Mechanical Principles of Biological Nanocomposites

Baohua Ji[1] and Huajian Gao[2,*]

[1]Biomechanics and Biomaterials Laboratory, Department of Applied Mechanics, Beijing Institute of Technology, Beijing 100081, P.R. China; email: bhji@bit.edu.cn

[2]Division of Engineering, Brown University, Providence, Rhode Island 02912; email: Huajian_Gao@brown.edu

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*Corresponding author.

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nanostructure, flaw tolerance, hierarchical structure, biomimicking materials, strength, toughness

Abstract

Biological nanocomposites, such as bone, tooth, shell, and wood, exhibit exceptional mechanical properties. Much recent effort has been directed at exploring the basic mechanical principles behind the microstructures of these natural materials to provide guidelines for the development of novel man-made nanocomposites. This article reviews some of the recent studies on mechanical properties of biological nanocomposites, including their stiffness, strength, toughness, interface properties, and elastic stability. The discussion is focused on the mechanical principles of biological nanocomposites, including the generic nanostructure of hard-mineral crystals embedded in a soft protein matrix, the flaw-tolerant design of the hard phase, the role of the soft matrix, the hybrid interface between protein and mineral, and the structural hierarchy. The review concludes with some discussion of and outlook on the development of biomimicking synthetic materials guided by the principles found in biological nanocomposites.

INTRODUCTION

Nature is a master of the synthesis and use of nanocomposites. With natural minerals and polymers that exhibit typically poor macroscale mechanical properties, strong composites, such as bone, tooth, shell, and wood, are made by mixing two or more phases, at least one of which is in the nanometer size range (1–16). Compared with their constituent phases, biological nanocomposites can achieve orders-of-magnitude improvements in strength and toughness; this mechanical-property amplification occurs often in a nonadditive manner that goes beyond the simple rules of mixture (17). Moreover, biological materials exhibit ordered, complex, hierarchical microstructures that current man-made composite materials cannot achieve.

The mechanical properties of biological materials have been a focal point of extensive studies over the past decades, leading to the emergence of a new research field that intimately connects biology, chemistry, and materials science (15, 18–20). Significant advances have been made in a number of disciplines and research areas, with phenomena and properties ranging from atomistic scales to molecular scales up to continuum scales. Of particular interest to the present review are those studies aimed at understanding the basic mechanical principles of biological nanocomposites through experimental, theoretical, and numerical investigations, as well as the development of biomimicking materials guided by the principles of biological nanocomposites.

Biological materials exhibit many levels of hierarchical structures from microscopic to macroscopic length scales. For instance, seashells have two to three orders of lamellar structure (1–3, 5), and bone has up to seven orders of hierarchy (6, 9, 10). In spite of these complex hierarchical structures, the smallest building blocks in biological materials—such as nanometer-sized hard inclusions embedded in a soft protein matrix in a staggered alignment pattern—are generally on the nanometer length scale (20, 21). For instance, tooth enamel (**Figure 1*a*,*d***) is made of long, more or less needle-like crystals approximately $15 \sim 20$ nm thick embedded in a soft matrix. Bone (**Figure 1*b***) has a nanostructure (**Figure 1*e***, mineralized fibrils) consisting of mineral crystals with a thickness of approximately a few nanometers staggered in a collagen matrix (4, 6, 9). Dentin (**Figure 1*a*,*e***) is a calcified tissue somewhat similar to bone, in which the collagen-rich organic matrix is reinforced by calcium phosphate mineral crystals (8, 11, 12). Nacre (**Figure 1*c***) has a brick-and-mortar lamellar structure (**Figure 1*f***), in which the thickness of the aragonite bricks is approximately a few hundred nanometers (1–3, 7). Wood is also a nanocomposite, with hard, crystalline cellulose fibrils dispersed in an amorphous hemicellulose-lignin matrix. For more details about the nano- to macroscale hierarchical structures of various biological nanocomposites, readers are referred to a number of existing review articles (9, 15, 22, 23).

During the past few decades, significant progress has been made on the understanding of the high toughness of biological nanocomposites from various points of view, including their hierarchical structures (3, 5, 24), the mechanical properties of protein on stress redistribution and energy dissipation (25–30), protein-mineral interface roughness (7), and the reduction of stress concentration at a crack tip (31). Recent investigations have also addressed (*a*) why the elementary structure of biocomposites is designed on the nanometer length scale (20, 32–34); (*b*) the strength of the protein-mineral interface (35, 36); and how toughness, elastic stability, and other mechanical properties are related to nanostructure (29, 33, 37–40). The reader is also referred to previous review articles (19, 23) discussing many theoretical and experimental studies on the mechanical properties of biological materials.

At present, most man-made polymer nanocomposites exhibit mechanical properties that fall short of their biological counterparts (16, 41). A critical challenge in current nanocomposite fabrication is the ability to realize materials that allow the exceptional mechanical properties

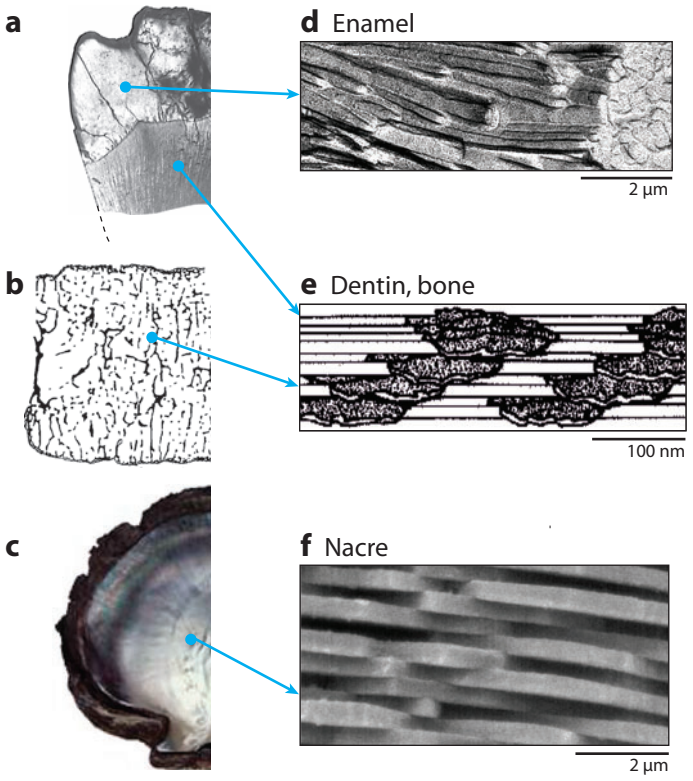

Figure 1

Hard biological tissues such as (*a*) tooth, (*b*) bone, and (*c*) nacre are nanocomposites with hard-mineral platelets in a soft protein matrix. (*d*) In the enamel of tooth, needle-like crystals 15∼20 nm in thickness are embedded in a soft matrix. (*e*) In cortical bone, plate-like mineral crystals a few nanometers thick are stacked in collagen molecules to form mineralized collagen fibrils. (*f*) In nacre, mineral platelets, which are a few hundreds of nanometers in thickness, interlock to form sheets that are stacked on top of each other in a staggered formation. From Reference 20 with permission.

(e.g., tensile strength, Young's modulus, and toughness) of nanoscale reinforcements to be manifested in the macroscale properties of bulk materials. More specifically, the deficiency in the properties of man-made nanocomposites is related largely to the difficulty of obtaining well-dispersed large-volume fractions of nanoparticles and to a lack of geometrical and structural control of a hierarchical structure. It is also difficult to realize an effective load transfer between the polymeric matrix and the nanoscale components, and we have an insufficient understanding of the interactions of the constituent phases at multiple length scales. Natural biological nanocomposites provide a useful platform to study possible strategies for resolving these issues.

In this review, we focus on some of the theoretical, numerical, and experimental studies on the basic mechanical principles of the hierarchical microstructures of biological nanocomposites. The purpose is to stimulate further discussions on possible ways to resolve the difficulties encountered in the syntheses of conventional nanocomposites and to suggest guidelines for the development of novel materials. For this purpose, we also include discussion on some of the recent efforts in the synthesis of biomimicking materials inspired by biological nanocomposites.

