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Mechanical Biocompatibility of Dental Implant Materials

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The aim of the present research is to investigate the dynamic mechanical behavior of dental implant materials and its effect on neighboring bone and soft tissues. A novel diagnostic system has been developed that can quantitatively measure the damping capacity of material samples and structures including natural teeth and dental implant components in vivo. Damping capacity values have been determined for natural teeth and a variety of different dental superstructures including those made of resin matrix composites, gold alloys, and porcelain fused to gold laminates. A comparison of these values indicates that the implant-supported structures are typically more conservative energetically than their natural tooth counterparts which include a thin ligament between the tooth and bone. Implications of this difference in dynamic behavior for unwanted physiological responses and the susceptibility to fatigue damage are discussed.

Keywords: *mechanical biocompatibility, energy dissipation, damping capacity, loss coefficient, implant materials, dynamic behavior, tooth intrusion*

1. Introduction

Damping capacity, the ability to dissipate mechanical energy, is an important property of the natural tooth complex and is primarily performed by the periodontal ligament. When a tooth is subjected to an impact force, a stress wave is transmitted through the tooth and into this ligament which serves as the connective tissue between the tooth and the bone. Because of the way it deforms, the periodontal ligament acts as a shock absorber, dissipating much of the energy associated with the impact.[1] This damping process can greatly reduce the resultant impact force transmitted to the surrounding bone. Dental implant prostheses often have no obvious means by which to dissipate significant amounts of mechanical energy because of the nature of the materials used. This difference in mechanical behavior may be particularly critical for bruxers and clenchers since they can impart relatively large impact forces on their teeth when sleeping.

There are numerous references in the dental implant literature regarding the importance of energy dissipation in dampening occlusal forces. The Brånemark System (Nobel Biocare USA, Yorba Linda, CA) originally advocated that the contact surfaces of prostheses be reconstructed with acrylic resin for this purpose.[2] Jempt[3] concurred with this recommendation after observing that complete restorations veneered with this material demonstrated an apparent reduction in the rate of fatigue fractures in the anchorage components. The IMZ System (Friadent, Irvine, Calif.) had an internal shock absorber made of a medical-grade polyoxymethylene (POM) incorporated into the implant design.[4] Several articles have addressed the potential significance of a periodontal ligament substitute for the rigid implant structure.[2-7] Although these studies have elucidated the importance of the mechanical damping in dental structures, few quantitative studies of energy dissipation in natural periodontum as well as that in implant prostheses have been reported.[8,9] However, research results that highlight the importance of dynamic loading and energy dissipation on the general process of bone remodeling are becoming more common.[10,11]

The relative extent to which a material dissipates elastic mechanical energy for this situation may be characterized using the loss coefficient, η, given by

$$\eta = \frac{D}{2\pi U} \tag{1}$$

where D is the total energy dissipated per unit volume and U is the total strain energy per unit volume generated at the maximum displacement.[1] The energy of an elastic wave attenuates after it has traveled a relatively short distance in materials with a high loss coefficient. By contrast, stress waves in materials that have a low loss coefficient can travel much farther before attenuation occurs.

A schematic depicting the propagation of stress waves from a superstructure into natural teeth is illustrated in Fig. 1. Stress waves from point of percussion, P, are attenuated in the three periodontal ligaments surrounding supporting teeth. A schematic depicting a superstructure supported by two implants and a natural tooth is illustrated in Fig. 2. As shown in Fig. 2a, stress waves may reach the tooth through the implants and bone as well as from the superstructure directly. The resultant impact force that is imparted on the periodontal ligament of the supporting tooth depends on the initial energy of the percussion and on the damping properties of the cast gold alloy superstructure ($\eta \approx 5\times10^{-3}$),[1] the titanium implant ($\eta \approx 10^{-4}$),[1] and bone ($\eta \approx 10^{-2}$).[12] By contrast, stress waves generated at the surface of natural teeth must pass through the soft tissue ligament before reaching neighboring teeth as illustrated in Fig. 1. Significant impact stresses may be transmitted into the supporting tooth as long as it remains in sufficient contact with the superstructure.

