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ISABELLA DA SILVA VIEIRA MARQUES

"A INFLUÊNCIA DA CICLAGEM MECÂNICA SOBRE O DESAJUSTE MARGINAL, TENSÃO E TORQUE DE AFROUXAMENTO NO SISTEMA IMPLANTOSSUPORTADO"

"INFLUENCE OF MECHANICAL CYCLING ON MARGINAL MISFIT, STRAIN AND LOOSENING TORQUE IN THE IMPLANT-SUPPORTED SYSTEM"

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UNIVERSIDADE ESTADUAL DE CAMPINAS
FACULDADE DE ODONTOLOGIA DE PIRACICABA

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Orientador: Prof. Dr. Marcelo Ferraz Mesquita

Co-orientadora: Profa. Dra. Jessica Mie Ferreira Koyama Takahashi

"INFLUENCE OF MECHANICAL CYCLING ON MARGINAL MISFIT, STRAIN AND LOOSENING TORQUE IN THE IMPLANT-SUPPORTED SYSTEM"

Dissertação de mestrado apresentada à Faculdade de Odontologia de Piracicaba da Universidade Estadual de Campinas para obtenção do título de Mestra em Clínica Odontológica, na área de Prótese Dental.

Master thesis presented to the Piracicaba Dental School of the University of Campinas to obtain the Master's degree in Clinical Dentistry in the area of Dental Prosthesis.

Este exemplar corresponde à versão final da Dissertação defendida pela aluna, e orientada pelo Prof. Dr. Marcelo Ferraz Mesquita.

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Resumo

O objetivo neste estudo foi avaliar a influência da ciclagem mecânica no torque de afrouxamento de parafusos protéticos, no desajuste marginal e na transmissão de tensões ao sistema prótese-implante. Foram confeccionadas 10 infraestruturas metálicas fundidas em titânio comercialmente puro, simulando próteses parciais fixas (PPF) de três elementos retidas por dois implantes e próteses totais fixas (PTF) retidas por cinco implantes. A partir das infraestruturas enceradas e fundidas, foram confecionados modelos que deram origem aos 3 grupos de avaliação (n = 10): grupo I (pré-fundição), grupo II (adaptado) e grupo III (desajuste simulado), tanto para a PPF quanto para a PTF. Para cada modelo de trabalho (resina) confeccionado, foram avaliados o desajuste marginal e torque de afrouxamento, antes e após a ciclagem mecânica (1 milhão de ciclos mecânicos, 2 Hz, 280 N). Os torques de afrouxamento iniciais dos parafusos protéticos e dos parafusos de *mini abutments* foram mensurados 24 horas após a aplicação do torque (10 Ncm e 20 Ncm, respectivamente) utilizando um torquímetro digital. O teste do parafuso único foi realizado e o desajuste vertical foi quantificado por meio de microscópio óptico. As tensões geradas aos análogos durante a fixação da infraestrutura foram avaliadas em um modelo de gesso utilizando *strain gauges*, sendo cada infraestrutura parafusada ao modelo com um torque padronizado de 10 Ncm. Os resultados obtidos foram submetidos à análise exploratória seguida dos testes estatísticos ANOVA e Tukey (α = 0,05) quando obtidos dados paramétricos e Mann Whitney e Wilcoxon quando encontrados dados não paramétricos. Para todos os grupos avaliados, a ciclagem mecânica não apresentou diferença significativa quando analisados torque de afrouxamento dos parafusos protéticos, desajuste e tensão, porém houve um aumento nos valores médios de torque de afrouxamento dos *mini abutments* nos grupos II e III para a PPF e no grupo III para a PTF. O teste de correlação de Pearson (α = 0,05) indicou correlação positiva entre desajuste e tensão para todas as situações avaliadas e ainda correlação negativa entre torque de afrouxamento e desajuste vertical, para os grupos II e III de ambos designs protéticos. Dentro das condições avaliadas e resultados obtidos, pode-se concluir que a ciclagem mecânica de infraestruturas de próteses múltiplas parafusadas não exerce influência relevante sobre o torque de afrouxamento, desajuste e tensão. Os parafusos protéticos apresentam maior tendência ao afrouxamento em próteses mal adaptadas. Próteses desadaptadas geram maior tensão ao sistema prótese-implante.

Palavras-Chave: Próteses e implantes, Infraestrutura, Titânio, Estresse mecânico, Torque.

The aim of this study was to evaluate the influence of cyclic loading on loosening torque, misfit and strain transmitted to prosthesis-implant system. Ten Commercially pure titanium frameworks were casted simulating fixed partial dentures (FPD) of three elements retained by two implants and mandibular full-arch fixed prostheses (FFP) as bars retained by five implants. From the waxed and casted frameworks, models were fabricated, which resulted in three evaluation groups (n = 10): group I (pre-casting), group II (passive fit) and group III (simulated misfit), for both prosthetic designs. For each fabricated acrylic resin cast, the misfit and loosening torque analyses were assessed before and after mechanical loading (10^6 mechanical cycles, 2 Hz, 280 N). The initial prosthetic and mini abutment screws loosening torque were measured 24 hours after the tightening torque (10 Ncm and 20 Ncm, respectively) using a digital torque meter. The single-screw test was performed and the vertical misfit was quantified using an optical microscope. The strain after fixation was evaluated in a plaster cast using strain gauges analysis with each framework tightened to the cast with 10 Ncm standardized torque. The results were submitted to exploratory analysis followed by statistical tests such as ANOVA and Tukey ($\alpha = 0.05$) when parametric data were found and Mann-Whitney and Wilcoxon when non-parametric data were obtained. For all groups, the cyclic loading had no significant difference when analyzed the prosthetic screw loosening torque, misfit and strain, but there was an increase of the mini abutment loosening torque mean values for FPD of groups II and III and for FFP of group III. The Pearson correlation test ($\alpha = 0.05$) indicated a positive correlation between misfit and strain for all evaluated situations and a negative correlation between loosening torque and vertical misfit for both prostheses designs of groups II and III. Within the evaluated conditions and obtained results, it can be concluded that cyclic loading of screw-retained multiple prosthesis does not exert significant influence on loosening torque, misfit and

strain. The prosthetic screws have a greater tendency to loosening in ill-fitting prosthesis, which induces higher strain values on prosthesis-implant system.

Key Words: Prostheses and implants, Infrastructure, Titanium, Mechanical stress, Torque.

Sumário

Introdução

Os implantes dentários osseointegrados têm menor resiliência do osso alveolar quando comparado com os dentes naturais, que estão ancorados pelo ligamento periodontal (Branemark, 1983). Assim, a relativa precisão entre a restauração implantossuportada e seus intermediários deve ser maior (Wee *et al.,* 1999), pois desajustes mínimos podem gerar altos níveis de tensão na interface osso-implante. Essa tensão excessiva pode resultar em complicações mecânicas e biológicas, tais como reações adversas do tecido peri-implantar, perda de osso marginal, perda de osseointegração, fratura dos componentes do sistema e afrouxamento dos parafusos protéticos (Branemark, 1983; Skalak, 1983; Jemt, 1991; Schwarz, 2000).

O assentamento passivo tem sido considerado um dos requisitos mais importantes para a manutenção da interface osso-implante, mas a relação entre o grau de desajuste e complicações mecânicas e biológicas ainda não está bem estabelecida na literatura (Sahin & Cehreli, 2001; Karl *et al.,* 2004; Taylor, 1998). Branemark, em 1983, considerou um *gap* vertical de 10 µm ou menos entre a infraestrutura e o pilar como sinal de assentamento passivo, enquanto Jemt, em 1991, sugeriu uma *gap* vertical de 150 µm como o limite aceitável para o desajuste da infraestrutura protética. Outros autores (Karl et al., 2004) sugerem a utilização do termo "assentamento passivo" quando uma infraestrutura não induz nenhum tipo de tensão sobre os componentes e sobre o osso circundante, na ausência de uma carga externa aplicada. No entanto, o valor clinicamente aceitável para a adaptação passiva em restaurações implantossuportadas ainda não está bem estabelecido.

Técnicas convencionais de fundição em monobloco não permitem a obtenção de infraestruturas rígidas com níveis aceitáveis de adaptação e distorções podem ocorrer em qualquer etapa durante a confecção da peça protética (Romero *et al.,* 2000; Mitha *et* *al.,* 2009). A maioria das distorções ocorre devido à alteração volumétrica dos materiais, como gessos, ceras, revestimentos, fundição do metal, material de moldagem e aplicação do revestimento estético (Nakaoka, 2007; Mitha *et al.,* 2009).

O tipo de liga utilizada para a confecção de infraestruturas protéticas é outro fator que pode influenciar na obtenção de ajuste passivo (Kano *et al*., 2007; Torres *et al*., 2007; Torres *et al*., 2011). Cilindros pré-fabricados com a cinta metálica permitem uma melhor adaptação marginal das coroas individuais em metais preciosos colocadas sobre implantes. A influência do processo de fundição na adaptação destas restaurações é mínima (Byrne *et al.,* 1998; Bhering *et al.,* 2012). No entanto, esta técnica não pode ser usada para todos os metais não-preciosos. Fundições em titânio comercialmente puro, por exemplo, devem ser realizadas utilizando combinação com cilindros totalmente calcináveis (Koke *et al*., 2004).

Nos últimos 50 anos, houve considerável evolução nos equipamentos e técnicas disponíveis para a fundição de titânio, embora apenas recentemente os profissionais tenham conseguido fundir este metal com qualidade satisfatória. O titânio e suas ligas proporcionam resistência, rigidez e ductilidade semelhantes às de outras ligas odontológicas. A baixa densidade de titânio faz a restauração mais leve e mais confortável para o paciente. Fornece ainda excelente resistência à corrosão e é considerado entre os metais, o mais biocompatível (Roach, 2007).

A adaptação marginal inadequada de próteses parciais parafusadas e o afrouxamento de parafusos protéticos são fatores correlacionados (Binon, 1996). Apesar do reaperto do parafuso ser prática comum nos acompanhamentos clínicos da prótese, sua realização pode provocar redução progressiva nos valores de torque de remoção e consequente instabilidade do sistema (Weiss *et al.*, 2000; Byrne *et al*., 2006). Assim, é importante ressaltar que a ocorrência de afrouxamento do parafuso protético pode sinalizar a futura falha de outros componentes (Byrne *et al.*, 1998; al-Turki *et al.*, 2002) e o desenvolvimento de complicações biológicas, além de sugerir a presença de sobrecarga sobre a reabilitação (Weiss *et al.*, 2000; Byrne *et al*., 2006).

A falta de adaptação passiva e a micromovimentação dos componentes protéticos transmitem tensões indesejáveis aos implantes e ao tecido ósseo adjacente (Gratton *et al*., 2001) que podem comprometer o sucesso da osseointegração. Mesmo diante da inserção dos parafusos protéticos, a correção dos desajustes e fechamento de fendas não é totalmente possível, sendo a forçada adaptação da peça protética um dos principais responsáveis pela geração de tensão excessiva (Cantwell & Hobkirk, 2004).

O sucesso do tratamento reabilitador com implantes é também associado à carga aplicada ao sistema, pois a localização e magnitude das forças oclusais afetam a quantidade e qualidade da tensão transmitida para os componentes do sistema osso-implante-prótese (Sahin *et al*., 2002; Sevimay *et al*., 2005). Além disso, a ciclagem mecânica em próteses implantossuportadas podem causar micromovimentos e fadiga dos componentes de união em próteses parafusadas. Baixos valores de pré-carga apresentam micromovimentos significativamente mais elevados na interface pilar-implante (Gratton *et al.,* 2001). Com isso, fatores relacionados à transmissão dos esforços oclusais para o osso, distribuição das tensões sobre os componentes protéticos e adaptação passiva da infraestrutura da prótese sobre o implante têm sido considerados essenciais para a longevidade do sistema implantossuportado (Jemt *et al*., 1999; de Torres *et al*., 2007).

Devido aos efeitos nocivos de desajustes e tensões excessivas sobre os componentes da reabilitação e os tecidos de suporte, a análise destas situações clínicas é necessária. Quanto aos métodos de análise, podem ser utilizados os métodos de elementos finitos (Sevimay *et al.* 2005), análise fotoelástica (Millington & Leung, 1995; Torres *et al.,* 2011) e extensometria (Karl *et al.,* 2009; Hegde *et al.,* 2009; Abduo & Lyons, 2012). O método da extensometria *(strain gauge)* tem sido utilizado para analisar o comportamento biomecânico em várias situações de confecção de próteses implantossuportadas. Muitos estudos (Karl *et al.,* 2009; Hegde *et al.,* 2009; Abduo *et al.,* 2011; Abduo & Lyons, 2012) têm enfatizado o desajuste presente na interface implante-infraestrutura de próteses implantossuportadas não-passivas a fim de buscar uma relação entre tensão e desajuste. *Strain gauge* é um sensor elétrico que quantifica a deformação da superfície, considerado um método de medida indireta, que analisa um efeito físico. O seu princípio de funcionamento baseia-se na variação da resistência elétrica transformado em níveis de deformação (Nishioka *et al.,* 2010).

Desta maneira, mediante a redução da força de união entre os componentes do sistema, a estabilidade da união pode ser comprometida e falhar clinicamente (McGlumphy *et al*., 1998). A presença de cargas mastigatórias nocivas e desajustes marginais excessivos podem promover micromovimentações entre os componentes protéticos, causando deslizamento das roscas de união, podendo ocasionar o desrosqueamento do parafuso protético (Burguete *et al*., 1994), e soltura da prótese. A possibilidade destas situações e as consequências clínicas das mesmas evidenciam a necessidade de avaliação dos efeitos da simulação mastigatória em infraestruturas múltiplas de próteses implantossuportadas sobre a adaptação passiva, força de torque de afrouxamento e geração de tensões sobre o sistema osso-implante-prótese.

Levando em consideração a carência na literatura de estudos que avaliem todos esses aspectos, o objetivo do presente estudo foi avaliar a influência da ciclagem mecânica no torque de afrouxamento de parafusos protéticos, no desajuste marginal de próteses múltiplas implantossuportadas e na transmissão de tensão ao sistema prótese-implante.

*Capítulo 1**

Implant-supported frameworks biomechanics under simulation of masticatory conditions

Cyclic loading on casted frameworks

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Implant-supported frameworks biomechanics under simulation of masticatory conditions

Abstract

This study aimed to evaluate the influence of cyclic loading in one-piece casted implant-supported frameworks on screw loosening, vertical misfit and strain transmitted to the implants. Ten one-piece casted 3-element implant-supported frameworks (FPD) and ten frameworks simulating bars for mandibular full-arch fixed prostheses (FFP) were casted in commercially pure titanium (cp Ti). The initial loosening torque for the prosthetic (PS) and mini abutment screws (MAS) were measured 24 hours after the tightening torque (10 Ncm and 20 Ncm, respectively) using a digital torque meter. The single-screw test was performed and the vertical misfit was quantified using an optical microscope. The strain transmitted to implants was measured using the strain gauge analysis with each framework tightened to the plaster cast with 10 Ncm standardized torque. The analyses were evaluated before and after cyclic loading. The results were statistically analyzed (ANOVA; Tukey test ($\alpha = 0.05$); Pearson correlation test). There were no statistically differences in any of the three variables before and after cyclic loading. There was correlation between the vertical misfit and strain before ($r = 0.86$; $p = 0.0014$) and after mechanical cycling ($r = 0.71$; $p = 0.0220$). Based on the results, cyclic loading did not influence screw loosening, vertical misfit and strain. For FPD, as misfit increased, strain also increased, but no correlation was found between vertical misfit and loosening torque for both designs.

Key words: dental implants, loosening torque, vertical misfit, strain gauge, distortion, titanium.

INTRODUCTION

Osseointegrated dental implants have lower resilience in the alveolar bone when compared to natural teeth, which are anchored by the periodontal ligament [1]. One factor that is expected to influence the longevity of implant-supported prosthesis is the lack of passive fit of the prosthesis framework. This situation may cause mechanical and biologic complications such as adverse reactions of peri-implant tissue, marginal bone loss, mechanical failure of the prosthesis, loss of osseointegration, fracture of system components and loosening of the prosthetic screws [1-4].

