, PET fabrics were used as reinforcement. Goldner and Urbaniak [1973] reported that the composite prosthesis is also successful in decreasing pain, improving stability, increasing hand function, and providing an adequate range of motion.

6.1.3.4 *Bone cement*

Proper fixation to the bones is as important as the design of joint replacement itself. One of the earliest methods is to press-fit the joint prosthesis into the bone using grouting material called 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. The most widely used bone cement is based on PMMA, also called acrylic bone cement [Saha and Pal, 1984]. It is self-polymerizing and contains solid PMMA powder and liquid MMA monomer. 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. The polymerization process is also associated with undesirable shrinkage of acrylic polymer. Another issue is the deterioration of cement/implant or cement/bone interface with time, leading to problems of mechanical failure and instability. Fatigue failure has been found to be a predominant *in vivo* failure mode of bone cement [Krause and Mathis, 1988]. Researchers have tried to improve bone cement mechanical

properties by reinforcing with stainless steel and Ti alloy wires, and polymer fibers such as UHMWPE [Wagner and Cohn, 1989], Kevlar, carbon [Pilliar *et al.*, 1976], and PMMA [Gilbert *et al.*, 1995]. Use of such fiber reinforcement also reduces the peak temperature during polymerization of the cement, and thus reducing the tissue necrosis [Topoleski *et al.*, 1992]. 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 [Park and Lakes, 1992] or surface-active glasses (Bioglass) with the bone [Tamura *et al.*, 1995]. 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.

6.1.4 *Bone replacement (synthetic bone graft) materials*

Synthetic bone grafts are necessary to fill bone defects or to replace fractured bones. 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 reinforced PE using HA particles [Bonfield *et al.*, 1981] and Bioglass [Hench and Ethridge, 1982], which are bioactive. It was reported that for HA particulate volume fractions above 40%, the composite is brittle. Moreover, 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. 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 [Liu *et al.*, 1998], PLLA [Higashi *et al.*, 1986], PHB [Knowles and Hastings, 1993], alginate and gelatin [Klein *et al.*, 1987] with bioactive particles. Upon implantation, as the matrix polymer resorbs, more and more bioactive particles come into contact with the growing tissues, thus achieving good integration of the biomaterial into the bone.

6.1.5 *Dental applications*

6.1.5.1 *Dental composite resin*

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 biomaterials 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. The choice of material is dependent on its ability to resemble the physical, mechanical and aesthetic 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 sometimes to mask discoloration (veneering) or to correct contour and alignment deficiencies. Dental composite resins, which are translucent with refractive index matching that of the enamel, have become the predominant such materials and are very commonly used to restore posterior teeth as well as anterior teeth. The dental composite resin comprises BIS–GMA as the matrix polymer and quartz, barium glass, and colloidal silica as fillers. 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 UV (ultraviolet) light from a lamp used by the dentist. In other types of composites a urethane dimethacryate rather than the BIS–GMA resin is used. The filler particle concentration varies from 33% to 78% by weight and the size varies from 0.05 μm 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 [Kennedy *et al.*, 1998]. Strong bonding between the fillers and resin is achieved using silane-coupling agents [Krause *et al.*, 1989].

6.1.5.2 *Dental post*

In cases when a 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 [Lovdahl and Nicholls, 1977], on which the core and crown are built (Fig. 6.2). 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

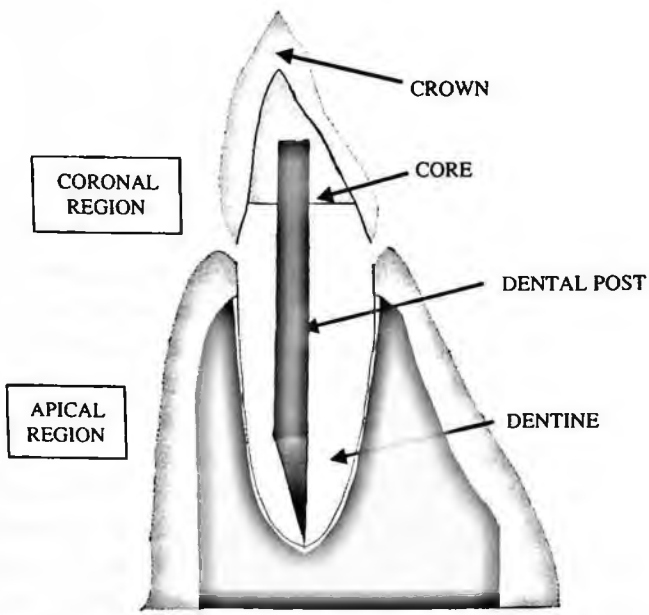


Fig. 6.2 A dental post restored tooth.

bone. 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 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 to 4%) and failure of restorations. Newer posts made of zirconia, short glass fiber reinforced polyester, and unidirectional carbon fiber reinforced epoxy composite posts [Issidor *et al.*, 1996] are introduced. These new posts are adequately rigid, resistant to corrosion and fatigue. 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.* [1992] indicated that 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. These findings contradict the belief that the conventional posts strengthen the tooth by evenly distributing the external forces acting on the tooth. Thus, 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 overcome the root fractures due to stress concentration. In other words, it is desirable to have a post with varying stiffness. In the next chapter, we will describe such a post, which has higher stiffness in the coronal region and lower one at the apical end.

6.1.5.3 *Dental implant and bridge*

In extreme case, the damaged or condemned tooth is extracted and replaced with a dental implant. Dental implants are artificial tooth roots that permanently replace missing teeth, and they are alternative to bridges or false teeth. The dental implants may be designed to enter the jawbone or to fit on to the bone surface. They are made of wide range of materials such as metals (Co–Cr–Mo alloys, Ti alloys, stainless steel, platinum, silver), ceramics (zirconia, alumina, glass, and carbon), polymers (UHMWPE, PMMA, PTFE, PS, and PET), and composites (SiC/carbon and CF/carbon) [Adams *et al.*, 1978; Louis and Dabadie, 1990]. 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, the 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 extreme cases, 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 [Bjork *et al.*, 1986], KF/PMMA [Henderson Jr. *et al.*, 1987], UHMWPE/PMMA [Davy *et al.*, 1992] and GF/PMMA [Miettinen and Vallittu, 1997] composite bridges and dentures [Freilich *et al.*, 2000].

6.1.5.4 *Bracket and archwire*

Orthodontic arch wires 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 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 the wire. It is important for the bracket to have high strength and stiffness to prevent distortion during tooth movement. This technique is also used to splint the traumatized teeth. Traditionally, arch wires are made of stainless steel and Ni–Ti alloys. Jancar *et al.* [1993] and Imai *et al.* [1999] proposed GF/PC, GF/Nylon, GF/PP, and GF/PMMA composite materials for arch wire applications. The advantages of using composite arch wires stated include aesthetics, easy forming in the clinic, and the possibility of varying stiffness without changing component dimensions [Zufall *et al.*, 1998].

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. 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). Their 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. Chopped glass fibers have been used as reinforcement to fabricate composite brackets. These devices, however, have found limited success in clinical treatment due to their inferior mechanical properties compared with metal brackets. Further development needs to use continuous fiber preforms as reinforcement.

6.2 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 provide mechanical support or serve as a scaffold for tissue growth.

6.2.1 *Bulk space fillers*

Bulk space fillers are used to restore cosmetic defects, atrophy, or hypoplasty to an aesthetically satisfactory condition. They are mostly used in the head and neck. The materials used in these applications include silicon rubber, PE, and PTFE. 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, have sufficient strength, and be very smooth. Polymers such as silicon rubber and PTFE [Park, 1984] 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 [1993] reported that composites comprising PET or PTFE fabrics and PU are more suitable for this purpose, as they could reduce the cartilage degeneration following the meniscectomy. At the same time, Pongor *et al.* [1992] 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.

6.2.2 *Encapsulants and carriers*

6.2.2.1 *Ureter prosthesis*

Ureter prostheses made of PVC, PE, nylon, PTFE, and silicon rubber were used without much long-term success. The lack of success was caused by the difficulty in 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 has proven to be difficult to overcome. Polyester fiber reinforced glycol methacrylate gel prostheses with a fabric backing was reported with some success [Kocvara *et al.*, 1967; Hench and Ethridge, 1982]. The fabric backing facilitated easy attachment of 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.

6.2.2.2 *Catheters*

Catheters (tubes) are widely used to access remote regions of the human body to administer fluids (e.g. nutrients, isotonic saline, glucose, medications, blood products) as well as to obtain data (e.g. artery pressure, gases, collecting blood samples for analysis). PU and silicon rubber are widely used materials for making catheters because of their flexibility and ease of fabrication into variety of sizes and lengths in order to accommodate the wide range of vessels to be cannulated. Silicon rubber is reinforced with silica particles to improve its tear strength and to decrease wettability. Andreopoulos *et al.* [1998] reported that with increasing silica particles

fraction 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 inhibition of infection. An ideal vascular catheter also must be flexible enough to allow bending and stretching of the veins as well as convenience of movement for the patient. Catheters that are initially supple may become brittle over time, resulting in vascular wall damage.

6.2.3 *Functional load-carrying and supporting implants*

6.2.3.1 *Tendons and ligaments*

Artificial tendons and ligaments are the best examples of load-bearing soft tissue implants. Tendon is a strong fibrous band of tissue that extends from a muscle to periosteum of the bone. Ligament is a connective tissue band that links bones in the vicinity of every synovial joint. Tendons and ligaments 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. Tendons have little regenerative capacity and require very long time to regenerate fully.

The use of biomaterials in tendon/ligament repair is one of the most demanding applications of prostheses in soft tissues. Biomaterials are used in a number of ways in tendon healing. They may be used to replace the tendon, to hold a damaged tendon in proper alignment, or to form a new 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 [Forster *et al.*, 1980; Mendes *et al.*, 1986; King *et al.*, 1996]. Permanent fixation of the implant is 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 in anchorage to the bone, and abrasion and wear of prostheses, which deteriorates strength in the long term and leads 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 [Seedham, 1993]. To reduce particle migration and improve handling properties, prostheses are coated with polymers such as silicon rubber, poly(2-hydroxyethyl methacrylate) (PHEMA), and PLA. Pradas and Calleja [1991] reported that by combining a 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.* [1995] developed ligament prosthesis by reinforcing a hydrogel matrix (PHEMA) with helically wound rigid PET fibers, and demonstrated that both the 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.

6.2.3.2 *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 [Ward and Minns, 1989]. The fabrics or meshes facilitate tissue ingrowth thus providing stability to the prosthesis. Recently, Werkmeister *et al.* [1998] developed PET fabrics coated with collagen and PU materials suitable for repairing hernia and abdominal wall (the 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 silicon rubber), 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 such as PMMA, PTFE, PE, and silicon rubber, 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 and glass ceramics). Migration of prostheses is the main problem reported and it is essential to apply a suitable surgical method. Researchers are also developing PE/PU flexible composite materials as tympanic membrane replacements [Teoh *et al.*, 1999]. Tympanic membrane transmits sound vibrations to the inner ear through three auditory ossicles.

6.3 Other Biomedical Applications

6.3.1 *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. The lightweight, corrosion resistance, fatigue resistance, aesthetics, and ease of fabrication of polymer composite materials made them ideal choice for modern limbs systems [Robin, 1981]. 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 [Coombes *et al.*, 1996]. The typical artificial leg system consists of three parts namely socket, shaft, and foot (Fig. 6.3). 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. More detailed descriptions on polymer composites used for socket fabrication will be given in the next chapter.

6.3.2 *Medical instrumentation*

High technology machines such as CT and MRI scanners are gaining wider usage for medical diagnostic purposes. These machines have larger bodies

Fig. 6.3 A transtibial prosthesis consisting of a socket, a shank, and a foot [from Huang and Ramakrishna, 1999].

fitted with moving tables for the patients. The moving table needs to be 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. These materials are also used in making surgical clamps, head rests frames, X-ray film cassettes and CT scan couches.

References

D. Adams, D. F. Williams, and J. Hill, Carbon fiber-reinforced carbon as a potential implant material, *Journal of Biomedical Materials Research*, 1978, **12**, 35–42.

M. Akay and N. Aslan, Numerical and experimental stress analysis of a polymeric composite hip joint prostheses, *Journal of Biomedical Materials Research*, 1996, **31**, 167–182.

L. Ambrosio, P. A. Netti, S. Iannace, S. J. Huang, and L. Nicolais, Composite hydrogels for intervertebral disc prostheses, *Journal of Materials Science: Materials in Medicine*, 1996, **7**, 251–254.

A. G. Andreopoulos, M. Evangelatou, and P. A. Tarantili, Properties of maxillofacial silicone elastomers reinforced with silica powder, *Journal of Biomaterials Applications*, 1998, **13**, 66–73.

Q. B. Bao, G. M. McCullen, P. A. Higham, J. H. Dumbleton, and H. A. Yuan, The artificial disc: Theory, design and materials, *Biomaterials*, 1996, **17**, 1157–1167.

N. Bjork, K. Ekstrand, and I. E. Ruyter, Implant-fixed, dental bridges from carbon/graphite fiber reinforced poly(methyl methacrylate), *Biomaterials*, 1986, **7**, 73–75.

J. Black, *Orthopaedic Biomaterials in Research and Practice*, Churchill Livingstone, New York, 1988, pp. 197–210.

S. Blazewicz, J. Chlopek, A. Litak, C. Wajler, and E. Staszkow, Experimental study of mechanical properties of composite carbon screws, *Biomaterials*, 1997, **18**, 437–439.

W. Bonfield, M. D. Grynpas, A. E. Tully, J. Bowman, and J. Abram, Hydroxyapatite reinforced polyethylene — A mechanically compatible implant material for bone replacement, *Biomaterials*, 1981, **2**, 185–186.

O. Bostman, E. K. Partio, E. Hirvensalo, *et al.*, Foreign-body reactions to polyglycolide screws, *Acta Orthopaedics Scandinavica*, 1992, **63**, 173–176.

J. S. Bradley, G. W. Hastings, and C. Johnson-Nurse, Carbon fiber reinforced epoxy as a high strength, low modulus material for internal fixation plates, *Biomaterials*, 1980, **1**, 38–40.

J. W. Brantigan, A. D. Steffee, and J. M. Geiger, A carbon fiber implant to aid interbody lumbar fusion Mechanical testing, *Spine*, 1991, **16**(6S), S277–S282.

K. H. Bridwell, R. L. DeWald, J. P. Lubicky, D. L. Spencer, K. W. Hammerberg, D. R. Benson, and M. G. Neuwirth, *The Textbook of Spinal Surgery*, J.B. Lippincott Company, Philadelphia, USA, 1991.

J. G. Cailleteau, M. R. Rieger, and J. Ed Akin, A comparison of intracanal stresses in a post-restored tooth utilizing the finite element method, *Journal of Endodontics*, 1992, **18**(11), 540–544.

