A Comparison of Fit of CNC-Milled Titanium and Zirconia Frameworks to Implants

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ABSTRACT

Background: Computer numeric controlled (CNC) milling was proven to be predictable method to fabricate accurately fitting implant titanium frameworks. However, no data are available regarding the fit of CNC-milled implant zirconia frameworks.

Purpose: To compare the precision of fit of implant frameworks milled from titanium and zirconia and relate it to peri-implant strain development after framework fixation.

Materials and Methods: A partially edentulous epoxy resin models received two Branemark implants in the areas of the lower left second premolar and second molar. From this model, 10 identical frameworks were fabricated by mean of CNC milling. Half of them were made from titanium and the other half from zirconia. Strain gauges were mounted close to the implants to qualitatively and quantitatively assess strain development as a result of framework fitting. In addition, the fit of the framework implant interface was measured using an optical microscope, when only one screw was tightened (passive fit) and when all screws were tightened (vertical fit). The data was statistically analyzed using the Mann–Whitney test.

Results: All frameworks produced measurable amounts of peri-implant strain. The zirconia frameworks produced significantly less strain than titanium. Combining the qualitative and quantitative information indicates that the implants were under vertical displacement rather than horizontal. The vertical fit was similar for zirconia (3.7 μm) and titanium (3.6 μm) frameworks; however, the zirconia frameworks exhibited a significantly finer passive fit (5.5 μm) than titanium frameworks (13.6 μm).

Conclusions: CNC milling produced zirconia and titanium frameworks with high accuracy. The difference between the two materials in terms of fit is expected to be of minimal clinical significance. The strain developed around the implants was more related to the framework fit rather than framework material.

KEY WORDS: distortion, framework fit, in vitro study, misfit, screw-retention, strain gauge

INTRODUCTION

Despite the impressive performance of oral implant prostheses in the management of edentulous and partially edentulous patients, they still suffer mechanical and biological complications in certain situations.[1] One of the mechanical factors that was expected to contribute to the level of complications of an implant prosthesis is the passive fit of the prosthesis framework.[1,2] Many authors have tried to define passive fit or the acceptable fit of implant frameworks. Branemark[2] considered the framework to be passively fitting if the gap between the framework and the abutment is 10 μm or less. Jemt[3] suggested that misfit should be smaller than 150 μm for the framework to be acceptable. Other authors suggested no strain development due to framework fixation.[4]

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DOI 10.1111/j.1708-8208.2010.00334.x

However, a genuine definition of passive fit from a biomechanical perspective is lacking. Based on this, many authors have argued against the significance of passive fit and concluded that well-controlled conventional crown and bridge techniques are adequate in providing long-term successful implant treatment.[4,5] Nevertheless, until clear guidelines are presented regarding the acceptable level of fit and the method to measure it, it is crucial to aim for the best framework fit possible to minimize the strain and gap developments.

Because every step of implant framework fabrication introduce certain degrees of inaccuracies, one of the methods to enhance the precision is omitting some of the fabrication steps and utilization of a well-developed industrialized engineering approach by means of Computer-Aided Design and Computer-Aided Manufacturing (CAD/CAM). Computer Numeric Controlled (CNC)-milled titanium framework (Nobel Biocare, Yorba Linda, CA, USA) was shown to exhibit better fit than cast gold[6,7] and silver palladium frameworks.[8]

In an aesthetically conscious society, there is continuous demand for highly aesthetic restorations. Recently, CNC-milled, screw-retained zirconia frameworks were proposed as an aesthetic alternative by the same manufacturer. Zirconia has the advantages of being aesthetic, highly biocompatible, unsupportive of plaque accumulation, and with superior mechanical properties. Those advantages have led to an exponential increase in zirconia as a material for dental prostheses.[9] Zirconia has a flexural strength of 900–1,400 MPa and a fracture toughness of up to 10 MPa/m$^{0.5}$. Such superior mechanical properties have increased the confidence and the scope of ceramic application in dentistry to include anterior and posterior fixed partial dentures.

Despite the general similarity of zirconia milling to titanium, there are significant differences in the fabrication procedure. The zirconia available for implant frameworks can be milled by "soft" machining, where oversized milling of pre-sintered restoration to the required design is followed by sintering at high temperature. The sintering procedure leads to about 25% shrinkage that has to be compensated for by the milling procedure.[9] However, the extent of shrinkage exerts an extra challenge to the software which has to be reliable to enable framework design that will shrink precisely to the required dimension.

