

Mechanical stability of dental materials

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17.1 Introduction

Over the course of time, dentistry, or dental care, has evolved from a primitive form of medicine to modern day use of preventative dental care, state of the art diagnostics, and modern treatment. During the past several centuries, dentistry has seen many changes and improvements. Materials that are specially fabricated and designed for use in dentistry are called dental materials. The science of dental material deals with the physical, mechanical, and biological properties of the components that comprise dental materials [1]. The study of the science of dental material is very important for the following reasons: (i) optimal selection of materials; (ii) for understanding the behavior of the materials; (iii) for safety considerations related to the materials; and (iv) recognition of materials, and proper care of prosthodontic/restoration.

17.2 Mechanical properties of tooth structures

Teeth are composed of a hard outer covering of enamel, a mid-layer of softer dentin, and an inner layer of pulp. The various layers of teeth can be considered different materials, and have different functions, structures, and properties. In addition, teeth located in different positions of the mouth have different forms and functions. The dental enamel is the hardest tissue in the body, and is composed of 92%–96% inorganic matter, 1%–2% organic material, and 3%–4% water by weight. Enamel is hard and brittle due to high mineral content, and its high elastic modulus and low tensile strength, respectively. In spite of the very strong nature of enamel, it cracks and fractures occasionally. Studies have shown that mechanical properties of enamel are dependent on the type and direction of the stress applied, as well as the prismatic orientation. For enamel, the hardness and elastic modulus decrease gradually from the surface of the enamel to the dentin-enamel junction (DEJ), and both exhibit positive relationships with calcium content. The hardness and elastic modulus of the heads of the enamel rods are distinctly higher than those of the tail and axial cross-section. Higher hardness and elastic modulus values can be obtained in the perpendicular direction. Increased crack resistance for enamel has been observed from outside to inside (Table 17.1).

The middle layer of teeth is soft dentin, which is composed of 70% inorganic material, 18% organic matrix, and 12% water (by wt), with properties and structural components that vary with location. The collagen phase of intertubular dentin contributes to a

Table 17.1 Mechanical properties of enamel and dentin

Mechanical properties of enamel					
Method and indenter	Site	Load	Hardness/GPa	Elastic modulus/GPa	Reference No
UIMS, Berkovich indenter	1st molar	50 mN 150 mN	4.88 ± 0.41 4.87 ± 0.29	80.94 ± 6.65 79.77 ± 8.86	[21]
Nanoindentation, spherical indenter	3rd molar	0.4 mN	5.7 ± 0.3	86.4 ± 11.7	[22]
Nanoindentation, Berkovich indenter	2nd, 3rd molar	400 or 800 nm in depth	Surface > 6 Near DEJ < 3	Surface > 115 Near DEJ < 70	[23]
Nanoindentation, Berkovich indenter	3rd molar	1000 µN 300 µN	Rod 4.3 ± 0.8 Sheath 1.1 ± 0.3	Rod 83.4 ± 7.1 Sheath 39.5 ± 4.1	[24]
<i>Mechanical properties of dentin</i>					
UIMS, Berkovich indenter	1st molar	50 mN 150 mN	0.95 ± 0.11 0.90 ± 0.09	20.55 ± 2.00 19.22 ± 1.84	[21]
Nanoindentation, Berkovich indenter	3rd molar, premolar, dry environment	200–300 µN	Peritubular dentin 1.34 ± 0.5; Intertubular 0.60 ± 0.2	Peritubular dentin 29.3 ± 6.7; Intertubular dentin 17.4 ± 3.5	[25]
UIMS, Berkovich indenter, moist environment	1st molar, dentin of crown	25 mN	Dentin nearest the pulp wall 0.52 ± 0.24; Dentin in the middle area 0.85 ± 0.19; Dentin nearest dentinoenamel junction 0.91 ± 0.15	Dentin nearest the pulp wall 11.59 ± 3.95; Dentin in the middle area 17.06 ± 3.09; Dentin nearest dentinoenamel junction 16.33 ± 3.83	[26]
Resonant ultrasound spectroscopy	3rd molar, dentin of crown	Vibration frequency: 0.5–1.4 MHz		Hydrated: 25.1 Dry: 28.1	[27]

lower modulus of elasticity than enamel, while the lower mineral content is associated with a decrease in dentin microhardness, as compared with enamel. Tubule orientation, collagen fibril, tubule density, and age of the human dentin are common factors that affect mechanical properties of dentin. The dentin with tubule orientation parallel to the loading direction exhibits higher mechanical properties than those with tubule orientation perpendicular to the loading direction, while the dentin close to the dentin-enamel junction (DEJ) has a higher fracture toughness and elastic modulus than that close to the pulp, due to the lower density and diameter of the tubule. This higher hardness of DEJ supports the integrity of enamel by preventing its fracture during its functioning. Aged dentin displays lower crack growth toughness due to the large fraction of tubules filled with carbonated apatite. For dentin, the variation of the hardness and elastic modulus decreases slowly toward the pulp cavity. Highly mineralized peritubular dentin has a higher Young's modulus than weakly mineralized intertubular dentin. A hydrated environment influences the mechanical behavior of dentin. For dentin, crack bridging, crack deflection and crack bifurcation in young hydrated dentin assists in energy dissipation and increases the crack growth resistance. The pulp is a soft tissue of tooth, and contains the nerve and blood supply that maintain the vitality of the tooth. Pulp occupies the central portion of the tooth, which is divided into a coronal pulp chamber and a radicular root canal.

The tensile strength of tooth structure shows that dentin is stronger (50 MPa) in tension as compared with enamel (10 MPa), and the compressive strength of enamel and dentin is comparable.