Stress waves will eventually diminish in the tooth and intrusion will cease once the contact area becomes sufficiently small, as depicted schematically in Fig. 2b. At this point, a tooth is stable until changes are made with regard to the manner in which it is joined to the prosthesis. Previous clinical observations have indicated that the absence of dynamic mechanical stimulus can result in extrusion of an intruded tooth. For example, reduction in mechanical stimulus when an energy dissipating element is subsequently included in a superstructure has led to reversal of intrusion.[8, 9] This phenomenon is shown schematically in Fig. 2c illustrating how the tooth receives very little mechanical stimulus from the superstructure once the area of contact reaches a sufficiently low value. Reversal of intrusion (emergence) under these conditions was observed in two separate cases when partial implant supported superstructures were modified so that the natural tooth received less mechanical stimulus.[8, 9] Hence, it appears that

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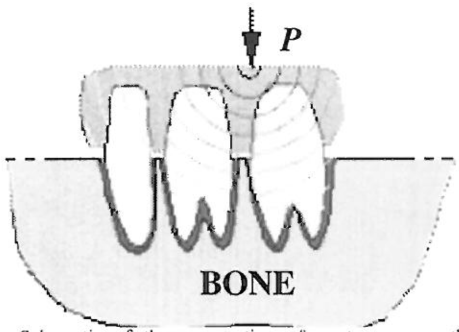

Fig. 1. Schematic of the propagation of a stress wave through a superstructure supported by natural abutments. The stress wave must pass through the periodontal ligaments surrounding the roots of the teeth.

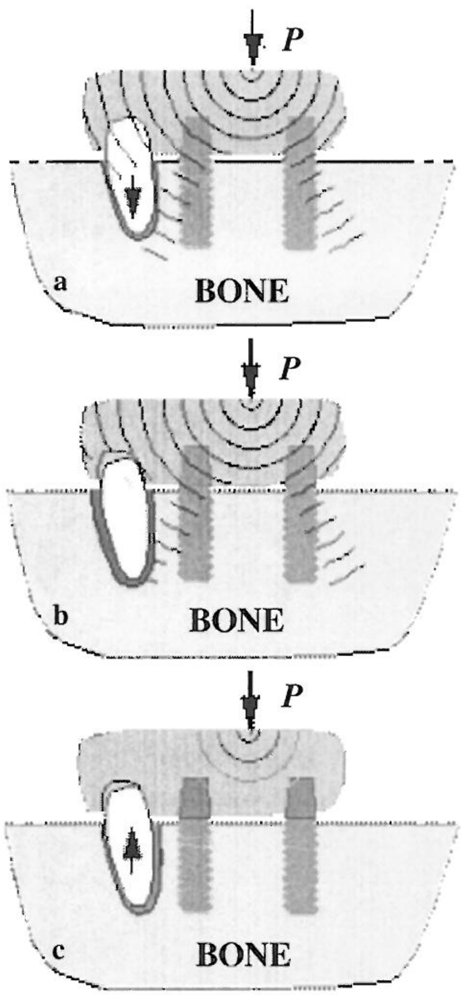

Fig. 2. Schematic depicting tooth intrusion and reversal as proposed by Sheets and Earthman.[8, 9] a) The integrated tooth is the primary recipient of stress waves resulting in intrusion. b) Stimulus/response equilibrium results once the tooth has intruded a sufficient distance from the superstructure. c) After modifying the implant structure to dissipate mechanical energy, extrusion results from a lack of mechanical stimulus until the tooth is reseated.

