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THE INFLUENCE OF THE PROSTHETIC MATERIALS STIFFNESS ON LOAD TRANSFER TO DENTAL IMPLANTS

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Abstract. *It is believed that the occlusal covering with less rigid materials, such as acrylic resins could reduce load transfer to the bone. The purpose of this research is to compare, by photoelastic and dynamic analysis, the load transfer to single implants, varying the occlusal materials. A photoelastic model was created, with an anatomy similar to a mandibular premolar region. An internal hexagon implant (3.75 x 11.5 mm) was put in this model. Three different types of prostheses were constructed with different degrees of stiffness: 1- Metallic (high); 2- Composite resin (moderate); 3- With a layer of ethyl-vinyl-acetylene (EVA, lower). By means of a circular polariscope, the stress generated after a compressive axial load of 100 N, was observed qualitatively for each prosthesis, by two calibrated examiners. For the dynamic analysis, a modeling of each implant-prosthesis was carried out with Voigt models. After that, a mathematical representation of the systems was developed and solved, taking into account the mechanical* *properties of the materials. To represent a physiologic masticatory load, a half cycle sinusoidal force was applied (t=0.25 s; Fmax=100 N), by the expression* $f(t) = 50 - 50.\cos(8\pi t)$ *. The photoelastic analysis showed stresses at the apical and cervical region of the implant, with similar intensity and pattern among the prostheses. The dynamic analysis, showed that the force applied over the prostheses were transferred to the implant instantaneously. The results of this study determine that the occlusal materials do not alter masticatory load transfer over single implants.*

Keywords*: osseointegrated implants – Dental Phrostesis – dynamic analysis – photoelasticity – biomechanics.*

1. Introduction

Occlusal overloads on implants are known to cause cervical bone losses (Brunski, 1999), fractures of implants and looseness of components (Bidez, 1992). Alternatives to reduce the forces transmitted to implants have been studied, including variations in implant positioning, implant design, prosthesis shape, occlusal requirements, prosthetic components and prosthetic materials (Skalak, 1983; Papavaliou *et al*., 1996; Spiekermann, 1995). It was postulated that the biomechanics of the implants would be improved if a mobility similar to the one allowed by the ligament periodontal was incorporated (Richter, 1989).

Kirsch (1983) proposed the use of implants containing a plastic component, which would imitate the properties of the periodontal ligament. It was also proposed that the use of prostheses in acrylic resin would contribute to dissipate a significant portion of the impact forces during mastication, due to the low stiffness of this material (Skalak, 1983). Based on that concept, the use of occlusal surfaces in resin has been recommended in clinical protocols, to control the load transfer to implants (Spiekermann, 1995; Misch, 1993). However, the biomechanical benefits of resins could not always be proved by the literature, and in some studies, other materials were more efficient (Inan and Kesin, 1999; Stegaroiu *et al.* 1998). Among in vivo studies, it was not possible to detect differences among the forces transmitted by prostheses in acrylic resin and other materials as metal or porcelain (Bassit, Lindstrom and Rangert, 2002; Duyck *et al.,* 2000; Hobkirk and Psarros, 1992). Still, a prospective clinical study of 509 implanted patients did not pointed to a correlation among the incidence of clinical implant failures and the occlusal materials (Naert *et al.*, 1992).

The photoelastic method allows a direct observation of stress distribution in structures, based on the ability of certain transparent materials to exhibit color patterns when stressed and observed under polarized light (Caputo, 1987; Guichet *et al.*, 2000). Dynamics studies analyze the movement of bodies under the action of forces, which allows to simulate determine the deformations, the time of transmission of the pulse of force and the intensity of the force transmitted (Den Hartog, 1985).

The present research has the objective to study the biomechanical behavior of single implants, varying the materials of the prosthetic crown. To achieve that, the stiffness of the prosthetic crown was varied in three levels: high, moderate and low.