Passive fit is assumed to be one of the most significant requirements for the maintenance of the bone-implant interface, but the relationship between the degree of fit and mechanical and biological complications is yet to be established [5-7]. Branemark [1] has considered a vertical gap of 10 µm or less between the framework and the abutment as sign of passive fit, while Jemt [4] has suggested a vertical gap of 150 µm as the acceptable limit for framework misfit. Other authors [6] suggest the use of the term “passive fit” when a framework induces absolute zero strain on the supporting implant components and the surrounding bone in the absence of an applied external load. Nonetheless, the clinically acceptable amount of superstructure passivity has yet to be determined for implant-supported restorations. The only method for determining the actual amount of superstructure passivity would be an *in vivo* analysis [5, 6]. However, this analysis is difficult to be performed, which makes *in vitro* studies very important to biomechanics understanding.

Conventional dental laboratory techniques do not fabricate a rigid framework assembly with an acceptable degree of fit accuracy. This distortion is mostly due to the inconsistency of volumetric and linear expansion of the materials used, such as impression material, waxes, gypsum products, investments and casting metal. Potential distortion can be generated at any step of the fabrication process, from the impression making to finishing and polishing [5, 8-10]. The prosthetic frameworks were initially made of precious alloys. However, due to the high cost of these alloys, the use of alternatives was a good solution for frameworks casting. One alternative material is titanium that exhibits excellent biocompatibility, great corrosion resistance, desirable physical properties, low cost, low density, and low thermal conductibility [11], nonetheless, it is a difficult metal to cast and weld [12].

Therefore, cyclic occlusal loading of the implant-prosthesis assembly induces micromotion of the joint components, which could wear down the microscopically rough areas of the contacting surfaces. Among the factors that contribute to screw instability are misfit of the prosthesis, insufficient tightening force, screw settling, biomechanical overload and differences in screw material or design [13-16].

The aim of this *in vitro* study is to evaluate the influence of cyclic loading on screw loosening, misfit and strain transmitted to implants in one-piece casted implant-supported frameworks with different designs and verify if there is correlation between the three of them before and after cyclic loading.

MATERIAL AND METHODS

For this study two stainless steel master models were made and from them obtained 10 waxed frameworks for each prosthesis design. For that, two or five mini abutment analogs (SIN - Sistema de Implante Nacional S/A, São Paulo, SP, Brazil) were installed over the models (for Fixed Partial Dentures - FPD - 3 elements or Full-arch Fixed Prostheses - FFP - 5 elements) using transversal screws. The FPD model aimed to represent the placement of two lower implants, a first premolar (A) and first molar (B), with 18mm interimplant distance. For the FFP, the mini abutment analogs (named A, B, C, D, and E) simulated a clinical situation of five implants placed between the mental foramina and arranged in the arch with 10 mm interimplant space. Over the mini abutment analogs calcinable cylinders were adapted for waxing of both framework designs (FPD and FFP), by progressive waxing technique. Afterwards, the waxed patterns were duplicated using a silicone matrix (Figure 1) (Flexitime Easy Putty, Correct Flow, Heraeus-Kulzer, Hanau, HE, Germany) and reproduced in low shrinkage acrylic resin

(Duralay II, Reliance Dental Mfg. Co, Worth, IL, United States). After all frameworks were waxed, plaster casts were fabricated with the assistance of a parallelometer. For this, modified analogs were used and embedded in gypsum in about 10 mm length, allowing the strain analysis. These replicas were made of titanium with 20 mm length and hollow interior to allow the elastic deformation to be detected by electrical resistances - strain gauges.

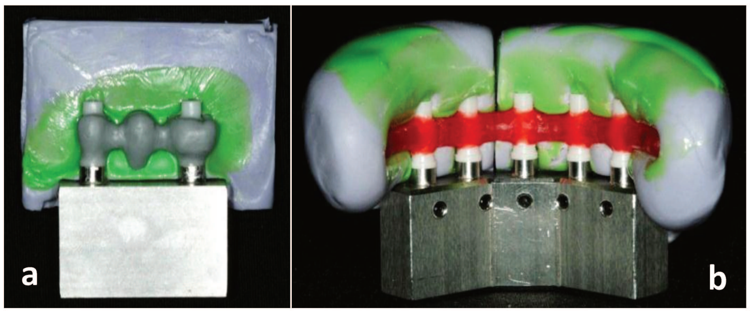

Figure 1. Waxed pattern duplication using a silicone matrix: a) FPD; b) FFP.

Acrylic resin casts were also fabricated with the assistance of a parallelometer, but external hexagon conventional analogs and mini abutments were used to manufacture them (Figure 2). These models were made to perform mechanical cycles, loosening torque and misfit analyses. The waxed frameworks were casted in cp Ti (Tritan, Dentaurum, Ispringen, BW, Germany). After cyclic loading, the frameworks were repositioned from the acrylic resin casts to the plaster casts to measure the final analyses (loosening torque, misfit and strain).

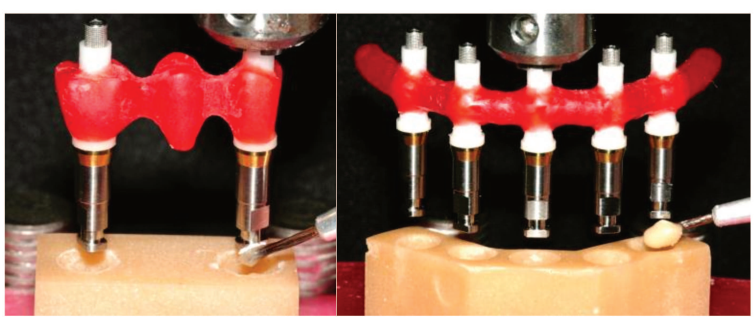

Figure 2. Acrylic resin casts obtainment on a parallelometer.

Loosening torque analysis

The loosening torque analysis was performed on the acrylic resin casts. The procedures for tightening and loosening torque were performed by one researcher using a digital torque meter with accuracy of 0.1 Ncm (TQ-8800 Torque Meter, Lutron, Taipei, Taiwan) coupled to a device allowing its vertical positioning, so lateral stresses would not be induced. The tightening torques of 20 Ncm and 10 Ncm were applied for mini abutments and prosthetic screws, respectively, and the loosening torque was evaluated for both. For FPD the tightening sequence for the FPD was first pillar A (premolar region) and then the pillar B (molar region). Therefore, for FFP the sequence was first the components of the ends (A and E), then the intermediate components (B and D) and finally the central component (C) [17-19].

Assessments of loosening torque was conducted in two stages, 24 hours after the tightening of the components (initial loosening torque) and after completion of mechanical cycling (final loosening torque). These measurements were conducted in order to verify the actual amount of force absorbed by the screws after the initial torque while measuring after mechanical cycling aimed to evaluate the effect of compressive load (masticatory simulation) and the misfit in maintaining the strength of torque of these same screws.

Vertical misfit analysis

An optic microscope (VMM-100-BT; Walter UHL, Asslar, HE, Germany) was used to measure the vertical misfit at 120X magnification. The measurements were performed just for one and calibrated examiner (intraclass correlation coefficient = 0.999). Both resin and plaster casts were used for these analyses, the resin casts to allow comparison with loosening torque and the plaster casts for comparison with strain analysis. For misfit analysis the specimens were tested by tightening the screws with 10 Ncm through a single-screw test [13, 20, 21] to simulate the maximum misfit of the framework. For FPD framework, the reading was performed on buccal and lingual sides in the cylinder

A (when B was tightened) and on buccal and lingual sides in cylinder B (when A was tightened). And for FFP frameworks, the reading was performed on buccal and lingual sides in the cylinders C and E (when A was tightened) and on buccal and lingual sides in the cylinders C and A (when E was tightened). To standardize the position for the readings, a silicon matrix was made so that all readings for all cylinders would be in the same position, in the middle point of each abutment, before and after cyclic loading. After all initial misfit readings procedures, the initial strain analyses were made.

Strain analysis

The strain analysis was performed on the plaster casts (Durone, Dentsply, Petrópolis, RJ, Brazil) for each framework before and after mechanical cycling. The measurement was performed by two strain gauges (PA-06-060BG-350L, Sensor Excel Engineering; Embu, SP, Brazil) for FPD and by five strain gauges for FFP, positioned directly on the internal face of each modified analog (Figure 3). The arrangement of strain gauges used was a quarter Wheatstone bridge, allowing the capture of deformation by electrical signals on a computer-controlling device (ADS2000; Lynx Tecnologia Eletrônica Ltda, São Paulo, SP, Brazil) processed by specific software (AqDados 7; Lynx Tecnologia Eletrônica Ltda, São Paulo, SP, Brazil). The framework was adapted over the modified analog of mini abutment and fixed according to manufacturer instructions with a torque of 10 Ncm. The sequence of tightening was performed by tightening the screw in the premolar position - A and then the molar position - B (FPD). For the FFP the sequence was A, E, B, D, C, according to the position of the pillars. At the end were obtained strain averages for each framework analyzed. The analyses were performed during ten minutes, with an intermediary five minutes period selected after strain stabilization for statistical analysis purpose.

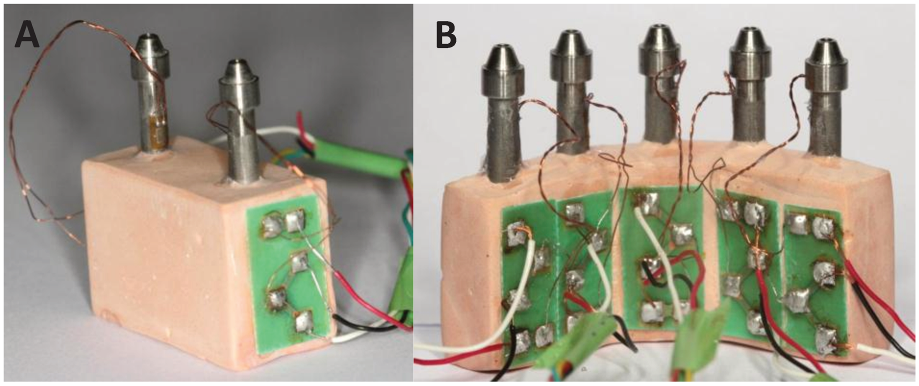

Figure 3. Strain gauges bonded on analogs surface with a quarter Wheatstone bridge. A) FPD; B) FFP.

Cyclic loading

After initial analyses, cyclic loading was applied obliquely on both partial and complete fixed implant-supported frameworks that were immersed in artificial saliva (1.5 mmol/L Ca, 3.0 mmol/L P, 20.0 mmol/L $NaHCO_3$; pH 7.0) [22] and submitted to mechanical fatigue testing. The frameworks were subjected to 10^6 mechanical cycles, with 2 Hz frequency, at 30° inclination (ISO 14801 - 2007) [23] in Fatigue Mechanics Simulator (ER 11000 Plus, ERIOS, São Paulo, SP, Brazil). The parameters were estimated supposing that an individual performs three episodes of chewing per day, each lasting 15 minutes, and at a chewing rate of 60 cycles per minute (1 Hz). This is equivalent to 2.700 chewing cycles per day or roughly 1 million cycles per year [24]. The simulation was performed with mechanical fatigue pressure of 6 bars corresponding to about 280 N for compressive load [25] applied to the distal region of framework.

All data obtained were submitted to ANOVA and Tukey's HSD test (α = 0.05). The correlation analyses were performed using Pearson correlation test (α = 0.05). All statistical tests were performed with SAS (version 9.1, The SAS Institute, Cary, NC).

RESULTS

Table I shows the mean and standard deviation values of the data obtained in this study.

Cyclic loading had no influence on marginal misfit, strain and loosening torque according to the ANOVA, for all analyzed casts and designs.

The comparison between the marginal misfit obtained in different casts was done and no difference was observed (left to right – Table I: p = 0.1235; p = 0.2830; p = 0.2498; p = 0.8107), what allows us to infer that the behavior of all variables can be analyzed together.

There was no correlation between misfit and loosening torque for both designs and screw types (r = -0.28256; p = 0.2274 – FPD; r = -0.41110, p = 0.0718 – FFP, for prosthetic screws) (r = 0.07709; p = 0.7467 – FPD; r = 0.13834; p = 0.5608 - FFP, for mini abutment screws).

The correlation tests performed between strain and marginal misfit values for FPD design showed difference statistically significant at initial and final times, (r = 0.86; p = 0.0014) (r = 0.71; p = 0.0220), which did not occur for FFP design (r = -0.09; p = 0.804) (r = -0.51; p = 0.133).

Table I. Mean and standard deviation values of the data obtained in the study.

	FPD		*p*-value	FFP		*p*-value
	Initial	Final		Initial	Final	
Strain - με	326.76 (161.73) [a]	421.99 (157.8) [a]	0.1992	224.92 (115.72) [a]	207.54 (86.33) [a]	0.7132
Loosening torque (PS) - Ncm	6.14 (1.07) [a]	5.45 (0.79) [a]	0.6752	4.08 (1.68) [a]	3.94 (1.27) [a]	0.9969
Loosening torque (MAS) - Ncm	16.83 (1.76) [a]	15.98 (1.62) [a]	0.5192	16.98 (1.58) [a]	16.93 (1.32) [a]	0.9998
Misfit (plaster cast) - µm	148.63 (53.03) [Aa]	141.42 (48) [Aa]	0.9918	311.48 (70.09) [Aa]	280.27 (60.53) [Aa]	0.8515
Misfit (resin cast) - µm	205.95 (65.08) [Aa]	187.20 (58.01) [Aa]	0.8789	384.78 (111.03) [Aa]	314.72 (95.58) [Aa]	0.2865
***p*-value**	0.1235	0.2830		0.2498	0.8107	

Means followed by different letters (capital, column; minor, line) differ statistically according to ANOVA/Tukey tests (α = 0.05).

DISCUSSION

In accordance with previous investigations [6, 26], none of the assessed frameworks exhibited a passive fit. Our results indicate that there is a considerable variation between the frameworks, as shown by the large scatter of the results in Table I. The present study suggests the inaccuracies that can occur in one-piece conventional casting procedures and that the vertical misfit applied in this study is a reasonable representation of what can occur clinically.

It remains difficult to determine the prognosis of an implant restoration in case of the presence of misfit based on an *in vitro* study. This approach is also impractical to provide an absolute guideline of what constitutes passive fit. However, this kind of study is relevant to provide a biomechanics understanding by dimensional measurements of fit, otherwise impossible by means of clinical trials.

The marginal misfit values presented no difference after cyclic loading for both prostheses designs. It could be because the tridimensional distortion that occurs after casting never follows a pattern, and it probably contributed to data variation. These results are in accordance with other studies that demonstrated an insufficient cervical fit and a great difficulty in casting procedures with titanium [10, 12, 20]. A 3-D study [10] was performed to evaluate the casting distortion of implant-supported frameworks and found that as-cast titanium frameworks are inaccurate and imprecise when judged against the 150 μm requirement for passivity of fit [4]. The same was observed in this study, in which misfit values near and above 150 μm were found in both FPD and FFP frameworks.

Regardless of the design of the prosthesis and the type of screw, the values of loosening torque were not significant different after cyclic loading. One explanation for this can be the fact that there are two factors that may influence these values, one is the presence of saliva and the other is the cyclic load. Studies [27, 28] affirm that the presence of saliva contributes with the maintenance of the preload, what suggests higher detorque values for the final analyses. This phenomenon can be explained by the formation of a corrosion layer between metallic surfaces at contact in the implant-abutment joint during immersion in fluoridated artificial saliva solution [28]. Even though the saliva composition used in this study did not contain fluoride, the actual influence of other components in saliva is not yet established. On the other hand, the cyclic loading on implant-supported prostheses may result in micromovements and fatigue of the metal in screwed prostheses that seem stable. In addition, screw joints on implants that present low preload values exhibit significant higher micromovements at the abutment/implant interface [29]. The results of this study are in agreement with other studies in the literature [15, 30] that also did not find influence of mechanical cycling in loosening torque. Nonetheless, Farina *et al*., 2012 [13] found that cyclic loading influenced the joint stability of multiple-unit prosthesis. This contradiction might be attributed to the different comparison made on its study between a target of 0.5 x 10^6 cycles and twice as many cycles (10^6), simulating six months and one year, whereas in the present study the comparison was between an unloaded situation and one year simulation. In addition, another difference might be explained by the fact that Farina *et al*., 2012 [13] performed axial loading and in the present study the specimens were submitted to oblique loading.

A standardized screw tightening sequence was used (A-E-B-D-C) in the present study. This sequence was used based on previous studies [17-19] that showed that the stress generated on the implants did not change when the screw tightening sequence was altered; however, the initial tightening of the end screw, as done in this study, could distribute more evenly the stress on the distal implants.

