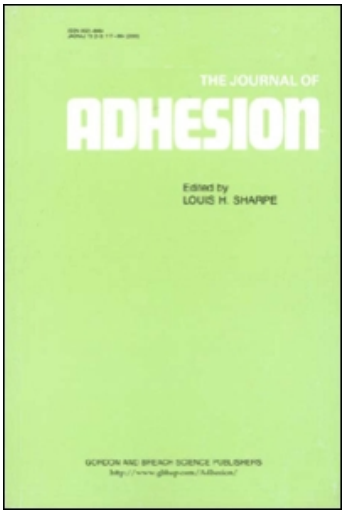

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The Artificial Tooth-Denture Base Joint

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The Artificial Tooth-Denture Base Joint

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Fracture mechanics methods have recently been applied to predict strength of adhesive systems. In this paper, the tooth–denture base bond strength is used to demonstrate the application of these techniques. Because of the complex geometric and material parameters involved, a finite element computer program was required to obtain the strain energy release rate values as the tooth debonds from the denture base. Adhesive strength parameters are shown to be a function of load application rate. Cinematography was used to determine debond propagation rate and demonstrate that debond rate is a function of load rate.

INTRODUCTION

In many technological applications one component of a system is bonded or cast to another. This introduces the possibility of interfacial or adhesive failure under service conditions. Costly and sometimes dangerous systems failures could be prevented by having experimental and analytical techniques for evaluating interfacial strength and for predicting the fracture resistance of practical joints.

This paper describes how fracture mechanics, commonly used to analyze and/or predict cohesive fracture, can also be used on adhesive systems. The artificial tooth–denture base unit in dentures was used as a practical system for the application of fracture mechanics and related analyses.[1]

Dentures are a very important category of biomedical devices. More than 35,000,000 are in use in the U.S.[2, 3] They are constructed predominantly from

pre-manufactured artificial teeth and denture base polymers (Figure 1).[3] Depending on the type of artificial teeth used, the tooth–denture base joints may range from purely mechanical fixtures to diffusion bonded compositional gradient domains (Figure 2).[4] Under dynamic conditions, such as chewing,

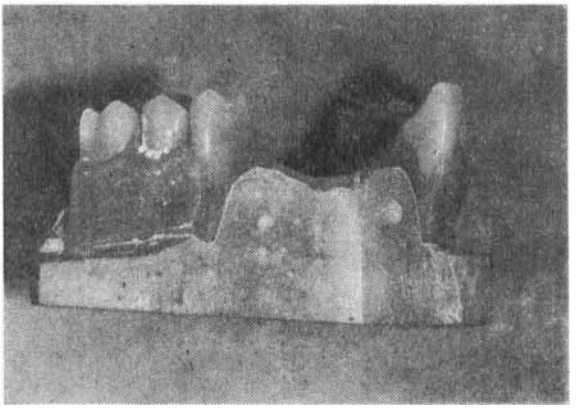

FIGURE 1 Cross section of denture with acrylic teeth and denture base. Denture is supported on a cut away plaster model of the patient's mouth.

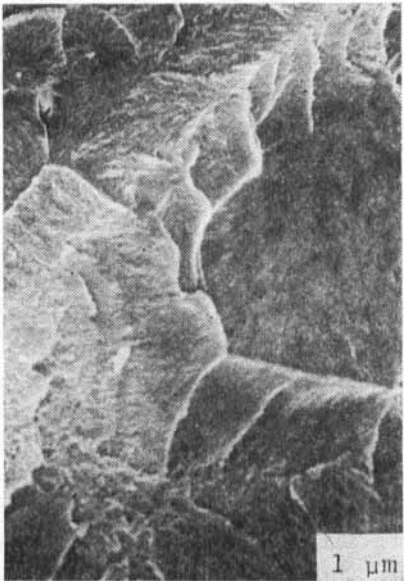

FIGURE 2 Cross section of acrylic tooth–denture base assembly. Bond line runs diagonally with tooth at upper left and denture resin at lower right.

these interfacial regions constitute the load transfer gradient domains which control the functionality and durability of dentures.[5]

BACKGROUND

Artificial acrylic teeth and acrylic denture base polymers were selected from the list of dental materials certified by the American Dental Association. Representative techniques[2] were used to fabricate dentures for studies on the mechanical functionality of these multicomponent systems. Cinematography (64 frames per second) was coupled with scanning electron microscopy to observe the effect of microstructure on the stress–strain behavior of the acrylic teeth bonded to the denture base polymer.

