32. Implantable Biomedical Devices and Biologically Inspired Materials

Hugh Bruck

Experimental mechanics is playing an important role in the development of new implantable biomedical devices through an advanced understanding of the microstructure/property relationship for biocompatible materials and their effect on the structure/performance of these devices. A similar understanding is also being applied to the development of new biologically inspired materials and systems that are analogs of biological counterparts. This chapter attempts to elucidate on the synergy between the research and development activities in these two areas through the application of experimental mechanics. Fundamental information is provided on the motivation for the science and technology required to develop these areas, and the associated contributions being made by the experimental mechanics community. The challenges that are encountered when investigating the unique mechanical behavior and properties of devices, materials, and systems are also presented. Specific examples are provided to illustrate these issues, and the application of experimental mechanics techniques, such as Photoelasticity, Digital Image Correlation, and Nanoindentation, to understand and characterize them at multiple length scales.

It is the purpose of this chapter to describe the application of experimental mechanics in understanding the mechanics of implantable biomedical devices, as well as biologically inspired materials and systems. In particular, the experimental techniques used to develop this understanding, and the fundamental scientific and technical insight that has been obtained into various aspects of processing/microstructure/property/structure/ performance relationships in these devices, materials, and systems will be reviewed.

Mechanical forces play an important role not only in synthetic materials and systems, but also in the natural world. They have guided many aspects of the structural development that is found in nature, primarily with respect to their effect on non-mechanical processes. For example, a plant must grow laterally in a manner that

the volume reduces the stress due to bending from wind loading, but it must also distribute a sufficient amount of this volume axially in order to have sufficient surface area to gather sunlight necessary to support photosynthetic chemical processes within that volume. The need to understand the tradeoffs that natural systems make has provided an enormous opportunity to the experimental mechanics community to modify techniques normally used on synthetic materials to the study of these natural systems. This knowledge of the relationship between natural structure and functionality can then be used to guide the way natural materials are used in a diverse array of applications, ranging from implantable biomedical devices to new materials and systems that are referred to as *Biomimetic* or *Biologically inspired/Biologically inspired* depending on whether the concepts originating from nature used in their developed are either copied (biomimetic) or serve as inspiration (biologically inspired) [32.1].

One notable characteristic of natural materials that is of interest to the experimental mechanics community is the way in which the structure functions and adapts at a wide range of length scales. Much of the interaction that biological materials and structures experience occurs through mechanical contact. Therefore, to develop biologically inspired materials and systems it is necessary to process synthetic materials to adopt the attributes of biological materials and structures that affect mechanical behavior through similar microstructure/ property/structure/performance relationships. This has also been synergistic with the development of implantable biomedical devices, where it is essential to process materials with appropriate microstructure/ property relationships that are biocompatible (*biomaterials*), and deploy these biomaterials in devices with structure/ performance relationships that are appropriate for mechanically interacting with the body of an animal/human.

32.1 Overview

32.1.1 History

Engineering can be defined as the effort by human beings to design and create materials and systems (i. e., technology) in order to improve their lives. Science can be defined as the effort by human beings to explain natural phenomena. To develop new technologies, Engineers employ scientific principles to manipulate raw materials that are found within their natural environment. The first engineered structures discovered by archaeologists were crude tools, utensils, and buildings consisting of natural materials such as stone, wood, and clay that were initially not refined [32.2]. Of these materials, only clay existed in both a liquid and solid phase. Therefore, it was the first material to be refined from its natural state by the addition of biological materials, such as straw, in order to form the first synthetic material: *a natural composite*.

With the advent of the Bronze and Iron Ages, synthetic materials refined from inorganic natural materials became the focus for engineering many new technologies. Metallurgy became the foundation for the science and engineering of materials. Although stone, wood, and clay were still employed by engineers, they were often used without additional refinement. The need for a deeper understanding of these natural materials was obviated by the ability to utilize extra quantities of material in order to create more conservative designs. This remained the case until the early 20th century, when the advent of synthetic polymers began to herald a new age in materials science and engineering: *the age of plastics*.

The development of plastics coincided with scientific advances in the characterization of materials through the use of microscopes. Characterizing the microstructure of materials enabled engineers to develop new theories on the physical behavior of materials, and to use these theories to alter the way these materials were refined in order to improve their physical properties. Engineers returned to the concept of natural composite materials, and began to extend them to new synthetic composites formed from metals, ceramics, and plastics. The development of these composite materials was critical to the greatest engineering feat to date: *landing a human being on the moon*. Concurrently, these new materials were also having an impact on another important scientific/engineering feat that has transformed the lives of human beings: *modern medicine*.

Much of the later half of the 20th century was focused on further development of advanced materials, with substantial motivation from the needs of modern medicine. One limitation of materials, which often restrict their applications in medicine, is the deterioration of their structural integrity due to damage accumula-

tion. This problem, which exists to some extent in all synthetic materials, has resulted in the extensive development of advanced techniques for the measurement and characterization of damage that have evolved from Non-destructive Evaluation and Fracture Mechanics to quantify and understand damage mechanisms in biomaterials. On the other hand, natural materials and structures have evolved to actively minimize the impact of damage accumulation through processes such as self-healing and remodeling, which have been widely considered impractical in synthetic materials until only recently [32.3, 4]. Adapting these natural processes to synthetic materials has led to a new field of materials and systems research near the end of the 20th century known as *biologically inspired materials and systems*.

The development of biologically inspired materials and systems has resulted from a new scientific and technical focus in materials research that has been placed on microtechnology and nanotechnology. This has been predicated by new characterization technologies, such as Atomic Force Microscopy that enabled visualization of natural phenomena down to the atomistic scale, as well as microtensile testers and nanoindenters for microscale and nanoscale mechanical characterization. At these size scales, new phenomena were revealed that presented scientists and engineers with a novel approach to engineering advanced materials and systems, such as biocompatible materials (*biomaterials*) for biomedical applications. Piehler has identified biomaterials as the dominant area of focus for materials research in the 21st century, one that will fuel significant economic expansion through the development of new implantable biomedical devices and biologically inspired materials and systems [32.5]. This new scientific and technical focus in materials research has provided a wonderful opportunity for experimental mechanics to play an important role in the understanding of processing/microstructure/property/structure/ performance relationships in implantable biomedical devices and biologically inspired materials and systems that are necessary for improving the development and deployment of both conventional synthetic materials and biomaterials.

Although there are many opportunities for experimental mechanics in developing implantable biomedical devices, the majority of research has primarily focused on three technologies: *prosthetics*, *biomechanical fixation*, and *deployable stents*. Similarly, the development of biologically inspired materials and systems has focused on four technologies: *functionally graded materials*, *self-healing polymer/polymer composites*, *active materials and structure*, and *biologically inspired ceramics*. The scientific and technical relationship between these implantable biomedical devices and biologically inspired materials and devices that are the current focus for the application of experimental mechanics can be seen in Fig. 32.1. It is the relationship that will serve as a basis for discussing the experimental techniques that have been developed for addressing the challenges of applying experimental mechanics to understanding processing/microstructure/ property/structure/performance relationships in the implantable biomedical devices and biologically inspired materials and system to be reviewed in this chapter.

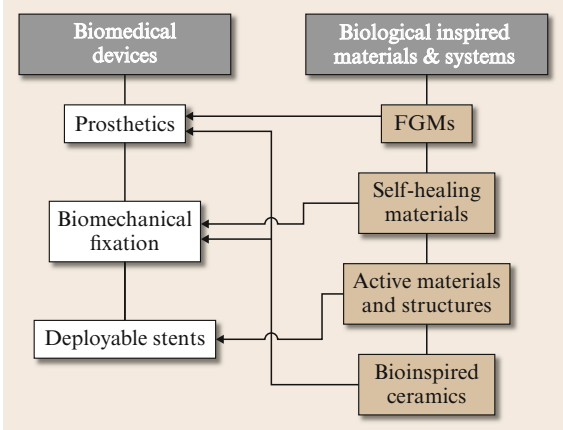

Fig. 32.1 Relationship of implantable biomedical devices to biologically inspired materials and systems that are the current focus of applications for experimental mechanics

32.1.2 Experimental Mechanics Challenges

Materials and Structural Characteristics

For implantable biomedical devices, the applications are typically in vivo (i. e., within a living organism). As such, biocompatibility plays an important role in the use of materials for any implantable biomedical device. Consequently, biomaterials are often selected whose primary characteristic are chemical and mechanical properties that are compatible with the biological function of the organism. Unfortunately, biomaterials do not often possess microstructures with chemical or mechanical characteristics that are optimal for biomedical applications. Therefore, designing new materials with the constraint of biocompatibility is a challenge that requires understanding the microstructure/property relationships in biomaterials using novel experimental

mechanics techniques, such as full-field deformation measurement at multiple length scales.

In addition to biocompatibility of physical properties, it is also necessary for implantable biomedical devices to have structural compatibility for optimal performance, which requires geometric complexity. Therefore, new testing simulators have also been developed to accommodate these geometric complex structures in order to determine the structure/performance relationship for these devices. For example, For example, the rigidity of deployable stents will determine the forces that they apply to the plaque as they are being deployed, as well as the arterial walls with which they will have contact once they are fully deployed and the plaque removed from the arterial walls. Therefore, point or circular loads are applied to the stents to ascertain the diametrical response in rubber tubes that model the mechanical response of the arterial walls.

Similar to biomaterials for implantable biomedical devices, biologically inspired materials and systems also require an understanding of microstructure/property relationships at multiple length scales in order to duplicate the function of natural counterparts. For example, the fracture toughness of the rigid biological ceramic composite microstructure found in mollusk shells (*nacre*) is superior to synthetic monolithic ceramic microstructures. Despite the inherently brittle nature of the ceramic phase (*aragonite*) that comprises 95% of the nacre, the microstructure formed when the aragonite is assembled in a *mortar-and-brick* composite microstructure using a natural polymer binder comprising the remaining 5% of the nacre enables the material to provide excellent resistance to crack initiation and propagation. The experimental mechanics community has been able to quantify the localized properties of the nacre microstructure and the transfer of load between the aragonite phases in order to understand the impact of the microstructure on crack growth mechanisms that has enabled the development of biologically inspired materials with similar improvements in fracture toughness.

Natural materials and systems have also evolved to balance multiple functional needs. For example, the microstructures of bone and bamboo are not the same throughout the structure. Instead, it gradually varies depending on location in order to balance the distribution of load to resist failure, minimize the local mass to optimize the volume and reduce energy consumption, and to optimize transport and storage processes necessary to support growth and repair of the structure. Similar multifunctional benefits are being realized by grading the microstructure in synthetic materials to create biologically inspired functionally graded materials. Therefore, the relationship between the gradient in microstructure/properties and various facets of the multifunctional performance has required the development of new testing and analysis techniques for experimental mechanics.

Experimental Mechanics Issues Based on the Microstructure, Property, Structure, Performance Characteristics

It should now be evident that implantable biomedical devices, as well as biologically inspired materials and systems, have posed a variety of challenges to the experimental mechanics community. Many of these challenges arise from understanding the mechanical behavior of these devices, materials, and systems in the context of their compatibility with and performance in biological environments (in vivo) rather than just external to these environments (in vitro). It is also clear that they are far more complex than the solid materials and structures that the mechanics community has conventionally focused on, since their physical properties are derived from structural interaction at multiple length scales through a wide variation in local physical properties in order to satisfy multiple functional requirements. Thus, the challenges that must be addressed can be generalized:

- wide range of physical properties,
- wide range of length scales (*hierarchical*),
- multifunctionality,
- environmental constraints (in vivo).

Each of these will now be discussed in more detail.

Wide Range of Physical Properties. The development of implantable biomedical devices involves working with materials as soft as an elastomer to as rigid as a ceramic. The design of implantable biomedical devices, such as implants using hard metals, ceramics, and composites for support and soft adhesives for fixation, often require understanding the impact of the differences of these material properties on the performance of the device, such as failure. To elucidate on the characteristics of these wide property variations, experimental mechanics approaches based on full-field deformation measurement techniques and micro/nanoscale indentation techniques are being developed.

In nature, physical properties can also vary dramatically within the microstructure of a given material. For example, the seashell can have very hard arago-

nite phases surrounded by a very compliant organic material [32.6]. Using standard experimental mechanics techniques, such as indentation, to characterize the wide range of properties, such as hardness, in these materials typically requires a transducer with high load resolution over a large load range. Additionally, understanding the deformation response of these types of materials can require techniques that have very high strain and displacement resolution over very large strain and displacement ranges, while having the additional capability of resolving discontinuities that result across the phase boundaries. These challenges are identical to those that are faced when engineering biologically inspired ceramics.

Wide Range of Length Scales (Hierarchical). The unique mechanical behavior of biological materials and systems and their potential biologically inspired counterparts can be attributed to microstructures that bridge many length scales (*hierarchical*) [32.8]. In calcified tissues, like bone, these length scales start with the collagen molecule at the nanoscale and organize into the haversian system at the microscale (Fig. 32.2) [32.7]. In bone tissues such as teeth, there are additional microscale structures defined by a dentin-enamel junction that serves as a natural barrier to cyclic fatigue crack growth.

Using the seashell as a more detailed example of the relationship of these length scales to mechanical behavior, there are biopolymer phases at the nanoscale whose mechanical behavior is critical to providing a crack-bridging mechanism that substantially increase resistance to crack propagation through formation of a nanovoid between the biopolymer ligaments whose growth is limited by a critical nanoscale crack tip opening displacement (CTOD) (Fig. 32.3) [32.6, 9]. Locally, on the nanoscale, the amount of resistance needs to be characterized by the surface energy and strength of the compliant organic material that are holding the aragonite phases together, as well as adhesive strength that will prevent a less energy intensive debonding process to occur. At the same time, the morphology of the aragonite phases forms a lamellar *brick and mortar* microstructure that distributes stress on the microscale that needs to be characterized to determine the relationship between the energy that can be released during formation of nanovoids and the direction of crack propagation in order to reconcile this behavior with the microscale distribution of resistance that is determined by the platelet morphology and binder thickness of the biopolymer. This behavior must then be reconciled with variations in the size of the phases on a mesoscale that can create further resistance to macroscale crack propagation through a gradient in resistance and a redistribution of stresses near the crack tip to guide energy away from the most favorable directions of crack propagation through the thickness towards the surface where a number of fine nanovoids are formed to dissipate energy. These spatial measurement challenges are identical to those faced when engineering biologically inspired FGMs.

Multifunctionality. The hierarchical nature of natural structures lead to an assembly of parts that make subsystems and collections of subsystems performing a variety of functions (i. e., *multifunctional*) [32.10]. Examples

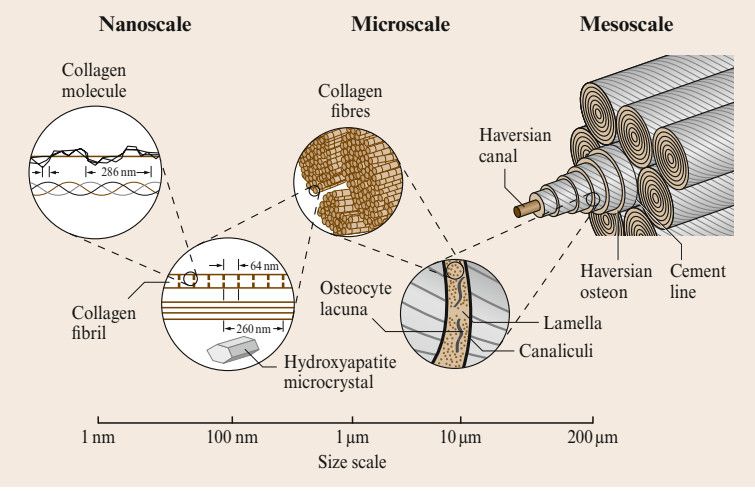

Fig. 32.2 Hierarchical structure in bone [32.7]

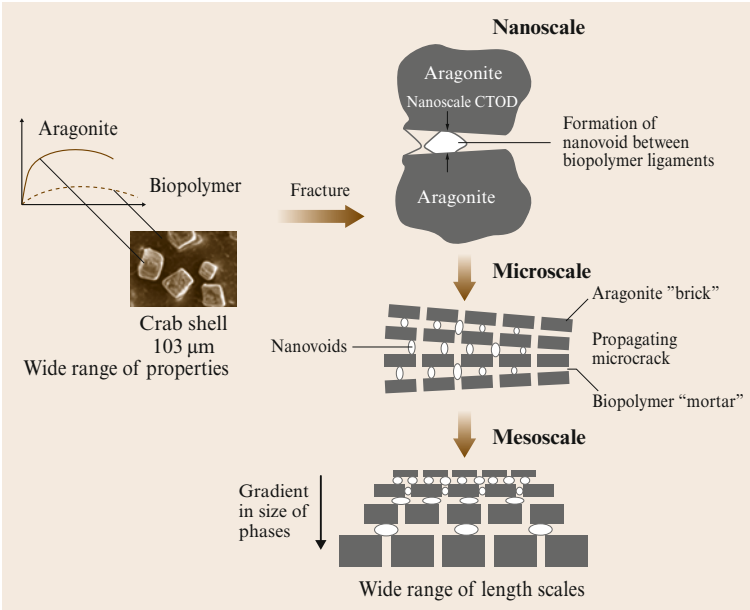

Fig. 32.3 Wide range of properties and length scales associated with fracture processes in seashells

of multifunctionality in natural structures can be seen in Table 32.1. The salient characteristics of these structures often dictate structural performance. For example, some natural objects are large and other objects are small, but the size of the structure of a particular animal does not necessarily have an influence on its speed. The smallest animal is not the fastest or slowest, and neither is the largest animal. The fastest mammal is the cheetah, whose size is somewhere in the middle. Having the right combination of skeletal structure for mechanical support and muscle structure for power are the largest contributors to the speed of an animal. This is why the average ant, even though it has the same muscle strength as a human is able to carry anywhere from five to over twenty times its body weight because the muscle force scales with cross-sectional area while the structural mass scales with the volume (i. e., relative lifting force = muscle strength × cross-sectional area/(density × volume × gravitational force)).

The multifunctional structural principles found in nature can lead to new designs that are not just much more efficient, lightweight, or responsive than traditional designs. They can also lead to multifunctional

Table 32.1 Multifunctionality in natural structures

Biological material	Multifunctionality
Bone	Lightweight structural support for body
	Storage of inorganic salts
	Blood cell and bone formation
Chitin-based exoskeleton in arthropods	Attachment for muscles
	Environmental protection
	Water barrier
Sea spicules	Light transmission
	Structural support
Tree trunks and roots	Mechanical Anchoring
	Nutrient transport

structures that are capable of harvesting energy from the environment, supporting mechanical and thermal loads from the environment, transporting fluids, etc. For example, the hierarchical-structure of wood enables the trunk and roots of a tree to transport nutrients in solution and gather solar energy to support growth and repair processes while concurrently serving as the mechanical support (Fig. 32.4) [32.11]. For the trunk, this support allows for the distribution of mass through complex geometries during the growth of the tree that increases the surface area of the tree both vertically and horizontally needed for gathering solar energy used in the chemical process known as photosynthesis to drive the production of new organic matter required for the growth. For the roots, complex geometries increase surface area to allow for mechanical anchoring of the tree in soils while increasing the absorption of water and nutrients from the soil as well. Thus, new power generation materials and systems can be designed with hierarchical-structures for optimal power generation and external anchoring to structures.

Another classic example of a multifunctional natural structure is bone. In a relative sense, bone can be dense (*cortical, aka compact*) and harder, primarily made up of collagen and calcium phosphate. However, bone can also contain pores (*cancellous, aka spongy or trabecular*) that reduce stiffness in a power law relationship with decreasing mass (e.g., Young's modulus $= 1240 \times \text{scaling factor} \times \text{density}^{1.8}$) that has also been correlated with hardness, while aiding in storage of inorganic salts and marrow for the formation of blood cells (Fig. 32.5) [32.12, 13]. The result is a lightweight microstructure with a balance of stiffness and strength for structural support of a body, and whose mechanical behavior creates low mechanical signals of 2000–3000 microstrain that can increase the rate of bone formation by 2.1-fold and the mineralizing surface by 2.4-fold in cancellous (porous) bone to strengthen it [32.14]. By using experimental mechanics to understand the mechanical principles of natural structures, new biologically inspired multifunctional materials and systems can be designed for optimum performance in a variety of applications.