All frameworks evaluated in this study generated strain around the implants. The values obtained for the strain analysis also revealed significant variation. Since the strain gauge has the limitation of recording the strain on the area where it is bonded, the recorded variation could be due to the inevitable minor difference in strain gauge location [16]. Another source of variation is the inevitable disparity in the manual tightening of the framework screws [16].

After mechanical cycling, the frameworks returned to the same plaster cast that was performed the initial strain analysis, to perform the final strain analysis. Other cast was used to avoid the influence of the acrylic resin aging during the masticatory simulation. Some studies [26, 31] believe that the resin can absorb a quantity of strain, changing the real values. Because of that, this study used plaster casts to perform this analysis.

This strain methodology is a novel approach used to evaluate the behavior of the strain found at the pillars. The mini abutment analogs were modified with hollow interior allowing the strains analysis. The replicas made of titanium are longer than the traditional to allow the elastic deformation that is detected by the resistance change in microvolt (μV) across the strain gauges, which can be converted subsequently to microstrain (με). With the information before and after cyclic loading, it is possible to compare the values obtained and analyze if masticatory simulation influences the strain between the frameworks. The results showed that the differences were not statistically significant at initial and final times, for either design. Some studies [32, 33] evaluated the strain of the frameworks under loading conditions. For this, the load is applied over the framework during the strain analysis, what differs from this study, that the frameworks were subjected to cyclic loading and after that, the strain was recorded with the tightening of the screws. There is a lack in the literature regarding the influence of cyclic loading on the tension, which gives the relevance for this study.

Regarding the relationship between vertical misfit and strain, it was found that the magnitude of framework strain rose markedly with increasing levels of misfit on FPD. The same relation was observed in others studies [26, 31] that showed a nearly linear relationship between strain and vertical misfit. This means that as misfit increases strain also increases.

Moreover when the relationship between vertical misfit and loosening torque was evaluated, this study found that it was not significant. Previous studies [34, 35] affirm that fitted prostheses are more suitable to present higher detorque values than prostheses with misfit, but this relation could not be linear.

CONCLUSION

According to the results and within the limitations of the present *in vitro* study, the following can be concluded:

Cyclic loading has no influence on frameworks' screw loosening, vertical misfit and strain.

FPD framework strain is affected by the vertical misfit. With increased misfit, the magnitude of peri-implant strain also increased.

The screw loosening is not influenced by the misfit.

ACKNOWLEDGMENTS

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Capítulo 2*

Biomechanics in screw-retained implant titanium frameworks at two levels of fit

Running head: Biomechanics of casted frameworks

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Abstract

Objective: The aim of this study was to evaluate the influence of cyclic loading in one-piece casted implant-supported frameworks on screw loosening, vertical misfit and strain transmitted to the implants at two levels of fit. ***Material and methods:*** Ten one-piece casted 3-element implant-supported frameworks were casted in commercially pure titanium (cp Ti). After that, two situations were created: passive fit (P) and non-passive fit (NP). A digital torque meter was used to measure the initial loosening torque for the prosthetic (PS) and mini abutment screws (MAS) 24 hours after tightening (10 Ncm and 20 Ncm, respectively). An optical microscope was used to quantify the vertical misfit performing the single-screw test. The strain gauge analyses were performed by the tightening of frameworks to the plaster casts with 10 Ncm standardized torque. The analyses were evaluated before and after cyclic loading. Data were analyzed by two-way ANOVA, Tukey's test and Pearson correlation test ($\alpha = 0.05$). ***Results:*** Cyclic loading did not influence any of the three variables, except the MAS loosening torque. There was no significant difference between the P and NP groups for strain and PS loosening torque. There was correlation between misfit and strain ($r = 0.63619$, $p < 0.001$) and also between misfit and PS loosening torque ($r = -0.31272$, $p = 0.0495$). ***Conclusions:*** Mechanical cycling does not influence the frameworks' screw loosening (PS), vertical misfit and strain. Ill-fitting prostheses overload the prosthetic screws and induce higher strain to the pillars.
Key words: cyclic loading, dental implants, misfit, strain gauge, torque.

Introduction

The proper fit between the implant and the prosthetic component is the primary desire during the manufacturing of implant-supported prostheses [1]. Most authors agree with the need of passive adaptation between prosthetic framework and implants [2]. Factors such as transmission of occlusal loads to the bone, stress distribution on the prosthetic components and passive adaptation of the prosthesis framework on the implant have been considered essential for the longevity of implants and prostheses [3-5].

The lack of passivity of implant-supported prosthesis can lead to mechanical and biological complications such as adverse reactions of peri-implant tissue, marginal bone loss, mechanical failure of the prosthesis, loss of osseointegration, fracture of system components and loosening of the prosthetic screws [1, 3, 5, 6]. Some authors have sought to define an acceptable level of fit, suggesting values between 10 [7] and 150 [8] µm as being clinically acceptable.

The non-passive fit and the micromovements of prosthetic components transmit undesirable strains to adjacent bone tissue and the implant [9] that could compromise the success of osseointegration. Even with the insertion of prosthetic screws, correction of misfits and closing of slits are not entirely obtained, being the force to adapt the prosthetic framework a major contributor to the excessive stress transmitted to the bone [10].

The success of rehabilitation treatment with implants is also associated with the load applied to the system, because the location and magnitude of occlusal forces affect the amount and quality of the strain transmitted to the components of bone-implant-prosthesis system [4].

Thus, by reducing the clamping force between the components of the system, the stability of the joint can be damaged and fail clinically [11]. The presence of harmful masticatory loads and excessive marginal misfit may promote micromovements between the prosthetic components causing sliding of the threads, which may result in the unscrewing of the prosthetic screw [12], and loosening of the prosthesis.

The possibility of these situations and their consequences highlight the need to evaluate multiple implant-supported frameworks on passive fit, loosening torque force and strain generation on the bone-implant-prosthesis system.

The aim of this study was to evaluate the influence of cyclic loading in one-piece casted implant-supported frameworks on screw loosening, vertical misfit and strain transmitted to the implants at two levels of fit.

Material and Methods

A stainless steel master model was fabricated and two mini abutment analogs (SIN - Sistema de Implante Nacional S/A, São Paulo, SP, Brazil) were fixed to it with transversal screws. The model aimed to represent the placement of two lower implants, a first premolar (A) and first molar (B), with 18 mm interimplant distance (Figure 1). Over the mini abutment analogs two calcinable cylinders were adapted for waxing the framework through the progressive waxing technique using gray wax (Yeti Thowax, Dental Produkte GmbH, Engen, BW, Germany). The waxed pattern was reproduced in low shrinkage acrylic resin (Duralay II, Reliance Dental Mfg. Co, Worth, IL, United States) using a silicone matrix (Flexitime Easy Putty, Correct Flow, Heraeus-Kulzer, Hanau, Germany). After all ten frameworks were waxed and casted in cp Ti (Tritan, Dentaurum, Inspringen, Germany), plaster and resin casts were fabricated with the assistance of a parallelometer at two situations, passive fit (P) and non-passive fit (NP) (Figure 2). For passive situation, the models were created without interposition between the framework and the mini abutments, featuring a juxtaposition of components (Figure 2A). For non-passive situation the models were made with the assistance of a perforated plate to simulate a misfit of 150 µm at molar region (Figure 2B). The created misfit was standardized in the model prior to the frameworks casting. There were two different models, the acrylic resin and the plaster casts, each designed for each type of analysis. For the manufacturing of the plaster casts, mini abutment analogs were modified allowing the strain analysis. The analogs were made of titanium with 20 mm length stem, presenting a hollow interior to enable its elastic deformation, which is detected by the electric resistances - strain gauges. The acrylic resin casts were manufactured with external hexagon analogs and mini abutments. These models were idealized for the mechanical cycles and for loosening torque and misfit analyses. After cyclic loading, the loosening torque and misfit final analyses were performed on resin casts and then the frameworks returned to the plaster casts to measure the strain final analysis.

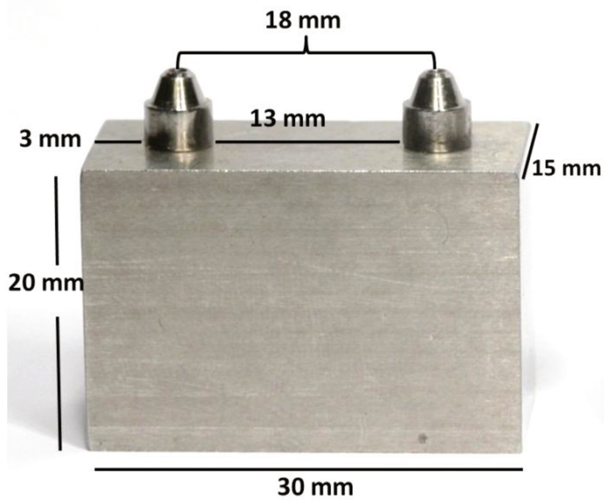

Figure 1. Stainless steel matrix.

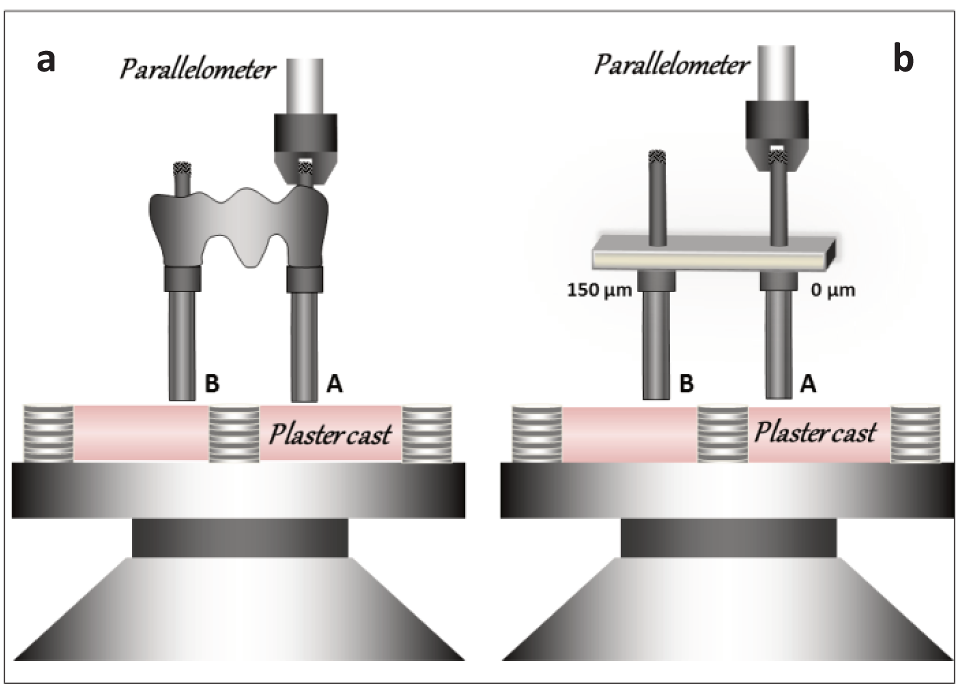

Figure 2. Schematic illustration of casts obtainment: a) Passive fit; b) Non-passive fit.

Loosening torque analysis

The loosening torque was evaluated using a digital torque meter with a 0.1 Ncm precision (TQ8800; Lutron, Taipei, Taiwan), coupled to a device allowing its vertical positioning to avoid lateral stress (Figure 3). Mini abutments (MAS) and prosthetic screws (PS) were used for this analysis. New screws were used for the initial analyses for both groups. One researcher performed the tightening and loosening torque. At first, the tightening torques of 20 Ncm and 10 Ncm were applied for mini abutments and prosthetic screws, respectively, according the manufacturer's instructions. After 24 hours, the loosening torque was quantified as initial time. The sequence for the procedure was first pillar A (premolar region) and then the pillar B (molar region), for both groups, P and NP. The final analysis followed the same parameters and was performed soon after the mechanical cycling.

Figure 3. Digital torque meter coupled to the device.

Vertical misfit analysis

The measurements of vertical misfit were performed using an optic microscope (VMM-100-BT; Walter UHL, Asslar, HE, Germany) at 120X magnification by a calibrated examiner (intraclass correlation coefficient = 0.999). Both resin and plaster casts were evaluated. The resin casts to allow the comparison with loosening torque, and the plaster casts the comparison with strain analysis. The technique used to measure the misfit was based on the single-screw test protocol [13-15], by tightening the prosthetic screws with 10 Ncm. Vertical misfits between the mini abutment platform and the inferior border of the denture framework were then measured, considering the buccal and lingual faces of the pillar A, when pillar B was tightened and buccal and lingual faces of the pillar B, when pillar A was tightened. The single-screw test simulates the maximum misfit in the framework. The position of the readings was standardized through a matrix so that all readings for all cylinders would be in the same position, in the middle point of each abutment, before and after cyclic loading. After all initial misfit readings procedures, the initial strain analyses were made.

Strain analysis

For this analysis, the plaster casts (Durone, Dentsply, Petrópolis, RJ, Brazil) were used for each framework before and after cyclic loading, in both groups, P and NP. Two strain gauges (PA-06-060BG-350L, Sensor Excel Engineering; Embu, SP, Brazil) were positioned directly on the internal side of the base of each modified analog to perform the analyses. A quarter of Wheatstone bridge arrangement was used to allow the capture of the real values of deformation by electrical signals on a computer-controlling device (ADS2000; Lynx Tecnologia Eletrônica Ltda, São Paulo, SP, Brazil), processed by an specific software (AqDados 7; Lynx Tecnologia Eletrônica Ltda, São Paulo, SP, Brazil). The framework was fixed on the modified analogs of mini abutment with a torque of 10 Ncm. The tightening was performed at first in a pillar A and then the pillar B. At the end, strain averages were obtained for each framework analyzed. The analyses were performed during ten minutes, with an intermediary five minutes period selected after strain stabilization for statistical analysis purpose.

Cyclic loading

An oblique cyclic loading was applied on the partial fixed implant-supported frameworks immersed in artificial saliva (1.5 mmol/L Ca, 3.0 mmol/L P, 20.0 mmol/L $NaHCO_3$; pH 7.0) [16] for mechanical fatigue testing after all initial analyses. The frameworks were submitted to 10^6 mechanical cycles, with 2 Hz frequency, at 30° inclination (ISO 14801 - 2007) in Fatigue Mechanics Simulator (ER 11000 Plus, ERIOS, São Paulo, SP, Brazil). The parameters were estimated supposing that an individual performs three episodes of chewing per day, each lasting 15 minutes, and at a chewing rate of 60 cycles per minute (1 Hz). This is equivalent to 2.700 chewing cycles per day and roughly 1 million cycles per year [17]. The pressure of 6 bars applied to the distal region of framework was used to simulate about 280 N of compressive load [18].

All data obtained were submitted to two way ANOVA and Tukey's HSD test ($\alpha = 0.05$). The correlation analyses were performed using the Pearson correlation test ($\alpha = 0.05$). All statistical tests were performed with SAS (version 9.1, The SAS Institute, Cary, NC).

Results

Table I and Table II show a visualization of the mean and standard deviation values of misfit, strain and loosening torque obtained in this study, and Table III shows the correlation between them.

The statistical analysis of misfit values (plaster and resin) did not show significant difference before and after cyclic loading for both situations. But the misfit values were statistically different between passive and non-passive groups at initial and final times (Tables I and II).

Regarding the strain, no difference was observed between the initial and final strain values for both passive and non-passive groups ($p = 0.9380$; $p = 0.9997$, respectively).

Table I. Mean and standard deviation values of misfit (plaster) and strain.

Groups	Misfit - µm		p-value	Strain - µε		p-value
	Initial	Final		Initial	Final	
Passive	76.35 (36.84) Ba	72.08 (34.92) Ba	0.9949	260.28 (184.71) Aa	304.40 (210.56) Aa	0.9380
Non-Passive	209.08 (35.23) Aa	206.28 (49.07) Aa	0.9985	398.10 (136.95) Aa	390.69 (138.67) Aa	0.9997
p-value	<0.0001	<0.0001		0.2871	0.6731	

Means followed by different letters (capital, column; minor, line) differ statistically according to ANOVA/Tukey tests (α = 0.05).

Table II shows the means of loosening torque (PS and MAS) obtained before and after the mechanical cycling. The statistical analysis revealed significant difference before and after cyclic loading for MAS in both groups and between groups at initial time.

Table II. Mean and standard deviation values of misfit (resin) and loosening torque.