The micromechanical behavior of dental polymers was observed by polarized light microscopy. Polymerization studies under the microscope indicated the influence of chemical composition on the internal morphology of multiphase acrylic polymers. Interfacial strains and failure phenomena were observed by phase contrast microscopy. These techniques were also used to investigate the contribution of diffusion and sorption processes on the performance of adhesive bonds in dental prostheses.

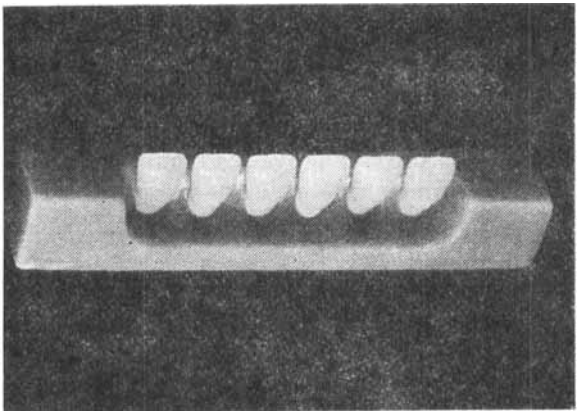

FIGURE 3 Specimen for quasi-static anchorage test.

There are two loading conditions of primary interest for dentures; normal use conditions and impact. The normal chewing rate is 70 to 80 cycles per minute which is applied in a nearly sinusoidal manner. Thus, as the teeth are nearing contact, the velocity is very low. This condition was approximated in the laboratory using cantilever flexural test specimens submerged in water at 37°C and loaded at a displacement velocity of 8.3×10^{-6} m/s[4, 6] (Figure 3).

Dynamic impact studies were completed using the dynamic impact specimens shown in Figure 4. These were loaded to failure at a deflection rate of 3.5 m/s to simulate failure events of dental polymers under conditions of accidental impact.

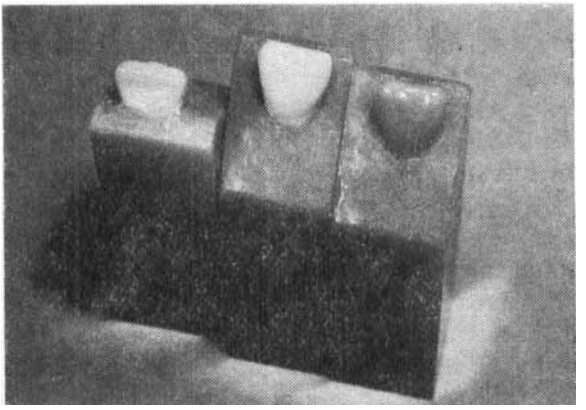

FIGURE 4 Specimens for dynamic impact test.

Dynamic tests have previously been used by Plati and Williams[7] to determine critical energy release rates for polymers in bulk. Similar fracture mechanics concepts were extended to the present consideration of polymers in the form of tooth–denture resins. Because of the complex geometry of the tooth, the finite element method, presently the most popular numerical stress analysis technique, was used to provide accurate values of strain energy release rate. With this technique, bodies with irregularly shaped boundaries, boundary conditions varying over the geometry, and nonhomogeneous materials can be easily analyzed.

Much effort has been expended in developing such techniques for numerical determination of stress intensity factors or energy release rates associated with fracture geometries.[8, 9] An excellent summary and description of available computer codes are given in Ref. 9. Most methods involve the direct use of finite element computer programs with variations in the technique of applying such programs. These techniques include:

1) Matching stress or displacement components near the crack tip with the corresponding singular terms from the theoretical equations as discussed in Refs. 1, 10 and 11.

2) Using the local energy release rate (contour integral) as proposed by Rice.[12]

3) Calculating the change in elastic potential as proposed by Buekner.[13]

4) Introducing special elements[14] at the crack tip which include one or several singular terms.