Environmental Constraints (In Vivo). The environments (e.g., the human body) under which implantable biomedical devices, as well as some biologically inspired materials and systems, operate present a unique challenge to the experimental mechanics community in developing testing techniques that can be conducted directly in these environments or in simulated environments for these materials and systems. For example, an implantable biomedical device for a soft collageneous tissue application, like tissue scaffolding, can be extremely permeable to water [32.15]. This degree of permeability will have an affect on the transport of oxy-

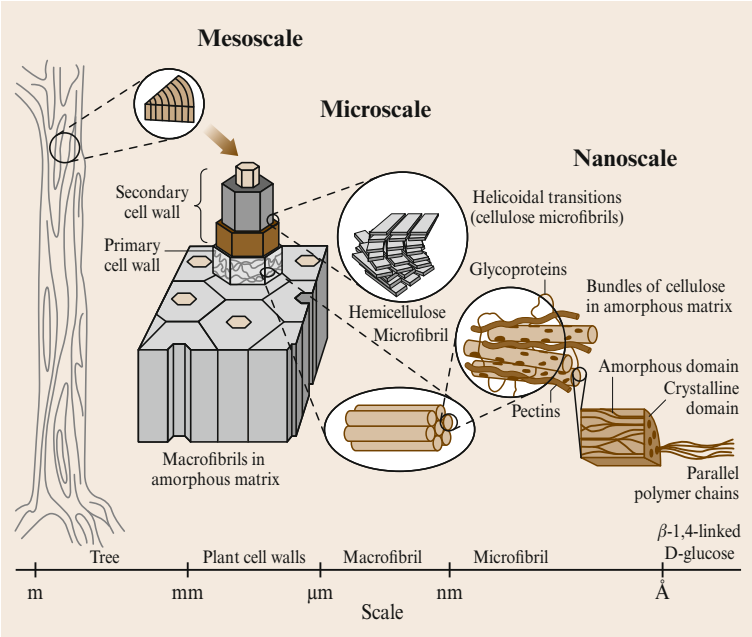

Fig. 32.4 Hierarchical structure of woode that leads to multifunctional behavior of tree trunk and root system [32.11]

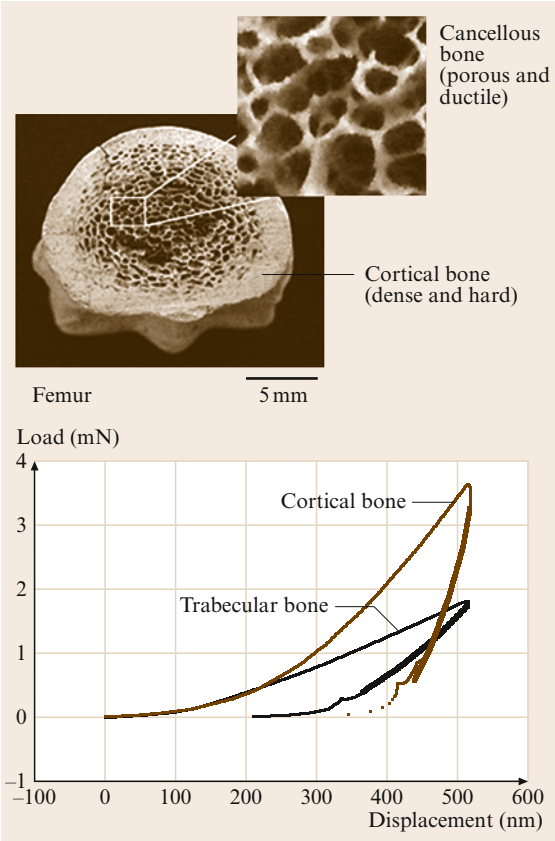

Fig. 32.5 Cross-section of a bone revealing porous and ductile microstructure (*cancellous*) used for storage of inorganic salts and marrow for formation of red blood cells and dense and hard microstructure (*cortical*). Also seen are typical nanoindentation measurements indicating a 4-fold difference in stiffness [32.13]

gen and nutrients to the soft tissue interacting with the device. Removing water begins a degradation process in the interaction that can rapidly affect the functionality of the device. However, the biological processes are electrically-driven as well, so removing the device from its living organism will also begin a degradation process in the interaction, albeit the rate of decay is much slower than when water is removed.

The degradation process can also be affected by different environmental factors associated with the applications of these devices, materials, and systems. For example, increasing the Ph and temperature associated with implanting implantable biomedical devices, such as polymer composites, in the human body can increase the rate of degradation for the implant materials [32.16].

Therefore, it is preferred for the experimentalist to conduct experiments on these materials in in vivo or simulated in vivo environments, rather than in vitro. The simulated in vivo environments are the easiest since they do not require surgical procedures and are more accessible to common experimental techniques employed by the experimental mechanics community; however they still present challenges when working with certain techniques, like optical methods whose noise level is very sensitive to the changes in optical characteristics presented by the aqueous and thermal conditions of these environments. For in vivo environments, it would be possible to utilize these techniques if accessibility issues can be overcome. When it is not possible to conduct experiments in either of these environments, it will be necessary to assess how the absence of an in vivo environment affects mechanical response in order to develop appropriate in vitro test methods.

Approaches for Addressing Issues

To address the experimental challenges posed by implantable biomedical devices and biologically inspired materials and systems, there are several general approaches being employed.

1. Characterization of mechanical properties and performance using new/modified testing techniques and simulators (e.g., *modified screw pullout tests, nano/microindentation*).
2. Modeling of the devices, materials and systems to analyze experimental results using new mechanics principles (e.g., *cohesive zone models for interfacial crack growth*).
3. Development of model structures and experiments to ascertain mechanical performance of devices and structures (e.g., *simulators for prosthetics, model arterial structures for deployable stents, model metal-ceramic composites for functionally graded materials*).
4. Characterization of the processing/ microstructure/property/structure/ performance relationships (e.g., *determining temperatures to process self-reinforced polymer composites for biomechanical fixation*).

These approaches represent application of the full range of testing techniques and mechanics principles employed by the experimental mechanics community. It is the intention of this chapter to provide details of the application of these approaches.

32.2 Implantable Biomedical Devices

With the advancement of medical knowledge, the development of implantable biomedical devices has also advanced. In some cases, advancements in the knowledge of the mechanics of materials and structures have led to the development of new medical procedures. For example, the ability to create metals with deployable geometries transformed cardiac surgery by enabling the developed of angioplasty for removing plaque buildup in arteries by insertion of a thin metallic wire that expands to break up the plaque and debond it from the arterial wall. Advances in materials and manufacturing technologies have also had a tremendous impact on replacing failed biological structures, such as prosthetics that have evolved from crude metal dentures that are removable, to more modern ceramic implants that are biocompatible and can be permanently fixated. Many applications of these implantable biomedical devices have required development and application of experimental mechanics techniques to facilitate their development, the details of which will now be discussed.

32.2.1 Applications

The primary application of implantable biomedical devices is for insertion into human beings or animals. For experimental mechanics, there are three primary classes of implantable biomedical devices of particular interest:

1. *prosthetics*,
2. *biomechanical fixation*, and
3. *deployable stents.*

While the range of applications is limited, the ways they are utilized inside of the human being are varied. For example, prosthetics such as artificial bone and teeth can be used in place of decaying natural counterparts. The placement of the prosthetics can also require new biomechanical fixation technologies such as biocompatible adhesives.

While it is sometimes necessary to replace decaying natural components of the human body, it is also possible to enhance their repair. For example, the removal of plaque buildups in arteries through surgical procedures such as angioplasty requires devices known as deployable stents that can be inserted into the narrowed artery and then expanded with pressures that are sufficient to remove the plaque buildup without damaging the arterial tissue. Furthermore, the stent can be left behind to prevent further plaque buildup using drug-eluting coatings to prevent re-stenosis.

The joining of hard materials associated with prosthetics with soft biological tissues, such as bone, has also been enhanced through the development of new biocompatible adhesives, such as injectable biodegradable polymers for tissue scaffolding [32.17]. Most biocompatible adhesives have the traditional mechanics issues associated with their development involving interfacial properties and stresses associated with dissimilar materials. For new tissue scaffolding materials, the adhesion concept involves using porous microstructures that have features that are compatible with the growth characteristics of new cells to create adhesion. By providing a flexible pathway for cell growth, adhesion is achieved through rapid in vivo formation of a more compatible interface with interfacial properties that are a combination of the hard and soft materials leading to improved performance. For example, tissue scaffolding has been employed extensively in oral and maxillofacial surgery to aid in regeneration of tissue around dental implants.

32.2.2 Brief Description of Examples

Prosthetics

One of the most important applications of mechanics in bioengineering has been in understanding the mechanical response of materials and structures that are being developed for *prostethetics*. Prosthetics are artificial structures and devices that are used to replace natural structures and systems in living organisms. They can be used to replace damaged or missing body parts, or to enhance the performance of the living organism. They are often deployed as bone implants, with classic examples being ball and socket components for hip replacements and dental implants for tooth restoration.

From a mechanics perspective, the development of prosthetics requires understanding the stresses that evolve within the natural body part due to the loading that that part will experience during use. This knowledge can be used to design and/or select biocompatible materials that will be sufficiently fracture-, fatigue-, and wear-resistant at the corresponding in vivo stress states and environment. One mechanics issue that has been extremely complex has been to develop biocompatible joining technologies for prosthetics that meet the aforementioned mechanical requirements.

Biomechanical Fixation

A critical issue for implantable biomedical devices such as prosthetics is the ability to fixate the device to a natural structure through adhesion that is compatible with the natural structure (*biomechanical fixation*) in order for it to properly function. The mechanics issues involved with biomechanical fixation fall into the more traditional category of *interfacial mechanics*. For biocompatible fixation, the unique issue is balancing the biocompatibility characteristics of the fixation device with the mechanical characteristics necessary for optimizing the interfacial adhesion between natural and synthetic materials.

A new biomechanical fixation device whose development is being guided through a more intimate understanding of mechanics is *tissue scaffolding*. To develop tissue scaffolding, there is a need to understand how the scaffolding can be designed to rapidly promote the growth of mechanically-sound tissue. The mechanical performance of this tissue is studied in an area of bioengineering known as *tissue mechanics*. The characterization of the evolution of the mechanical behavior of the tissue scaffold and new tissue has presented unique challenges for the experimental mechanics community to resolve. The porous nature of the scaffolding makes it difficult to measure the loads that are distributed within it to optimize its design, while it is also necessary to characterize in vivo the evolution of mechanical properties for the combined scaffold/new tissue system. Also, the adhesion of cells to the scaffold material, and its impact on the growth of new tissue, is also an area of active research for the development of implantable biomedical devices in general.

Deployable Stents

A very popular implantable biomedical device where mechanics has played an important role is the development of arterial stents. Arterial stents are small, deployable metal structures with complex geometries that enable them to expand after they are inserted into an artery clogged by plaque build-up in order to break up the plaque and restore the proper flow of blood. The stresses experienced by the stent when it is expanded must be characterized in order guide the design the geometry of the stent to mitigate stress concentrations, and to select appropriate biocompatible materials that can withstand the corresponding stress state.

Originally, the pressures that induced the expansion of the metal stents were generated by a balloon in the center of the collapsed stent that could be pressurized after insertion. However, more advanced materials have been recently investigated for deployable stents that are self-deployable. The mechanics of these materials, known as shape memory alloys (SMAs), are vastly different than the original metals. Upon heating, the SMAs can experience extremely large deformations associated with martensitic phase transformations. Because of these large thermally-induced deformations, SMAs are considered *active* materials, while the original metals are considered *passive*. Therefore, they just need to be designed to deploy and achieve the desired pressures within a very narrow range of temperatures that will not harm the arterial tissue in the human body.

32.2.3 Prosthetics

Classification of Prosthetics

Prosthetics generally fall into two categories:

1. Orthopaedic (bone and joints),
2. Dental (teeth).

For orthopaedic applications, the structures are artificial load-bearing members designed to replace failed natural counterparts. A critical mechanics issue is the joining of these structures to the biological system, which often entails additional mechanics issues involving the joints themselves. In some cases, natural joints may fail and be replaced with artificial ones that may require modifying the natural structures that are being connected to the joint (i. e., biomechanical fixation), where pull-out forces associated with interfacial strength need to be characterized and will be discussed in more detail in terms of biomechanical fixation in Sect. 32.2.4. Finding materials and joining concepts that are amenable to the cyclic transfer of loads across the joints is usually the most important experimental mechanics consideration in developing orthopaedic prosthetics.

Dental prosthetics differ from general orthopaedic prosthetics in that they employ a dental implant to serve as a *root* for a permanent tooth restoration known as the *crown*. Typically, these implants are designed to integrate with existing natural structure (e.g., jawbone), without need for a periodontal ligament. The interfacial mechanics issues associated with this design are typically the most important to understand, and will be discussed in more detail in terms of biomechanical fixation in Sect. 32.2.4. Additionally, the mechanical performance of the tooth restoration is much different than orthopaedic prostheses in that the surface is subjected to compressive point loads rather than distributed pressure loads. As such, the development of hard ma-

terials that can bear these loads without failing while being amenable to manufacturing technologies required to create complex shapes from these materials is the most important consideration.

Types of Prosthetic Materials

Prosthetic materials are generally made from:

1. metals,
2. ceramics,
3. polymers, and
4. composites.

The primary materials used in prosthetic applications are as follows (Table 32.2) [32.19, 20]:

Because of the time involved in approving new materials for biomedical applications, many of the materials in Table 32.2 have been used for decades and serve as the basis for much of the current research into the mechanics of not only prosthetics, but of implantable biomedical devices in general. The specific selection of an appropriate prosthetic material for a particular implant application will require a balance of mechanical properties, inert in vivo response, and availability [32.21].

Experimental Mechanics Characterization Techniques

There are three techniques generally used for characterizing prosthetics and their associated materials

1. Standard tension/compression testing.
2. Load frames with multiple degrees of freedom (DOF).
3. Micro and nanoindentation.

Standard tension/compression testing can be conducted to elucidate on the properties of biomaterials, but they need to be modified to assess performance.

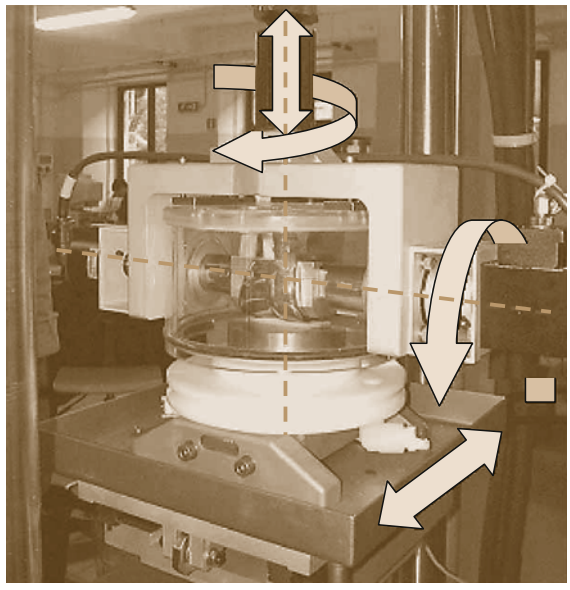

Fig. 32.6 Four DOF knee simulator from MTS (MTS Systems, Minneapolis) [32.18]

Load frames with multiple DOF are often employed to simulate the cyclic loading conditions that a prosthetic will be subjected to when it is deployed to determine its performance (Fig. 32.6) [32.18]. There have also been unique test fixtures designed to simulate natural boundary conditions, such as a wear simulator for shoulder prostheses, a new glenoid component test fixture for force controlled fatigue testing through the movement of the femoral head of shoulder prostheses, and a transverse plane shear test fixture for knee prosthetics [32.22–24]. To further elucidate on the details of the processing/structure/property relationship in biomaterials at the micro- and nanoscales, indentation and scratch tests can be employed.

Table 32.2 Classification of prosthetic materials

Class of prosthetic material	Type
Metal	Ti6Al4V, CrNiMo stainless steel, NiTi, CoCrMo, gold
Ceramic	Hydroxyapatite, alumina, zirconia, tricalcium phosphate, glass-ceramic, pyrolytic carbon
Polymer	Silicone, PMMA, polyurethane, polyethylene, acrylic, hydrogel, nylon, polypropylene, polylactic acid, polycaprolactone, polyglycolides, polydioxanone, trimethylene carbonate, polyorthoester
Composite	Carbon fiber/epoxy, hydroxyapatite/PMMA, bioactive glass/PMMA, self-reinforced PMMA, self-reinforced PLA

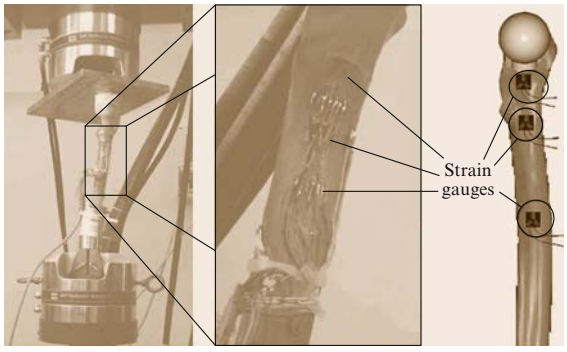

Fig. 32.7 Strain gages used to determine strain distribution in femur after implantation of a hip prostheses [32.18]

Diagnostic techniques for mechanical testing of prosthetics also cover a wide range of experimental mechanics methods. For harder prosthetic materials, standard strain gage technology will be employed to understand the strain distribution in the prosthetic and/or the natural structure it is joined with, and can be used in vivo (Fig. 32.7) [32.25]. Uniaxial strain gages are also commonly used to characterize strains on the hard biological materials, such as bone, to which prosthetics are fixated, although triaxial rosettes are sometimes needed for more complete information unless the principal directions are along the axis of the bone, as in the case of stress shielding for a fixator plate on a femur [32.26]. For softer materials, video extensometry techniques like DIC can be employed [32.27]. Maximum pressure distributions associated with load transfer within articulating prosthetic joints are obtained using pressure-sensitive films with microbubbles that are designed to rupture and release liquids that former darker stains as the maximum pressure increases locally [32.28]. Photoelasticity has also been popular for characterizing the interfacial stresses generated at the interface of the prosthetic and the natural structure [32.29–31].

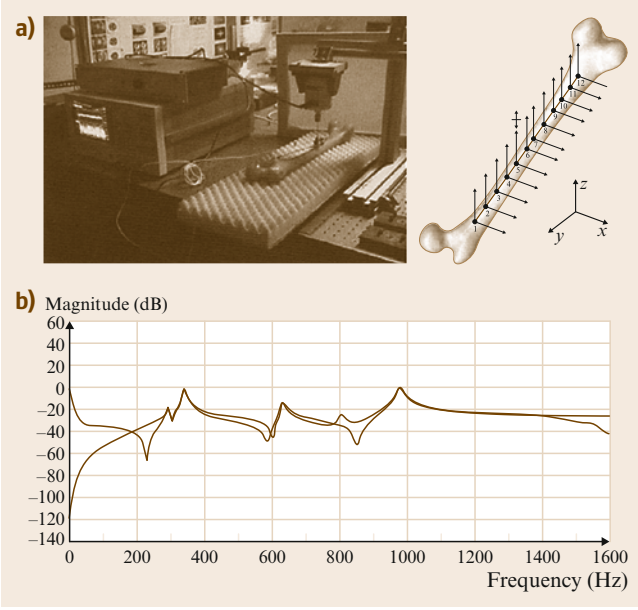

Fig. 32.8 **(a)** Modal testing system for composite femur, and **(b)** comparison of experimental and numerical modal response indicating good correlation [32.32]

Microstructure/Property/Structure/Performance Relationships

Modal Response. The development of prosthetics using advanced materials, such as synthetic composites, requires modeling tools to relate that need experimental validation for microstructure/property/structure/performance relationships. It has been proposed that the modal response of the entire prosthetic structure can provide superior verification for these models over strain gage measurements since modal parameters (frequency, damping ratio, and mode shape) are intrinsic properties of the whole structure (Fig. 32.8) [32.32]. Modal analysis can also be used to characterize performance by detecting early loosening of prosthetics due to interfacial failure as the modal response transitions from linear to non-linear.