THEORETICAL MODELING, NUMERICAL SIMULATIONS, AND EXPERIMENTS ON MECHANICS OF BIOLOGICAL NANOCOMPOSITES

Aspect ratio: the length-to-thickness ratio of a mineral crystal

TSC model: tension-shear chain model

In this section, we discuss some recent progress in theoretical, numerical, and experimental studies of the mechanical properties of biological nanocomposites, especially those related closely to their microstructural features, such as the characteristic length scale and geometry (aspect ratio) of the hard phase (minerals), the mechanics of the soft phase (proteins), the interface between the hard phases and soft phases, and the hierarchical structure. We show that all these parameters should be considered to balance various mechanical properties, such as stiffness, strength, toughness, and elastic stability, in the design of a biomimicking nanocomposite. Bioinspiration not only should result from observations of natural structures but also should involve a thorough investigation and understanding of the structure-function relationships in biological materials. The characterization of biological materials within a rigorous materials science approach is crucial to the elucidation of the fundamental mechanical principles of these materials.

Tension-Shear Chain Model for the Generic Nanostructure of Biological Nanocomposites

On the basis of the prior work of Jäger & Fratzl (21) on the staggered arrangement of bone fibrils and experimental observations of various nanostructures of biological materials (4, 5), a tension-shear chain (TSC) model was proposed (20, 33) as a one-dimensional composite model for the nanostructure of biological nanocomposites, as illustrated in **Figure 2**. Mineral crystals possess large aspect ratios and are much harder than the soft protein matrix, and the tensile zone of the protein matrix near the ends of mineral crystals is therefore assumed to carry no mechanical

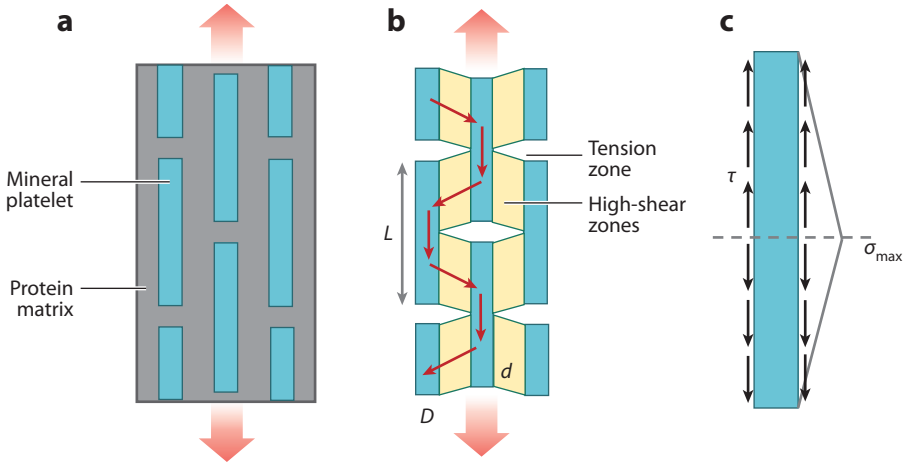

Figure 2

The tension-shear chain model for the nanostructure of biological nanocomposites. The yellow regions denote the higher-shear zones of protein, whereas the turquoise region denotes the mineral crystals. (*a*) Perfectly staggered mineral inclusions embedded in a protein matrix. (*b*) A tension-shear chain model of biocomposites in which the tensile regions of protein are eliminated to simplify the load transfer within the composite structure. (*c*) The free-body diagram of a mineral crystal. L and D are, respectively, the length and thickness of the mineral crystal, d is the thickness of protein, τ is the shear stress in the protein layer, and σ_{max} denotes the maximum tensile stress in the mineral. From Reference 20 with permission.

load. The load transfer is accomplished largely by the shearing of the protein matrix between the long sides of mineral platelets. Therefore, the composite model of **Figure 2*a*** can be simplified to that of **Figure 2*b***, which shows that the tensile zone of the soft phase is eliminated from the composite structure. Under an applied tensile stress, the mineral platelets carry most of the tensile load, whereas the protein matrix transfers load between mineral crystals via shear. The path of load transfer in the composite is thus simplified to a one-dimensional serial spring system consisting of mineral elements (tension) interspersed among protein elements (shear). The TSC can be regarded as the primary structure of biological materials (20, 33).

The large aspect ratio of the mineral platelets ensures that the force transferred between the platelets is distributed over a large shear region so as to cause only a small stress within the protein phase. If one assumes that the stress distribution along the length of the mineral crystals is linear, as shown in **Figure 2*c***, the maximum and average tensile stress in the mineral can be written as

$$\sigma_m = \rho\tau_p, \quad \bar{\sigma}_m = \rho\tau_p/2, \qquad 1.$$

where $\rho = L/h$ (L is the length and h the thickness of the platelets) is the aspect ratio of mineral and τ_p is the shear stress in the protein. These equations show that the tensile stress transferred by the protein has been amplified ρ times from its shear stress, i.e., a small shear in the matrix transfers to a large tension in the mineral. The shear stress in the protein can be regarded as uniform due to the large difference in elasticity between the matrix and the inclusions.

The TSC model predicts the effective modulus of the nanocomposite structure to be

$$\frac{1}{E} = \frac{4(1-\Phi)}{\mu_p \Phi^2 \rho^2} + \frac{1}{\Phi E_m}, \qquad 2.$$

where Φ is the volume fraction of mineral, μ_p is the shear modulus of protein, and E_m is the Young's modulus of mineral. The predictions of Equation 2 are consistent with micromechanical finite element method (FEM) calculations at various volume fractions and aspect ratios of mineral (33, 42).

Experimental studies have provided strong evidence for tension-shear microdeformation mechanisms in biological materials, e.g., mineral versus collagen protein matrix in the nanostructure of bone and dentin (15, 43), cellulose fibrils versus hemicellulose/lignin matrix in the nanostructure of wood (15, 44, 45), and mineral versus globular protein matrix in the nanostructure of seashells (7).

In addition, the TSC model suggests one type of structure-splitting design of biological materials (20, 36, 46). Due to the chain mode of load transfer, a local change in deformation and stress in mineral has been localized effectively without propagation. In other words, the local deformation and damage of an individual mineral are shielded from other parts of the material. The structure of the material is effectively split in the direction perpendicular to the loading direction, whereas integrity along the loading direction of the whole structure is kept by the TSC via the large aspect ratio and staggered alignment of mineral crystals (4, 5, 7, 21). A strong interface between protein and mineral is extremely crucial for the integrity of the chain structure.

In the TSC model, the shear stress distribution along the interface between mineral and protein is assumed to be uniform. If this restriction is relaxed, we return to a shear lag–type model (42, 47), in which the shear stress is not necessarily uniform. Kotha et al. (42) derived an analytical solution of shear stress along the interface in the biological nanostructure by using the shear lag model. The protein matrix between the platelets is modeled to be shear springs that transfer load from one platelet to the other. These investigators showed that the larger the modulus ratio between the mineral and protein and the volume fraction of the protein, the more uniform is the shear stress along the interface. Recently, Zuo & Wei (48) extended the shear lag model to analyze the

FEM: finite element method

Structure-splitting design: the splitting of a structure in one dimension while retaining integrity in other dimensions to enhance mechanical strength and flaw tolerance

Shear lag model: a model in mechanics of composites that centers on the load transfer from matrix to reinforcement via shear stress

effective modulus of higher-level hierarchical structures. Consistent with the findings by Kotha et al. (42), Zuo & Wei (48) showed that the shear stress at the protein-mineral interface becomes more uniform at higher modulus ratios.

More recently, Liu et al. (G. Liu, B. Ji, Y. Huang & K.C. Hwang, work in progress) adopted a more rigorous approach to analyze the stress and strain fields on the basis of two-dimensional elasticity theory. Using a perturbation approach, they solved not only the interfacial stress between protein and mineral but also the stress and strain fields in the biological nanostructure. Their analysis showed that the larger the Young's modulus ratio, the more uniform is the distribution of the shear stress along the interface. These researchers further demonstrated that the shear stress distribution along the interface is approximately uniform at an aspect ratio of approximately 40 for a Young's modulus ratio of approximately 1000, as in the case of real biological nanocomposites, which fully supports the basic assumption of the TSC model.