Groups	Misfit - µm		p-value	Loosening Torque (MAS) - Ncm		p-value	Loosening Torque (PS) - Ncm		p-value
	Initial	Final		Initial	Final		Initial	Final	
Passive	57.08 (23.16) Ba	61.05 (20.40) Ba	0.9891	14.80 (1.86) Bb	18.19 (0.76) Aa	p<0,001	5.98 (1.34) Aa	4.96 (1.20) Aa	0.2723
Non-Passive	217.75 (31.67) Aa	212.60 (35.43) Aa	0.977	16.82 (1.57) Ab	18.84 (1.15) Aa	0.0133	4.57 (1.27) Aa	4.66 (1.16) Aa	0.9982
p-value	<0.0001	<0.0001		0.0138	0.7235		0.0707	0.8965	

Means followed by different letters (capital, column; minor, line) differ statistically according to ANOVA/Tukey tests (α= 0.05).

When the correlation between the vertical misfit and strain was analyzed by Pearsons' correlation test, a positive relationship was found ($r = 0.63619$ and $p < 0.001$) (Table III). The correlation between misfit and loosening torque (MAS) was not statistically significant ($r = 0.30263$; $p = 0.0577$), while for the prosthetic screws this correlation was significant, featuring an inverse relationship ($r = -0.31272$; $p = 0.0495$).

Discussion

Screw loosening has been described as a significant problem in dental implant therapy [19]. The screw may work as a spring as it is stretched by the preload maintained by the friction between the threads. The higher the preload, the higher the screws' resistance to loosening. When applying insertion torque, it elongates causing stress on the stem and threads [20, 21]. Next, it tends to suffer elastic recovery, which creates the force responsible for maintaining the abutment united to the implant [22]. The preload value may be influenced by several factors such as friction coefficient [22], lubrication [23], rate of applied torque [24], biomechanical overload [11], adaptation between the implant and the abutments [25].

In this study, the loosening torque measurements aimed to check the actual amount of force absorbed by the screws after the initial torque and to evaluate the effect of compressive load (masticatory simulation) at two levels of fit in maintaining the strength of torque of the screws.

There was an increase in MAS detorque values after cyclic loading in both fit situations. Some authors [23, 26] stated that the presence of saliva can influence loosening torque values of the screws, contributing to maintain the values of preload. Authors [26] affirm that a corrosion layer can be formed between abutment and implant screws at static contact in the implant-abutment joint during immersion in fluoridated artificial saliva. An electrochemical reaction between implant, abutment and artificial saliva compounds may lead to this corrosive process. Consequently, that reaction layer can be responsible for the increase of detorque values (MAS) recorded on mini abutments after the cyclic loading upon the immersion of the specimens in fluoridated artificial saliva solution [26]. Although the saliva used in this study did not contain fluoride, the real influence of the other components of saliva is not well established yet. Another explanation can be the fact that an intervening humid media eliminates most of internal shear forces produced on threads, considering that all the internal implant portions were filled with artificial saliva [23]. Furthermore, perhaps lubrication of the abutment screw provided greater preload and thus higher stability of the joints, resulting in less screw vibration and micromovements during cyclic testing.

The same did not occur with the PS. The occlusal load primarily affects the prosthetic screw, as it is more externally located as well as being more delicate and less tightened than mini abutment screw. Because of that, the prosthetic screw may be more influenced by these factors. While a factor such as the presence of saliva helps to maintain the preload, another, the mechanical loading, leads to a decrease of the loosening torque values.

Farina *et al.*, 2012 [15] observed that the mechanical cycling influenced the loosening torque values when comparing the duration of loading. A target of 10^6 cycles leaded to a decrease on loosening torque values in comparison with 0.5×10^6 cycles. In addition, this difference in results between the present study and that of Farina *et al.*, 2012 [15] might also be explained by the different loading conditions. Whereas the present study the specimens were submitted to oblique loading, Farina et al., 2012 performed axial loading.

Although there was no significant difference between the loosening torque values (PS) comparing the groups P and NP, the NP group showed lower loosening torque values. According to Spazzin et al., 2009 [13], the non-passive scenario reduces the loosening torque of the prosthetic screws. This decrease in the joint stability may be due to the increase of the static stress present in non-passive dentures [13, 27]. The inverse correlation between vertical misfit and loosening torque, found in the current study, corroborates this statement indicating that as the misfit increases, detorque values slightly decrease.

In this study, the cyclic loading did not influence the marginal misfit values in both fit situations. This finding is in accordance with Bhering *et al*., 2012 [28] that also found no difference between the misfit values before and after cyclic loading for two abutment types in single unit implant-supported prostheses. The cyclic loading was applied through a loading piston. The masticatory simulated forces might have closed the gap between the abutment platform and the prosthetic cylinder. However, as soon as the force was ceased, the misfit would be reestablished.

When data were analyzed for a correlation between misfit and strain, a positive relationship was found, due to the magnitude of framework strain strongly ascending with increased levels of misfit. The same was observed in other studies [29-31] that showed a linear relationship between vertical misfit and strain. Thus, it can be inferred that as misfit increases, the strain transmitted to the bone also increases.

A perfect passive fit of fixed implant-supported prostheses is very difficult to achieve clinically. Conventional laboratory techniques do not allow prostheses to be manufactured without misfit. This distortion is due to linear and volumetric expansion of the materials used in the casting, from the plaster to the casting metal, at any step of the process [32]. Though empiric, clinically acceptable levels of misfit were established between 10 and 150 µm. The 10 µm level was proposed in 1983 by Branemark [7] in order to allow bone maturation and remodeling in response to occlusal loads, however, values these low are very difficult to be achieved. In 1991, Jemt [8] defined passivity as the level that did not cause long-term complications and suggested values less than 150 µm as clinically acceptable. The mean passive fit values found in the present study are among those currently considered as clinically acceptable. With the results, it is possible to realize that even among the models passively constructed, there was a substantial misfit, what may be due to the models material and casting limitations. Because titanium does not admit overcasting procedures, conventional one-piece casting must be performed. By using this technique, with entirely calcinable cylinders, the connection surface of the framework may suffer distortion and present irregularities resulting, which is mostly inevitable. Irregularities such as defects and positive blisters may influence the perfect adaptation between frameworks and abutments and contribute to additional fit problems.

The measurement of misfit can be a prediction of the strain induced at the implant-bone interface [29-31], but it is difficult to obtain *in vivo*, which gives the importance of *in vitro* studies. *In vitro* models that describe the differences in interface strain for differing levels of prosthetic misfit would be useful [33]. All frameworks evaluated in this study generated strain around the implants, even in the passive group. The values obtained for the strain analysis presented a significant variation.

The P and NP groups did not show statistical difference on the strain analyses. Strain analysis presents a high variation and casting procedure may induce a wide range of distortion. Additionally, the technique used in this study concerning the manufacture of the models has a limitation, because the non-passive models were manufactured prior to casting, and thus, did not consider the casting induced distortion. Furthermore, strain gauge methodology has limitations. The gauge foils are highly sensitive to the location, thus tiny differences in the placement might have influenced the results.

Nonetheless, the increase in strain magnitude in non-passive frameworks is not adequate for the bone tissue and may cause deleterious effects on the interface bone-implant [1, 2, 4]. Based on studies in animals, although only values above 3000 με were reported as critical to bone repair, causing fatigue failure [34], clinically non-harmful strain values are yet to be determined.

Within the limitations of this *in vitro* study, it was found that cyclic loading for masticatory simulation does not influence the prosthetic screw loosening, framework vertical misfit and strain, but increases abutment screw loosening torque. Ill-fitting prostheses overload the prosthetic screws and induce higher strain levels to the pillars. Regarding the correlation, the increase of vertical misfit reduces prosthetic screw loosening and increases strain on the system.

Acknowledgments

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Declaration of interest

The authors report no conflicts of interest. The authors are responsible for the content and writing of the paper.

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*Capítulo 3**

Detorque and vertical misfit of titanium frameworks under simulation of masticatory conditions with different misfit levels

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Detorque and vertical misfit of titanium frameworks under simulation of masticatory conditions with different misfit levels

ABSTRACT

Purpose: The aim of this study was to evaluate screw loosening and vertical misfit in one-piece casted multiple implant-supported frameworks at two fit situations upon cyclic loading.

Materials and methods: Ten frameworks were casted in commercially pure titanium (cp Ti) simulating bars for mandibular full-arch fixed prostheses. Two levels of fit were created by fabricating resin casts to obtain experimental groups: passive fit (P); non-passive fit (NP). At initial time, the loosening torque of the prosthetic (PS) and mini abutment (MAS) screws were measured 24 hours after the tightening torque (10 Ncm and 20 Ncm, respectively) using a digital torque meter. The vertical misfit was quantified using an optical microscope by the single-screw test. The analyses were performed before and after cyclic loading. The results were statistically analyzed by Mann-Whitney and Wilcoxon tests (α=0.05) and Pearson correlation test.

Results: Cyclic loading influenced the MAS loosening torque ($p = 0.0011$) in non-passive group. By contrast, prosthetic screw loosening and misfit values were not influenced by mechanical cycling. The P group showed higher values of PS loosening torque at initial time than NP group ($p = 0.03$). There was correlation between misfit and PS loosening torque ($r = -0.49$; $p = 0.0011$), but no correlation was observed between misfit and MAS loosening torque.

Conclusions: One year of masticatory simulation by cyclic loading has no influence on prosthetic and abutment screw loosening, and also on vertical misfit of passive fit frameworks. However, it increases abutment screw loosening torque for non-passive ones. Passive frameworks present higher prosthetic loosening torque. Ill-fitting prosthesis leads to screw loosening with greater facility.

Key words: implant prosthesis, vertical misfit, loosening torque, cyclic loading.

Introduction

An osseointegrated implant can be defined as a body directly connected to living remodeling bone without any intermediate soft tissue component, providing directly transferred loads to the anchoring bone [1]. The lack of periodontal ligament can enable bone resorption when inappropriate stress is applied. Therefore, the passive fit has been suggested to be a prerequisite for successful long-term osseointegration [2]. However, a complete passive fit of multiple one-piece casted implant-supported frameworks has been hard to achieve. These complications may lead to mechanical failures, such as loosening of the abutment or the prosthetic screws, fracture of components in the system, or biologic failures including adverse tissue reactions, such as pain, tenderness, marginal bone loss and loss of osseointegration [3-6].

Although the clinically acceptable degree of marginal misfit is not yet well established in the literature, some authors have sought to define an acceptable level. A 10 µm gap was considered in 1983 by Branemark [1], in order to allow maturation and bone remodeling as response to occlusal loads. In 1991, Jemt [4] defined passivity as the level that did not cause long-term complications and suggested values lower than 150 µm as clinically acceptable. Furthermore, others authors defined passive fit as the absence of strain pattern during framework fixation [7].

Screw loosening is a current problem that affects screwed implant-supported prosthesis [4, 8, 9]. The prosthetic screw instability may lead to more serious complications such as fracture of the screws and loosening of the prosthesis [10]. When dealing with multiple prostheses, it is difficult to realize when one screw is loose, which may result in overloading of the others. This overload on the tightened screws may compromise the surrounded area of the implants, culminating in bone loss [11, 12].

Another important factor to be evaluated is the cyclic loading. The mechanical cycling of implant-supported prostheses may cause micromovements and fatigue of the joint components of screwed prostheses. Moreover, low preload values exhibit significant higher micromovements at the abutment-implant interface [13].

In order to analyze the screw-retained prosthesis biomechanics, the aim of this study was to evaluate the influence of cyclic loading in one-piece casted implant-supported frameworks on screw loosening and vertical misfit with two fit situations. The hypotheses tested were that (i) the cyclic loading decreases the loosening torque and the vertical misfit and (ii) the different levels of fit influence the screws stability on prosthesis-implant system.

Material and Methods

Ten waxed frameworks were obtained from a stainless steel master model, simulating the clinical situation of a complete fixed denture supported by five implants (labeled A, B, C, D, and E) positioned between the mental foramens with an arch arrangement of 10 mm interimplant space and a prosthesis cantilever extension of 15 mm. Five mini abutment analogs (SIN - Sistema de Implante Nacional S/A, São Paulo, SP, Brazil) were installed in the master model using transversal screws. Calcinable cylinders were fixed to the analogs for waxing the framework. The waxed pattern was reproduced in low shrinkage acrylic resin (Duralay II, Reliance Dental Mfg. Co, Worth, IL, United States) using a silicone matrix (Flexitime Easy Putty, Correct Flow, Heraeus-Kulzer, Hanau, HE, Germany). The waxed frameworks were casted in cp Ti (Tritan, Dentaurum, Ispringen, BW, Germany). After all frameworks were casted, acrylic resin models (Clássico, São Paulo, SP, Brazil) were fabricated with the assistance of a parallelometer. Two fit situations were simulated, passive (P) and non-passive (NP) (Figure 1). External hexagon analogs and mini abutments were used to the casts manufacturing. For P situation, ten models were created with juxtaposition between the mini abutments and the framework cylinder. For NP situation, the ten models were made using a perforated plate to simulate a misfit of 150 µm at E pillar position. This misfit was created at the model, prior to the framework casting.

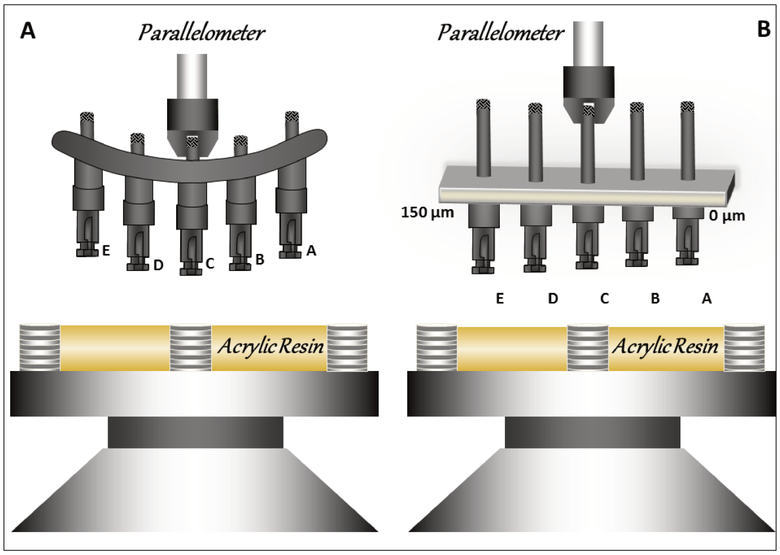

Figure 1. Casts manufacturing of passive fit (A) and non-passive fit (B) groups.

Loosening torque analysis

The loosening torque of mini abutments (MAS) and prosthetic screws (PS) were evaluated using a digital torque meter with a 0.1 Ncm precision (TQ8800; Lutron, Taipei, Taiwan), coupled to a device allowing its vertical position, so not to induce lateral stress. Only one researcher performed the tightening and loosening torque analyses. The tightening torque was applied following the manufacturer's instructions of 20 Ncm for mini abutment and 10 Ncm for prosthetic screw. The tightening and loosening sequences were the same: first the components of the ends (A and E), then the intermediate components (B and D) and finally the central component (C). The initial analysis was performed 24 hours after the tightening torque. The same initial parameters were used for the final analysis, which was performed soon after the mechanical cycling.

Vertical misfit analysis

Vertical misfit measurements were performed with an optic microscope (VMM-100-BT; Walter UHL, Asslar, HE, Germany) at 120X magnification. A calibrated examiner (ICC = 0.999) performed the analyses. The single-screw test protocol [11, 14, 15] was the technique used to measure the misfit, simulating the maximum misfit in the framework, by tightening the prosthetic screw with 10 Ncm. Vertical misfit between mini abutment platform and the inferior border of the denture framework were measured on buccal and lingual sides of cylinders C and E (when A was tightened) and of cylinders C and A (when E was tightened). A matrix was used to standardize the position of the readings. This matrix allowed the same position for all readings, in the central point of each abutment, before and after cyclic loading.

Cyclic loading

Cyclic loading was applied obliquely on the frameworks' E pillar (Figure 2). The test was performed with the frameworks immersed in artificial saliva (1.5 mmol/L Ca, 3.0 mmol/L P, 20.0 mmol/L $NaHCO_3$; pH 7.0) [16]. The frameworks were submitted to 10^6 mechanical cycles, with 2 Hz frequency, at 30° inclination (ISO 14801 - 2007) [17] in Fatigue Mechanics Simulator (ER 11000 Plus, ERIOS, São Paulo, SP, Brazil). The parameters were estimated supposing that an individual performs three episodes of chewing per day, each lasting 15 minutes, and at a chewing rate of 60 cycles per minute (1 Hz), what is equivalent to 2.700 chewing cycles per day or roughly 1 million cycles per year [18]. The simulation was performed with mechanical fatigue pressure of 6 bars corresponding to about 280 N for compressive load [19].