The other significant difference between the two materials is the elastic modulus. Zirconia has an elastic modulus of more than twice that of titanium. This means that for the same level of accuracy, there is a greater possibility for this stiff material to induce greater peri-implant strain after framework fixation.

To date, there is no study, to our knowledge, assessing the fit of milled implant zirconia frameworks. The purpose of this study was to measure and compare the precision of fit of implant fixed partial denture frameworks fabricated by CNC-milled titanium and CNC-milled zirconia, and to investigate the implication of misfit in terms of strain development. The hypothesis tested was that CNC milling of titanium and zirconia provides a similar level of framework fit.

MATERIALS AND METHODS

Master Model Fabrication

With the aid of a silicone mold, an artificial partially edentulous mandible was fabricated using epoxy resin material (Masterflow 622, Heavy Duty Epoxy Resin Grout, Degussa, Seven Hills, NSW, Australia) that exhibits an elastic modulus similar to that of cortical bone (15.0 GPa). This assumption was based on an earlier investigation conclusion that an entirely isotropic mandible model, with an elastic modulus of the cortical bone is a close representative of natural human mandible.[10] Because the epoxy resin is isotropic and the modulus of elasticity is close to the cortical bone, it was assumed that the ridge quality would be equivalent to a Lekholm–Zarb type I quality classification.[11] After mixing the epoxy resin according to manufacturer's instructions and pouring it into the silicone mold, a waiting time of 7 days was allowed for the resin to fully cure. The edentulous area was sectioned to facilitate future use of the mandible as a definitive master model for framework fabrication.

Two Branemark implants (Mk III TiUnite, Nobel Biocare AB, Göteborg, Sweden) with a length of 11.5 mm and a diameter of 4.0 mm were inserted in the edentulous area in the positions of lower left second premolar, implant A, and lower left second molar, implant B. The external hex implant system was preferred because it allows the direct measurements of the vertical gap without the possibility of binding with the internal surfaces like the case for conical abutments. Therefore, only the vertical inaccuracies were recorded. The implant holes were drilled in clinically favorable positions and were slightly larger than the diameter of the implants. To ensure optimal parallelism, the drilling

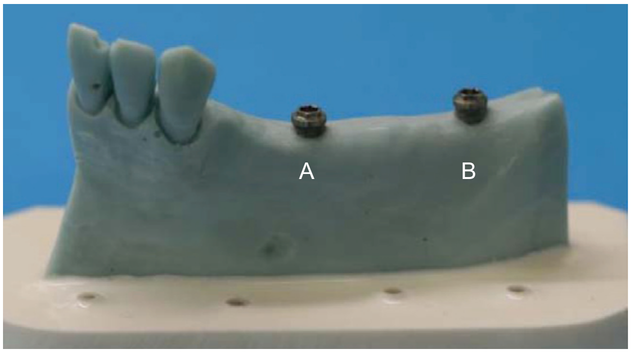

Figure 1 The mandible master model illustrating the location of implant A and implant B.

was performed using a milling system attached to a surveyor table (Amann Girrbach AG, Austria). The two implants were fixed in the mandible with freshly mixed epoxy resin material (Masterflow 622, Degussa), and the model was left for a further 7 days. This ensured a firm bond between the implants and the mandible without stressing the peri-implant structures. A self-cured epoxy resin base (Easycast®, Barnes Products, Bankstown, NSW, Australia) was then fabricated under the mandible to facilitate the handling of the master model (Figure 1).

Framework Fabrication

Implant level plastic copings (Nobel Biocare AB) were adapted over the implants and used for acrylic resin framework (Pattern Resin LS, GC America Inc., Alsip, USA) fabrication by an experienced dental technician. The design of the framework was a simple beam with a cross-sectional dimension of 4.0 × 4.0 mm.

Following single scanning procedures of the framework and the master model by a Procera touch scanner (Forte, Nobel Biocare AB), 10 identical frameworks were fabricated by the same manufacturer (Nobel Biocare, Tokyo, Japan). Half were made from grade 2 commercially pure titanium. The rest of the frameworks were fabricated from partially sintered zirconia material. The zirconia milling was performed at the pre-sintered stage and at an enlarged dimension. This step was followed by sintering the frameworks that led to subsequent shrinkage (Figure 2)

After receipt from the manufacturer, the frameworks were not modified in any way. All the frameworks were subjected to two different experimental tests: strain gauge analysis and gap measurements. To ensure the repeatability of the results, all the measurements were performed by one operator.