17.3 Dental materials and their properties

Currently, different types of materials are available for dentistry according to their intended purposes. These include temporary dressings, dental restorations (fillings, crowns, and bridges), endodontic materials (used in root canal therapy), impression materials, prosthetic materials (dentures), dental implants, and many others. Dental materials have been classified on various bases, but the most common classification of dental materials is based on their nature; namely, metallic or nonmetallic ([Table 17.2](#)) [2, 3]:

Metals are usually hard, lustrous, dense, good conductors of heat and electricity, ductile, and malleable. They are electropositive, and furnish positive ions in solution. Pure metals are infrequently used in dentistry. Occasionally, pure gold (Au) is used as a high quality metal, but high cost is its serious limitation. The addition of Cu to a gold alloy increases its hardness, but too much copper leads to corrosion of the alloy in the oral cavity. The hardest and most durable alloys seem to be combinations of Au, Ag, Cu, and Pt. The dental amalgam consists of Ag, Sn, Cu, and Hg. The high copper amalgams are clinically superior to the traditional ones in marginal integrity and mechanical resistance.

Nonmetallic dental materials have gained popularity owing to their favorable properties. The gypsum products, in general, are important accessory materials used in many clinical and laboratory procedures. Impression plaster and plaster are based on ordinary gypsum plaster, while stones (either with high strength, low expansion or with

Table 17.2 Classification of dental materials on the basis of their nature

Metallic materials	Dental alloys		
	Dental amalgams		
Nonmetallic materials	Gypsum products	Impression plaster	
		Plaster	
		Stone	
		Stone, high-strength, low-expansion	
		Stone, high-strength, high- expansion	
	Dental porcelains (silicates)		
	Dental cements	Inorganic dental cements (zinc phosphate, zinc silicophosphate)	
		Organic cements	Phenolate-based cements
			Polycarboxylate-based cements
			Glass-ionomer cements
			Polymer-based cements
	Impression materials	Nonelastic	Impression plaster
			Impression compound
			Zinc oxide /eugenol pastes
			Impression waxes
		Elastic	Hydrocolloids
			Non aqueous elastomers

high strength, high expansion) are based on high-strength gypsum plasters. Dental porcelains consist of SiO_2 , Al_2O_3 , K_2O , Na_2O , ZrO_2 , B_2O_3 , Rb_2O , SnO_2 , BaO , ZnO , CaO , and TiO_2 [4, 5]. Dental cements are used in very small quantities, but represent the most important materials in clinical dentistry nowadays because of their applications as (i) luting agents to bond preformed restorations and orthodontic attachments on or in the tooth; (ii) cavity liners and bases serving as protection of the dental pulp, foundation, and anchors for restorations; and (iii) restorative materials. Both organic and inorganic cements are used in dentistry. More recently, with the introduction of glass-ionomer cements, interest in restorative applications has increased substantially.

Impression materials are used for making replicas (impressions, imprints) of oral structures. These materials must be plastic while the replica is being made; later, a physical or chemical change is required to convert the substance into a solid that keeps

the shape of the negative replica of the hard and soft tissues in the mouth. The impression compound consists of about 40% natural resins, which make the mixture thermoplastic. Shellac is often used; waxes also offer good thermoplastic properties; while stearic acid acts as a lubricant and plasticizer. The dental compound is thermoplastic: it is used warm (45°C), and then cooled to the mouth's temperature (37°C), at which point it is fairly rigid. The setting mechanism, then, is a physical, rather than chemical, reaction. Accordingly, the results are relatively poor [6].

In addition to the preceding classification, dental materials can also be classified as:

1. Preventive dental materials
2. Restorative dental materials
 - 2.1. Direct restorative dental materials
 - 2.2. Indirect restorative dental materials
3. Auxiliary dental materials

The basic aim of preventive materials is to provide resistance to the progression of an active carious lesion that causes demineralization of the dental enamel. The preventive materials may be chemotherapeutic agents (dentifrices, mouth washes, and cavity varnishes), pit and fissure sealants, and fluoride releasing cements. Preventive materials may either be used for a short period or long period. Restorative materials help to repair or replace defective tooth structure. These include amalgams, bonding agents, resin-based composites, compomers, ceramics, liners, cements, and dental polymers. Direct restorative materials are used intra-orally, while indirect restorative materials are used extra-orally. Auxiliary dental materials are used in the process of fabricating dental prostheses. These include impression materials, dental waxes, gypsum casts, finishing and polishing abrasives, acrylic resins for impressions, and acid-etching solutions [7].

The ideal dental material should possess the following properties:

1. *Biocompatibility*: The dental materials should be nontoxic, nonirritating, and nonallergic.
2. *Mechanically stable and durable*: The materials should be strong and resist fracturing.
3. *Resistant to corrosion*: The materials should not deteriorate over time in oral saliva.
4. *Dimensionally stable*: The materials should be stable enough to withstand extreme temperatures and solvents.
5. *Minimal conduction*: The materials should be insulators against thermal or electric charges.
6. *Aesthetic*: The material should look like oral tissues.
7. *Easy to manipulate*: Minimum/reasonable effort and time should be needed to implant the dental material in an oral cavity.
8. *Tissue adherence*: Dental materials should adhere to tissue and retain onto and seal broken tooth structures.
9. *Tasteless and odorless*: No unpleasant effect should be produced in the patient.
10. *Cleanable and repairable*: Dental materials should be easy maintainable and fixed.
11. *Cost-effective*: Should be inexpensive and readily available.

Considering the characteristics of an ideal dental material, it is apparent that no single material can fulfill all of the clinical needs. A wide spectrum of properties is present within each basic material type; nevertheless, there is a "family resemblance" among the varieties of each material type. These properties must be taken into account during the selection of material used in dentistry. For example, metal exhibits a

wide range of strengths, melting ranges, and so on. Metals also resemble one another in their ductility, thermal and electrical conductivities, and metallic luster; while non-metals are devoid of these properties. Similarly, ceramics can be characterized as strong yet brittle, and polymers tend to be flexible (low elastic modulus) and weak [8]. The properties of dental materials can be classified into five major headings, as detailed in Table 17.3.

Physical properties are of great importance for dental materials. These properties are needed to assess the characteristics of improvement in materials under development. The physical properties of a tooth set the standards for materials attached to a tooth. For example, if a restorative material possesses similar properties to those of natural tooth structures, it should perform as well as the original tooth. Despite the fact that physical properties of dental filling materials are of great interest in modern dentistry, a closer look at the problem of long-term durability of dental fillings proves that the thermal stress resistance should not be neglected. Moreover, in certain cases, such as thermal shock damage to a filling-tooth structure, thermal stress resistance could be of greater account than that produced by a strictly mechanical load. Heat effects cannot be neglected, because of a growing use of radiation in dental therapy.