Griffith criterion: a criterion for crack propagation in brittle materials

Flaw tolerance: a material state in which preexisting cracks do not propagate even as the material is stretched to its limiting strength

VIB model: virtual-internal bond model

Critical Length Scale for Flaw-Tolerance Design

The experimental observation that the smallest structure in biological materials is almost always at the nanoscale has puzzled the materials science community for a very long time (3–6, 9, 21). Inspired by the convergent design of biological nanostructures, Gao et al. (20) addressed this fundamental question from the point of view of fracture mechanics. They found that the nanoscale dimension of a mineral may be the result of fracture strength optimization. When the mineral size exceeds this length scale, the fracture strength is sensitive to the structural size, and the material is sensitive to crack-like flaws and fails by flaw propagation under stress concentration at crack tips. As the mineral size drops below this length scale, the strength of a perfect mineral platelet is maintained in an optimal way in spite of defects, and the failure is governed by the theoretical strength of material rather than by the classical Griffith criterion of fracture propagation (see **Figure 3*a*,*b***). These arguments have prompted Gao et al. (20) to make the following postulate for biological materials: The nanometer size of mineral crystals in biocomposites may have been selected to ensure optimum fracture strength and maximum tolerance of flaws (for robustness). Nature finds this secret by evolution and hides material defects by designing the elementary structure of biomaterials at the nanoscale to achieve the level of robustness required for survival.

The flaw-tolerance idea was demonstrated numerically by finite element simulations based on the virtual-internal bond (VIB) model (49, 50), with results showing that the failure mode changes from brittle to ductile-like failure with decreasing dimension of the structure (20, 32, 33). At a critical length scale, the stress in the materials becomes nearly uniform at failure (see **Figure 3*c***).

The critical length for flaw tolerance was first derived from a platelet model with a semisnail surface crack (see **Figure 3*a*,*b***) using linear elastic fracture mechanics (20),

$$h^* \approx \alpha \frac{\gamma E_m}{\sigma_{th}^2}, \qquad 3.$$

where γ is the fracture energy, σ_{th} is the theoretical strength of material, and α is a constant. This length scale was also studied via a strip model with a semi-infinite crack using the VIB constitutive law (32). Later, the idea was further explained and demonstrated by using a strip model with center and edge cracks of arbitrary crack sizes (34). These analyses showed that once the dimension of the structure shrinks down to the critical size, the strip will not be sensitive to the existence of a crack-like flaw.

The flaw-tolerance idea was verified by atomistic simulations with more realistic material constitutive models by considering the discrete characteristics of materials at the nanoscale

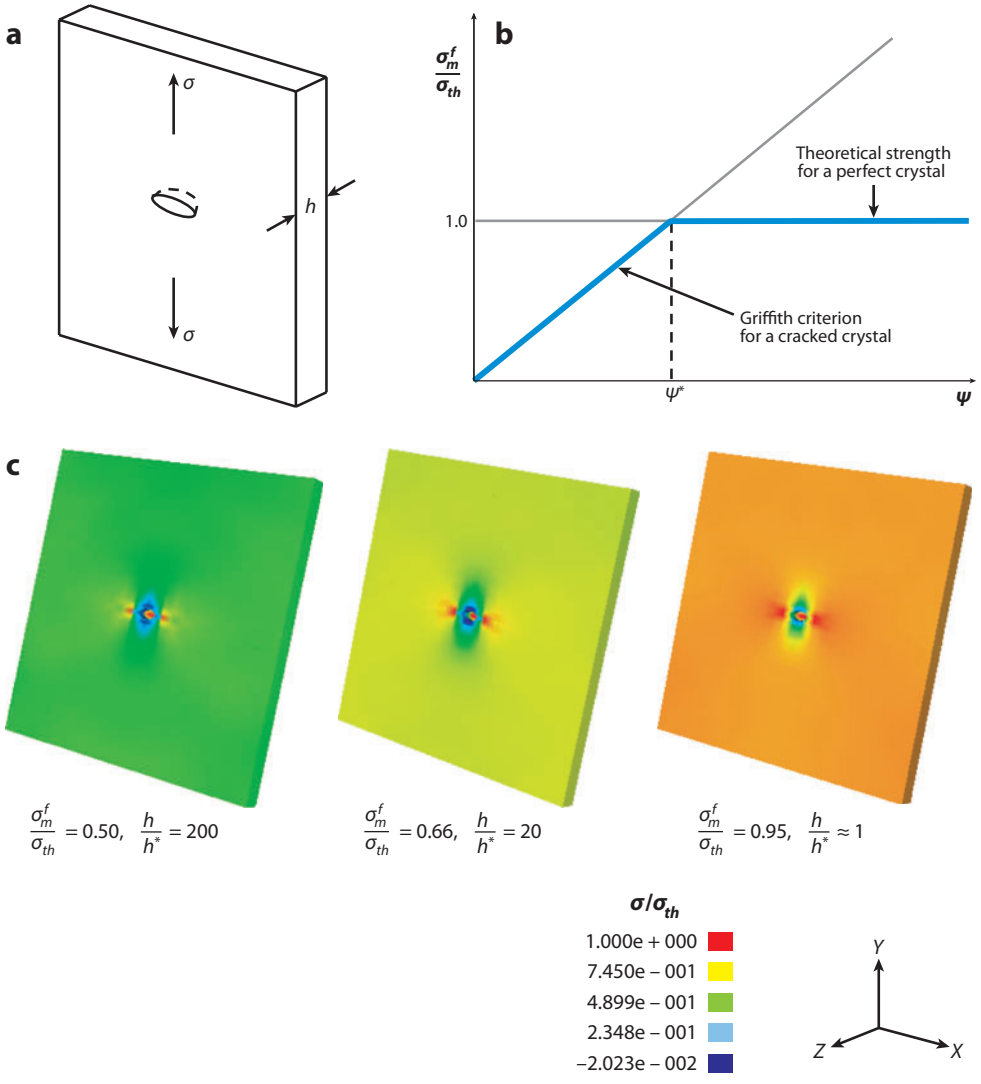

Figure 3

The length scale for optimized fracture strength in mineral platelets. (*a*) A schematic diagram of a mineral platelet with a thumbnail crack. (*b*) Comparison of the fracture strength of a cracked mineral platelet calculated from the Griffith criterion with the strength of a perfect, defect-free crystal. The comparison shows that there exists a transition in failure mechanism from Griffith crack propagation at macroscopic scales to uniform bond rupture near theoretical strength at nanoscopic scales. (*c*) The calculated stress field near the critical thickness for optimum fracture strength of a mineral platelet with a thumbnail surface crack. The color map of normal stress σ_{22} is obtained from a three-dimensional finite element method (FEM) simulation based on the virtual-internal bond (VIB) model. At large thicknesses ($h/h^* = 20, 200$), the stress concentration at the crack tip significantly reduces the fracture strength σ_m^f from the theoretical strength σ_{th}. Near the critical thickness h^*, the stress concentration vanishes, and the strength approaches the material's theoretical strength. From Reference 20 with permission.

Flaw insensitivity: a material state in which the classical singular-stress field is replaced by a uniform stress distribution near the crack tip

MD: molecular dynamics

Bone-like materials: materials possessing a microstructure similar to that of bone

(37, 39, 51). In addition, experimental studies also demonstrated the unique characteristics of failure in nanomaterials. For instance, Heinrich et al. (52) showed that the fracture strength of thin films made of brittle materials increases when the film thickness is reduced, and the fracture pattern becomes more uniform, exhibiting more ductile behavior. A brittle-to-ductile transition of fracture was also observed in experiments with nanoscale glassy materials (53).

A direct study on flaw tolerance was carried out recently by Kumar et al. (54), who performed in situ fracture experiments on single-edge-notched thin-film specimens inside a transmission electron microscope. In these experiments, freestanding thin-film specimens 50 μm long, 3.5 μm wide, and 100–125 nm thick were made from evaporated aluminum with an average grain size of 50 nm. Focused ion beam milling was used to cut U-shaped notches with tip radius of approximately 50 nm and length varying from 500 nm to 800 nm. The specimens failed at nominal stress range of 450–475 MPa, but the failure occurred far away from the notch at places with no apparent stress concentration, rather than at the notch tip, whereas in situ electron microscopy showed evidence of little or no plastic deformation at the notch tip. These experiments provide strong evidence of flaw-insensitive fracture at the nanoscale.

In contrast, studies on the mechanical strength of single-wall carbon nanotubes, SiC, and β-Si_3N_4 nanowhiskers (55–61) seem to show that nanodefects in these nearly perfect nanostructures can strongly affect their strength. A possible explanation is that the characteristic length of flaw tolerance for these materials is on the atomic scale so that the resulting nanostructures remain sensitive to the presence of nanodefects. The Young's modulus of various brittle materials can vary, depending on the atomic structure and the purity of materials, in quite a wide range between a few gigapascals to approximately 1000 GPa. For example, the Young's modulus of $CaCO_3$ can be as low as 50 GPa, and that of diamond can reach as high as 1200 GPa. Typical estimates of the theoretical strength can range between 1% and 10% of the Young's modulus. If we take $\gamma = 1$ J m^{-2}, $E = 50-1000$ GPa, and $\sigma_{th} = (1-10\%)E$, the characteristic length for flaw tolerance can be estimated to vary roughly in the range of 2 Å–400 nm. For instance, molecular dynamics (MD) simulations by Khare et al. (56) showed that the Young's modulus and theoretical strength of a nanotube are 1000 GPa and 120 GPa, respectively, resulting in a critical length of less than 2 Å, according to Equation 3.