Figure 2. Frameworks positioned for cyclic loading.

An exploratory analysis of data was done and nonparametric tests were required. The Mann-Whitney test ($\alpha = 0.05$) was applied to compare P and NP groups for misfit and loosening torque analyses. The Wilcoxon test ($\alpha = 0.05$) was applied to compare before and after cyclic loading. The data from misfit were correlated with loosening torque analysis, using the Pearson correlation test ($\alpha = 0.05$). All statistical tests were performed with SAS (version 9.1, The SAS Institute, Cary, NC).

Results

Tables I and II show the means and standard deviation of misfit and loosening torque values obtained at initial and final times in both groups.

Regarding the MAS loosening torque, non-passive prostheses showed an increase on values after cyclic loading ($p = 0.0011$). No difference was observed on PS loosening torque for both groups. The passive group presented better stability with higher initial loosening torque ($p = 0.0333$), but the difference was not significant on final time.

Table I. Mean and standard deviation of loosening torque values.

Groups	Loosening Torque (MAS) - Ncm		p-value	Loosening Torque (PS) - Ncm		p-value
	Initial	**Final**		**Initial**	**Final**	
Passive	15.38 (1.92)Aa	16.77 (2.00)Aa	0.2227	5.62 (0.74)Aa	5.12 (0.94)Aa	0.6447
Non-Passive	15.59 (1.16)Ab	18.53 (1.03)Aa	0.0011	4.40 (0.80)Ba	4.53 (1.25)Aa	0.9909
p-value	0.9916	0.0824		0.0333	0.5116	

Means followed by different letters differ statistically (capital, column – Mann-Whitney Test; minor, line – Wilcoxon Test), α=0.05.

For misfit analysis, the cyclic loading had no influence on both groups, but there was a significant difference between the P and NP groups on both times.

A correlation between the misfit and PS loosening torque was found ($r = -0.4963$; $p = 0.0011$), but no correlation was observed between the misfit and MAS loosening torque values ($r = 0.15782$; $p = 0.3307$).

Table II. Mean and standard deviation of misfit values.

Groups	Misfit (resin)		p-value
	Initial	Final	
Passive	108.69 (45.64)Ba	95.84 (39.11)Ba	0.9752
Non-Passive	449.55 (99.73)Aa	416.03 (73.70)Aa	0.6990
p-value	<0.0001	<0.0001	

Means followed by different letters differ statistically (capital, column – Mann-Whitney Test; minor, line – Wilcoxon Test), α=0.05.

Discussion

According to the results, the first hypothesis was rejected since the loosening torque and misfit values did not decrease after cyclic loading. The second hypothesis was partially accepted since the different levels of fit influenced only the prosthetic screws stability before cyclic loading.

The desired goal of tightening any screw joint is to generate sufficient clamping force to keep the component parts together. The clamping force is the compressive force that two joint members exert on each other and is created by the force that the screws are exerting on them [15]. It can be achieved by applying a proper amount of tightening torque to the screw to generate the utmost preload (elongation of the screw) below the fatigue limits. When the total load applied to the screw (preload and external load) exceeds the fatigue parameters, the ability to fix the components together decreases dramatically [20]. The preload value may be influenced by several factors, such as elastic modulus of the screws, type of opposing joint surface, abutment design, friction coefficient, lubrication, rate of applied torque, and adaptation between the hexagon and the abutment [12, 15]. In this study, the loosening torque measurements were performed aiming to check the actual amount of force absorbed by the screws after the initial torque and the measurement following the mechanical cycling aimed to evaluate the effect of compressive load (masticatory simulation) at two levels of fit in maintaining the strength of torque of the same screws.

Many studies [10-12, 15] have sought to evaluate the behavior of prosthetic screws. This study assessed the behavior of both screws involved in multiple prostheses (prosthetic and abutment screws) to whole analyze the prosthesis biomechanics upon cyclic loading.

As regards the MAS loosening torque, the P and NP group showed no statistically significant differences at initial and final times. However, for non-passive group, it was observed that the loosening torque values increased after mechanical loading. It can be explained by the fact that the artificial saliva has an important contribution for the maintenance of preload [21, 22]. Duarte *et al*., 2012 [21] found that a corrosive layer is formed between the implant and abutment screw, which improves the clamping force on the joint after immersion in a fluoridated artificial saliva solution. Despite of the different saliva composition used in the present study, the real influence of the other components is not yet well known. The maintenance of preload is also unclear for Nigro *et al.,* 2010 [22], who discussed on three possible reasons: the first is that an intervening humid media certainly eliminates most of internal shear forces produced on threads; the second is that wet and dry regions can generate non-uniform friction resulting in final values higher than the initial ones; and the third is that wetting can contribute to more apical thread levels, which hinders the embedment relaxation (settling), and loss of applied torque is minimal. Beyond that, the non-passive group showed higher increasing on loosening values, which can be inferred that cyclic loading may induces a better gearing between the screw threads in this situation.

For PS loosening torque values, no difference was observed between times for both groups, but the difference appeared when the groups were compared at initial time. The loosening torque values were lower in the frameworks with a higher vertical misfit than those presented by the fitting ones. Therefore, when multiple prostheses are tightened, beyond the forces inherent to a screwed joint, residual static stresses are created because of the misfit [23]. Based on these statements, it can be suggested that residual stresses created in non-passive fit frameworks after the screws tightening decreases the stability of the screwed prosthetic joints.

Furthermore, the cyclic loading was not sufficient to generate difference between the times of analysis. The results of this study are in agreement with other studies in the literature [24, 25] that also did not find influence of mechanical cycling in loosening torque.

A perfect fit is very difficult to achieve in multiple implant-supported prostheses. According to Mitha *et al.* 2009 [26], as-cast titanium frameworks are inaccurate and imprecise when judged against the 150 µm [4] requirement for passivity of fit. These distortions are tridimensional and can be attributed to factors inherent to the casting process. Thus, in order to reduce distortion, other techniques must be used, such as, casting in separate units or sectioning followed by indexing [26] or laser-welding [27, 28].

In this study, the passive group presented high misfit values, although within the clinically acceptable limit suggested by Jemt [4] in 1991. This certain degree of inaccuracy may have occurred because casting in Ti is a delicate process. When this metal is casted, there is only one cylinder type that can be used for the manufacturing of implant-supported prostheses, which is the entirely calcinable one. Due to such limitation, the casting process can generate unavoidable irregularities at the bottom of the cylinders, such as defects and positive blisters, which may influence the perfect fit between frameworks and abutments. Another possible contributing factor for the high misfit values is the single screw test protocol, which generates the greatest misfit possible on the non-screwed side.

The cyclic loading had no influence on misfit values for any of the groups. The results found in the current study corroborate with Bhering *et al.* 2012 [24] who found that the misfit values of two abutment types did not differ before and after cyclic loading of single unit implant-supported prostheses. The loading piston may exert a force to close the space between the platform and the abutment cylinder, but when this force had ceased, the misfit may have returned to their original state.

An inverse relationship was observed between vertical misfit and PS loosening torque, which indicates that as misfit increases, detorque values slightly decrease. These findings are in accordance with others authors [20] who found that the inadequate marginal adaptation of screwed prostheses and prosthetic screw loosening are correlated factors. Repeated tightening and loosening of the screw is a common practice during the manufacturing and in the clinical followings of prosthesis and its performance may indicate a gradual reduction in the removal torque and consequent instability of the system, besides being unpleasant to the patient and costly for the professional [8]. Thus, it is important to emphasize that prosthetic screw loosening may be a sign of a future failure of other components [8, 10] and the development of biological complications, besides suggesting the presence of overload on rehabilitation [8, 29].

Conclusion

Based on the results and within the limitations of this study, it was concluded that:

Cyclic loading for one year of masticatory simulation has no influence on prosthetic and abutment screw loosening and vertical misfit of passive fit frameworks, but increased abutment screw loosening torque for non-passive situation.

Ill-fitting prosthesis leads to screw loosening with greater facility.

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Strain gauge analysis of multiple frameworks at two levels of fit under cyclic loading conditions

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Strain gauge analysis of multiple frameworks at two levels of fit under cyclic loading conditions

Abstract

The aim of this study was to evaluate the influence of cyclic loading on strain of commercially pure titanium (cp Ti) one-piece casted 5-unit implant-supported frameworks at two fit situations. Ten frameworks simulating bars for full-mouth fixed prostheses were casted in cp Ti. Two fit conditions were created: passive fit group (P) and non-passive fit group (NP). The single-screw test was performed and the vertical misfit was quantified using an optical microscope. The strain after fixation was measured using strain gauge analysis with each framework tightened to the plaster cast with 10 Ncm standardized torque. The P group and NP group were evaluated before and after cyclic loading. Data were analyzed by Mann-Whitney and Wilcoxon tests ($\alpha = 0.05$) and Pearson correlation test. There were no statistically differences in strain values before and after cyclic loading for both P and NP groups ($p = 0.9997$; $p = 0.8694$, respectively). NP group presented higher strain than P group at initial ($p = 0.0011$) and final ($p = 0.012$) times. There was correlation between the vertical misfit and strain ($r = 0.64333$; $p < 0.0001$). Cyclic loading does not influence strain of the framework. Ill-fitting prostheses induce higher strain than the adapted ones, regardless of cyclic loading for one-year masticatory simulation.

Key words: implant prosthesis, strain gauge, cyclic loading, vertical misfit, titanium.

Introduction

Osseointegrated dental implants are a successful treatment for rehabilitation of partially and completely edentulous patients (Branemark, 1983). Nonetheless, passive fit in certain situations such as multiple one-piece cast implant-supported frameworks is hard to achieve. Biomechanical concerns associated with the connection of non-passive frameworks include stress in the prosthesis, the implant and the surrounding bone

(Branemark, 1983; Skalak, 1983). As the strain and stress concentration increases in the supporting tissues, the magnitude of the forces affects all components of bone-implant-prosthesis system (Nishioka et al., 2009; Nishioka et al., 2010).

The clinically acceptable degree of misfit is still divergent in the literature. Authors have considered as an acceptable level values ranging from 10 µm (Branemark, 1983) to 150 µm (Jemt, 1991), while others have defined passive fit as the absence of strain upon framework fixation (Karl et al., 2004).

The strain gauge method has been used to analyze biomechanical behavior in several situations of prosthetic fabrication. Many studies (Abduo et al., 2011; Abduo and Lyons, 2012; Hegde et al., 2009; Karl et al., 2009) have focused on misfit present at the implant-framework interface in non-passive implant-supported prosthesis in order to seek a relationship between strain and misfit. Strain gauge is considered an indirect measurement method, which analyzes a physical effect. The mechanical deformation of a structure is measured through electrical signals captured by a transducer. A strain gauge is an electrical sensor which quantifies surface deformation. Its operating principle is based on the variation of electrical resistance transformed into levels of deformation (Nishioka et al., 2010).

Moreover, mechanical cycling on implant-supported prostheses may cause micromovements and fatigue of the joint components in screwed prostheses. Low preload values exhibit significant higher micromovements at the abutment-implant interface (Gratton et al., 2001).

The aim of this study was to evaluate the influence of cyclic loading in one-piece casted 5-unit implant-supported frameworks in commercialy pure titanium (cp Ti) on vertical misfit and strain during fixation at two fit situations. The hypotheses tested were that (i) the cyclic loading decreases the vertical misfit and the strain and (ii) the different levels of fit influence the strain transmitted to prosthesis-implant system.

Material and methods

A stainless steel master model (Figure 1) was used to obtain 10 waxed frameworks that simulated a clinical situation of five implants (named A, B, C, D, and E) positioned between the mental foramens with an arch arrangement of 10 mm interimplant space (center to center) and a cantilever extension of 15 mm. Five mini abutment analogs (SIN - Sistema de Implante Nacional S/A, São Paulo, SP, Brazil) were fixed to the master model with transversal screws. Calcinable cylinders were adapted over them for waxing the framework with a progressive waxing technique. The frameworks were reproduced in low shrinkage acrylic resin (Duralay II, Reliance Dental Mfg. Co, Worth, IL, United States) using a silicone matrix (Flexitime Easy Putty, Correct Flow, Heraeus-Kulzer, Hanau, Germany). The waxed frameworks were casted in cp Ti (Tritan, Dentaurum, Ispringen, Germany). After all frameworks casted, plaster models (Durone, Dentsply, Petrópolis, RJ, Brazil) were fabricated with the assistance of a parallelometer at two fit situations, passive (P) and non-passive (NP) (Figure 2). In the P group (Figure 2A), the casts were fabricated featuring a juxtaposition of mini abutments and framework cylinders. For non-passive situation (Figure 2B), the casts were made with the aid of a perforated plaque that simulated 150 µm gap at E pillar region. For casts manufacturing, modified analogs (Figure 3A) were used, embedded in gypsum in about 10 mm. These analogs were machined in titanium with 20 mm length and hollow interior to allow elastic deformation to be detected by the strain gauges. Acrylic resin casts were manufactured with conventional analogs (Figure 3B) following the same parameters, in order to perform cyclic loading.

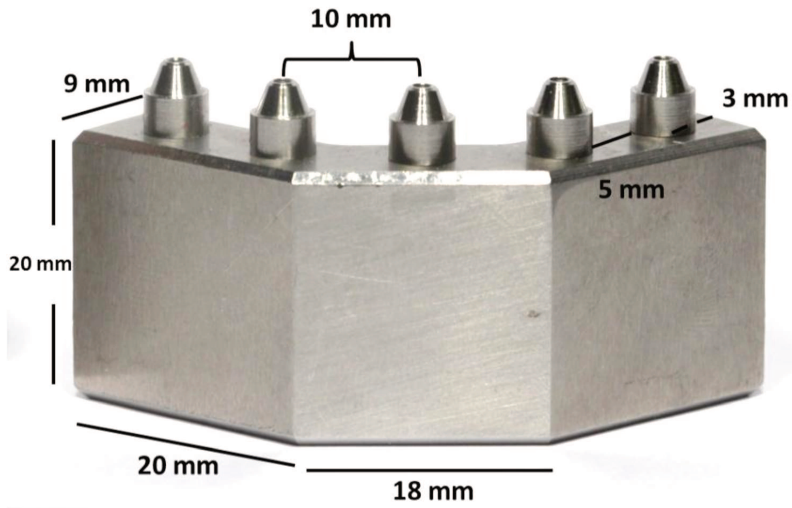

Figure 1. Stainless steel master model and disposition of the abutment analogs.

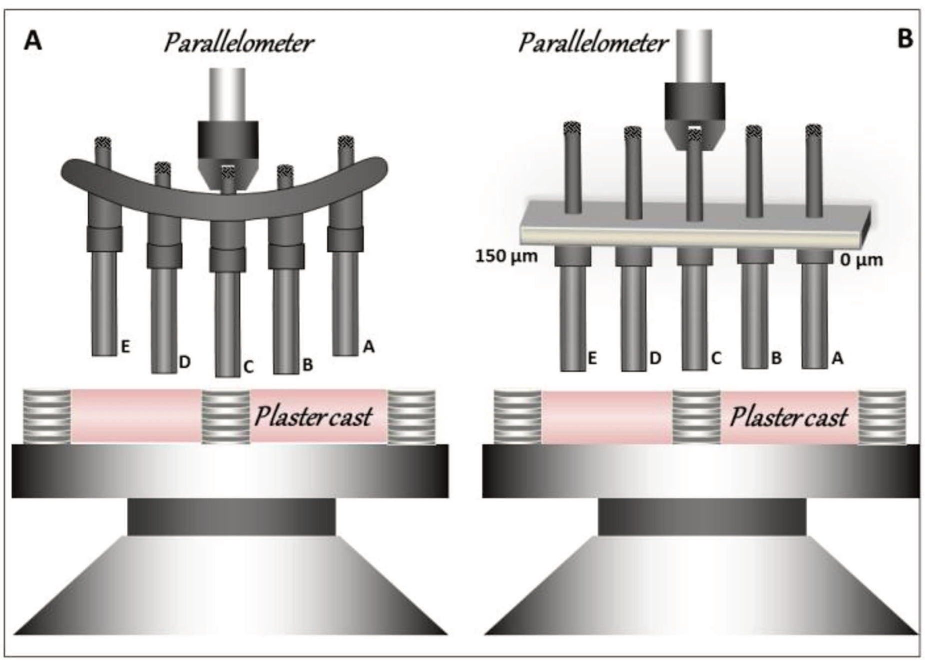

Figure 2. Schematic illustration of the two fit conditions: (A) passive fit; (B) non-passive fit.