F. K. Chang, J. L. Perez, and J. A. Davidson, Stiffness and strength tailoring of a hip prostheses made of advanced composite materials, *Journal of Biomedical Materials Research*, 1990, **24**, 873–899.

J. Choueka, J. L. Charvet, H. Alexander, Y. H. Oh, G. Joseph, N. C. Blumenthal, and W. C. LaCourse, Effect of annealing temperature on the degradation of reinforcing fibers for absorbable implants, *Journal of Biomedical Materials Research*, 1995, **29**, 1309–1315.

P. Christel, A. Meunier, and S. Leclercq, Development of a carbon-carbon hip prosthesis, *Journal of Biomedical Materials Research*, 1987, **21**, 191–218.

P. Ciappetta, S. Boriani, and G. P. Fava, A carbon fiber reinforced polymer cage for vertebral body replacement: A technical note, *Neurosurgery*, 1997, **41**(5), 1203–1206.

A. G. A. Coombes, C. D. Greenwood, and J. J. Shorter, Plastic materials for external prostheses and orthoses, *Human Biomaterials Applications*, eds. D. L. Wise, D. J. Trantolo, D. E. Altobelli, M. J. Yaszemski, and J. D. Gresser, Humana Press Inc., Totowa, New Jersey, USA, 1996, pp. 215–255.

K. W. M. Davy, S. Parker, M. Braden, I. M. Ward, and H. Ladizesky, Reinforcement of polymers of 2,2 bis-4(2-hydroxy-3-methacryloyloxy propoxy) phenyl propane by ultra-high modulus polyethylene fibers, *Biomaterials*, 1992, **13**(1), 17–19.

M. Deng and S. W. Shalaby, Properties of self-reinforced ultra-high-molecular weight polyethylene composites, *Biomaterials*, 1997, **18**, 645–655.

M. van der Elst, C. P. A. T. Klein, P. Patka, and H. J. T. M. Haarman, Biodegradable Fracture Fixation Devices, *Biomaterials and Bioengineering Handbook*, ed. D. L. Wise, Marcel Dekker, Inc., New York, 2000, 509–524.

I. W. Forster, Z. A. Ralis, and D. H. Jenkins, Filamentous carbon fiber induction of new tendon: Tissue reactions and environmental conditions, *Evaluation of Biomaterials*, eds. G. D. Winter, J. L. Leray, and K. de Groot, John Wiley & Sons Ltd, New York, 1980, pp. 367–371.

M. A. Freilich, J. C. Meiers, J. P. Duncan, and A. Jon Goldberg, *Fiber-Reinforced Composites in Clinical Dentistry*, Qintessence Publishing Co, Inc., Illinois, USA, 2000.

T. Friden and U. Rydholm, Severe aseptic synovitis of the knee after biodegradable internal fixation, *Acta Orthopaedics Scandinavica*, 1992, **63–1**, 94–97.

K. Fujihara, Z. M. Huang, S. Ramakrishna, K. Satkunanantham, and H. Hamada, Development of Braided Carbon/PEEK Composite Bone Plates, *Advanced Composites Letter*, 2001, **10**, 449–456.

J. L. Gilbert, D. S. Ney, and E. P. Lautenschlager, Self-reinforced composite poly(methyl methacrylate): Static and fatigue properties, *Biomaterials*, 1995, **16**(14), 1043–1055.

N. Gillett, S. A. Brown, J. H. Dumbleton, and R. P. Pool, The use of short carbon fiber reinforced thermoplastic plates for fracture fixation, *Biomaterials*, 1985, **6**, 113–121.

J. L. Goldner and J. R. Urbaniak, The clinical experience with silicone-Dacron metacarpophalangeal and interphalangeal joint prostheses, *Journal of Biomedical Materials Research Symposium*, 1973, **4**, 137–163.

W. C. Hayes and B. Snyder, Towards a Quantitative Formulation of Wolff's Law in Trabecular Bone, *Mechanical Properties of Bone*, ed. S. C. Cowin, *The Joint ASME-ASCE Applied Mechanics, Fluids Engineering and Bioengineering Conference, AMD-Vol. 45*, Boulder, Colorado, 1981.

L. L. Hench and E. C. Ethridge, *Biomaterials: An Interfacial Approach*, Academic Press, New York, USA, 1982.

J. D. Henderson Jr., R. H. Mullarky, and D. E. Ryan, Tissue biocompatibility of kevlar aramid fibers and polymethylmethacrylate, composites in rabbits, *Journal of Biomedical Materials Research*, 1987, **21**, 59–64.

S. Higashi, T. Yamamuro, T. Nakamura, Y. Ikada, S. H. Hyon, and K. Jamshidi, Polymer-hydroxyapatite composites for biodegradable bone fillers, *Biomaterials*, 1986, **7**, pp. 183–187.

E. Hirvensalo, O. Bostman, E. Partio, *et al.*, Absorbable polyglycolide pins in fixation of displaced fractures of the radial head, *Arch Ortop Trauma Surg*, 1990, **109**, 258–261.

R. Huiskes, *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**, 124–134.

M. S. Hunt, Development of carbon fiber/polysulfone orthopaedic implants, *Materials & Design*, 1987, **8**(2), 113–119.

Z. M. Huang and S. Ramakrishna, Development of knitted fabric reinforced composite materials for prosthetic application, *Advanced Composites Letters*, 1999, **8**(6), 289–294.

A. Ignatius, K. Unterricker, K. Wenger, M. Richter, and L. Claes, A new composite made of polyurethane and glass ceramic in a loaded implant model: A biomechanical and histological analysis, *Journal of Materials Science: Materials in Medicine*, 1997, **8**, 753–756.

T. Imai, F. Watari, S. Yamagata, M. Kobayashi, K. Nagayama, and S. Nakamura, Effects of water immersion on mechanical properties of new esthetic orthodontic wire, *American Journal of Orthodontics and Dentofacial Orthopedics*, November 1999, 533–538.

S. Iannace, G. Sabatini, L. Ambrosio, and L. Nicolais, Mechanical behavior of composite artificial tendons and ligaments, *Biomaterials*, 1995, **16**(9), 675–680.

N. Inoue, Y. Hirasawa, T. Hirai, and T. Katayama, Composite materials in biomedical engineering, *Materiaux & Techniques*, 1994, No. 4–5, 23–26.

F. Issidor, P. Odman, and K. Brondum, Intermitttent loading of teeth restored using prefabricated carbon fiber posts, *International Journal of Prosthodontics*, 1996, **9**(2), 131–136.

J. Jancar, A. T. Dibenedetto, and A. J. Goldberg, Thermoplastic fiber-reinforced composites for dentistry Part II Effect of moisture on flexural properties of unidirectional composites, *Journal of Materials Science: Materials in Medicine*, 1993, **4**, 562–568.

J. Kettunen, A. Makela, H. Miettinen, T. Nevalainen, M. Heikkila, P. Tormala, and P. Rokkanen, Fixation of femoral shaft osteotomy with an intramedullary composite rod: An experimental study on dogs with a two-year follow-up, *Journal of Biomaterials Science Polymer Edition*, 1999, **10**(1), 33–45.

C. P. A. T. Klein, H. B. M. van der Lubbe, and K. de Groot, A plastic composite of alginate with calcium phosphate granules as implant material: An *in vivo* study, *Biomaterials*, 1987, **8**, 308–310.

K. C. Kennedy, T. Chen, and R. P. Kusy, Behavior of photopolymerized silicate-glass-fiber-reinforced dimethacrylate composites subjected to hydrothermal ageing, *Journal of Materials Science: Materials in Medicine*, 1998, **9**, 243–248.

M. W. King, N. Poddevin, R. Guidoin, Y. Marois, B. Cronier, A. Y. Belanger, and J. P. Delagoutte, Designing textile structures to repair and replace knee ligaments, *Canadian Textile Journal*, 1996, **113**(3), 53–57.

J. C. Knowles and G. W. Hastings, *In vitro* and *in vivo* investigation of a range of phosphate glass-reinforced polyhydroxybutyrate-based degradable composites, *Journal of Materials Science: Materials in Medicine*, 1993, **4**, 102–106.

S. Kocvara, C. H. Kliment, J. Kubat, M. Stol, Z. Ott, and J. Dvorak, Gel-fabric prostheses of the ureter, *Journal of Biomedical Materials Research*, 1967, **1**, 325–336.

W. Krause and R. S. Mathis, Fatigue properties of acrylic bone cements: Review of the literature, *Journal of Biomedical Materials Research: Applied Biomaterials*, 1988, **22**(A1), 37–53.

W. R. Krause, S. H. Park, and R. A. Straup, Mechanical properties of BIS-GMA resin short glass fiber composites , *Journal of Biomedical Materials Research*, 1989, **23**, 1195–1211.

S. M. Kumta, R. Spinner, and P. C. Leung, Absorbable intramedullary implants for hand fractures, *Journal of Bone Joint Surgery*, 1992, **74b**, 563–566.

T. W. Lin, A. A. Corvelli, C. G. Frondoza, J. C. Roberts, and D. S. Hungerford, Glass peek composite promotes proliferation and osteocalcin production of human osteoblastic cells, *Journal of Biomedical Materials Research*, 1997, **36**(2), 137–144.

Q. Liu, J. R. de Wijn, and C. A. van Blitterwijk, Composite biomaterials with chemical bonding between hydroxyapatite filler particles and PEG/PBT copolymer matrix, *Journal of Biomedical Materials Research*, 1998, **40**(3), 490–497.

J. P. Louis and M. Dabadie, Fibrous carbon implants for the maintenance of bone volume after tooth avulsion: first clinical results, *Biomaterials*, 1990, **11**, 525–528.

P. E. Lovdahl and J. I. Nicholls, Pin-retained amalgam cores Vs cast-gold dowel-cores, *Journal of Prosthetic Dentistry*, 1977, **38**(5), 507–514.

M. Marcolongo, P. Ducheyne, J. Garino, and E. Schepers, Bioactive glass fiber/polymeric composites bond to bone tissue, *Journal of Biomedical Materials Research*, 1998, **9**(1), 161–170.

D. G. Mendes, M. Iusim, D. Angel, A. Rotem, D. Mordehovich, M. Roffman, S. Lieberson, and J. Boss, Ligament and tendon substitution with composite carbon fiber strands, *Journal of Biomedical Materials Research*, 1986, **20**, 699–708.

K. Messner and J. Gillquist, Prosthetic replacement of the rabbit medical menicus, *Journal of Biomedical Materials Research*, 1993, **27**, 1165–1173.

V. M. Miettinen and P. K. Vallittu, Release of residual methyl methacrylate into water from glass fiber-poly (methyl methacrylate) composite used in dentures, *Biomaterials*, 1977, **18**, 2, 181–185.

B. J.-L. Moyen, P. J. Lahey, E. H. Weinberg, and W. H. Harris, Effects on intact femora of dogs of the application and removal of metal plates, *Journal of Bone and Joint Surgery*, 1978, **60A**(7), 940–947.

J. B. Park, *Biomaterials Science and Engineering*, Plenum Press, New York, 1984.

J. B. Park and R. S. Lakes, *Biomaterials: An Introduction*, Plenum Press, New York, 1992, pp. 169–183.

G. Peluso, L. Ambrosio, M. Cinquegrani, L. Nicolis, S. Saiello, and G. Tajana, Rat peritoneal immune response to carbon fiber reinforced epoxy composite implants, *Biomaterials*, 1991, **12**, 231–235.

R. M. Pilliar, R. Blackwell, I. Macnab, and H. U. Cameron, Carbon fiber-reinforced bone cement in orthopedic surgery, *Journal of Biomedical Materials Research*, 1976, **10**, 893–906.

P. Pongor, J. Betts, D. S. Muckle, and G. Bentley, Woven carbon surface replacement in the knee: independent clinical review, *Biomaterials*, 1992, **13**, 15, 1070–1076.

M. N. Pradas and R. D. Calleja, Reproduction in a polymer composite of some mechanical features of tendons and ligaments, *High Performance Biomaterials: A Comprehensive Guide to Medical and Pharmaceutical Applications*, ed. M. Szycher, Technomic Publishing Co., Inc., Lancaster, USA, 1991, 519–523.

S. Ramakrishna, J. Mayer, E. Wintermantel, and K. W. Leong, Biomedical applications of polymer-composite materials: A review, *Composites Science and Technology*, 2001, **61**, 1189–1224.

G. C. Robin, Below-knee drop-foot braces: Stresses during use and evaluation of design, *Biomechanics of Medical Devices*, ed. D. N. Ghista, Marcel Dekker Inc., 1981, pp. 535–567.

P. Rokkanen, O. Bostman, S. Vainionpaa, *et al.*, Biodegradable implants in fracture fixation, early results of treatment of fractures of the ankle, *Lancet*, 1985, **1**, 1422–1424.

N. Rushton and T. Rae, The intra-articular response to particulate carbon fiber reinforced high density polyethylene and its constituents: an experimental study in mice, *Biomaterials*, 1984, **5**, 352–356.

S. Saha and S. Pal, Mechanical properties of bone cement: a review, *Journal of Biomedical Materials Research*, 1984, **18**, 435–462.

S. Santavirta, Y. T. Konttinen, T. Saito, *et al.*, Immune response to polyglycolide acid implants, *Journal of Bone Joint Surgery*, 1990, **72-B**, 597–600.

Kh. G. Schmitt-Thomas, Z. G. Yang, and T. Hiermer, Performace characterization of polymeric composite implant rod subjected to torsion, *Proceedings of ICCM-11*, Gold Coast, Australia, 1997, V277–V286.

E. Sclippa and K. Piekarski, Carbon fiber reinforced polyethylene for possible orthopedic uses, *Journal of Biomedical Materials Research*, 1973, **7**, 59–70.

B. B. Seedham, Ligament reconstruction with reference to the anterior cruciate ligament of the knee, *Mechanics of Human Joints: Physiology, Pathophysiology, and Treatment*, eds. V. Wright and E. L. Radin, Marcel Dekker Inc., New York, USA, 1993, 163–201.

C. Silverton, A. O. Rosenberg, R. M. Barden, M. B. Sheinkop, and J. O. Galante, The prosthesis-bone interface adjacent to tibial components inserted without cement, *Journal of Bone and Joint Surgery*, 1996, **78-A**(3), 340–347.

J. A. Simoes, A. T. Marques, and G. Jeronimidis, Design of a controlled-stiffness composite proximal femoral prosthesis, *Composites Science and Technology*, 1999, **60**, 559–567.

J. Tamura, K. Kawanabe, T. Yamamuro, T. Nakamura, T. Kokubo, S. Yoshihara, and T. Shibuya, Bioactive bone cement: The effect of amounts of glass powder and histologic changes with time, *Journal of Biomedical Materials Research*, 1995, **29**, 551–559.

S. H. Teoh, Z. G. Tang, and S. Ramakrishna, Development of thin elastomeric composite membranes for biomedical applications, *Journal of Materials Science: Materials in Medicine*, 1999, **10**, 343–352.

