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Mechanisms and risk factors for fracture predilection in endodontically treated teeth

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The prognosis of root-filled teeth depends not only on the success of the endodontic treatment but also on the amount of remaining dentine tissue, and the nature of final restoration. Fractures of restored endodontically treated teeth are a common occurrence in clinical practice. This article outlines the mechanisms and risk factors for fracture predilection in endodontically treated teeth. Different mechanisms of fracture resistance in dentine and the biomechanical causes of fracture predilection in restored endodontically treated teeth are described. Furthermore, dentinal, restorative, chemical, microbial, and age-induced factors that predispose restored endodontically treated teeth to fracture are also reviewed.

'The structure and composition of teeth is perfectly adapted to the functional demands of the mouth, and are superior in comparison to any artificial material. So first of all, do no harm.'

Tooth fracture has been described as a major problem in dentistry, and is the third most common cause of tooth loss after dental caries and periodontal disease (1, 2). Many *in vivo* studies have highlighted endodontic treatment as the major etiological factor for tooth fracture (3–9). Generally, an endodontically treated tooth undergoes coronal and radicular tissue loss due to prior pathology, endodontic treatment, and/or restorative procedures. There is evidence that these teeth have reduced levels of proprioception (9, 10), which could impair normal protective reflexes. This finding, however, cannot explain the reduction in mechanical properties of dentine *in vitro*. Even though loss of tooth structure is an important factor that diminishes fracture resistance in endodontically treated tooth, this reason does not fully explain why endodontically treated teeth with minimal loss of tooth structure become susceptible to fracture, particularly over time.

Historically, the increased susceptibility of fracture in endodontically treated teeth had been attributed to the postulated increased brittleness of dentine due to loss of moisture (11). This hypothesis was first put forward by G.V. Black, and later given credence by Helfer et al. (11), who reported that the moisture content of dentine from endodontically treated teeth was about 9% less than teeth with vital pulp. However, other studies contradict this view (12). Papa et al. (13) reported insignificant difference in the moisture content between endodontically treated teeth and teeth with vital pulp. They emphasized the importance of conserving the bulk of dentine to maintain the structural integrity of post-endodontically restored teeth (14). Other studies have also emphasized that the loss of tooth structure is the key reason for the increase in fracture predilection of endodontically treated teeth (15, 16). These contradictory views in literature are, to some extent, a result of direct comparison of root-filled teeth with teeth with vital pulp, and due to the difficulties in establishing appropriate controls and standardized test procedures. Specimen shape that represents the bulk structural behavior of dentine and the effect of hydration in dentine were also overlooked (17).

Traditionally, biomechanical experiments conducted to determine the fracture resistances of intact and endodontically treated teeth have tended to focus on the strength of the teeth (16). A number of such studies have suggested that there are no major differences in the mechanical properties of teeth with vital pulp and root-filled teeth (12, 14). As pointed out by Kahler et al. (17) the inference from such studies has been that the apparent brittleness of root-filled teeth would manifest itself as changes in strength or modulus of elasticity. It should be noted that strength is merely the ability to resist deformation or show stiffness to loads, measured under well-controlled situations. On the other hand, toughness is the ability to absorb energy without fracturing. In material science, stiffness and toughness are mechanical properties that cannot be increased together indefinitely. It is inherently difficult in artificial materials with very high initial stiffness (strength) to accommodate a long plastic yield (toughness) (18, 19). Natural mineralized tissues, such as dentine, are a result of long-term optimization, controlled by the selection processes of evolution. Therefore, understanding the mechanisms of fracture resistance operating in the dental tissues would provide better insight into the risk factors that predispose endodontically treated teeth to fracture.

Fractures in endodontically treated teeth have been understood to be multifactorial in origin. The causes of fracture in endodontically treated teeth can be broadly classified as iatrogenic and non-iatrogenic, and are outlined in Fig. 1. In this article, the mechanisms and risk factors for fracture predilection in endodontically treated teeth are described under the following headings:

1. *The mechanisms of fracture resistance in dentine*

(1.1) Biomaterial considerations of dentine substrate
(1.2) Biomechanical considerations in intact and post–core restored teeth

2. *The risk factors for fracture predilection in endodontically treated teeth*

(2.1) Chemical factors: effects of endodontic irrigants and medicaments on dentine
(2.2) Microbial factors: effects of bacteria-dentine interaction
(2.3) Dentine factors: effects of tooth structural loss
(2.4) Restorative factors: effects of post and core restorations
(2.5) Age factors: effects of age changes in dentine

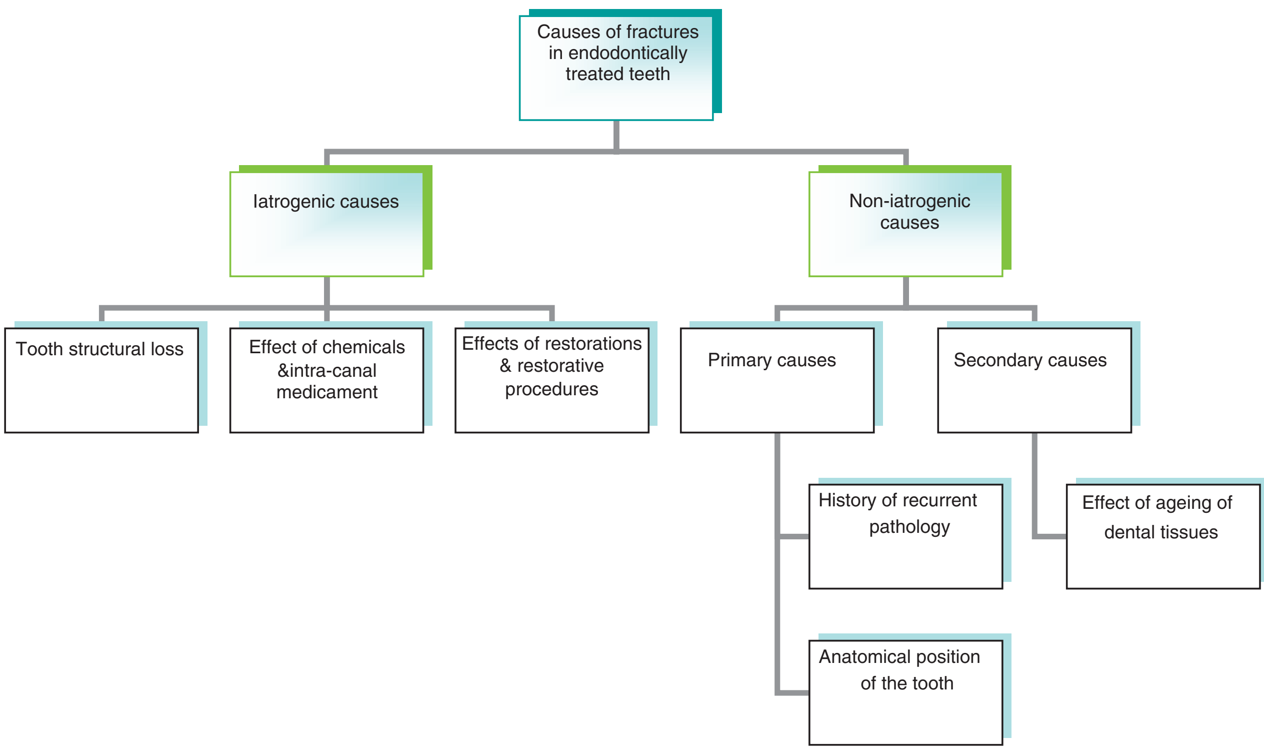


Fig. 1. Outline of the causes of fracture in endodontically treated teeth.

Mechanisms of fracture resistance in dentine

Biomaterial considerations of dentine substrate

Dentine is a natural, hydrated, mineralized hard tissue that forms the major bulk of a tooth. It consists of thousands of microscopic tubules with diameter ranging from 0.5to 4.0 μm, and the typical density of dentinal tubules ranges from 10 000 to 96 000 tubules per mm^2 (20). Mature dentine is a composite material made up of an organic fraction (30 wt%), which is mainly collagen and an inter-penetrant inorganic fraction (60 wt%) and water (10 wt%) (21–26). The inorganic phase in dentine is mainly composed of poorly crystalline-carbonated hydroxyapatite with a needle and/or plate like morphology (10 × 50 nm), which exists both within the collagen fibrils (intra-fibrillarly mineralized) and between fibers (inter-fibrillarly mineralized) on a nanometric scale. Ninety percent of the organic phase is collagen, which is exclusively Type I (27, 28). Type I collagen is a strong, three-dimensional fibrous polymer that usually exists in an aqueous biological environment. It is often associated with proteoglycans, which contains a large amount of bound water (19, 29). The water content of dentine is believed to vary with location. The general conjecture is that there are two types of water in dentine. One type is associated with the apatite crystal of the inorganic phase, and collageneous and non-collageneous matrix proteins of the organic phase. They are mostly 'tightly' bound in nature. The second type is the free or 'unbound' water, and this type of water fills the dentinal tubules and other porosities in the dentine matrix. The free water is associated with inorganic ions such as calcium and phosphate and aids in their transport within the dentine matrix. Free water can be removed by heating at 100°C, but bound water can only be substantially removed by heating at 600°C (30). The precise distribution and effect of different types of water on dentine structure have not been extensively explored in the past (31). The role of different constituents on the mechanical characteristics of dentine is shown in Fig. 2.

An interesting phenomenon that has intrigued material scientists is the ability of biological materials, such as dentine, to exhibit superior strength, and

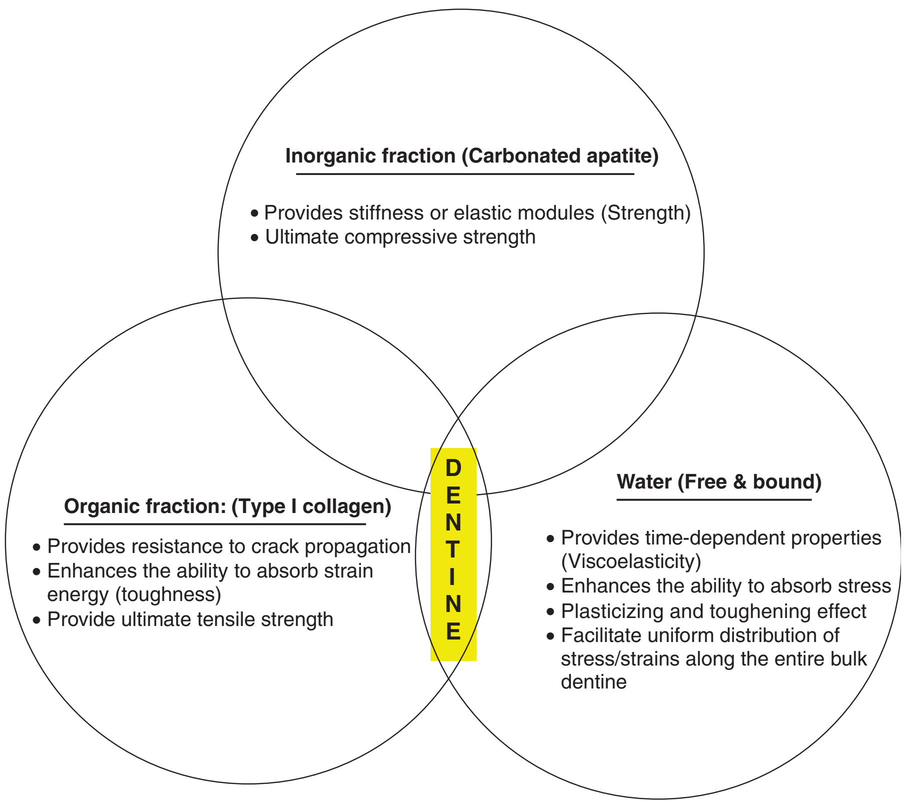


Fig. 2. The role of different constituents on the mechanical integrity of structural dentine.

fracture toughness despite the relatively poor mechanical properties of their constituent elements. On the one hand, there is the mineral component, which is brittle and fragile, while on the other the protein component is soft and tender. Protein plays a crucial role not only as a template for mineral biodeposition but also in the mechanical properties of biological materials. Basically, the protein matrix behaves like a soft wrap around the mineral platelets and protects them from the peak stresses caused by the external load and homogenizes stress distribution within the composite structure. At the most elementary structure level, natural biocomposites exhibit a generic microstructure consisting of staggered mineral bricks shown in Fig. 3A. It was proposed that under an applied tensile stress, the mineral platelets carry the tensile load while the protein matrix transfers the load between mineral crystals via shear (32). The strength of the protein phase in a biological material is amplified by the large aspect ratio of mineral platelets. Besides, the larger volume concentration of protein significantly reduces impact damage to the protein–mineral interface (Fig. 3B). An optimum balance between stiffness and dynamic toughness is crucial for the mechanical stability of a biological structure such as dentine (19).