Strain Gauge Placement

Four linear strain gauges (CEA-06-015UW-120, Vishay Micromeasurements Group, Shelton, CT, USA; resistance 120.0 ± 0.2% Ω; gauge factor: 2.07 ± 0.1%; gauge length: 0.38 mm) were bonded with cyanoacrylate resin (Master Bond, Shanghai, China) on the crestal region around each implant, on the mesial, distal, buccal, and lingual aspect. The strain gauge foils were oriented radial to the implant as illustrated in Figure 3. The rationale for the location of the gauges is that peak strains around implants are known to occur in the crestal bone around the osseointegrated implants.[12] A Wheatstone bridge electrical circuit arrangement provided by PowerLab data acquisition system (Sydney, NSW, Australia) was used to record the resistance change in microvolt (μV) across the strain gauges that can be converted subsequently to microstrain (με).

Strain Qualification

Strain qualification was performed as described in previous investigation by the authors.[13] In summary, the qualitative step was applied to simplify the model by recording the compression or tension response of each strain gauge to constant forces, which can subsequently assist with the interpretation of deformation around the implants that occurs as a result of attaching the framework to the implants. A force magnitude of 30 N was

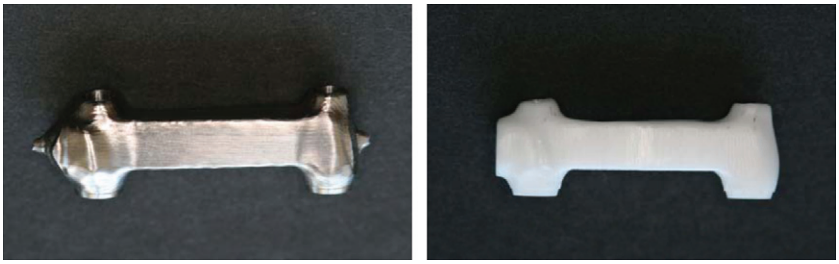

Figure 2 The final titanium and zirconia frameworks used for this study.

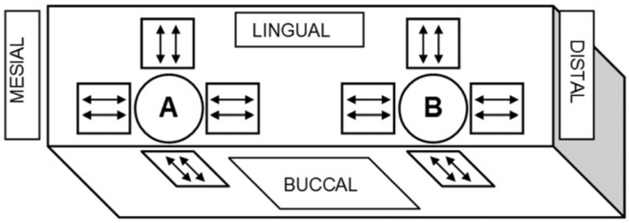

Figure 3 Diagram representing the location of strain gauges. The arrows indicate the direction of strain measured by the gauges.

arbitrarily selected to standardize the strain gauges response and was controlled by Correx tension gauge (Haag-Streit AG, Koeniz, Switzerland). On each implant, a force of 30 N in eight horizontal directions was applied; the mesio–distal axis; bucco–lingual axis, and diagonally. Further, the strain pattern was verified by vertically pulling and compressing the implants. In addition, the strain pattern was verified by tightening the retaining screws to each abutment to a torque of 35 Ncm using a manual torque controller (Nobel Biocare AB).

All the values for the strain gauges were color coded in figures. The strain gauges showing positive readings (tension) were color coded with blue, and the strain gauges with negative readings (compression) were color coded red (Figure 4A). The intensity of the response was further arbitrarily classified into maximal (above 100 με), moderate (50–100 με), minimal (less than 50 με), or nil (0 με). The color intensity was changed in relation to changes in the response intensity for every diagram specific to force direction (Figure 4B).

Strain Quantification

The protocol for tightening involved torqueing the retaining screw on implant A until finger pressure resistance was met, followed by tightening the screw into implant B until finger pressure resistance was met. A torque of 35 Ncm was then applied first on implant A then on implant B using a manual torque controller (Nobel Biocare AB). For each framework, a new set of retaining screws were used. The readings were simultaneously monitored during the torqueing procedure for 1 minute. Prolonged connection was avoided to prevent overheating of the strain gauges. To standardize the readings, the strain values were zeroed before framework fitting.