MD simulations of Lu et al. (51) showed that the fracture strength of amorphous carbon nanospecimens is insensitive to initial cracks with diameters smaller than approximately 40 Å, i.e., the material exhibits flaw tolerance at the nanoscale. These researchers also found that the amorphous carbon nanospecimens fracture very differently from diamonds: (*a*) Failure is gradual instead of catastrophic, and (*b*) it is accompanied by void-like defect growth and coalescence. This fracture behavior appears to result from the structural disorder of amorphous carbon.

In summary, a critical length scale roughly at a nanometer size makes nominally brittle mineral crystals insensitive to crack-like flaws. The flaw tolerance corresponds to optimizing the strength of a material in the presence of crack-like flaws. This optimal state of a material can be achieved simply by size reduction. Load transfer in the staggered nanostructure of bone-like materials follows a TSC, in which the tension-loaded brittle mineral crystals are susceptible to brittle fracture. Through evolution, Nature finds a way to render the nominally brittle mineral crystals insensitive to crack-like flaws by confining them to the nanoscale. This design strategy, when combined with the extraordinary energy-absorbing capabilities of the soft protein matrix, leads to robust biocomposites.

Below, in a subsection entitled Roles of the Hierarchical Structure, we show that the flaw-tolerance property at the nanoscale can propagate to larger length scales by the use of hierarchical design (15, 19, 20).

Roles of the Aspect Ratio of Mineral Crystals

The aspect ratio of mineral crystals can affect the mechanical properties of the nanostructure of biological materials from many aspects, such as load transfer in the nanostructure, stiffness, strength, and elastic stability.

Stiffness is an important issue for the support and protection of biological materials. According to the TSC model, the large aspect ratio of mineral crystals in bone is designed to compensate for the softness of the matrix (33, 62). Equation 2 shows that an aspect ratio of 30–40 would provide a magnification of three orders of magnitude over the stiffness of protein and bring the stiffness of the composite close to that of the mineral. In contrast, according to Equation 1, the larger the aspect ratio, the larger will be the force that the protein matrix transfers to the mineral, causing high tensile stress in the mineral. Therefore, to secure the integrity of the mineral and the TSC structure, the aspect ratio cannot be infinitely large. There is an optimum value for the aspect ratio,

$$\rho^* = \frac{\sigma_m^f}{\tau_p^f} = \frac{1}{\tau_p^f}\sqrt{\frac{\alpha^2 \gamma E_m}{h}}, \qquad 4.$$

which balances the limiting strengths of the mineral and the mineral-protein interface.

The optimum aspect ratio should also consider the elastic buckling of the structure. Mineral crystals with large aspect ratio are susceptible to buckling under compressive loading due to their slender geometry. Ji et al. (40) addressed the stability of the nanostructure of biological materials under compression from the point of view that the mineral nanocrystals are stabilized by their staggered alignment in the surrounding protein matrix (see **Figure 4**). Ji et al. found a transition of buckling strength from an aspect ratio–dependent regime to a lower threshold value independent of the mineral aspect ratio. Typical values of the aspect ratio of nanocrystals of biological nanocomposites, such as bone and nacre, fall in this transition regime. The existence of a buckling strength independent of the detailed geometrical parameters of mineral is critically important from the point of view of structural robustness.

The previous study (40) did not consider the influence of neighboring platelets on the stability of a chosen mineral crystal. The coordinated buckling of mineral nanocrystals in biological materials is expected to be more complex due to the staggered alignment of mineral crystals. Studies on the coordinated buckling of mineral nanocrystals that consider their staggered alignment will be very helpful for understanding the compressive strength of the nanostructure of biological materials.

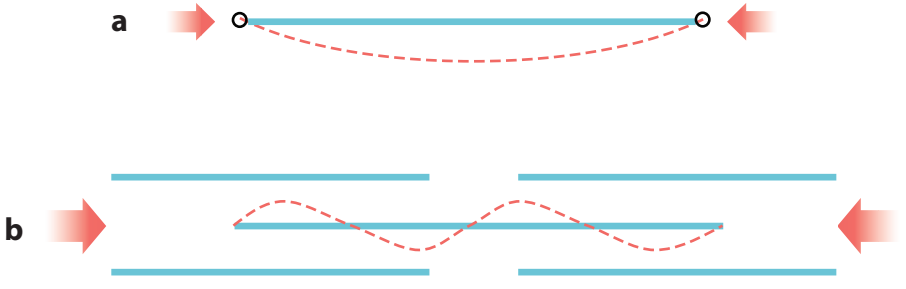

Figure 4

Elastic stability of slender mineral platelets in the nanostructure of biological nanocomposites. (*a*) Classical Euler buckling mode of a rod. (*b*) Schematic illustration of the buckling mode (*dotted red lines*) of mineral crystals in biocomposite nanostructures subject to the support of protein within the structural confinement by the neighboring crystals. The arrows denote compressive load. From Reference 40 with permission.

Dugdale model: a model of crack-tip field describing the transition from small-scale yielding to large-scale yielding

The optimal value of the aspect ratio can also be understood from the point of view of most effective load transfer in the nanocomposite structure of biological materials. Chen et al. (63) considered a generalized shear lag model and derived a characteristic length for stress transfer in a unit cell of biological nanocomposites. These investigators showed that the assumption of uniform shear stress in the TSC model of biocomposites is valid only if the length of the mineral platelets is below a characteristic value. Numerical simulations were carried out to confirm that stress transfer in biocomposites is most efficient when the length of the mineral platelets is equal to the characteristic length. In the elastic shear lag model, a characteristic length was derived as (63)

$$L_c = \sqrt{\frac{Eh_f h_m}{G_m}}. \qquad 5.$$

The axial stress of the fiber increases over the above characteristic distance and then saturates at a constant value; meanwhile, the shear stress along the interface decreases over the characteristic distance and vanishes beyond this distance. In this sense, the characteristic length L_c serves as a fundamental length scale over which stress transfer occurs along the interface. Considering large nonlinear deformation of the protein layer, Chen et al. (63) adopted a Dugdale shear cohesive zone model (generalized shear lag model) to show that there is a similar characteristic length for load transfer with uniform shear stress along the interface (63).

The generalized shear lag model (63) suggested that the concept of a characteristic length for stress transfer should be valid for biocomposites in which the matrix is strongly nonlinear in the presence of large deformation in the protein matrix. In this model, stress transfer in a unit cell of staggered biocomposites can be divided into two regimes. When the length of the mineral platelet is below the characteristic length L_c, the shear stress along the mineral-protein interface is uniform, and the apparent strength of the unit cell is proportional to the length of the mineral platelet. However, when the length of the mineral platelet is above L_c, the zone of stress transfer is saturated at L_c, and the apparent strength of the unit cell reaches a saturation value and no longer changes with the platelet length. Combining these two regimes shows that that stress transfer in the TSC is most efficient when the length of the mineral platelets is equal to L_c. In the TSC model of staggered biocomposites (20, 33, 36, 40, 62), the shear stress is assumed to be uniformly distributed along the interface. The generalized shear lag model (63) provides a rigorous justification for this assumption but shows that it is valid only when the platelets are shorter than a characteristic length that also depends on the modulus ratio between mineral and protein.

The Role of Protein Matrix

Protein plays a very important role in biological nanocomposites. First, it acts as a template for the nucleation and deposition of mineral crystals during the biomineralization process. The protein matrix governs the size and assembly of mineral crystals on several length scales. From a mechanical point of view, the protein matrix is crucial to the mechanical properties of biomaterials in many aspects, such as transferring load through shear deformation, dissipating energy through its large deformation and viscoelastic properties, and trapping cracks through matrix softness. For example, protein layers in nacre are approximately 20 nm thick and can act as an effective barrier for crack propagation (19, 23, 33, 39). In addition, protein layers in biocomposites can act as sources of microcrack nucleation to redistribute stress and to dissipate fracture energy (29).