L. D. T. Topoleski, P. Ducheyne, and J. M. Cuckler, The fracture toughness of titanium-fiber-reinforced bone cement, *Journal of Biomedical Materials Research*, 1992, **26**, 1599–1617.

P. Tormala, P. Rokkanen, J. Laiho, *et al.*, Material for osteosynthesis devices, *US Patent 4,743,257*, 1988.

P. Tormala, P. Rokkanen, S. Vainionpaa, *et al.*, Surgical materials and devices, *US Patent 4,968,317*, 1990.

H. K. Uhthoff and M. Finnegan, The effects of metal plates on post-traumatic remodeling and bone mass, *Journal of Bone and Joint Surgery*, 1983, **65B**(1), 66–71.

J. R. Urbaniak, D. S. Bright, and J. E. Hopkins, Replacement of intervertebral discs in chimpanzees by silicone-Dacron implants: A preliminary report, *Journal of Biomedical Materials Research Symposium*, 1973, **4**, 165–186.

A. D. C. Valdevit, N. Inoue, B. A. MacWilliams, and L. L. Anderson, Methods for mechanical testing of spinal constructs, *Spine*, 1996, **10**(2), 231–248.

H. D. Wagner and D. Cohn, Use of high-performance polyethylene fibers as a reinforcing phase in poly(methylmethacrylate) bone cement, *Biomaterials*, 1989, **10**, 139–141.

P. S. Walker and G. W. Blunn, Biomechanical principles of total knee replacement design, *Basic Orthopaedic Biomechanics*, eds. V. C. Mow and W. C. Hayes, Lippincott-Raven Publishers, Philadelphia, USA, 1997, pp. 461–493.

R. Ward and R. J. Minns, Woven carbon fiber mesh patch versus Dacron mesh in the repair of experimental defects in the lumbar fascia of rabbits, *Biomaterials*, 1989, **10**, 425–428.

L. M. Wenz, K. Merritt, S. A. Brown, A. Moet, and R. D. Steffee, *In vitro* biocompatibility of polyetheretherketone and polysulphone composites, *Journal of Biomedical Materials Research*, 1990, **24**, 207–215.

J. A. Werkmeister, G. A. Edwards, F. Casagranda, J. F. White, and J. A. M. Ramshaw, Evaluation of a collagen-based biosynthetic material for the repair of abdominal wall defects, *Journal of Biomedical Materials Research*, 1998, **39**(3), 429–436.

E. Whiteside, The effect of stem fit on bone hypertrophy and pain relief in cementless total hip arthroplasty, *Clinical Orthopedics*, 1989, **247**, 138–147.

D. F. Williams, A. McNamara, and R. M. Turner, Potential of Polyetheretherketone (PEEK) and carbon-fiber-reinforced PEEK in medical applications, *Journal of Materials Science Letters*, 1987, **6**, 188–190.

S. L. Y. Woo, W. H. Akeson, B. Levenetz, R. D. Coutts, J. V. Matthews, and D. Amiel, Potential application of graphite fiber and methyl methacrylate resin composites as internal fixation plates, *Journal of Biomedical Materials Research*, 1974, **8**, 321–338.

T. M. Wright, T. Fukubayashi, and A. H. Burstein, The effect of carbon fiber reinforcement on contact area, contact pressure, and time-dependent deformation in polyethylene tibial components, *Journal of Biomedical Materials Research*, 1981, **15**, 719–730.

M. Zimmerman, J. R. Parsons, and H. Alexander, The design and analysis of a laminated partially degradable composite bone plate for fracture fixation, *Journal of Biomedical Materials Research: Applied Biomaterials*, 1987, **21**(A3), 345–361.

S. W. Zufall, K. C. Kennedy, and R. P. Kusy, Frictional characteristics of composite orthodontic arch wires against stainless steel and ceramic brackets in the passive and active configurations, *Journal of Materials Science: Materials in Medicine*, 1998, **9**, 611–620.

Chapter 7

CASE STUDIES

In previous chapters, fundamental aspects of biocomposites together with their overall use in biomedical fields have been described. In this chapter, we illustrate some specific applications of biocomposites mainly for hard tissue repairing and substitute purposes.

7.1 Dental Applications

In this section, development of a functionally graded composite dental post, orthodontic composite archwires and brackets are presented. In these applications, efforts were focused to use composite materials. The key points of this section are braided fabric reinforcement and interface control to give functional and high mechanical property on products.

7.1.1 *Functionally graded composite dental post*

The main role of the dental post is to provide retention to the core of an endodontically treated tooth (Fig. 6.2). When occlusial force is applied to the crown, the force is transferred to dentin through core and post. In such cases, stress concentration occurs at the apical end, which initiates root fracture. This phenomenon is dependent on post geometry, material choice of the post and adhesion between a post and dentin. There are two main different geometries of post, i.e. tapered and paralleled posts. In order to avoid root fracture, the stiffness of a post is decreased with tapered geometry. However, it is clear that tapered posts exhibit wedging effect at root

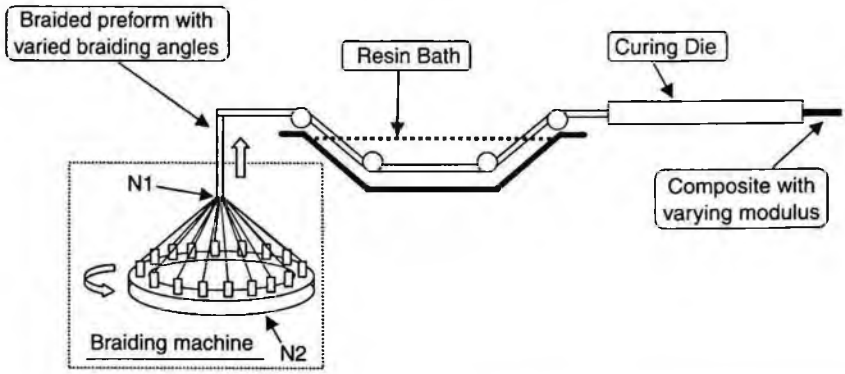
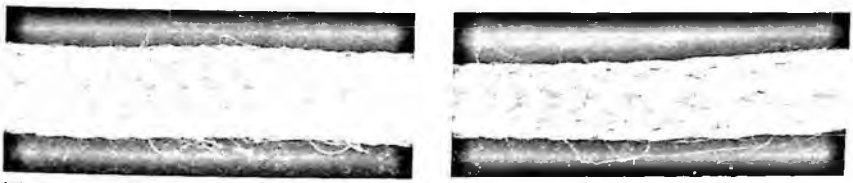


Fig. 7.1 Schematic representation of the fabrication of functionally graded composite dental post.

canal and produce stress concentration around the tip of post [Standlee *et al.*, 1972; Cooney *et al.*, 1986]. In terms of material of post, stainless-steel, titanium and ceramics have around 10–17 times higher stiffness than dentin (12 GPa). The modulus mismatch between dentin and post is one of the prime reasons to cause stress concentration at the root of the teeth. Therefore, an ideal dental post should have parallel geometry without taper and its stiffness property should be close to dentin. In the last decade, uni-directional carbon/epoxy composite post (Compositpost: R.T.D. Meylan, France) has focused attention as an alternative to high stiffness posts to avoid the risk of root fracture [Isidor *et al.*, 1996; Asmussen *et al.*, 1999]. It was clear that Compositpost possessed the similar stiffness of dentin and showed excellent fatigue property as compared with metal posts. The dental post needs to be stiff in the coronal region, i.e. in the region of the core, so that the core is not stressed excessively when occlusial force is applied to the crown [Ramakrishna *et al.*, 2001]. Hence, an optimal dental post would be one, which has a high stiffness in the coronal region and this stiffness gradually reducing to the value of dentine at the apical end. The high stiffness will take away the stress from the core and the gradually reduction of stiffness would unload the stress from the post to the dentine uniformly. The gradual unloading would eliminate stress concentration and reduce the interfacial shear stress. The requirements of an ideal dental post were achieved by braided composite material developed by authors.



(a) Preform with constant braiding angle. (b) Preform with graded braiding angle.

Fig. 7.2 Photographs showing braided preforms with (a) constant and (b) graded braiding angles.

7.1.1.1 *Fabrication method of functionally graded composite dental post*

The fabrication process (Fig. 7.1) of dental post consists of two important components. One, the preforming process, where the fiber yarns are placed at the required graded angles and two, the composite forming process, where, the preform is impregnated with matrix and consolidated. Dental post is essentially a composite rod with diameter ranging from 0.8 mm to 2.0 mm and length of 20 mm. The preforming was done using the braiding process and the composite forming by the pultrusion process. The fibers used were carbon for the core and glass for sheathing. The rods made using this composite are a hybrid of braided glass sheathing for aesthetic appearance and braided carbon fibers to satisfy the stiffness requirements of the intended application. The primary parameters, which control the braiding angle during braiding, are the braider yarn carrier speed and the take-up speed. It is the ratio of the carrier speed and take-up speed, which defines the braiding angle. In the conventional braiding process, this ratio is kept constant and hence braided preforms of constant braid angle are produced, as seen in Fig. 7.2(a). In the present approach, this ratio was varied continuously to achieve a graded braiding angle, as shown in Fig. 7.2(b). For the present study, a KOKUBUN braiding machine (Model: 102-C13) with 20 carriers was used. In this machine, two gears mesh to provide the required speed ratio between the take-up and the carrier. The braided preform has a 1×1 intersection repeating unit pattern (diamond braiding structure, i.e. ten carriers were used to make a braided preform). The present 10 carriers machine could achieve a braiding angle from about 10° to 43°. In practice, braiding angle can range from 5° which is almost parallel yarn braid to approximately 85° in a hoop yarn braid. However, because of the geometry

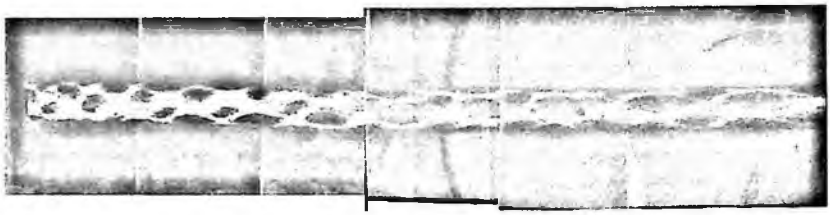


Fig. 7.3 Photograph of a fabricated composite dental post.

limitations of yarn jamming, braiding angle that can be achieved for a particular braided preform is controlled by factors such as number of carriers and yarn size. Braided preform used as reinforcement was continuous and was good for pulling. Here, the continuous pultrusion process was modified into an intermitted process. For the present study, the wetted preform was pulled through a long and straight Teflon tube with a circular cross-section. Once the tube length was filled with the preform, the wetted preform was cut and pulled through another Teflon tube. Later, these tubes were placed in an oven at 60°C for 6 hours to ensure complete curing. After curing, the tube was cut and length of FGM (Functionally Graded Material) in the post form was removed. This length was cut at the appropriate section to get the dental post, as shown in Fig. 7.3.

7.1.1.2 *Mechanical performance*

Finite element method was used to calculate the stress state within the restored tooth. The stress analysis was performed on a maxillary central incisor. The finite element model (Fig. 7.4) includes alveolar bone, dentin, periodontal ligament, dental post, gutta-percha, core and crown. The geometry of the tooth was derived by cross-sectioning a typical central incisor and plotting the geometry using a profile-meter. Later, the restoration and the bone support were constructed over this cross-section. Stainless steel post was considered in this study to form the datum for evaluating the performance of the functionally graded composite dental post. The stainless steel post was modeled as isotropic material with uniform Young's modulus along the length. The functionally graded composite post was modeled as transversely isotropic with Young's modulus in the longitudinal direction varying from 20 GPa at the apical end to 80 GPa at the coronal in seven

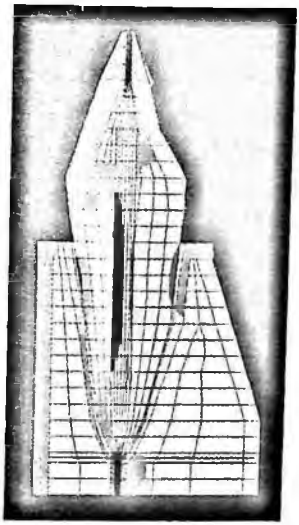


Fig. 7.4 A finite element mesh of an endodontically treated tooth with a dental post.

steps. The post was 13.0 mm in length and 1.25 mm in diameter. The *3D* finite element model consists of 3,308 hexagonal-8 elements and has 3,759 nodes. The mastication loading was class-I occlusion 45-degrees to the longitudinal axis with a magnitude of 100 N. The boundary conditions at the bottom end of the bone were restrained in all degrees of freedom to simulate the alveolar bone holding the tooth. ABAQUS standard version 5.7 general-purpose finite element analysis program was executed on an Indigo 2, Silicon Graphics work station. Normal and shear stresses were calculated and are presented in Figs. 7.5 and 7.6, respectively. It is seen that the peak tensile and shear stresses for a functionally graded dental post reduce to about half the values of those for conventional stainless steel dental post. The reduction in tensile stress in brittle material, like dentine, is advantageous as dentine can resist larger cyclic loading and chance of root (dentine) failure is reduced. The reduction in the interface shear stress reduces the chances of the post loosening from dentine as the stress on bonding cement comes down. Both the root fracture and post loosening constitute major causes of restoration failures. The functionally graded dental post has an advantage by reducing the severity of these problems.

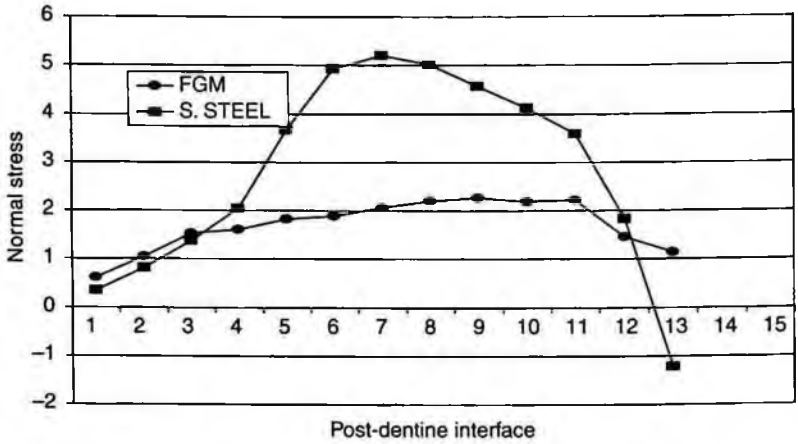


Fig. 7.5 Normal stress distribution along post-dentine interface.