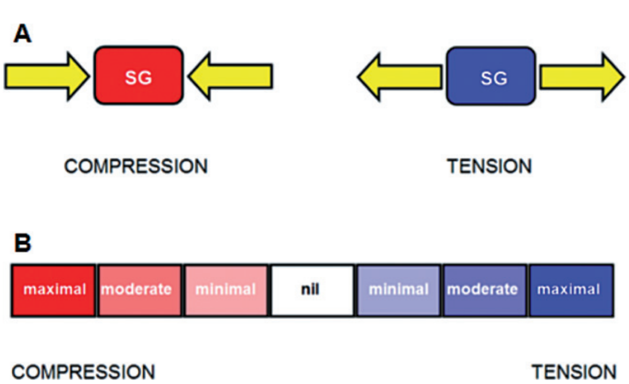

Figure 4 Color coding for the compressive and tensile response of the strain gauge (SG) (A). The scale for the different intensity of strain (B).

Gap Measurements

A Branemark implant level impression coping was adapted over each implant and connected with light cured resin verification jig (Megatray, Megadenta Dentalprodukte, Radeberg, Germany). After curing, the jig was sectioned centrally and reconnected with self-curing resin material (Pattern Resin LS, GC America Inc., Alsip, IL, USA). After 24 hours, implant replicas (Nobel Biocare AB) were attached to the impression copings of the verification jig and type IV dental stone (Fuji Rock, GC Corporation, Tokyo, Japan) was poured around the replicas. The purpose of this model was to facilitate specimen placement under the microscope and subsequent measurement of the vertical gap between the implants and the frameworks.

The fit of every framework was assessed for two conditions: vertical fit and passive fit. The passive fit is the vertical gap that is formed on the non-tightened implant as a result of manual tightening of the retaining screw on the other. The tightening procedure was performed till the first fixation of the screw was felt, as described for the Sheffield test.[14] The vertical fit is the vertical gap that is formed between the implants and the frameworks as a result of tightening both retaining screws.[15] The purpose of measuring the vertical fit and passive fit of the frameworks is to assess the vertical gap reduction as a result of retaining screw tightening.

The gaps were measured microscopically with a Nikon Measurescope (Nippon, Kogaku, Japan) at 50× magnification and 1 μm of measurement accuracy. For each condition, the gaps were measured on four locations around each implant: mesial, buccal, distal, and lingual. To ensure the precision of gap measurements, three readings were obtained for each location. A model base was made from gypsum and designed at an angle to allow for readings of the proximal surfaces of

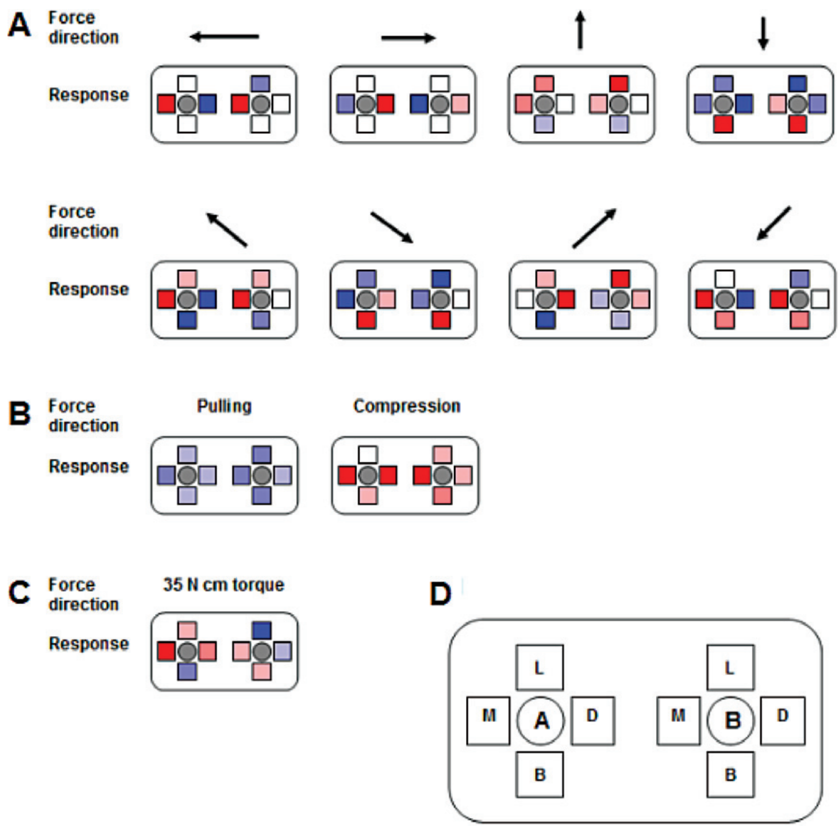

Figure 5 The response of all the strain gauges to the horizontal force (A) and vertical force (B) application. The response of strain gauges after torqueing (C). The location of strain gauges: M, mesial; B, buccal; D, distal; L, lingual (D).

each implant (Figure 5). With a sharp scalpel the implant replica was marked on each of the measurement points.