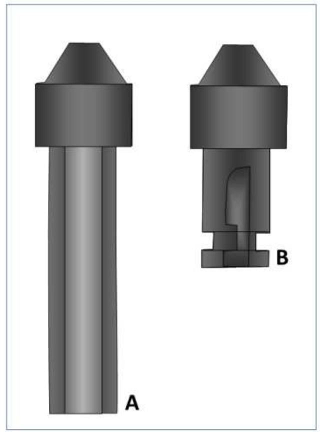

Figure 3. Schematic illustration of modified (A) and conventional (B) analogs.

Vertical misfit analysis

Vertical misfit measurements were performed with an optic microscope (VMM-100-BT; Walter UHL, Asslar, Germany) at 120X magnification by a calibrated examiner (ICC = 0.999). The single-screw test protocol (Farina et al., 2012; Sartori et al., 2004; Spazzin et al., 2009) was the technique used to measure the misfit in order to simulate the maximum misfit in the framework, by tightening the prosthetic screws with 10 Ncm. Vertical misfit between mini abutment platform and the inferior border of the framework was then measured on buccal and lingual sides in the cylinders C and E (when A was tightened) and on buccal and lingual sides in the cylinders C and A (when E was tightened). A silicon matrix was used to allow the same position for all readings of all cylinders, in the middle point of each abutment, before and after cyclic loading.

Strain analysis

The measurement was performed by five strain gauges (PA-06-060BG-350L, Sensor Excel Engineering; Embu, São Paulo, SP, Brazil), one corresponding to each modified analog. The strain gauges were bonded on the internal face of the base of the modified analogs. The strain gauge arrangement was a quarter of Wheatstone bridge (Figure 4), allowing the capture of deformation by electrical signals on a computer-controlling device (ADS2000; Lynx Tecnologia Eletrônica Ltda, São Paulo, SP, Brazil), processed by specific software (AqDados 7; Lynx Tecnologia Eletrônica Ltda, São Paulo, SP, Brazil). The framework was adapted over the modified mini abutment analog and fixed according to manufacturer's instructions with a torque of 10 Ncm. The tightening sequence was A, E, B, D, C, according to pillars position. All of the strain gauges were zeroed and calibrated prior to each analysis. The magnitude of deformation on each strain gauge was recorded in microstrain units (με). Strain average was obtained for each framework analyzed. The analyses were performed during ten minutes, with five minutes interval selected after stabilization for statistical analysis purposes.

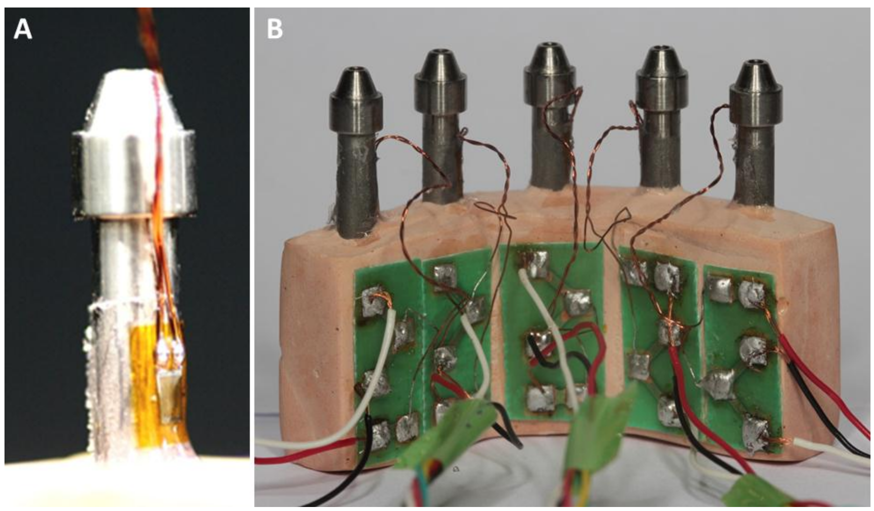

Figure 4. A) Closer view of the strain gauge bonded; B) Strain gauge model with a quarter Wheatstone bridge.

Cyclic loading

The complete fixed implant-supported frameworks were immersed in artificial saliva (1.5 mmol/L Ca, 3.0 mmol/L P, 20.0 mmol/L $NaHCO_3$; pH 7.0) (Pigozzo et al., 2009) and submitted to mechanical fatigue testing upon oblique loading. The frameworks were subjected to 10^6 mechanical cycles, with 2 Hz frequency, at 30° inclination (ISO 14801 - 2007) in Fatigue Mechanics Simulator (ER 11000 Plus, ERIOS, São Paulo, SP, Brazil). The parameters were estimated supposing that an individual performs three episodes of chewing per day, each lasting 15 minutes, and at a chewing rate of 60 cycles per minute (1 Hz), what is equivalent to 2.700 chewing cycles per day and roughly 1 million cycles per year (Wiskott et al., 1995). A pressure of 6 bars was used, corresponding to about 280 N of oblique compressive load (Mericske-Stern and Zarb, 1996) applied to the distal region of framework, at pillar E.

An exploratory analysis of data was done and nonparametric tests were required. The Mann-Whitney test ($\alpha = 0.05$) was applied to compare P and NP groups for misfit and strain analyses. The Wilcoxon test ($\alpha = 0.05$) was applied to compare initial and final times, before and after cyclic loading. The data from misfit analysis were correlated with strain, using the Pearson correlation test ($\alpha = 0.05$). All statistical tests were performed with Bioestat (Bioestat, version 5.0, Instituto de Desenvolvimento Sustentável Mamirauá, Belém, PA).

Results

Table I and Table II show the means of misfit and strain values obtained at initial and final times in both groups, respectively.

For misfit analysis, there was statistically significant difference before and after cyclic loading for both groups. The misfit values were significantly different between P and NP groups.

For strain analyses, test showed significant difference between P and NP group at initial and final times. No difference was observed between the initial and final times for both P and NP groups.

There was a positive correlation between misfit and strain values ($r = 0.64333$; $p < 0.0001$).

Table I. Mean and standard deviation of misfit values.

Groups	Misfit - µm		p-value
	Initial	Final	
Passive	157.86 (61.68)Ba	140.44 (59.31)Bb	0.0125
Non-Passive	383.11 (45.90)Aa	379.15 (43.45)Ab	0.0144
p-value	0.0002	0.0002	

Means followed by different letters differ statistically (capital, column – Mann-Whitney Test; minor, line – Wilcoxon Test).

Table II. Mean and standard deviation of strain values.

Groups	Strain - µε		p-value
	Initial	Final	
Passive	177.40 (92.70)Ba	187.41 (113.33)Ba	0.9997
Non-Passive	433.83 (155.22)Aa	389.20 (216.02)Aa	0.8694
p-value	0.0011	0.012	

Means followed by different letters differ statistically (capital, column – Mann-Whitney Test; minor, line – Wilcoxon Test).

Discussion

According to the results, the first hypothesis was partially accepted since the misfit values decreased after cyclic loading, but there was no difference for strain values. The second hypothesis was accepted since the different levels of fit influenced the strain transmitted to prosthesis-implant system.

The mechanism of stress transference around bone is physiologically complex and any laboratory model is an approximation of the clinical situation, evaluating the behavior of strain distribution. This study used strain gauge analysis to compare the strain behavior during the fixation at two fit situations before and after cyclic loading.

Strain gauge is a controllable method for quantitative analysis of strain, currently used for this purpose (Abduo et al., 2011; Abduo and Lyons, 2012; Watanabe et al., 2000). The foil gauge was bonded on the implant analog surface to avoid the influence of other materials, working in the same way as if it had been bonded to the framework, due to the similarity of material.

The results of this study showed that strain magnitude varied according to the increase of misfit with NP group presenting higher strain values than P group. These findings are in accordance with other studies (Abduo and Lyons, 2012; Abduo and Swain, 2012; Hegde et al., 2009) that reported that prosthesis misfit would induce strain within the peri-implant structures. The greater the misfit, the greater the stress within the implant system (Millington and Leung, 1995).

When both P and NP groups were subjected to cyclic loading, it had no influence on strain values. Although mechanical cycling on implant-supported prostheses may cause fatigue of the joint components in screwed prostheses (Gratton et al., 2001), changes on system may not be found. The framework seems not to have suffered much alteration through mechanical cycling, which may explain the absence of difference between the times. When evaluated the misfit after cyclic loading, the values decreased in both groups, which might have been a consequence of framework distortion and wear due to its tendency to close the marginal gap during masticatory simulation.

In this study, all frameworks generated deformation on the supporting analogs. Therefore, it could be stated that implant strain is an unavoidable consequence of tightening the retaining screws of the prosthesis. This finding corroborates with other strain gauges studies (Clelland et al., 1995; Hegde et al., 2009) that have assessed the fit of implant frameworks and found that framework fixation induces strain. As a consequence of these findings some authors (Karl et al., 2004; Sahin and Cehreli, 2001) have suggested the revision of passive fit definition based on strain.

During the fixation, the abutment screw exerts a compressive force to keep the contact between the abutment and the implant surface. Due to the characteristics inherent to framework casting, component fit is not perfect but needs to be clinically acceptable. Torque of the prosthesis-abutment set induces stress, which is transmitted to the surrounding bone (Nishioka et al., 2010). Physiologically, clinical and laboratory studies indicate that permanent mechanical stimulation is necessary. Magnitudes of deformation above 100 με are required to prevent bone resorption (Wiskott and Belser, 1999); however, the stimulation values must not exceed the physiological limit of 3000 με (Frost, 1992).

According to Frost (Frost, 1992), based on studies in animals, bone's response to mechanical usage can be divided into 4 windows named as disuse, adapted, mild overload and pathologic overload. The biologic response depends on the vigor of mechanical usage. The disuse window occurs at 50 με, the adapted window occurs when peak bone strains range from 50 to 1500 με, mild overload occurs at 1500 to 3000 με, and the pathologic overload window occurs when strains exceed 3000 με.

Although this is not a real clinical simulation, the present methodology was used to avoid some noise on the analysis and evaluate the comparison of the behavior between passive fit and non-passive fit groups. Nevertheless, *in vivo* studies, despite the difficulty to be performed, are necessary to analyze clinical behavior. The increase in strain magnitude observed in non-passive frameworks does not seem to be adequate for bone tissue and may cause deleterious effects to the interface bone-implant (Sahin and Cehreli, 2001; Skalak, 1983). To prevent such effects laboratory methods have been used to correct ill-fitting implant supported frameworks, such as laser welding (Nunez-Pantoja et al., 2011; Tiossi et al., 2012; Tiossi et al., 2008) and electroerosion (Sartori et al., 2004) in order to decrease the strains transferred to the implant-bone interface. Furthermore, CAD-CAM system has also been used to improve the passive fit of multiple prostheses (Abduo and Lyons, 2012; Abduo and Swain, 2012).

Within the limitations of this study, it was found that ill-fitting prostheses induce higher strain than adapted ones. Strain increases as misfit increases. Cyclic loading for one-year masticatory simulation influences the frameworks on misfit, but has no influence on strain.

Conflict of interest statement

The authors had no conflict of interest in working on or writing this article.

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O objetivo neste estudo foi avaliar a biomecânica de reabilitações implantossuportadas, por meio da análise de torque de afrouxamento, desajuste e tensão, de próteses múltiplas parafusadas frente à simulação de um ano de uso clínico em diferentes situações de adaptação.

De modo geral, esperava-se uma ação prejudicial da ciclagem mecânica sobre a biomecânica de próteses múltiplas parafusadas. Entretanto, os resultados apresentados neste estudo mostram que a simulação em laboratório de um ano de uso clínico de próteses implantossuportadas não teve um papel relevante frente às análises de torque de afrouxamento, desajuste e tensão. Paradoxalmente, quando avaliado o comportamento de próteses parciais e totais fixas dos grupos adaptado (GII) e desajuste simulado (GIII), os valores de torque de afrouxamento dos *mini abutments* aumentaram após a ciclagem. Este comportamento pode estar relacionado com o fato de que as análises iniciais não tiveram a influência da saliva, enquanto que as análises finais foram realizadas após a ciclagem mecânica, na qual houve a imersão das infraestruturas em saliva artificial. Segundo Duarte *et al*., 2012, quando superfícies de parafusos são avaliadas quimicamente, uma camada corrosiva pode ser formada entre as superfícies metálicas em condições estáticas na união implante-abutment durante imersão em solução de saliva artificial. A corrosão pode ocorrer devido a uma reação eletroquímica que acontece entre o implante, *abutment* e compostos de saliva artificial. Com isso, esta camada reacional pode ser responsável pela maior resistência ao afrouxamento apresentada pelos *mini abutments* após a imersão em solução de saliva artificial fluoretada. Apesar de não ter sido utilizada a mesma composição de saliva no presente estudo, o efeito dos outros componentes ainda é incerto.

Muito citada na literatura (Sartori *et al.*, 2004; Spazzin *et al.*, 2009; Spazzin *et al.*, 2010; Farina *et al.*, 2012), a técnica do parafuso único foi utilizada para aferição do desajuste marginal. Essa técnica possibilita a comparação entre os desajustes criados, uma vez que seria muito difícil analisar os desajustes com todos os parafusos apertados, o que tenderia ao fechamento dos *gaps*. Porém, quando esta técnica é aplicada, as distorções e imperfeições decorrentes da fundição nas plataformas dos cilindros podem ocasionar valores superestimados para o desajuste das infraestruturas. Por isso, quando uma situação clínica é analisada, os desajustes apresentados são consideravelmente menores, porém às expensas da geração de tensões indesejáveis ao osso (Tramontino, 2008).

Os valores de desajuste encontrados para o grupo I (desajuste decorrente da fundição), não se apresentaram dentro do limite clinicamente aceitável de 150 µm proposto por Jemt (Jemt, 1991). Esses achados corroboram com outros estudos (Sartori *et al.*, 2004; Nakaoka, 2007; Mitha *et al.*, 2009; Nunez-Pantoja *et al.*, 2011), que demonstram uma grande dificuldade durante a fundição convencional em titânio, apresentando ainda uma adaptação cervical insuficiente, o que resulta em distorções e imprecisões neste tipo de prótese. No que concerne ao grupo II, em que os modelos foram construídos de maneira passiva, foram encontrados valores dentro dos limites clinicamente aceitáveis (Jemt, 1991), apesar de ainda apresentarem um certo grau de desajuste. Este desajuste pode ter sido consequência de imperfeições na infraestrutura, decorrentes da técnica convencional de fundição em titânio comercialmente puro. Esse metal não permite um procedimento chamado de sobrefundição, que, se possível, tenderia à redução de irregularidades, bolhas positivas e porosidades nas bases dos cilindros. Quando utilizados cilindros totalmente calcináveis, a probabilidade dessas imperfeições ocorrerem é maior, o que pode prejudicar a perfeita adaptação entre as infraestruturas e os pilares.

Um método já consagrado na literatura para avaliar a tensão em próteses mal adaptadas é a extensometria (Hegde *et al.*, 2009; Abduo *et al.*, 2011; Abduo & Lyons, 2012; Abduo & Swain, 2012). Nesta técnica utilizam-se sensores (*strain gauges*) capazes de registrar a deformação elástica da superfície na qual estão colados. Neste trabalho, cada infraestrutura foi posicionada sobre os análogos nos modelos de gesso e foi realizado o aperto de todos os parafusos, de acordo com a sequência A-B para a prótese parcial fixa e A, E, B, D e C para a prótese total fixa. A fixação foi realizada em dois estágios. O primeiro aperto foi realizado até resistência inicial em todos os parafusos e posteriormente, o segundo aperto foi aplicado o torque de 10 Ncm (conforme indicado pelo fabricante) respeitando a sequência pré-estabelecida.

A aferição da tensão foi realizada por meio de dois extensômetros elétricos de resistência (PA-06-060BG-350L, Excel Engenharia de Sensores; Embu, São Paulo), posicionados diretamente um em cada análogo modificado de *mini abutment* para a prótese parcial fixa e cinco extensômetros posicionados um em cada análogo para a prótese total fixa. O arranjo escolhido para os extensômetros em cada fixação foi de 1/4 de ponte de Wheatstone, a fim de obter-se o valor real de cada pilar, permitindo a captação das deformações por meio de sinais elétricos em um aparelho controlado por computador (ADS2000; Lynx Tecnologia Eletrônica Ltda, São Paulo, SP), e processado por um software específico (AqDados 7; Lynx Tecnologia Eletrônica Ltda, São Paulo, SP). A unidade utilizada para os dados foi με. Ao final da análise foram obtidas médias de tensões para cada infraestrutura analisada.