an optimum level of mechanical stimulus is required to maintain osteoclastic/osteoblastic equilibrium. Excessive stimulus results in intrusion while insufficient stimulus brings about extrusion. This behavior is consistent with the hypothesis proposed by Fung[13] that a certain range of stress is required for bone growth while either an insufficient or excessive amount causes bone resorption. This relationship between bone remodeling and stress is shown schematically in Fig. 3. For dynamic impact loading, the stress amplitudes transmitted can depend strongly on the damping capacity of the structure. In an earlier study, it was found that a 10% increase in damping capacity can result in a 40% decrease in the impact loads transmitted through a dental implant structure in vitro.[9] Although this difference in stress amplitude is not generally great enough to cause overload fracture, it can be sufficient to substantially decrease fatigue life of the implant, superstructure, or bone. Thus, it appears that a moderate level of damping in the implant structure is necessary for optimized mechanical biocompatibility and fatigue resistance. Instrumentation was developed in our laboratory to measure quantitative values of η for natural teeth and dental implant structures. Results obtained with this instrumentation to achieve a better understanding of the factors that determine the mechanical biocompatibility of dental implant materials are described in the following.

2. Tests and Procedures

Diagnostic dental instrumentation and software, referred to as the Periometer, was used for quantitative measurement of mechanical damping in a range of materials and dental structures.[9, 14-16] A schematic of the present energy dissipation test system is illustrated in Fig. 4. The experimental system consists of hardware and software that are interfaced to a Siemens Periotest that has been modified to quantitatively and reproducibly measure the energy damping capacity of dental structures. This system performs measurements of the damping capacity defined above as the loss coefficient, η. The energy dissipation measured by the Periometer is representative of realistic damping achieved in the human mouth since the loading paradigm of the Periometer is consistent with impact force amplitudes and rates produced by mastication.

A stress wave propagates through the Periotest probe upon its percussion with a specimen. Conservation of energy dictates that the elastic strain energy associated with this stress wave is given by

$$E_r = U - D - D_p \tag{2}$$

where U is the input energy equal to the kinetic energy of the probe just before impact, D is the energy dissipated by the specimen, and D_p is energy dissipated at the interface between the probe and the specimen as well as any sources of energy loss in the probe. Substituting into Eqn. (1), the loss coefficient for the inelastic specimen becomes

$$\eta = \frac{1}{2\pi}\left(1 - \frac{E_r + D_p}{U}\right) \tag{3}$$

A calibration of the present technique is preformed periodically to determine accurate values of D_p and U using model material specimens.

In Vitro Testing:

Testing was conducted using a 40 x 25 x 27 mm simulated bone model made of resin acrylic (Dentsply, Inc., York, PA). Two Nobel Biocare implants were placed into

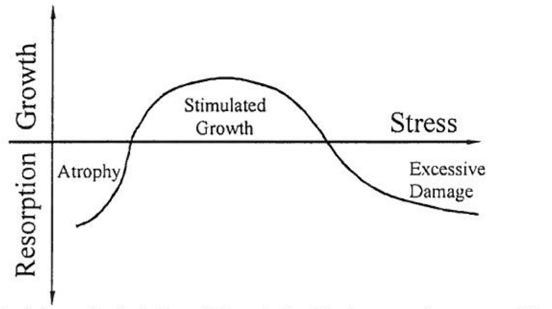

Fig. 3. Schematic depiction of the relationship between bone remodeling and applied mechanical stress (adapted from Fung[13]).

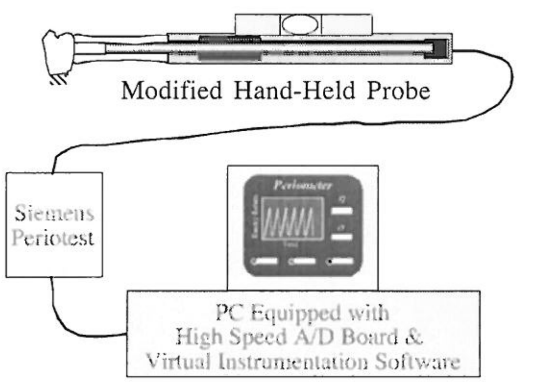

Fig. 4. Schematic depiction of the Periometer Instrumentation used in the present study.

the model and separated by approximately 9.0 mm in order to mimic intraoral implant positioning covering a missing site. Each implant was then fitted with a prefabricated DIA abutment (Nobel Biocare USA, Yorba Linda, CA) on which prosthetic superstructures are mounted for testing.