5) Global energy release rate (calculating the strain energy stored in the body for two different values of crack length and then calculating the energy release rate using a central difference technique).[1, 15]

As has been shown by several authors, convergence of stresses and strains produced by normal finite element codes in the vicinity of boundary corners, or material discontinuities with mesh refinement, is not guaranteed. Discussions of this problem may be found in Birkhoff,[16] Babuska,[17], Fix,[18] Fix *et al.*,[19] and Kellogg.[20] On the other hand, the accuracy of numerically calculated energy release rates, stresses, and displacements has been evaluated by comparing the numerically calculated values with analytical results for several different geometries and loadings as reported in Refs. 1, 11 and 21 for homogeneous materials and in Ref. 1 for adhesive materials. The conclusions in these reports are generally that even for relatively coarse grid networks, any of the above methods can be used with standard finite element computer programs (such as constant strain element) to produce satisfactory values for the stress intensity factor. With reasonable care in selecting a grid network, the accuracy of these numerical techniques is normally greater than 95%. With a properly selected grid, the global energy release rate method has produced three digit accuracy for many test cases.[1, 10, 21] It is, therefore, concluded that finite element programs can be used to obtain accurate values of the elastic stress intensity factor or, equivalently, the energy release rate for at least a large family of cohesive and adhesive fracture geometries in spite of the well known mathematical singularities.[22]

In the present denture study, only the central cross section of the tooth was modeled (Figure 5). Plane strain assumptions were used with the resulting

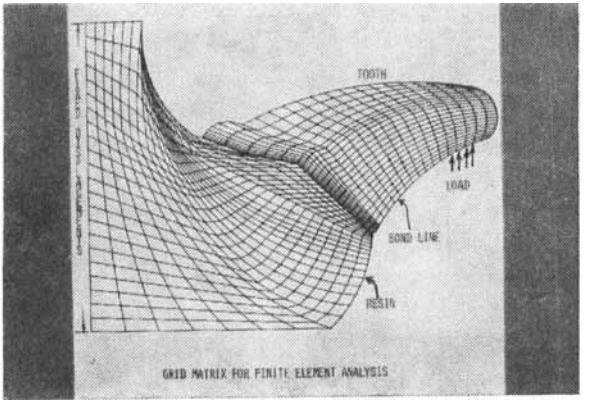

FIGURE 5 Grid matrix for finite element stress analysis.

values of strain energy modified to account for tooth thickness at each crack depth. A total of seven different crack depths were analyzed. At each crack depth, the value of strain energy was determined. Strain energy release rates were then calculated using a central difference technique.[1]

The displacements for each node in the left hand portion of the grid were fixed to simulate clamping by the test fixture jaws. A 25 kg force was applied to the tooth in a direction parallel to the resin base. The elastic moduli for tooth and resin materials were 3300 MPa and 2900 MPa respectively. A Poisson's ratio of 0.35 was used for both materials.

The difference in strain energy for two successive crack depths was divided by the new surface area generated by the increment of crack depth to calculate the reference strain energy release rate, G (Figure 6). For any given crack

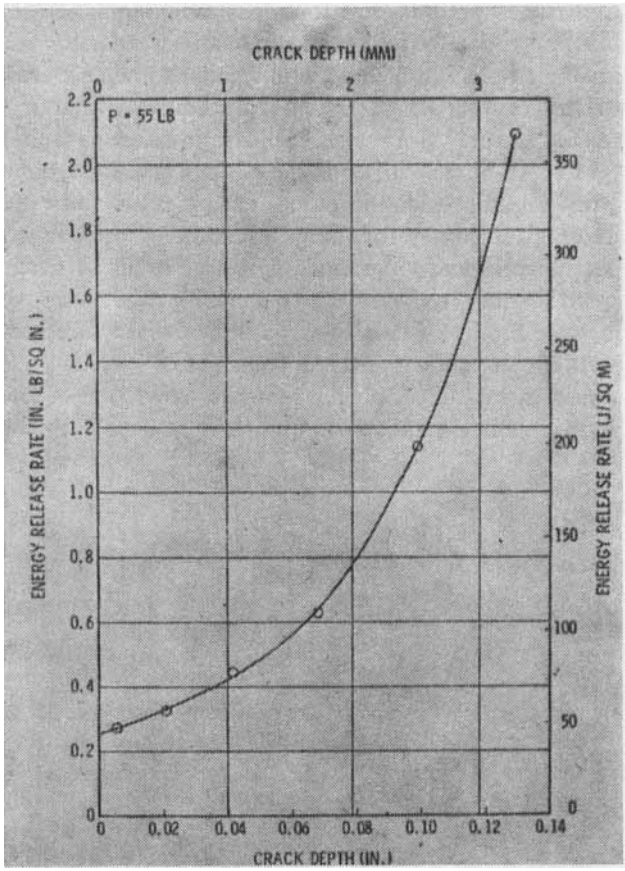

FIGURE 6 Results of finite element stress analysis.