Para a construção dos modelos do grupo III, foram utilizadas placas metálicas perfuradas nas quais os análogos foram parafusados para a obtenção de um desajuste de 150 μm. Este dispositivo apresenta duas ou cinco perfurações para o parafusamento dos análogos, dependendo do design (PPF ou PTF), sendo pré-determinado um desnível gradativo entre as mesmas. Uma vez parafusados os análogos, este desnível gradativo faz com que os análogos das extremidades apresentem variação da posição vertical de 150 μm, padronizando assim o desajuste vertical dos modelos. Esta metodologia foi adotada a fim de exacerbar o desajuste criado no grupo III, padronizando os desajustes no modelos e não para as infraestruturas. Porém, pode ser considerada uma limitação deste tipo de confecção o fato de que não ser levada em consideração a distorção da infraestrutura após a fundição. Com essa padronização dos modelos, foi possível realizar as análises de extensometria entre os tempos inicial e final sem a variação de local de colagem no caso de modelos diferentes.

Para a análise do torque de afrouxamento, foram avaliados tanto os parafusos protéticos quanto os parafusos de *mini abutment*, devido este ser pouco avaliado na literatura. Para a padronização do posicionamento vertical do torquímetro digital foi utilizado um dispositivo a fim de evitar tensões laterais durante as aferições. Desta maneira, há uma maior segurança ao se afirmar que somente forças axiais foram aplicadas e não obliquas.

Com relação aos valores de torque de afrouxamento dos parafusos protéticos, observou-se uma maior estabilidade no grupo II (adaptado) do que nos grupos I e III, sendo mais acentuado para próteses totais fixas. Isto demonstra que próteses parafusadas múltiplas necessitam de um bom assentamento para se evitar falhas mecânicas. Esses achados corroboram com muitos estudos (Millington & Leung, 1995; Spazzin *et al.*, 2009; Spazzin *et al.*, 2010) que também afirmam que há uma maior soltura de parafusos quando analisadas próteses mal adaptadas, pois podem gerar tensões estáticas residuais, na qual a magnitude depende da amplitude do desajuste.

Conclusão

Considerando as limitações deste estudo e os resultados obtidos, pode-se concluir que:

Os parafusos protéticos apresentam tendência maior ao afrouxamento em próteses mal adaptadas.

A simulação de um ano de uso clínico de próteses múltiplas parafusadas não exerce influência relevante sobre o torque de afrouxamento, desajuste e tensão.

Próteses desadaptadas induzem maiores níveis de tensão que próteses adaptadas, o que pressupõe que à medida em que o desajuste aumenta a tensão também aumenta.

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DETALHAMENTO DA METODOLOGIA

Confecção das infraestruturas protéticas

Enceramento dos corpos de prova

Para a realização deste estudo foram confeccionados dois modelos mestre em aço inoxidável. Esses modelos mestre possuem as seguintes dimensões, de acordo com a Figura 1.

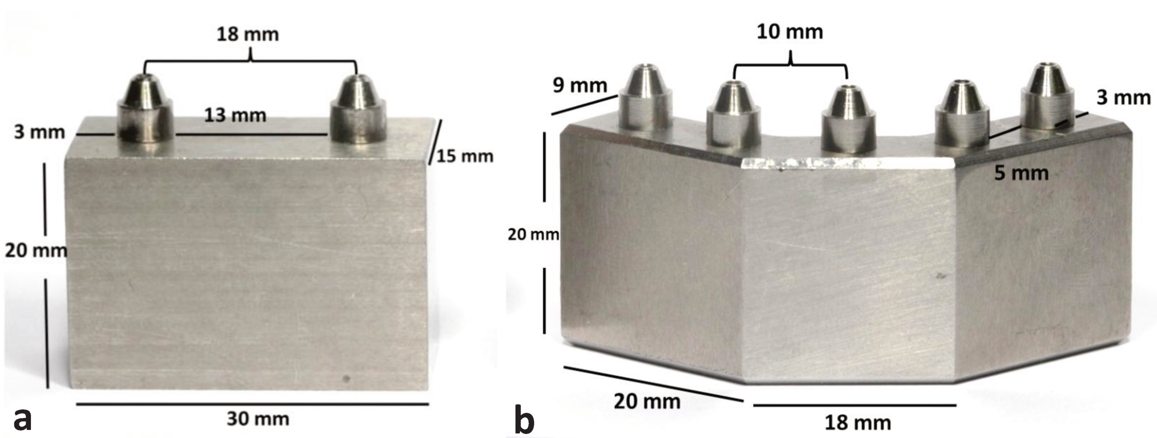

Figura 1. Modelos mestre metálicos. a) Prótese Parcial Fixa (PPF); b) Prótese Total Fixa (PTF).

A partir dos modelos mestre metálicos foram obtidos os corpos de prova. Duas réplicas do análogo *mini abutment* (SIN - Sistema de Implante Nacional S/A, São Paulo, SP, Brasil) foram fixadas no modelo mestre da PPF por meio de dois parafusos transversais e cinco réplicas do análogo de *mini abutment* foram fixadas no modelo mestre da PTF por meio de cinco parafusos transversais. O posicionamento dos análogos no modelo mestre da PPF teve como objetivo representar o posicionamento de dois implantes inferiores, um primeiro pré-molar e um primeiro molar inferior.

Nos análogos de *mini abutment* foram adaptados dois cilindros calcináveis (SIN - Sistema de Implante Nacional S/A, São Paulo, SP, Brasil) para enceramento de uma infraestrutura padrão (modelo PPF), por meio da técnica de enceramento progressivo utilizando cera cinza para escultura (Cera Yeti Thowax, Dental Produkte GmbH, Engen, Germany) (Figura 2).

Figura 2. Padrão de enceramento em cera cinza (PPF).

Já para o enceramento do grupo PTF, foram confeccionados bastões de resina acrílica de baixa contração, com secção transversal de 5 mm, adquiridos através da moldagem de sprues de 5 mm com silicone de condensação (Zetalabor, Zhermack, Rovigo, Itália). Esses bastões foram unidos aos cilindros calcináveis adaptados aos análogos através da técnica de Nealon, conferindo à estrutura um formato de arco (Figura 3).

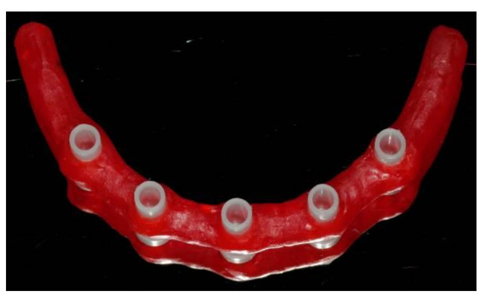

Figura 3. Padrão de enceramento em resina acrílica (PTF).

Após o enceramento foi realizada a duplicação do enceramento padrão utilizando silicone de adição (*Flexitime Easy Putty, Correct Flow*, Heraeus-Kulzer, Hanau, Alemanha) para padronização das infraestruturas que posteriormente foram enceradas em resina acrílica de baixa contração. A configuração externa desse padrão moldado em silicone de adição foi realizada em duas ou três etapas, de forma a servir como um molde partido para a confecção das infraestruturas. Primeiramente, moldou-se um dos lados do conjunto, e após a polimerização do silicone, isolou-se com vaselina sólida as extremidades desse molde. Em seguida, moldou-se o outro lado do conjunto atentando-se ao fato de que o mesmo estivesse bem justaposto à extremidade isolada do molde (Figura 4).

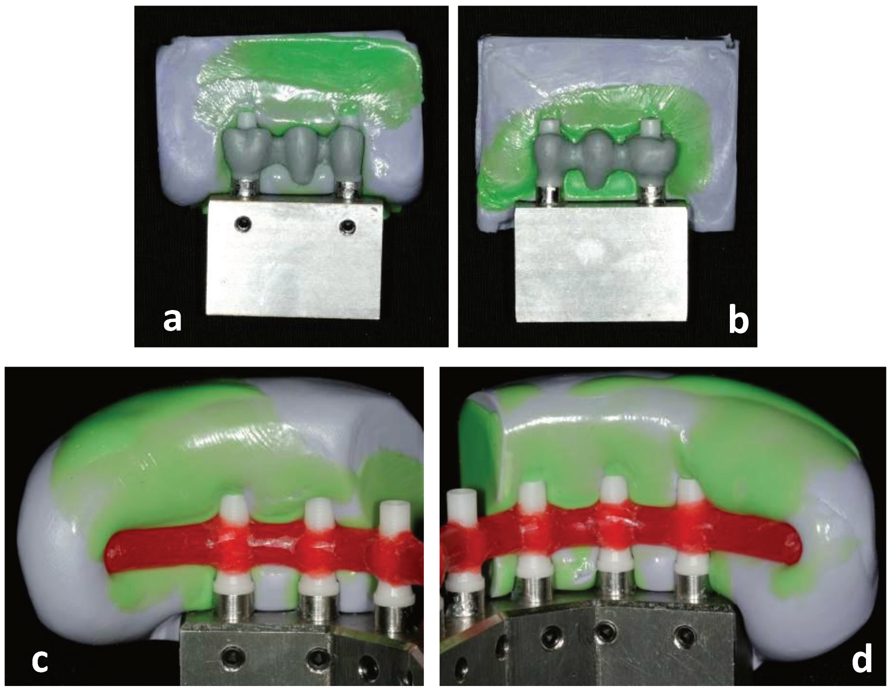

Figura 4. Molde em silicone de adição. a,b) PPF; c,d) PTF.

Por meio desse molde de silicone as infraestruturas foram enceradas em resina Duralay II (Reliance Dental Mfg. Co, Worth, IL, Estados Unidos), totalizando 10 unidades de PPF e 10 unidades de PTF. Durante todo enceramento, a adequada adaptação do modelo mestre em relação ao molde de silicone foi verificada, para certificar a padronização das infraestruturas enceradas. O modelo mestre, com os cilindros calcináveis previamente parafusados aos análogos, foi posicionado no molde de silicone e ambas as partes preenchidas com resina acrílica por meio da técnica de Nealon, com auxílio de um pincel. As partes do molde de silicone foram posteriormente unidas, esperando-se 5 minutos para adequada polimerização da resina acrílica, antes da abertura do mesmo (Figura 5).

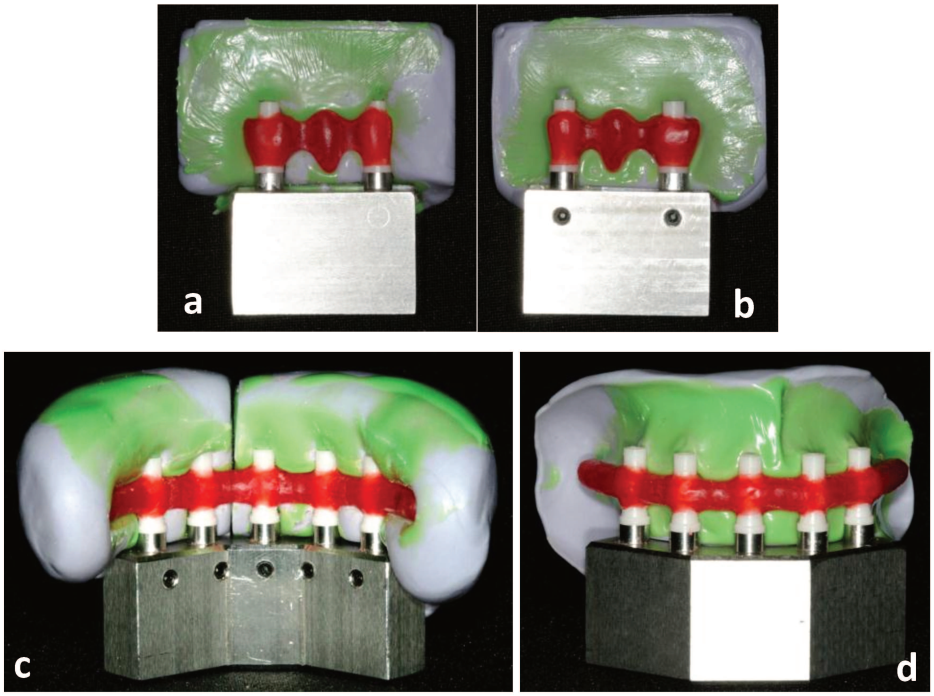

Figura 5. Obtenção das infraestruturas em resina acrílica. a, b) PPF; c, d) PTF.

Após o processo de enceramento da PPF, a região entre pré-molares de todos os enceramentos realizados foi seccionada com o uso de um disco de aço monoface para evitar tensões devido à contração de polimerização da resina acrílica. As regiões seccionadas foram novamente posicionadas no molde de silicone, parafusadas aos análogos e unidas com resina acrílica. Da mesma maneira foi realizado para os enceramentos da PTF, sendo a infraestrutura seccionada em quatro regiões, entre os análogos. Com o uso de pontas de tungstênio realizou-se o acabamento das infraestruturas em resina (Figura 6).

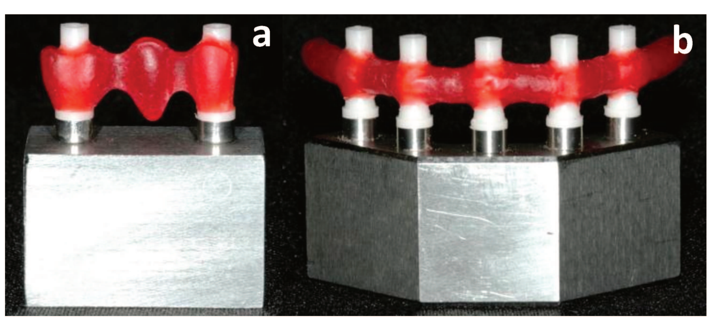

Figura 6. Enceramento concluído sobre a matriz metálica. a) PPF; b) PTF.

Depois de enceradas todas as infraestruturas, foram confeccionados os modelos de trabalho do grupo I. Para isso, blocos de resina acrílica (Class Mold, Clássico, São Paulo, SP, Brasil) com perfurações em posição equivalente aos análogos foram pré-confeccionados (Figura 7) a fim de minimizar os efeitos da contração de polimerização da resina acrílica.

Os análogos de implante, os *mini abutments* e as infraestruturas enceradas foram então unidos através de parafusos protéticos, de modo que este conjunto fosse posicionado no bloco de resina acrílica, com o auxílio de um delineador (Figuras 7a e 8a). Essa união foi realizada por meio da técnica de Nealon até o completo preenchimento das perfurações com resina acrílica (Figuras 7b, 8a e 8b). Após a polimerização da resina acrílica, cada infraestrutura encerada foi testada sobre o modelo obtido, para verificação do assentamento passivo e adaptação das mesmas sobre os análogos.

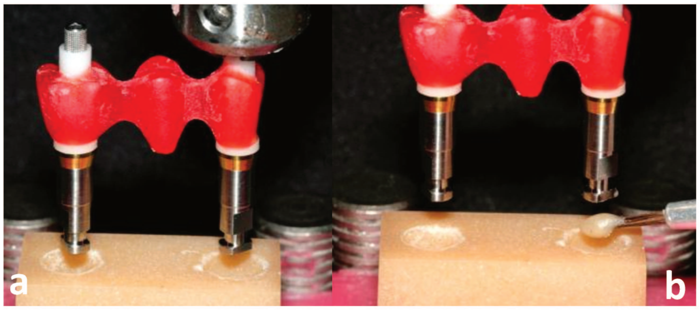

Figura 7. Confecção dos modelos de trabalho do grupo I. a, b) PPF.

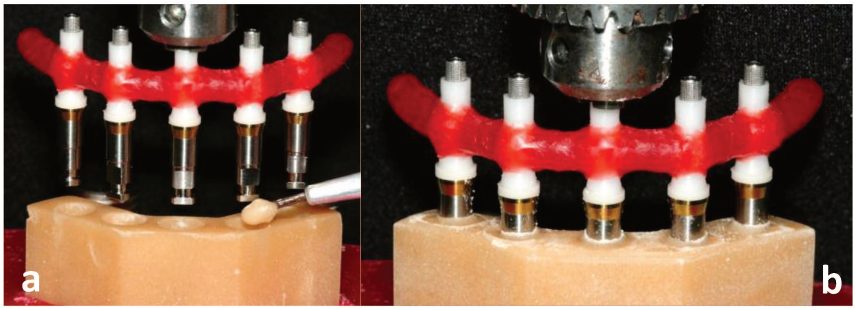

Figura 8. Confecção dos modelos de trabalho do grupo I. a, b) PTF.