Four superstructure materials were tested on the simulated bone model: Gradia™ ceramic particulate reinforced (CPR) resin matrix composite fused to gold (GC America), belleGlass™ CPR resin matrix composite fused to gold (Kerr Manufacturing Co.), porcelain fused to gold (PFG), and cast gold. For the Gradia and belleGlass superstructures, the composite is laminated to the underlying cast gold at room temperature. By contrast, the porcelain for the PFG structure is fused to the underlying cast gold at 900°C. Each superstructure consists of three sites with measurements taken only on the first and third which are directly above the supporting implants. Each superstructure was seated using TempBond temporary cement (Kerr Manufacturing Co.) and allowed to cure for 10 minutes prior to testing. Measurements were made with the Perioimeter probe oriented in the buccal direction (normal to the tooth root axis) and the occlusal direction (aligned with the tooth root axis). These measurements were facilitated by flat surfaces on the occlusal side of the in vitro model superstructure. The superstructures were then tested 26 separate times for each site, percussion direction, and superstructure material. Since each Periometer determination consists of ten percussions by the hand held probe, 260 loss coefficient measurements were made for each site. For in vivo testing, measurements were only performed in the buccal direction.

In Vivo Testing:

Subject JS, a 77 year-old male, was tested at sites 4-6, 12, 14, 18, 19, 28, and 30. The locations of these sites within the periodontum are indicated in the chart in Fig. 5. Sites 4, 5, 6, 18, and 19 are supported by Nobel Biocare implants and custom milled gold abutments. Frialit-2 implants (Friadent North America, Inc., Irvine, CA) support sites 12, 14, 28, and 30 with custom milled gold abutments. The four superstructure materials tested in vitro (Gradia fused to gold, belleGlass fused to gold, PFG, and cast gold) were also tested in vivo for JS. In addition to the sites supported by implants, PFG crowns supported by natural tooth abutments were also tested at sites 7-9, 11, 20, 22, and 24-27. Each implant-supported superstructure was cemented onto the respective sites using TempBond temporary cement, with testing initiating after a ten-minute curing period. The superstructure at each site was tested three times at the buccal aspect and an average reading for each site was obtained from the corresponding 30 individual measurements by the Periometer.

3. Results

In Vitro Testing:

Loss coefficient values for different superstructures tested in vitro are illustrated in Fig. 5. As shown in this histogram, the composite materials give higher values of the loss coefficient compared to those for cast gold or porcelain fused to gold superstructures. It therefore appears that the polymer matrices of the composites significantly enhance the damping capacity of these superstructures. It can also be seen that the average η values determined with impact in the buccal direction are substantially higher than those with the impact in the occlusal direction. Overall, these results indicate that the energy dissipation by the composite superstructures is as much as twice that for cast gold or porcelain fused to gold. The presence of many ceramic-polymer interfaces within the composite probably also enhances the damping of the composite.

The data in Fig. 5 indicate that the average loss coefficient for PFG superstructures are somewhat greater than those for cast gold superstructures. This may seem counter-intuitive since the loss coefficient for solid porcelain ($\eta \approx 5\text{x}10^{-4}$)[1] is considerable less than that for cast gold ($\eta \approx 5\text{x}10^{-3}$).[1] However, it is likely that the microstructure of the cast gold in the PFG superstructure is softened during the fusion process at 900°C so that its damping capacity is substantially increased. Tests are currently underway to investigate this hypothesis. It is reasonable that the lower loss coefficient of the relatively thin porcelain layer has only a moderate contribution to the overall loss coefficient in comparison with the changes in the underlying gold.