2. Material and methods

2.1 photoelastic analysis

To obtain a photoelastic model, a mandible section of the premolar area, with a triangular base, was obtained in type II gypsum, from a didactic cast of the Department of Prosthodontics of FOUSP, São Paulo-SP, Brazil. A perforation of 15 x 5 mm was drilled in the cast. In that hole, it was positioned with cianoacrilate and type IV gypsum, an internal hexagon implant analog (013070, Conexão, São Paulo - SP, Brazil), as indicated in Figure 1.A. A layer of colorless nail varnish applied on the cast, and soon afterwards, an impression coping was positioned (Fig. 1.B). An impression was done then with a silicone rubber (Silibor, Clássico, São Paulo-SP, Brazil) (Fig. 1.C). After 24 h the cast in silicone was removed and an internal hexagon implant (11.5 x 4.0 mm, Conexão) was screwed in the impression abutment trapped in the impression material. The photoelastic resin (PL-2, Vishay Measurements Group, Raleigh, USA) was manipulated and then flowed in the impression, that was taken to a vacuum camera for 15 min (Fast Vac, JB, Brazil), to eliminate bubbles (Fig. 1.D). After 48 h, the model was liberated and in a preliminary photoelastic evaluation, it was considered free from stress (Fig. 1.E). Three single prostheses were made, a copper-aluminum one (CuAl),[18] one in composite resin (CR)[10] and one with a layer of ethyl-vinyl–acetylene (EVA), considered respectively as of high, moderate or low stiffness. The materials used in the prostheses were selected for their elastic modulus of and Poisson coefficient (Table 1). For the accomplishment of the prostheses, golden UCLA abutments were used (056072, Conexão), and then frameworks with h = 10 mm ϕ= 7.5 mm were constructed (Figure 2).

Material	**m (Kg)**	**ϕ (m)**	**h (m)**	**K (10^6 N/m)**	***c* (10^6 Ns/m)**	**E (GPa)**	**v**
CuAl	2.78×10^{-3}	7.5×10^{-3}	10×10^{-3}	486.0	-	-	-
CR	0.29×10^{-3}	7.5×10^{-3}	5.0×10^{-3}	176.7	-	-	-
CuAl	1.32×10^{-3}	7.5×10^{-3}	5.0×10^{-3}	971.9	-	-	-
EVA	0.01×10^{-3}	7.5×10^{-3}	2.5×10^{-3}	1.8	0.424	0.1*	0.49 (Crede, 1965)
CR	0.14×10^{-3}	7.5×10^{-3}	2.5×10^{-3}	353.4	-	20	0.28 (Stegaroiu, 1998)
CuAl	1.29×10^{-3}	7.5×10^{-3}	5.0×10^{-3}	971.9	-	110	0.33 (Carvalho, 1986)

TABLE 1 – Data considered for the calculations. CuAl – Copper aluminum alloy (NPG, NPG +, dental Aalba, USA); CR - Composite resin . (Synfony, 3M, USA); EVA - ethyl-vinyl-acetylene (Kreateva, Brasil). * http://www.maropolymeronline.com/Properties/modulus_values.asp

Marginal gaps measurements, were conducted, in six regions of each implant-component interface, by means of an optical microscope with 170 x magnification (Toolmaker microscope; Mitutoyo, Tokyo, Japan), and the maximum distance gaps should not exceed 10 µm (Rangert, Jemt and Jorneus, 1989). After the installation of the prostheses, a 20 N.cm torque control (Lifecore Biomedical Inc, Chasca, MN) was used to attach the framework to the implant. The model was submitted to photoelastic analysis, with a circular polariscope, at two moments: (i) before test and (ii) after applying a load of 100 N. The 100N load (Soumeire and Dejou, 1999**)**, applied to the implant, was obtained through a testing machine (Versatest, Mecmesin, England). Prior to each of these photoelastic analyses, the polymeric model was kept at 50ºC for one hour to relax the residual stress generated during model fabrication or testing. During the photoelastic analysis, the model remained immersed in mineral oil, to reduce light refraction (Nishimura *et al.*, 1999**)**. Results were registered by a high-resolution digital photographic camera (Cybershot W1, Sony, Japan). The interpretation of the results was accomplished, by two calibrated examiners, according to French *et al.* (1999): 1- The larger the number of fringes, the greater the magnitude of the stress; 2- the closer the fringes are to each other, the larger the stress concentration.

