Dental Materials and Their Selection

Fourth Edition

Edited by

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Preface

This fourth edition of *Dental Materials and Their Selection* continues the objective of presenting a concise consensus of information on basic and applied dental materials. This text is envisioned as a review of basic principles and an update on current dental products for predoctoral and graduate dental students along with dental practitioners, dental hygienists, and dental technicians. Keeping this text comprehensive yet concise is a challenge. The field of dental materials has grown significantly, but the time available for teaching and studying this subject has not. Therefore, the approach maintained since the very first edition of this book has been to stress basic principles and limit the coverage of historic techniques and materials in order to keep the size of the book relatively constant.

The main purpose of this fourth edition is to update chapters, especially on the following topics: color and appearance, surface phenomena and adhesion to tooth structure, polymeric restorative materials, dental amalgams, dental porcelains, orthodontic wires, and implant and bone augmentation materials. This edition also includes updated references and revision of some clinical decision scenarios. I welcome and appreciate comments on the technical coverage and educational application of this edition.

I would like to thank the many people who, in addition to the contributors, have made this fourth edition possible: Elizabeth Rodriguiz for her invaluable efforts in manuscript preparation, Chris Jung for his excellent illustrations (especially Fig 3-19), and Dr Sumant Ram for his indispensable assistance. I would also like to welcome Drs Joseph B. Dennison and Peter X. Ma as authors in this fourth edition with their contributions on polymeric restorative materials and polymers and polymerization, respectively. Several contributors to the third edition are also recognized for their valuable contributions: Dr Raymond L. Bertolotti, Dr Gerald N. Glickman, Dr Eugene F. Huget, Dr Ann-Marie L. Neme, and Dr Denis C. Smith. Finally, I want to acknowledge the staff at Quintessence for their expert assistance in preparing this book for publication.

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Physical Properties and Biocompatibility

Physical properties determine how materials respond to their environment, and *biocompatibility* relates to the effect a material has on the surrounding tissue. Physical properties are classified according to the scheme in Fig 2-1.

Mechanical Properties

The concepts of stress, strain, modulus of elasticity, plastic deformation, and other properties were introduced in chapter 1, and are discussed here in greater depth.

Stress

Consider again the block of material described in the previous chapter (see Fig 1-2), which measured 1.0 cm $\times$ 1.0 cm = 1.0 cm² (0.394 in $\times$ 0.394 in = 0.155 in²) in cross section and was subjected to a 10.0-kN (2,250-lb) load. As was pointed out previously, the stress experienced by that block is 100 MPa (14,500 psi). Stress (σ) is the force (F) divided by the cross-sectional area (A):

$$\sigma = \frac{F}{A}$$

Consider now a similar block but with smaller dimensions—0.50 cm $\times$ 0.50 cm (0.197 in $\times$ 0.197 in) in cross section (area = 0.25 cm² or 0.0388 in²). If this new block is subjected to the same 10.0-kN (2,250-lb) tensile load, the stress is 10 kN (2,250 lb) divided by 0.25 cm² (0.0388 in²) = 400 MPa (58,000 psi).

The usefulness of the concept of **stress** is apparent. It is not sufficient merely to state the load or force that is being applied to a dental material, because the stress

that is produced in the material depends just as much on the cross-sectional area on which the load is acting as it does on the load itself. For instance, if the block that measured 1.0 cm $\times$ 1.0 cm in cross section is subjected to a load of 40 kN (9,000 lb) instead of 10 kN (2,250 lb), the stress is 400 MPa (58,000 psi). If the cross-sectional area is made four times smaller (one-fourth as large), or if the load is made four times larger, the stress is increased by a factor of four. Thus, the stress is said to be inversely proportional to the cross-sectional area and directly proportional to the load.

The basic types of stresses produced in dental structures under a force are **tensile**, **compressive**, and **shear**. All three are active in a beam loaded in the center. If the value of these stresses exceeds the strength of the material, the structure will fail. It is therefore important to know the strength values of materials. It is rare that an object will be subject to the pure tensile, compressive, or shear stresses experienced by test specimens in a materials testing laboratory. As shown in Fig 2-2, however, wherever bending forces are present, tensile stresses are also present in critical areas, which could result in failure.