Interfacial Damage. Since prostheses can be modular (e.g., head-stem in hip implant), internal interface integrity plays an important role in its performance, the experimental characterization of the structure and properties of the interface are important for understanding failure [32.33]. In particular, characteristics of the modular interface, such as surface roughness, can change due to fretting and corrosion from wear at the interface causing inflammation [32.34]. As a result, the cyclic contact loading of modular orthopaedic prostheses can decrease the size of surface asperities after a few cycles, but corrosion can increase the size depending on the magnitude of the contact load. Using finite element analysis, significant residual stresses have been predicted at the trough of the asperities due to elastic interaction of plastically deforming peaks. Thus, a damage mechanism was proposed using the experimental and numerical results that explained the evolution of the measured surface roughness by incorporating the residual-stress devel-

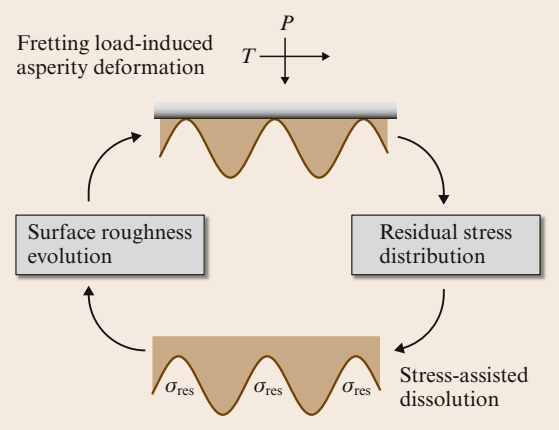

Fig. 32.9 Experimental and numerically-derived damage mechanism at the interface of orthopaedic implants resulting from residual stresses due to contact loading and stress-assisted dissolution [32.33]

opment governed by the yield stress of the implant material normalized by the contact stress, along with the stress-assisted dissolution of the implant material (Fig. 32.9).

Surface Damage. The performance of dental implants also involves concentrated contact loading at the surface of the tooth restoration instead of distributed loading across the interface. This loading condition has been found to be accurately emulated using Hertzian indentation testing [32.36]. Using this testing technique, it has been determined that there are two failure modes of importance:

1. a fracture mode consisting of *Hertzian cone cracks that* develop outside of the contact zone due to brittle fracture in response to tensile stresses, and
2. a quasi-plastic deformation mode consisting of *microdamage* beneath the contact zone in response to shear stresses.

There is a progressive transition from the first to the second mode depending on the microstructural heterogeneity, and is manifested through non-linearity in the indentation stress-strain curve. Threshold loads for the initiation of fracture and deformation modes have been related to indenter radius in order to assess prospective tooth restoration materials with different microstructure/property relationships for various masticatory conditions.

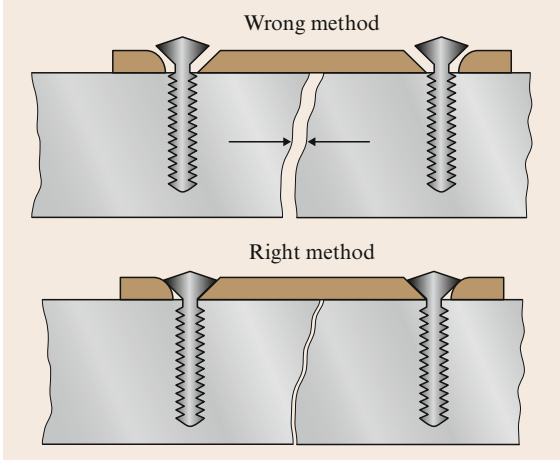

Fig. 32.10 Compression across fracture produced by rigid plate fixator [32.35]

Axial Flexibility. For temporary prosthetics, like a dynamic compression plate, the mechanics of the device are designed to produce compressive stress on a fracture to promote healing through membranous bone repair (Fig. 32.10). For these prosthetics to work properly, the fundamental principle is to keep the two pieces of fractured bone from moving relative to each other through rigid fixation. However, there are degrees of this motion that need to be constrained less than others, in particular the axial direction where the rigidity can cause stress shielding during the healing process that reduces the density of the remodeled bone. For example, a thin Ti6Al4V plate that has low stiffness in the axial and bending directions is not adequate, but a thin tubular plate with low stiffness in the axial direction but moderate stiffness in the bending and torsional directions produce superior mechanical and structural properties that do not degrade even after the application is prolonged for 9 months [32.35]. Composite materials would seem to possess the ideal microstructure to tailor the directional properties for this application. While promising, conventional carbon-fiber/polymer composites have been found to be brittle and degrade rapidly when exposed to body fluids [32.16, 37]. Alternatively, the use of bioresorbable screws with rigid plates have been found to potentially take advantage of the degradation behavior in reducing axial stiffness as the bone heals, while elastic inserts inside of the screw slots have also been found to structurally produce an *axially flexible plate* through the geometry of the slot and the stiffness of the elastic insert [32.38, 39].

32.2.4 Biomechanical Fixation

Types of Biomechanical Fixation

A critical issue for successful deployment of prosthetics has been the development and characterization of fixation concepts between the prosthetic and the natural structure with which it functions, typically bone. These concepts generally fall into two categories:

1. biocompatible adhesives,
2. mechanical fixation (screws, roots).

The biocompatible adhesives can take many forms, and are most advantageous since they do not require any modification of natural structures. For example, the most common biocompatible adhesives are epoxies, such as DePuy CMW or Zimmer bone cements, which not only possess excellent biocompatibility, but can also develop strong adhesion with metal, ceramic, and biological materials [32.40]. However, they are inherently weak and can be exothermic when they set causing bone necrosis. Alternatively, soft tissue scaffolding materials have been developed to promote cell growth at the interface between prosthetics and natural structures, and are bioresorbable so they disappear once the desired growth has been achieved.

While adhesive materials can be ideal for most joining applications, it is not always possible to achieve the desired performance. Therefore, more conventional mechanical joining techniques are often employed. These include the use of screws as fixators, as well as rivets, staples, and other fastening devices.

Experimental Mechanics Characterization Techniques

The most common technique for characterizing biomechanical fixation performance is pull-out testing. In pull-out testing, the need is to identify the critical force required to exceed the interfacial strength and pull out a screw, nail or root that is being used to anchor a prostheses. The resulting pull-out response is a combination of the interfacial properties, in particular the shear strength, and the geometry of the interface. Alternatively, lap-shear testing can be used to isolate the shear strength component in the pull-out response.

Novel microscale testing techniques, such as microfabricated arrays of needles, are also being developed to better understand how the adhesion mechanics evolve between new biomechanical fixation devices, such as tissue scaffolding, and the biological materials, such as individual cells and tissues. However, the more conventional techniques of pull-out and lap-shear tests are still needed to assess the interfacial properties and performance.

Processing/Microstructure/Property/Structure/Performance Relationships

Cell–Substrate Interaction. Because biomechanical fixation is unique in that it involves bonding synthetic biomaterials with biological materials, such as individual cells and tissue, it is essential to develop an understanding of the fundamental physics governing the microstructure/property/structure/performance relationship involved in biomechanical fixation. In order to develop this relationship, novel experimental mechanics techniques have been developed, such as a microfabricated array of elastomeric microneedles that isolate mechanical force on deforming cells to study the mechanics of cell-substrate interaction and the effects of the topology, flexibility, and surface chemistry of the substrate on this interaction to establish the role that cells play in the interfacial strength (Fig. 32.11) [32.41]. By measuring the deflection of the posts δ the contractile forces F in cells can be quantified and related to the cell morphology and focal adhesions using the following standard bending equation

$$F = \left(\frac{3EI}{L^3}\right)\delta\,, \tag{32.1}$$

where E, I, and L are the Young's modulus, moment of inertia and length of the posts.

Characterization of the scaffold itself usually involves compression testing, and analyzing the results for initial stiffness, collapse stress, and densification using a standard cellular solid model with an open-cell tetrakaidecahedron microstructure to describe the scaffold mechanics [32.42]. The collapse stress σ_{cl} depends on the initial relative density of the scaffold ρ^*/ρ_{s} and the yield stress of the scaffold material σ_{y} as follows:

$$\frac{\sigma_{\mathrm{cl}}}{\sigma_{\mathrm{y}}} = 0.3\left(\frac{\rho^*}{\rho_{\mathrm{s}}}\right)^{\frac{3}{2}}\,. \tag{32.2}$$

The densification of the scaffold microstructure after the collapse stress is exceeded provides insight into the permeability changes that will occur according to the formula [32.43]

$$k = A'd^2\left(1-\frac{\rho^*}{\rho_{\mathrm{s}}}\right)^{\frac{3}{2}}\,, \tag{32.3}$$

where k is the permeability, A' is a dimensionless system constant, and d is the scaffold mean pore size, and the relative density changes are obtained from the

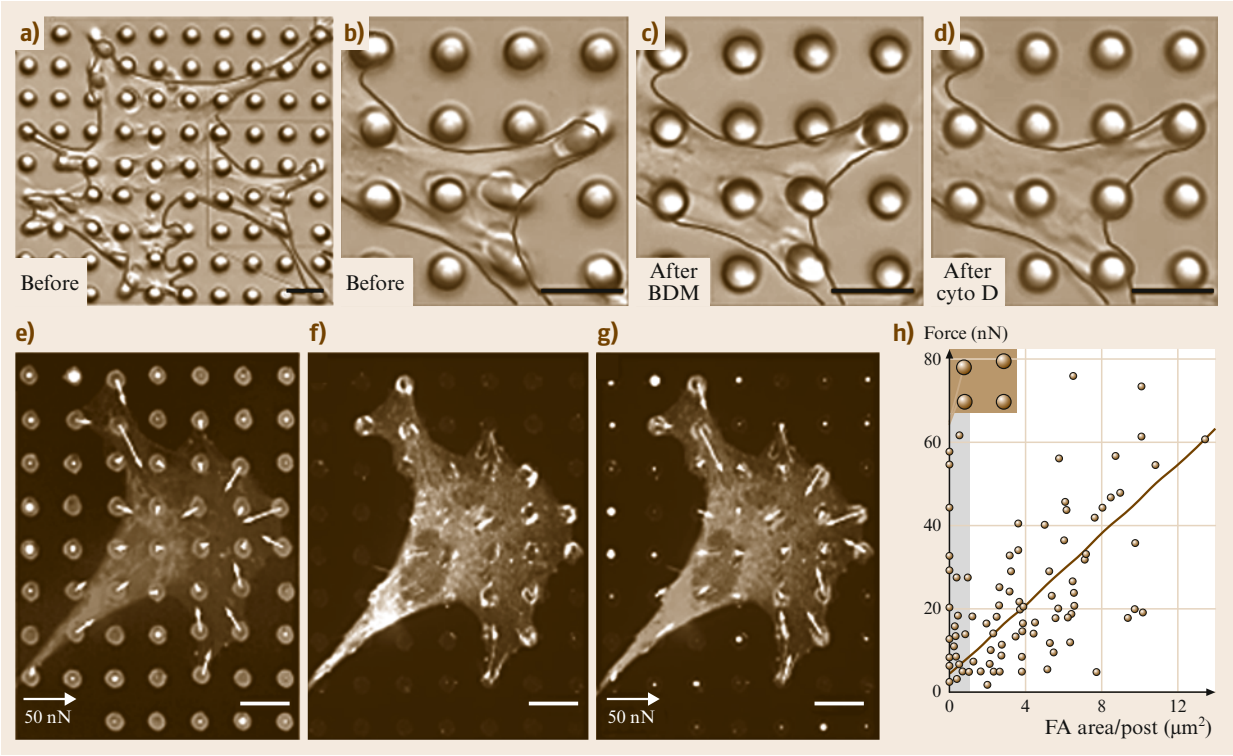

Fig. 32.11a–h Measurement of contractile forces in cells using a microfabricated array of elastomeric microneedles. The plot in the lower left (e) is the plot of the force generated on each post as a function of the total area of focal adhesion staining per post [32.41]

compression test. Changes in permeability are especially important since they affect scaffold performance by controlling the transport of nutrients and waste products within the structure that affect processes like cell adhesion and degradation [32.44].

Alternatively, the effects of scaffold properties and microstructural characteristics on cell adhesion can be qualitatively assessed by counting the number of cells that adhere to a scaffold, where more cells imply stronger interfacial adhesion [32.45]. From these studies, it was concluded that cell attachment and viability are primarily influenced by the specific surface area over a limited range of pore sizes (95.9–150.5 μm). As a general rule, it has been determined that adhesion rarely occurs when pore sizes are less than 20 μm or greater than 120 μm [32.46].

Pull-Out Strength and Failure Mechanisms. More traditional experimental mechanics testing techniques, such as pull-out tests, have been employed to characterize the interfacial strength and failure mechanisms for conventional prosthetic biomechanical fixation, such as bone fasteners. Using this test, it has been possible to characterize the ultimate pull-out forces for commercial orthopaedic self-tapping screws and machine screws [32.47]. Three failure mechanisms were identified depending on the level of pull-out force.

1. Bone-thread shear (low);
2. Bone splitting (intermediate);
3. Bone fragmentation (high).

As expected, the ultimate pull-out force was maximum at the mid-length of a bone and minimum at the distal end. With machine screws, the ultimate shear stress ranged from 3.5 to 5.8 ksi depending on the size of the screw and interference fit, which produced a maximum pull-out force at 50% difference between the diameter of the screw and the diameter of the drilled hole. Self-tapping screws produced about 50% greater shear strengths than machine screws.

More advanced techniques are also being developed for characterizing the mechanics associated with

biocompatible interfaces. Fiber bragg gratings (FBGs) combined with strain gages have been employed to characterize the strains associated with a fixator plate-bone interface [32.48]. The plates are amenable to strain gages, but the optical FBGs were determined to be superior to electrical resistance strain gages because it is less intrusive for biological materials and smaller. From tests on a synthetic femur specimen, It was determined that strain shielding is more pronounced in the distal region of the plate, which is 130% higher than for the intact femur and is 640% higher at the medial-distal side of the plated femur.

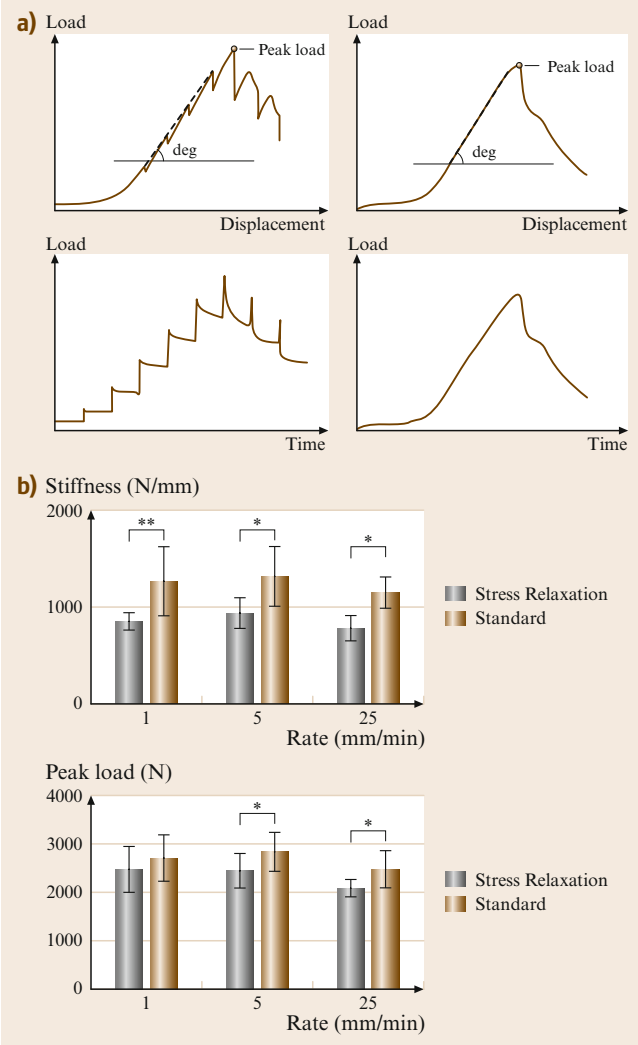

Fig. 32.12 **(a)** Comparison of new testing method for screw pullout with the standard testing method, and **(b)** comparison of stiffness and peak load measurements obtained from the two methods from lumbar spines [32.49]

Another complication in assessing the pull-out strength of screws is the complex mechanical behavior of bone. For example, the viscoelastic properties of bone can result in stress relaxation that can affect the pullout strength of orthopaedic screws (Fig. 32.12) [32.49]. The viscoelastic behavior is more likely to be present in vivo, and will significantly reduce the pullout strength and stiffness compared to in vitro tests where the behavior is more elastic. Thus, in vitro tests will overpredict mechanical performance due to the viscoelastic properties of bone in which the screws are implanted. To quantify these differences, a new screw pullout test was developed were the load was periodically held in order to quantify the relaxation effects and obtain the stiffness from the peaks of steps.

Stress at Geometrically Complex Interface. In contrast to the biomechanical fixation obtained with screws, the fixation of dental prosthetics requires understanding the stresses and displacements associated with a wedge-like root-periodontium interface that is geometrically complex [32.30]. In this investigation, a photoelastic specimen was prepared, and loading applied according to a model of the loading condition on a tooth associated with orthodontic appliances (Fig. 32.13). Although a 5° angle is more typical of the interface, a 15° angle was employed for a valid photoelastic test. The variation of interfacial stress was correlated to theoretical solutions obtained from elementary elasticity theory in Timoshenko and Goodier. Although these results are for a simplified 2-D model assuming elastic behavior, they can still be used to gain insight into the critical loads and displacements associated with the biomechanical fixation of a dental prosthetic. Recently, more detailed nonlinear two-dimensional interface element for finite element analyses of the bone-dental prosthetic interface have also been developed and validated using photoelasticity [32.31].