In Vivo Testing:

A histogram of average loss coefficient for different superstructures acquired in vivo with human subject JS is illustrated in Fig. 6. The superstructures tested were all supported by implants except for those corresponding to the farthest column to the right which were PFG crowns supported by natural teeth. All Periometer measurements were made in the buccal direction. The average loss coefficient values plotted in this figure are consistent with the in vitro data in Fig. 5, although slightly greater overall. We also note that, although loss coefficient values for the resin matrix composite superstructures are greater than those

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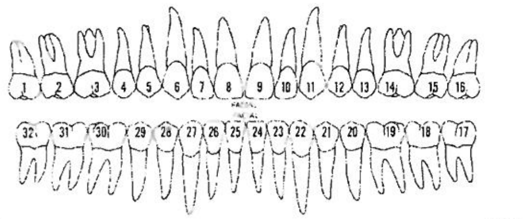

RIGHT LEFT

Fig. 5. Convention for the site numbers used in the present work.

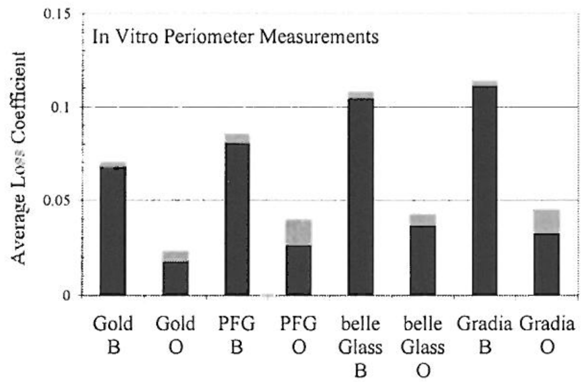

Fig. 5. Average loss coefficient data for different orientations and superstructure materials on the present in vitro implant model. The shaded portions of the columns represent the standard deviations while the height of the black columns indicates the average value of the loss coefficient.

for gold and PFG, they are still somewhat less than those for PFG crowns on natural tooth abutments.

4. Discussion and Conclusions

A comparison of the in vivo and in vitro data obtained in the present study indicates good agreement with regard to the inherent damping capacities of four superstructures consisting of different materials. The lowest values of loss coefficient were observed for the cast gold superstructures contrary to the traditional clinical view that these superstructures are more forgiving than porcelain fused to gold superstructures. Somewhat higher values were observed for porcelain fused to gold. The effect of the porcelain fusion process on the damping properties of the underlying cast gold is not known. However, it is reasonable to assume that the damping capacity of underlying cast gold increases due to microstructural softening that would occur at the fusion temperature of 900°C.

The greatest values of loss coefficient among the superstructure materials were obtained for particulate reinforced polymer matrix composites, Gradia and belleGlass. This result is consistent with the fact that the polymer matrix of these materials has a relatively high damping capacity ($\eta \approx 10^{-2}$)[17] compared to cast gold or porcelain.

The present work indicates that composite superstructures provide energy dissipation that is closest to that measured for natural dentition. It is therefore likely that these materials provide the best biocompatibility in terms of avoiding excessive stimulus to natural tissues as well as preventing fatigue damage in the implant or bone due to repeated impact loading. However, long term studies are needed to determine the longevity and reliability of the

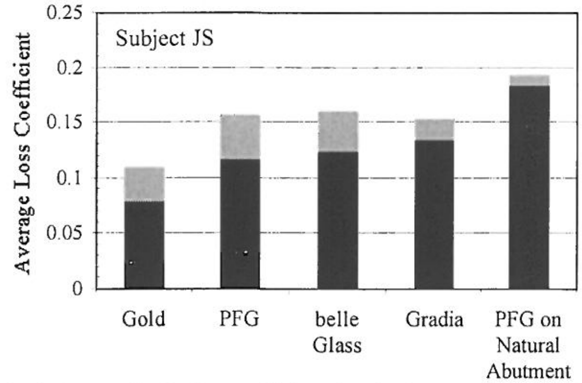

Fig. 6. Average in vivo loss coefficient data for human subject JS. The shaded portions of the columns represent the standard deviations while the height of the black columns indicates the average values of loss coefficient. All measurements were made in the buccal direction.

composite superstructures themselves under severe parafunctional conditions in comparison to gold and PFG superstructures.

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