To understand the underlying mechanisms of the unique mechanical properties of proteins, many studies have been conducted at the molecular level (30, 64–70). The hierarchical structures of proteins are ideally suited for absorbing and dissipating fracture energy. Proteins in nacre deform by gradual unfolding of their domain structures (30). This unfolding can require a large

amount of deformation because of deformation on the tertiary and secondary levels. Thus, the molecular structure of a protein is ideally suited for absorbing fracture energy. In bone, sacrificial bonds, in which Ca^{2+} ions cross-link peptides with negative electric charges (70), bind functional groups along different segments of collagen and along the protein-mineral interface, increasing the effective stress in proteins and leading to high fracture energy. The sacrificial bonds not only build up a large effective stress in protein but also allow protein deformation and interface slipping to occur simultaneously under similar stress levels, making it possible to engineer a maximum amount of fracture energy (33, 39).

Multiscale modeling: the calculation of material properties or system behavior on one level using information or models from different levels

In comparison with globular proteins in nacre, the molecular structure of collagen has been studied more intensively (64–69). Collagen plays an important role in many biological tissues, including tendon, bone, tooth, and cartilage (64). Severe mechanical tensile loading of collagen occurs under many physiological conditions, as in bone and joints (71, 72). Despite significant research efforts over the past few decades, the deformation mechanisms under mechanical load and, in particular, the relationship between those mechanisms and collagen's molecular and intermolecular properties are still not completely understood. For instance, single-molecule experiments on (type I) collagen molecules employing atomic force microscopy (73, 74) or optical tweezers (75, 76) showed that collagen molecules are flexible rather than rigid, rod-like molecules.

Buehler (64–68) and coworker (69) performed a series of joint atomistic and continuum studies on the mechanics of collagen molecules in the protein matrix in bone, dentin, tendon, etc. They developed a hierarchical multiscale modeling scheme based on atomistic and molecular simulations to describe the mechanical properties of collagen under large stretch, leading to permanent deformation or fracture. Their study suggests that the design of collagen fibrils in a staggered array of ultralong tropocollagen molecules provides large strength and energy dissipation during deformation. The mechanical behavior of a collagen fibril can be understood quantitatively in terms of two length scales, one characterizing when deformation changes from homogeneous intermolecular shear to propagation of slip pulses and the other characterizing when covalent bonds in the tropocollagen molecules start to fracture (64). A deformation map of collagen fibrils was suggested to show how the mechanical response is controlled by these two characteristic length scales. They also showed that cross-linking can enhance the strength of collagen fibrils.

Recently, Buehler and coworkers (77) also developed an atomistically informed continuum model of collagenous tissue. Results from full atomistic and molecular modeling are linked with a continuum theory of a fiber-reinforced composite, with handshaking from the fibril scale to the fiber and continuum scales in a hierarchical multiscale simulation approach (see **Figure 5**). This model can be used to study the continuum-level response of collagenous tissue as a function of the cross-link density, making a link between nanoscale collagen features and material properties at larger tissue scales.

Protein-Mineral Interface

In composite materials, the volume of interface between reinforcements and matrix increases as the size of the reinforcements decreases (16). For example, in the case of particles or fibers, the interface area per unit volume is inversely proportional to the characteristic dimension of the reinforcements (16, 78). For nanocomposites, as the mineral bits have nanoscale size, the volume fraction of the protein-mineral interface can be enormous. For example, in a raindrop-size volume of a nanocomposite, the area of interfacial region can be as large as a football field (79). This means that interfaces play more crucial roles in the overall mechanical properties of nanocomposites than in their conventional counterparts.

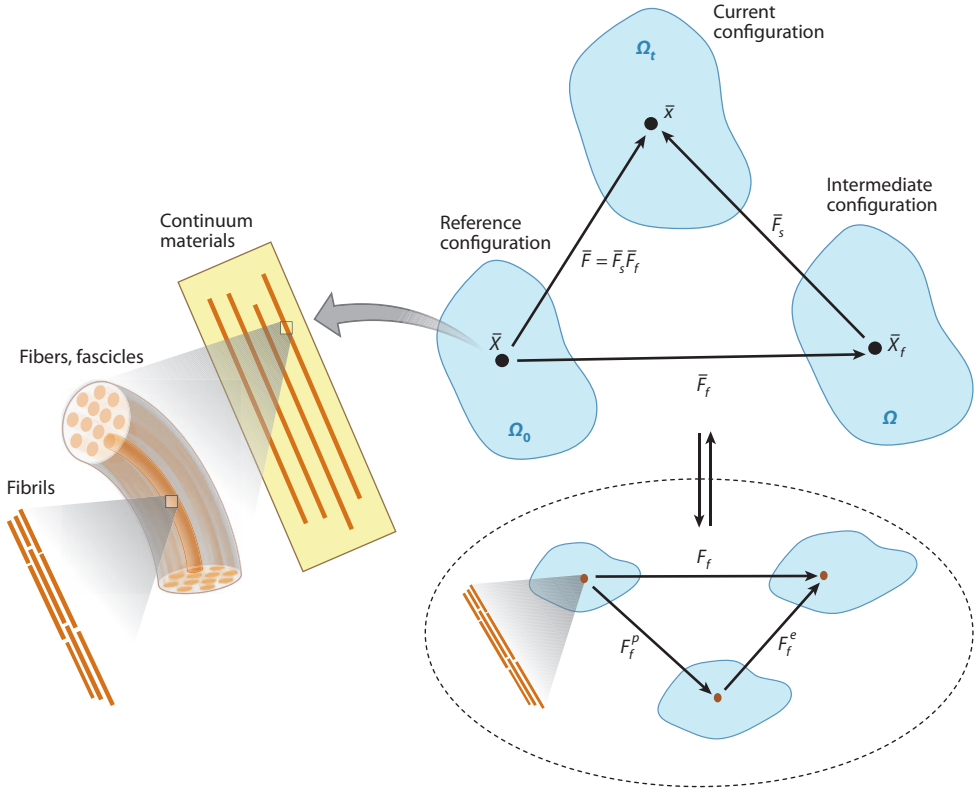

Figure 5

Multiscale modeling of the mechanics of a collagen fiber. Here, F is the deformation gradient defined by $\bar{F} = \partial \bar{x}(\bar{X}, t)/\partial \bar{X}$, where $\bar{x} = \bar{x}(\bar{X}, t)$ is the position of a material point in the current configuration Ω_t at time t, Ω_0 is the reference configuration, and Ω_f is the intermediate configuration. $\bar{F}_f$ is the local uniaxial deformation gradient, and $\bar{F}_s$ describes the remaining macroscopic shear deformation and rigid body rotation. F_f^e and F_f^p represent the uniaxial elastic and plastic deformation gradients, respectively. From Reference 77 with permission.

To understand the principles of interface design in biological nanocomposites (79–81), many studies have been performed to investigate the strength of the hybrid interface between protein and mineral. Several experiments have been performed to study the interaction between mineral and protein at small scales, with the suggestion that the static Coulomb force may dominate such an interaction in biological materials (82–85). Jäger et al. (86) recently studied the binding sites of protein molecules on the surface of minerals with the solid-state nuclear magnetic resonance (NMR) method.

Theoretical and numerical studies with continuum and atomistic fracture modeling have also been performed to study the interface strength in biological nanocomposites. In particular, a soft-hard bilayer fracture model was developed to study the strength τ_{int}^* of the hybrid interface, for which a scaling law was derived as (36)

$$\tau_{\text{int}}^* = \frac{1}{\rho}\sqrt{\frac{3\Delta\gamma E_m}{C h_m}}, \qquad 6.$$

NMR: nuclear magnetic resonance

where $C = \frac{3}{2}\frac{(1-\Phi)E_m}{\Phi\mu_p\rho^2} + 1$, and Φ is the volume fraction of mineral in the composite, which is approximately equal to $h_m/(h_m + h_p)$; h_p is the thickness of the protein.

Equation 6 shows that the interface strength depends on both the size and the geometry of mineral crystals and can be optimized through the miniaturization of mineral at the nanoscale. Although determining the strength of the interface and its size dependency directly at the nanoscale remains a challenge with the presently available experimental techniques, there is qualitative experimental evidence in support of the theoretical result. For example, it has been observed (7, 87, 88) that mineral platelets of relatively large size, e.g., those in mother of pearl (approximately 500 nm in thickness), need mineral bridges or asperities on their surface to enhance the interface strength. Experimental studies (87, 89) have shown that the pullout of mineral platelets due to interface failure is a major failure mode of nacre. However, there has been no observation of mineral bridges or asperities on the surface of the bone mineral. These findings suggest that the small size of mineral crystals may also effectively enhance the protein-mineral interface strength.