Processing/Microstructure/Property Relationship at Nanoscale. While the structure/performance relationship for biomechanical fixation can be characterized using novel microscale and conventional macroscale testing techniques, more advanced techniques, such as nanoindentation, have been used, to extend this characterization to the understanding the processing/microstructure/property relationship at the nanoscale. Using nanoindentation across the surface of a degradable polymer, polylactic acid (PLA),

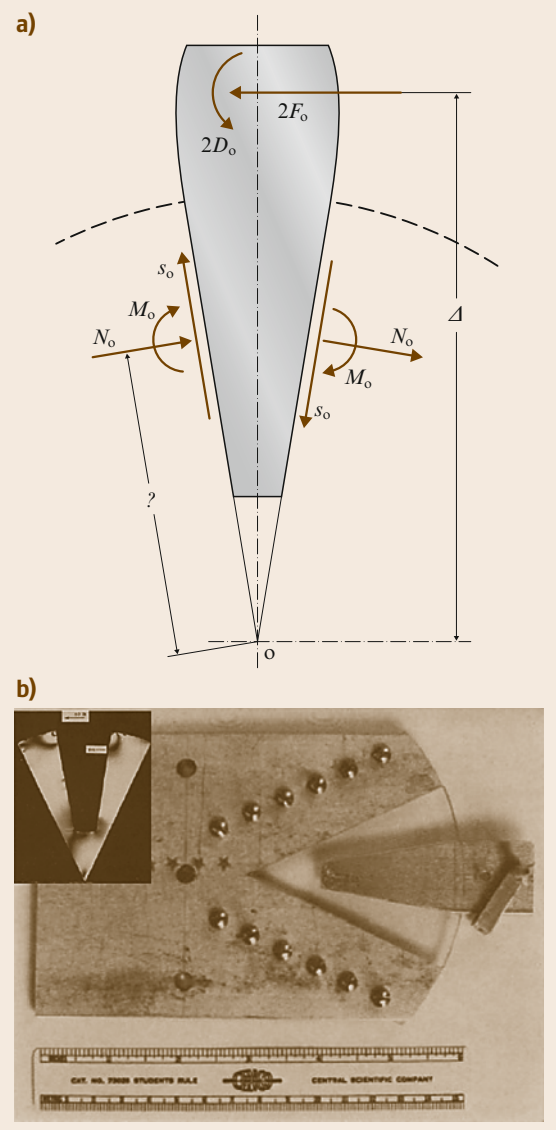

Fig. 32.13 **(a)** Model of loading conditions on tooth from orthodontic appliance, and **(b)** experimental photoelastic model used to measure the root-periodontium interfacial stresses and displacements and resulting fringe field [32.30]

which is being used for orthopaedic screws to provide temporary biomechanical fixation, it was possible to characterize that mechanical changes occurred sooner at the nanoscale than at the macroscale in a phosphate buffered saline solution after 12 weeks [32.50]. There were also statistical differences between the hardness at the surface and interior of the specimen. It was concluded that nanoindentation can be used to discern changes in mechanical properties sooner than macroscale tests.

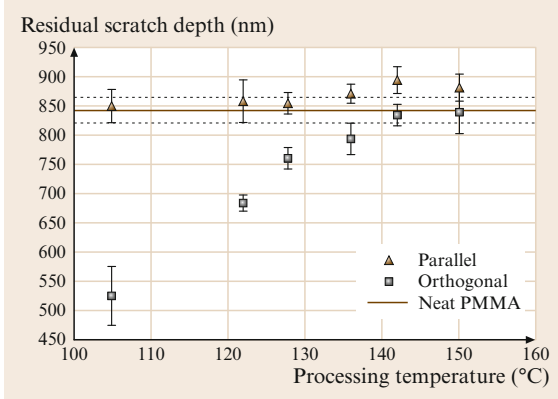

Fig. 32.14 Variation of the plastic deformation with processing temperature in self-reinforced PMMA both parallel and transverse to the fiber direction indicating increasing resistance to plastic deformation at decreasing processing temperatures due to molecular orientation [32.51]

The effects of processing conditions on the mechanical behavior of polymer composites for prosthetic applications have also been investigated using nanoindentation [32.51]. A self-reinforced PMMA composite was processed at a range of temperatures by laying fibers in a mold and heating for twelve minutes. At low processing temperatures, the fibers retain their orientation, and do not form a strong matrix bond. At higher processing temperatures, the outer surfaces of the fibers are able to bond better, at the expense of retained molecular orientation in the fibers. The nanoindenter was used to characterize these mechanical changes by scratching parallel and perpendicular to the fiber direction (Fig. 32.14). Scratching parallel produced similar amounts of plastic deformation as control specimens. However, scratching perpendicular to the fibers resulted in increased resistance to plastic deformation at low processing temperatures. It was concluded that the molecular orientation in the fibers can be indirectly ascertained from the deformation induced by scratching.

32.2.5 Deployable Stents

Experimental Mechanics Characterization Techniques

Deployable stents are geometrically complex tubular structures that are designed to undergo very large de-

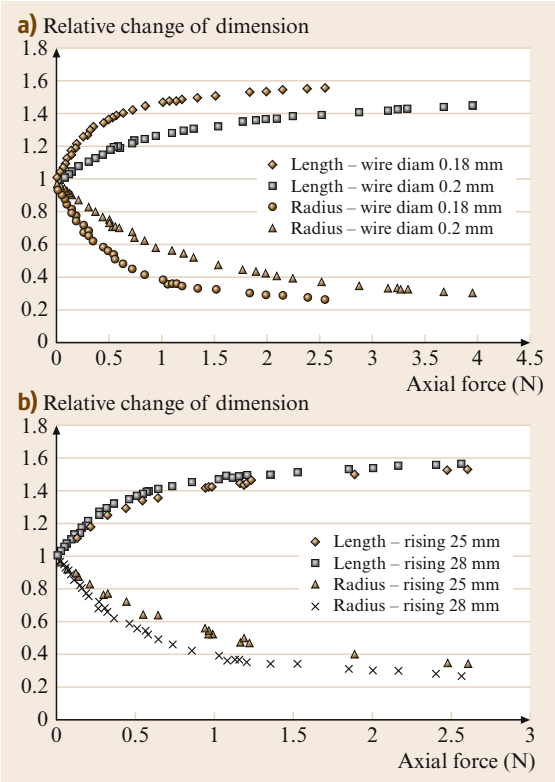

Fig. 32.15a,b Variation of rigidity for SMA stents with **(a)** wire diameter and **(b)** rise of the windings indicating more sensitivity to wire diameter [32.53]

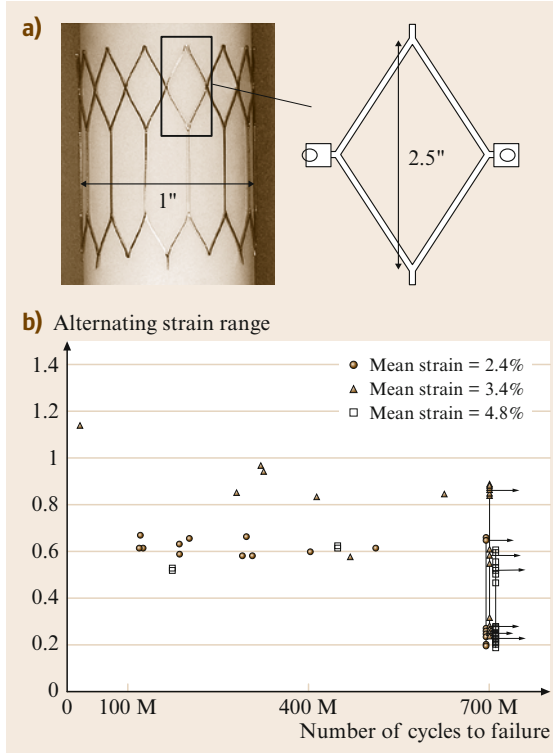

Fig. 32.16 **(a)** Fatigue test specimen design for deployable stent, and **(b)** fatigue performance data [32.54]

formations in order to expand and unblock clogged arteries, or reinforce weakened arteries. They are capable of being expanded up to 4 times their original diameter. Mechanical testing of these stents usually consist of applying point loads or circular loads to obtain qualitative comparisons of different stent designs through their diametrical response [32.52]. However, this mechanical characterization is not adequate for providing the quantitative pressure-diameter relationships necessary to analyze of the coupled artery-stent design problem because it lacks the external resistance provided by the clogged artery.

To gain insight into these relationships, specialized test methods have been developed where the stents are inserted into rubber tubes that are used to simulate the arterial wall [32.55]. When external pressure applied, the resulting change in diameter can be measured and related to the pressure. A more direct measure can be obtained by placing a stent within a collar with tabs that are pulled apart to contract the collar and apply radial pressure to the stint [32.56]. The resulting change in load N with respect to stent diameter D can be used to determine the stiffness of the stent K as follows:

$$K = \left(\frac{2}{bD_0}\right)\frac{\mathrm{d}N}{\mathrm{d}D}\,, \tag{32.4}$$

where b is the width of the collar.

A test method has also been developed to obtain pressure-diameter relationships by simulating the in vivo mechanical response of stents using wrinkled polyethylene bags that are capable of expanding without stretching [32.57]. The bags could be placed internally or externally to generate pressure, such that the resistance from expansion comes only from the stent itself and not the polyethylene film. The strains associated with the change in diameter were measured and related to the pressure and axial strain. From this experimental method, it was determined that the stiffness of the stents changed dramatically as they expanded, and could cause excessive wall stress in an artery if it is *overexpanded*. Using an analogous model of a beam on an elastic foun-

dation and a thin-wall tube under pressure, the coupled response of the stent and aorta could be modeled based on the experimental data [32.58]. By formally stating the coupled problem, the specific need has been established for an appropriate pressure-diameter relationship for the artery.

Property/Structure/Performance Relationship

Recently, interest in stent technology has shifted towards shape memory alloys (SMAs). Mechanical testing of these stents has focused on how the geometric characteristics of the stents affect their rigidity [32.53]. These tests have determined that the diameter of the wire plays a more important role in rigidity than the rise of the windings (Fig. 32.15). The rigidity also increases with increasing temperature due to the transformation from Martensite to a stiffer Austenite phase. However, the temperature at which this occurs increases as the pressure on the stent increases, so that transformation may be suppressed if too much force is required to expand the stent in the artery.

Not only will the static performance of deployable stents be important, but the fatigue performance will also dictate the life expectancy of the stent, where a pulse of 75 beats per minute over a 10 year period will produce roughly 400 million cycles of service [32.54]. Therefore, a fatigue testing protocol has been developed to determine safety factors for stents made from materials such as SMAs. Specimens are designed that reflected the areas where there would be high strain, so that the local testing conditions in the specimen and the stent were identical (Fig. 32.16). Tests conducted to assess the effects of mean strain on the fatigue life of the stent material indicated that the mid- to high-cycle fatigue life of the material appears to increase with increasing mean strain in the range of 2–4%.

32.3 Biologically Inspired Materials and Systems

The science and engineering of synthetic materials has been traditionally separated into classes of structures, length scales, and functionality that are used to differentiate disciplines such as experimental mechanics and materials science from each other. However, biological materials do not conform to disciplinary boundaries, since they possess structures that span across a full range of length scales in order to react to a variety of environmental stimuli with optimal functionality. A number of technological breakthroughs may be achieved through mimicry of the multi-scale optimization of structure and functionality in natural materials, such as material systems with morphogenesis capabilities and biomorphic explorer robots with versatile mobility [32.59]. The translation of this multi-scale optimization of structure and functionality to the science of advanced materials is a part of the new field where science and engineering are joined together: *biomimetics*. Biomimetics refers to human-made processes, substances, devices, or systems that imitate nature, and has led to the development of new *biologically inspired materials* based on biological analogs. The research in this area can either be focused on the investigation of natural materials, or on the processes that optimize the structure of materials in a manner similar to that occurring in nature. Therefore, biology can become the basis for developing new processes required for synthesizing materials, while mechanics can be used to interpret the manner in which the properties and functionality associated with the structure of natural materials have enabled them to adapt to environmental stimuli.

32.3.1 Applications

The use of biologically inspired materials and systems is quite varied, ranging from new adhesive materials to specialized robotic applications. The primary mechanical benefit in these applications often involves more durable and reliable materials, structures, and systems. For example, nacre-like ceramics are being developed that can provide superior fracture resistance in applications such as ceramic armors and thermal barrier coatings (TBCs). In the case of ceramic armors, the constituents may consistent of synthetic ceramics and polymers instead of natural. For the TBCs, it may even require using a higher temperature metal in place of the polymer, in essence resulting in a bio-inspired ceramic matrix composite (CMC).

Not only can the properties of materials be improved, but the joining of dissimilar materials can also be enhanced. For example, it is possible to create new adhesion concepts that mimic the behavior of gecko feet by micromachining pillars on a flexible surface to provide new mechanically-based adhesion mechanisms

based on the pillar radius and aspect ratio for improved joining to chemically inert materials, such as ceramics [32.60]. Furthermore, the geometry of the interface can be altered to provide mesoscale and macroscale structural features that enhance joining, like the features associated with the growth of tree trunks into soil and around obstacles.

Structurally, the mechanical behavior of a component can be enhanced through variations in the distribution of materials. For example, the microstructure of a metal-ceramic composite can be gradually varied from a strong outer surface, resistant to the high stresses associated with loading events such as cutting, to a more ductile inner surface, resistant to the cracks that may grow from the global deformations associated with loading events such as flexural. This is similar in principle to the way that bamboo distributions strong fiber reinforcement from high concentrations at the outer surface of a piece of culm to low concentrations at the inner surface.

In addition to the improved mechanical behavior that can be obtained from biological materials, it is also possible to couple the mechanical response with others, such as thermal or chemical, in order to create new multifunctional structures that are more compact, reliable, efficient, and autonomous by embedding sensors, controls, actuators, or power storage elements in polymers and polymer composites for applications such as morphing structures or robotics. For example, active materials such as shape memory alloys (SMAs) can undergo very large deformations through thermally-activated martensitic phase transformations and generate very high thermomechanical contractile forces when embedded in the form of a wire inside a polymer or polymer composite, much like the chemomechanical contractile forces produced by muscle tissues embedded within a body. This can enable the structural support in a morphing or robotic structure to change its shape for reduced aerodynamic resistance or for locomotion to mimic the performance of natural counterparts.

Materials can also be designed with entirely new transduction principles as well. For example, microencapsulated monomers can be distributed in polymers that are impregnated with a catalyst. When a crack begins to grow in the polymer, the microencapsulation is breached, releasing the monomer and initiating a polymerization reaction that *self heals* the surfaces of the cracks, reversing the damage process. This is similar in principle to the self-healing of natural structures through processes like bleeding.

32.3.2 Brief Description of Examples

Functionally Graded Material (FGMs)

In nature, many natural structures exhibit very complex microstructural distributions. For example, the connectivity and size of porosity will change from the outer surface to inner volume of a bone. Similarly, the concentration of fibers can change within the cross section of plants like bamboo. The experimental mechanics community has attempted to quantify and model the effects of these microstructures in order to gain inspiration for developing new materials, known as *functionally graded materials* (FGMs), for optimal structural performance. In particular, these materials provide an optimal distribution of phases at the microstructural level in order to minimize the amount of material and weight needed to fabricate the structure. Furthermore, the microstructural distributions can be tailored to minimize stress distributions around stress concentrations, which nature has done with holes in bones, known as *foramen*, and which bioengineers are now doing with the interfaces of prosthetics implanted in bone.

Active Materials

Conventional materials are characterized by mechanical behavior that is fairly inert or passive (i. e., they only move when a load is applied to them). Many biological materials and systems do not display this behavior, with even single cells capable of locomotion. This has led to a desire to engineer *active materials*, which are synthetic materials capable of responding to their environment. For example, one active material known as a *Shape Memory Alloy*, can undergo a martensitic phase transformation when heated that enables it to undergo substantial structural reconfiguration not associated with conventional thermal expansion. Active materials have also been synthesized from *polymers (electroactive)*, and *ceramics (piezoelectric)*. Combinations of materials, such as ferromagnetic particles and a liquid, can be used to create *magnetorheologic fluids*. For the experimental mechanics community, there has been a great deal of interest in quantifying the mechanical limitations of these active materials and overcoming these limitations through composite structures.

Self-Healing Polymers/Polymer Composites

Just as many biological materials and systems are active, they are also capable of self-healing. This behavior makes it very difficult to permanently damage them. It also enables them to be reconfigured, as is the case with

bone tissue remodeling. Developing similar concepts in synthetic materials has provided the experimental mechanics community with a unique opportunity to contribute to the engineering of these biologically inspired materials. In particular, efforts have focused on quantifying the mechanics of the interactions of cracks with microencapsulated monomers that are the healing agent for the material in order to design an appropriate size and thickness for the encapsulation in order for the crack to penetrate the microcapsule and release the healing agent [32.61].

Biologically Inspired Ceramics

One of the most unique materials found in nature are seashells. Seashells are made from calcium carbonate, which makes it chemically the same as chalk. However, the mechanical behavior of shells and chalk vary tremendously. Shells are very hard and difficult to break, while chalk tends to be much softer and more brittle. Thus, shells make very good protective coatings for living organisms, while chalk is good to write with on hard surfaces. The difference in the mechanical behavior of chalk and shells is attributed to the microstructures of the two materials. The calcium carbonate atoms in chalk tend to arrange in more of a granular, porous microstructure, exhibiting the stochastic fracture behavior observed in conventional ceramics and governed by the size of the pores. In the *nacre* of mollusk shells, these same atoms tend to configure into micron-sized, lamellar blocks known as *aragonite,* which are held together with a nanometer-thick compliant organic material in a brick-and-motor microstructure. It has been of tremendous importance to the experimental mechanics community to understand the effect of this microstructure on crack initiation and growth processes in order to quantify the benefits of creating similar, biologically inspired microstructures in conventional ceramics with processing techniques similar to *biomineralization* [32.62].

32.3.3 Functionally Graded Materials

Classes of Functionally Graded Materials

Biological materials have served as inspiration for a new class of materials that has become of significant interest to the mechanics and materials communities: *functionally graded materials (FGMs)*. FGMs are defined as materials featuring engineered gradual transitions in microstructure and/or composition, the presence of which is motivated by functional requirements that vary with location within the component [32.63]. Recent advances in the selection of materials over the past few decades have provided engineers with the new opportunities to engineer materials using FGM concepts. Currently, most structures are engineered by using a large number of uniform materials that are selected based on functional requirements that vary with location. For example, hard and wear-resistant materials are used to keep the edge of a knife sharp for cutting purposes, but to improve durability it is necessary to use strong and tough materials for the body of the blade. Abrupt transitions in material properties within a structure that result from these functional requirements lead to undesirable concentrations of stress capable of compromising structural performance by promoting crack growth along the interface. In nature, these stresses are controlled by gradually varying the material behavior through a structure, resulting in a FGM.

In a variety of biological structures, from insect wings to bamboo, evidence can be found that functionally graded materials have been selected through natural evolution to optimize structural performance through the unique coupling of material and stress distribution [32.64, 65]. FGM concepts have also been inadvertently exploited for years in synthetic structural material systems as common as dual-hard steels, through the use of surface heat treatment processes [32.61]. However, the advent of new materials and manufacturing processes now permit approaches for developing *engineered* materials with tailored functionally graded architectures, such as the *inverse design procedure* [32.66]. Thus, component design and fabrication have been synergistically combined, not just for the manufacturing of FGMs, but for the establishment of an entirely new approach to engineering structures.

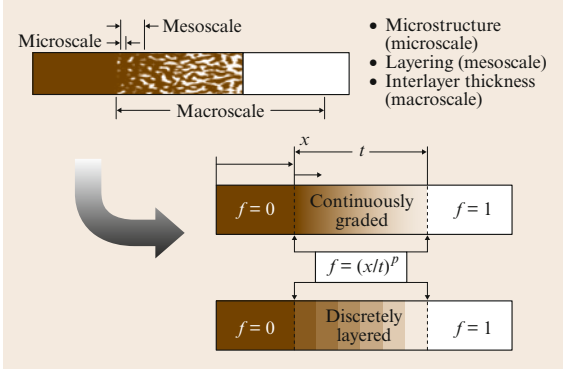

Fig. 32.17 Functionally gradient architectures and various length scales related to the architectural features involving microstructure, layering, and interlayer thickness [32.1]

The end result is the capability of engineering parts that correspond to designer-prescribed properties; e.g., parts with negative Poisson's ratios [32.67]. This approach has already being used to accurately design FGMs for aerospace applications [32.68]. FGMs have also been successfully developed for bone replacement in prosthetic applications [32.69].

Property Distributions

The gradual material variation results in a functionally graded architecture described with a continuously graded or discretely layered interlayer that has several relevant length scales, seen in Fig. 32.17, and a variation in properties in the interlayer, typically denoted by one of the following models [32.70].

$$f(x) = [x/t]^p \quad \text{(power law)}\,, \tag{32.5}$$

$$f(x) = a_0 + a_1 x + a_2 x^2 \quad \text{(polynomial)}\,, \tag{32.6}$$

$$f(x) = a_0 + a_1\,\mathrm{e}^{\delta x} \quad \text{(exponential)}\,, \tag{32.7}$$

where $f(x)$ is the value of the property at a microscale location x in the interlayer, t is the thickness of the interlayer on the macroscale, and a_0, a_1, a_2 and δ are constants describing the magnitude and rate of change in properties. When properties vary through discrete layering, the resulting gradient architecture will no longer have property variation on the microscale but rather possess mesoscale variations that are controlled by the number and relative thicknesses of the discrete layers.