The final peri-implant strain values were plotted against the passive and vertical fit gap values. In order to determine the presence of a relationship, the correlation coefficient (R^2) was determined.

Statistics

Mean values were obtained for each strain gauge and gap measurement for every framework material. All the statistical analyses were performed by statistical program (SPSS for Windows, version 10, SPSS Inc., Chicago, IL, USA). The Mann–Whitney test was used to verify the effect of framework material on the strain and gap values for every condition. Because of the small sample size number, the level of statistical significance was set at $p < .1$.

RESULTS

Strain Qualification

Most of the strain gauges responded to the forces, irrespective of the orientation of the force applied. Some surfaces experienced more strain than others when loaded to the same force magnitude (Figure 5A). In general, the strain gauges aligned parallel to the force direction recorded higher strain values than strain gauges aligned perpendicular to the applied force. The majority of strain gauges showed results as expected in relation to the direction of the force applied and few had insignificant readings. After application of compressive vertical forces on the implants, all the strain gauges showed compressive activities and vice versa (Figure 5B). As the retaining screws were tightened without the framework in place, the strain gauges showed mixed activities (Figure 5C).

Strain Quantification

During the tightening procedure, the strains initially fluctuated and then reached a plateau as the system settled. A trend line was added to the mean strain graphs (Figure 6) to help visualize the effect of the different framework material on the pattern of deformation of peri-implant structures for each implant.

Around the two implants, the trend lines showed similar deformation patterns for the two materials,

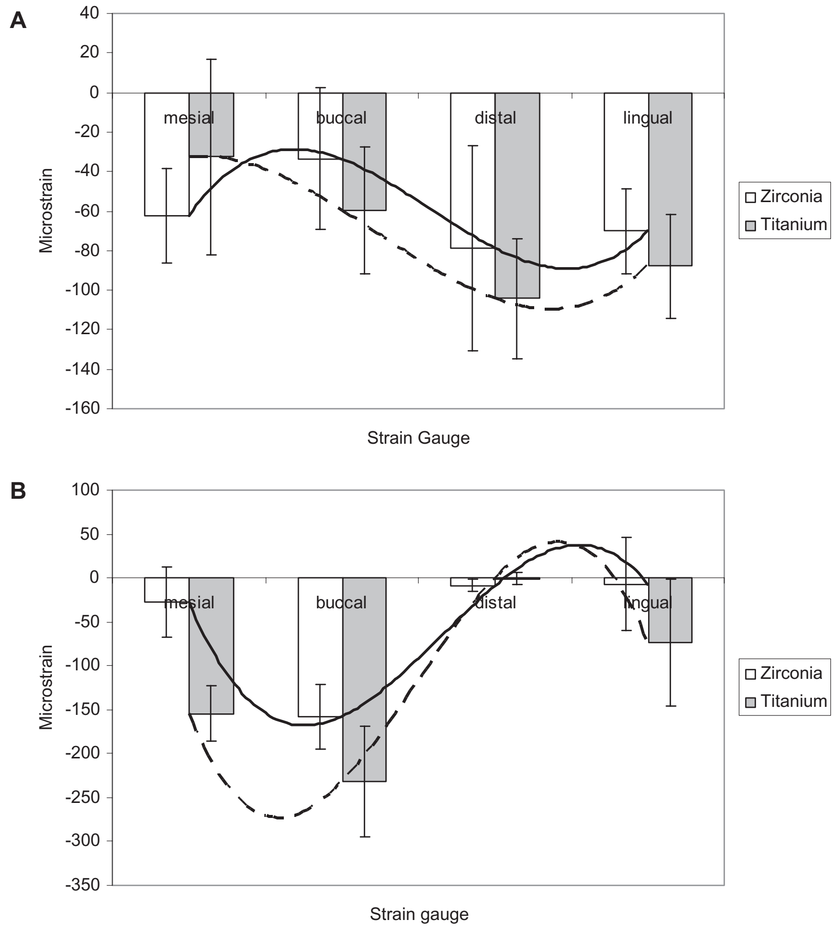

Figure 6 The mean and standard deviation of strain values at the various strain gauge positions about implant A (A) and implant B (B).

however, with greater magnitude of strain for titanium (Figure 6, A and B). The only significant statistical difference in strain magnitude was from the mesial ($p = .009$) and lingual ($p = .076$) strain gauges about implant B. For the overall peri-implant strain magnitude, the zirconia frameworks produced significantly less deformation than titanium frameworks ($p = .009$).