An atomistic biomimicking model system taking into account the chain structure of protein molecules was developed to validate the continuum model of the protein-mineral interface (35), assuming that the chain structure of proteins is a crucial factor for the strength of the protein-mineral interface. In addition, a corresponding biomimicking experiment for verifying this study can be contemplated. MD simulations based on this model showed that the chain length of molecules plays an important role in interface strength. The underlying mechanism is that the chain can perform cyclic stick-slip motion on the substrate during the fracture and sliding of the interface (see **Figure 6**). The stick-slip process is accompanied by a buckling-like instability of chain molecules, opening up an important channel of energy dissipation for the high strength and toughness of the mineral-protein interface. This study provides a unique microscopic understanding of failure along the hybrid interface. Further analyses with chemical reaction kinetic theory and worm-like chain model showed that the increase in chain length can critically raise the strength of the hybrid interface (35). The interface strength calculated by MD simulations was consistent with that from the previous continuum results (36).

To consider the morphology of the interface and complex structure of protein molecules more accurately, all-atom MD simulations with a more realistic molecular structure were conducted. Katti and coworkers (90) studied the mechanics of collagen molecules at the mineral-protein interface with MD simulations. Their simulations showed that the interaction between the mineral and collagen can significantly affect the load-deformation response of collagen molecules. Dubey & Tomar (91) studied the roles of interfacial arrangement of collagen molecules with respect to the hydroxyapatite crystal surface in the strength of hydroxyapatite-tropocollagen (HAP-TC) composite layers. Their results showed that the presence of water molecules and their strong capability to form hydrogen bonds in the HAP-TC composite substantially increase its interfacial strength.

Studies of the hybrid interface in biological materials are very challenging due to their complex microstructures. Although previous experimental and theoretical investigations have provided many valuable insights into the design of protein-mineral interfaces at the nanoscale, there are still numerous issues that are not yet fully understood about the physics of the hybrid interface and hierarchical interface design. Recent experiments (92) have shown that in some cases biological nanocomposites adopt the same strategy of interface design at the nanoscale as do those at higher levels. Clearly, much more effort is needed.

Roles of the Hierarchical Structure

One of the most important features of biological materials is that they are always hierarchically structured. Hierarchical structuring allows the adaptation and optimization of material at each

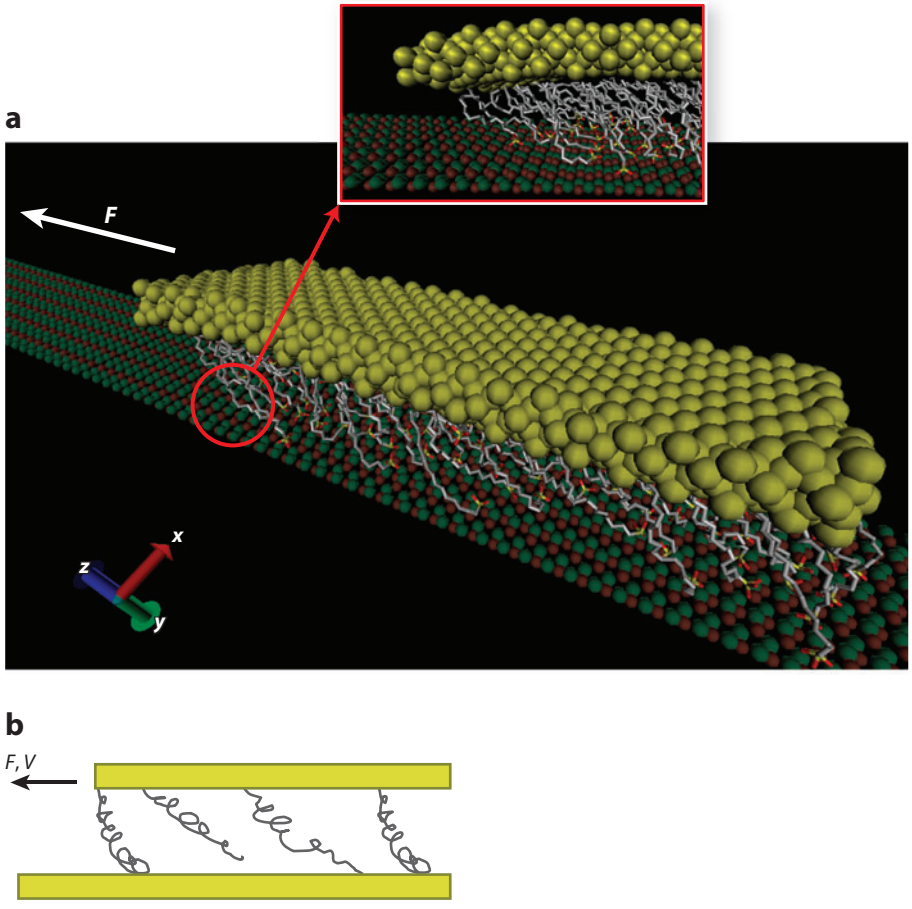

Figure 6

Atomistic modeling of protein-mineral interface strength showing the mechanical behavior of chain molecules and their interaction with the substrate during interface failure. (*a*) A snapshot of the sliding of alkane chains (*silver*) on the substrate; the inset shows a crack-like interface flaw nucleated at the leading edge of the molecular monolayer. (*b*) Schematic illustration of the stick-slip motion of alkane chains due to bond breaking and reforming dynamics. *F* denotes the pulling force and *V* the pulling velocity. From Reference 35 with permission.

level of hierarchy for outstanding performance (93). For example, the extraordinary toughness of bone is due to the combined action of structural elements at both nanometer (20, 43, 93) and micrometer (94) levels. Gao (19) developed a theoretical model to understand the roles of structural hierarchy in stiffness, strength, toughness, and flaw-tolerance properties. This model involves a hypothetical hierarchical material with multiple levels of self-similar structures mimicking the staggered nanostructure of bone and has been referred to as the fractal bone model (19, 95).

Fractal bone model: a hypothetical hierarchical material with multiple levels of self-similar structures mimicking the staggered nanostructure of bone

In this model, the hard plates at each level are assumed to be made of a staggered hard-soft microstructure from one level below, as shown in **Figure 7**. The total number of hierarchical levels is *N*. At each level of hierarchy, the roles of the hard materials and the soft materials are similar to those in the nanostructure of bone, i.e., the slender hard plates provide structural rigidity, whereas the soft matrix absorbs and dissipates fracture energy associated with crack-like flaws in the size

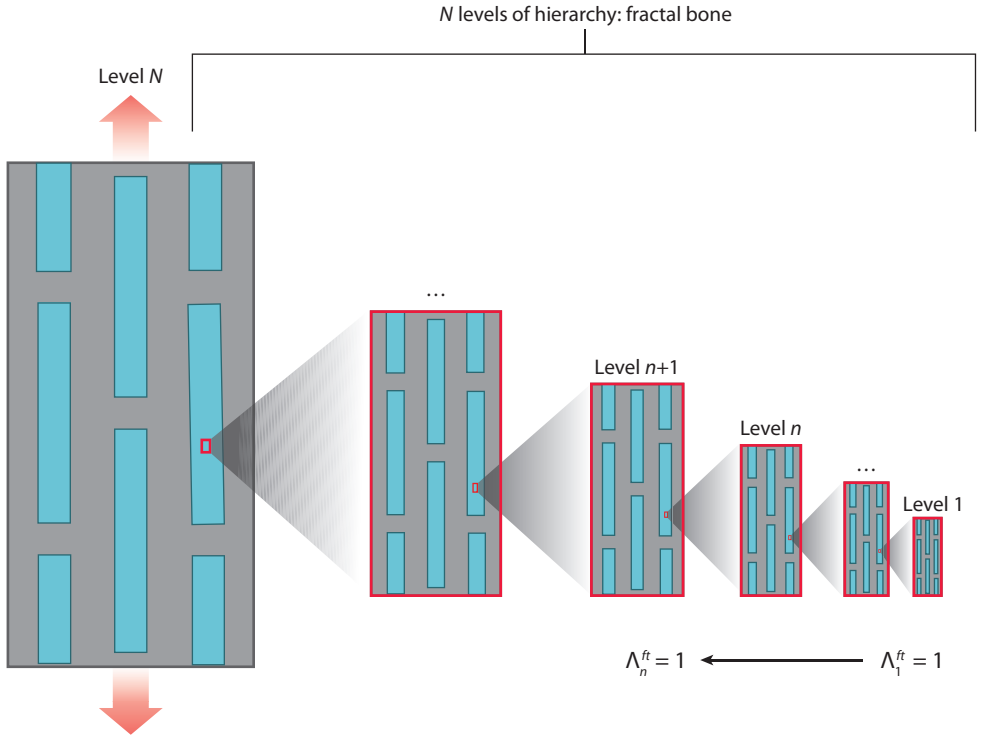

Figure 7

The *N*-level hierarchical structures of a fractal bone. Every level of structure is similar to the elementary structure of bone and nacre, with a slender hard phase aligned in a parallel staggered pattern in a soft matrix. The hard phase at the $(n + 1)$th level is made of the hard-soft microstructure at the nth level. The principle of flaw tolerance is used to determine the characteristic size of all levels using a bottom-up approach (19). Here, Λ_n^{ft} denotes the flaw tolerance number at the nth level. From Reference 19 with permission.