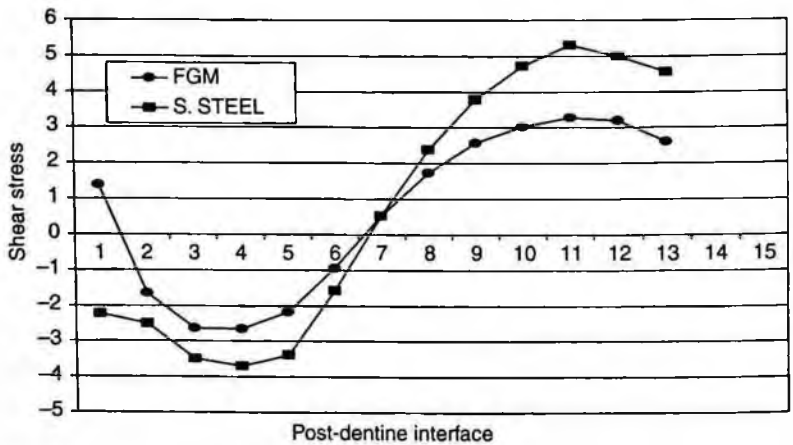


Fig. 7.6 Shear stress distribution along post-dentine interface.

7.1.2 *Composite orthodontic archwires and brackets*

Orthodontics is a branch of dentistry concerned with the growth, guidance, correction and maintenance of the dento-facial complex. The area of orthodontic practice includes the diagnosis, prevention, interception and treatment of all forms of malocclusion of the teeth and alterations in the supporting structures. The practice involves the application of corrective

appliances to move teeth, commonly called braces. The components of braces (a fixed appliance) are orthodontic archwires and orthodontic brackets [Bennett and McLaughlin, 1993], as shown in Fig. 7.7. In recent years, patients receiving orthodontic treatment are concerned with their aesthetic appearance. Current materials used in orthodontic archwires are metals. Type of metal wires is dependent on the stage of treatment, less stiff wires, such as nickel–titanium and beta–titanium are used in the initial stage and stiffer stainless-steel and Co–Cr alloy are used in the final stage and their detailed mechanical performance is shown in some review literatures [Kapila and Sachdeva, 1989]. As well as archwires, metal brackets were once the main appliance in the treatment. In these days, the patient's critical concern in orthodontic treatment is the aesthetic performance of archwires and brackets. Metallic color appearance of metal archwires and brackets gives an unnatural impression of patients. Hence, tooth-colored or translucent archwires and brackets provide superior presentation for the patient.

For this reason, in bracket application, products such as clear polymer brackets and ceramic brackets were introduced. Clear polymer bracket (polycarbonate) was first reported by Newman [1965] to make a labial appliance almost invisible. Although the aesthetic appearance of polymer brackets is satisfactory, they are easily fractured when the orthodontic force is applied through archwire [Rains *et al.*, 1977]. Due to this reason, ceramic brackets (mono- and polycrystalline alumina) [Swantz, 1988] have been widely acceptable instead of polymer brackets in late 1986 and patients

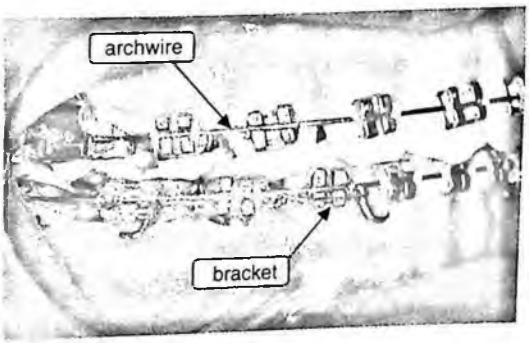


Fig. 7.7 Photograph of orthodontic archwires and brackets. (Original photo is referred from Bennett and McLaughlin, 1993.)

appreciated their better aesthetic qualities. Ceramic brackets have shortcomings of brittle fracture and are expensive to manufacture [Birnie and Edin, 1990]. Especially, remnants of fractured bracket by occlusial force may be harmful to patients. Although similar aesthetic importance is needed on archwires as well, no aesthetic wires are available except Teflon coated stainless-steel wire [Postlethwaiste, 1992].

These facts made researchers pay attention to polymer composite materials into the development of the archwires and brackets to replace conventional products. With regards to composite brackets, the trend currently heads to fabricate discontinuous chopped glass fiber reinforced polycarbonate made of injection molding. Their mechanical evaluation, such as, friction with archwire [Bazakidou *et al.*, 1997] and bonding performance with enamel [Crow, 1995; Guan *et al.*, 2001], has been conducted by many researchers. However, since mechanical improvement using chopped fiber is still not enough, polycarbonate bracket which has a rectangular metal piece inserted into a slot of a bracket, is popularly used as an alternative of composite brackets [Feldner *et al.*, 1994]. Hence, in terms of composite brackets, a method to obtain more effective reinforcement of polymer bracket using braided preform without diminishing the aesthetic property was persued.

Composite archwires have been made of continuous glass fiber reinforced composite using pultrusion fabrication process. The main materials were epoxy resin, polyethylene-terephthalate-glycol (PETG) and poly(1,4) cyclohexylene dimethylene terephthalate glycol (PCTG) [Goldberg and Burstone, 1992], and poly-methyl-methacrylate (PMMA) reinforced by *S*-glass fiber which has higher strength than *E*-glass fiber. Photopolymerized dental resins, such as urethanedimethacrylate (UDMA), bisphenol-A glycidyl methacrylate (bis-GMA), however have become popular matrix since they have been accepted for teeth restoration. The group of Watari *et al.* gave the biocompatible function to glass fibers and fabricated composite archwires with photopolymerized resins [Watari *et al.*, 1998]. However, the interface property, which controls mechanical property of composite archwires was not considered. Silane coupling agent is popularly applied to glass fiber in order to encourage the chemical bonding between fiber and matrix. It should be mentioned that the extent of its effect on mechanical property of composite is dependent on silane treatment condition such as

silane concentration [Suzuki *et al.*, 1992]. Therefore, influence of silane concentration was focused on composite archwire.

7.1.2.1 *Fabrication of aesthetic composite archwires*

A technique, which can directly fabricate curved archwires based on tube-shrinkage process is used. The fabrication process is pictorially described in Fig. 7.8. A number of resin-impregnated fiberglass yarns were bundled together, and then introduced into a plastic (polyolefin) tube which is heat shrinkable. As the inner diameter of the tube was larger than that of the yarn bundle, fiber damages could be controlled to a minimum at this stage. In the next step, hot air was applied to the tube from one end (at the top grip) to the other (at the bottom grip). While, the heated tube shrank, the extra resin is pushed out of the bottom end of the tube. Then, the shrunk tube with the resin-impregnated fiber yarn bundle inside was loaded into the female part of a mold, where the required curvature to the archwire is given, before

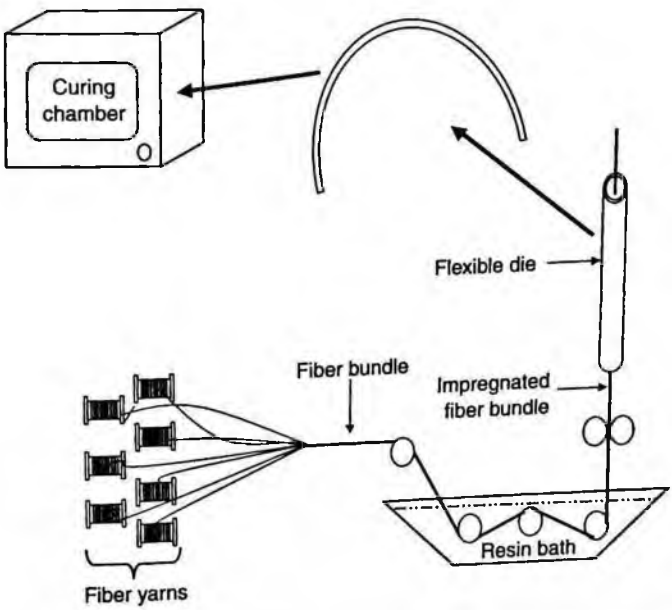


Fig. 7.8 Fabrication process of aesthetic composite archwires.

being placed into an oven. Finally, the tube is carefully removed and the composite archwire extracted.

It is well known that interface property of composite materials plays an important role. In this study, aesthetic composite archwires were fabricated from glass/epoxy unidirectional composites. Three different groups of glass fibers, i.e. (1) fibers treated with γ-aminopropyltrimethoxysilane (aminosilane), (2) fibers treated with γ-glycidoxypropyltrimethoxysilane (epoxysilane) and (3) non-treated fiber were used. Furthermore, three different silane concentrations were applied to glass fiber in each silane coupling agent. The treated fibers were impregnated with the epoxy resin and pulled through a polyolefin tube (0.5 mm diameter) heated at 100°C for 4 hours. An average volume fraction of aesthetic composite wire was 48%. Color appearance of developed aesthetic composite archwire is shown in Fig. 7.9 and the cross-section photo of composite archwire (Fig. 7.10) shows good fiber dispersion and negligible void amounts were confirmed.

The mechanical properties of orthodontic archwire are discussed in terms of bending because this mode of deformation is considered more representative of clinical use. The actual force imposed on the teeth is expressed as the flexural load on the unloading curve in a hysterisis test. This is because the teeth give rise to the stress relaxation in an orthodontic

Fig. 7.9 Color appearance of fabricated aesthetic composite archwire. The color of the wire is close to that of a tooth.

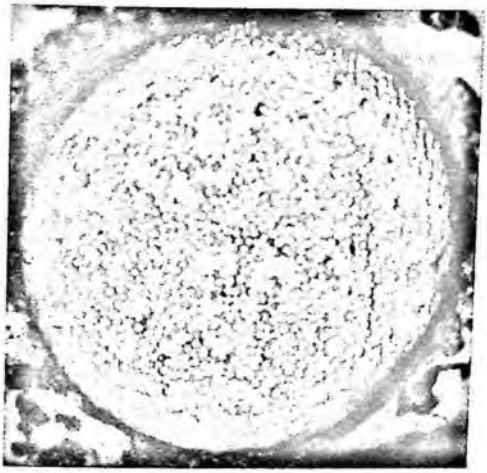


Fig. 7.10 Cross-sectional photo of aesthetic composite archwire.

wire by yielding movement backwards against the imposing force originated by the elastic recovery of orthodontic wire. Three point bending tests were carried out for the developed composite archwires with 1.0 mm/min cross-head speed. A 14 mm span length was chosen in accordance with the distance between the labial mid-line to the first premolar on the lower dental arch. In each silane concentration, the specimens were deflected to 1.5 mm displacement and unloaded to zero load, with the same cross-head speed to obtain the hysterisis curve. Fourteen specimens were prepared in each glass fiber group. Commercially available 0.018 inch (0.45 mm) Ni–Ti round wire (Reflex® wire; TP Orthodontics, Inc) was also tested for comparison purpose. Figure 7.11 shows that silane concentration does influence the bending stiffness, as well as the loading capacity of the wire. In epoxysilane treatment, the load capacity had a peak (2.8 N) at 1.0wt% silane concentration and this value was around 30% higher than that of 0wt% silane concentration. Similarly, specimens treated by aminosilane also showed load increment from 0wt% to 1.0wt%, however, this value was saturated with further a increase of silane concentration. It must be mentioned that Ni–Ti round wire indicated lower load value than composite archwires which was not even treated by coupling agent. The bending stiffness of the wire improved with 1.0wt% epoxysilane, giving the highest

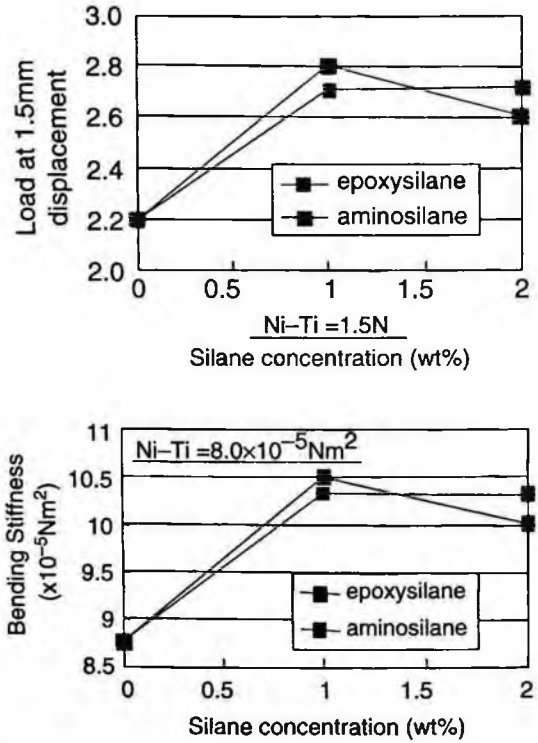


Fig. 7.11 Relationship between bending properties ((a) bending load at 1.5 mm displacement and (b) bending stiffness) and silane concentration.

value and then dropping at 2.0wt%. In the case of aminosilane, the bending stiffness for both silane concentrations (1.0wt% and 2.0wt%) was almost the same. Ni–Ti wire also showed lower stiffness than composite archwires. Hence, it was clear that 1.0wt% epoxysilane treatment was suitable to obtain high bending performance as glass/epoxy composite archwires. Furthermore, glass/epoxy composite archwires developed by the authors can be an alternative to Ni–Ti wire.