Strain

When a block of material is subjected to a tensile stress as described in the preceding paragraph, it temporarily becomes longer. This temporary increase in length is called **strain**. The following examples illustrate conditions of strain. Consider the block with the cross section of 0.50 cm $\times$ 0.50 cm = 0.25 cm² (0.197 in $\times$ 0.197 in = 0.0388 in²), and assume it has a length of 25.0 cm (9.843 in) when no load is applied. If the length is measured while the 10.0-kN (2,250-lb) tensile load is being applied and the

◉ Glossary

biocompatible A material is considered biocompatible if it does not produce harmful or toxic reactions in the tissues it contacts or adverse systemic reactions as a result of elements, ions, and/or compounds it releases.

biomaterial A material that is designed to be implanted in the body for the repair or replacement of a diseased or damaged tissue.

brittle Characteristic of a material that tends to fracture without appreciable plastic deformation.

compressive stress Two forces applied toward one another in the same straight line.

ductile Characteristic of a material to be plastically strained in tension.

elasticity Ability to sustain deformation without permanent change in size or shape.

electric resistivity Ability of a material to resist conduction of an electric current.

elongation Overall deformation (elastic + plastic) as a result of tensile force application.

fatigue Tendency to fracture under cyclic stresses.

fracture toughness The resistance of a material to brittle fracture when a crack is present in (or at the surface of) the material. Fracture toughness is a measure of the amount of energy absorbed during propagation of a crack. Materials with lower fracture toughness are more prone to fail catastrophically during service than those with higher fracture toughness.

hardness Resistance to permanent indentation on the surface.

linear coefficient of thermal expansion Change in length per unit of original length for a 1°C temperature change.

modulus of elasticity Stiffness of a material within the elastic range. Numerically, it is the ratio of stress to strain.

percolation The pumping of oral fluids in and out at the tooth-restoration interface as the restoration contracts and expands with temperature changes. Percolation occurs when the thermal expansion coefficient of the restoration is markedly different from that of tooth structure.

proportional limit (elastic limit) The maximum stress at which the straight-line relationship between stress and strain is valid.

resilience Energy needed to deform a material to the proportional limit.

shear stress Two forces applied toward one another but not in the same straight line.

strain (nominal) Change in length per unit of original length.

stress (nominal) Force per unit area.

tensile stress Two forces applied away from one another in the same straight line.

thermal conductivity The quantity of heat passing through a material 1 cm thick with a cross section of 1 cm^2, having a temperature difference of 1°C.

thermal diffusivity Measure of the heat transfer of a material in the time-dependent state.

toughness Amount of energy needed for fracture.

ultimate tensile strength The maximum strength obtained based on the original dimensions of the sample.

viscoelastic Having both elastic and viscous properties.

viscous Resistant to flow (referring to a fluid).

work hardening The increase in strength and hardness, with accompanying decrease in ductility, that occurs in a ductile metal as it is plastically deformed. Also called *strain hardening*.

yield strength Strength measured at the stress at which a small amount of plastic strain occurs. Also called *yield point*.

◉ Discussion Questions

1. What is galvanic action in the mouth, and how can it be minimized?

2. Why is thermal diffusivity more relevant to insulation of the pulp than thermal conductivity?

3. Explain why fixed partial dentures may fail under tensile stresses due to biting forces that appear to be compressive.

4. Explain how two materials with the same measured tensile strength could have markedly different fracture toughness values.

5. How is the leakage of oral fluids around a composite restoration related to the coefficient of thermal expansion and temperature changes in the mouth?

6. Why is time so important in the behavior of viscoelastic materials such as impression materials and waxes?

7. Why is there a hierarchy of biocompatibility tests?

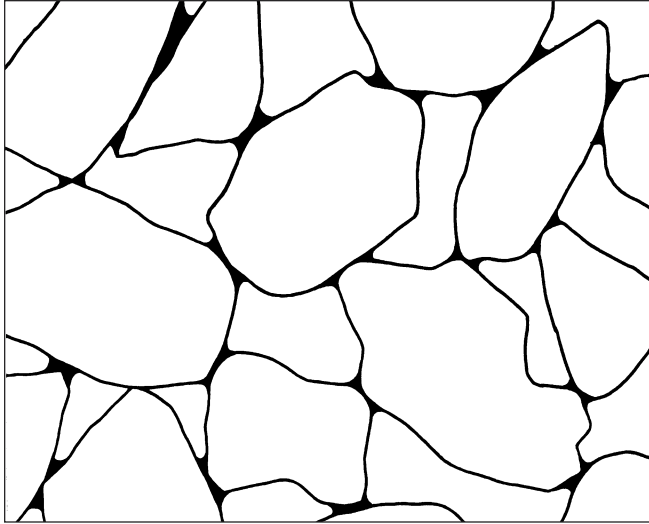8. Why can biocompatibility tests be difficult to interpret?