Qualitative and Quantitative Comparison

After combining the quantitative data with the previously obtained qualitative data, the two materials showed a high degree of similarity in relation to the direction of deformation recorded by all the strain gauges (Figure 7).

Gap Measurements

The results are summarized in Table 1. In relation to the vertical fit (Figure 8A), there was no significant difference between the two materials for each location or the overall vertical gap. After passive fit gap measurements (Figure 8B), the titanium frameworks on implant A exhibited significantly greater gap than zirconia frameworks ($p = .008$). On implant B, there was no significant difference. For overall passive fit, the zirconia frameworks had a significantly superior fit ($p = .008$).

Relation between the Strain and the Passive Fit

After comparing the passive fit gap measurements and the final strain values for all the frameworks, it appears

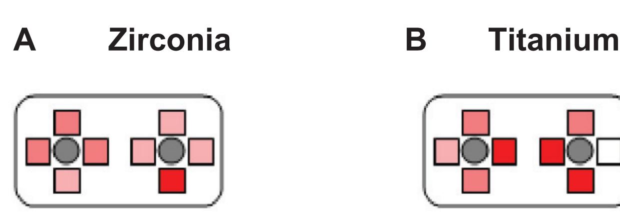

Figure 7 Strain gauge response after fitting the zirconia (A) and titanium (B) frameworks.

TABLE 1 The Mean Vertical Gap and Standard Deviation for Titanium and Zirconia at Different Screwing Conditions (μm)

Screwing Condition	Titanium Mean (SD)	Zirconia Mean (SD)	*p* Value
Vertical fit	3.6 (0.9)	3.7 (1.1)	0.829
Passive fit	13.6 (10.1)	5.5 (2.1)	0.009

there was a tendency for greater peri-implant strains with greater vertical gap ($R^2 = .7$). This relationship was more evident for titanium than for zirconia (Figure 9).

DISCUSSION

For tooth-supported frameworks, in vitro studies have failed to show the superiority of fit of the CAD/CAM produced copings compared with conventional cast copings. This is true for milled titanium[16] and zirconia frameworks.[17] On the contrary, milled titanium implant frameworks have been shown to exhibit a superior fit to conventional cast frameworks.[6–8] Our results confirm that a highly accurate fit is possible with CNC-milled titanium frameworks and provide initial results regarding the fit of implant zirconia frameworks. This outcome can be attributed to the precision possible with this controlled industrialized procedure and the exclusion of several error-introducing steps such as waxing, investing, and casting.[7,18]

In relation to the implant-framework interface, the CNC milling procedure has the advantage of reproducing the dimension of machined implant components as it is an integral part of the CAD software. This alleviates the risk of compromising the tolerance of the implant-framework interface by relying on the scanning system. Our results indicate that the interface between the zirconia or titanium framework and implant platform was 3.7 μm and 3.6 μm, respectively, which is comparable to the implant-abutment interface for single implant crown situation.[19]

According to gap-related passive fit criteria provided by Branemark[2] and Jemt,[3] the CNC-milled titanium and zirconia frameworks assessed in the present study exhibited passive fit. However, following the strain-related passive fit criteria,[4,5] all the frameworks induced measurable strain values, hence, using this criteria; none of the frameworks exhibited passive fit. This is in accordance with all previous strain gauge analysis investigations.[4,20] Therefore, it seems that peri-implant strain development is an inevitable consequence of