7.1.2.2 *Aesthetic braided composite brackets*

Fabrication of aesthetic braided composite brackets was conducted by conventional pultrusion method as shown in Fig. 7.12. Flattened tubular glass braided fabric was immersed into a resin bath of transparent epoxy and

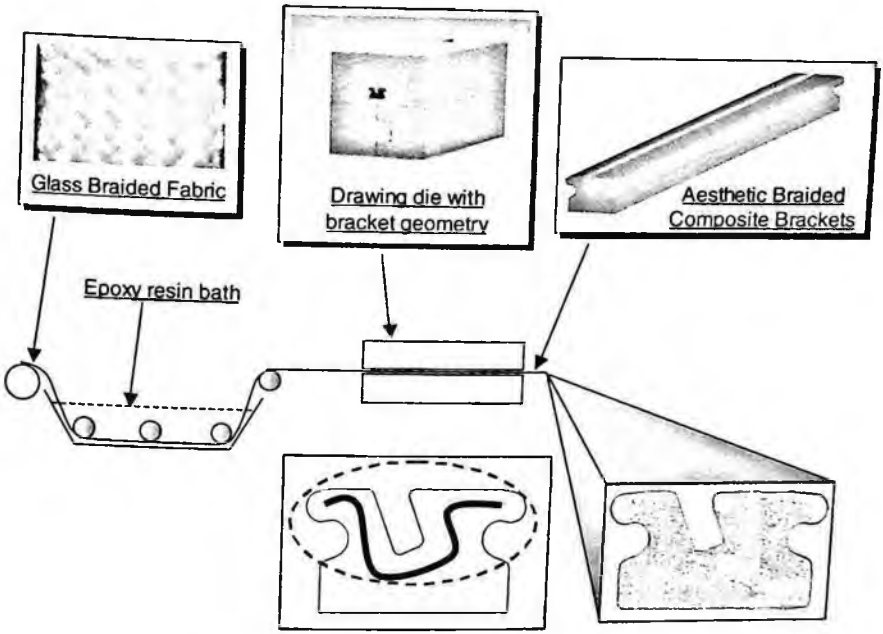


Fig. 7.12 Fabrication of aesthetic braided composite brackets.

continuously inserted into a pultrusion die with 50 mm length whose geometry is a bracket's cross-sectional profile. Since pultrusion process is used to produce composite bracket with constant cross-section, bracket geometry was chosen to be single-wing and flat-base brackets. In order to reinforce tie-wings and archwire slot, flattened braided preform was filling the tie-wings and archwire slot, thus the braided preform takes U-shape in the molding. Pultruded glass braided preform was consolidated with epoxy resin in the molding and the consolidated bracket rod was cut into the specimen with 3.2–3.4 mm mesial-distal width. Translucency of the developed aesthetic braided composite bracket was satisfactory with acceptable distribution of glass fibers. Volume fraction of aesthetic braided composite bracket was 50%. In this study, four different brackets, i.e. (1) epoxy (2) aesthetic braided composite (3) stainless-steel (3M Co. Ltd., Unitek Victory Series), and (4) ceramic (polycrystalline alumina) (3M Co. Ltd., Unitek Transcend 6000) were prepared, as shown in Fig. 7.13, to investigate the mechanical performance of the developed composite bracket. All brackets have 0.022 inch (0.55 mm) slot and are intended to use for mandibular

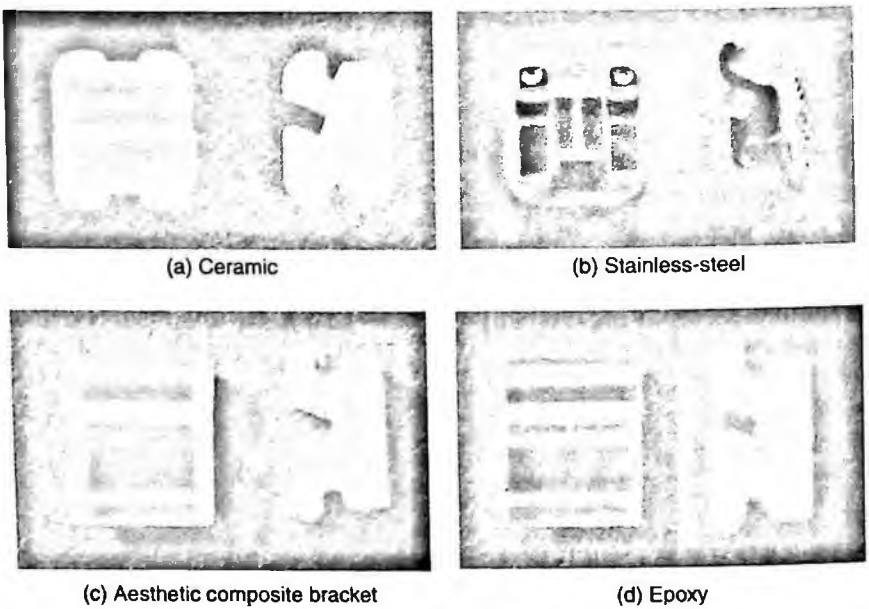


Fig. 7.13 Photographs of four different types of brackets.

premolars. It must be mentioned that stainless-steel brackets have double edgewise design although the other brackets have single edgewise design. As seen in Fig. 7.14, brackets were bonded to the bracket holder with certain angle in order to apply the force to only one tie-wing. The force was applied to the specimen with 0.5 mm/min cross-head speed.

Figure 7.15 shows load-displacement curves of each tested bracket. Epoxy bracket depicted ductile load increment with increasing displacement. Around 0.8 mm displacement, slight load drop was recognized and the load increased again. Since gradual load decrease was recognized after the maximum load, testing was stopped at 0.3 mm. Similarly, aesthetic braided composite bracket also showed non-linear load increase against the displacement and showed gradual load drop after the maximum loading. Stainless-steel bracket showed larger load increment with increasing displacement than epoxy and composite brackets and gradually reached the maximum loading with increasing displacement. In the case of composite and stainless-steel brackets, testing was also stopped at 0.3 mm displacement and the load was released in the same manner used for the epoxy

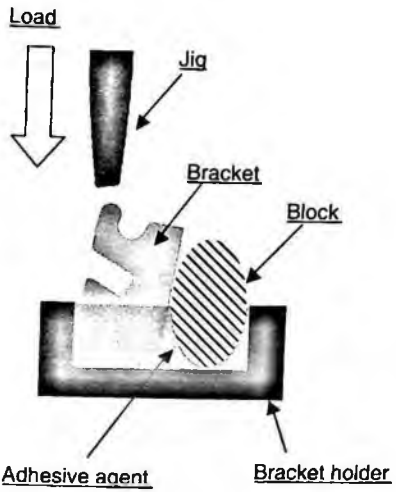


Fig. 7.14 Schematic drawing of tie-wing test of brackets.

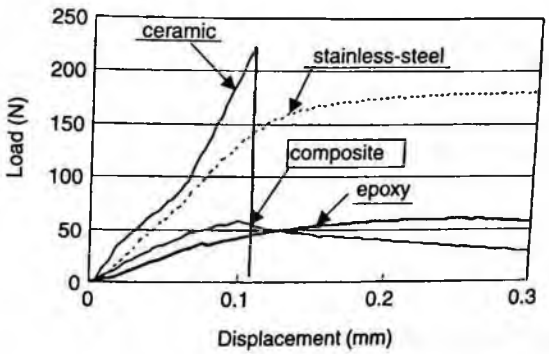


Fig. 7.15 Load-displacement curves of each tested bracket.

bracket. On the contrary, ceramic bracket displayed linear load increase and showed brittle load drop.

In the clinical situation, an archwire with 0.5 mm diameter is inserted into the slot (0.55 mm width) of a bracket. Therefore, large deformation of tie wing are not of practical significance. Since ceramic bracket fractured near 0.1 mm displacement, load value at 0.1 mm displacement in each bracket was chosen and compared in Fig. 7.16. Ceramic bracket had the

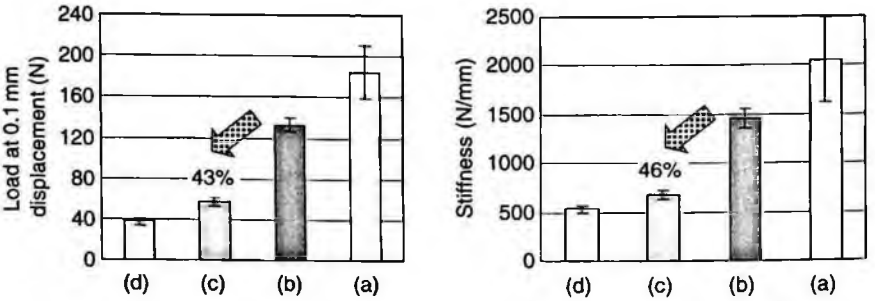


Fig. 7.16 Comparison of load at 0.1mm displacement and stiffness.

highest load and stiffness values, however, it must be mentioned that this value may be dropped easily if ceramic bracket is scratched by orthodontic instrument. In this case, induced micro cracks easily propagate in ceramic bracket because of its low fracture toughness. As compared with stainless-steel bracket, aesthetic composite bracket could sustain 43% of the load sustained by stainless steel bracket, and its stiffness was 46% of stainless steel. The present aesthetic composite bracket was fabricated with glass fiber untreated by silane coupling agent. Hence, there is much possibility to increase these values by applying proper surface treatment to glass fibers. In conclusion, the prototype aesthetic composite brackets fabricated with glass braided preform appeared to have superior fracture resistance on application of shear force on the gingival tie-wing.

7.2 Orthopedics Applications

In orthopedics applications, composite materials have to possess good biocompatible harmony with human tissue and excellent mechanical property under body fluid. Since there is still further discussion in the usage of thermoset resin as matrix of composite, the usage of high performance thermoplastics polymer such as poly-ether-ether-ketone (PEEK) is highly recommended. In order to use PEEK as matrix on textile composites, the bottle neck is the difficult fabrication process to obtain product with good quality. In this section, development of micro-braided fabrication technique to solve existing fabrication problem is presented and successfully

fabricated braided carbon/PEEK composite compression bone plate is introduced. It can be considered that this fabrication process can be applied to other orthopedic fixations, such as the intramedullary nail, the spine disk and rod and the total hip replacement.

7.2.1 *Braided carbon/PEEK composite compression bone plate*

Compression bone plate is used to fix diaphyseal fracture of a long bone. The function of the compression bone plate is to transfer the compression force between bone fragments to support the body and protect the fracture area with proper alignment of the fragments throughout the healing process (normally taking around 1–1.5 years). The conventional compression plates are made of metals. The elastic modulus of human cortical bone is in the range of 15 GPa to 26 GPa, much lower than that of metals. This material modulus mismatch leads to a situation that the metal plates provoke the decrease of bone mineral mass [Olerud and Danckwardt-Lilliestrom, 1968] and occasionally cause bone refracture [Hidaka and Gustilo, 1984] after the plate removal. This phenomenon is widely recognized as 'Stress-Shielding Effect' or 'Stress-Protection' [Ali *et al.*, 1990]. In order to avoid 'Stress-Shielding Effect', it is desirable to use plates whose mechanical properties are close to those of the cortical bone. In the past, although many efforts were conducted to fabricate composite compression bone plates, these plates were made of unidirectional fiber laminates [Bradley and Hastings, 1980]. Therefore, the braided Carbon/PEEK composite compression bone plates of three different thicknesses and with three different braiding angles were comparatively studied in terms of their bending performance [Fujihara *et al.*, 1993].

Flat braided fabrics were preformed using micro-braided yarns, as seen in Fig. 7.17. The unique feature of the micro-braided yarn is that the reinforcing and matrix fibers are easily mixed using a simple braiding technique and a consistent reinforcement can be achieved after a hot-press fabrication. Braided fabrics with three different braiding angles, i.e. 5°, 10° and 15° were prepared to investigate the influence of braiding angle on bending property of composite plates. In order to see the influence of plate thickness on bending properties, three different thicknesses (2.6, 3.2 and 3.8 mm: corresponds to 8, 10 and 12 layers of a fabric) were prepared. After inserting braided fabrics into the mold, the mold was put into a hot press machine. Plate

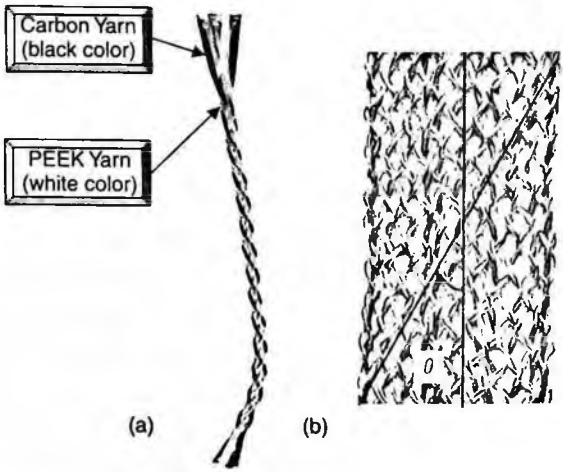


Fig. 7.17 Photograph of a flat braided fabric [b] made of micro-braided yarns [a]. (θ indicates a braiding angle and a braided fabric has certain range of braiding angle with constant preform width.)

Fig. 7.18 Photograph of a braided carbon/PEEK composite compression plate.

fabrication was conducted at 380°C for 20 minutes with 4.6 MPa pressure. Fabricated braided Carbon/PEEK composite bone plate (see Fig. 7.18) had an averaged volume fraction of 53% in all types of specimens. A stainless-steel narrow Dynamic Compression Plate (DCP) of AO Institute, which is used widely in surgery, was also tested for comparative purpose. The geometry of the AO plate is 103 mm long, 13 mm wide and 3.8 mm thick. In this study, static 4-point bending tests were conducted with a cross-head speed of 1.0 mm/min at room temperature. The compression composite bone plate has six holes and the curvature geometry to fix diaphyseal part of a long bone. Upper and lower span lengths are 41 mm and 73 mm respectively.

The measured bending results are plotted against plate thickness, as shown in Fig. 7.19, in which the percentage values are relative to those of the stainless-steel plate. Thus, the bending performance of the stainless-steel

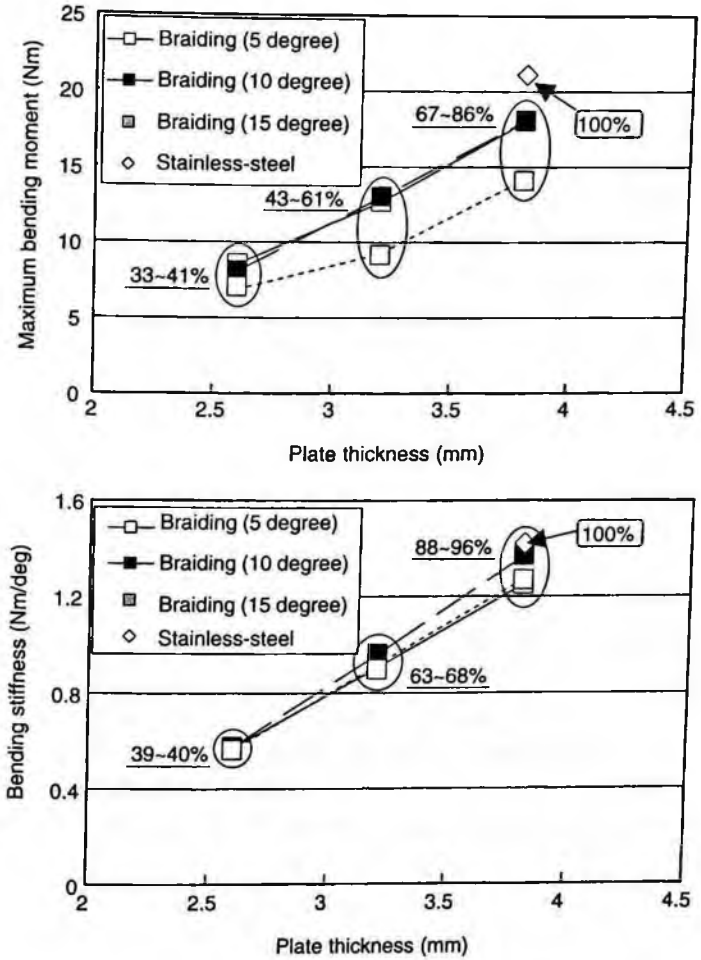


Fig. 7.19 Bending data comparison against plate thickness.

plate was taken as 100%. The braided composite plates showed an increase in the maximum bending moment with the increase of plate thickness for all the braiding angles. The plates which have the minimum thickness showed only 8% difference in the bending moment among different braiding angle specimens. This difference was more distinct for the thicker specimens, i.e. 18% for the 3.2 mm and 19% for the 3.8 mm thick specimens among different braiding angles. The bending stiffness of the braided composite

plates increased with the increase in thickness, and the specimens of 3.8 mm thickness indicated quite a close stiffness to that of the stainless-steel plate. There was no significant stiffness variation for the 2.6 mm thick specimens with the different braiding angles. This was not applicable, however, to the specimens with larger thicknesses.