Fig 15-2 Biscuit stage of vitreous sintering, involving flow of glass to form bridges between particles. (Reprinted with permission from Van Vlack.[3])
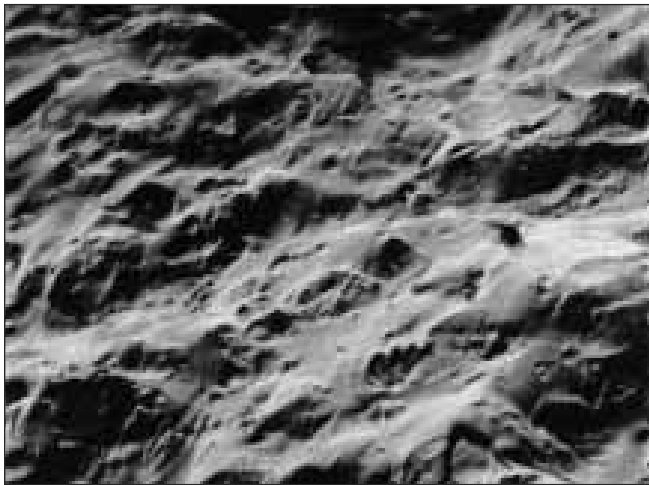

Fig 15-3a Fracture surface of gingival porcelain fired under vacuum (scanning electron micrograph; original magnification ×300). (Reprinted with permission from Meyer et al.[4])

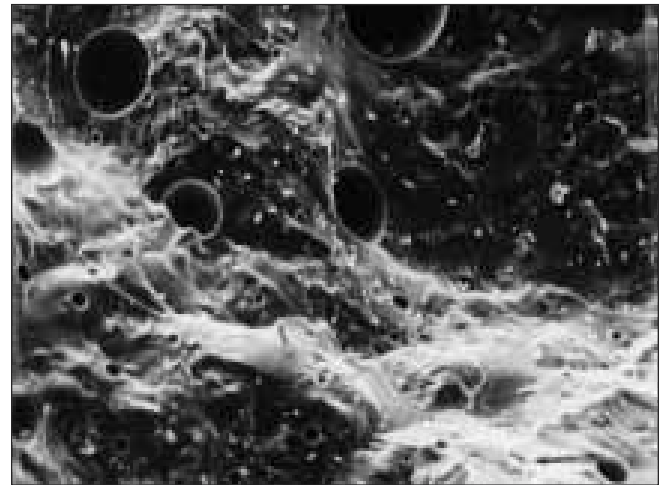**Fig 15-3b** Fracture surface of gingival porcelain fired under normal atmospheric pressure (scanning electron micrograph; original magnification ×300). (Reprinted with permission from Meyer et al.[4])

cent particles bond together in a process called **sintering**. Although there is no meeting of the porcelain powder particles, they join together by flow on contact as a result of surface energy (Fig 15-2).

Firing in a vacuum furnace greatly reduces the porosity of the final product, as shown in Figs 15-3a and 15-3b. The first firing of porcelain is called the **bisque** or *biscuit bake*. After the incisal layer is added, the porcelain is brought to the final stage, called the **glaze bake**. When the glazing temperature of the porcelain is reached, a layer of glass is formed on the surface. After glazing, the crown is removed from the furnace and cooled under an inverted glass or beaker. An alterna-

tive approach is to add a thin layer of a low-fusing glass or glaze to the surface and fire to the flow temperature of the glaze.