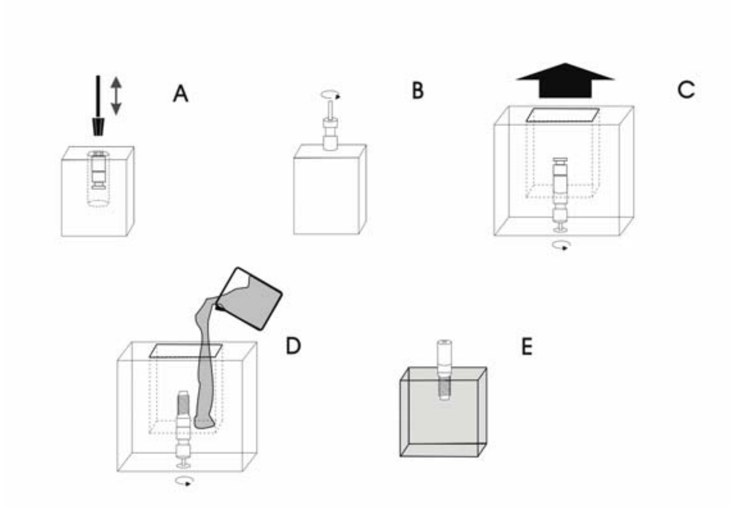


FIGURE 1 – Methods used to fabricate the photoelastic model. A) hole made in a gypsum cast and analog positioning; B) impression coping installed; C) impression; D) implant connected to the impression coping; pouring the photoelastic resin; E) manufacture of prosthesis.

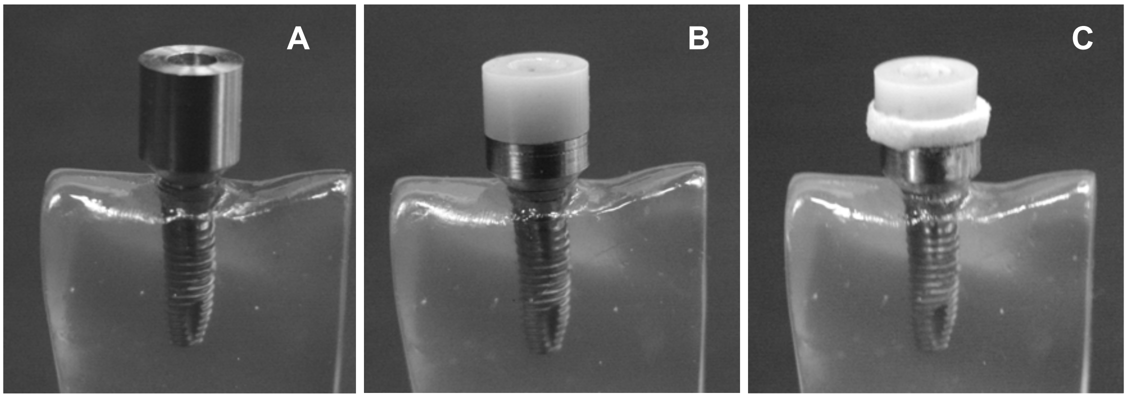


FIGURE 2 – Prosthesis with high (A), average (B) or low stiffness (C), positioned over the photoelastic model. The methodology of dynamic analysis consists in the transformation of the biological model into a physical model (Voigt models) and soon afterwards into a mathematical model (Figure 3).

2.2 dynamic analysis

The titanium was considered as a stable and rigid base and the bone was not modeled. The elastic constants were calculated assuming that in a body submitted to axial compression, stress (σ) and strain (ε) are related by the elastic modulus (E) Eq. (1):