Electrical properties, including galvanism and electrical conductivity, are also accountable for dental materials, especially during the impregnation of the resin monomers in dentin bonding systems for hybrid layer formation. The electrical properties of dental materials correlate with mechanisms of the monomer migration under an applied electric field. The optical properties of restorative material should be similar to those of the natural teeth. Color, metamerism, fluorescence, opacity, translucency, and transparency are the desired optical properties of dental materials, so that the desired aesthetic appearance of a restoration will match the surrounding dental tissues.

Table 17.3 Properties of dental materials

Physical	Thermal	Electrical	Optical	Mechanical
Density	Coefficient of thermal expansion	Galvanism	Color and aesthetic	Stress
Viscosity	Thermal conductivity	Electrical conductivity	Metamerism	Tensile, compressive and shear stress
Hardness	Boiling and melting point(s)		Fluorescence	Strain
Abrasion resistance	Vapor pressure		Opacity, translucency, transparency	Elongation and compression
Solubility and sorption				Resilience
Adhesion				Toughness
Wettability				

One of the significant properties of dental materials is the ability to withstand various mechanical forces placed on them during use as restorations, impressions, models, appliances, and tools. Mechanical properties of all dental materials are frequently evaluated to predict their clinical performance. Mechanical properties deal with forces that act on bodies and the resultant motion, deformation, or stress that those bodies experience. Mechanical properties measure response, both elastic (reversible on force removal) and plastic (irreversible on force removal), of materials under an applied force or distribution of forces. Mechanical properties of direct restorative materials are crucial, not only to serve and allow similarity with human enamel and dentine, but also to compare composites and determine objective criteria for their selection. These properties are considered collectively because no single mechanical property can give a true measure of quality or performance. Improving the performance of one property might alter the performance of another property, so it is essential to understand the principles involved in variety of mechanical properties to optimize the clinical services of dental materials. All mechanical properties are measure(s) of the resistance of a material to deformation, crack growth, or fracture under an applied force or pressure, and the induced stress. These properties are most commonly expressed in terms of stress and strain.

17.4 Stress and strain

Stress is defined as force per unit area. It is normally defined in terms of mechanical stress, which is the force divided by the perpendicular crosssectional area over which the force is applied. For a given force, the smaller the area it's applied to, the larger the value of the stress; and increasing force over a smaller area will also cause increased stress. Because stress is directly correlated with the force applied, and inversely related to the area over which the force acts, it is an important consideration. Usually, in the case of dental restoration, the area over which the force is applied is very small, and the force applied is high; thus the possibility of greater stress will always be there. Stresses can be produced in the range of thousands of MPa when the contact area of a cusp or dental explorer is used to apply the force. There are many types of stresses, according to the direction of the applied force; each type of stress is accompanied by the same type of strain (Fig. 17.1).

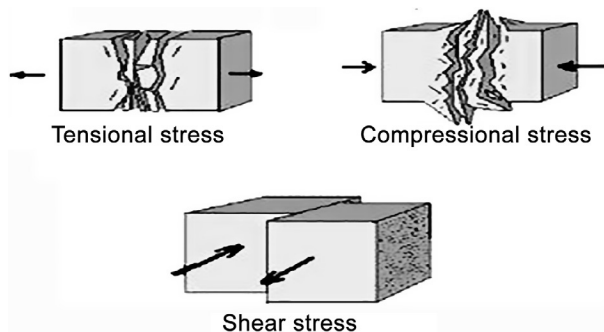


Fig. 17.1 Types of stress.

1. Tensile stress is defined as the force per unit area induced in the body in response to externally applied force, which tends to elongate or stretch the body. It is accompanied by tensile strain. Usually, most dental materials are quite brittle, and are highly susceptible to crack initiation when subjected to tensile stress, such as when they are subjected to flexural loading. Although some brittle materials are strong, they fracture with little warning, because little or no plastic deformation occurs to indicate high levels of stress.
2. Compressive strain is the force per unit area induced in the body in response to externally applied force, which tends to compress, or shorten, the body. It is also accompanied by compressive strain. Investment materials, and restorative materials and models should have high compressive strength.
3. Shear stress is force-per-unit-area induced in the body in response to externally applied force that is applied to one part of the body in one direction, while the rest is being pushed in the opposite direction. In an oral environment, shear failure is unlikely to occur for at least four reasons:
 - i. Brittle materials used for restored tooth surfaces have a rough, curved surface.
 - ii. The presence of bevels, chamfers, or changes in the curvature of the bonded tooth surface would also make shear failure of a bonded material highly unlikely.
 - iii. To produce shear failure, the applied force must be located immediately adjacent to the interface. This is quite difficult to accomplish, even under experimental conditions under which polished, flat interfaces are used. The farther the load is applied from the interface, the more likely it is that tensile failure occurs, rather than shear failure, because the potential for bending stresses would increase.
 - iv. Because the tensile strength of brittle materials is usually well below their shear strength values, tensile failure is more likely to occur.

Strain is defined as the change in length-per-unit length when stress is applied. It has no unit because it is the ratio of change in the length/original length.

$$\text{Strain} = \frac{\text{Change in length}}{\text{Original length}} = \frac{\Delta L}{L_0} \quad (17.1)$$

The magnitude of strain varies according to material used, and with the amount of the load applied. Usually, deformation of the body occurs when a force is applied to it. The tensile or pulling force causes elongation deformation in the axis of a body, whereas pushing or compressive force cause shortening or compression of the body in the axis of the loading. Strain is the important parameter during the selection of dental material.

The stress-strain ratio of a dental material is important to determine its mechanical behavior. The stress-strain curve is drawn by generating a stress-strain proportional relationship. If permanent deformation does not occur, it demonstrates material elasticity. This proportion occurs until a limit point called the proportional limit point and deformation as an elastic deformation (Fig. 17.2). At the proportional limit point, the maximum stress of a material will withstand without permanent deformation, and elasticity of a dental material can be calculated by the ratio of the stress-strain curve within the elastic limit. This proportionality is defined as the modulus of elasticity, or Young's modulus, and measures the stiffness of dental material. However, when the applied load exceeds this point, irreversible deformation occurs, resulting in permanent plastic deformation. Each material presents a resistance to deformation, and after this point, it will result in its rupture. At this point, the ultimate strength value is obtained [9].