The elastic modulus or Young's modulus of a material is defined as the ratio of stress to strain within the elastic limit in a stress–strain curve. The elastic modulus is an indicator of stiffness of a material. The ultimate strength of a material is the maximum stress that a material can withstand before failure. It is important to note that the above parameters are different from toughness, which is the true indicator of a material's ability to resist fracture. Toughness is defined as the total energy absorbed by a structure before it fractures, and can be determined by calculating the area under the stress–strain curve (strain energy). A tough material is found to withstand larger strains before failure (Fig. 4). The main determinant of stiffness in dental hard tissue is the mineral concentration in the tissue. During mineralization, when more and more mineral component displaces water, the hard tissue becomes stiffer (19). It is observed that high levels of mineralization result in high values of elastic modulus and static strength, while also resulting in low values of strain energy. A low value of strain energy renders the highly mineralized tissue more brittle (18). The toughness and/or increased strain energy is provided by the organic fraction, particularly the collagen in dentine

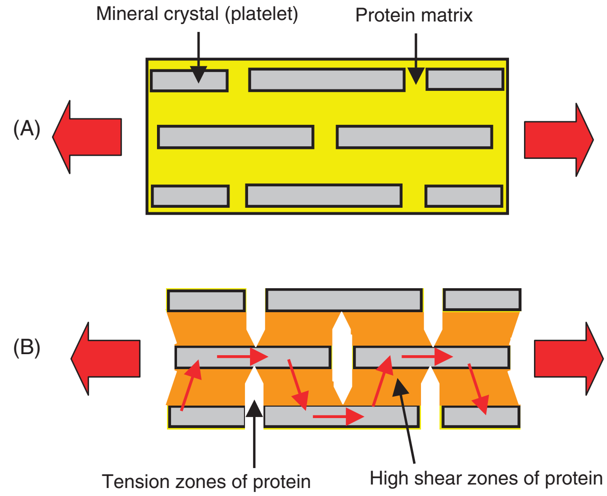


Fig. 3. Schematic diagram showing a model of natural biocomposites. (A) Shows staggered mineral crystals embedded in protein matrix. (B) A simplified model showing the load-carrying structure of the mineral – protein composites. Most of the load is carried by the mineral platelets whereas the protein transfers load via the high shear zones between mineral platelets (32).

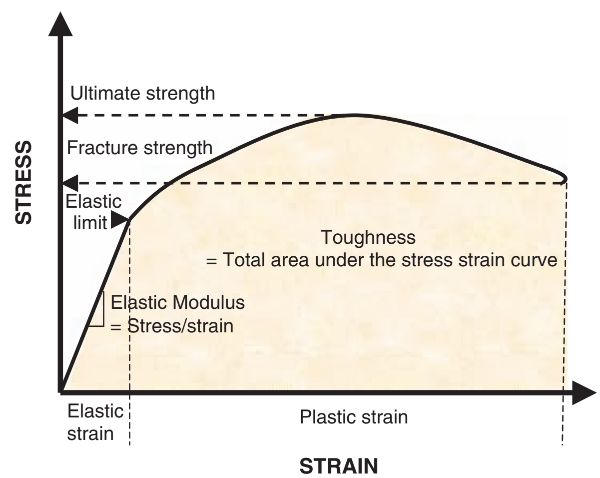


Fig. 4. A schematic stress–strain curve denoting different mechanical parameters.

(33). The degree of mineralization of the collagen substrate in dentine varies continuously with location, and this gradient in the mineralization is believed to optimize the functionality of the overall tooth structure (26). It should be realized that stiffness is different from hardness. Hardness is defined as the resistance of a material to permanent deformation, usually to penetration by an indenter. Hardness is measured using indentation methods that are quite different from the methods used to measure stiffness. No obvious relationship has been observed between mineralization and hardness in dentine (33).

Collagen is a structural protein and the mechanical properties of this material are closely related to the molecular mechanisms rather than to its overall chemistry or morphology (18). Dry collagen is brittle and stiff, with an elastic modulus of about 6 GPa. Addition of water softens the collagen progressively. The interaction of water and dentine matrix has been found to occur in a well-defined manner (18). It has been demonstrated that a mono-layer of water molecules will be adsorbed to the surface of hydroxyapatite by hydrogen bonds. Additional water would be held by weak van der Waals forces (31). The most strongly interacting water molecule is probably incorporated as an integral part of the triple helix of the collagen structure. It is usually incorporated as two water molecules for each tripeptide. When the water content exceeds two molecules per tripeptide, the molecule starts to swell laterally. At this level of hydration, water is said to act as a plasticizer, keeping the matrix soft and pliable. The collagen fibrils of dentin collagen are made up of smaller microfibrills separated by spaces filled with water (18). Dehydration leads to the loss of these interfibrillar spaces and shrinkage of the overall diameter of the fibrils. As the collagen polypeptide chains contact each other, they can form a variety of molecular associations that cannot be formed in an aqueous environment. For example, more van der Waals forces could develop between collagen peptides. Groups capable of forming interpeptide hydrogen bonds but previously unable to do so due to the preferential H-bonding with water could form H-bonding in the absence of water. These interpeptide forces tend to stabilize the structure of dried collagen and increase its stiffness (18). Interest in the role of water in the mechanical properties of the dentine has been rekindled lately. It is realized that water plays a pivotal role in the mineralization, various physiological processes, and biomechanical characteristics of calcified tissues such as dentine (34, 35).

The nanoindentation technique makes use of a much smaller indenting stylus than conventional microhardness testers, so it is possible to sample material properties with smaller applied loads and finer spatial resolution. Moreover, from the load displacement behavior, it is possible to determine the elastic modulus of the material. Kinney et al. (33) used nanoindentation-based experiments and suggested that dry dentine exhibited an elastic modulus of 23.9 GPa, while wet dentine exhibited an elastic modulus of 20 GPa. The same groups also calculated the isotropic elastic modulus of dry dentine to be 28.3 GPa and wet dentine to be 24.4 GPa using resonant ultrasound spectroscopy (RUS) (36). Jameson et al. (37) showed the mean weight loss upon dehydration of dentine bars at 20°C and 50% relative humidity after 7 days to be 3.33%, which equated to a water loss of around 30% of the total water in dentine. This water was fully restorable by rehydration in deionized water. Further, in this study, 85% of the total water lost during dehydration took place in the first 30 min. The water loss by this approach is thought to be from the free water present in the dentine tubules, porosities, and dentine surface (37, 38). A significant increase in stiffness, and a decrease in toughness of dentine were also reported after dehydration at 20°C for 7 days (39). Kahler et al. (17) studied the work of fracture and fracture patterns in bovine dentine and found fully hydrated dentine to be significantly tougher than dentine dehydrated at 22°C and 50% relative humidity for 7 days. The hardness of normal dentine did not appear to depend on water content. Nevertheless, several studies have highlighted time-dependent properties or viscoelastic behavior in dentine. Viscoelastic characteristics in dentine include (1) an increase in strain with time when stress is held constant (creep); (2) a decrease in stress with time when strain is held constant (stress relaxation); (3) the stiffness depends on the rate at which the load is applied; (4) hysteresis (a phase lag) occurs if cyclic loading is applied, leading to dissipation of mechanical energy; (5) acoustic wave experience attenuation; (6) rebound of an object following an impact is less than 100%; and (7) during rolling, frictional resistance occurs (40, 41). Loss of free water will compromise all characteristics of visoelastic behavior (34).

The mechanical properties of mineralized tissues can be studied as localized material properties or bulk structural properties. Localized material or tissue-level properties of mineralized tissues are determined by conducting standard mechanical tests on uniformly shaped and/or sized tissue samples. These tissue-level properties are relatively independent of the geometry of the tissue. The bulk structural properties of a tissue are determined by examining the mechanical behavior of the mineralized tissue as an anatomical unit (18). A bulk dentine structure will include pulp space and

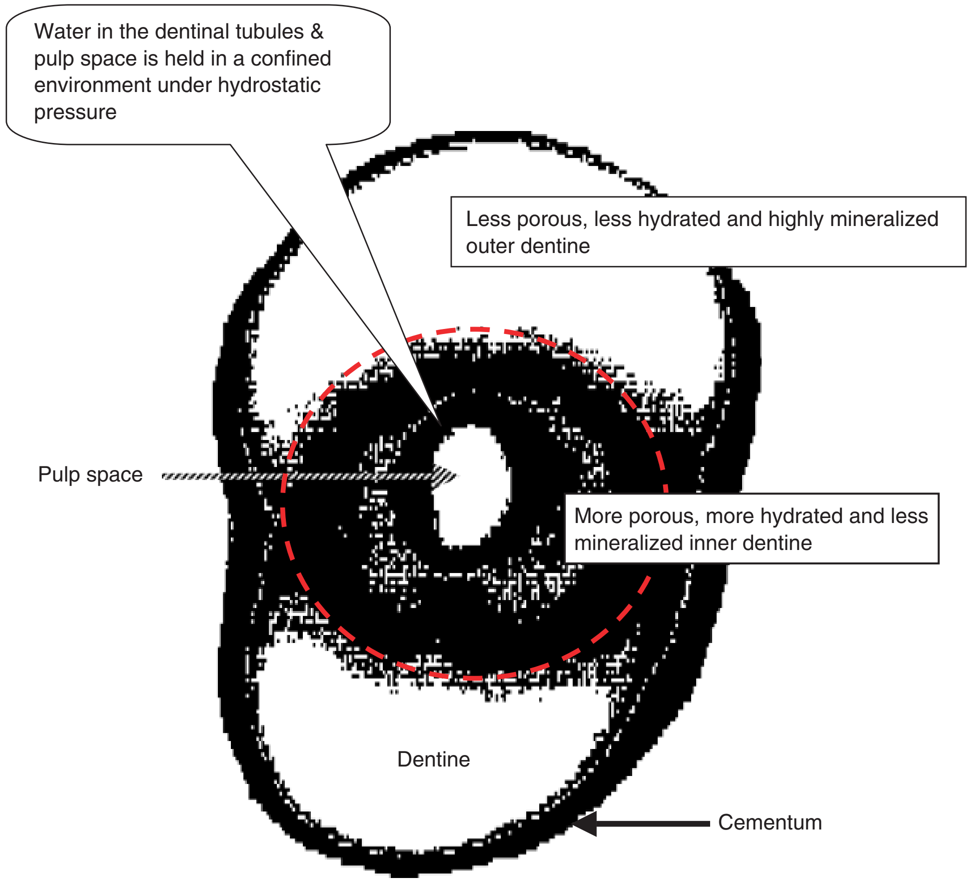


Fig. 5. A cross-sectional view of root dentine highlighting the confined nature of free water within bulk dentine.

dentinal tubules, which are filled with water at a particular hydrostatic pressure. The highly mineralized peritubular dentine, and the less permeable enamel and cementum on the outer aspect of bulk dentine confer a 'confined environment' to the free water in the dentinal tubules and pulp space (38). The biomechanical response of fully hydrated and dehydrated bulk dentine (24°C, 55% relative humidity for 72 h) has been studied using moiré interferometry. The entire facio-lingual section (plano-parallel sagittal section) of single rooted anterior teeth was used. These samples allow maintenance of the major bulk of dentine structure in the facio-lingual plane, and at the same time includes the dentinal tubules, pulp space, outer enamel, and cementum within the specimens. This type of sample preparation aided in retaining free water in a confined environment (Fig. 5). When compressive loads were applied on such fully hydrated dentine specimens, the free water in dentinal tubules and pulp space resulted in a stress–strain response characteristic of tough material, while the loss of this free water resulted in a response characteristic of brittle material. In addition, the free water in the dentinal tubules and root canal lumen was observed to facilitate a homogenous lateral strain transfer within the bulk dentine. This may be due to the micromovement of free water within the dentinal tubules (42). Pashley had earlier suggested that the fluid-filled dentinal tubules could function to hydraulically transfer and dissipate occlusal forces applied to teeth (23). Kinney et al. (25), from a theoretical mechanics perspective, have suggested that the structural stability of dentine is not only the function of mineralization but also its moisture content.