range of the corresponding hierarchical level. The same flaw-tolerance criterion is applied to all hierarchical levels using a bottom-up design approach. First, the characteristic size scale of the lowest level of structure, i.e., the nanostructure, is determined by Equation 3. Then the properties at the next level above are determined from the current level of structures, and the criterion of flaw tolerance is applied to determine the characteristic size at the next level. The same process of design is repeated until all *N* levels of structural hierarchy are determined.

With the hierarchical model, an analytical formula of multilevel stiffnesses of the fractal bone was derived by generalizing Equation 2 to the fractal bone (19). Increasing the number of hierarchical levels generally increases the overall stiffness of a composite. In the same spirit, analytical formulas for multilevel strengths and multilevel fracture energies of the fractal bone were derived, which then yielded characteristic structural sizes for multilevel flaw tolerance. The results showed that, although each additional level of structural hierarchy degrades the strength of the material, the fracture energy and flaw-tolerant structure sizes will increase exponentially with the increase in hierarchical levels. These calculations demonstrate the potential of a bottom-up design methodology to improve the capability of materials to withstand crack-like flaws.

Gorbatikh et al. (96) also studied the effect of hierarchical structure on fracture toughness by using reinforcements of a bimodal size distribution to model two levels of hierarchy. Their results showed that failure initiation can be shifted from stress concentration sites of the

Bottom-up design: an approach seeking to have smaller (usually molecular) components built up into more complex assemblies

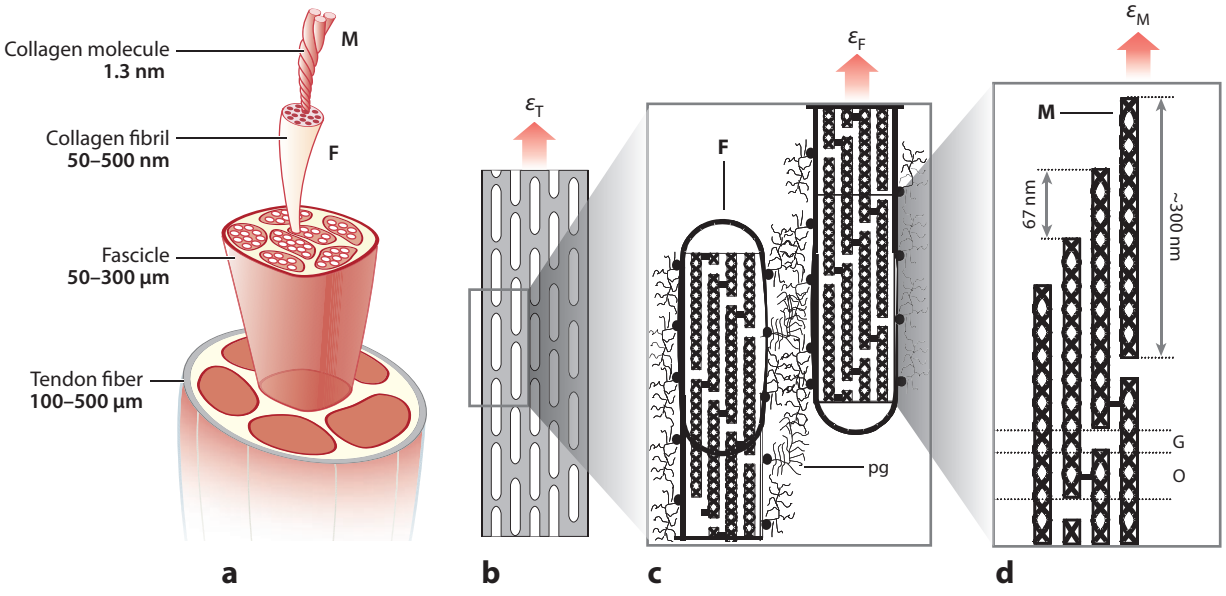

Figure 8

Multiscale deformation mechanism of tendon. (*a*) In a simplified tendon structure, tendon is made of a number of parallel fascicles containing collagen fibrils (F). (*b*) The tendon fascicle can be viewed as a composite of collagen fibrils in a proteoglycan-rich matrix, subjected to a strain ε_T. (*c*) Some of the strain is taken up by a deformation of the proteoglycan (pg) matrix. The remaining strain, ε_F, is transmitted to the fibrils (F). (*d*) Triple-helical collagen molecules (M) are packed within fibrils in a staggered way. The strain in the molecules, ε_M, may be different from the strain in the fibril, ε_F, which is caused largely by the sliding of neighboring molecules. Here, G and O denote the gap zone and the overlap zone, respectively. From Reference 98 with permission.

higher-level structure to the lower level. A most interesting finding was that arrangement of low-level structures in bands between higher-level structures will cause the stress intensity factor at the tip of a crack to be negative, effectively shielding crack initiation and propagation.

Fratzl and coworkers (15, 20, 21, 43, 44, 92–94, 97, 98) carried out systematic experimental studies on the hierarchical mechanics of biological materials, including tendon, bone, and wood. Their experiments revealed multiscale deformation mechanisms in biological materials and provide strong support to the theoretical studies discussed here.

Figure 8 shows the multiscale deformation mechanism of tendon. Collagen fibrils in tendons typically have a diameter of a few hundred nanometers and are decorated with proteoglycans forming a matrix between fibrils (**Figure 8*b*,*c***). Fibrils are assembled into fascicles and, finally, into a tendon (98). Experimental observations found that the extension of collagen fibrils inside the tendon is always considerably smaller than the total extension of the tendon (99), indicating that considerable deformation must occur in the proteoglycan-rich matrix (100) that mediates deformation by shearing between fibrils (**Figure 8*c***). Similarly, at the level of the collagen fibrils, deformation is mediated largely by gliding between neighboring molecules (99).

Gupta et al. (43) experimentally observed the deformation mechanisms at two levels in the structural hierarchy of bone in response to an external tensile load, i.e., at the levels of a fibril array and mineralized collagen fibrils. They reported that the stiff mineralized fibrils deform in tension and transfer the stress between adjacent fibrils by shearing in the thin layers of extrafibrillar matrix. Within each mineralized fibril, the stiff mineral platelets deform in tension and transfer the stress between adjacent platelets through shearing in the interparticle collagen matrix, thus having

a multiscale deformation similar to that in tendon. These observations support the assumption of deformation modes in the fractal bone model.

BIOMIMICKING SYNTHESES OF NOVEL NANOCOMPOSITES

LbL assembly method: layer-by-layer assembly method

Materials scientists are increasingly seeking to create synthetic materials based on the mechanical-design principles found in biological materials (101). Many experiments have been carried out for the synthesis of novel biomimicking nanocomposites in the laboratory. These biomimicking activities strongly validated the mechanical principles found in biomaterials. Here we discuss a few typical examples of man-made novel nanocomposites.