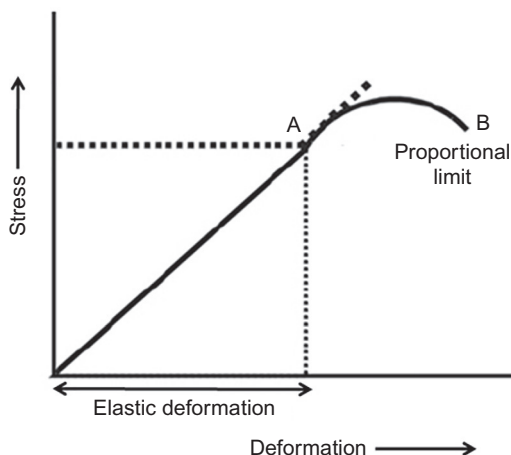


Fig. 17.2 Stress-strain curve depicting elastic deformation and proportional limit.

17.5 Strength properties

Strength is the important factor in the design of a dental prosthesis, which ensures that the prosthesis serves its intended functions effectively and safely over extended periods of time. Strength is the stress at which the material begins to function as a plastic. Strength is the force-per-unit area induced in a body in response to some externally applied force. It is force/area measured in kg/cm^2 or pound/in^2 , or pascal. Strength defines the ability of the prosthesis to resist induced stress without fracture or permanent deformation (plastic strain) [10, 11].

There are three types of strength:

1. *Transverse strength*: It is the strength of the beam to a force applied on the center of the beam when it is supported at its two ends.
2. *Fatigue strength*: It is the resistance of the material when the material is constantly subjected to a change in shape and dimensions due to the frequent application of force, such as a clasp arm or acrylic denture, for example, clasp material has high fatigue strength.
3. *Impact strength*: It is the ability of the material to resist the break of sudden impact, for example, denture material has high fatigue strength.

Strength properties are presented in Fig. 17.3.

17.5.1 Proportional limit

The proportional limit is the maximum stress that a dental material sustains without any deviation, or the magnitude of elastic stress above which plastic deformation occurs. Below the proportional limit, no permanent deformation occurs; and when the stress is removed, the structure returns to its original dimension. Proportional limit (PL) can be easily deciphered in Fig. 17.3:

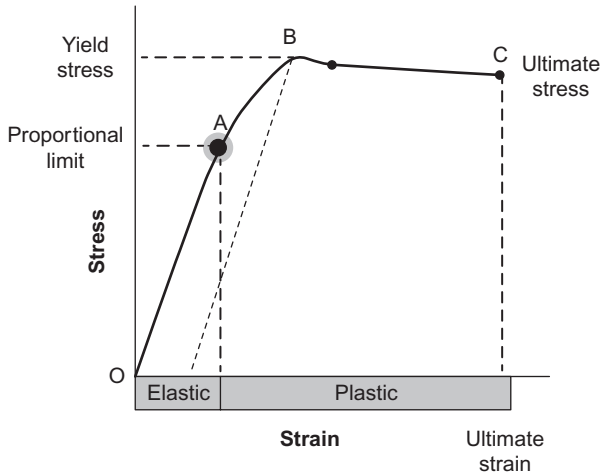


Fig. 17.3 Stress-strain curve representing elastic and plastic behavior.

In preceding figure, as the stress increases, the strain also increases. From O to A, a linear relation is established between stress and strain. As the stress doubles, the strain also doubles. After point A, no linear proportional relation is maintained between stress and strain. Here, the value “A” is identified as the proportional limit. So, the proportional limit is defined as the highest stress at which the stress-strain curve is a straight line. Below the proportional limit, there is no permanent deformation in a structure, that is, the object returns to its original position after the removal of applied force. Thus, the material is elastic in nature below the proportional limit, and the curve before the proportional limit is called the “elastic region,” and above the proportional limit, is called the “plastic limit.” The connectors of plastic dentures should have a high proportional limit. Materials such as cobalt/chromium (alloy) have high proportional limits; thus, they should be used to fabricate connectors, because they can withstand high stresses without permanent deformation.

17.5.2 Elastic limit

The elastic limit (EL) is the maximum stress produced in the material after which the material returns to its original dimension if the force is removed. For linearly elastic materials, the elastic and proportional limits represent the same stress within the structure. On increasing the stress beyond the PL or EL, the material will deform and on removing stress, but the material will not return to its origin dimension, which is called “permanent deformation.” On further application of more force, the material will break.

17.5.3 Yield strength

Yield strength is defined as the stress at which the material exhibits a specified limiting deviation from the proportionality of stress to strain. It is the amount of stress required to produce a predetermined amount of permanent strain, usually 0.1

or 0.2%, which is called the “percent offset.” Yield strength is a property often used to describe the stress at which the material begins to function in a plastic manner. In the figure, the yield strength is represented by the point of intersection of the parallel line to the stress-strain curve. The proportional limit, elastic limit, and yield strength are defined differently; but their values are fairly close to each other in many cases. Yield strength cannot be measured for brittle materials such as composites and ceramics.

17.5.4 Diametral tensile strength

Diametral tensile strength (DTS) is the indirect measure of tensile strength by breaking (applying compression) a disk of brittle material placed on its side. A diametral compression test is used for brittle materials, and for materials that exhibit predominantly elastic deformation without plastic deformation. DTS is measured by effective forces that are directed away from the center of the disk, thus making the specimen break under tensile forces, even though the applied force is compressive in nature. Fracture occurs along the vertical plane (Fig. 17.4).