depth, a, the value of strain energy release rate required for crack propagation (G_c) is determined from the load required to cause debond propagation, P_{cr}, as:

$$G_c = \left(\frac{P_{cr}}{P_0}\right)^2 G_0, \tag{1}$$

where G_0 is the reference value taken from Figure 6, and P_0 is the load used in the reference energy release rate curve (25 kg).

MATERIALS

The teeth selected for this study were acrylic maxillary central incisors (Trubyte Biotone 3N, Dentsply International). These were randomly selected from production lots prior to handling and packaging to obtain controlled surfaces free from carding wax and denture processing materials. The teeth were cleaned in a laboratory ultrasonic unit, rinsed, and dried in a dust free desiccator at 23°C.

The gross structure of the teeth simulated natural tooth structure. A cross-linked "enamel" layer was laminated during their manufacture to a tougher, higher molecular weight "body". The gel molding process employed a swollen gel of high molecular weight suspension polymerized methacrylate polymer "beads" with methacrylate monomers. The vestigal bead structure is observable in Figure 2.

TruPour Denture Resin (Dentsply International) was used as a denture base mounting for the teeth in static and dynamic impact test specimens. The well defined composite structure of this denture resin is apparent in Figure 2. This resulted from the dispersion of medium and high molecular weight methacrylate and polymer beads in a cross-linking methacrylate monomer matrix.

TEST SPECIMEN BONDING

Dentures and tooth–denture base bonded specimens were constructed using a denture base casting technique.[4] The acrylic teeth were mounted in a wax model of the gums (Figure 1). The teeth and wax denture base model were encased in a rigid dental plaster mold. After the mold hardened, the wax was melted, leaving the artificial teeth retained in position in the mold. The bases of the teeth were cleaned with detergent and boiling water. After rinsing and drying the mold and teeth, the denture base dispersion was poured into the cavity left by the wax mold. After a 30 minute, 60°C, 20 psi polymerization

cycle, the completed tooth–denture base units were separated from the plaster mold. Using this method, anchorage (Figure 3) and impact (Figure 4) test specimens were constructed to simulate the typical acrylic denture shown in cross-section in Figure 1. In dentures, the base material overlaps the tooth to provide a mechanical lock. However, this mechanical lock feature was eliminated in the present study to permit a more direct evaluation of the adhesive system. Additional control specimens were prepared by integrally molding the tooth-base assembly entirely from denture base resin so that no bond line existed between the tooth and the denture base support.

APPARATUS

The cantilever anchorage fixture shown in Figure 7 was mounted on an Instron universal testing machine. The behavior of simulated denture specimens in this apparatus had been previously demonstrated[4] to correlate well with practical experience on tooth retention and fracture resistance as observed over a 15-year period with actual dentures.

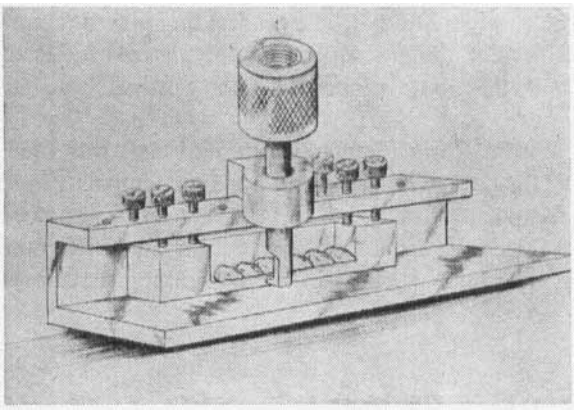

FIGURE 7 Fixture for quasi-static anchorage test.