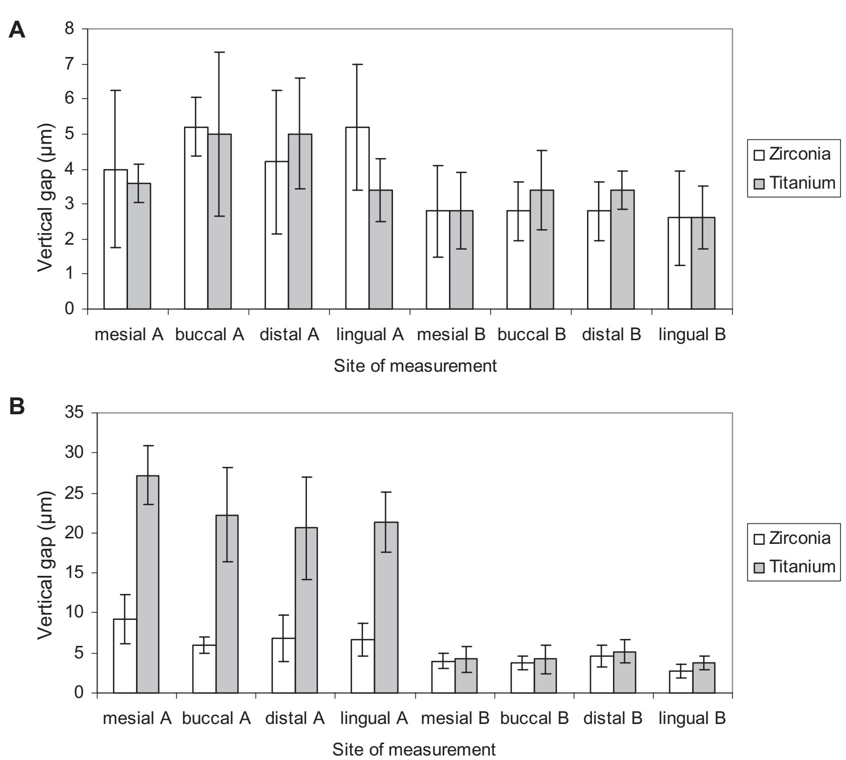

Figure 8 The mean vertical gap and standard deviation in the vertical (A) and passive (B) fit conditions.

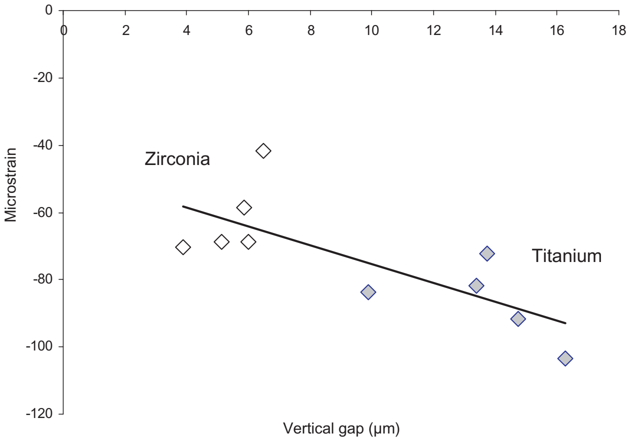

Figure 9 The relation between peri-implant strain and passive fit gap values.

tightening the retaining screws even for good fitting frameworks. Based on this, our study suggests reviewing the definition of strain-related passive fit and reinforces other conclusions that a true definition of passive fit from the biomechanical perspective is needed.[4,5] After combining the qualitative and quantitative information, it was evident that all the strain gauges showed compressive responses which means the strain direction of one strain gauge was not opposed by inverse activities from the opposing strain gauge. This indicated that the implants were primarily under vertical displacement. Therefore, it can be speculated that from an experimental context, the strain pattern is a better indicator of the degree of fit rather than the strain magnitude.

From our strain and gap results, it can be speculated that CNC-milled zirconia frameworks exhibited a statistically significant superior fit to CNC-milled titanium. Therefore, the hypothesis that CNC milling of the two materials provide similar level of fit was rejected. The slight distortion of the titanium frameworks can be explained from the engineering principle "machining-induced distortion,"[21] where the grinding process of the metal surface leaves significant work hardening-induced stress in the superficial layer of the metal. Titanium machining is typically carried out under high feed rates and high cutting speeds, and the milling edge of the titanium blanks suffers from low heat dissipation resulting in high cutting temperature.[22] These milling conditions exert thermal and mechanical loads on the workpiece, which stresses the subsurface material leading to distortion within the external layer.[21] In addition, titanium is relatively elastic and exhibits greater reversible deformation after cutting.[23] All these factors were proposed as explanations for the possible distortion of milled titanium. Nevertheless, the resulting distortion from milling titanium is likely to be less than the distortion from conventional casting procedures. This assumption is based on previous studies that compared the fit of conventional cast frameworks with CNC-milled titanium frameworks.[6–8]