17.5.5 Flexural strength

Flexural strength/bend strength, or transverse strength, is the ability to bend before breaking. Most restoration and prosthesis fractures develop progressively over many stress cycles after initiation of a crack from critical flow, and subsequently by propagation of the crack until a sudden unexpected fracture occurs. This phenomenon is called “fatigue failure.” Flexure strength is the strength required to break a rod or rectangular specimen when supported on each end and loaded in the center, causing the specimen to bend (3-point bend) (Fig. 17.5). The specimen bends until it breaks. Thus, it is a measure of how a material behaves when under multiple stresses. Flexure strength of dental material is evaluated by using

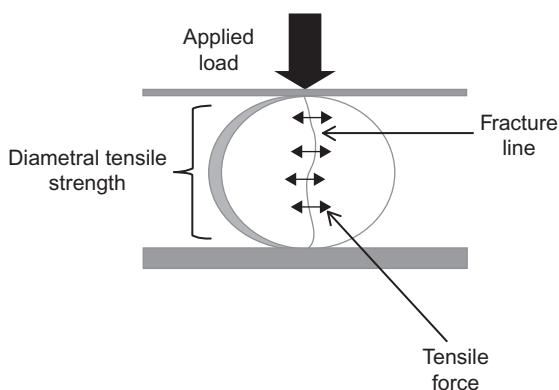


Fig. 17.4 Schematic diagram of diametral tensile strength (DTS) test.

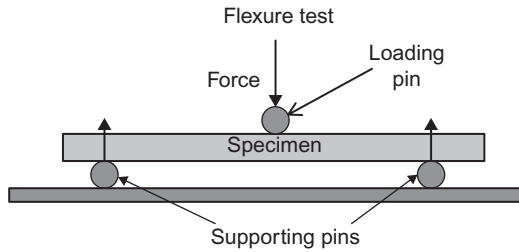


Fig. 17.5 Schematic diagram of flexure test.

a bar-shaped specimen with dimension of 25 mm in length $\times$ 2 mm in width and $\times$ 2 mm in height. Specimens are placed on a support, and a load is applied at the center (called the three-point bending test) [12, 13].

Flexure strength (σ) can be calculated as:

$$\sigma = \frac{3PL}{2Wt^2} \quad (17.2)$$

where σ is max. flexure strength (MPa), P is load at fracture (N), L is distance between two supports (mm), W is width of specimen (mm), and t is thickness of specimen (mm).

17.5.6 Fatigue strength

Dental materials are subjected to many cycles of repeated stress during mastication. Materials subjected to such a stress undergo material fatigue, which means they fracture at a stress lower than the value required for a single load application. It is the endurance limit that is, maximum stress cycles that can be maintained without failure. Thus, the fatigue strength is defined as the stress at which the material fails under repeated loading [14, 15]. Fatigue strength can be determined by subjecting a material to a cyclic stress of maximum known value and determining the number of cycles required to produce failure.

17.5.7 Impact strength

Impact strength is the measure of energy absorbed by a material when it is broken by a sudden blow. Ideally, denture base resins should have high impact strength to prevent breakage when they are accidentally dropped. Plasticizer addition increases the impact strength. Although strength is an important factor, it is not a reliable property for estimating the survival probabilities over time of prostheses made of brittle material, because strength increases with specimen size and stressing rate, decreases with the number of stress cycles, and is strongly affected by surface processing damage. Thus, strength is not a true property of a material compared with fracture toughness, which more accurately describes the resistance to cracking of brittle materials.

17.6 Mechanical properties based on elastic deformation

There are several important mechanical properties measuring elastic deformation or reversible deformation. These include:

1. Elastic modulus (Young's modulus, or modulus of elasticity, or Hook's law)
2. Dynamic Young's modulus
3. Flexibility
4. Resilience
5. Poisson's ratio

17.6.1 Elastic modulus (Young's modulus or modulus of elasticity)

Young's modulus describes the relative stiffness of a material, which is measured by the slope of elastic of a stress and strain graph. It is calculated by the ratio of stress value to its corresponding strain value. A constant of proportionality will result, which is known as the modulus of elasticity, or Young's modulus (E).

$$E = \frac{\text{Stress}}{\text{Strain}} \tag{17.3}$$

where stress(σ) = $\frac{F}{A}$, and Stress (ϵ) = $\frac{\Delta l}{l_0}$

“Elastic modulus” describes the relative stiffness or rigidity of a material, that is, a stiff material will have a high module of elasticity, while a flexible material will have a low module of elasticity. Elastic modulus has a unit of giga-newton/m² (GN/m²). This phenomenon can play a role in burnishing the margins of a crown (high elasticity), but the impression materials should have a low modulus of elasticity to enable removal from the undercut areas in the mouth. The modulus of elasticity should not be so low that the material cannot withstand tearing. Enamel has a higher elastic modulus (3–4 times) than dentin, and is stiffer and more brittle. Although dentin is more flexible and tougher, ceramic has a higher modulus than polymers and composites [16]. Materials with their elastic modulus in GN/m² are presented in Table 17.4.

Table 17.4 Materials with their elastic modulus in G N/m²

Material	Elastic modulus in G N/m ²
Enamel	84.1
Dentin	18.3
Feld spathic porcelain	69.0
Composite resin	16.6
Acrylic denture resin	2.65

17.6.2 *Dynamic Young's modulus*

Elastic modulus can be measured by a dynamic method. The velocity at which sound travels through a solid can be readily measured by ultrasonic transducers and appropriate receivers. The velocity of the sound wave and the density of the material can be used to calculate the “elastic modulus” and “Poisson ratio” values. This method of determining “dynamic elastic moduli” is less complicated than conventional tensile or compressive tests. If a shear stress is induced instead of a uniaxial tensile or compressive stress, the resulting shear strain could be used to define a shear modulus for the material. The shear modulus (G), can be calculated from the elastic modulus (E) and Poisson's ratio (ν), using equation

$$G = \frac{E}{2(1+\nu)} = \frac{E}{2(1+0.3)} = 0.38E \quad (17.4)$$

Typically, the value of Poisson's ratio is 0.3, and thus shear modulus (G) is usually 38% of the elastic modulus value.

17.6.3 *Flexibility*

Flexibility is defined as the flexural strain that occurs when the material is stressed to its proportional unit. A larger strain or deformation with slight stresses is called “flexibility,” and is an important consideration in orthodontic appliances. For materials used to fabricate dental appliances and restorations, a high value of elastic limit is a necessary requirement. This is because the structure is expected to return to its original shape after it has been stressed and the force removed. Maximum flexibility is the flexural strain that occurs when the material is stressed to its proportional limit.