Kruzic et al. (43) studied the fracture toughness properties of dentine, based on resistance-curve (R-curve) behavior i.e. fracture resistance increasing with crack extension, particularly with different toughening mechanisms operating in the dentine. On the basis of this fracture mechanics based experiments, two types of toughening mechanisms have been suggested in dentine: (i) *intrinsic* and (ii) *extrinsic*. The *intrinsic toughening mechanisms* operate ahead of the crack tip and act to enhance the material's inherent resistance to microstructural damage and cracking. The *extrinsic toughening mechanisms* operate primarily behind the crack tip by promoting crack-tip shielding, which reduces the local stress intensity experienced at the crack tip. Basically, toughness is increased by mechanisms that increase the amount of energy required for fracture or methods that prevent strain energy from reaching the crack tip. It is suggested that the viscous effects within the material will slow down the rate of delivery of energy to the crack tip so that the crack can be propagated only slowly and with difficulty. The movement of water from one site to another within the material falls in this category and seems to be a mechanism for toughening dentine (18). Microcracking, crack blunting, and crack bridging by ligament formation and collagen fibrils are examples of extrinsic toughening mechanisms in dentine (Fig. 6) (43). Microcracking causes dilation and increase the compliance of the region surrounding the crack. Sharpness of the crack tip focuses strain energy onto the next susceptible bond and is an important factor governing fracture propagation. Crack blunting causes the stresses at the crack tip to be defocused. In crack bridging, as the crack opens, fibers or filaments extend across it, dissipating energy by their own deformation or by friction as they pull out from the bulk of the material (43, 18). In addition, strain energy may not be transmitted to the crack tip if the shear stiffness of the matrix material is too low (evidenced by a J-shaped stress–strain curve, commonly found in soft tissues) (18).

Intrinsic mechanisms, such as crack blunting, tend to affect the initiation toughness, whereas extrinsic mechanisms, such as crack bridging, promote crack-growth toughness. Hydration also increases the fracture toughness of dentin by extensive crack blunting, which elevates the crack-initiation toughness, and additionally from enhanced uncracked-ligament bridging, which promotes the crack-growth toughness. In comparison, dehydrated dentin shows little blunting, which results in a lower crack-initiation toughness; however, with crack extension, significant crack bridging occurs, although the bridging does not develop as quickly as in the hydrated state (43, 44). These investigations highlight that the collagen microstructure and the water of hydration are the foremost factors that contribute to the fracture toughness of dentine.

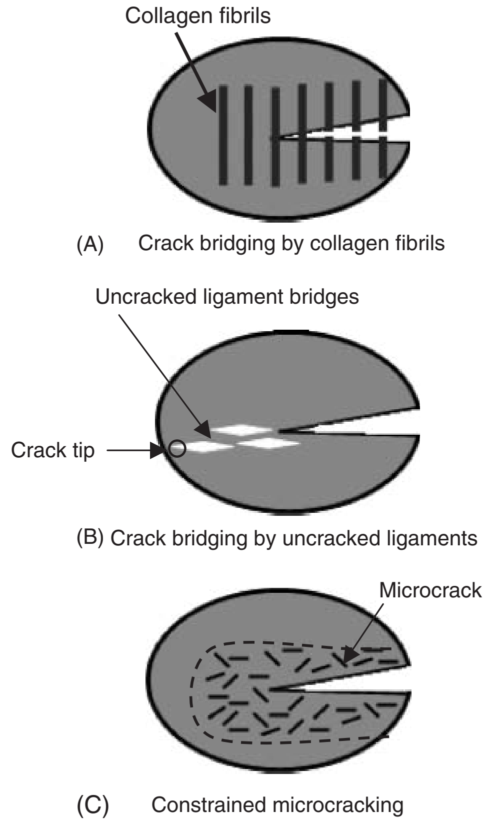


Fig. 6. Schematic diagrams showing the different fracture toughening mechanisms operating in dentine (43).

Biomechanical considerations in intact and post–core restored teeth

The primary responsibility of teeth in the oral cavity is to serve as a mechanical device for the mastication of food. Biomechanics is the study of the mechanics of living structures, especially the energy, forces, and their effects on structures such as teeth. Stress is produced within a structure due to the load acting on it. The direction of the load applied and the shape of the structure influence the nature of stress distribution within the structure. It is established that many detrimental effects produced during restorative procedures are due to the lack of understanding of biomechanical principles underlying the treatment. Examining the nature of stress distribution within intact tooth structure will aid in understanding how natural tooth structures resist mechanical forces in the mouth. Similarly, investigation on the nature of stress distribution in teeth restored using post–core restorations will aid in understanding the biomechanical response of post–core restored teeth to functional forces in the mouth.

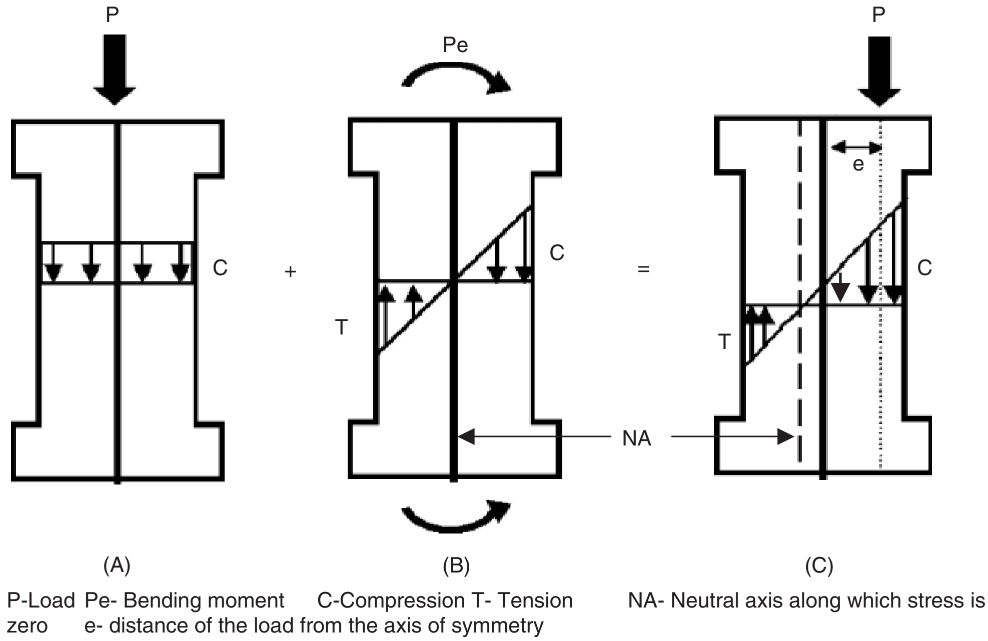
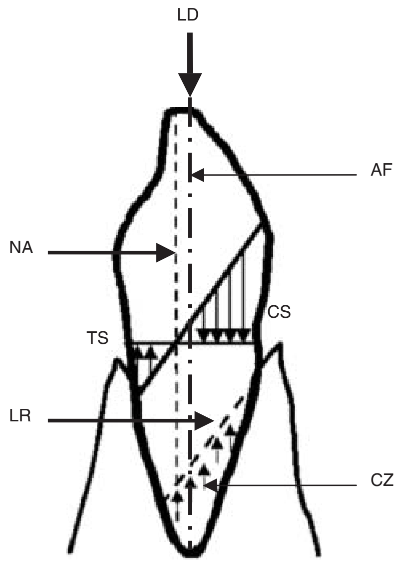


Fig. 7. Schematic diagrams showing the nature of stress distributions in a column. (A) shows the distribution of compressive stress when a load is applied along the axis of symmetry. (B) shows the distribution of bending stress when bending moment is applied. Bending stress resulted in compressive stress on one side and tensile stress on the other with a neutral stress plane intervening. (C) shows the distribution of bending stress when a load is applied at a distance 'e' from the axis of symmetry. This resulted in higher compressive stress when compared with the tensile stress. The neutral axis shifts in this case and no longer coincides with the axis of symmetry (15).

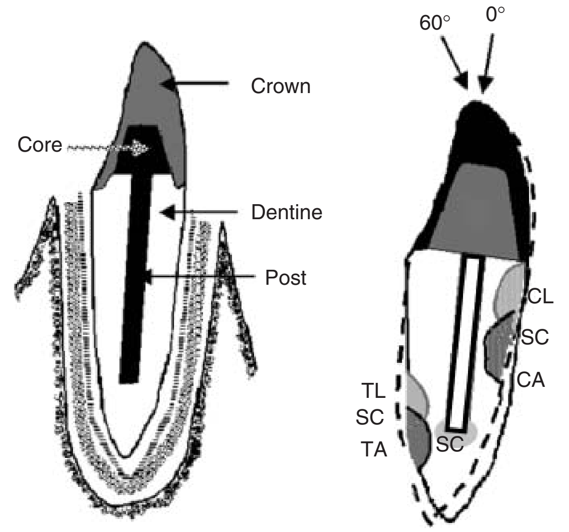
The intact natural tooth is found to experience flexing or bending stress when biting forces act on it. Bending stress distribution in a columnar structure subjected to an eccentric load (i.e. loads acting away from the line of symmetry) is shown in Fig. 7C. The column tends to bend, resulting in compressive stress on one side and tensile stress on the other side. These stresses are the highest at the borders and diminish to zero toward the middle of the cross-section (Fig. 7B). The bending stresses (Fig. 7B), coupled with the compressive axial stress (Fig. 7A), yield a stress distribution as shown in Fig. 7C, displaying higher compressive stresses as compared with tensile stresses. In a tooth under eccentric load, the compressive stresses along one side are substantially higher in comparison with the tensile stresses along the other side in the facio-lingual direction of the root. The apical region of the root showed a notable reduction in bending and manifested particularly compressive stress (15). This increased propensity to compressive stresses in comparison with tensile is due to the shape and angulation of the tooth and supporting bone reactions rather than the eccentricity of loading (Fig. 8) (15).

The stress distribution pattern in a tooth restored using post, core, and crown is distinctly dissimilar to that of an intact tooth. In a post–core restored tooth, the 'post–core-crown-tooth system' bends or flexes as a single unit during mastication. This difference in the 'flexing pattern' of a post–core restored tooth in comparison with a normal intact tooth may be a suggested cause for periodontal bone loss in teeth with metal post (45). The key differences between intact tooth and tooth restored using post–core are (1) occurrence of regions of stress concentration and (2) increase in the tensile stresses produced within the remaining tooth structure of a post–core restored tooth. Figure 9 shows the schematic representation of the regions of stress concentration in a post–core-restored incisor. The stress concentration intensity and tensile stresses have been noted to increase significantly when biting loads are angled away from the long axis of the tooth (16). The factors responsible for this dissimilar stress distribution pattern in a post–core-restored teeth are (1) the greater stiffness of the endodontic post and core restoration, (2) the angulation of the post with respect to the line of action of occlusal load, and (3) increased flexure of the remaining reduced tooth structure. These factors would result in regions of stress concentration and high tensile stresses



LD-Load
TS-Tensile stress
CZ-Compressive zone
CS-Compressive stress
NA-Neutral axis along which stress is zero
AF-Axis of force
LR-Line resulting from the reactant stresses produced by the initial contact of tooth with supporting bone.

Fig. 8. Schematic illustration of bending stress distribution within tooth (15).



CL-Compressive stress concentration regions for axial loads (0°)
CA-Compressive stress concentration for loads at 60° lingual to the long axis of the tooth
TL-Tensile stress concentration for axial loads (0°)
TA-Tensile stress concentration for loads at 60° lingual to the long axis of the tooth
SC-Regions of stress concentrations

Fig. 9. Schematic diagram illustrating stress concentration regions in post–core restored teeth (16).

in the remaining tooth structure. Stress concentrations at the cervical region are mostly because of the increased flexure of the compromised tooth structure, while stress concentrations at the apical region are generally due to taper of the root canal and characteristics of the post. The regions of high stress concentration are also associated with the apical termination of the post (16). Imperfections such as a notch, ledge, or crack created in the dentine during root canal preparation or sharp threading from a post or a pin will also give rise to localized stress concentration regions that can be the locus for a potential fatigue failure (Fig. 10) (47). A smooth root canal shape is therefore recommended to eliminate stress concentration sites.