Experimental Mechanics Characterization Techniques

A variety of experimental mechanics diagnostic techniques are being employed to understand FGMs. These include

1. x-ray diffraction,
2. neutron diffraction,
3. laser fluorescence spectroscopy,
4. digital image correlation,
5. coherent gradient sensing,
6. photoelasticity,
7. moiré interferometry.

These techniques are being employed on a full-field or array basis to understand the thermomechanical residual strain and stress distributions that form at the graded interface, as well as at the tip of stationary and propagating cracks.

To further advance the development of FGMs, it will be absolutely essential to use experimental mechanics to characterize the coupling between material and stress distributions. In particular, full-field deformation measurement techniques, such as Digital Image Correlation, combined with advanced microscopy techniques, such as Electron Microscopy and Atomic Force Microscopy, and localized property characterization techniques, such as microtensile testing and nanoindentation, are being developed to elucidate on the unconventional structure/property/stress relationship produced by this coupling. Non-traditional testing methods, such as hybrid numerical-experimental techniques, are also being developed because of the inherent inhomogeneous behavior of these materials.

Failure Mechanics

Fracture Resistant Materials. Recently, material and mechanical characterization of synthetic and natural FGMs has become more extensive. Most of the work on synthetic materials has focused on controlling stress and crack growth in metal/ceramic composites for applications such as prosthetics, while the natural materials have been limited to understanding the relationship between stress and microstructural distributions in bone, bamboo, and shells [32.71–80]. *Rabin* et al. successfully employed FGMs to minimize the peak axial thermal residual stress that develops in metal-ceramic joints during powder processing [32.81–83] (Fig. 32.18).

New theoretical concepts are also being developed and experimentally characterized in order to understand the fracture mechanics of FGMs. *Eischen* has shown that conventional theoretical fracture mechanics concepts, such as the stress intensity factor, can be used to describe the stress fields around a crack tip in a FGM, as [32.84]

$$\sigma_x + \sigma_y = \left[\frac{K_\mathrm{I}}{2\pi} r^{-1/2} f_0(\theta) + C_1 f_1(\theta)\right] + O(r^{1/2})\,, \tag{32.8}$$

where K_I is the mode-I stress intensity factor, and the functions of angle θ are identical to the homogeneous case. The J-integral concept has also been modified in order to preserve the path independence in the presence of the inhomogeneities occurring in the interlayer [32.85]. At the microscale, *Dao* et al. have focused on a physically based computational micromechanics model to study the effects of random and discrete microstructures on the development of resid-

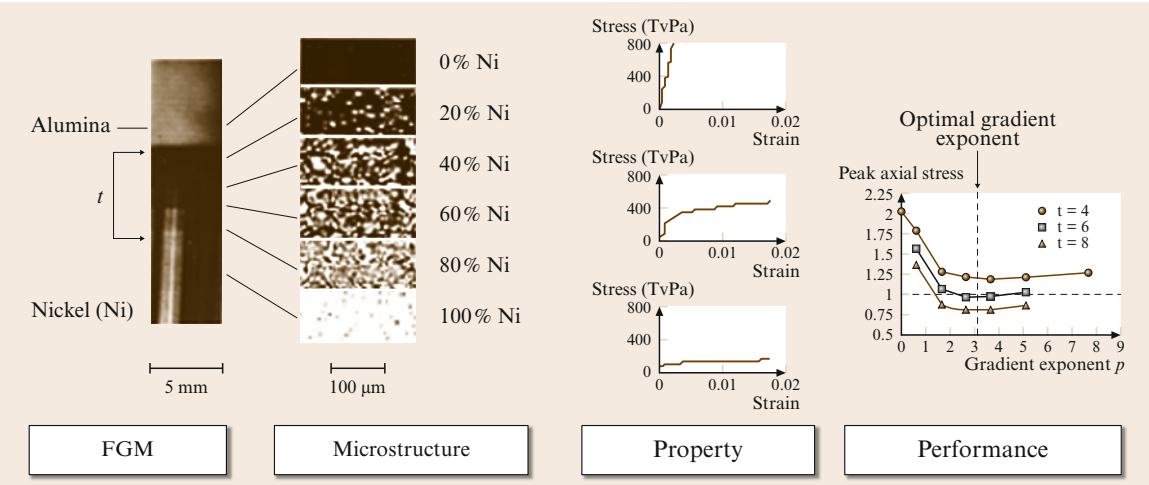

Fig. 32.18 Schematic of design approach for minimization of peak axial stresses in powder-processed metal ceramic FGM joints through an optimal gradient

ual stresses in functionally graded materials [32.86]. New finite element methods have also been developed to provide insight into the complex stress distributions that can develop within and near the graded interlayer as [32.87]

$$\mathbf{K}^{\mathrm{e}} = \int_{V_e} \mathbf{B}^{\mathrm{T}}(x, y)\mathbf{C}(x, y)\mathbf{B}(x, y)\mathrm{d}V \,, \tag{32.9}$$

where $\mathbf{K}^{\mathrm{e}}$ is the element stiffness matrix, $\mathbf{B}(x, y)$ is the matrix of shape function derivatives, and $\mathbf{C}(x, y)$ is the compliance matrix derived from the property gradient, $E(x, y) = E_0\,\mathrm{e}^{(\alpha x+\beta y)}$. FGMs have provided a plethora of unique challenges that theoretical, experimental, and computational mechanics have had to overcome in order to characterize and model the behavior of these biologically inspired materials.

Dynamic Mechanical Response. Gradient architectures have also been developed to optimize the energy-absorbing capabilities of armors, inspired by the superior impact resistance provided by wood, bone, and the shells of sea and land animals [32.88]. For these applications, it has been important to understand the dynamic failure of gradient microstructures caused by impact loading events. This failure is governed by the response of stress waves generated by the impact loading as they interact with the gradient microstrucure [32.89–94]. For example, it was determined that a stress wave with magnitude f_{i} will reflect with magnitude f_{r} at time t from a graded interface as [32.89],

$$\frac{f_{\mathrm{r}}}{f_{\mathrm{i}}} = \frac{1}{2}\int_0^{\frac{x}{d}} (\bar{\kappa}-1)n\tau^{n-1}[1+(\bar{\kappa}-1)\tau^n]^{-1}\,\mathrm{d}\tau$$
$$= \frac{1}{2}\ln\left[1+(\bar{\kappa}-1)\left(\frac{x}{d}\right)^n\right] \,, \tag{32.10}$$

$$\frac{tc_0}{d} = 2\int_0^{\frac{x}{d}} [1+(\bar{\kappa}-1)\tau^n]^{-1}\,\mathrm{d}\tau \,, \tag{32.11}$$

where d is the interlayer thickness, c_0 is the wave speed in base material 1, $\bar{\kappa}$ is the acoustic impedance of base material 2 normalized by the acoustic impedance of base material 1. From (32.11), it was determined that the gradient architecture creates a *time delay* effect that allows for peak stresses to form more slowly and for more energy to be absorbed before crack initiation when a gradient microstructure is present versus a sharp interface.

Once a crack initiates due to stress wave loading and begins to grow, the mechanics governing crack propagation differ substantially from static loading. Higher order asymptotic analyses are required to incorporate the effects of gradient property distributions into the localized stress fields around a dynamically growing crack tip [32.95]. The results of this analysis for various values of the material non-homogeneity parameter, $\alpha = \delta\cos\varphi$, where δ describes the gradient in material properties in (32.7) and φ is the property gradation direction, can be seen in Fig. 32.19. The

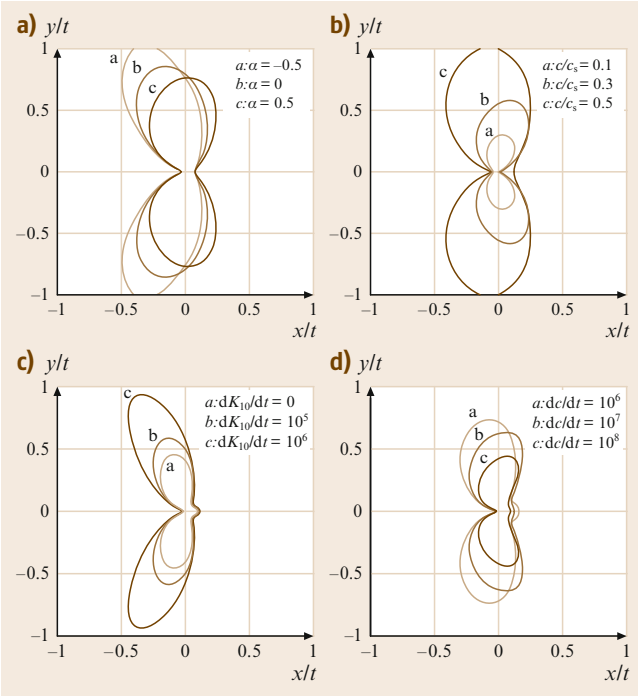

Fig. 32.19a–d Maximum shear stress contours around crack tip at (0, 0) with crack lying along the negative x-axis for **(a)** various material non-homogeneity parameters, **(b)** for different crack speeds at a non-homogeneity parameter of 0.57, **(c)** effect of rate of change of dynamic stress intensity factor of 1 MPa $m^{1/2}$ for crack velocity of 650 m/s, **(d)** various levels of crack acceleration [32.95]

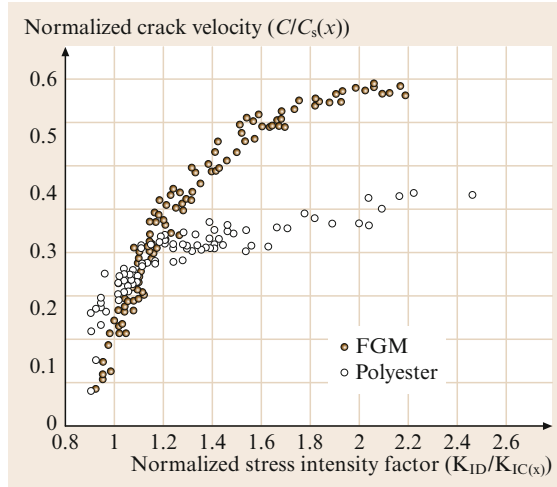

Fig. 32.20 Experimentally determined dynamic constitutive fracture relationship between crack velocity and dynamic stress intensity factor in FGMs that is distinctly different in nature from homogeneous counterparts [32.98]

data in Fig. 32.19a illustrates that for a homogeneous material, the contours tilt backward due to inertial effects, but the material gradient can compensate and tilt the contours forward. In Fig. 32.19b–d, it can be seen that they will tilt forward more and decrease in size as the crack propagates slowly, will tilt backward more and increase in size as the stress-intensity factor changes, and tilt forward more and decrease in size again as the crack accelerates. Numerical methods using cohesive-volumetric finite elements have also been employed to model dynamic crack propagation in FGMS, and parametric studies of dynamic fracture in Ti/TiB FGM specimens have shown a great sensitivity of crack motion to the gradient of bilinear cohesive failure parameters, σ_{max} (the maximum stress in the CVFE) and Δ_{nc} (the critical normal displacement component in the CVFE) [32.96]. Cohesive failure parameters have been measured using digital image correlation near slow-speed cracks exhibiting stable growth in a poly(ethylene carbon monoxide) copolymer, ECO, with a gradient in ductile-to-brittle transition controlled through UV exposure [32.97].

A dynamic constitutive fracture relationship between the crack velocity and dynamic SIF has also been established experimentally using photoelasticity on polyester-based FGMs (Fig. 32.20) [32.98]. This relationship has three regions:

1. a *stem,* which is nearly vertical and indicates that the velocity is independent of the stress intensity factor,
2. a transition region, and
3. a *plateau* where the crack reaches a terminal velocity (usually around 0.6 of the shear wave speed) that does not change substantially with stress intensity factor.

Comparing the results of an FGM with a homogeneous counterpart in Fig. 32.20 indicates that at a dynamic stress intensity factor approximately 10% greater than the fracture toughness the crack speed in the FGM increases to almost 2-fold of the speed in the homogeneous counterpart. Experimental results from a variety of experiments have indicated that this relationship is unique in nature for FGMs, and is distinctly different from homogeneous counterparts [32.99]. The dynamic stress intensity factor has also been experimentally measured in compositionally graded glass-filled epoxies at low impact velocities using the technique of coherent gradient sensing [32.100]. Using the crack tip stress field analyses from [32.84] and [32.95], it was

observed that the dynamic stress intensity factor continuously increases with crack growth when the filler volume fraction is monotonically increasing, and continuously decreases when the gradient is of the opposite sense similar to a homogeneous material (Fig. 32.21).

Geometrically Complex Interfaces. Another area where functionally graded materials have been making an impact is to enhance the toughness of interfaces with more geometrically complex features. For example, one of the simplest geometrically complex interfaces is a circular inclusion. A classic circular inclusion is a hole, which is conventionally characterized as an inclusion with no mechanical properties. For a two-dimensional elastic material, this case produces the classic stress concentration of three times the far field stress applied to the structure. In nature, holes exist within structures, such as bones, in order to provide internal access for arteries and veins. One such hole is known as a *foramen*, and the study of the graded material property distribution around the foramen has become the basis for a new concept in stress concentration reduction for holes in synthetic structures, like rivet holes in the skin of aircraft structures [32.101–103]. Using the full-field deformation technique of Moire inerferometry, the strains around the foramen were characterized and were related to variations in the elastic modulus to determine an optimal gradient for reducing the stress concentration. Assuming a power law relationship between density of the material ρ and the elastic modulus and yield strength, $E : \rho^b$ and $\sigma_y : \rho^\beta$, the optimal gradient was found to depend on the ratio of the power law exponents, β/b. At some point ($\beta/b \approx 0.75$), when strength does not increase fast enough with density compared to the modulus, then it pays to divert the stresses away from the hole, as is the case of a foramen in bone ($\beta/b \approx 0.5$).

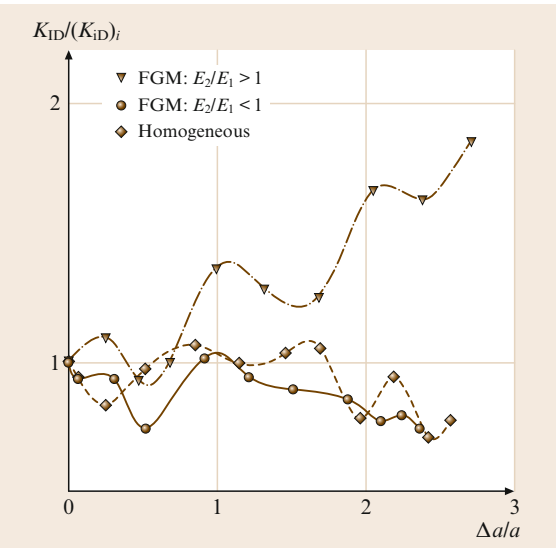

Fig. 32.21 Experimentally measured dynamic stress intensity factor versus crack length in FGMs indicating increases with increasing filler volume fraction ($E_2 > E_1$) in and decreases when the gradient is in the opposite sense, similar to a homogeneous material [32.100]

A variation on the use of FGMs at geometrically complex interfaces has been the engineering of the geometric complexity to increase the fracture resistance of the interface by redistributing the stress and strain at the interface. For example, development of a new multi-material multi-stage molding technology enabled the geometric features of the interface to be precisely engineered at the milliscale [32.104]. Experimental measurements of deformation fields associated with circular and square features using DIC determined that the strength of the interface could be enhanced by 40% using the features to convert the failure mode from the traditional K-dominant fracture to ligament failure. It was also determined that mechanical interlocking mechanisms attributed to the geometric features enabled the interface to retain at least 30% of its strength in the absence of chemical bonding.

Another investigation conducted into free-edge stress singularities associated with bio-inspired designs for dissimilar material joints was conducted based on the growth patterns when trees grow around an obstruction [32.105]. Using photoelasticity, model polycarbonate-aluminum specimens were prepared with varying degrees of joint angles associated with the free-edge of the interface between the dissimilar materials. The convexity associated with these angles was determined to eliminate the free-edge stress singularity, increasing the tensile load capacity by 81% and reducing the material volume by 15% over traditional butt-joint specimens with sever free-edge stress singularities.

32.3.4 Self-Healing Polymers/ Polymer Composites

Microstructure/Property/ Performance Relationship

One of the most unique characteristics of biological systems has been their ability to heal. Recently, advancements have been made in the formulation of

Table 32.3 Summary of biological sources of inspiration for self-healing and potential approaches using composite/polymer engineering [32.106]

Biological source of inspiration	Composite/polymer engineering
Molecular cross-linking	Remendable polymers
Bleeding	Microcapsules, hollow microfibers
Blood clotting	Nanoparticles, healing resin
Bone reparation	Short microfibers

polymers and polymer composites to impart this ability to synthetic materials [32.106]. *Material heal thyself* has now become a reality. Most of the work conducted on self-healing polymers/polymer composites has focused on the impregnation of polymers with a Rhuthenium-based catalyst known as Gibbs' catalyst, that will polymerize monomers that are microencapsulated and dispersed in the polymer. When a crack propagates through the polymer, it breaks open the microcapsules, and then enables the monomer to polymerize and heal the fracture surfaces in a process inspired by *bleeding*. The size and concentration of the microcapsules has a significant effect on the healing efficiency. Variations on this *bleeding inspired* concept include vascular networks consisting of hollow microfibers instead of hollow spheres, and use of microfibers or nanoparticles to mimic the behavior of bone reparation or blood clotting [32.4, 106–108]. A summary of biological sources of inspiration for self-healing processes and some potential approaches to developing them using composite/polymer engineering can be seen in Table 32.3 [32.106].

Remendable Polymers. Another attractive self-healing approach for pure polymers is to use mending cross-linked molecular chains activated by a thermal input to remend the fracture surfaces [32.109]. Remendable polymers have been engineered at the molecular level to exhibit this self healing behavior. Using nuclear magnetic resonance spectroscopy, the disconnecting and reconnecting of intermonomer linkages have been tracked during heating and cooling of the material. These polymers are engineered by forming a macromolecular network formed in its entirety by reversible cross-linking covalent bonds, where the degree of cross-linking is so high that they are relatively stiff and strong. Healing efficiencies of over 50% were reported for these polymers. Photochemical healing has also been achieved via re-photocycloaddition of cinnamoyl groups that form during crack growth in a polymer engineered using a tricinnamate monomer [32.110].