17.6.4 *Resilience*

Resilience can be defined as the amount of energy per unit volume that is sustained on loading, and released upon unloading, of a test specimen. It is popularly associated with springiness of a material, but it means precisely the amount of energy absorbed within a unit volume of structure when it is stressed to its proportional limit. For example, when an acrobat falls on a trapeze net, the energy fall is absorbed by the resilience of the net, and when this energy is released, the acrobat is again into the air. This is a stress-strain relationship that illustrates the concepts of resilience and toughness. Resilience of two or more materials can be compared by observing the areas under the elastic region of their stress-strain curve. Material with a longer elastic area has higher resilience. The restorative material should exhibit a moderately high elastic modulus, and relatively low resilience, thus limiting the elastic strain.

17.6.5 *Poisson's ratio*

When a tensile stress or compressive stress is applied to a cylinder or rod, there is simultaneous axial and lateral strain within the elastic range. The ratio of the lateral to the axial strain is called Poisson's ratio. It has no unit, as it is the ratio of two strains. Most rigid materials such as enamel, dentin, amalgam, and composite have

a Poisson's ratio of about 0.3. More ductile materials such as soft gold alloy showed a higher degree of reduction in the crosssectional area and have a higher Poisson's ratio.

17.7 Other mechanical properties

Other mechanical properties of dental materials include:

17.7.1 Toughness

Toughness is defined as the amount of energy required (both plastic and elastic deformation energies) to fracture the material. It is the inherent property of material that describes the nature of the material to break. Toughness is determined by measuring total area under the stress-strain curve (Fig. 17.6). Toughness increases with an increase in elasticity and ductility. On increasing strength, ductility increases, and toughness will also increase. Thus, a tough material is generally strong, but strong material is not necessary to tough [17].

17.7.2 Fracture toughness

The fracture toughness of all dental materials is one of the major concerns in clinical applications of dentistry. Fracture toughness is another mechanical property that describes the ability of a dental material containing a crack to resist fracture. "Fracture toughness" describes the resistance of brittle materials to the propagation of flaws under an applied stress, and it assumes that the longer the flaw, the lower is the stress needed to cause fracture. The ability of a flaw to cause fracture depends on the fracture toughness of the material. Thus, fracture toughness is proportional to energy consumed in plastic deformation. Fracture toughness is expressed in units of stress times the square root of crack length, that is, $\text{MPa M}^{1/2}$ or $\text{MN M}^{3/2}$ (Table 17.5). It is one of the most important mechanical properties of any dental material for design applications [18]. Fracture toughness values may serve as a basis for comparison, selection, structural flaw tolerance assessment, and quality assurance of materials.

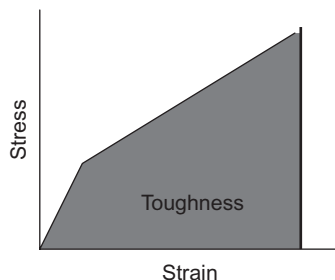


Fig. 17.6 Area of toughness in stress-strain curve.

Table 17.5 Fracture toughness of commonly used dental materials

Material	Fracture toughness
Enamel	0.7–1.3
Dentin	3.1
Amalgam	1.3–1.6
Ceramic	1.2–3.0
Composite	1.4–2.3
Porcelain	0.9–1.0

17.7.3 Brittleness

It is the relative inability of a material to sustain plastic deformation before fracture will occur. For example, amalgam ceramics and composites are brittle at an oral temperature between 5°C and 55°C. They sustain little or no plastic strain before fracture. Brittle material may or may not be weak. Dental materials with low, or 0% elongation, will have little or no burnishability because they have no plastic deformation. For example, a cobalt chromium partial dental alloy has 1.5% elongation, but ultimate tensile strength of 870 MPa.

17.7.4 Ductility and malleability

Ductility is the ability of a material to sustain a large permanent deformation under a tensile load up to the point of fracture, or the relative ability of a material to be stretched plastically at room temperature without fracturing. Ductility can be measured by the amount of permanent deformation indicated by the stress-strain curve. Three methods have been reported to measure ductility. These include: (i) percentage elongation after fracture, (ii) reduction in the area of the fractured region, and (iii) the cold bend test.

The percentage of elongation after fracture is the most common method, which compares the increase in length of a wire/rod after fracture, in tension, to its length before fracture. Two marks are made on the wire/rod at a specified distance, said to be the “gauge” length. The standard gauge length for dental material is 51 mm. The wire/rod is then pulled apart under the tensile load, the fracture ends are fitted together, and the length is measured. The ratio of the original length to the increase in length is measured. This ratio, if expressed in a percentage, is called percentage elongation. The second method, that is, the reduction in area in the fractured region, utilizes the necking, or cone-shaped constriction occurs at the fractured end of a ductile wire after rupture under a tensile load. The percentage of decrease in the cross-section area of the fractured end in comparison with the original area of the wire or rod is called a reduction in area. In a cold bend test, material is clamped in a vise and bent around a mandrel of a specified radius. The number of bends to fracture is counted; the greater the number is, the higher is the ductility. Malleability is the ability of a material to sustain stress and not rupture under compression, as in

hammering or rolling into a sheet. Gold is the most malleable and ductile metal, and the second is silver.

17.7.5 Hardness

The definition of hardness varies according to the nature of the material. In mineralogy, the relative hardness of a substance is based on its ability to resist scratching. In metallurgy, and in most other disciplines, the concept of hardness that is most generally accepted is its resistance to indentation. The indentation produced on the surface of a material from an applied force of a sharp point or an abrasive particle results from the interaction of numerous properties. The properties that are related to the hardness of a material are the strength proportional limit and ductility. A hardness test can be applied on the basis of the method of application, size of the indenter, and amount of load applied to the indenter [19].