Fracture from a biomechanical perspective is a very complex process that involves the nucleation and growth of micro and macro cracks. Knowledge of how cracks are formed and grow within materials is important to understand how a structure fractures. Even microscopic cracks can grow over time, eventually resulting in the fracture of the structure. Therefore, structures with cracks that are not superficially visible could fail catastrophically. For fracture to occur in a material, the following factors must be present simultaneously: (1) stress concentrator: this can be a crack, or a geometric notch such as a sharp corner, thread, hole, etc,. (2) tensile stress: the tensile stress must be of a magnitude high enough to provide microscopic plastic deformation at the tip of the stress concentration. The tensile stress need not be an applied stress on the structure, but may be a residual stress inside the structure. It should be understood that material properties such as yield strength and tensile strength have virtually no bearing on the vulnerability of a material to crack extension and fracture. The increase in magnitude of tensile stresses and concentration of stresses will render the remaining tooth structure prone to fracture. Close congruence has been reported between the regions of stress concentration observed in photoelastic models and oblique fracture in extracted teeth subjected to *in vitro* fracture resistance tests (16). Besides, the tensile strength of dentine is much lower than its compressive strength (47). Finite-element analyses (FEA) was also used to study the stress distribution pattern in teeth restored using a post–core system. The FEA studies have highlighted similar altered stress distribution patterns in tooth structure after the placement of post, core, and crown (48–50).

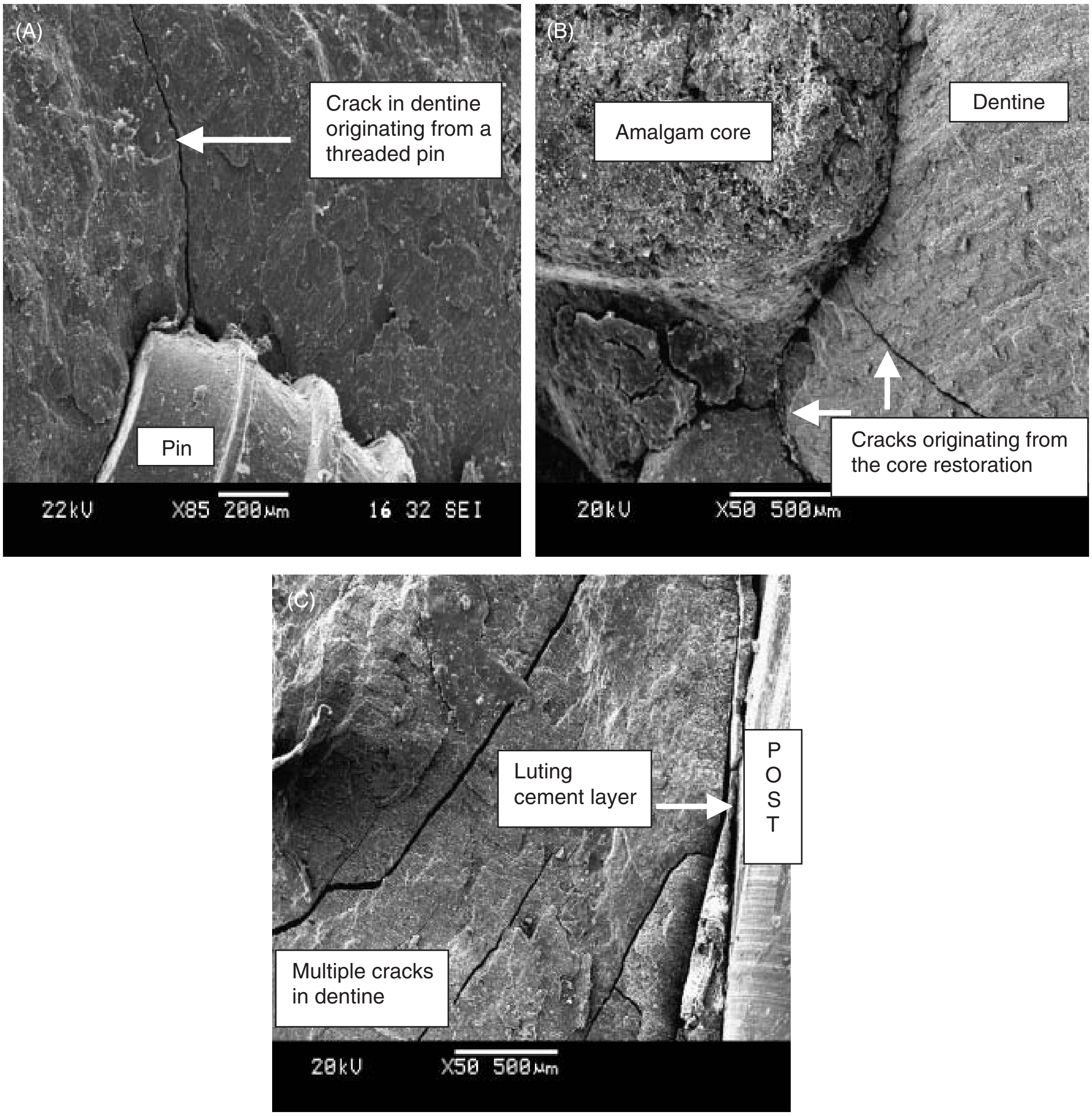


Fig. 10. Scanning electron microscopic images showing cracks emanating from (A) threaded pin (B) amalgam core and (C) endodontic post (with permission from reference (46)).

These investigations have emphasized the fact that tooth structural loss and presence of post–core altered the stress distribution pattern within the remaining tooth structure. Post and core restoration does not allow uniform distribution of stresses within the remaining tooth structure or reinforce the remaining tooth structure (51).

Risk factors for fractures in endodontically treated teeth

For better understanding, the risk factors for fractures in endodontically treated teeth can be classified as primary and secondary causes. Primary causes usually predispose teeth to fracture immediately, while secondary causes predispose the tooth to fracture after a period of time (Fig. 11).

Chemical factors: effects of endodontic irrigants and medicaments on dentine

Sodium hypochlorite is commonly used as a root canal irrigant when performing endodontic treatment. Concentrations of 0.5–5.25% are used for two main purposes: (1) to dissolve pulp tissue and (2) to destroy bacteria (52, 53). In a survey carried out in Australia, the most commonly used concentration was a 1%

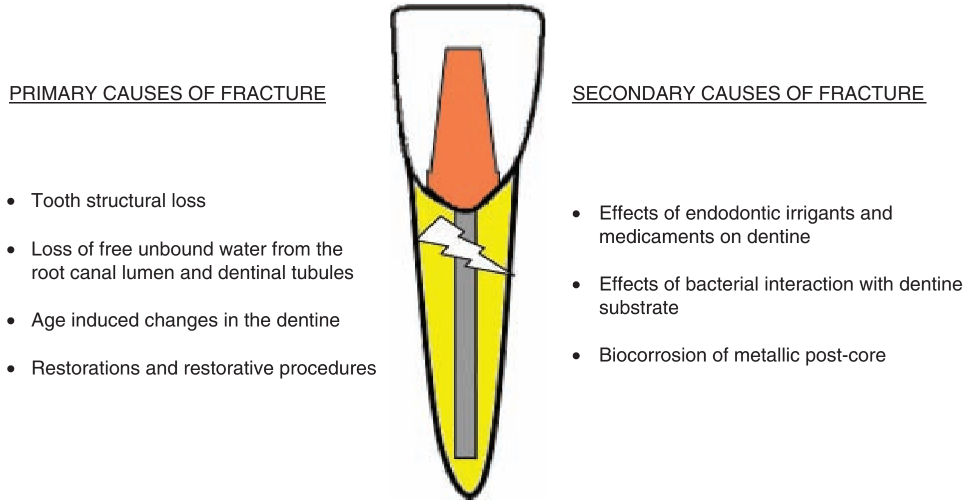


Fig. 11. Causes of fracture in restored endodontically treated teeth.

solution of sodium hypochlorite (54). Recently, there has been a tendency among clinicians to use a higher concentration of sodium hypochlorite in order to achieve a more ‘thorough therapeutic effect’ in a shorter treatment time. Sodium hypochlorite, which is a very reactive chemical, when applied at a high concentration for a long period, along with its desired therapeutic effects, has undesired effects on the root canal dentine. Recently, there have been several reports of the adverse effects of sodium hypochlorite on physical properties such as flexural strength, elastic modulus, and microhardness of the dentine (55–57). These changes in the physical properties of dentine arise due to the changes in the inorganic and the organic phases of the dentine (58, 59). Moreover, in their zeal to ensure complete and rapid disinfection, clinicians vary not only the concentration of the irrigant but also the volume, duration, flow rate as well as temperature. These factors would further aggravate the deleterious effect of sodium hypochlorite on the root canal dentine substrate.

Ethylenediaminetetraacetic acid (EDTA) is also an endodontic irrigant utilized to remove the smear layer formed after root canal preparation. The common concentrations used are 15–17% (60). While smear layer removal continues to be controversial, there can be many benefits in removing this layer of organo-mineral, which is not sufficiently adherent to dentine surfaces to prevent leakage (61). Removing the smear layer not only helps to improve the seal of root fillings, it also removes bacteria, toxins, and remnant pulpal tissues that may be in the smear layer. Five-minute exposure of the root canal to EDTA would remove the smear layer and open the dentinal tubules to a depth of 20–30 μm. In a study by Calt & Serper, (62) dentine was irrigated with 10 mL of 17% EDTA for 1 and 10 min. This was followed by 10 mL of 5% sodium hypochlorite irrigation. While the 1 min EDTA irrigation proved to be effective in removing the smear layer, the 10 min EDTA irrigation group had excessive peritubular and intertubular dentinal erosion. In another study, the Vickers microhardness of root canal dentine irrigated with 5.25% NaOCl, 2.5% NaOCl, 3% H_2O_2, 17% EDTA, and 0.2% chlorhexidine gluconate for 15 min each was studied. Except for chlorhexidine, all irrigants were found to reduce dentine surface hardness. Although most clinicians are aware of the effects of acid on dentine, the effects of EDTA on the dentine surface are less commonly understood. Atomic force microscopy has shown that the intertubular surface dentine etched by phosphoric (3 and 5 mM) and citric acids (5 mM) were smooth, whereas 0.5 M (17%) EDTA treatment produced intertubular dentine surfaces that were significantly rough. The acid-treated groups had an average root mean square roughness of 15 nm, whereas the EDTA-treated groups had an average value of 32 nm (63). It should be noted that the intrafibrillar minerals contribute significantly to the elastic modulus and hardness of dentine (33). These studies confirm that EDTA has profound effects on the dentine substrate.

What happens when EDTA and sodium hypochlorite are used together? 17% EDTA irrigation following sodium hypochlorite irrigation resulted in the opening of dentinal tubular orifices, destruction of intertubular dentine, and reduction of dentine microhardness (57, 64, 65). Many *in vitro* studies have shown that this combination removed the inorganic phase as well as the organic phase of dentine, giving rise to dire effects. It was shown that after the removal of the inorganic phase, applying sodium hypochlorite to it would also remove the organic phase, resulting in a porous dentine surface with multiple channels. This porous dentine surface was observed 40 s after exposure to sodium hypochlorite. It was unique and was not seen after acid etching alone. Furthermore, there was a concomitant loss of mechanical strength by 75% to that of untreated samples (65). A similar drastic effect was also observed in dentine when exposed to EDTA (66). Niu et al. conducted a scanning electron microscopic study of the root canal dentine surface subsequent to final irrigation with EDTA and sodium hypochlorite solutions. They evaluated different irrigation regimes such as initial irrigation with 3 mL of 15% EDTA for 1 or 3 min, followed by 3 mL of 6% sodium hypochlorite for 2 min. Whether 1 min or 3 min irrigation with EDTA was performed, as long as the two irrigants EDTA and sodium hypochlorite were used, there was erosion of dentine (65). Furthermore, mechanical testing of dentine specimens treated with calcium hydroxide, mineral trioxide aggregate, and sodium hypochlorite for 5 weeks demonstrated a 32% mean decrease in strength after calcium hydroxide treatment, a 33% decrease in strength after mineral trioxide aggregate treatment, and a 59% decrease in strength after sodium hypochlorite treatment (67). These experiments highlighted the potential deleterious mechanical effects that can be produced on root canal dentine substrate by the extensive application of concentrated intracanal irrigants and medicaments. The chemically affected radicular dentine can form potential sites for cracking and subsequent fatigue failures. In another investigation, dentine specimens subjected to cyclic fatigue in two different pH environments were tested. Human dentine cantilever-beams were fatigued under load control in pH 6 and pH 7 buffer, with a load ratio (R = minimum load/maximum load) of 0.1 and a frequency of 2 Hz, for stresses between 5.5 and 55 MPa. The material loss was determined at high- and low-stress locations before and after cyclic loading. It was found that the mean material loss in high-stress areas was greater than in low-stress areas, and losses were greater at pH 6 than at pH 7. This study suggests that the residue of low pH chemicals or sealers in the root canal will accelerate mechanical stress-induced erosion of dentin substrate (68). Experiments have also highlighted 3.8% of post-debridement retention of endodontic irrigants in the root canal (69). This means that irrigant-induced effects on the dentine substrate may occur for a longer period of time than what is anticipated.