Microencapsulated Healing Agents. One characteristic of natural materials that differs substantially from synthetic is their ability to heal after small amounts of damage. This ability is conveyed through the growth of new tissue by a collagen-based DNA-controlled mechanism. While the building blocks of natural materials has been inherently designed for this capability, there is no such mechanism available in synthetic material systems. A bio-inspired alternative was developed in the late 90s employing the technique of microencapsulation to embed healing agents inside of polymers and polymer composites that would be released whenever a crack compromised the encapsulation, creating an au-

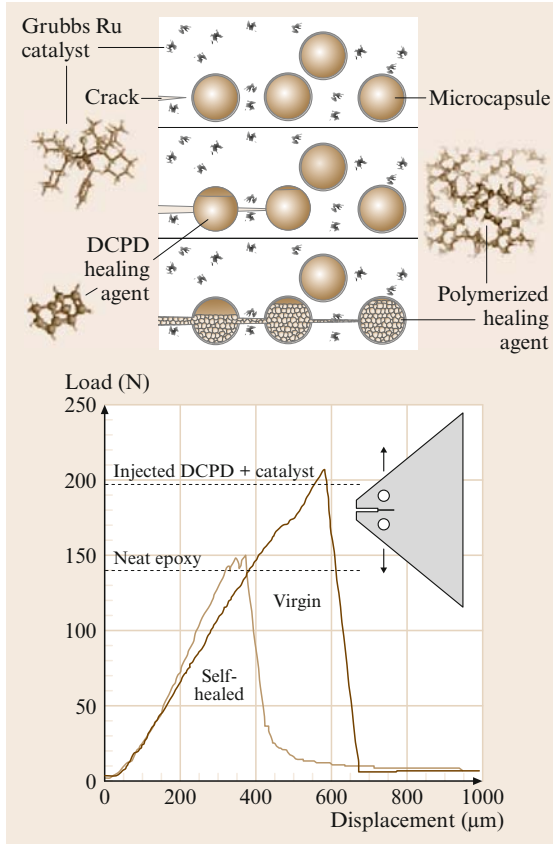

Fig. 32.22 Fundamental concept for microencapsulated healing agents and healing efficiency [32.61]

tonomic healing response with efficiencies greater than 70% (Fig. 32.22) [32.3, 61, 111]. This novel approach raised a number of mechanics issues to optimize the design of this bio-inspired material that are being addressed by the experimental mechanics community that will now be discussed.

For the design of microencapsulated healing agents there are two fundamental mechanics issues:

1. how can they be compromised by propagating cracks, and
2. what is their healing efficiency.

The first is a microscale issue while the second is macroscale. For the first issue, there are a number of design characteristics for the microencapsulation that can be optimally engineered:

1. thickness of encapsulation,
2. properties of encapsulation,
3. size of encapsulation, and
4. the dispersion of the microscapsules.

An additional issue that is being investigated in order to enhance the commercial appeal of these bio-inspired materials is replacement of expensive ingredients, such as the Ruthenium-based Grubbs' catalyst.

Characterization of Fracture Mechanics

To determine the optimal design characteristics, the fundamental interaction between microcapsules and propagating cracks must be investigated. Currently, computational experiments are being performed to understand this interaction at a very basic level. However, the development of microscale and nanoscale mechanical and microstructural characterization techniques, such as microscale DIC, are now enabling the assumptions in these computational experiments to be verified by direct measurement of mechanical properties on and near microcapsules, by determining path of crack tips near microcapsules, and by measuring the local deformation fields near the crack tip and the microcapsules.

For the healing efficiency, very fundamental mechanical characterization experiments have been performed using unconventional fracture specimens with tapered geometries to arrest the development of cracks in more brittle polymers [32.111]. The use of brittle polymers is dictated by the interaction of the crack with the microcapsules, where more ductile or rubbery polymers permit cracks to deflect between microcapsules rather than through them. The fracture resistance of ductile or rubbery polymers also preclude the necessity of healing agents at this time, unless an embrittling mechanism is present such as an oxidizing atmosphere like ozone or an aging mechanism like chain decomposition. Cracks can be grown either monotonically or fatigue, and then the change in load-bearing capacity with respect to healing time can be assessed. For optimal healing efficiency, almost 100% of the load bearing capacity can be recovered. It has also been demonstrated that the microencapsulation can be designed such that the fracture toughness and strength of the pure polymer is not compromised. In addition to the healing efficiency, it is also desirable to increase the speed of the healing process. For general fatigue fracture, this is generally not an issue. However, for overloads, significant crack propagation can be reversed if the healing time is faster than the time between overloads. Currently, healing agents can achieve near optimal efficiency as quickly as a few hours. It is envisioned that healing times of minutes, similar to the cure time of fast-curing epoxies, will be achieved in the near future using cheaper ingredients then are currently employed.

32.3.5 Active Materials and Systems

Classes of Active Materials and Structures

Active Materials and Structures (aka, *smart or intelligent materials and structures*) are defined by their ability to mechanically deform in response to an external stimulus such as an electrical signal, similar to muscle tissue. They basically fall into three categories.

1. SMA and SMA composites.
2. Piezoceramics.
3. Electroactive polymers (EAPs).

Each class is essentially defined by the characteristic of the main active material ingredient (i. e., metal, ceramic, polymer). The active response of each class is summarized in Table 32.4 and compared with muscle tissue.

Table 32.4 Comparison of active response for each class of active material in comparison to natural muscle tissue

Class of active material	Active strain (%)	Active stress (MPa)	Frequency response (Hz)
EAP	300	10	1000
SMA	10	200	100
Piezo	1	100	100 000
Muscle	30	1	1000

SMA and SMA Composites

In many experimental investigations of biologically inspired materials and systems, characterization has focused on relating the distribution of material properties, such as hardness, to the microstructure, independent of the stress distribution in the structure. For advanced structural systems, known as smart structures, the material properties can actually depend on this stress distribution. In particular, smart structures can be fabricated from materials, such as shape memory alloys (SMAs). SMAs experience martensite-austenite phase transformations that enable them to fully recover inelastic deformations due to martensite detwinning when they are heated above their austenite finish temperature (known as the shape memory effect), and will exhibit pseudoelastic behavior when deformed above these temperatures (Fig. 32.23). This novel thermomechanical behavior has led to extensive use of these materials in biologically inspired and biomedical applications, such as deployable stents [32.53]. They can also be embedded within other materials, such as composites where functionally grading the distribution of SMA wire reinforcement has been demonstrated to theoretically increase buckling strength by controlling the distribution of recovery stresses that are generated when the wires are heated [32.112].

By thermomechanically treating SMAs, they can be trained to deform to two different states upon heating and cooling, an effect known as the two-way shape memory effect (TWSME), which is a deformation behavior similar to the chemomechanical constriction of muscle tissue. Functionally graded smart materials are now being fabricated on the microscale from SMA thin films with compositional gradients for microelectromechanical systems (MEMS) applications [32.113]. By grading these films, a unique coupling develops between the microstructure and residual stresses that enable these materials to exhibit thermomechanical behavior as fabricated that would not otherwise be possible, such as repeatable actuation behavior (i. e., the TWSME). Using the biologically inspired actuation behavior of these graded films, new biomimetic concepts can be developed for novel structures fabricated from these materials, such as micropumps for MEMS applications.

A similar coupling has also been achieved by functionally grading the distribution of SMA wires exhibiting the one-way SME in polyurethanes to produce an *equivalent* TWSME that mimics the actuation behavior of biological structures such as the fibrous muscle tissue found in a heart or a bird's wings (Fig. 32.24) [32.114]. The equivalent TWSME for an SMA wire can be determined for wires embedded in a beam structure from the bending curvature k_2 and elongation e of the beam associated with the location of the SMA wire d the geometric properties of the beam, and the properties of the wire and beam material as [32.115]

$$k_2 = \frac{\alpha\beta d\gamma(\gamma\beta-\gamma-\beta)}{(\gamma+\beta)\gamma-\alpha\beta d^2(\gamma\beta-\gamma-\beta)}\left[\varepsilon^{\mathrm{t}}+\alpha^{\mathrm{a}}(T-T_0)\right] , \tag{32.12}$$

$$e = \frac{\gamma^2\beta}{(\gamma+\beta)\gamma-\alpha\beta d^2(\gamma\beta-\gamma-\beta)}\left[\varepsilon^{\mathrm{t}}+\alpha^{\mathrm{a}}(T-T_0)\right] , \tag{32.13}$$

where $T-T_0$ is the temperature change in the SMA wire, ε^{t} is the stress-free transformation strain, α^{a} is the

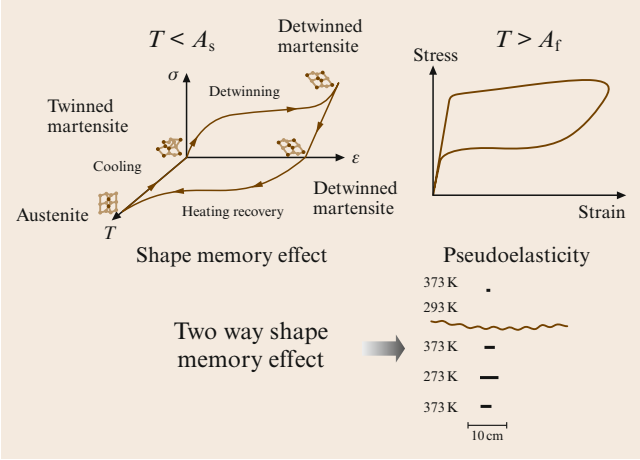

Fig. 32.23 Novel thermomechanical behavior of Shape Memory Alloys exploited for numerous biologically inspired and biomedical actuation applications. related to the deformation and transformation of martensite and austenite phases at different temperatures

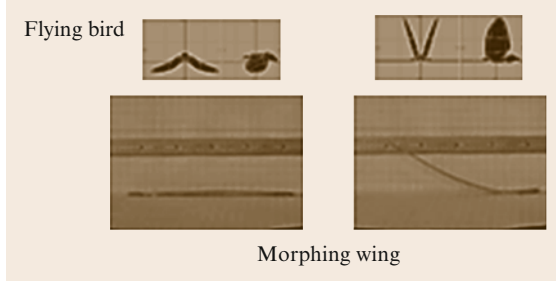

Fig. 32.24 Biologically inspired morphing wing structure using graded one-way SMA wire distribution in polyurethane that reversibly deform when heated

coefficient of thermal expansion, γ is the fraction of the recovery strain transferred to the matrix, and

$$\alpha \equiv A/I_2 \,,$$
$$\beta \equiv E^{\mathrm{a}} A^{\mathrm{a}}/(EA) \,, \qquad (32.14)$$

where A is the cross-sectional area of the beam, A^{a} is the cross-sectional area of the wire, E is the Young's modulus for the beam, E^{a} is the Young's modulus of the beam, and I_2 is the moment of inertia of the beam cross section. The experimental mechanics technique known as digital image correlation has been used to quantify the transfer of the recovery strain between the SMA wire and the matrix for comparison with 3-D thermomechanical finite element analysis (Fig. 32.25) [32.116].

Piezoceramics

Another active material system that has seen limited use in biologically inspired structural applications has been *piezoceramics*. Piezoceramics produce deformation by coupling electric fields with mechanical and thermal loadings (*piezoelectric effect*). In reverse, they can produce electrical signals upon mechanical loading similar to act as a force sensor, similar to bone [32.117]. The deformations associated with these fields are very fast (>10 kHz) and very small (<1%), but they can be configured to amplify their relative displacement in the form of *stacks*. However, because of the inherently small strain associated with their piezoelectric response, their experimental mechanics applications have been limited to primarily vibration control in the form of *patches* which are attached externally to a passive structure [32.118]. They have also been popular as strain and force transducers, where they exhibit greater sensitivity and signal-to-noise ratios than conventional strain gages or load cells [32.119].

In order to overcome some of these limitations for applying piezoceramics, there has been a great deal of historical work by the experimental mechanics community to design mechanical amplifiers based on flextensional principles for piezoceramic structures to enhance the deformation of stacks for low-frequency high-power applications such as sonar [32.120]. Applying these amplification principles at the microscale, microamplifiers for piezoceramics have been fabricated using LIGA with a mean static amplification factor of 5.48 and frequency response of 82 kHz [32.121].

In addition to the electromechanical strain limitations, piezomaterials tend to be very brittle, with the fracture toughness of the most common piezoceramic (PZT) being below 1 MPa $\mathrm{m}^{1/2}$. Their fracture behavior is also very sensitive to applied electric fields, such that a positive electric field increases the apparent fracture toughness for cracks propagating parallel to the poling direction, while a negative electric field has an opposite effect which can be modeled using a work energy based criterion for domain switching and a Reuss-type approximation for poly-domain piezoelectrics [32.122]. Thus, there is interest in the experimental mechan-

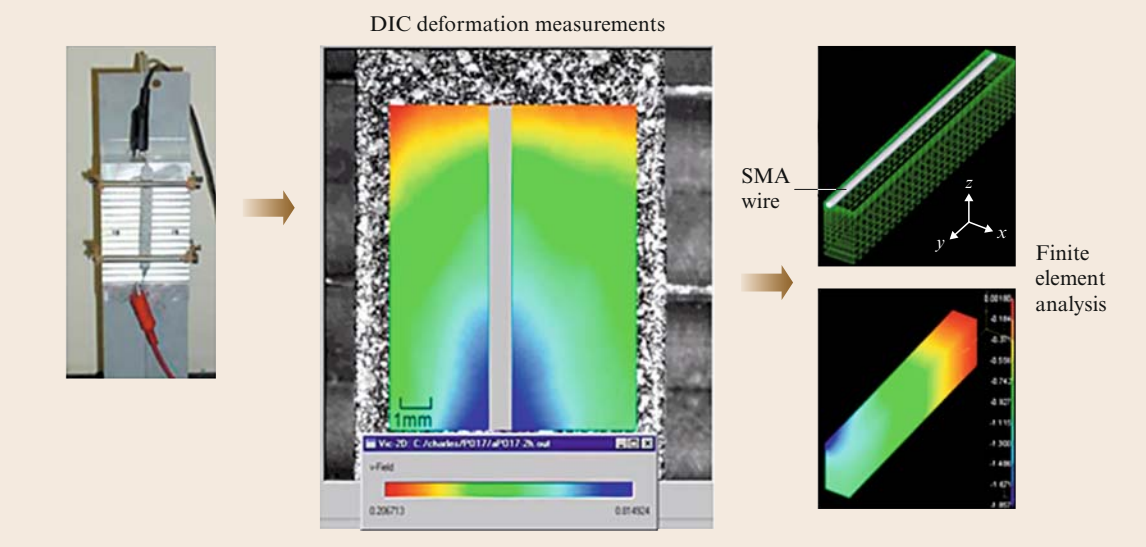

Fig. 32.25 Quantification of the transfer of recovery strain between an SMA wire and matrix material using the experimental mechanics technique of digital image correlation and comparison with 3-D thermomechanical FEA [32.116]

ics community for better understanding these fracture mechanisms in piezoceramics in order to increase their fracture toughness, so they can be deployed in more mechanically demanding applications. For example, the addition of rare earth elements can increase the fracture toughness above 1 MPa $m^{1/2}$ without significant effect on the piezoelectric properties [32.123]. The addition of Pt particles to PZT also enhances mechanical properties, but at the expense of piezoelectric properties, which is minimized through the use of functionally graded microstructures [32.124].

Electroactive Polymers (EAPs)

To complete the development of active materials, polymers have also been developed to provide mechanical behavior that can be altered with electrical signals [32.126]. These materials are called *electroactive polymers* (EAPs). Electric EAPs also come in the following forms: *ferroelectric polymers*, *electrostatic graft elastomers*, *electrostrictive paper*, *electro-viscoelastic elastomers*, and *liquid crystal elastomers* (LCEs). Alternatively, ionic EAPs can produce significant volumetric changes through reversible counter-ion insertion and expulsion during redox cycling (reduction and oxidation reactions through exchange of ions with electrolytes) at low voltages (1–10 V). However, ionic EAPs are a combined solid-fluid system, have slow response time (>0.1 Hz), have difficulty holding their strain, and produce relatively low actuation forces. The forms of ionic EAPs include *ionic polymer gel* (IPG), *ionomeric polymer-metal composite* (IPMC), *conductive polymer* (CP), and *carbon nanotube* (CNT).

EAPs can exhibit mechanical properties and actuation behavior that are very close to muscle tissue. The simplest design of an EAP is to sandwich a polymer with a low elastic stiffness and high dielectric constant

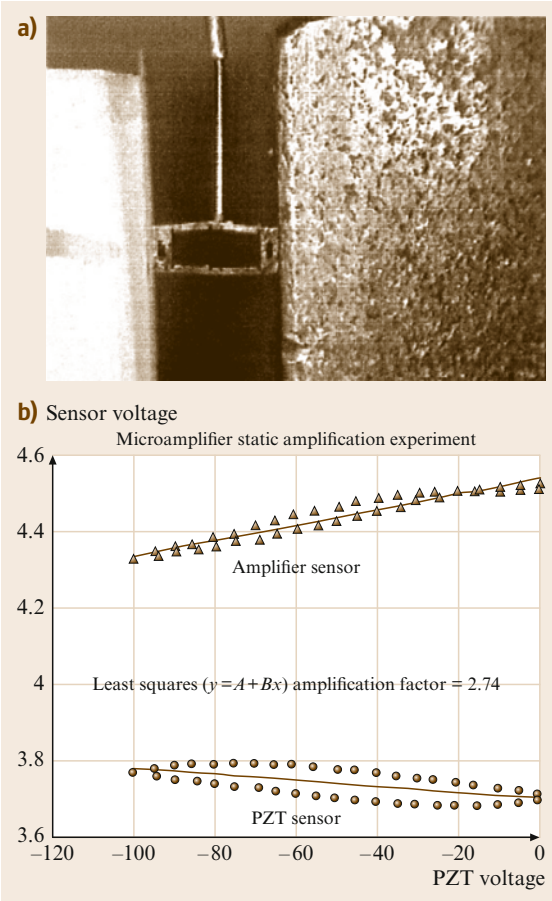

Fig. 32.26 **(a)** Microamplifier fabricated using LIGA sandwiched between piezoceramics, and **(b)** the static amplification of the piezoceramic displacement [32.119]

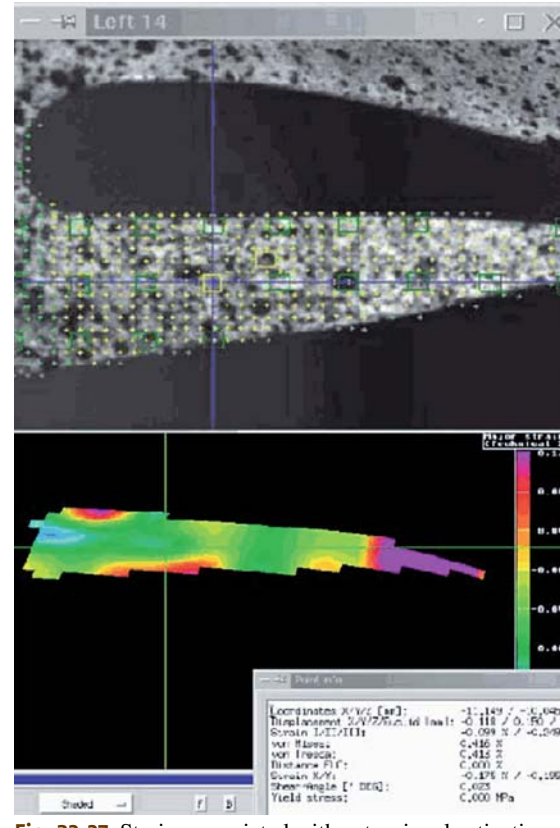

Fig. 32.27 Strains associated with extensional activation of a synthetic muscle formed from EAPs [32.125]

in between two electrodes, creating an electric EAP which is known as an *electrostatically stricted polymer* (ESSP). By applying a high voltage V to electrodes separated by a thickness t of a polymer with dielectric constant ε and area A an actuation force F will be produced as

$$F = \frac{\varepsilon\varepsilon_0 A V^2}{2t^2} , \qquad (32.15)$$

where ε_0 is the dielectric constant of a vacuum. The resulting deformation will depend on the constitutive behavior of the polymer. The response of these electroactive polymers is fast (ms levels), and the response can be held for long periods of time under DC activation. However, the required voltage levels are high ($\approx$150 MV/m), and the glass transition temperature is inadequate for low temperature applications. Recently, 3-D digital imaging techniques have been employed to look at the deformations associated with synthetic muscles formed from EAPs (Fig. 32.27) [32.125].