On the basis of the method of application, a hardness test can be classified into (i) static loading (slowly applied), and (ii) dynamic loading (rapidly applied). On the basis of the size of the indenter, a hardness test can also be classified into two types: (i) macroindentation (large indenter tip), and (ii) microindentation (small indenter tip). Furthermore, on the basis of the amount of load applied to the indenter, the hardness test can be classified into two types: (i) macrohardness (>1 kg load), and (ii) microhardness (<1 kg load). The following tests are performed to measure surface hardness [20]:

1. Barcol test
2. Brinell test
3. Rockwell test
4. Scholl test
5. Vickers test
6. Knoop test

The selection of the test is based on the material to be tested. Knoop and Vickers tests are called microhardness tests, while the Brinell and Rockwell tests are macrohardness tests. The Brinell test is one of the oldest tests to measure hardness of the metals and metallic materials used in dentistry. This test is related to the proportional limit and ultimate tensile strength. In this test, a hardened steel ball is pressed under a specified load into the polished surface of a metal. The load is divided by the area of the projected surface of the indentation, and the quotient is referred to as the Brinell hardness number (BHN). This method is best suited for ductile materials, but not for brittle materials, and is limited to cold-worked materials.

The Rockwell method is a rapid method for determination of hardness of dental materials, and is similar to the Brinell method. In this method, instead of a steel ball, a conical diamond point is used. The depth of the penetration is directly measured by a dial gauge on an instrument. In this method, indentation points with different sizes are available for testing a variety of different materials. Rockwell hardness number (RHN) is designated according to the particular indenter and load applied. This method is also not fit for brittle metals.

The Vickers test is similar to the Brinell, test but instead of a steel ball, a square-based pyramidal indenter is used, and thus a square impression is obtained on the material. This test is fit for measuring hardness of casting gold alloys, as well as tooth structures, and also suitable for determining the hardness of brittle materials. This test has the advantage of measuring hardness of small area and hard materials, while it is limited to materials that require a higher polished surface.

In the Knoop test, a diamond indenting tool is used to cut into a geometric configuration. The impression is rhombic in outline, and the length of the largest diagonal is measured. The Knoop hardness number (KHN) is measured by dividing the projected area by the load value. When the indentation is made, and the indentation is removed, the shape of the Knoop indenter causes elastic recovery of the projected impression to occur along the short diagonal. The stresses are therefore distributed in a matter such that only the dimension of the minor axis is subjected to change by relaxation. This test can determine the hardness of brittle materials, but the material requires a highly polished surface. Knoop hardness numbers of commonly-used dental materials are listed in [Table 17.6](#).

The hardness value is virtually independent of the ductility of the material tested. The load to be used may be varied over a wide range, from 1 g to more than 1 kg, so that values for both hard and soft materials can be obtained by this test.

17.8 Criteria for selection of restorative materials

Selection of the appropriate material for dental restoration has become more and more difficult, owing to the increasing variety of restoration materials. Usually, the choice of the material for a single tooth restoration is mainly based on the following factors:

- i. The physical properties of the presently available restorative materials and their ability to maintain the physical properties of the tooth.
- ii. The ability of the restorative material to maintain its occlusion.
- iii. Relative caries susceptibility.
- iv. Hygienic condition of the mouth.
- v. Friability of enamel and sensitivity of dentin.
- vi. Age, physical condition, and mentality of the patient.
- vii. Cost effectiveness.

Table 17.6 Knoop hardness number of commonly used dental material

Material	KHN
Enamel	343
Dentin	68
Cementum	40
Denture acrylic	21
Zinc phosphate cement	38
Porcelain	460

Physical properties of materials are very important during selection. In this regard, gold is the material of choice, because it can be used in thin sections to protect and reinforce the remaining tooth structure, reestablish the ideal contour, as well as the anatomy, and rebuild the occlusion with high accuracy. Ceramic is the second choice, having the limitation of being too brittle to be designed in thin sections, so that more of the remaining tooth structure must be removed to allow sufficient room for adequate thickness of ceramic. Gold is also a material of choice over ceramics for restoration and maintenance of occlusions. Ceramic has the limitation of abrading opposing enamel, but is desirable to occlude porcelain to porcelain. Composite and glass ionomers are not suitable for restoration and occlusion, because their wear is too high. Hygienic conditions must be maintained during restoration. Gold and ceramic are not a wise selection for patients with habitually unclean mouths. Composite restorations also fail under unhygienic conditions. Amalgam may undergo progressive and severe tarnish and corrosion. Cohesive gold and ceramic veneers are avoidable due to the chances of friability of enamel, while in such cases, composite resins can be used. Neonates or children, older people, and sick and nervous patients cannot tolerate prolonged and difficult restoring operations; for example, by direct gold. For these patients, an amalgam is suitable for posterior teeth, and a composite for anterior teeth. The financial condition of the patient may modify the selection of the restorative material, favoring the extensive use of amalgams. Apart from these, aesthetic factors are also important during the selection of restorative materials. These factors include available space in the buccal area, brightness value of the neighboring teeth, translucency of the neighboring teeth, and degree of discoloration of the abutment tooth.

Visual criteria are very important for the selection of material. If the teeth show high translucency (a high amount of light transmission), a glass-ceramic is favored as the restoration material of choice. It ensures maximum light flux, and therefore crowns exhibit a translucent effect. This translucent effect may further be amplified by the use of translucent veneering ceramics. In contrast, for a deeply opaque reference tooth (with a low amount of light transmission), zirconia may be recommended as a core material due to its material properties, uniform tooth-colored appearance, and in general, reduced translucency. However, for multi-unit fixed dental prostheses (FDPs), additional factors have to be considered for the choice of material. Full-ceramic as well as metal-ceramic crowns may be indicated for single tooth restorations. The five-year survival rate for metal-ceramic crowns is 95.6%, for glass-infiltrated alumina crowns 94.5%, and for reinforced glass-ceramic crowns, 95.4%. Consequently, both full-ceramic, as well as metal-ceramic crowns may be indicated for single-tooth restorations. The failures of all-ceramic FDPs manufactured from weaker ceramics are due to fractures of the framework. The failures of metal-ceramic FDPs occur due to biological and technical complications.