Microbial factors: effects of bacteria—dentine interaction

Many chemical treatments can modify biological substrates and render them amenable to bacterial adherence or susceptible to degradation (Figs 12 and 13). When EDTA is used on the root canal dentine, it is pragmatic to believe that a layer of collagen and other extracellular matrix proteins will be exposed on the surface of the root canal (70). Collagen is important for the binding of some oral bacteria including *Enterococcus faecalis* (71, 72). Mayrand and Grenier tested *Bacteroides gingivalis*, *B. asaccharolyticus*, *B. melaninogenicus*, *B. loescheii*, *B. intermedius*, *B. endodontalis*, *Actinobacillus actinomycetemcomitans*, *Fusobacterium nucleatum*, *Capnocytophaga ochracea*, *C. gingivalis*, *C. sputigena,* and *Eubacterium saburreum* for collagenolytic activity. Except *Actinobacillus actinomycetemcomitans* and *Eubacterium saburreum*, all tested bacterial strains have been found to be capable of degrading protein substrate. *Bacteroides gingivalis* (*Porphyromonas gingivalis*) is even capable of degrading a protein substrate sterilized using ethylene oxide. Their study

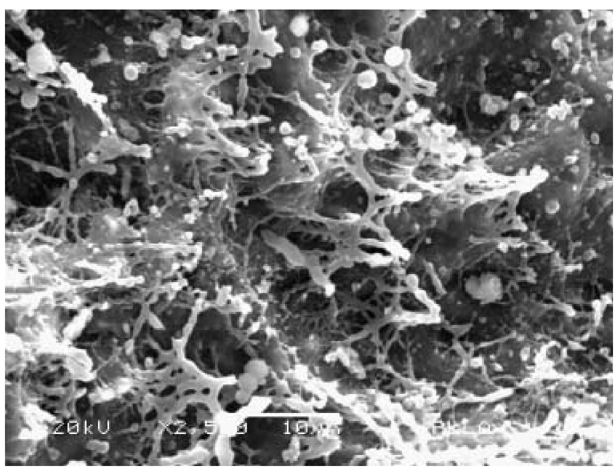


Fig. 12. SEM image from an endodontically treated tooth extracted for periodontal pathology showing bacterial colonies adhering to root canal dentine.

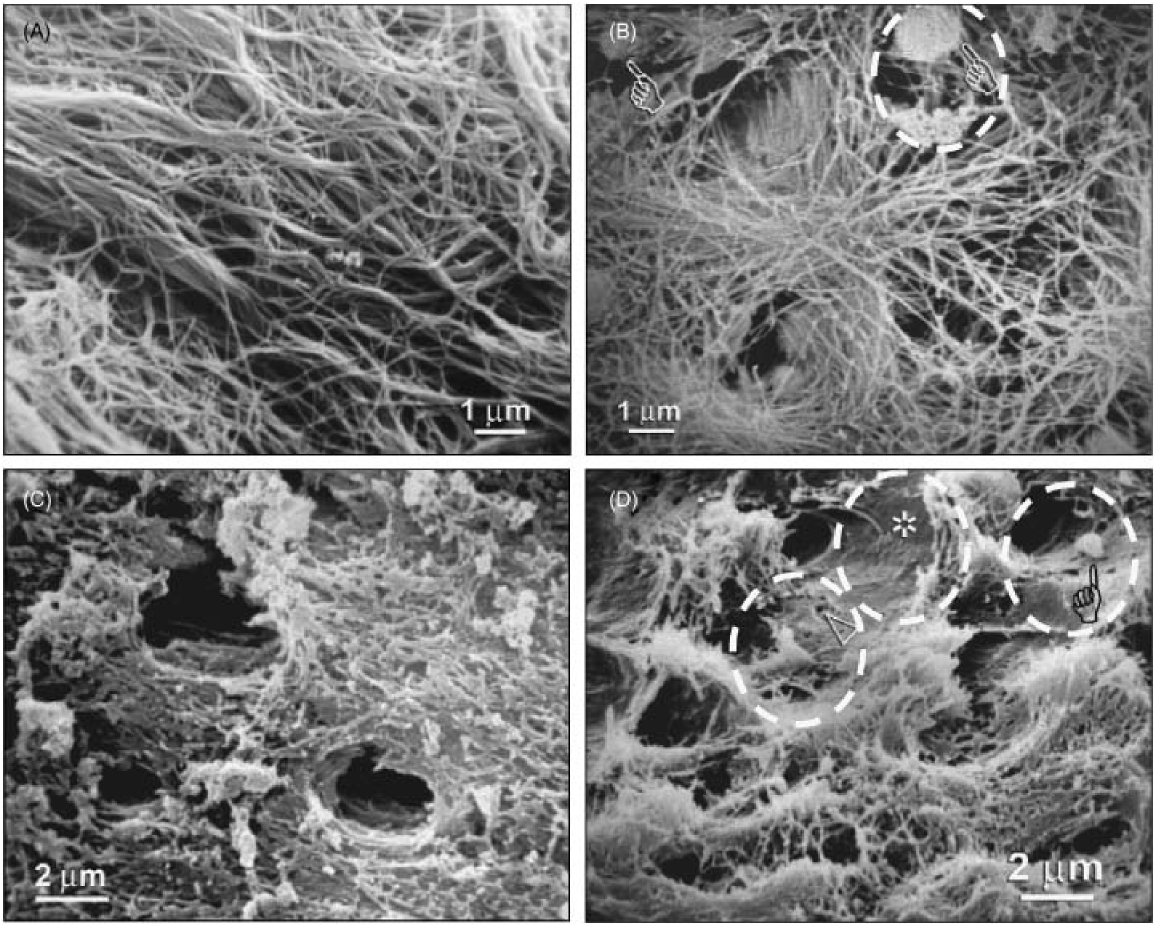


Fig. 13. Representative SEM micrographs of the status of the collagen fibrils that were present along the coronal third of the root canal walls (i.e. post spaces) of endodontically treated teeth. In the control group, the surfaces were treated with zinc phosphate cement for comparison with the other groups that were retrieved after clinical aging and examined without additional demineralization. (A) control group. A three-dimensional network of collagen fibrils that covered the entire root dentin surface could be observed. (B) 3–5-year group. The collagen fibrillar network was sparser in appearance, with the entrapment of globular structures (pointers) within the intertubular collagen matrix and the dentinal tubules. (C) 6–9-year group. The structural integrity of the collagen network was lost. Individual collagen fibrils could not be recognized. (D) 10–12-year group. The demineralized collagen matrix was partially missing from the root canal surface, exposing the underlying unetched mineralized dentin (asterisk) and the lateral branches of the dentinal tubules (open arrowhead). Bacteria-like structures were also present (pointer) (with permission from reference (75)).

indicated that collagenolytic activity can either be the result of the combined activities of both a specific collagenase and non-specific proteases or non-specific proteases only, although in the latter case, the time taken to hydrolyze collagen can be ten times longer than with a specific collagenase (73).

Microbe-induced degradation or modification of collagen can cause deterioration of the mechanical properties such as strength and toughness of dentine, and weakening of the restoration and/or cement dentine interface. Bacteria-induced collagenolytic activity can break chemical bonds at the crack tip and aid in crack propagation through the dentine substrate. Kayaoglu et al. (74) have recently reported that a high prevailing pH in the root canal gives rise to increased bacteria adhesion to collagen. Interestingly, a scanning electron microscopy of extracted post-restored tooth specimens has revealed progressive degradation of the demineralized collagen matrices (DCMs) created on the root canal dentine. The DCMs were observed to become less dense after 3–5 years, losing structural integrity after 6–9 years, and partially disappearing after 10–12 years (Fig. 13). Transmission electron microscopic examination of the same specimens revealed evidence of collagenolytic activity within the DCMs, with loss of cross-banding and unraveling into microfibrils, and gelatinolytic activity that resulted in disintegration of the microfibrils. It is suggested that bacterial colonization and the release of the bacterial enzymes and of host-derived matrix metalloproteinases may contribute to the degradation of the collagen fibrils in the root dentine after clinical function (75). Keeping in view the significant contribution of collagen microstructure to the toughening mechanisms in the dentine, bacteria-induced degradation of the collagen substrate may be a significant potential secondary cause of fracture predilection in endodontically treated teeth.

Dentine factors: effects of tooth structural loss

Sections '*Biomaterial considerations of dentine substrate*' and '*Biomechanical considerations in intact and post–core restored teeth*' and 1.3 describe how major constituents in the dentine, especially the collagen microstructure and water of hydration, contribute to the mechanical integrity of a dentine structure. It is also imperative to realize that the pulp space in an intact tooth with vital pulp is made up of a connective tissue system consisting of cells and fibers both embedded in the extracellular matrix. The extracellular matrix proteins have very high water-holding properties, and the total water content of the pulp is more than 90%. Also, a physiological intra-pulpal pressure of 10–28 mm Hg constantly drives the pulpal fluid outward along the dentinal tubules (22). In endodontically treated teeth, the hydrophilic pulp tissue is extirpated, and the root canal lumen and the dentinal tubules are disinfected and dehydrated before root canal obturation. The loss of water-rich pulp tissues and free water from the dentine surface, porosities, and dentinal tubules can contribute to the reduction in mechanical integrity of endodontically treated teeth (39). In the hydrated dentine bulk, the outer and the inner region showed consistent strains in the lateral direction. This did not vary significantly with an increase in stress within physiological limits. In the dehydrated dentine bulk, the difference in lateral strains between the outer and inner dentine was conspicuous (39, 42). Earlier experiments have also displayed shrinkage and compressive strains in unconstrained dentine bulk subjected to dehydration. Moreover, the residual strain in the outer dentine caused by dehydration increased substantially with applied stress (43). This author has noted that the behavior of dentine observed in this study is similar to a fluid-filled cellular solid (38). Correspondingly, when bulk dentine is compressed, the water-filled dentinal tubules are compressed and the water it contained is squeezed out in the direction of dentinal tubules. Water has a viscosity, and so work is carried out to force out the water through the dentinal tubules. This will lead to dual effects in hydrated dentine: (1) an inherent plasticity effect and (2) distinct strain response in the directions parallel and perpendicular to the dentinal tubules (44). Upon dehydration, which means the loss of water from the dentinal tubules and pulp space, the 'water-induced effects' are lost and hence the bulk dentine displayed increased stiffening and low plasticity (42). Furthermore, it is also shown that the fully hydrated dentine material displayed significantly higher crack-initiation-toughness and crack-growth-toughness than dehydrated dentine (43, 44). The above alterations in the mechanical characteristics of dentine, together with the variation in the biomechanical response, predispose endodontically treated tooth to fracture (16).