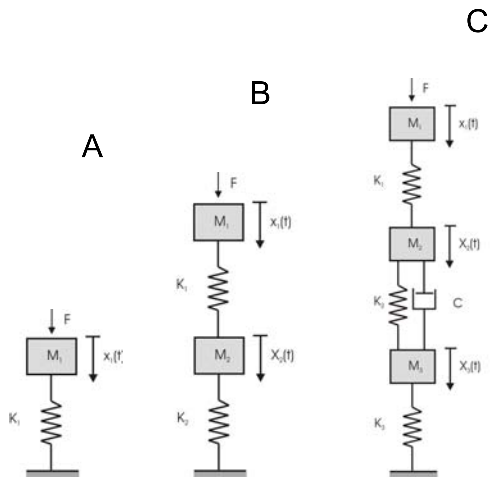
$$\sigma = E\varepsilon \Rightarrow \frac{F}{A} = E\frac{x}{l_0} \Rightarrow \frac{F}{x} = E\frac{A}{l_0}$$

In the other hand, in a mass-spring model, force and deformation are related by the Hook´s law Eq. (2):

$$F = kx \Rightarrow \frac{F}{x} = k$$

Joining Equations (1) and (2) results in the Equation (3)

$$k = \frac{EA}{l_0} = \frac{\pi.\phi^2 E}{4.l_0}.$$



A $\left\{ m_1 \ddot{x}_1 + k_1 x_1 = u \right.$

B $\left\{ \begin{array}{l} m_1 \ddot{x}_1 + k(x_1 - x_2) = u \\ m_2 \ddot{x}_2 + k_1(x_2 - x_1) + k_2 x_2 = 0 \end{array} \right.$

C $\left\{ \begin{array}{l} m_1 \ddot{x}_1 + k(x_1 - x_2) = u \\ m_2 \ddot{x}_2 + k_1(x_2 - x_1) + k_2(x_2 - x_3) + c\ (x_2 - x_3) = 0 \\ m_3 \ddot{x}_3 + k_2(x_3 - x_2) + k_3 x_3 + c(x_3 - x_2) = 0 \end{array} \right.$

FIGURE 3 – Physical and Mathematical representation of the prosthesis. A - metallic; B – composite resin; C - EVA. Each mass is equal to a material layer, with corresponding behavior as spring and/or shock absorber. k = equivalent rigidity, elastic constant; F & u= applied force; m = mass; x = displacement; c = damping constant.

The damping properties (c) were only considered for the EVA layer, because they are much smaller for the other materials (metallic alloy and composite resin). The calculation of the damping coefficient for EVA (c) was performed by the following method (Crede, 1969):

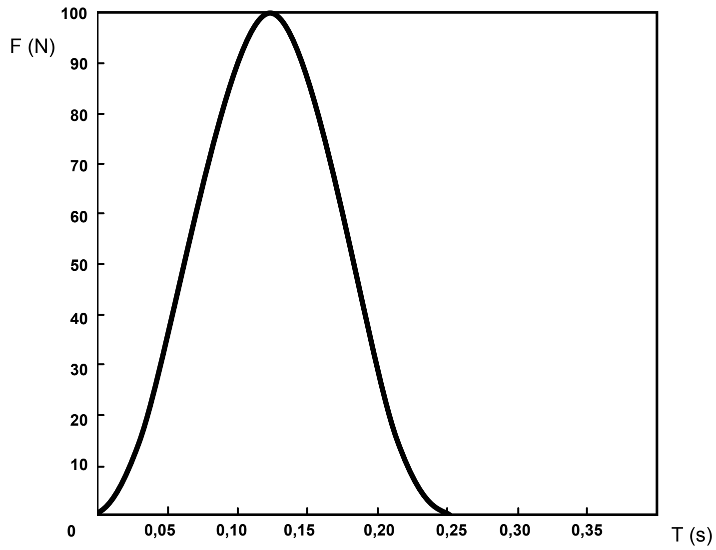
$$\zeta = \frac{c}{c_c} \cong 0{,}05$$

and $c_c = 2\sqrt{m.k} = = 2\sqrt{1.10^{-5}.1{,}8.10^{6}} = 8{,}48\,\frac{N.s}{m}$,

$$\therefore \; c = \zeta.c_c = 0{,}05.8{,}48 = 0{,}424 \,.$$

The force (*f(t)*), was considered a sinusoidal pulse with 0.25 s (Brunski, 1988) length and 100 N of maximum force(Ueda, 2004) (Graph 1), in other words (Eq.4):

$$F = 50\text{-}50.\cos\left(8\pi t\right)$$



GRAPHIC 1 – Load applied to the dynamic systems.