The holding fixture for dynamic tests, shown in Figure 8, was mounted in a standard Izod impact test apparatus.[23] This fixture employed strain gages (EA–06–125BB–120, Micro-Measurements, Romulus, Mich.), whose signals were amplified by a bridge amplifier meter (BAM–1M, Ellis Associates, Pelham, N.Y.) and detected by a digital storage oscilloscope (1090A Nicolet Explorer, model 93A, Nicolet Instrument Corp., Madison, Wis.). The output from the oscilloscope memory was fed to a Houston, Omnigraphic x–y

recorder, model 2000 (Houston Instrument Div., Bausch & Lomb, Bellair, Texas) to obtain a hard copy force-deflection record.

Failure events in static testing were observed using a motion picture camera, Maurer model 05, 3 inch lens, with 16 mm high speed Ektachrome (tungsten) film. Fracture surfaces were examined with the aid of a Cambridge Stereoscan scanning electron microscope.

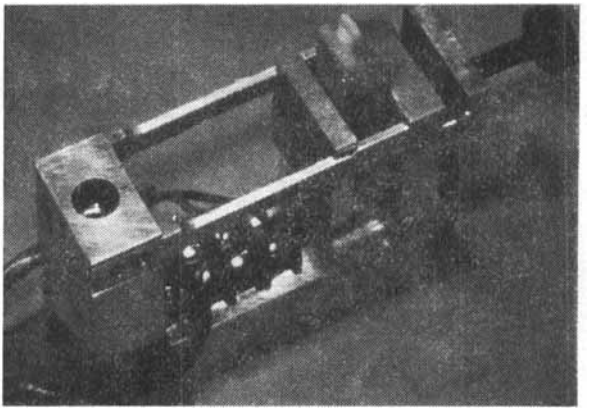

FIGURE 8 Fixture for dynamic impact test.

RESULTS AND DISCUSSION

The fracture surface topography of the acrylic teeth indicated a relatively uniform, one-phase, high molecular weight microstructure. The denture resins, however, retained a relatively heterogeneous morphology after polymerization. Their multiphase microstructure contributed to their fracture sensitivity under use conditions which were simulated by flexural tests at quasi-static and impact strain rates.

Under the quasi-static and dynamic impact conditions of this study, the tooth–denture resin assembly was substantially stronger flexurally and more impact resistant than the same configuration molded entirely from denture resin. Under static conditions, the denture resin had a flexural strength of 75 MPa. The polymer from which the acrylic teeth were molded had a flexural strength of 130 MPa (Figures 9 and 10). These values were obtained on specimens fabricated and tested under conditions as nearly optimum as possible.[24] Slow rate testing was also completed for a system in which the entire tooth–denture base assembly was constructed from the denture base resin. In this system, fracture propagated through the "tooth" region above

the denture base support at an average applied stress of 45 MPa, thus indicating a stress concentration factor of approximately 1.7 exists in the tooth geometry.

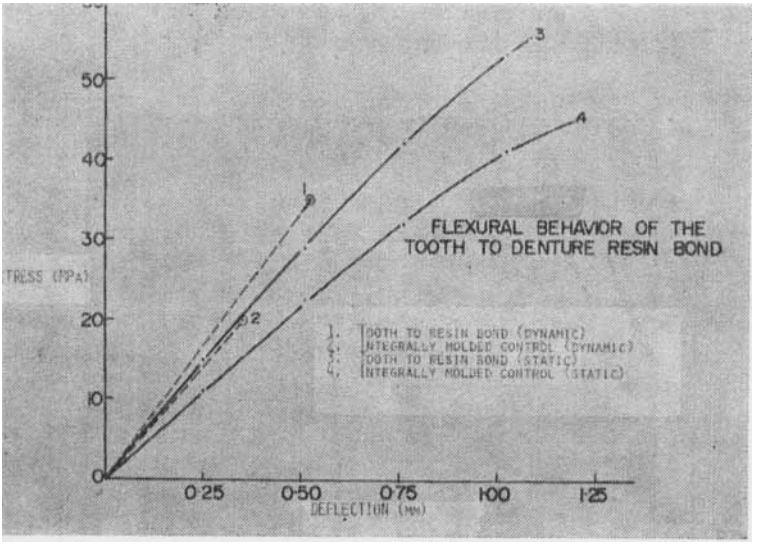

FIGURE 9 Flexural behavior under static and dynamic conditions.