Restorative factors: effects of post–core restorations

The most common iatrogenic causes of tooth fracture are (1) due to tooth structural loss during treatment procedures and (2) due to the effects of restorations and restorative procedures on the remaining tooth structure (76). Figure 14 shows the different fracture predisposing factors in a post–core-restored tooth. Previous investigations have suggested that the strength of a tooth is directly related to the amount of remaining tooth structure. Hence, preservation of tooth structure has been recognized to be crucial for the successful management of structurally compromised endodontically treated teeth. Unfortunately, access cavity preparation by itself compromises the mechanical integrity provided by the roof of the pulp chamber and allows greater flexure of the tooth during function (77). Even heavy obturation forces and obturation techniques that generate heavy apical forces (e.g.,: lateral condensation) may also cause tooth fracture (3, 76). Nevertheless, endodontic procedures have been shown to reduce the relative tooth stiffness by only 5%. This was less than that of an occlusal cavity preparation, which reduced the relative stiffness by 20%. The largest losses in stiffness were related to the

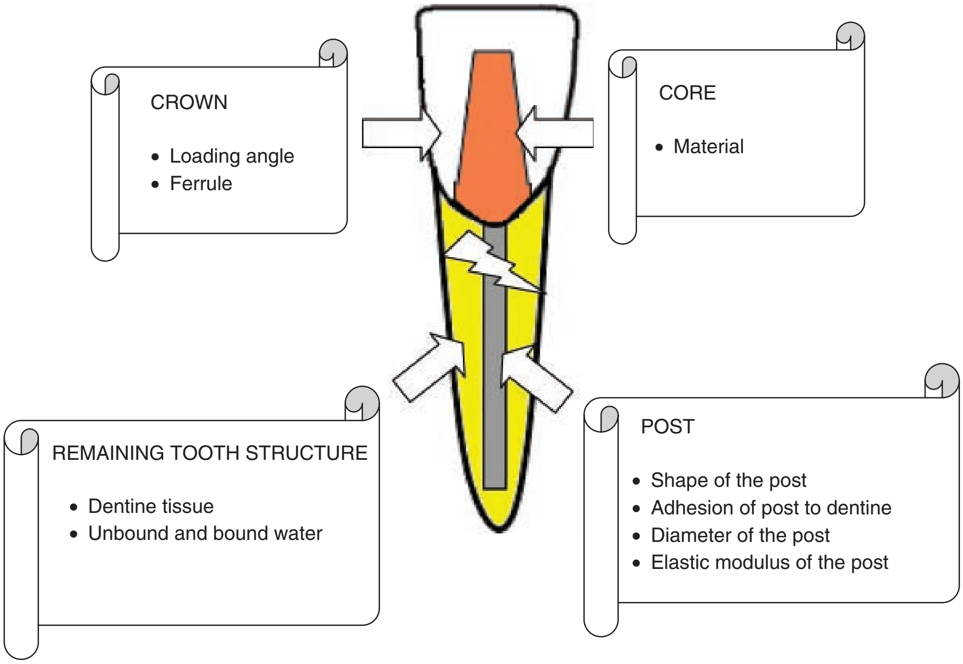


Fig. 14. Fracture predisposing factors in post–core restoration.

loss of marginal ridge integrity, and mesio-occluso-distal (MOD) cavity preparation, which resulted in a 63% loss of relative stiffness (78). Recently, FEA has been applied to investigate the extent to which dentine thickness, radius of root canal curvature, and external root morphology influence tooth fracture susceptibility. This study concluded that tooth fracture is unpredictable, and removal of dentine does not always result in increased fracture susceptibility. It was highlighted that many factors interact in influencing the fracture susceptibility and pattern. However, any one variable can easily predominate over the rest (79).

The most essential component for the reinforcement of the endodontically treated tooth is the ferrule effect. Root factures were common in restored endodontically treated teeth without sufficient ferrule effect (80). The ferrule is an encircling band of metal that embraces the coronal surface of the tooth structure. Teeth prepared with adequate ferrule effectively resist functional forces and enhances the fracture strength of post–core restored endodontically treated teeth. The ferrule length obtained will be influenced by the 'biologic width', which is the dimension of the junctional epithelial and connective tissue attachment to the root above the alveolar crest. It is generally accepted that if unpredictable bone loss and inflammation are to be avoided, the crown margin should be at least 2 mm from the alveolar crest. It has been recommended that at least 3 mm should be left to avoid impingement on the coronal attachment of the periodontal connective tissue (81). Therefore, at least 4.5 mm of supra-alveolar tooth structure may be required to provide an effective ferrule. Several retrospective studies have demonstrated improved long-term prognosis for endodontically treated posterior teeth that received full cuspal coverage restoration (82–85). A detailed review of ferrule effect has been reported by Stankiewicz & Wilson (81). Based on this review, it was concluded that ferrule is desirable, but should not be provided at the expense of the remaining tooth/root structure.

The significance of the amount of remaining dentine thickness (buccal to the post space) in resisting root fracture has been investigated. In this study, 40 extracted maxillary central incisors were used. Four groups were prepared with dowel channels having 1, 2, 3, and 1 mm with a 60° bevel (collar) of remaining buccal dentin. Cast post and cores were cemented into all tested teeth and no crowns were placed. Compressive loads were applied until failure. Based on this experiment, the authors concluded that although significant differences in the failure loads for the four groups were not found, dowel channels with 1 mm of the remaining buccal dentin walls were apparently more prone to fracture under horizontal impact than those that had 2 or 3 mm of dentin thickness. It was also concluded from this study that the addition of a metal collar did not enhance the resistance to root fracture (84). The effect of a cervical metal collar has been re-examined later by Barkhordar et al. (86). This study was based on that of Tjan & Whang (84) but used a modified collar design. It was found that a metal collar significantly increased resistance to root fracture. They also observed different fracture patterns in the collared teeth compared with those without collars. The collared group predominantly underwent patterns of horizontal fracture, whereas the teeth without collars mainly exhibited patterns of vertical fracture (wedging) (86).

Considerable controversy exists regarding the ideal choice of material and design of post and core to improve fracture resistance in post–core-restored teeth. One study recommended endodontic posts of high modulus (86), while another study supported posts having a modulus of elasticity close to that of dentine (87). It is believed that carbon and glass fiber posts have a transverse elastic modulus that is close to that of dentin and therefore are less damaging to the remaining tooth structure (88, 89). Two opposing views of stiffness have been expressed. Some investigators advocate posts with mechanical properties similar to those of dentine. With elastic posts, the tooth, cement, and post will all deform during function. Failure will appear at the weakest point, which would be the adhesive joints at the core–dentine and post–cement–dentine interfaces. Hence, the mode of failure will be loss of marginal seal, core fracture, post fracture, or loss of retention (90). The less the remaining coronal tooth structure, the greater will be the stress at the adhesive interface. *In vitro* studies have shown elastic posts to have a lower tendency to cause root fracture than posts of higher stiffness. Nevertheless, the reinforcement effect after cementation of a complete crown with ferrule effect makes the difference between stiff and elastic posts less obvious. It has been demonstrated that there was no significant difference in fracture resistance between teeth restored with four post and core systems: serrated, parallel-sided, cast posts, and cores; prefabricated, stainless steel, serrated and parallel-sided posts, and resin-composite cores; prefabricated carbon fiber posts and resin-composite cores; and ceramic posts and resin-composite cores. In this study, the teeth from each group received endodontic therapy and a full-coverage metal crown, which was cemented onto each tooth. The specimens were subjected to a compressive load at a 45° angle to its axis until failure (91). However, it should be noted that a stiff post placed in a tooth with minimal coronal dentine would withstand greater biting stresses. Fatigue-induced failure in this system would occur at higher stress levels and after a considerably longer time compared with a low-modulus post, but the risk of an irreparable root fracture would be higher (92–94).

Several *in vitro* studies have determined the resistance to fracture of post–core-restored teeth under static loads and have found lower (93, 94), the same (95, 91), or higher (96) strength for teeth restored with fiber posts compared with teeth restored with metal posts. Some studies indicate that a long post reduces stress (97, 98) and the risk of root fracture (99, 100) (Fig. 15). However, other studies attribute less importance to post length (101, 102). The shape of the post has also been investigated, and several studies have highlighted the dangers of using tapered posts (87–89, 96, 97, 103). Although it was believed that parallel-sided posts can distribute functional loads passively to the remaining tooth structure (98, 104), some studies have observed only minimal advantage with parallel-sided posts when compared with tapered posts (91, 95). The most important factor for preventing fracture, however, is not the post design but the final crown. Studies have confirmed that the difference in fracture rate between various posts disappeared when the crowns were placed (105–107). This was confirmed by a retrospective clinical study that showed that the presence of a post had little effect on the fracture rate of a crowned tooth (99). Recently, an FEA-based investigation was conducted to analyze the stresses in post–core-restored teeth, and to elucidate the effect of material, shape, bonding, modulus of elasticity, diameter, and length of the post on stress distribution. This study concluded that (1) bonded posts and parallel-sided posts resulted in less dentine stress than non-bonded posts and tapered posts. (2) dentine stress was reduced with increasing diameter and modulus of elasticity of a bonded post. (3) a decrease in post length increased dentine stress, but shifted the maximum stress to a location apical to the post. This investigation also highlighted that the shear stresses were smaller than the tensile stresses by a factor of approximately 3–6. The maximum shear stresses were primarily located at the post cement/dentine interface, while the maximum tensile and von Mises stresses were primarily located at

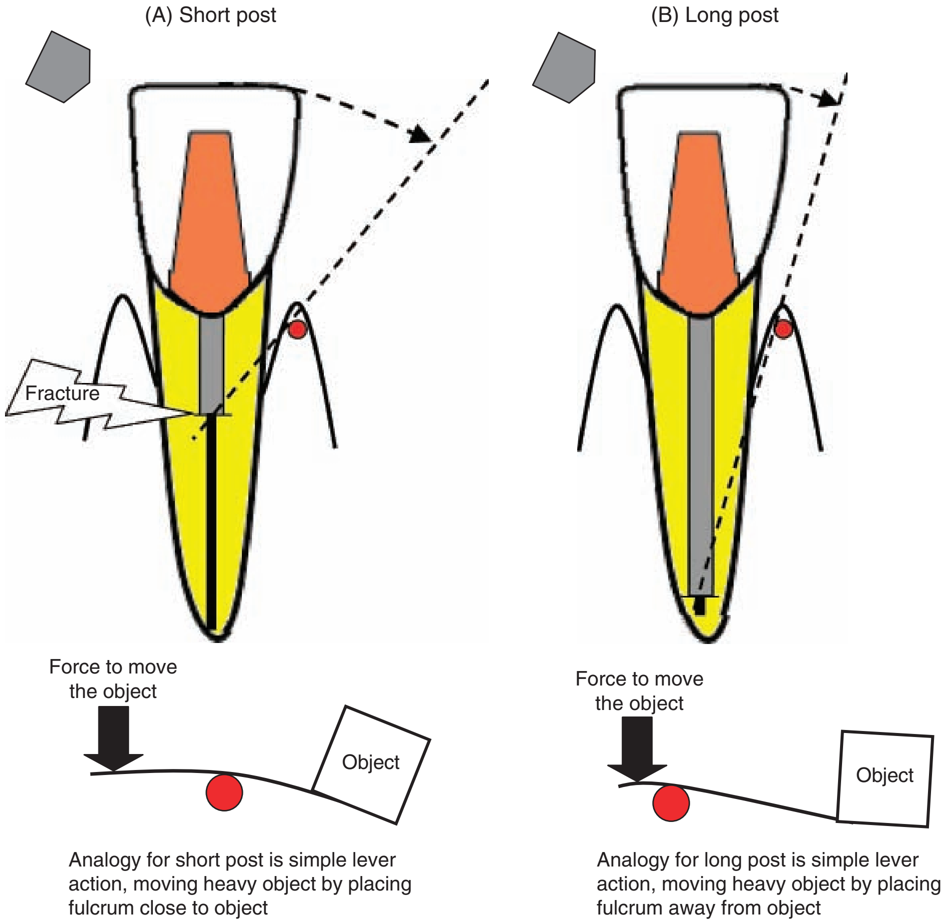


Fig. 15. Showing the lever arm from the occlusal aspect of a tooth to the height of bony attachment in case of (A) short post and (B) long post.

the peripheral portions of the tooth. This tensile and von Mises stresses culminate in increased risk of root fracture, while the shear stresses culminate in increased risk of loss of retention of the post (Fig. 16) (108).