The bending stiffness of a compression plate has a close relationship with bone healing process, the previous animal and clinical trials have suggested an appropriate choice for the plate stiffness to achieve good bone healing. One of the examples discussed the relationship between the plate stiffness and the bone strength after a fixation treatment was given by Bradley *et al.* [1979]. In their work, fractured dog femurs were treated using a stainless-steel compression plate and a carbon/polysulfone UD laminate plate, respectively. The result tells that when the stiffness of composite plate is around 10–25% value of stainless-steel, the healed bone strengths are much higher than those treated by even stiffer composite plates. However, since the stress environment of the dog femur is not as severe as that of human femur or tibia due to the weight and bone site differences, the stiffness values of composite plates for human femur and tibia should be higher. For instance, 53% [Tayton and Bradley, 1983] and 43% [Tayton *et al.*, 1982] of stainless-steel plate stiffness were recommended to obtain a quick human bone healing. In the case of human forearm where less stress environment than that of femur and tibia is expected, the appropriate stiffness of composite plates is around 30% that of the stainless steel plate [Ali *et al.*, 1990]. The braided composite compression bone plate developed in this study showed 40% stiffness value with 2.6 mm thickness. Thus, this plate can be appropriate for forearm fixation. On the other hand, since the braided composite bone plates with 3.2 mm thickness showed 63–68% stiffness of the stainless steel plate, these composite plates may be used to fix femur and tibia bones. It is noted that in both cases, the current braided composite bone plates are thinner than previously developed composite plates.

7.3 Prosthetic Socket Application

7.3.1 *Potential of composites for socket application*

Accidents (such as traffic accidents, natural disasters, etc), disease, congenital disorders, and wars give rise to a lot of amputations each year. Artificial

limbs can help those patients with amputations to resume their active life and to restore their confidence. The artificial limbs (prostheses) are customized devices. Namely, every limb is fabricated individually for each amputee. In general, a below knee prosthesis consists of three parts, i.e. a socket, a shank, and a foot (Fig. 6.3). The most highly customized and important part of the prosthesis is the socket, which is a thin-walled receptacle that fits over the remaining portion of an amputated limb and provides an interface between the artificial limb and the amputee's body.

According to their fabrication process, sockets can be divided into two categories, i.e. direct and indirect sockets. As the name suggested, a direct socket is to be directly fabricated on the stump of a patient, whereas an indirect socket must be made on a mold which is the reflection of the patient's stump. Many materials including wood and metals have once been used in socket fabrication. However, the majority of the sockets in current usage has been made by using some plastic materials. One popular way to obtain a pure plastic (single-phase) socket is through sheet casting. A major drawback with such a single-phase plastic socket is that it lacks stiffness and strength, and hence another rigid carriage is generally needed. Such a socket, although usually more comfortable to a patient due to its relatively soft structure, only serves as a cushion between the patient's amputee and the rigid carriage in general.

The situation can be significantly improved by reinforcing the polymer plastics with textile preforms. It has been recognized that knitted fabrics are best applicable to make a multi-phase indirect socket. The flexibility of knitted fabrics makes them to be easily and tightly wrapped on a mold. After impregnated with a polymer matrix, the multilayer knitted fabric composite socket is strong and stiff enough to sustain all the functions of the patient's amputee. No additional carriage will be required. As the constituent materials, the stitching parameters, the number of layers, etc can be selectively changed, a suitable composite socket can be developed which is soft and comfortable to a patient so that no cushion is needed and yet is strong enough to sustain all the patient's loads.

7.3.2 *Fabrication of a composite indirect socket*

As mentioned above, an indirect socket must be fabricated on a plastic mold. To make the plastic mold, a three-stage process is observed: measurement,

fabrication, and rectification. The first stage is physical measurement. A skilled prosthetist takes down all the required amputee's data and a negative mold is formed by wrapping plaster-of-Paris onto the stump of the patient. The second stage is to create a preliminary positive mold based on the negative mold by using plaster mixed with water. The last stage is to rectify the preliminary mold. This stage is necessary for the fabrication of a comfortable socket. Generally, a socket exactly the same as the negative mold cannot serve the patient's amputee comfortably, as the negative mold has not been formed under the patient's standing and walking conditions. The recorded measurements and the information acquired from past experience form a guide to the amount of rectification for the preliminary mold, and provide the desired dimensions, i.e. the dimensions of the final socket.

The fabrication of a composite socket begins by wrapping several layers of knitted fabrics on the plaster mold, vacuuming the fabrics enclosed in a plastic bag, and impregnating the vacuumed fabrics with a polymer resin (mostly polyester or acrylic resin). The most commonly used fibers are glass (*E*-glass) and nylon fibers. In a typical fabrication successful for clinic applications, three kinds of fabrics are used as reinforcement. One is a glass fiber fabric, another is a combined glass and nylon fiber fabric, and the third is a nylon fiber fabric, with pure glass fabric at the innermost layer and the pure nylon fabric at the outermost layer. The glass fabric has the coarsest whereas the nylon fabric has the finest stitch density, with the combined glass and nylon fiber fabric in between. Each fabric contributes two layers, giving a total of six layers for the reinforcement. The fabric-draped mold is then enclosed by a plastic film (PVA sheet) with one hole at the top for the resin to come in and another hole at the bottom connected to a vacuum pump. After applying a vacuum pressure, the resin is filled into the fabrics through the top hole and the resin-impregnated fabrics are cured at room temperature. The cured composite is sucked tightly onto the plaster

Table 7.1 Tensile properties of a new direct socket material [Huang and Ramakrishna, 1999].

Young's Modulus (GPa)		Ultimate Strength (MPa)		Ultimate Strain (%)	
Longitudinal	Transverse	Longitudinal	Transverse	Longitudinal	Transverse
2.67	1.51	36.1	28.6	1.68	3.39

Longitudinal = socket (tube) axial; transverse = transverse to the socket axis

Table 7.2 Bending properties of a new direct socket material [Huang and Ramakrishna, 1999].

Young's Modulus (GPa)		Ultimate Strength (MPa)		Ultimate Strain (%)	
Longitudinal	Transverse	Longitudinal	Transverse	Longitudinal	Transverse
1.64	1.18	46.3	37.2	3.65	5.43

mold, giving a final socket shape. The mechanical properties of so-obtained socket material are summarized in Tables 7.1 and 7.2 respectively. It is noted that the socket is generally not considered as a primary load-carrying element, as the patient's amputee generally does not apply very high loads. The composite sockets are used mainly for their easy fabrication and long durability.

7.3.3 *Fabrication of a direct socket*

Although indirect sockets are currently predominant in clinical applications, several drawbacks have been realized, as a consequence of a plaster mold used in fabrication. First, the quality of the mold and hence the final socket fabrication depends on the prosthetist's skills. Second, the mold preparation and the socket wearing require the patient to visit the hospital at least two times. Third, the mold preparation gives additional cost to the patient. To overcome these drawbacks, direct sockets have been proposed. Their benefit is to combine the casting, manufacturing and fitting of a definitive transtibal prosthetic socket in one visit.

Several commercial direct socket materials such as ALPs socket pre-forms (ALPs Socket-Pro™) and ICEX socket system (ICEX Socket System) are available on the market. An ICEX pre-form is pre-impregnated carbon or glass braided fabric with a water-activated resin. It utilizes air-pressure to cast the ICEX fiber fabric directly onto the patient's residual limb. The resin pre-impregnated fabric will soon become a rigid form once it is sucked into water. The fabrication begins by wrapping the water-sucked fabric onto the patient's stump. Meanwhile, air-pressure is manually applied to fit the wrapped fabric onto the stump. Much fewer fabric layers (usually one or two layers) have to be used during the fabrication, because the resin hardens very quickly (in about 5 minutes). An ICEX direct socket reinforced with braided carbon fiber fabric, which can fit users of up to 100 kg in weight is shown in Fig. 7.20.

Fig. 7.20 An ICEX direct socket made of braided carbon fiber fabric and a water-activated resin [Ossur Company Ltd., http:///www.ossur.com/].

It is noted that not all kinds of amputees can be fitted on direct sockets. A sufficiently long residual limb is necessary for a direct socket to be fabricated. Further, several limitations make the present direct socket less comparable with a carefully fabricated indirect socket. First, the air-pressure is applied through an enclosed bag which does not face perfectly all parts of the socket material during the fabrication. Second, the pre-impregnated braided carbon fabric does not show enough formability before curing so that the cured product is not as good as the indirect sockets both in fitting to the patient stump and outside the product shape. Third, the curing period of the water-activated resin is too limited to achieve a perfect socket fabrication. Finally, the socket suction is performed while the patient is not under standing and walking conditions. Further development and modification on direct sockets is required.

7.3.4 *Outline of a new direct socket*

Modifications to the ALPs or ICEX sockets may focus on two parts [Huang and Ramakrishna, 1999]. First, a new hydrostatic pressure casting system may be employed, with which the pressure can be applied more uniformly and the patient's stump is under standing and walking condition during the

Fig. 7.21 A hydraulic casting system [Huang and Ramakrishna, 1999].

fabrication. Another modification may concern with material. The basic requirements for the new indirect socket material are: (a) to have sufficient strength in its final form; (b) to have a longer period of curing time for the prosthetist to shape the socket before it hardens; and (c) to have none or little exothermic release during its hardening.

The proposal of a new direct socket possessed three distinguished features [Huang and Ramakrishna, 1999]. First, a hydrostatic casting system was used (Fig. 7.21), which allows more uniform application of the casting pressure; second, a UV activated resin was incorporated, which takes much longer time to cure than a water-activated resin; and third, a sleeve shape of two layers of rib knitted fabrics (Fig. 7.22) were employed as reinforcement, which are more flexible than braided fabric.

Prior to the socket fabrication, the fabrics were impregnated with the resin without any UV light. The socket fabrication began by wrapping the resin-impregnated fabric onto the stump of the patient. The hydraulic pressure (Fig. 7.21) was applied while the resin was in curing with a UV

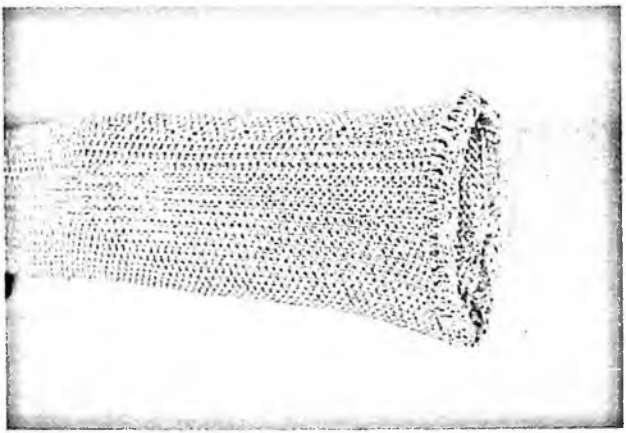


Fig. 7.22 A sleeve form of rib knit Kevlar 49 fiber fabric used in this study.

Fig. 7.23 A prototype direct socket using two layers of Kevlar 49 fiber rib knitted fabrics and a UV activated resin [Huang and Ramakrishna, 1999].

light. The casting was applied when the patient was in standing/walking condition, so that the socket could be formed more practically and would be softer and more comfortable to the patient's later wearing. The prototype socket is shown in Fig. 7.23.

Mechanical characterization was performed in terms of a flat panel sample, which was fabricated in a set up having the similar condition as that for the socket fabrication. Measured properties were comparable to those shown in Tables 7.1 and 7.2 [Huang and Ramakrishna, 1999]. Further investigation on incorporating the hydrostatic casting and the UV lighting systems into a compacted set up is required, and durability study is also necessary.

7.4 External Fixator Application

7.4.1 *Introduction*

A human bone can be fractured into segments/pieces. When such a case happens, doctor needs to use additional internal or external (Fig. 7.24) or both kinds of fixators to initially fix the fractured segments or pieces into their proper positions, and then to keep the positions during the bone healing. Typical internal fixators are bone plates, whereas typical external fixators are of Ilizarov type [Ilizarov, 1992]. The Ilizarov system is also used in limb lengthening and the correction of congenital and pathological orthopedic deformities. Limb correction is a gradual process, which lengthens and re-aligns the bone to restore normal function [Ilizarov, 1992].

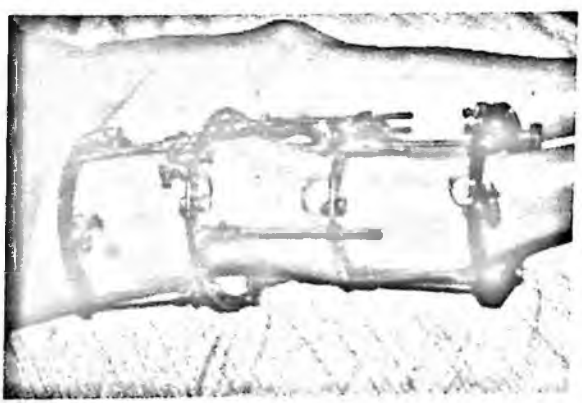


Fig. 7.24 Use of an external fixator for a patient where conservative reduction and treatment is insufficient.

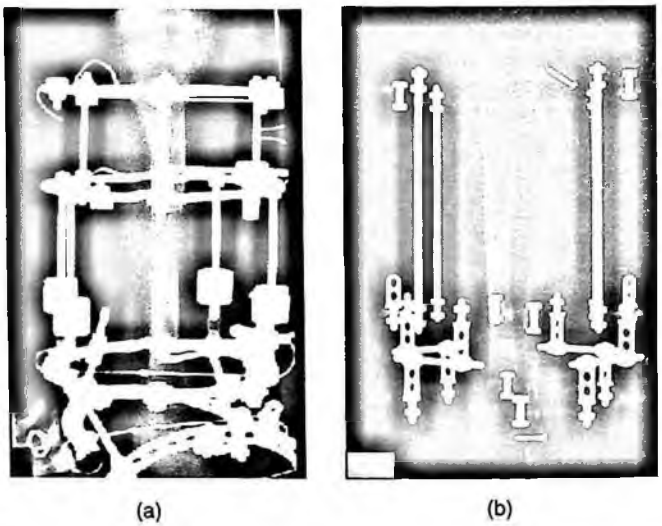


Fig. 7.25 Radiograph of (a) Stainless Steel Ilizarov external fixator system, (b) the system using random short carbon fiber composite rings.