Kotov and coworkers (41, 81, 102) conducted systematic studies on syntheses of nacre-mimetic nanocomposites. Believing that the ordered brick-and-mortar arrangement of organic and inorganic layers and the ionic cross-linking of macromolecules in organic layers are two essential structural features of nacre, they reproduced both of these structural features by sequential deposition of polyelectrolytes and clays with the layer-by-layer (LbL) assembly method (81). The tensile strength of the prepared multilayers approached that of nacre, whereas their ultimate Young modulus was similar to that of lamellar bones. Podsiadlo et al. (102) studied the roles of strength and rigidity in the mechanical properties of nacre-mimetic nanocomposites. Later, Podsiadlo et al. (41) took an alternative approach in the design of a montmorillonite clay platelet–poly(vinyl alcohol) matrix nacre-mimetic nanocomposite by tailoring the chemistry of the platelet-matrix interface to enhance load transfer. They showed that a high level of ordering of the nanoscale building blocks, combined with dense covalent and hydrogen bonding and stiffening of the polymer chains, led to highly effective load transfer between the nanosheets and the polymer matrix. These polyelectrolyte multilayers are structurally and functionally close replicas of natural biological nanocomposites.

Bonderer et al. (103) carried out a deliberate microstructural design of a multilayered alumina platelet–reinforced chitosan nanocomposite to resemble the inner nacreous layer of many seashells. The size and aspect ratio of the reinforcing platelets, as well as the mechanical properties of individual inorganic and organic phases, were deliberately chosen to mimic the design principles of strong and tough biological structures. The alumina platelets possess higher ultimate tensile strength than the aragonite platelets found in nacre. This study demonstrated the biomimicking concept that the weak constituents found in nature can be replaced with more advanced synthetic engineered materials, with the goal of producing structural composite materials with mechanical properties that exceed both those of nacre and those of the state-of-the-art synthetic materials.

Munch et al. (104) created complex hierarchical architectures in their biomimicking nacre materials by using the freeze-casting method. In their syntheses, they could refine the lamellae thickness, control their macroscopic orientation, manipulate the chemistry and roughness of the interlamellae interfaces, and generate a given density of inorganic bridges. Their results illustrated the importance of hierarchical design in promoting toughening mechanisms at multiple length scales as a way to create materials with unique combinations of strength and fracture resistance.

Lipowsky et al. (105) developed a zinc oxide and poly(amino acid) system to synthesize a nacre-mimetic nanocomposite. They showed that the combination of nanocrystalline ZnO with poly(amino acids) greatly enhances mechanical performance, i.e., it results in a significant increase in hardness without any negative impact on the stiffness. Gehrke et al. (106) suggested a retrosynthetic approach to make a nacre-mimetic material. Kauffmann et al. (107) made a superhard titanium nitride/silicon nitride coating with a microstructure mimicking nacre. Biological materials have also inspired the development of novel nanocomposites with superior thermomechanical properties (108, 109).

Bioinspired materials: man-made materials synthesized by using the principles found in biological materials

Although strong and stiff man-made nanocomposites mimicking nacre and bone have been developed, the microstructure of the most advanced composites has yet to achieve the order and sophisticated hierarchy of biological materials. There are still many challenging problems in the biomimicking synthesis, e.g., control of the size, geometry, alignment of nanostructures, and higher levels of hierarchy. Fratzl (93) suggested a systematic approach to develop biomimicking technologies by storing all available successful biomimetic solutions into large databases, where they can then be retrieved by materials scientists and engineers. Initial attempts have been made to establish a system into which all known biomimetic solutions can be placed, classified in terms of function (110, 111). Such tools will become extremely valuable for the development of bioinspired, novel nanocomposites.

CONCLUSION AND PERSPECTIVES

Nanostructure and complex hierarchy, the two main features of biological nanocomposites, are widely recognized as the bases of the superior mechanical strength of these materials. A number of mechanical principles in the structural design of biological nanocomposites have been suggested. The first principle is the staggered arrangement of hard phases in soft matrices, which, combined with a large modulus ratio between the soft phases and the hard phases, leads to a special load transfer path and structural splitting arrangement. The second principle is the flaw-tolerant design of the hard phase in the nanocomposite structure as well as in higher levels of hierarchy, which is a novel concept in material design. The third principle is the design of the geometry of the hard phase, e.g., the aspect ratio and the characteristic length for load transfer. The fourth principle is the mechanism of large deformation and energy dissipation in the soft phase. The fifth principle involves the design of a strong hybrid interface between soft materials and hard materials. The sixth principle is the hierarchical structure design for strength optimization, fracture toughness, robustness, and the adaptation of a complex loading environment.

Many important breakthroughs in the understanding of the basic principles of biological nanocomposites can be attributed to parallel theoretical, numerical, and experimental studies. For example, the deformation mechanism in the nanostructure of biological materials predicted by the TSC model was observed soon after the theoretical finding. Experimental studies guided the theoretical modeling by revealing that natural materials use similar deformation mechanisms at higher levels of hierarchies. The flaw-tolerant concept and the brittle-to-ductile fracture mode transfer of nanomaterials were demonstrated by large-scale MD simulations and nanoscale experiments. The most exciting discovery is that many synthetic nanocomposites using the principles from biological materials possess mechanical properties approaching or exceeding those of their natural counterparts. We expect that more breakthroughs will be made in this area in the near future.

Besides materials synthesis, the theoretical analysis and description of biomimicking nanocomposites at multiple material scales represent major scientific and engineering challenges and opportunities. For the accurate design of multiscale hierarchical materials, the integration of predictive theoretical and numerical studies with experimental methods represents a new frontier in materials research. An integrated approach with concurrent experimentation and computer simulation could evolve into a new paradigm of materials research. In the modeling and simulation efforts, a central theme is to develop a multiscale structure-property paradigm, i.e., to link the different scales of materials by modeling and describing the cross-scale interactions and to predict their overall mechanical properties. The presence of complex hierarchical structures of biological materials makes the development of these rigorous links between structure and properties a frontier of multiscale mechanical modeling of materials.

SUMMARY POINTS

1. The smallest building blocks in biological materials—such as nanometer-sized mineral crystals embedded in a soft protein matrix in a staggered alignment pattern—are generally on the nanometer length scale.
2. In the nanostructure of biological nanocomposites, the mineral crystals have large aspect ratios and are much harder than the soft protein matrix. Under an applied tensile stress, the mineral platelets carry most of the tensile load, whereas the protein matrix transfers the load between mineral crystals via shear. The path of load transfer in the composite is thus simplified to a one-dimensional serial spring system called the tension-shear chain (TSC). The TSC model is regarded as a classic model of the primary structure of biological nanocomposites.
3. At a critical length, generally at the nanometer scale, the theoretical strength of a perfect mineral platelet is maintained in spite of defects, and failure is governed by the theoretical strength of material rather than by the classical Griffith criterion of fracture propagation. It is therefore hypothesized that the nanometer size of the mineral crystals in biological nanocomposites may have been selected to ensure optimum fracture strength and maximum tolerance of flaws (for robustness). Nature finds this secret and hides material defects by designing the elementary structure of biomaterials at the nanoscale to achieve the level of robustness required for survival.
4. The aspect ratio of mineral crystals can affect the mechanical properties of the nanostructure of biological materials in many aspects, such as load transfer in the nanostructure, stiffness, strength, and elastic stability. The optimal value of the aspect ratio should balance many aspects of the mechanical properties of the nanostructure, e.g., the limiting strength of mineral, protein, and interface; the elastic stability of the nanostructure; and an effective load transfer in the nanocomposite structure of biological materials.
5. The protein matrix is crucial to the mechanical properties of biological nanocomposites in many aspects, such as transferring load through shear deformation, trapping cracks through matrix softness, and dissipating energy through its large deformation and viscoelastic properties. The hierarchical structures of protein molecules are ideally suited for inducing large deformations and absorbing and dissipating fracture energy.
6. The interface properties should play more crucial roles in the overall mechanical properties of nanocomposites than in their conventional counterparts. Atomistic and continuum studies showed that the interface strength depends on both the size and the geometry of mineral crystals and can be optimized through the miniaturization of minerals at the nanoscale.
7. One of the most important features of biological materials is their hierarchical structuring, which allows for the adaptation and optimization of the materials at each level of hierarchy for outstanding performance. Increasing the number of hierarchical levels generally increases the overall stiffness of the composite. In addition, the fracture energy and flaw-tolerant structure sizes will exponentially increase with the increase in hierarchical levels. These calculations demonstrate the potential of a bottom-up design methodology to improve the capability of materials against crack-like flaws.

8. Materials scientists are increasingly seeking to create synthetic materials based on the mechanical-design principles found in biological materials. Many experiments have been carried out for the synthesis of biomimicking novel nanocomposites in the laboratory. These biomimicking activities strongly validate the mechanical principles found in biomaterials.

DISCLOSURE STATEMENT

The authors are not aware of any affiliations, memberships, funding, or financial holdings that might be perceived as affecting the objectivity of this review.

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