Properties

Porcelain enamels have a **vitreous** structure consisting of an irregular network of silica produced by the presence of large alkali metal ions, such as sodium, potassium, and lithium (Fig 15-4a). This amorphous structure produces physical properties typical of a glass, including brittleness and lack of a definite melting temperature. Glasses are brittle due to their irregular structure and the absence of slip planes, which are present in a
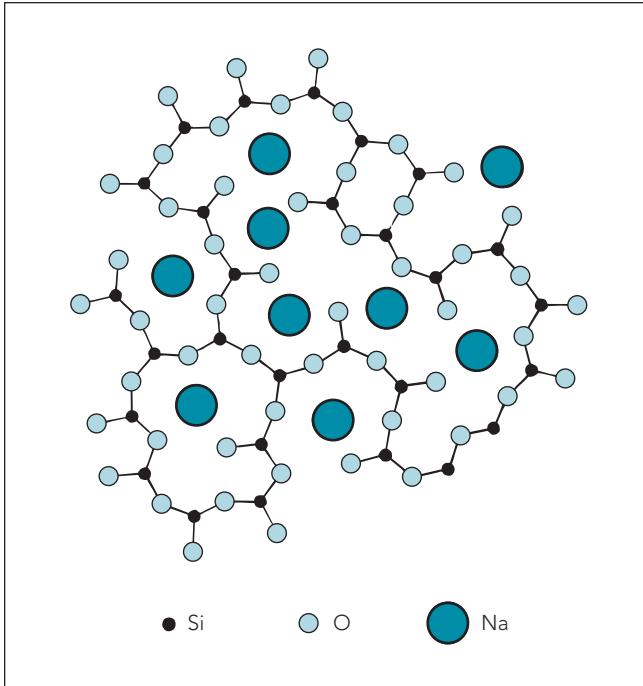
Fig 15-4a Irregularity of glass structure (silicon and oxygen) due to the presence of large alkali cations (sodium). (Modified with permission from Warren.[5])
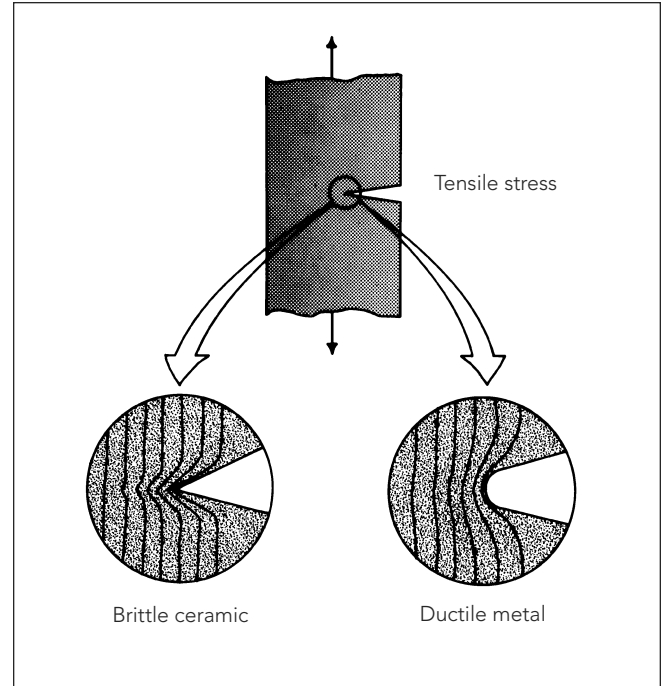

Fig 15-4b Effects of tensile forces on crack propagation in ceramics and metals, resulting in brittleness and ductility, respectively.

true crystalline material. The strength of glasses and brittle materials is governed by the presence of small flaws or cracks. When stressed in tension, according to the **crack propagation theory**, small flaws tend to open up and propagate, resulting in a low tensile strength. This is less of a factor with ductile metals because the stress concentration around the tip of the flaws is reduced by elongation of the metal, as illustrated in Fig 15-4b. However, glasses are much stronger in compression, because **residual compressive stresses** tend to close up flaws. Therefore, the tensile strengths of vitreous dental porcelains are around 35 MPa (5,000 psi) as compared with compressive strengths of 517 MPa (75,000 psi).

The strength of dental porcelains is traditionally tested in flexure as a beam and reported as **modulus of rupture**. The modulus of rupture of a vitreous body or enamel porcelain is about 90 MPa (13,000 psi). The strengths of vacuum-fired porcelains are higher due to fewer flaws.

Vitreous dental porcelains do not have a definite melting temperature but undergo a gradual decrease in viscosity when heated. A sharp decrease in viscosity occurs around the **glass-transition temperature** (T_g), as shown in Fig 15-5. Below T_g, the glass has the properties of a solid. Above T_g, glass flows more readily, and vitreous sintering takes place.

A typical TE curve of a porcelain bar is shown in Fig 15-6. The TE is linear up to around T_g, but above T_g there is a rapid increase in the rate of expansion when the glass has a more liquid structure. If heating is continued, the bar will reach the softening temperature and collapse. The TE of dental porcelains for bonding to metals is especially important in relation to the TE of the metal involved. Generally, the metal and porcelain should be matched in coefficients of TE values. If the TE curves of the metal and porcelain are too far apart, undesirable thermal stresses will result in fracture of the porcelain, which is the weaker material. The porcelain and metal are therefore said to be incompatible.