The acrylic tooth–denture resin assembly failed at the tooth–denture base joint with an average applied stress of 55 MPa. The much weaker tooth–denture configuration may be explained by invoking qualitative fracture mechanics behavior criteria. At the surface adjacent to the tooth–denture base bond, the stresses as determined from linear elastic analyses are mathematically infinite.[1] In real materials, of course, infinite stresses are not realized due

Flexural behavior of denture materials

Property	Tooth polymer	Denture polymer	Tooth–denture bond	Integrally molded tooth–denture resin
Ultimate static flexural strength (MPa)	130	75	55	45
Ultimate dynamic flexural strength (MPa)	150	85	35	20
Ultimate static deflection (m)	0.008	0.006	0.001	0.001
Ultimate dynamic deflection (m)	0.002	0.0006	0.0005	0.0003

FIGURE 10 Flexural properties of polymer and tooth–denture base bond at ambient temperature. Static = 8.3×10^{-6} m/s, dynamic = 3.5 m/s.

to such mechanisms as plastic flow. However at such points, brittle fracture initiates and propagates at much lower average stress levels than predicted by bulk mechanical behavior.

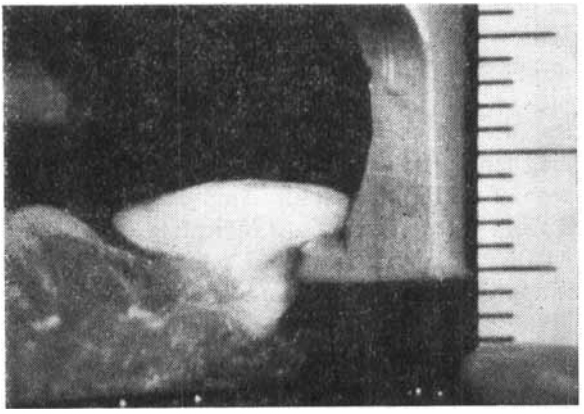

FIGURE 11 Fracture initiation under quasi-static condition at stress whitened tooth–denture resin juncture.

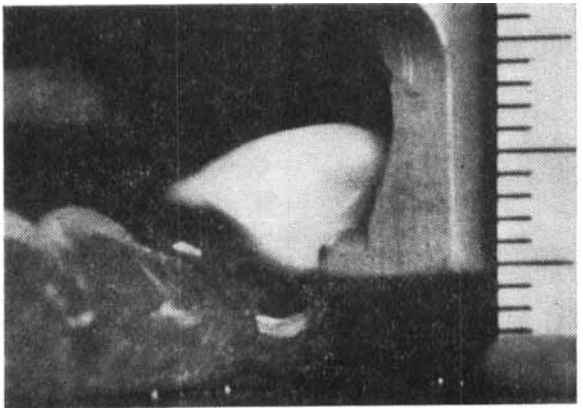

FIGURE 12 Fracture trajectory under quasi-static conditions.

The cinematography of the static flexural tests confirmed the influence of sample geometry on fracture resistance. Figures 11 and 12 are sequential frames from a test series in which clear denture resin was used to permit observation of fracture initiation and crack trajectory. Fractures initiated in a stress-whitened region of the denture resin or in the ridge lap of the tooth

at the tooth–denture resin margin. The crack trajectory followed the interfacial region producing fracture alternating between tooth and denture resin. This is shown more clearly in Figure 13. The fracture initiation and propagation phenomena have been described previously.[4, 5]

Interestingly, the failure phenomena were observable by cinematography at the relatively slow speed of 64 frames per second. Even though the cracks initiated and propagated much faster than the camera could record on a single test filming, parts of the events were captured. By compiling and editing several filmings, the events were clarified by sequential frame-by-frame inspection. Using large film size and slow speeds permitted better lighting and greater depth of field in the photos. An average crack propagation rate was calculated from these observations indicating that the crack travelled approximately 0.01 meter in 0.01 to 0.001 seconds. This corresponds to an average rate of approximately 1 to 10 m/s.

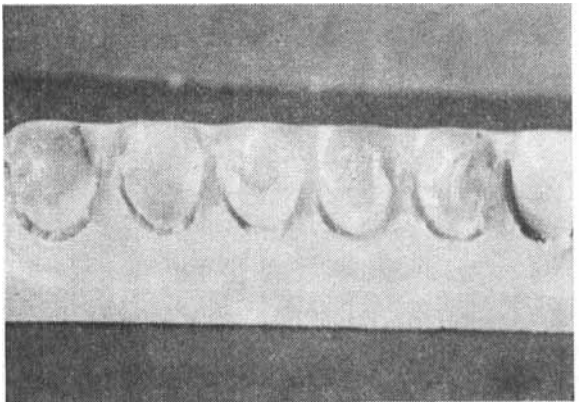

FIGURE 13 Fracture surfaces—quasi-static conditions.