Confecção dos modelos para extensometria

Para a realização dos ensaios de extensometria foram usinadas réplicas modificadas (Tramontino *et al.*, 2009) dos análogos de *mini abutment,* possibilitando a colagem dos extensômetros (PA-06-060-BG-350L, Excel Sensores Ltda, Embu, São Paulo) nos mesmos. A padronização da configuração do design do pilar *mini abutment* foi feita tendo como modelo um análogo de *mini abutment* (SIN - Sistema de Implante Nacional S/A, São Paulo, SP, Brasil), sendo que as réplicas foram confeccionadas em titânio, com hastes de 20 mm de comprimento (Figura 9). Os análogos modificados foram confeccionados com as hastes fresadas, de modo que seu interior ficasse oco para permitir a deformação elástica do mesmo. A deformação elástica dos análogos modificados é detectada pelos extensômetros elétricos de resistência – *strain gauges,* que traduzem essas deformações elásticas em leituras de microdeformação.

Figura 9. Análogo modificado e análogo original.

As hastes dos análogos modificados foram jateadas com óxido de alumínio, com partículas de 100 µm e pressão de 0.55 MPa, antes da confecção dos modelos para proporcionar uma superfície levemente rugosa, mais favorável para a colagem dos *strain gauges.* Os modelos de trabalho para a análise de tensão foram confeccionados em gesso pedra tipo IV (Durone IV, Dentsply, Nova York, Estados Unidos). Para tal, foi obtido um molde do modelo mestre metálico em silicone de condensação (Zetalabor, Zhermack, Rovigo, Itália). Para a confecção do modelo, foi vertido o gesso no molde de silicone de condensação, manipulado à vácuo (Multivac 4, Degussa, Hanau, Alemanha) numa proporção de 50 g de gesso para 9,5 mL de água. Em seguida, o conjunto, constituído da infraestrutura parafusada ao análogo foi levado ao delineador, e os análogos imersos no gesso aproximadamente 10 mm (Figura 10). O modelo foi removido do molde de silicone após 45 minutos, período de cristalização do gesso, de acordo com as recomendações do fabricante.

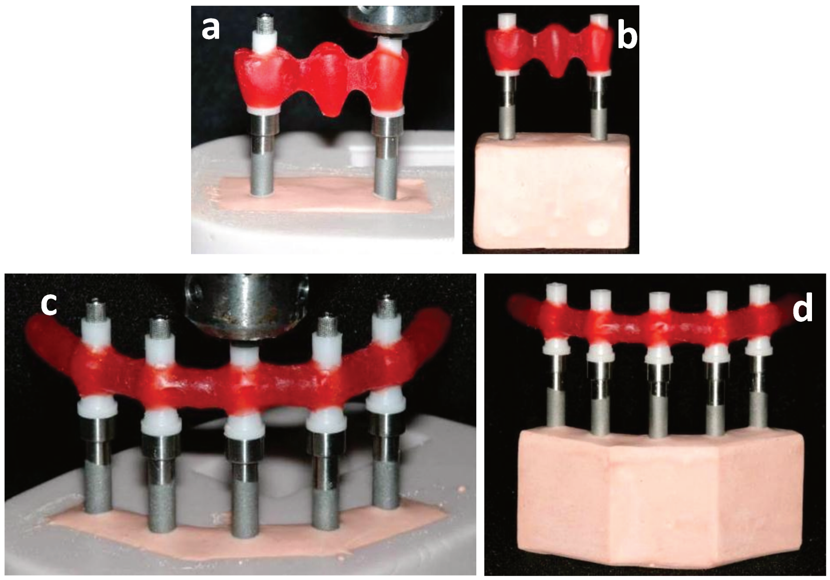

Figura 10. Confecção dos modelos para a análise de tensão. a, b) PPF; c, d) PTF.

Após a confecção do modelo para a análise de tensão, deu-se início a colagem dos *strain gauges* e a montagem dos fios, configurando um circuito montado em 1/4 de ponte de Wheatstone.

Para a realização da colagem do *strain gauge*, foi utilizada uma placa de vidro limpa com álcool isopropílico, na qual o *strain gauge* foi posicionado em uma fita adesiva. A haste do análogo modificado foi submetida à limpeza com álcool isopropílico, sendo então o *strain gauge* juntamente com a fita adesiva posicionados no análogo. Interposto a eles, uma gota de adesivo a base de cianocrilato (Loctite Super bonder, Henkel, Düsseldorf, Alemanha) foi dispensada a fim de realizar a colagem. Leve pressão foi aplicada sobre o *strain gauge*, com o auxílio de uma fita de teflon até a polimerização do adesivo. Após a polimerização, foi verificado se a colagem foi efetiva e, por fim, protegeu-se o *strain gauge* com resina de silicone (Figura 11).

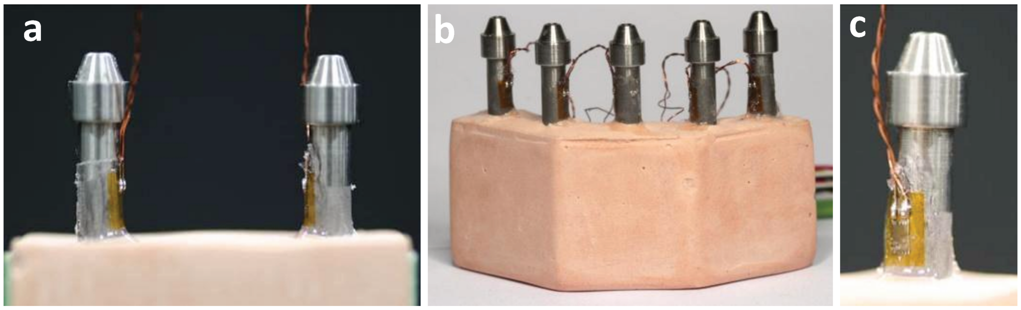

Figura 11. *Strain gauges* posicionados após sua colagem. a) PPF; b) PTF; c) Vista aproximada.

Para a montagem do circuito, os fios oriundos de cada *strain gauge* foram soldados aos terminais, de modo que fosse configurado 1/4 de ponte de Wheatstone (Figura 12). Os fios que saem dos terminais serão acoplados ao equipamento ADS 2000 (Lynx Tecnologia Eletrônica Ltda, São Paulo, SP, Brasil) no momento da análise de tensão.

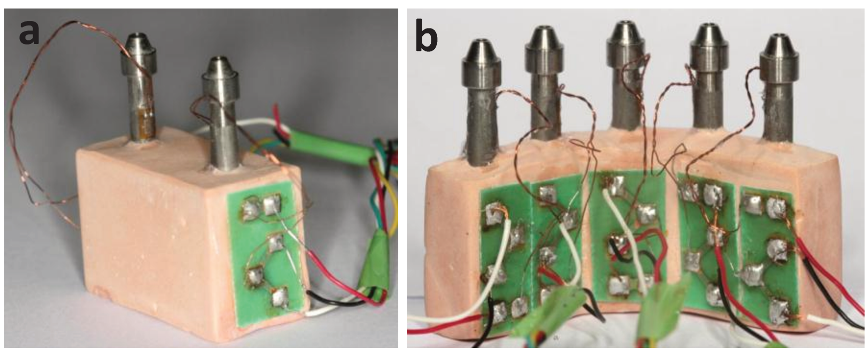

Figura 12. Montagem dos fios em 1/4 de ponte. a) PPF; b) PTF.

Depois de confeccionados todos os modelos do grupo I, as estruturas em Duralay II (Reliance Dental Mfg. Co, Worth, IL, Estados Unidos) foram enviadas ao laboratório protético Central do Titânio (São Paulo, SP, Brasil) para iniciar o processo de inclusão e fundição em titânio comercialmente puro grau I (Tritan, Dentaurum, Pforzeim, Alemanha).

Infraestruturas Fundidas

Após a fundição (Figura 13) realizada pelo laboratório, juntamente com o acabamento e polimento das infraestruturas, foi possível iniciar as análises do grupo I (torque de afrouxamento, desajuste marginal e tensão) e posteriormente realizar a confecção dos modelos dos outros grupos, II e III, para o design da PPF e então, prosseguir com as análises. Ainda foram realizadas tomadas radiográficas (90 KV, 15 mA, 0,6 segundos, à uma distância de 10 a 13 mm) de todas as infraestruturas, a fim de analisar possíveis porosidades (Zavanelli & Henriques, 2001). As análises das infraestruturas da PTF serão realizadas depois de finalizadas as análises da PPF.

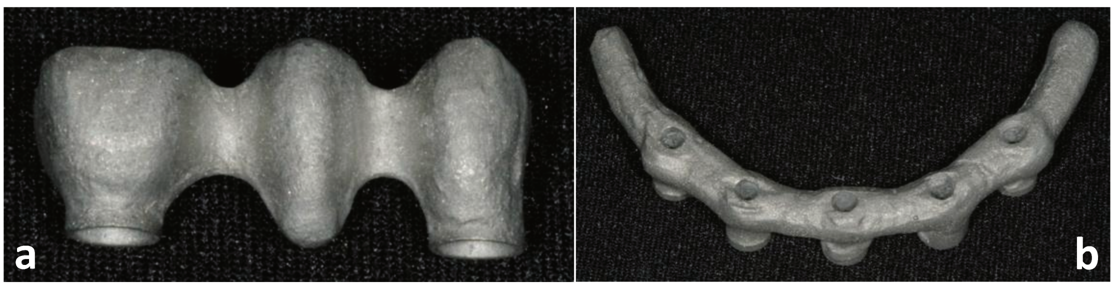

Figura 13. Infraestruturas fundidas. a) PPF; b) PTF.

Torque de Afrouxamento

Para a análise de torque de afrouxamento, realizou-se primeiramente a limpeza dos parafusos protéticos, parafusos de *mini abutment* e análogos de implante com álcool isopropílico. A força necessária para o torque de afrouxamento dos parafusos protéticos foi medida por meio de um torquímetro digital de precisão (Torque Meter TQ-8800; Lutron, Taipei, Taiwan) acoplado a um dispositivo (Figura 14), que permitiu o seu posicionamento vertical, de modo a não induzir tensões laterais.

Inicialmente foi aplicado o torque de 20 Ncm nos parafusos de *mini abutment* e de 10 Ncm nos parafusos protéticos. A sequência de aperto para a PPF foi primeiramente o pilar A e em seguida o pilar B.

A força do torque de afrouxamento inicial foi aferida após 24 horas tanto nos parafusos protéticos quanto nos parafusos de *mini abutment* na mesma sequência do aperto. Os parafusos protéticos foram utilizados uma única vez para determinação da força de torque de afrouxamento.

A força de afrouxamento final seguiu os mesmos parâmetros da análise inicial, sendo realizada após a ciclagem mecânica.

Figura 14. Torquímetro digital acoplado no dispositivo.

Leitura do Desajuste Marginal

Os desajustes marginais verticais foram avaliados tanto no modelo de trabalho, em resina acrílica, quanto no modelo de tensão, em gesso, por meio do teste do parafuso único (Sartori *et al.*, 2004; Tramontino, 2008; Spazzin *et al,.* 2011), mensurados por visualização direta em aumento de 120 vezes em microscópio de medição com precisão de 1,0 µm (UHL VMM-100-BT; Reino Unido), equipado com câmera digital (KC-512NT; Kodo BR Eletrônica Ltda, São Paulo, SP) e unidade analisadora (QC 220-HH Quadra-Check 200; Metronics Inc., Bedford, Estados Unidos). As leituras foram realizadas por um mesmo avaliador, em um ponto central na face vestibular e na face lingual, nas interfaces *mini abutment* e infraestrutura.

Para a leitura dos modelos de PPF, quando parafusado o pilar A, a leitura era realizada tanto na face vestibular quanto na face lingual do pilar B, e quando parafusado o pilar B, a leitura era realizada tanto na vestibular quanto na lingual do pilar A.

As leituras de desajuste foram realizadas sempre antes e após a ciclagem mecânica, em todos os grupos.

Análise de tensão

As análises de tensão foram realizadas em cada modelo de gesso especial (Durone*,* Dentsply*;* Rio de Janeiro, Petrópolis, Brazil) antes e após a ciclagem mecânica, em todos os grupos. A medição da tensão foi realizada por meio de dois extensômetros elétricos de resistência (PA-06-060BG-350L, Excel Engenharia de Sensores; Embu, São Paulo), posicionados diretamente em cada análogo modificado do pilar *mini abutment* (Figuras 11 e 12).

O arranjo dos extensômetros em cada fixação obedeceu à formação de 1/4 de ponte de Wheatstone, permitindo a captação das deformações, por meio de sinais elétricos em um aparelho controlado por computador (ADS2000; Lynx Tecnologia Eletrônica Ltda, São Paulo, SP) (Figura 15), e processado por um software específico (AqDados 7; Lynx Tecnologia Eletrônica Ltda, São Paulo, SP).

As infraestruturas foram adaptadas conforme recomendações do fabricante, por retenção aos análogos dos pilares modificados do tipo *mini abutment* mediante torque de 10 Ncm, nos parafusos protéticos correspondentes. A sequência de aperto dos parafusos foi realizada apertando primeiro o parafuso no pilar A e em seguida no pilar B. Ao final foram obtidas médias de tensões para cada infraestrutura analisada.

Figura 15. ADS2000 (Lynx Tecnologia Eletrônica Ltda) acoplado a um computador.

Ciclagem Mecânica

Após as análises iniciais de cada grupo, os modelos foram submetidos a 10^6 ciclos mecânicos, realizados com 2 Hz de frequência, com inclinação de 30º (ISO 14801 - 2007) em Simulador de Fadiga Mecânica (ERIOS, modelo ER – 11000 – Plus) (Figura 16 a). Os parâmetros utilizados foram estimados a partir da pressuposição de que um indivíduo realiza três episódios de mastigação por dia, cada um com duração de 15 minutos, que seriam equivalentes a 2.700 ciclos por dia e aproximadamente 10^6 ciclos por ano (Wiskott *et al.*, 1995). A simulação de fadiga mecânica será realizada com pressão de 6 bar correspondentes a aproximadamente 280 N de carga compressiva (Mericske-Stern & Zarb, 1996), aplicada sobre a região distal das infraestruturas (Figura 16 b).

Figura 16. a) Simulador de fadiga mecânica (ERIOS, modelo ER – 11000 – Plus); b) Modelo de trabalho imerso em saliva artificial sob simulação mastigatória.

Depois de realizadas as análises iniciais do grupo I, os modelos do grupo II foram confeccionados (Figuras 17 e 18), seguindo o mesmo padrão de confecção do grupo I, consistindo a diferença no fato desses modelos serem confeccionados com as estruturas já fundidas, promovendo assim uma justaposição entre as infraestruturas e os *mini abutments*. Depois dos modelos confeccionados, foram realizadas as análises iniciais, conforme descrito anteriormente.

Figura 17. Confecção dos modelos do grupo II para a Prótese Parcial Fixa. a) Modelo de resina; b) Modelo de tensão.

Figura 18. Confecção dos modelos do grupo II para a Prótese Parcial Fixa. a) Modelo de resina; b) Modelo de tensão.

O mesmo se aplica à confecção dos modelos do grupo III, sendo que para isso foi utilizada uma placa metálica perfurada, na qual os análogos foram parafusados para a obtenção do desajuste de 150 µm (Figura 19).

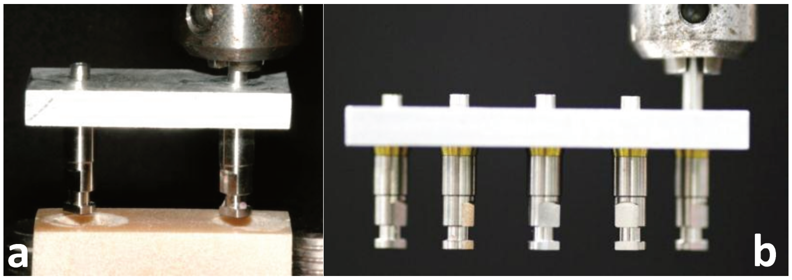

Figura 19. Confecção dos modelos do grupo III. a) PPF; b) PTF.

Após as análises iniciais, as infraestruturas foram submetidas à ciclagem mecânica. Assim que os ciclos mecânicos finalizaram, foram realizadas as análises finais de torque de afrouxamento, desajuste vertical e tensão.

Anexo 1

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Anexo 2

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