17.9 Conclusion

An important factor in the selection of a dental prosthesis is the mechanical property of the dental material, to ensure that the prosthesis serves its intended functions

effectively, safely, and for a reasonable time period. Mechanical properties explain the ability of dental materials to withstand the various mechanical forces placed on them during use as restorations, impressions, models, appliances, and tools. In general, stress and strain are important mechanical properties representing measurements of: (1) elastic or reversible deformation (i.e., proportional limit, resilience, and modulus of elasticity); (2) plastic or irreversible deformation (e.g., percent elongation and hardness); or (3) a combination of elastic and plastic deformation, such as toughness and yield strength. The potential knowledge of mechanical stability of dental materials allows us to differentiate the potential causes of clinical failures that may be attributed to material deficiencies, dentist errors, technician errors, or patient factors.

References

- [1] P.E. Petersen, Sociobehavioural risk factors in dental caries—international perspectives, *Community Dent. Oral Epidemiol.* 33 (2005) 274–279.
- [2] K.J. Anusavice, Recent developments in restorative dental ceramics, *J. Am. Dent. Assoc.* 24 (1993) 73–84.
- [3] J.E. Frencen, T. Pilot, Y. Songpaisan, P. Phantumvanit, Atraumatic restorative treatment (ART): rationale, technique, and development, *J. Public Health Dent.* 56 (1996) 135–140.
- [4] H. Forss, E. Widstrom, From amalgam to composite: selection of restorative materials and restorative longevity in Finland, *Acta Odontol. Scand.* 59 (2001) 57–62.
- [5] J.A. Griggs, Recent advances in materials for all-ceramic restorations, *Dent. Clin. N. Am.* 51 (3) (2007) 713–727.
- [6] M.A. Rosenblum, A. Schulman, A review of allceramic restorations, *J. Am. Dent. Assoc.* 128 (3) (1997) 297–307.
- [7] V. Deligeorgi, I.A. Mjör, N.H. Wilson, An overview of reasons for the placement and replacement of restorations, *Prim. Dent. Care* 8 (1) (2001) 5–11.
- [8] L. Wang, P.H.P. D'alpino, L.G. Lopes, J.C. Pereira, Mechanical properties of dental restorative materials: relative contribution of laboratory tests, *J. Appl. Oral Sci.* 11 (3) (2003) 162–167.
- [9] J.M. Powers, R.L. Sakaguchi, *Craig's Restorative Dental Materials*, twelfth ed., Elsevier, New Delhi, 2006, pp. 443–464.
- [10] S.O. Hondrum, A review of the strength properties of dental ceramics, *J. Prosthet. Dent.* 67 (6) (1992) 859–865.
- [11] J.R. Kelly, Perspective on strength, *Dent. Mater.* 11 (1995) 103–110.
- [12] R.R. Seghi, T. Daher, A. Caputo, Relative flexural strength of dental restorative ceramics, *Dent. Mater.* 6 (1990) 181–184.
- [13] R.R. Seghi, J.A. Sorensen, Relative flexural strength of six new ceramic materials, *Int. J. Prosthodont.* 8 (1995) 239–246.
- [14] D.R. Carter, W.E. Caler, D.M. Spengler, V.H. Frankel, Uniaxial fatigue of human cortical bone. The influence of tissue physical characteristics, *J. Biomech.* 14 (7) (1981) 461–470.
- [15] R.K. Nalla, V. Imbeni, J.H. Kinney, M. Staninec, S.J. Marshall, R.O. Ritchie, In vitro fatigue behavior of human dentin with implications for life prediction, *J. Biomed. Mater. Res.* A 66 (1) (2003) 10–20.
- [16] M. Braem, P. Lambrechts, D.V. Van, G. Vanherle, The impact of composite structure on its elastic response, *J. Dent. Res.* 65 (1986) 648–653.

- [17] M.H. Atala, E.B. Gul, How to strengthen dental ceramics, *Int. J. Dent. Sci. Res.* 3 (1) (2015) 24–27.
- [18] M.F. Ashby, D.R.H. Jones, *Engineering Materials I: An Introduction to Their Properties and Applications*, Pergamon Press, Inc, New York, 1980.
- [19] M.S. Bobji, S.K. Biswas, Deconvolution of hardness from data obtained from nano-indentation of rough surfaces, *J. Mater. Res.* 14 (6) (1999) 2259–2268.
- [20] T. Attin, P.R. Schmidlin, F. Wegehaupt, A. Wiegand, Influence of study design on the impact off bleaching agents on dental enamel microhardness: a review, *Dent. Mater.* 25 (2) (2009) 143–157.
- [21] E. Mahoney, A. Holt, M. Swain, N. Kilpatrick, The hardness and modulus of elasticity of primary molar teeth: an ultra-micro-indentation study, *J. Dent.* 28 (8) (2000) 589–594.
- [22] S.F. Ang, T. Scholz, A. Klocke, G.A. Schneider, Determination of the elastic/plastic transition of human enamel by nanoindentation, *Dent. Mater.* 25 (11) (2009) 1403–1410.
- [23] J.L. Cuy, A.B. Mann, K.J. Livi, M.F. Teaford, T.P. Weihs, Nanoindentation mapping of the mechanical properties of human molar tooth enamel, *Arch. Oral Biol.* 47 (4) (2002) 281–291.
- [24] L. Angker, M.V. Swain, N. Kilpatrick, Characterising the micro-mechanical behaviour of the carious dentine of primary teeth using nano-indentation, *J. Biomech.* 38 (7) (2005) 1535–1542.
- [25] D. Ziskind, M. Hasday, S.R. Cohen, et al., Young's modulus of peritubular and intertubular human dentin by nano-indentation tests, *J. Struct. Biol.* 174 (1) (2011) 23–30.
- [26] L. Angker, C. Nijhof, M.V. Swain, N. Kilpatrick, Correlating the mechanical properties to the mineral content of carious dentine—a comparative study using an ultra-micro indentation system (UMIS) and SEM–BSE signals, *Arch. Oral Biol.* 49 (5) (2004) 369–378.
- [27] J.H. Kinney, J. Gladden, G.W. Marshall, S.J. Marshall, J.H. So, J.D. Maynard, Resonant ultrasound spectroscopy measurements of the elastic constants of human dentin, *J. Biomech.* 37 (4) (2004) 437–441.