32.3.6 Biologically Inspired Ceramics

Ceramics are a wonderful engineering material. Extremely hard, stiff, fairly light, chemically inert, and thermally-resistant, they have found many uses, especially in harsh environments and protective applications. However, they suffer from a major drawback: they are very brittle and easily fracture. This is due to the bonding and microstructure of ceramics. Ceramics are ionically bonded, which is a very strong and highly directional bonding. Thus, it takes a great deal of local force to rearrange the bonding and to create defects, which is necessary to generate plasticity through the traditional mechanisms of dislocation motion in a crystal. However, the lack of defects creates grain morphologies that result in significant amounts of porosity in the microstructure. This porosity becomes sites for initiating crack growth, and the strength of the material is ultimately determined by the loading necessary to generate cracks from these pores rather than to propagate dislocations through a crystal structure.

To overcome the drawback of brittleness manifested through low fracture toughness, high-performance ceramics are being engineered with reduced grain sizes that reduce the size and amount of pores, as well as to create more tortuous crack paths for intergranular crack growth mechanisms that lead to more crack bifurcation and diffuse microcracking to consume more energy through the formation of greater crack surface area. While this significantly strengthens the ceramic, it still does not permit the loads or generates the defects that are necessary to initiate the motion of dislocations to consume significantly greater fracture energy.

Nacre Inspired Ceramics

The inherent limitation on dislocation motion due to the type of bonding present in ceramic has significantly limited progress with the current ceramic engineering approach Composites have been an attractive alternative, where very ductile, relatively strong metals, such as aluminum, can significantly enhance fracture toughness of ceramics, such as silicon carbide, at the expense of thermal resistance, while ceramic-ceramic composites, such as carbon–carbon, can create mesoscopic interfaces to provide additional surface area at another length scale to consume more fracture energy. On the other hand, rigid biological ceramic composites, such as the *nacre* of mollusk shells, possess fracture mechanisms that increase the absorption of fracture energy through the use of very thin layers of an biopolymer (*proteinaceous*) phase between platelets of calcium carbonate known as *aragonite* that generate crack bridging and tunneling mechanisms at the nanoscale through the molecular strength of the molecules in the biopolymer [32.5, 128]. Digital image correlation has been

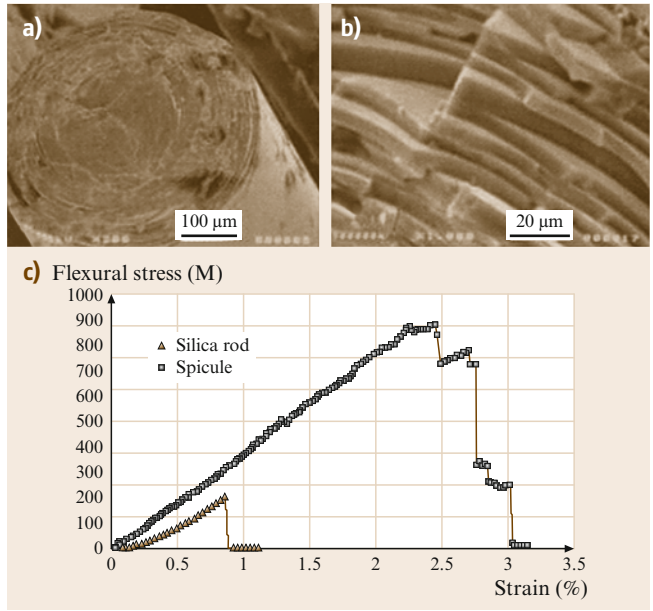

Fig. 32.28a–c Micrographs of **(a)** amorphous silica rod and **(b)** a sponge spicule, along with the associated four-point bend data **(c)** indicating a 4-fold increase in strength and a 3-fold increase in failure strain [32.127]

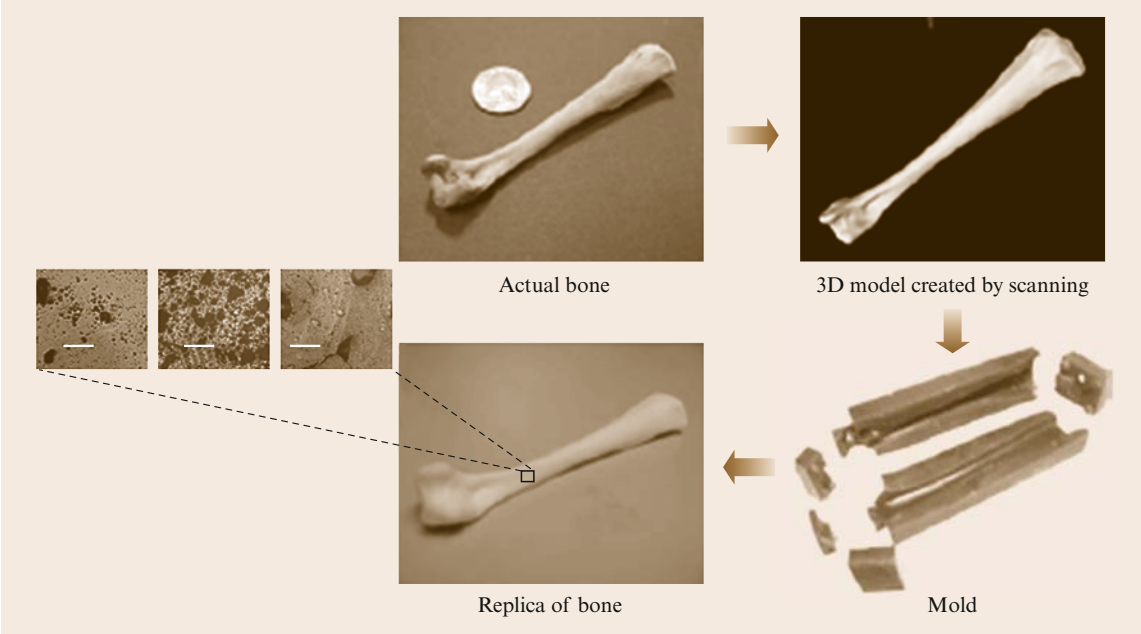

Fig. 32.29 Replamineform inspired bone structures (RIBS) fabricated using a new multistage molding technology to create geometric complex features on the milliscale and advanced ceramic gelcasting technology with carbon fugitives that form replamineform inspired microstructures [32.130]

used at the microscale to determine that sliding of the platelets during crack growth can give rise to a rising crack resistance curve characteristic of a material that can stabilize and even arrest cracks [32.129]. The geometric features of the ceramic platelets alone can substantially alter the fracture resistance, as evidenced by a 4-fold increase in strength and 3-fold increase in failure strain when comparing a sponge specula made of silica with a standard amorphous silica rod (Fig. 32.28) [32.127].

At the nanoscale, the characteristics of nacre inspired ceramics leads to an insensitivity of fracture processes to flaws that are present in the biopolymer phase, which is governed by the aspect ratio of the platelets ρ^* and the shear strength of the biopolymer τ_p^f as

$$\rho^* = \frac{1}{\tau_p^f}\sqrt{\frac{\alpha E_m \gamma}{h}}\,, \tag{32.16}$$

where E_m is the Young's modulus of the platelet, γ is the surface energy, h is the thickness of the platelet, and α depends on the crack geometry [32.9]. Nacre also exhibit a much higher fracture toughness in water, where a ten-fold increase can occur through the activation of more plastic work at the platelet interfaces normal to the crack propagation, and are extremely temperature dependent [32.6, 131]. Synthetic methods for processing these lamellar microstructures have been developed using Portland-cement to create a self-aligned calcium carbonate plates or prismatic morphologies, however they have yet to be mechanically characterized in a manner similar to Nacre [32.132].

Replamineform Ceramics

Another natural ceramic that possesses unique microstructural characteristics is coral. It has been a source of inspiration in bone healing and remodeling since the 1970s due to the interconnected porosity that develops during the biomineralization process [32.133]. Pores with the appropriate diameter ($>100\,\mu$m) allow the in-growth of vascularized bone tissue for prosthetics and prosthetic fixation [32.134]. The *replamineform* process was developed to duplicate the interconnected, controlled pore size microstructure of particular coral species (such as Porites and Goniopora) [32.135, 136]. Processes have also been developed to fabricate biologically inspired ceramics that are potential prosthetic materials or structures using hydroxyapatite particles and taking advantage of the

physics of ice formation [32.137]. These materials are four times stronger than conventional prosthetic materials. Also, replamineform inspired bone structures (RIBS) have been formed by combining a new multi-piece molding technology for creating geometrically complex milliscale features with an advanced ceramic gelcasting technology using carbon fugitives to create replamineform inspired microstructures (Fig. 32.29) [32.130].

Experimental mechanics investigations on RIBS have indicated that 35% total porosity in bone replicas can decrease strength 86%, decrease stiffness 75%, and reduce total deformation 30% in comparison to bone replicas with 13% total porosity fabricated without carbon. The reduction in stiffness was consistent with a model developed from microscale finite element analysis of overlapping solid spheres that produce microstructures similar to those observed in these specimens. Furthermore, the increased porosity of the material permits axial cracks to grow and bifurcate more easily at substantially reduced deformation and load levels.

32.4 Conclusions

32.4.1 State-of-the-Art for Experimental Mechanics

Experimental mechanics is being applied to the development of new implantable biomedical devices through understanding the processing/microstructure/property relationship in biocompatible materials and the structure/performance of the devices that are being made from them. It is also being used to provide insight into the fundamental processing/microstructure/property/structure/performance relationship at multiple length scales for new biologically inspired materials and systems that are analogs of biological counterparts. To elucidate on the synergy between the research and development activities in these areas through the application of experimental mechanics can be seen through specific examples that are motivated by the following challenges that are faced in understanding the mechanics of these unique materials and systems: *Wide range of properties, wide range of length scales, thermal and chemical constraints (in vivo)*, and *multifunctionality*.

The application of experimental mechanics to implantable biomedical devices has accelerated their development by understanding fundamental performance issues. Prosthetics have benefited from a detailed understanding of the microstructure/property relationship in composites from which they are made, as well as the development of simulators and testing techniques for characterizing the structure/performance relationship when they are deployed. Characterizing the processing/microstructure/property relationship in biomaterials using micro/nanoidentation have facilitated the development of biodegradable polymer composites for biomechanical fixation. The design of deployable stents has been improved by characterizing their performance using new test methods and relating it to the materials, geometry of the stents, and the mechanical interaction of system components. The novel use of experimental mechanics has also led to the development of new implantable biomedical devices, such as SMA stents that can self-expand through heating instead of requiring a balloon as the original passive metal stents used in angioplasty.

All of the experimental mechanics knowledge obtained from implantable biomedical devices and the biological systems they are used in is now being applied to developing new biologically inspired materials and systems. Graded natural structures formed from adaptation to changing environmental conditions and defects, such as foramen holes, have provided inspiration for creating Functionally Graded Materials out of composites formed from polymers, metals, and ceramics that provide greater fracture resistance and tougher geometrically complex interfaces. The actuation behavior of soft tissues such as muscle is providing inspiration to create active structures using Shape Memory Alloy metals, piezoceramics, and Electroactive Polymers that provide more compact, reliable, efficient, and autonomous replacements for conventional actuation systems. The self-healing behavior of biological materials are also inspiring the development of similar attributes in polymers and polymer composites using engineered polymer molecules and microencapsulated healing agents that mimic the bleeding process. Finally, the microstructures of bone and seashells are serving as templates for creating more fracture resistant and biocompatible ceramics materials and structures.

32.4.2 Future Experimental Mechanics Research Issues

It is clear from the experimental mechanics research that has been conducted to date on implantable biomedical devices and biologically inspired materials and systems, that the following research issues need to be addressed to further the development of these devices, materials, and systems.

1. Further development of 3-D measurement techniques are required to understand the mechanics associated with the geometric complexity of biologically inspired systems and advanced implantable biomedical devices.
2. Further development of non-destructive and non-invasive sub-surface measurement techniques are required to elucidate more directly on the interfacial interactions between natural structures and implantable biomedical devices both in vitro and in vivo.
3. Advancement of multiscale measurement techniques to elucidate on the link between nanoscale, microscale, and mesoscale mechanisms involved in the transfer of load and resulting failure in biologically inspired materials.
4. More dynamic measurement techniques to resolve time-dependent processes involved in polymer-based implantable biomedical devices and biologically inspired materials and systems.
5. More advanced loading systems and test methods are required for improved experimental in vitro simulation of the performance of implantable biomedical devices.

32.5 Further Reading

- Overview of experimental mechanics testing methods for prosthetics [32.18].
- Overview of tissue scaffolding [32.15, 17, 44].
- Overview of biomaterials [32.8, 20].
- Overview of hierarchically structured materials [32.6, 10, 11].
- Overview of mechanics and fundamental principles in biologically inspired materials and systems [32.1, 2, 21].
- Overview of processing, characterization, and modeling of FGMs [32.63, 70].
- Overview of dynamic failure in FGMs [32.98].
- Perspective on self-healing polymers/polymer composites [32.106].

References

32.1 H.A. Bruck, J.J. Evans, M. Peterson: The role of mechanics in biological and biologically inspired Materials, Exp. Mech. **42**, 361–371 (2002)

32.2 I. Amato: *Stuff: The Materials the World is Made of* (Basic Books, New York 1997)

32.3 S.R. White, N.R. Sottos, P.H. Geubelle, J.S. Moore, M.R. Kessler, S.R. Sriram, E.N. Brown, S. Viswanathan: Autonomic healing of polymer composites, Nature **409**, 794–797 (2001)

32.4 C. Dry: Procedures developed for self-repair of polymer matrix composite materials, Compos. Struct. **35**, 263–269 (1996)

32.5 H.R. Piehler: The future of medicine: biomaterials, MRS Bull. **25**, 67–69 (2000)

32.6 S. Karmat, H. Kessler, R. Ballarini, M. Nassirou, A.H. Heuer: Fracture mechanisms of the Strombus gigas conch shell: II-micromechanics analyses of multiple cracking and large-scale crack bridging, Acta Mater. **52**, 2395–2406 (2004)

32.7 J.L. Katz, A. Misra, P. Spencer, Y. Wang, S. Bumrerraj, T. Nomura, S.J. Eppell, M. Tabib-Azar: Multiscale mechanics of hierarchical structure/property relationships in calcified tissues and tissue/material interfaces, Mater. Sci. Eng. C **27**, 450–468 (2007)

32.8 J.F.V. Vincent: *Structural Biomaterials* (Princeton Univ. Press, Princeton 1991)

32.9 H. Gao, B. Ji, I.L. Jager, E. Arzt, P. Fratzl: Materials become insensitive to flaws at the nanoscale: lessons from nature, Proc. Natl. Acad. Sci., Vol. 100 (2003) pp. 5597–2600

32.10 E. Baer, A. Hiltner, A.R. Morgan: Biological and synthetic hierarchical composites, Phys. Today **45**, 60–67 (1992)

32.11 A.V. Srinivasan, G.K. Haritos, F.L. Hedberg: Biomimetics: advancing man-made materials through guidance from nature, Appl. Mech. Rev. **44**, 463–481 (1991)

32.12 G. Yang, J. Kabel, B. Van Rietbergen, A. Odgaard, R. Huiskes, S. Cowin: The anisotropic Hooke's law for cancellous bone and wood, J. Elast. **53**, 125–146 (1999)

32.13 P.K. Zysset, X.E. Guo, C.E. Hoffler, K.E. Moore, S.A. Goldstein: Elastic modulus and hardness of cortical and trabecular bone lamellae measured by nanoindentation in the human femur, J. Biomech. **32**, 1005–1012 (1999)

32.14 C. Rubin, A.S. Turner, S. Bain, C. Mallinckrodt, K. McLeod: Low mechanical signals strengthen bones, Nature **412**, 603–604 (2001)

32.15 P.K.D.V. Yarlagadda, M. Chandrasekharan, J.Y.M. Shyan: Recent advances and current developments in tissue scaffolding, Bio-med. Mater. Eng. **15**, 159–177 (2005)

32.16 S. Saidpour: Assessment of carbon fibre composite fracture fixation plate using finite element analysis, J. Biomed. Eng. **34**, 1157–1163 (2006)

32.17 J.S. Temenoff, A.G. Mikos: Injectable biodegradable materials for orthopaedic tissue engineering, Biomaterials **21**, 2405–2412 (2000)

32.18 R. Pietrabissa, V. Qauglini, T. Villa: Experimental methods in testing of tissues and implants, Meccanica **37**, 477–488 (2002)

32.19 G. Heimke, P. Griss: Ceramic implant materials, Med. Biol. Eng. Comput. **18**, 503–510 (1980)

32.20 J.B. Park, R.S. Lakes: *Biomaterials* (Plenum, New York 1992)

32.21 E.N. Brown, M.L. Peterson, K.J. Grande-Allen: Biological systems and materials: a review of the field of biomechanics and the role of the society for experimental mechanics, Exp. Techn. **2**, 21–29 (2006)

32.22 A.J. van der Pijl, W. Swieszkowski, H.E.N. Bersee: Design of a wear simulator for in vitro should prostheses testing, Exp. Techn. **5**, 45–48 (2004)

32.23 A.J. Rapoff, W.M. Johnson, S. Venkataraman: Transverse plane shear test fixture for total knee system, Exp. Techn. **3**, 37–39 (2003)

32.24 R. Oosterom, H.E.N. Bersee: Force controlled fatigue testing of shoulder prostheses, Exp. Techn. **5**, 33–37 (2004)

32.25 V.L. Roberts: Strain-gage techniques in biomechanics, Exp. Mech. **6**, 19A–22A (1966)

32.26 L. Cristofolini, M. Viceconti: Comparison of uniaxial and triaxial rosette gages for strain measurement on the femur, Exp. Mech. **37**, 350–354 (1997)

32.27 C. Franck, S. Hong, S.A. Maskarinee, D.A. Tirrell, G. Ravichandran: Three-dimensional full-field measurements of large deformations in soft materials using confocal microscopy and digital volume correlation, Exp. Mech. **47**(3), 427–438 (2007)

32.28 A.B. Liggins, W.R. Hardie, J.B. Finlay: The spatial and pressure resolution of Fuji pressure-sensitive films, Exp. Mech. **35**, 166–173 (1995)

32.29 R.D. Peindl, M.E. Harrow, P.M. Connor, D.M. Banks, D.R. D'Alessandro: Photoelastic stress freezing analysis of total shoulder replacement systems, Exp. Mech. **44**, 228–234 (2004)

32.30 R.J. Nikolai, J.W. Schweiker: Investigation of root-periodontium interface stresses and displacements for orthodontic application, Exp. Mech. **12**, 406–413 (1972)

32.31 H. Vaillancourt, D. McCammond, R.M. Pilliar: Validation of a nonlinear two-dimensional interface element for finite-element analysis, Exp. Mech. **36**, 49–54 (1996)

32.32 J.F. Dias Rodrigues, H. Lopes, F.Q. de Melo, J.A. Simões: Experimental modal analysis of a synthetic composite femur, Exp. Mech. **44**, 29–32 (2004)

32.33 J.-J. Ryu, V. Daval, P. Shrotriya: Onset of surface damage in modular orthopedic implants: influence of normal contact loading and stress-assisted dissolution, Exp. Mech. **47**(3), 395–403 (2007)

32.34 N.J. Hallab, J.J. Jacobs: Orthopaedic implant fretting corrosion, Corros. Rev. **21**, 183–213 (2003)

32.35 S.L.-Y. Woo, K.S. Lothringer, W.H. Akeson, R.D. Coutts, Y.K. Woo, B.R. Simon, M.A. Gomez: Less rigid internal fixation plates: historical perspectives and new concepts, J. Orthop. Res. **1**, 431–449 (2005)

32.36 I.M. Peterson, A. Pajares, B.R. Lawn, V.P. Thompson, E.D. Rekow: Mechanical characterization of dental ceramics by Hertzian contacts, J. Dent. Res. **77**, 589–602 (1998)

32.37 K. Tayton, C. Johnson-Nurs, B. McKibbin, J. Bradley, G. Hastings: The use of semi-rigid carbon-fibre reinforced plates for fixation of human factors, J. Bone Joint Surg. **64-B**, 105–111 (1982)