This relatively slow crack propagation rate was approximately one-tenth of the observed average dynamic crack propagation rate. Under dynamic conditions, the applied load as a function of time was recorded and stored in the digital oscilloscope. The failure time was typically 125 ± 25 micro-seconds. This corresponded to an average crack propagation rate of 0.01 m/0.000125 s or approximately 100 m/s.

The crack initiation sites and crack trajectories in the bonded and control assemblies were similar under dynamic impact and static conditions. However, the flexural strengths observed for both bonded and control specimens were much lower under dynamic impact conditions (Figures 9, 10). This provided additional insight into the fracture behavior of the interfacial region.

Localized high stress concentrations were not sufficiently relieved by deformation processes under high strain rate impact conditions and caused fracture at relatively low average stress levels. In parallel tests on ASTM Izod specimens not subject to these stress concentration geometries, the critical bending tensile stresses increased as expected under dynamic conditions.

The information of flexural modulus and crack trajectories was used to construct the computer grid (Figure 5) which yielded the strain energy release rate data plotted in Figure 6. The strain energy release rate curve indicated that a crack driving force (G) of 50 J/m^2 would be required to propagate a small crack, surface scratch, or inherent flaw with a depth in the range 6×10^{-5} m to 1.5×10^{-4} m. For inherent flaws where $G_0 = 50$ J/m^2, the tooth–denture bond strength of 55 MPa for static loads and 35 MPa for dynamic loads were obtained. These strength values correspond to critical loads of X_1 and X_2 where:

$$X_1 = \text{load at static failure} = 25 \text{ kg}$$
$$X_2 = \text{load at dynamic failure} = 16 \text{ kg}$$

Thus, the critical static and dynamic strain energy release rates are found using Eq. (1)

$$G_c^s = \left(\frac{X_1}{25}\right)^2 50 = 50 \text{ J/m}^2$$

and

$$G_c^d = \left(\frac{X_2}{25}\right)^2 50 = 20 \text{ J/m}^2$$

and therefore G_c is dependent on load rate. This load rate dependence is discussed further in Ref. 1.

If, on the other hand, a value of G_c is known from other laboratory testing, the energy release rate curve can be used to determine the flaw size that could be tolerated for the 25 kg applied load. For example, if G_c has a value of 100 J/m^2 for a given load rate, cracks 1.6 mm deep could be tolerated. However, if G_c drops below about 50 J/m^2, the 25 kg load will always cause failure of the tooth–denture base assembly. This method of stress analysis provides an additional predictive method for the behavior of materials in dental prosthesis configurations.

SUMMARY AND CONCLUSIONS

A combination of static and dynamic mechanical tests has been developed as predictive tools for optimizing dental restorative and prosthetic materials and their use configurations. These tests included: flexural behavior under static

and dynamic (impact) conditions, cinefractography, scanning electron and polarized light fractography, and finite element stress analysis.

These tests were used to investigate the fracture surface morphology, stress–strain behavior, and fracture mechanics behavior of the tooth–denture resin attachment system.

Strong geometrical (fracture mechanical) and microstructural dependencies were observed for the mechanical behavior of the acrylic tooth–denture base assembly. This indicated that high values of bulk mechanical properties in dental materials are not sufficient in themselves to guarantee their desired performance in dental restorations or appliances under use conditions. Both chemical constitution and processing conditions critically influenced the performance of dental articles.

Proper consideration of the fracture mechanics, chemical constitution, and microstructural parameters provided a meaningful insight into the structure–property relationships of dental materials and a means to predict and optimize their clinical performance.

While the material presented here has dealt almost exclusively with the tooth–denture base system, the methods and procedures are of much more general application. Adhesive fracture mechanics in conjunction with carefully selected and planned tests yield fundamental system parameters that can be used to predict behavior of practical adhesive geometries, optimize joint configuration, etc. The authors and others are currently applying the technique to dental, medical, aerospace and a variety of other problems.

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