Most clinically used Ilizarov external fixator components are made from stainless steel, titanium, or aluminum metals, which are radio-opaque (Fig. 7.25a). X-ray radiographic examination of bone fractures and bone healing processes is a valuable method in the treatment and measurement of patients. One of the problems commonly experienced during treatment of patients with fractures is the difficulty in interpretation of radiographs taken through casts of fixation devices where the synthetic material interferes with the quality of radiograph produced, as a consequence of bone shielding [Wytch *et al.*, 1991]. To facilitate clinical evaluation of the fracture healing process, shielding of the fracture site by the fixation device must be minimized. By using composite components, the bone shielding problem can be satisfactorily addressed. In this chapter, we only focus on a composite ring.

7.4.2 *Composite ring fabrication*

Due to its complicated structural geometry, it would be costly to fabricate a composite ring with continuous fiber reinforcement. Instead, chopped carbon (T300) fibers (in 6 to 8 mm in length) and thermoplastic PEEK matrix

were used as constituent materials. Sample composite ring was fabricated using a hand lay-up method. A three-piece compression mold made of stainless steel was employed. The bottom piece holds all the pins, which will generate holes of the ring. The central piece of the mold functions as the female part, and the top piece as the male part. The PEEK pellets were placed in the mold layer by layer with a chopped carbon fiber layer in between. The mold was then put into the hot press machine under pressure and heat treatment, according to the temperature versus time cycle as shown in Fig. 7.26. The composite panel (ring) obtained is shown in Fig. 7.27, which has a thickness of 2 mm and a fiber volume fraction of 0.186. As expected, this composite ring is radio transparent, giving no radio-opaque (Fig. 7.27b).

7.4.3 *Performance characterization*

Since the rings carry a large compressive load under the wire tension it was necessary to measure their in-plane compressive strength. Axial compressive stiffness was also checked for comparison with the metal ring. The test for the in-plane compressive strength of a ring was carried out according to the ASTM F1746-96 standard. The in-plane test results are shown in

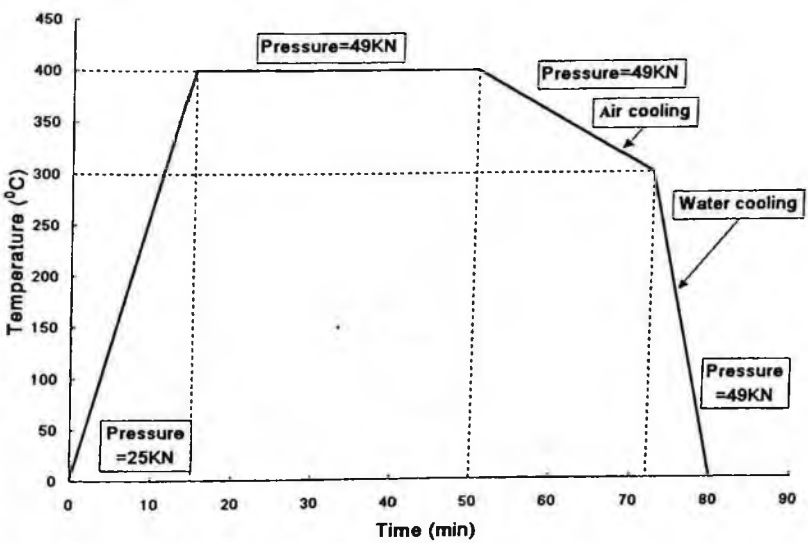


Fig. 7.26 Temperature *vs* time cycle for processing of CF/PEEK composite ring.

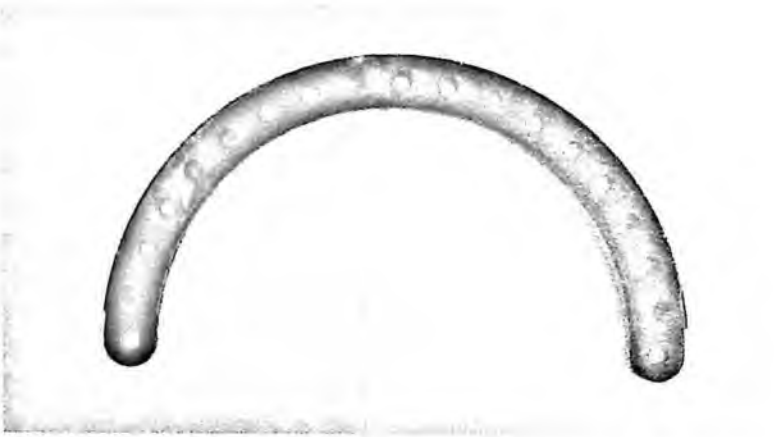


Fig. 7.27 A half-ring of an external fixator made from chopped carbon fibers and thermoplastic PEEK matrix through a hot press.

Table 7.3 In-plane compressive test results of chopped CF/PEEK composite ring.

Compressive Stiffness (N/mm)	96.34
Compressive Yield Strength (N)	1135
Maximum Compressive strength (N)	1426

Table 7.3. During the test, the sample failed in a brittle matter, initiated near a hole region.

Axial compression is the most important mode of loading for a bone-device construct [Carter, 1985], as shown in Fig. 7.28. Compressive stiffness and fixation stability may be increased by the use of more wires and by decreasing the spacing between rings. The testing procedure followed ISO1438 Standard, which is intended to evaluate the stability of various configurations of external fixator devices. In the test, four rings of 200-mm diameter were used, with two rings on each side of the fracture gap. Perspex® tubes of 30 mm outer diameter and 4 mm wall thickness were used as bone analogues. A gap of 20 mm between the fragments was maintained to ensure that the entire load could be transmitted through the fixator. The wires were tensioned to 1200 N. Axial compressive stiffness of the frame using the composite rings was compared with that of a frame having clinically used stainless steel rings. Table 7.4 shows the axial and

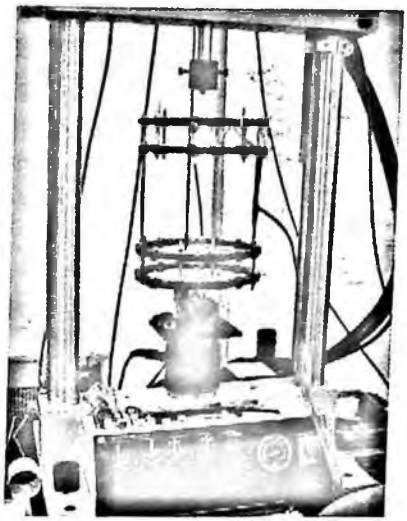


Fig. 7.28 Frame assembly picture of an axial compressive test for chopped CF/PEEK composite rings.

Table 7.4 Summary of results from the Axial Compressive tests on external fixator rings.

	Radiolucency	Weight (g)	Axial Stiffness (N/mm)	Comparative Stiffness
Stainless Steel	No	105	60.92	3674.56
Short Chopped Carbon	Yes	70	39.91	2545.65

comparative stiffness of the two systems tested. The axial compressive stiffness [Fredrick, 1990] was defined as the force per unit deflection, whereas the comparative stiffness was calculated on the basis of the deflection in the direction of the bone axis as per

$$\text{Comparative stiffness} = \frac{\text{force} \times \text{ring radius}}{\text{deflection}}. \tag{7.1}$$

Although the composite ring has inferior mechanical properties compared with the stainless steel ring, the composite ring has less mass density and can be made with a bigger size (thicker) without increasing the ring weight. In this way, the composite ring stiffness and load carrying capacity can be redesigned according to the worst-case loading condition. Such

design can be accomplished based on FEM (finite element method) analysis, and has been discussed by Baidya *et al.* [2003]. It has been concluded there that the thickness and the width of the composite ring should be increased by 75% and 57% respectively in order that (1) the bending stiffness of the composite ring is equal to that of the stainless-steel ring, (2) the worst-case loading stress is well below the yield limit of the material, and (3) the deformation is less than 1% of the nominal diameter of the ring.

7.5 Conclusion

Fibrous composite materials used in dental and orthopedic applications have been introduced. In dental application, composite materials are required with good aesthetic and mechanical properties besides good biocompatibility. On contrast, orthopedic application needs materials, which have excellent mechanical property and biocompatibility in order to sustain severe mechanical environment. In both applications, effective reinforcing methods using textile performs and interface property of composite have not been proven. Consequently, there are more avenues for the feasibility of composite materials usage in dental and orthopedic applications. Therefore, dental composite products (dental post, orthodontic archwires and brackets) that paid attention to the usage of textile preform and proper interface control were firstly introduced. It was clear that textile preform is greatly useful to obtain the functional mechanical requirement in dental post and orthodontic brackets. The importance of interface control to obtain the best mechanical performance was also shown in aesthetic composite archwire. Novel fabrication method of textile carbon/PEEK composites which may be applied to several orthopedic applications was introduced. Fabricated braided carbon/PEEK composite compression plate showed potential mechanical performance as alternative of stainless-steel plate.

References

M. S. Ali, T. A. French, G. W. Hastings, T. Rae, N. Rushton, E. R. S. Ross, and C. H. Wynn-Jones, Carbon fibre composite bone plates, *Journal of Bone Joint Surgery*, 1990, **72B**(4), 586–591.

E. Asmussen, A. Peutzfeldt, and T. Heitmann, Stiffness, elastic limit, and strength of newer types of endodontic posts, *J Dent*, 1999, **27**, 275–278.

K. P. Baidya, A. Ritchie, Z. M. Huang, S. Ramakrishna, and M. Rahman, An investigation on the polymer composite medical device-external fixator, *Journal of Reinforced Plastics & Composites*, 2003 (in press).

E. Bazakidou, R. S. Nanda, M. G. Duncanson, and P. Sinha. Evaluation of frictional resistance in esthetic brackets, *American Journal of Orthodontics and Dentofacial Orthopedics*, 1997, **112**, 138–144.

J. C. Bennett and R. P. McLaughlin, *Orthodontic Treatment Mechanics and the Preadjusted Appliance*, Wolfe Publishing, 1993.

D. Birnie and M. Edin, Orthodontic materials update-ceramic brackets, *Br J Ortho*, 1990, **17**, 71–75.

J. S. Bradley and G. W. Hastings, Carbon fibre reinforced plastics for orthopaedic implants, Mechanical Properties of Biomaterials, eds. G. W. Hastings and D. F. Williams, 1980, pp. 379–386.

G. W. Bradley, G. B. Mckenna, H. K. Dunn, A. U. Daniels, and W. O. Statton, Effects of flexural rigidity of plates on bone healing, *Journal of Bone Joint Surgery*, 1979, **61A**(6), 866–872.

D. R. Carter, Biomechanics of Bone, *The Biomechanics of Trauma Nahum*, eds. A. M. Nahum and J. Melvin, Norwalk, Connecticut, Appleton-Century-Crofts, 1985, pp. 157.

J. P. Cooney, A. A. Caputo, and K. C. Trabert, Retention and stress distribution of tapered-end endodontic posts, *Journal of Prosthetic Dentistry*, 1986, **55**, 540–546.

V. Crow, *Ex vivo* shear bond strength of fiberglass reinforced aesthetic brackets, *British Journal of Orthodontics*, 1995, **22**, 325–330.

J. C. Feldner, N. K. Sarker, J. J. Sheridan, and D. M. Lancaster, *In vitro* torque-deformation characteristics of orthodontic polycarbonate brackets, *American Journal of Orthodontics and Dentofacial Orthopedics*, 1994, **106**, 265–272.

J. K. Fredrick, Technical note: Evaluation of new Ilizarov rings, *Bulletin of the Hospital for Joint Diseases Orthopaedic Institute*, 1990, **50**, 8.

K. Fujihara, Z. M. Huang, S. Ramakrishna, K. Satkunanantham, and H. Hamada, Performance study of braided carbon/PEEK composite compression bone plates, *Biomaterials*, 2003 (in press).

A. J. Goldberg and C. J. Burstone, The use of continuous fiber reinforcement in dentistry, *Dent Mat*, 1992, **8**, 197–202.

G. Guan, T. Takano-Yamamoto, M. Miyamoto, T. Yamashiro, H. Noguchi, K. Ishikawa, and K. Suzuki, An approach to enhance the interface adhesion between an orthodontic plastic bracket and adhesive, *European Journal of Orthodontics*, 2001, **23**, 425–432.

S. Hidaka and R. B. Gustilo, Refracture of bones of the forearm after plate removal, *Journal of Bone Joint Surgery*, 1984, **64A**(8), 1241–1243.

Z. M. Huang and S. Ramakrishna, Development of knitted fabric reinforced composite material for prosthetic application, *Advanced Composites Letters*, 1999, **8**(6), 289–294.

G. Ilizarov, *The transosseous osteosynthesis: Theoretical and clinical aspects of the regeneration and growth of tissues*, Springer-Verlag, Berlin, New York, 1992.

F. Isidor, P. Odman, and K. Brondum, Intermittent loading of teeth restored using prefabricated carbon fiber posts, *The International Journal of Prosthodontics*, 1996, **9**(2), 131–136.

S. Kapila and R. Sachdeva, Mechanical properties and clinical applications of orthodontic wires, *Am J Ortho*, 1989, **96**, 100–109.

G. V. Newman, Epoxy adhesives for orthodontic attachments, *Progress Report of Am J Ortho*, 1965, **51**, 901–912.

S. Olerud and G. Danckwardt-Lilliestrom, Fracture healing in compression osteosynthesis in the dog, *Journal of Bone Joint Surgery*, 1968, **50B**(4), 844–851.

K. M. Postlethwaiste, Advances in fixed appliance design and use: 1. Bracket and archwires, *Dental Update*, 1992, pp. 276–280.

M. D. Rains, S. J. Chaconas, A. A. Caputo, and R. Rand, Stress analysis of plastic bracket configurations, *Journal of Clinical Orthodontics*, 1977, **11**(2), 120–125.

S. Ramakrishna, V. K. Ganesh, S. H. Teoh, P. L. Loh, and C. L. Chew, Fiber reinforced composite product with graded stiffness. US patent No 6,287,122 B1, 11 September 2001.

J. P. Standlee, A. A. Caputo, E. W. Collard, and M. H. Pollack, Analysis of stress distribution by endodontic posts, *Oral Surgery Oral Medicine Oral Pathology*, 1972, **33**, 952–960.

Y. Suzuki, Z. Maekawa, H. Hamada, M. Kibune, M. Hojo, and N. Ikuta, Influence of absorption behavior of a silane coupling agent on interlaminar fracture in glass

fibre fabric-reinforced unsaturated polyester laminates, *Journal of Materials Science*, 1992, **27**, 6782–6790.

M. L. Swantz, Ceramic brackets, *Journal of Clinical Orthodontics*, 1988, **22**, 82–89.

K. Tayton and J. Bradley, How stiff should emi-rigid fixation of the human tibia be? *Journal of Bone Joint Surgery*, 1983, **65B**, 312–315.

K. Tayton, C. Johnson-Nurse, B. Mckibbin, J. Bradleym, and G. W. Hastings, The use of semi-rigid carbon fibre reinforced plastic plates for fixation of human fractures, *Journal of Bone Joint Surgery*, 1982, **64B**(1), 105–111.