It should be noted that the stress distribution between the post and the remaining tooth structure (post to root dynamics) is not improved by increasing the diameter of the post beyond a point. Having noted that the thickness of the dentinal wall at the root circumference is the most critical factor to resist lateral forces and avoid fracture, it is difficult to recommend any treatment that recommends removal of tooth structure from the canal walls during post-endodontic restoration or positioning of a wider post, as increasing the diameter led to little or no improvement in the post to root retention (109). Another FEA-based study suggested that the core material was the most important parameter in post–core restoration. The effects of changes in the sizes of the post diameters produced only minor differences in the developed stress values, and were not as significant as the core material effects. The loading angle was also found to be an important parameter, which should be taken into consideration in post–core restorations. The increase in loading angle profoundly affected and increased the stress values in the cases considered (110).

Another factor suggested to cause fractures in restored endodontically treated teeth is corrosion. A post and core assembly fabricated using dissimilar metals can cause corrosion by the galvanic effect operating between the two dissimilar metals (111). Non-noble metals such as stainless steel can promote or be subjected to corrosion in non-biologic (112) and biologic environments such as the oral cavity (113). Corrosion mechanisms are complex and are associated with factors such as microbial biofilms, low oxygen tension, and electrical potentials (114) that exist in the oral environment. Corrosion of a metal can induce

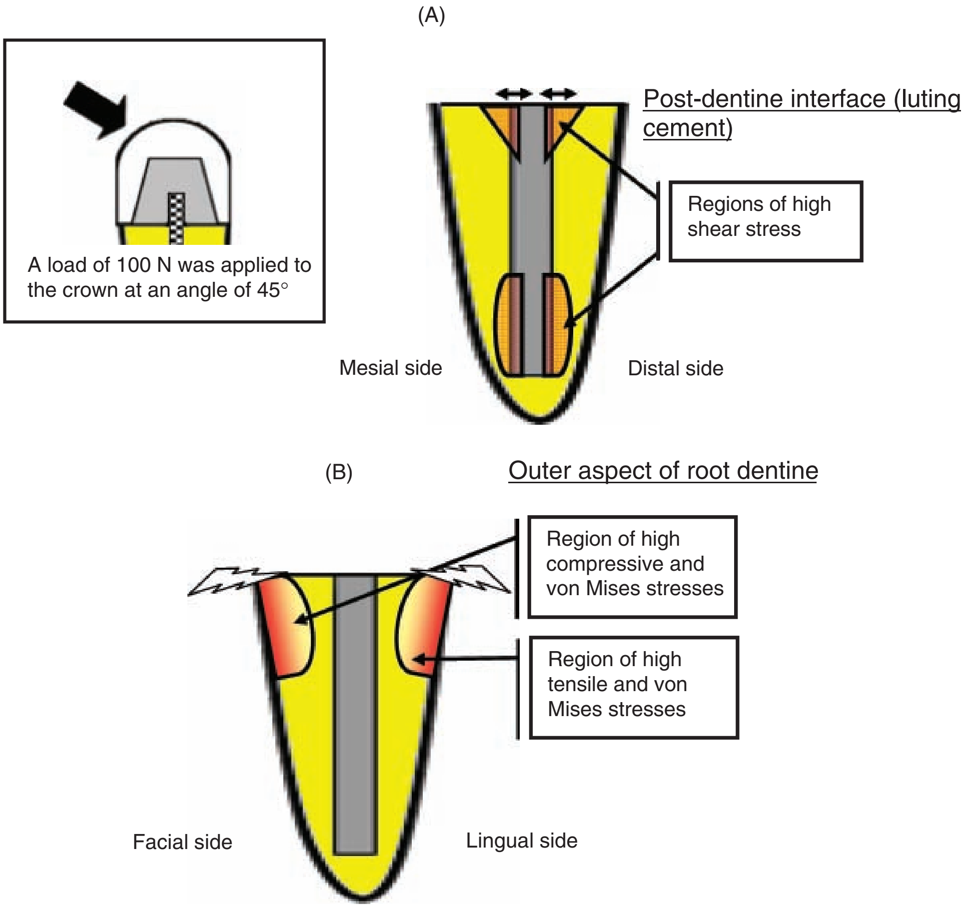


Fig. 16. Schematic diagram obtained from FEM analysis showing the typical distribution of (A) shear and (B) tensile, compressive and von Mises stresses in a dowel–core restored tooth (108).

corrosion expansion stresses (CES) and produce considerable physical damages. These stresses have been implicated as a factor responsible for root fracture. CES in a confined spaces produced by the corrosion of metallic posts can induce a wedging effect and subsequently result in root fracture (115). Therefore, it has been recommended that metals of different electrochemical potential should not be used in a tooth to preclude corrosion (116). Zinc phosphate cement has been selected for cementation of posts for many years. It is still the luting agent of choice for most conventional fixed prosthesis because of its easy handling characteristics and adequate long-term clinical results (117). Zinc phosphate cement adheres by mechanical interlocking to irregularities in the dentine and the prosthesis. Resin-based cements adhere both mechanically and chemically to tooth structure and studies have reported significantly higher retention and resistance to fatigue for resin cements than zinc phosphate cements and resin-modified glass ionomer cements (118). The advantage of using resin cements for post cementation is supported by data reporting the modulus of elasticity of resin based cements as approaching that of dentine. A cement layer elastically compatible with dentine, forming an inner tube bonded to the intraradicular tooth structure, would have the potential to reinforce clinically thin-walled roots (119, 120). However, it has been suggested that the resin-based cements will not miraculously improve the prognosis of a structurally compromised tooth and cannot be universally recommended (121).

There are many prefabricated post systems available in the market today. The disadvantage of prefabricated posts is that the root canal is designed to receive the post rather than the post being designed to fit within the root. Active, threaded posts have the greatest retention; however, due to the threads indenting into the dentine, threaded posts can induce stresses in the root dentine. This could lead to crack initiation and might induce root fracture at a later time. Root morphology is also an important parameter in determining the optimum post length without endangering the structural integrity of the tooth. The root should have at least 1 mm of tooth structure remaining around the post in all directions to resist fracture or perforation (122). Adequate knowledge of the root morphology for each tooth is therefore mandatory for dowel placement. Maxillary first molars have deep concavities on the furcal surface of 94% of mesio-buccal roots, 31% of disto-buccal roots, and 17% of palatal roots. Mandibular first molars have root concavities on the furcal surface of all mesial roots and 99% of distal roots (123). Maxillary first premolars have deep mesial concavities and slender roots with thin dentine (77). These anatomical restrictions should be kept in mind when restoring such teeth.

Anatomical location influences the fracture predilection of endodontically treated teeth. Intact pulpless anterior teeth that have not lost further tooth structure beyond the endodontic access preparation are at minimal risk for fracture. However, posterior teeth bear greater occlusal loads than the anterior teeth during mastication, and adequate restorations must be planned to protect these teeth against fracture. It has been suggested that posts should be avoided in posterior teeth as the roots are often narrow and/or curved. Subsequently, post-space preparation can lead to root perforation. Removal of radicular dentine to accommodate post will further weaken the tooth and may lead to fracture. Posts are not usually required for the retention of the core in posterior teeth as there is normally sufficient coronal tooth structure present and mechanical undercuts from the pulp chamber (124). Previous biomechanical studies have focused mostly on post–core restoration in single-rooted teeth. It is significant to realize that the biomechanical response of posterior teeth is different from that of anterior teeth. Therefore, factors such as ideal post length, post diameter, etc., observed in an anterior tooth, may not be accurate for a posterior tooth.

Failure of the crown–tooth interface at the crown margin has been realized to be the earliest sign of failure in post–core-restored tooth (92). With time, this interfacial failure at the crown margin may penetrate into the tooth, leading to core fracture or the failure of the core–crown interface and/or core–tooth interface. Once this progressing interfacial failure reaches the post, it may cause the fracture of the post. Alternatively, it may also cause the failure of post–tooth interface. With extensive interfacial failure, there will be rocking of the post–core component within the remaining tooth structure. Subsequently, the post–core–crown–tooth structure ceases to function as a single functional unit. This mismatch in the biomechanical response further predisposes tooth to fracture. Therefore, interfacial properties at the crown–tooth, crown–core, core–tooth, core–post, and post–tooth interfaces and bulk mechanical properties of the crown, core, and post materials will determine the nature of fracture in post–core-restored tooth. The anatomy of the tooth, shape and stiffness of the post, direction and magnitude of external forces, and the amount of remaining tooth structure will also influence the nature of the final tooth fracture (Fig. 17) (93, 125).

Abutment teeth for long-span, fixed bridges and distal extension, removable partial dentures (RPDs) receive greater transverse loads during function. The horizontal and torquing forces endured by endodontically treated abutments of fixed or RPDs predispose them to fracture (126–130). Nyman & Lindhe reported that fractures in abutment teeth occurred more frequently in endodontically treated teeth fitted with posts, serving as terminal abutments for free end segments (127). Testori et al. recommended against post–core-restored endodontically treated teeth as abutments for cantilever bridges. In addition, it has been suggested that attaching RPDs with extracoronal attachments to endodontically treated teeth places these teeth at significant risk. It has been found that abutment teeth for distal extension RPDs are more likely to experience high bending stresses in a mesio-distal direction than either tooth-borne or non-cantilevered fixed partial denture (FPD) abutments. The stresses generated in the tooth structure by such loads are found to be greater than those of comparable bucco-lingual loading, as most of these abutment teeth are much narrower mesio-distally than bucco-lingually (129). Figure 18 shows the different types of forces acting in restored endodontically treated teeth subjected to cantilever loading (128). Therefore, based on the available data, the use of an endodontically treated tooth to support a precision attachment RPD, a distal extension RPD, or a cantilevered FPD cannot be considered to be highly predictable. Furthermore, endodontically treated teeth may not be suitable as abutments in individuals with a history of bruxism, or those requiring long-span fixed bridges (129, 130).

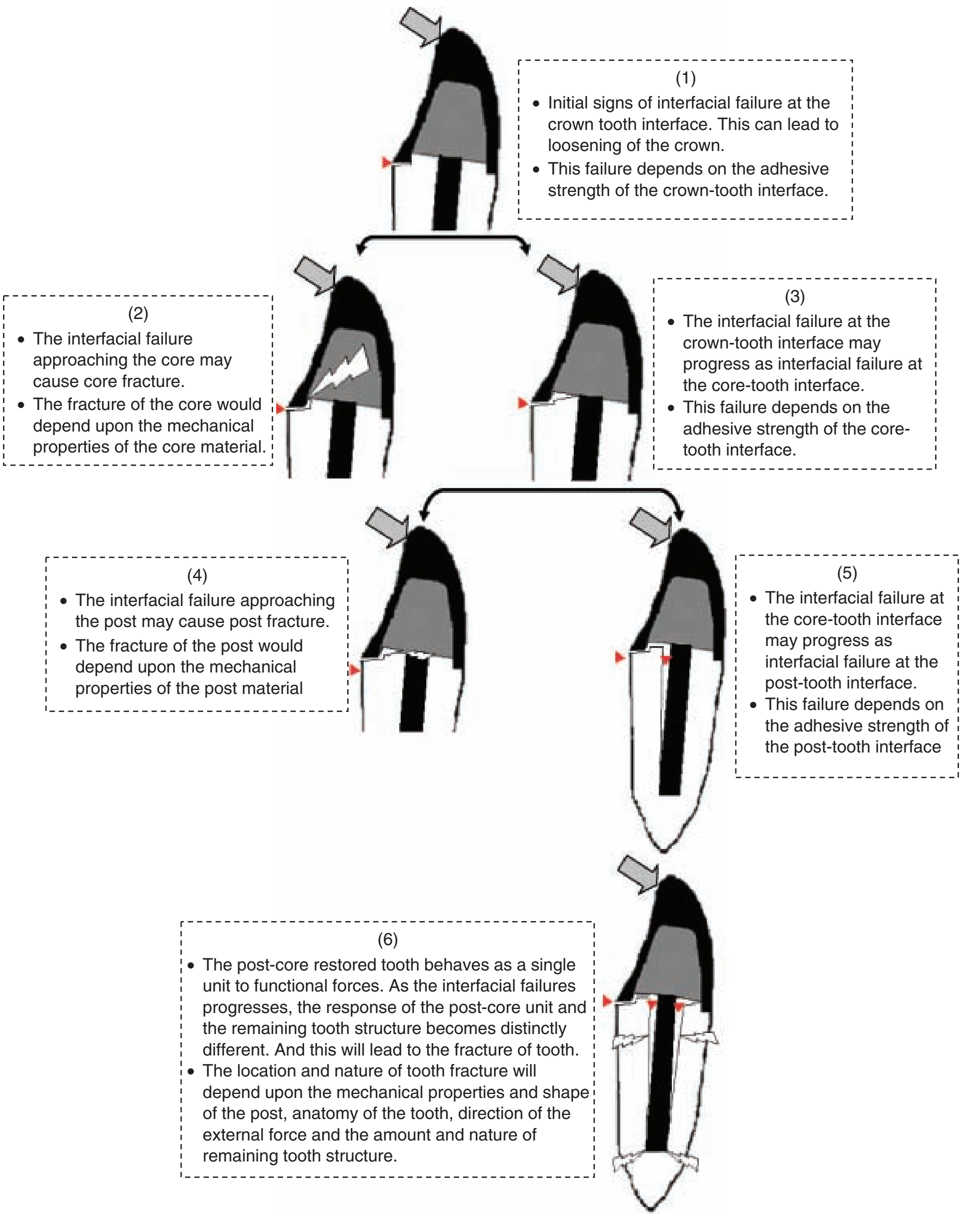


Fig. 17. Schematic diagram showing the progression of interfacial failures and fractures in post–core restored tooth.