32.38 D.A. Rikli, R. Curtis, C. Shilling, J. Goldhahn: The potential of bioresorbable plates and screws in distal radius fracture fixation, Injury **33**, B77–B83 (2002)

32.39 A. Foux, A. Yeadon, H.K. Uhthoff: Improved fracture healing with less rigid plates: a biomechanical study in dogs, Clin. Orthop. Rel. Res. **339**, 232–245 (1995)

32.40 S. Saha, S. Pal: Mechanical properties of bone cement: a review, J. Biomed. Mater. Res. **18**, 435–462 (2004)

32.41 J.L. Tan, J. Tien, D.M. Pirone, D.S. Gray, K. Bhadriraju, C.S. Chen: Cells lying on a bed of microneedles: an approach to isolate mechanical force, Appl. Phys. Sci. **100**, 1484–1489 (2003)

32.42 M.F. Ashby: The mechanical properties of cellular solids, Metall. Trans. **14A**, 1755–1769 (1983)

32.43 W.F. Brace: Permeability from resistivity and pore shape, J. Geophys. Res. **82**, 3343–3349 (1977)

32.44 C.M. Agrawal, R.B. Ray: Biodegradable polymeric scaffolds for muscoskeletal tissue engineering, J. Biomed. Mater. Res. **55**, 141–150 (2001)

32.45 F.J. O'Brien, B.A. Harley, I.V. Yannas, L.J. Gibson: The effect of pore size on cell adhesion in collagen-GAG scaffolds, Biomaterials **26**, 433–441 (2005)

32.46 I.V. Yannas, E. Lee, D.P. Orgill, E.M. Skrabut, G.F. Murphy: Synthesis and characterization of a model extracellular-matrix that induces partial regeneration of adult mammalian skin, Proc. Natl. Acad. Sci., Vol. 86 (1989) pp. 933–937

32.47 D. Bynum Jr., W.B. Ledtter, C.L. Boyd, D.R. Ray: Holding characteristics of fasteners in bone, Exp. Mech. **11**, 363–369 (1971)

32.48 P.M. Talaia, A. Ramos, I. Abe, M.W. Schiller, P. Lopes, R.N. Nogueira, J.L. Pinto, R. Claramunt, J.A. Simões: Plated and intact femur strains in fracture fixation using fiber bragg gratings and strain gauges, Exp. Mech. **47**(3), 355–363 (2007)

32.49 S. Inceoglu, R.F. McLain, S. Cayli, C. Kilincer, L. Ferrara: A new screw pullout test incorporating the effects of stress relaxation, Exp. Techn. **4**, 19–21 (2005)

32.50 D.D. Wright-Charlesworth, D.M. Miller, I. Miskioglu, J.A. King: Nanoindentation of injection molded PLA and self-reinforced composite PLA after in vitro conditioning for three months, J. Biomed. Mater. Res. **74A**, 388–396 (2005)

32.51 D.D. Wright-Charlesworth, W.J. Peers, I. Miskioglu, L.L. Loo: Nanomechanical properties of self-reinforced composite poly(methylmethacrylate) as a function of processing temperature, J. Biomed. Mater. Res. **74A**, 306–314 (2005)

32.52 F. Flueckiger, H. Sternthal, G.E. Klein, M. Aschauer, D. Szolar, G. Kleinhappl: Strength, elasticity, plasticity of expandable metal stents: in vitro studies with three types of stress, J. Vasc. Interv. Radiol. **5**, 745–750 (1994)

32.53 J. Hanus, J. Zahora: Measurement and comparison of mechanical properties of Nitinol stents, Phys. Scr. **T118**, 264–267 (2005)

32.54 K.E. Perry, C. Kugler: Non-zero mean fatigue testing of NiTi, Exp. Techn. **1**, 37–38 (2002)

32.55 S.C. Schrader, R. Bear: Evaluation of the compressive mechanical properties of endoluminal metal stents, Cathet. Cardiovasc. Diagn. **44**, 179–187 (1998)

32.56 J.P. Nuutinen, C. Clerc, P. Tormala: Theoretical and experimental evaluation of the radial force of self-expanding braided bioabsorbable stents, J. Biomater. Sci. Polym. Ed. **14**, 677–687 (2003)

32.57 R. Wang, K. Ravi-Chandar: Mechanical response of a metallic aortic stent – part II: pressure-diameter relationship, J. Appl. Mech. **71**, 697–705 (2004)

32.58 R. Wang, K. Ravi-Chandar: Mechanical response of a metallic aortic stent- part II: A beam-on-elastic foundation model, J. Appl. Mech. **71**, 706–712 (2004)

32.59 D.S. Goldin, S.L. Venneri, A.K. Noor: The great out of the small, Mech. Eng. **122**, 70–79 (2000)

32.60 C. Greiner, A.D. Campo, E. Arzt: Adhesion of bioinspired micropatterned surfaces: effects of pillar radius, aspect ratio, and preload, Langmuir **23**, 3495–3502 (2007)

32.61 E.N. Brown, S.R. White, N.R. Sottos: Microcapsule induced toughening in a self-healing polymer composite, J. Mater. Sci. **39**, 1703–1710 (2004)

32.62 C.E. Flynn, S.W. Lee, B.R. Peelle, A.M. Belcher: Viruses as vehicles for growth, organization and assembly of materials, Acta Mater. **51**, 5867–5880 (2003)

32.63 S. Suresh, A. Mortenson: *Fundamentals of Functionally Graded Materials* (Institute of Materials, London 1998)

32.64 S. Amada, Y. Ichikawa, T. Munekata, Y. Nagese, H. Shimizu: Fiber texture and mechanical graded structure of bamboo, Compos. B **28B**, 13–20 (1997)

32.65 P. Kreuz, W. Arnold, A.B. Kesel: Acoustic microscopic analysis of the biological structure of insect wing membranes with emphasis on their waxy surface, Ann. Biomed. Eng. **29**, 1054–1058 (2001)

32.66 M. Niino, S. Maeda: Recent development status of functionally gradient materials, ISIJ Int. **30**, 699–703 (1990)

32.67 P.M. Bendsoe, N. Kikuchi: Generating optimal topologies in structural design using a homogenization method, Comput. Methods Appl. Mech. Eng. **71**, 197–224 (1998)

32.68 T. Hirano, T. Yamada, J. Teraki, A. Kumakawa, M. Niino, K. Wakashima: Improvement in design accuracy of functionally gradient material for space plane applications, Proc. 7th Int. Symp. Space Technol. Sci. (Tokyo 1990)

32.69 C. Schiller, M. Siedler, F. Peters, M. Epple: Functionally graded materials of biodegradable polyesters and bone-like calcium phosphates for bone replacement, Ceram. Trans. **114**, 97–108 (2001)

32.70 A.J. Markworth, K.S. Ramesh, W.P. Parks Jr.: Review: Modelling studies applied to functionally graded materials, J. Mater. Sci. **30**, 2183–2193 (1995)

32.71 N. Noda: Thermal stresses in functionally graded materials, J. Thermal Stresses **22**, 477–512 (1999)

32.72 C.E. Rousseau, H.V. Tippur: Influence of elastic gradient profiles on dynamically loaded functionally graded materials: Cracks along the gradient, Int. J. Solids Struct. **38**, 7839–7856 (2001)

32.73 A.M. Afsar, H. Sekine: Optimum material distribution for prescribed apparent fracture toughness in thick-walled FGM circular pipes, Int. J. Press. Vessels Piping **78**, 471–484 (2001)

32.74 T.J. Chung, A. Neubrand, J. Rodel: Effect of residual stress on the fracture toughness of Al_2O_3/Al gradient materials, Euro Ceram. VII, PT1–3 **206**, 965–968 (2002)

32.75 K.S. Ravichandran: Thermal residual stresses in a functionally graded material system, Mater. Sci. Eng. A **201**, 269–276 (1995)

32.76 Y.D. Lee, F. Erdogan: Residual/thermal stresses in FGM and laminated thermal barrier coatings, Int. J. Fract. **69**, 145–165 (1994)

32.77 S. Suresh, A.E. Giannakopoulos, M. Olsson: Elastoplastic analysis of thermal cycling: layered mater-

ials with sharp interface, J. Mech. Phys. Solids **42**, 979–1018 (1994)
32.78 A.E. Giannakopoulos, S. Suresh, M. Olsson: Elastoplastic analysis of thermal cycling: layered materials with compositional gradients, J. Mech. Phys. Solids **43**, 1335–1354 (1995)
32.79 M. Finot, S. Suresh: Small and large deformation of thick and thin-film multi-layers: effects of layer geometry, plasticity and compositional gradients, J. Mech. Phys. Solids **44**, 683–721 (1996)
32.80 Q.R. Hou, J. Gao: Thermal stress relaxation by a composition-graded intermediate layer, Mod. Phys. Lett. **14**, 685–692 (2000)
32.81 B.H. Rabin, R.L. Williamson, H.A. Bruck, X.L. Wang, T.R. Watkins, D.R. Clarke: Residual strains in an Al_2O_3-Ni joint bonded with a composite interlayer: experimental measurements and FEM analysis, J. Am. Ceram. Soc. **81**, 1541–1549 (1998)
32.82 B.H. Rabin, R.J. Heaps: Powder processing of $Ni-Al_2O_3$ FGM, Ceram. Trans. **34**, 173–180 (1993)
32.83 R.L. Williamson, B.H. Rabin, J.T. Drake: Finite element analysis of thermal residual stresses at graded ceramic–metal interfaces part I: model description and geometrical effects, J. Appl. Phys. **74**, 1310–1320 (1993)
32.84 J.W. Eischen: Fracture of nonhomogeneous materials, Int. J. Fract. **34**, 3–22 (1987)
32.85 G. Anlas, M.H. Santare, J. Lambros: Numerical calculation of stress intensity factors in functionally graded materials, Int. J. Fract. **104**, 131–143 (2000)
32.86 M. Dao, P. Gu, A. Maewal, R.J. Asaro: A micromechanical study of residual stresses in functionally graded materials, Acta Mater. **45**, 3265–3276 (1997)
32.87 M.H. Santare, J. Lambros: Use of graded finite elements to model the behavior of nonhomogeneous materials, J. Appl. Mech.-Trans. ASME **67**, 819–822 (2000)
32.88 E.S.C. Chin: Army focused research team on functionally graded armor composites, Mater. Sci. Eng. A **259**, 155–161 (1999)
32.89 H.A. Bruck: A one-dimensional model for designing functionally graded materials to attenuate stress waves, Int. J. Solids Struct. **37**, 6383–6395 (2000)
32.90 X. Han, G.R. Liu, K.Y. Lam: Transient waves in plates of functionally graded materials, Int. J. Numer. Methods Eng. **52**, 851–865 (2001)
32.91 G.R. Liu, X. Han, Y.G. Xu, K.Y. Lam: Material characterization of functionally graded material by means of elastic waves and a progressive-learning neural network, Compos. Sci. Technol. **61**, 1401–1411 (2001)
32.92 J.E. Lefebvre, V. Zhang, J. Gazalet, T. Gryba, V. Sadaune: Acoustic wave propagation in continuous functionally graded plates: an extension of the legendre polynomial approach, IEEE Trans. Ultrason. Ferroelectr. Frequ. Control **48**, 1332–1340 (2001)
32.93 Y. Li, K.T. Ramesh, E.S.C. Chin: Dynamic characterization of layered and graded structures under impulsive loading, Int. J. Solids Struct. **38**, 6045–6061 (2001)
32.94 P.R. Marur, H.V. Tippur: Evaluation of mechanical properties of functionally graded materials, J. Test. Eval. **26**, 539–545 (1998)
32.95 V. Parameswaram, A. Shukla: Crack tip stress fields for dynamic fracture in functionally gradient materials, Mech. Mater. **31**, 579–596 (1999)
32.96 S.S.V. Kandula, J. Abanto-bueno, P.H. Geubelle, J. Lambros: Cohesive modeling of dynamic fracture in functionally graded materials, Int. J. Fract. **132**, 275–296 (2005)
32.97 J. Abanto-Bueno, J. Lambros: Experimental determination of cohesive failure properties of a photodegradable copolymer, Exp. Mech. **45**, 144–152 (2005)
32.98 A. Shukla, N. Jain, R. Chona: Dynamic fracture studies in functionally graded materials, Strain **43**, 76–95 (2007)
32.99 V. Parameswaran, A. Shukla: Dynamic fracture of a functionally gradient material having discrete property variation, J. Mater. Sci. **33**, 3303–3311 (1998)
32.100 C.-E. Rousseau, H.V. Tippur: Dynamic fracture of compositionally graded materials with cracks along the elastic gradient: experiments and analysis, Mech. Mater. **33**, 403–421 (2001)
32.101 J. Huang, A.J. Rapoff: Optimization design of plates with holes by mimicking bones through nonaxisymmetric functionally graded materials, Proc. Inst. Mech. Eng. Part L – J. Mater. – Des. Appl., Vol. 217 (2003) pp. 23–27
32.102 B. Garita, A. Rapoff: Biomimetic designs from bone, Exp. Techn. **1**, 36–39 (2003)
32.103 N. Gotzen, A.R. Cross, P.G. Ifju, A.J. Rapoff: Understanding stress concentration about a nutrient foramen, J. Biomech. **36**, 1511–1521 (2003)
32.104 R.M. Gouker, S.K. Gupta, H.A. Bruck, T. Holzchuh: Manufacturing of multi-material compliant mechanisms using multi-material molding, Int. J. Adv. Manuf. Technol. **28**, 1–27 (2006)
32.105 L.R. Xu, H. Kuai, S. Sengupta: Dissimilar material joints with and without free-edge stress singularities part I: a biologically inspired design, Exp. Mech. **44**, 608–615 (2004)
32.106 R.S. Trask, H.R. Williams, I.P. Bond: Self-healing polymer composites: mimicking nature to enhance performance, Bioinspir. Biomimetics **2**, P1–P9 (2007)
32.107 J.Y. Lee, G.A. Buxton, A.C. Balazs: Using nanoparticles to create self-healing composites, J. Chem. Phys. **121**, 5531–5540 (2004)
32.108 S.M. Bleay, C.B. Loader, V.J. Hayes, L. Humberstone, P.T. Curtis: A smart repair system for polymer matrix composites, Compos. A **32**, 1767–1776 (2001)

32.109 X.X. Chen, M.A. Dam, K. Ono, A. Mal, H.B. Shen, S.R. Nutt, K. Sheran, F. Wudl: A thermally remendable cross-linked polymeric material, Science **295**, 1698–1702 (2002)

32.110 C.M. Chung, Y.S. Roh, S.Y. Cho, J.G. Kim: Crack healing in polymeric materials via photochemical [2+2] cycloaddition, Chem. Mater. **16**, 3982–3984 (2004)

32.111 E.N. Brown, N.R. Sottos, S.R. White: Fracture testing of a self-healing polymer composite, Exp. Mech. **42**, 372–379 (2002)

32.112 V. Birman: Stability of functionally graded shape memory alloy sandwich panels, Smart Mater. Struct. **6**, 278–286 (1997)

32.113 K. Ho, G.P. Carman: Sputter deposition of NiTi thin film shape memory alloy using a heated target, Thin Solid Films **370**, 18–29 (2000)

32.114 H.A. Bruck, C.L. Moore, T. Valentine: Characterization and modeling of bending actuation in polyurethanes with graded distributions of one-way shape memory alloy wires, Exp. Mech. **44**, 62–70 (2004)

32.115 H.A. Bruck, C.L. Moore, T. Valentine: Repeatable bending actuation in polyurethanes using opposing embedded one-way shape memory alloy wires exhibiting large strain recovery, Smart Mater. Struct. **11**, 509–518 (2002)

32.116 C.L. Moore, H.A. Bruck: A fundamental investigation into large strain recovery of one-way shape memory alloy wires embedded in flexible polyurethanes, Smart Mater. Struct. **11**, 130–139 (2002)

32.117 E. Fukada, I. Yasuda: On the piezoelectric effect of bone, J. Phys. Soc. Jpn. **12**, 1158–1162 (1957)

32.118 G. Song, V. Sethi, H.N. Li: Vibration control of civil structures using piezoceramic smart materials: a review, Eng. Struct. **28**, 1513–1524 (2006)

32.119 J. Sirohi, I. Chopra: Fundamental understanding of piezoelectric strain sensors, J. Intell. Mater. Syst. Struct. **11**, 246–257 (2000)

32.120 K.D. Rolt: History of the flextensional electroacoustic transducer, J. Acoust. Soc. Am. **87**, 1340–1349 (1990)

32.121 B.J. Pokines, E. Garcia: A smart material microamplification mechanisms fabricated using LIGA, Smart Mater. Struct. **7**, 105–112 (1998)

32.122 A. Garg, D.C. Agrawal: Effect of rare earth (Er, Gd, Eu, Nd, and La) and bismuth additives on the mechanical and piezoelectric properties of lead zirconate titanate ceramics, Mater. Sci. Eng. B **86**, 134–143 (2001)

32.123 R. Rajapakse, X. Zeng: Toughening of conducting cracks due to domain switching, Acta Mater. **49**, 877–885 (2001)

32.124 K. Takagi, J.F. Li, S. Yokoyama, R. Watanabe: Fabrication and evaluation of PZT/Pt piezoelectric composites and functionally graded actuators, J. Eur. Ceram. Soc. **23**, 1577–1583 (2003)

32.125 J. Tyson, T. Schmidt, K. Galanulis: Biomechanics deformation and strain measurements with 3D image correlation phtogrammetry, Exp. Techn. **5**, 39–42 (2002)

32.126 Y. Bar-Cohen: Electroactive polymers as artificial muscles: reality and challenges, Proc. 42nd AIAA Struct. Struct. Dyn. Mater. Conf. (Seattle 2001)

32.127 G. Mayer, M. Sarikaya: Rigid biological composite materials: structural examples for biomimetic design, Exp. Mech. **42**, 394–403 (2002)

32.128 G. Mayer: Rigid biological systems as models for synthetic composites, Science **310**, 1144–1147 (2005)

32.129 F. Barthelat, H. Espinosa: An experimental investigation of deformation and fracture of nacre-mother of pearl, Exp. Mech. **47**(3), 311–324 (2007)

32.130 L.S. Gyger Jr., P. Kulkarni, H.A. Bruck, S.K. Gupta, O.C. Wilson Jr.: Replamineform inspired bones structures (RIBS) using multi-piece molds and advanced ceramic gelcasting technology, Mater. Sci. Eng. C **27**, 646–653 (2007)

32.131 A.P. Jackson, J.F.V. Vincent, R.M. Turner: The mechanical design of nacre, Proc. R. Soc. London: Ser. B. Biol. Sci, Vol. 234 (1988) pp. 415–440

32.132 S.L. Tracy, H.M. Jennings: The growth of self-aligned calcium carbonate precipitates on inorganic substrates, J. Mater. Sci. **33**, 4075–4077 (1998)

32.133 E.W. White: Biomaterials innovation: it's a long road to the operating room, Mater. Res. Innov. **1**, 57–63 (1997)

32.134 C. Müller-Mai, C. Voigt, S.R. de Almeida Reis, H. Herbst, U.M. Gross: Substitution of natural coral by cortical bone and bone marrow in the rat femur. 2. SEM, TEM, and in situ hybridization, J. Mater. Sci. Mater. Med. **7**, 479–488 (1996)

32.135 E.W. White, J.N. Weber, D.M. Roy, E.L. Owen, R.T. Chiroff, R.A. White: Replamineform porous biomaterials for hard tissue implant applications, J. Biomed. Mater. Res. **9**, 23–27 (1975)

32.136 R.T. Chiroff, E.W. White, J.N. Weber, D.M. Roy: Restoration of articular surfaces overlying replamineform porous biomaterials, J. Biomed. Mater. Res. **9**, 29 (1975)

32.137 S. Deville, E. Saiz, R.K. Nalla, A.P. Tomsia: Freezing as a path to build complex composites, Science **311**, 515–518 (2006)