, where the speed of application of the force is $\omega_d = 8\pi \quad rad/s$. The inertial forces - those which try to maintain the movement of a body, after the cessation of the loading force (Halliday, Resnick, Walker, 2004) **-** induced to the metallic prosthesis, was calculated by the expression: $F_{inertial} = -m.a$. The matrixes shown in Figure 3 were solved by means of a calculation program (GNUOctave for Windows v.2.1.50), after the insertion of the calculated parameters (c, k, F), and the masses obtained from the photoelastic models. By this method, the force transfer to the base was analyzed during time, along with the displacements within each material layer of the prostheses.

3. Results

In the Figure 4, it was possible to observe the photoelastic results. A similarity in the stress fringe pattern among the prostheses can be noticed. Those fringes were concentrated around the medium third and the apical of the implants, what denotes an axial transmission of the force. The results of the dynamic analysis are shown in table 2. The transmitted maximum force was always equal to 100 N and the f(t) graphs of the applied and transmitted force were always similar (Graph 1). It can be noticed that the resin layer of the prosthesis in composite resin deformed over the metallic support. In the situation with the prosthesis in EVA, the resin layer compresses the EVA against the metallic base.

The inertial force calculated resulted to be by far smaller than the applied force: (Eq.5)

$$\frac{F_{inertial}}{F_{transmitted}} = \frac{4.10^{-6}N}{100N} = 4.10^{-8}.$$

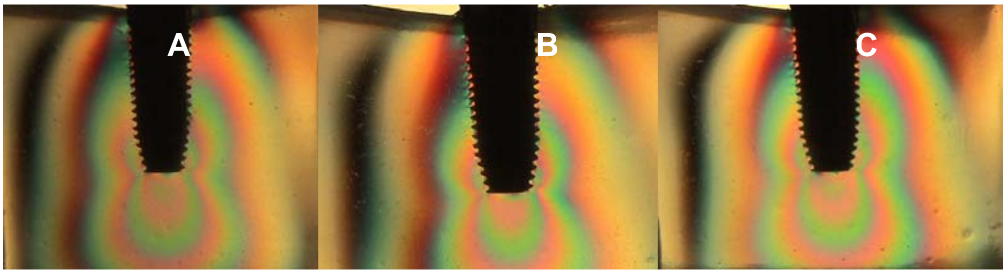


FIGURE 4 – Photoelastic fringe pattern obtained with the incidence of a 100 N over a metallic (A), in composite resin (B) and in EVA (C) prosthesis.

Prosthesis	Material	Deformation (µm)	Elongation (%)	Force transmitted (N)	Pulse length (s)
CuAl	metal	0.2	0.002	100	0.25
CR	resin	0.6	0.013	100	0.25
	metal	0.1	0.002		
EVA	resin	0.3	0.011	100	0.25
	EVA	55.6	2.220		
	metal	0.1	0.002		

TABLE 2 – Results obtained through the dynamic analysis.

4. Discussion

In this study it was possible to analyze how the loads are transferred to a single implant, depending on the stiffness of the prosthetic materials. The methods used (photoelastic analysis and dynamics) supplied complemental information to each other.

The similarity in the pattern of stresses among the photoelastic models can be justified for Newton's third law (law of the action and reaction), which enunciates that, for an applied force on a body, a reaction force appears with the same intensity and direction (Halliday, Resnick, Walker, 2004). In that way, as the applied force was set at 100 N, the stress induced to the model by the prostheses was similar, corresponding to the load of 100 N. In other words, the mechanical properties of the prosthesis did not change the stress induced to the implant. McGlumphy, Campagni and Peterson (1989) approximated this same conclusion when they analyzed photoelastically the biomechanics of single implant that contained or not an intramobile element.