Age factors: effects of age changes on dentine

Alteration of normal dentine to form transparent dentine is a common age-induced process. *Physiologic transparent (or sclerotic) dentine* appears to form without trauma or caries attack as a natural consequence of aging, whereas *pathologic transparent dentine* is often seen subjacent to caries. The dentinal tubules in transparent dentine are gradually filled up with a mineral phase over time, beginning at the apical end of the root and often extending into the coronal dentine. The large intratubular mineral crystals deposited within the tubules in transparent dentine are chemically similar to intertubular mineral. It was suggested that a 'dissolution–reprecipitation' mechanism is responsible for its formation (131). In the past, it was believed that transparency required a vital pulp (132). This belief has been largely discounted. It now appears that endodontically restored teeth have the same or a greater rate of transparent dentine formation as teeth with vital pulp (133). The elastic properties of

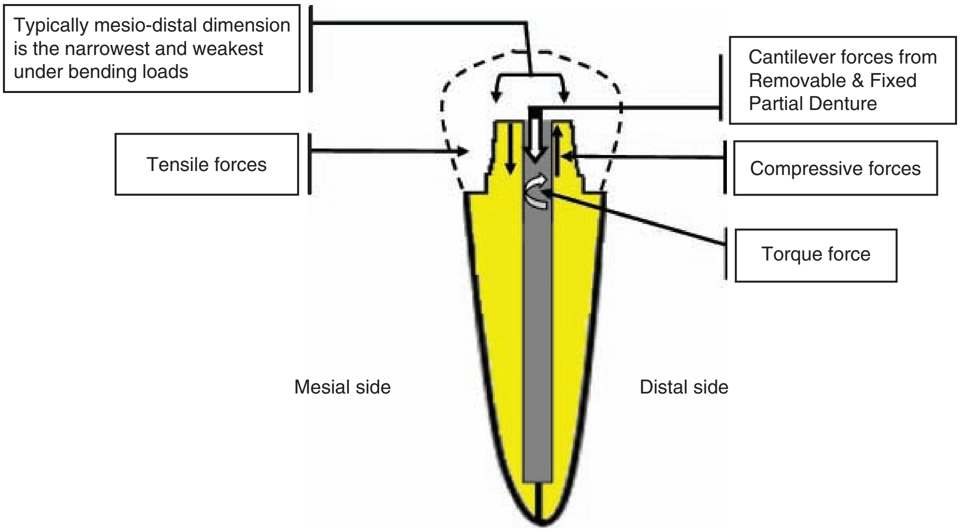


Fig. 18. Effect of bending forces from cantilever loading of endodontically treated tooth (128).

transparent dentine were not significantly different from normal dentine. However, transparent dentine, unlike normal dentine, exhibits almost no yielding (plastic strain) before failure. The fracture toughness in transparent dentine is approximately 20% lower and their stress–strain response is characteristic of brittle behavior (134). It has also been reported that the tensile fatigue strength of aged dentine is lower when compared with young dentine (135). Hence, restorative procedures in aged individuals might require modification to accommodate the reduced fracture toughness of dentine tissue.

What are the causes of reduced fracture toughness in transparent dentine? One cause is the presence of reduced water content compared with normal dentine. Another proposed theory is that mineral accretion within the tubules lowers the tendency for microcrack nucleation there. In the absence of significant microcrack nucleation ahead of the main crack tip, uncracked ligaments are less likely to form, which in turn lowers the fracture toughness of dentine (Fig. 19) (134). Experiments conducted to study crack propagation in different dentine show that fatigue crack growth in old and dehydrated dentine occurred more rapidly than in young and hydrated dentin. For equivalent crack driving forces, the fatigue crack growth in old dentin occurred at a rate over 100 times that in young hydrated dentin. Dehydration resulted in an increase in the growth rate of nearly 100 times as well. SEM examination of fracture surface showed a rough surface topography in young hydrated dentine. There was also

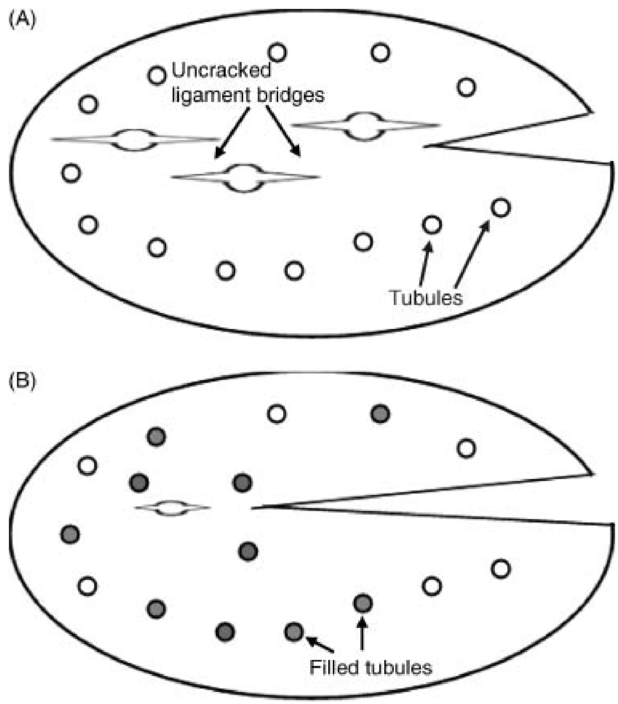


Fig. 19. Schematic illustration of the difference in proposed fracture mechanisms in (A) normal dentine and (B) transparent dentine. In normal dentine, microcrack formation at peritubular cuffs ahead of the crack tip leads to the formation of uncracked ligaments due to imperfect link-ups between the microcracks and the main crack tip. With transparent dentine, mineral accretion seals up some of the tubules, leading to fewer stress concentration points ahead of the crack tip and hence, fewer (if any) uncracked ligament bridges are formed. The absence of such bridges which normally increase the toughness by sustaining part of the applied loads that would otherwise be used for crack extension, result in diminished fracture toughness of transparent dentine (134).

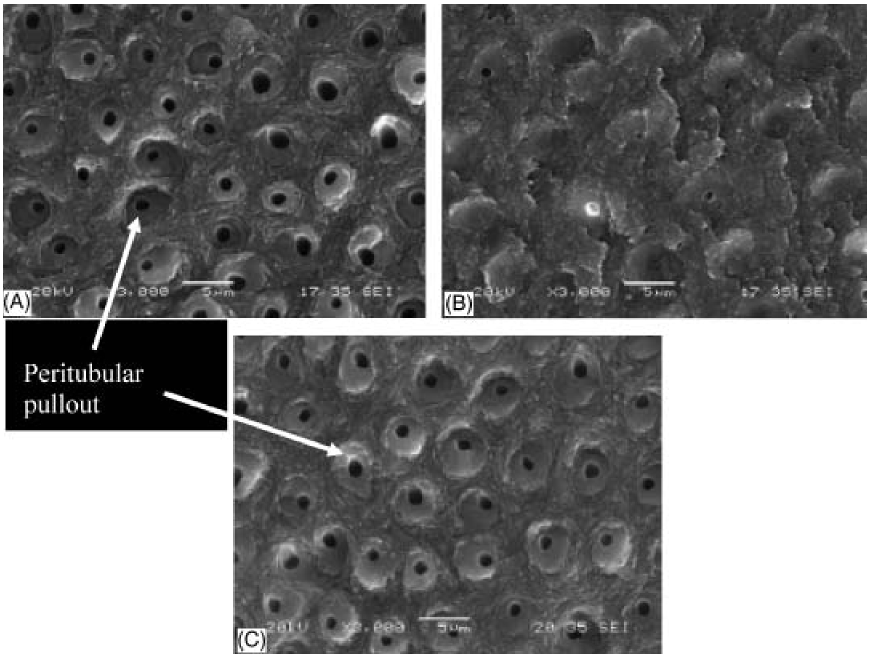


Fig. 20. Typical micrographs from the fatigue fracture surface of young and old dentin CT specimens. The direction of crack growth is from top to bottom in all three micrographs: (A) the fatigue fracture surface from a young hydrated dentin specimen. This dentin is from a female patient 20 years of age. Note the difference in fracture plane of many of the tubules in relation to the intertubular dentin; (B) the fatigue fracture surface from an old hydrated dentin specimen. This dentin is from a male patient 50 years of age. Most of the tubules are occluded and there is very little evidence of peritubular pullout; (C) the fatigue fracture surface from a young dehydrated dentin specimen. This dentin is from a male patient 19 years of age (with permission from reference (136)).

characteristic evidence of peritubular dentine cuff pullout in the young hydrated dentine. Fracture surfaces of the old dentine did not show evidence of peritubular cuff pullout was rather uncommon. The fracture surface of the old dentine appeared smooth (Fig. 20). It is suggested that the peritubular dentine cuffs contribute to energy dissipation during crack extension in young hydrated dentine. Also, the increase in mineral content of dentine with aging, or redistribution of mineral with dissolution and precipitation may result in an increase in cohesion between the intertubular and peritubular dentine constituents, thereby decreasing the relative interfacial sliding with crack growth (136). Age-related modifications of collagen such as cross-linking have also been suggested to cause deterioration of the mechanical properties such as strength in mineralized tissues (137).

Concluding remarks

Dentine is primarily a collagen-rich organic matrix reinforced by calcium phosphate mineral particles. The constituents of dentine material are efficiently optimized to different mechanical demands in the mouth. Gradients in the elastic modulus of dentine have been attributed to the distribution of the mineral phase, while different toughening mechanisms in the dentine have been attributed to collagen microstructure and water content. In order to replace the mechanical function of dentine from a restorative perspective, it is not only important to study its localized tissue properties but also its bulk structural behavior. Nonetheless, more research is necessary to comprehend the mechanisms by which tooth structures resist functional forces in the mouth.

Often, endodontically treated teeth experience tissue loss due to prior pathology or treatment procedures. The loss of dentine tissue will compromise the mechanical integrity of the remaining tooth structure. In a post–core-restored tooth, the post, core, crown, and the remaining tooth structure respond to biting forces as a single functional unit. Yet, there is an alteration in the stress distribution pattern within the remaining tooth structure. Regions of stress concentration and high tensile stresses readily occur in post–

core-restored tooth. The intensity of stress concentration and tensile stresses will depend upon (1) the material properties of the crown, post, and core material, (2) the shape of the post, (3) the adhesive strength at the crown–tooth, core–tooth, and core–post, post–tooth interfaces, (4) the magnitude and direction of occlusal loads, (5) the amount of available tooth structure, and (6) the anatomy of the tooth. The stress concentrations and high tensile stresses will predispose such teeth to fracture.

In the past, many conclusions on the biomechanical behavior of post–core restored tooth have been based on static experiments. Although these findings can be helpful as a first step, additional experiments are indicated under clinically realistic and dynamic loading conditions. Development of adhesive-based integrated restorations that minimize stress concentrations and tensile stresses in the remaining tooth structure is recommended. The role of aging and long-term effects of root canal reagents and surviving microbes on interfacial failures and fractures of restored endodontically treated require further investigations. All these studies will not only enhance our understanding of the factors that predispose endodontically treated teeth to fracture but also aid in better restorative management of broken-down teeth, endodontic retreatment cases, and geriatric patients.

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