F. Watari, S. Yamagata, T. Imai, and S. Nakamura, The fabrication and properties of aesthetic FRP wires for use in orthodontics, *Journal of Materials Science*, 1998, **33**, 5661–5664.

R. Wytch, G. P. Ashcroft, G. Mckenzie, D. Wardlaw, and W. M. Ledingham, Radiographic assessment of splinting bandages, *Injury: the British Journal of Accident Injury*, 1991, **22**(1), 41–44.

GLOSSARY

Apical	Near the apex or extremity of a conical structure, such as the tip of the root of a tooth
AAPM	American Association of Physicists in Medicine
ABF	Aortobifemoral bypass graft or surgery
Acetabulum	The socket portion of the hip joint
ADA	American Dental Association, American Diabetic Association, American Dietetic Association, or Americans with Disabilities Act
ADR	Adverse drug reaction
AFB	Aortofemoral bypass graft or surgery
Allograft	Transplanted tissue or organ between unrelated individuals of the same species
Alumina	Aluminium oxide
Alveolar bone	Bone structure that supports and surrounds the roots of teeth
AMA	American Medical Association
Amalgam	Alloy of two or more metals, one of which is mercury
AMS 800	Type of artificial urinary sphincter
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 alveolar bone and ossification of the periodontal membrane

Anterior	Direction referring to the front side of the body
Anterior cervical plates	System of plates and screws placed anteriorly in the spine for fixation of unstable spine fractures and dislocations or to stabilize the spine after surgery
Antibiotic beads	Any beadlike material impregnated with antibiotics for use in treating bone and joints infections. The beads, typically composed of ploy(methyl methacrylate), are packed into the area of infection. The antibiotics help treat the infection, and the bead packing material provides mechanical support in an area of missing or weakened bone
Arch Wire	Wire which is attached to brackets to move teeth
Arthritis	Inflammation of joints
Arthrodesis	Fusion or fixation of a joint
Arthroplasty	Surgical repair of a joint
Arthroplasty	Generic term for any joint surgery that is designed
Articular cartilage	Cartilage at the ends of bones in joints which serve as the articulating, bearing surface
Atrophy	Wasting away of tissues or organs
Autograft	Transplanted tissue or organ transferred from one part of a body to another part of the same body
Band	Metal ring that is placed on teeth to hold onto parts of appliance
Biocompatibility	Acceptance of an implant by surrounding tissues and by the body as a whole
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
Bonding	Process of attaching brackets to teeth using a safe glue

Bone cement	Biomaterial used to secure a firm fixation of joint prostheses, such as hip and knee joints
Bracket	Metal or ceramic or polymer part that is glued onto a tooth and serves as a means of fastening the arch wire
Bridge	Dental prosthesis used to restore a space where there is missing teeth
Callus	The hard substance that is formed around a bone fracture during healing
Cancellous bone	Reticular or spongy tissue of bone where spicules or trabeculae form the interconnecting latticework that is surrounded by connective tissue or bone marrow
Catheter	Instrument (tube) for gaining access to and draining or sampling fluids in the body
Cavity	Small hole in one of the teeth caused by tooth decay
Celestin tube	Nylon reinforced latex tube used to bypass esophageal tumors
Cochlear implant	Type of surgically implanted hearing aid used to treat sensorineural hearing loss
Collagen	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	Physical defect existing since birth
Cortical bone	Compact hard bone with osteons
Core	Preparation built upon a tooth on which a crown is cemented
Crown	Part of tooth that is exposed above the gum line or covered with enamel
CT	Computed tomography, an X-ray technique for producing cross-sectional image of the body

Dacron	Polyethylene terephthalate polyester that is made into fibers
Dental caries	Tooth decay caused by acid-forming microorganisms
Dental Implant	Replacement for one of the missing teeth
Dental restoration	Another name for dental fillings
Dentine	Main substance of the tooth
Dermatitis	Inflammation of skin
Dura mater	Dense, tough connective tissue over the surface of the brain
Elastin	Elastic fibrous mucoprotein in connective tissue
Enamel	Hard, white substance that covers the dentine of the crown of a tooth
Endosseous	Referring to dental implants fixed to the jaw bone
Endodontist	Dentist who specializes in root canal pulp chamber in teeth treatment
Endosteal	Related to the membrane lining the inside of the bone cavities
Extracorporeal	Outside the body
Femur	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
Gingiva	Gum tissue; the dense fibrous tissue overlying the alveolar bone in the mouth and surrounding the necks of teeth
Graft	Transplant
Ground substance	Amorphous polysaccharide material in which cells and fibers are embedded
Hard tissue	General term for calcified structures in the body, such as bone
Heterograft	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
Hydroxyapatite (HA)	Mineral component of bone and teeth
Ilizarov technique	Technique used most often in reconstructive settings to lengthen limbs, transport bone segments, and correct angular deformities
In vitro condition	Simulated *in vivo* condition in the laboratory
In vivo condition	Inside the living body
Intervertebral disc	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	Orthopedic rod or nail inserted into the intramedullary marrow cavity of the bone to promote healing of long bone fractures
Intraosseous implant	Implant inserted into the bone
Kirschener wire	Metal surgical wires
Kyphosis	Abnormally increased convexity in the curvature of the lumbar spine
Ligament	Sheet or band of fibrous connective tissue that join bone to bone, offering support to the joint
Long bones	Bones with large aspect ratio and with distinctive shaped ends, such as femur
Lordosis	Abnormally increased concavity in the curvature of the lumbar spine
LTI carbon	Low-temperature isotropic carbon
Lumen	Space within a tubular structure
Mandibular bone	Lower jaw of the mouth
Maxillary bone	Upper jaw of the mouth
Medullary cavity	Marrow cavity inside the long bones
Myocardium	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
Orthodontics	Treatment performed to correct bite and make smile look good
Orthopedics	Medical field concerned with the skeletal system
Orthotics	Science and engineering of making and fitting orthopedic appliances used externally to the body
Ossicles	Small bones of the middle ear which transmit sound from ear drum to the body
Osteoarthritis	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	Abnormal reduction of the density and increase in porosity of bone due to demineralization
Osteotomy	Cutting of bone to correct a deformity
Periodontal ligament	Periodontium; the connective tissue (ligament) joining the tooth to the alveolar bone
Posterior	Direction referring to the back side of the body
Proplast	Composite material made of fibrous PTFE and carbon
Prosthesis	Device that replaces tissues or organs of the body
Proximal	Nearest the trunk or point of origin; opposed to distal
Pulp	Soft inner portion of tooth, consisting of nerve and blood vessels
Pyrolitic carbon	Isotropic carbon coated onto a substrate in a fluidized bed
Resorption	Dissolution or removal of a substance
Root	Part of tooth with in the gums

Rheumatoid arthritis	Chronic and progressive inflammation of the connective tissue of joints, leading to deformation and disability
Scoliosis	Abnormal lateral (sideward) curvature of a portion of the spine
Silastic	Medical grade silicone rubber
Silica	Ceramic SiO_2
Spondylolisthesis	Forward bending of the body at one of the lower vertebrae
Spondylosis	Any of various degenerative diseases of the spine
Stapes	One of the ossicles of the middle ear
Stenosis	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
Subcutaneous	Beneath the skin
Subperiosteal	Underneath the periosteum
Synovial fluid	Clear viscous fluid that lubricates the surfaces of joints and tendons, secreted by the synovial membrane
Tendon	Band or cord of fibrous tissue connecting muscle to bone
THR	Total hip replacement
Torque	Rotating movement on crown
Thromboembolism	Obstruction in the vascular (blood circulating) system caused by a dislodged thrombus
Thrombosis	Formation of a thrombus, blood clot
Thrombus	A fibrinous blood clot attached at the site of thrombosis
TKR	Total knee replacement
Trachea	Cylinder-shaped tube lined with rings of cartilage that is 115 mm long, from the larynx to the bronchial tubes; the windpipe

Transplantation	Transfer of a tissue or organ from one body to another, or from one location in a body to another
Typodont	Plastic model of a typical mouth showing alignment of teeth
Ureter	Tube that conducts urine from the kidney to the bladder
Urethra	Canal leading from the bladder to the outside for discharging urine
Vascular	Blood vessels
Vitallium	Co-Cr alloy, Howmedica Inc.
Vitreaous carbon	Term generally applied to isotropic carbon with very small crystallites
Wolff's law	Principle relating the internal structure and architecture of bone to external mechanical stimuli
Xenograft	Transplanted tissue or organ transferred from an individual of another species

Definitions in Composites

Absorption	Process in which one material (absorbent) takes in or absorbs another (absorbate)
Aging	Effect, on materials, of exposure to an environment for a period of time
Angleply	Laminae or laminate, which is not uniaxial
Anisotropic	Material possessing directional dependent mechanical and/or physical properties
Aramid	Polymer material having general structure of aromatic rings alternating with amide linkages
Aspect Ratio	Ratio of long dimension to the short dimension

A-stage	Early stage in the reaction of thermosetting polymers in which the material is still soluble in certain liquids or capable of becoming liquid upon heating
Autoclave	Closed vessel for producing an environment of fluid pressure, with or without heat
Braid Angle	Acute angle measured from the axis of braiding
Braid or Braided Fabric	Textile fabric made by intertwining of yarns in bias direction
Braiding	Textile process where two or more yarns are intertwined in the bias direction to form an integrated structure
B-stage	Intermediate (partially cured) stage in the reaction of a thermosetting polymer in which the material swells when in contact with certain liquids and softens when heated but does not dissolve or fuse completely
Coupling Agent	Chemical substance designed to react with both the reinforcement and matrix phases of a composite material to form or promote a stronger bond at the interface
Course	Knitting term for rows of loops or stitches running across a knitted fabric
Course Density	Number of courses per unit length of fabric
Crimp	Waviness or undulation of fibers or yarns
C-stage	Final stage of curing reaction of a thermosetting polymer. After this stage the material is insoluble and infusible
Cure	To change the properties of thermosetting resin irreversibly by chemical reaction (cross-linking). It can be accomplished by addition of cross-linking (curing) agents, with or without catalyst, and with or without heat
Debond	Deliberate separation of interface, e.g. between fiber and matrix

Degradation	Deleterious change in physical, chemical and/or mechanical properties of the material
Delamination	Separation of layers of material (laminae) in a laminate
Denier	Unit for measurement of linear density of fiber, filament, yarn, or strand. Denier is equivalent numerically to the number of grams per 9,000 meters length
Desorption	Desorption is the reverse of absorption, adsorption or both. It is a process in which an absorbed or adsorbed material is released from another material
End	Individual fiber, strand, roving or yarn
Fiber Content	Amount of fiber present in a composite. It is usually expressed as a percentage volume fraction or weight fraction of the composite
Filament Wound	Composite component made by the filament winding method of fabrication
Fill	Same as weft. A set of yarn running at right angles to the warp in a woven fabric
Glass Transition Temperature	Approximate midpoint of the temperature range over which glass transition (a reversible change in amorphous polymer or in amorphous regions of a partially crystalline polymer from a viscous or rubbery condition to a hard and relatively brittle one) takes place
Interface	Boundary between physically distinguishable constituents of a composite material
Interlaminar	Between the laminae of a composite laminate
Interlaminar Shear	Shearing force tending to produce a relative displacement between two laminae in a composite laminate along the plane of their interface
Intralaminar	Within the laminae of a composite laminate
Isotropic	Material having uniform properties in all directions

Kevlar	Trade mark of E. I. du Pont de Nemours and Company for a new family of high strength aramid fibers
Knitted Fabric	Textile fabric constructed by interlocking a series of loops of one or more yarns
Laid-in Yarn	System of longitudinal yarn inserted between the yarns of textile fabric
Lamina	Single ply or layer in a laminate
Laminae	Plural of lamina
Laminate	Term mainly used in describing fiber-reinforced composites. It is a consolidated collection of plies (laminae), which are oriented or placed in required directions
Lay-up	Process in which the laminae or plies are assembled in desired directions before consolidation into a composite laminate
Mat	Fibrous material consisting of randomly oriented fibers which are loosely held together
Matrix content	Amount of matrix material present in a composite
Nonwoven fabrics	Planar assemblies of fibers held together either by mechanical interlocking, by fusing (in the case of thermoplastic fibers), or by bonding with a chemical (solvent or cementing medium) means
Orthotropic	Having three mutually perpendicular planes of elastic symmetry
Plasticizer	Lower molecular weight material deliberately added to a polymer to separate the molecular chains, there by improving its processability
Porosity	Condition of trapped pockets of air, gas, or vacuum within a solid material. It is a ratio nonsolid volume to the solid volume of the material
Preform	Assembly of dry fibers or fabrics which has been prepared according to the geometry and other requirements of the product

Prepreg	Precombined form of reinforcement and matrix materials
Resin Starved Area	Area of composite part where the polymer has a noncontinuous smooth coverage of the fiber(s)
Roving	Mainly used in referring to glass fibers. It contains a number of strands, tows or ends with little or no twist
Sizing	Material, with which filaments are treated, which contains ingredients that provide lubricity to the filament surface. Sizing prevents filament abrasive damage during handling. Some times sizing system also contains a coupling agent that improves the bond between filaments and the matrix polymer .
Strand	Untwisted bundle of fibers
Tex	Unit for measurement of linear density of fiber, filament, yarn, or strand. The Tex number is the weight in grams of a one-kilometer length
Tow	Untwisted bundle of continuous filaments. Mainly used in referring to carbon fibers
Void	Pocket of air, gas or vacuum within a composite
Wale	Series of loops in successive courses lying lengthwise in the fabric, formed by the action of one needle, in knitted fabric
Wale Density	Number of wales per unit length of fabric
Warp	Set of yarn running along the lengthwise direction of the woven fabric
Weft	Set of yarn running at right angles to the warp in a woven fabric
Woven Fabric	Textile fabric composed of interlaced yarns. The specific manner in which the yarns are interlaced determines the weave structure or pattern

Yarn	Widely used term for describing a bundle of continuous filaments or fibers, which are usually twisted. Yarn is a basic material, which is made into textile fabric

Acronyms

BIS-GMA	bis-phenol A glycidyl methacrylate
C	carbon
CF	carbon fibers
GF	glass fibers
HA	hydroxyapatite
HDPE	high density polyethylene
KF	Kevlar fiber
LCP	liquid crystalline polymer
LDPE	low density polyethylene
MMA	methylmethacrylate
PA	polyacetal
PBT	poly(butylene terephthalate)
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	polyglycolic acid
PHB	polyhydroxybutyrate
PHEMA	poly(HEMA) or Poly(2-hydroxyethyl methacrylate)
PLA	polylactic acid
PLDLA	poly L-DL-lactide
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	ultrahighmolecular weight polyethylene

Many years of cumulative research has been conducted on the usage of fiber-reinforced composites for biomedical application, but no one source exists where this topic is dealt with systematically. This book addresses polymer composites applied to bioengineering in a comprehensive manner.

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