Through the dynamic analysis, it was possible to simulate the mechanical behavior of prostheses if they were submitted to a masticatory pulse. In the same way that occurred in the photoelastic analysis, the similarity in the results among the prosthetic materials can be justified by Newton's third law. According to Skalak (1983), the viscoelastic behavior of an acrylic resin as occlusal material would be enough to delay the transmission of the force and to reduce its peak. However, the composite resin used in this study did not provoke any alteration in the transmission of the force. It could be imagined, then, that the composite resin would be too rigid to evidence that phenomenon, and that a material with a lower stiffness should be used to observe it. Though neither the magnitude nor the time of transmission of the force were altered by the use of EVA, which main difference was the presence of a large deformation, in comparison with the other materials, to transfer the force applied to it. The results of this study agree to the one of Van Rossen and Braak (1990), who determined, by the finite element method, that a variation of the elastic modulus of a prosthetic abutment between 0,15 and 110 GPa (titanium) did not induce different stresses after the application of a force on an single implant. Contradicting those results, Richter (1989) proposed that the use of a less rigid material in the prosthesis would approximate the biomechanics of the implants to the one of the teeth. He confirmed that theory in a subsequent study, in which it was verified, by the finite element method, a reduction of 20 times in the stress generated in a model under axial loads (Richter, 1990). The lack of consensus among biomechanical theories and the results of this study, brings doubts on the real biomechanical effects of the prosthetic materials in clinical procedures. However, some of the main assertions taken into account in literature were based on subjective parameters, or personal manifestations without corresponding scientific demonstrations (Skalak, 1983; Sheets and Earthman, 1993; Misch, 1999).

In dental prostheses, the damping coefficient (*c*) does not seem to alter the load behavior, as seen in the EVA model. Therefore, the only way to reduce loads over single implants would be with a displacement restraint caused by neighboring teeth. In a clinical situation, neighboring natural teeth to a prosthetic space can limit the displacement of the prosthesis to the one that happens due to the compression of the periodontal ligament under loads - 0,025 mm approximately - (Mosham and Berkovitz, 1982). A reduction in the transmitted force could occur when using prosthesis with low stiffness, admitting liquid or pasty feeding, that would not compress the prosthesis. Being known that a deformation of 56 µm was necessary to transmit 100 N, a deformation of 25 µm would achieve a maximum loading of 55 N, in other words, a reduction in 45%. The prosthesis in EVA would only apply to the implant the 100 N load when compressed by a sufficiently solid food. However, that phenomenon presents small deformation magnitudes and should be confirmed by further experiments. For being dependent of the displacement limitation by neighboring teeth, that mechanism would not exist in larger fixed prostheses, in which the prosthetic material would be deformed freely.

The reduced magnitude of the inertial forces in this study are a result of the reduced accelerations created during dental loading. As a consequence, there is almost no delay between the application of the force, the deformation of components and its transfer, and a quasi static situation is defined. Therefore, to predict the biomechanical behavior of forces in dental studies, such as those with the finite element method, static principles would simplify the analysis as affirmed by Spiekermann (1995) and Skalak (1983). As consequence of this results one can also affirm that mastication does not apply an impact to teeth, as previously stated by Stegaroiu *et al.* (1999). Also, the mandible closure velocity is reduced during the final phase of mastication (Bates, Stafford and Harrison, 1976). Therefore, the incidence of impact forces during dental loading would only happen at a traumatic shock or at an unadverted mastication of a solid body (Davis, Rimrott and Zarb, 1988). The studies of Gracis *et al.* (1991) that concluded that acrylic resins may diminish load peaks and delay the load transfer, have considered a impact loads, which can be the reason of the differences between the results obtained from in vivo studies (Bassit, Lindstrom and Rangert, 2002; Duyck *et al.,* 2000; Hobkirk and Psarros, 1992).

The accomplishment of clinical tests with prostheses in low rigidity, through methodologies that allow the evaluation of the transmission of forces, as it happens with transducers or strain gauges, would bring relevant complementary information to this study.

5. Conclusions

The material of the prosthetic crown, which determines the stiffness degree, did not interfere in the stress patterns around of implants after the loading application over single implant prosthesis.

The use of prosthetic materials with lower stiffness was not capable to diminish or to delay the forces transmitted to single implants.

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