The zirconia frameworks are subjected to fabrication-induced inaccuracies mainly resulting from approximately 25% sintering shrinkage.[9] Studies on tooth-supported zirconia frameworks found that milling zirconia in the post-sintered state is more predictable than the pre-sintered state.[24,25] However, in relation to implant framework, our results did not confirm the findings of these studies and showed that zirconia frameworks had a comparable vertical fit and superior passive fit to the titanium frameworks. This reflects the high accuracy achieved with CNC-milled zirconia and indicates efficiency of the available CAD system in compensating for sintering shrinkage. Because milling is performed at the pre-sintered state, minimal pressure and heat production is needed, reducing potential for distortion. From our data it can be suggested that milling zirconia in the pre-sintered state can lead to less distortion than milling titanium.

From a clinical perspective, the difference in strain and gap magnitudes is most likely of minimal clinical significance.[26] Despite the excellent fit of zirconia frameworks, more research is needed regarding the vulnerability of fit to several other factors, namely veneering, variation in span, and aging.[27] In addition, several long-term studies on zirconia fixed partial dentures have expressed concern regarding the chipping of veneering porcelain on zirconia frameworks.[9]

At the in vitro level, there are two concepts for measuring the fit of an implant framework: measurement of the exact dimensional distortion[6–8] or assessment of misfit-induced strains.[4,20] In this study, an attempt was made to combine the two concepts in order to determine the actual fit and the implication of any misfit within the peri-implant structures. After comparing the peri-implant strain values with the passive fit gap measurements, there was a direct relationship that indicated an increase in strain with greater passive fit gap values. These findings are in accordance with other studies that reported that prosthesis misfit would induce strain within the peri-implant structure.[20,28] The greater the misfit, the larger the magnitude of the forces as the framework screws are tightened[28] and consequently, the stress within the implant complex.[29] On the basis of a finite element study, when optimal fit was assumed, the stress was uniformly distributed in all components of the implant complex, producing minimal peak stress in each component.[30]

As the lower modulus of elasticity of titanium (two times lower than zirconia) did not lead to less peri-implant strain, it can then be assumed that the peri-implant strain is largely dependent on prosthesis fit rather than prosthesis material. However, the lower modulus of elasticity for titanium might be the reason for the greater reduction in the gaps for titanium than for the zirconia frameworks as the two retaining screws were tightened. Therefore, it is reasonable to state that a more accurate procedure is required for zirconia than titanium. This is even more important knowing that there are no corrective techniques available for zirconia frameworks compared with metal frameworks.

Strain gauge analysis has been used extensively in implant framework fit assessment.[4,20] It has the advantage of high sensitivity even to minimal force application. This means minor changes in tightening torque procedure may have a profound effect on the final strain reading. However, the efficiency of strain-related fit assessment is offset by the small sample number which might be the reason for the observed variation within the strain magnitude. To minimize this effect, every reading was repeated three times to ensure similar range of recording. Therefore, the variation obtained from each strain gauge reading is expected to be mainly due to minor variations in the fit of the frameworks.

Another limitation of the presented experimental set up is gap assessment on different model than the scanned model. The additional model was necessary to allow accurate positioning of the specimen on the microscope table. However, the minimal values obtained from gap measurements that fulfilled some of the passive fit definitions indicated that it did not create significant distortion. Optimal parallelism of implants and simple hypothetical framework design were chosen to reduce the variables that can affect the final strain development.

Because the experimental set-up omitted the process of impression taking and model pouring, it should be stated that greater inaccuracies might be expected clinically due to distortion from these omitted steps. However, with clinical verification procedures of the master model before framework fabrication, this effect can be minimized.

CONCLUSIONS

Within the limitations of the present study, the following can be concluded:

1. CNC milling is a predictable approach in providing accurately fitting frameworks with consistent dimensional values. The recorded inaccuracies are most likely of minimal clinical significance.
2. In relation to the gap values, titanium and zirconia CNC-milled frameworks fulfilled the criteria for passive fit. However, in relation to strain development, none of the examined frameworks exhibited absolute passive fit.
3. There are indications from the strain gauge analysis and gap measurements that zirconia frameworks have the tendency to exhibit superior fit than titanium frameworks.
4. Clinical studies are needed to disclose any difference in performance between the two implant framework materials.

ACKNOWLEDGMENTS

The first author would like to acknowledge the support of the University of Otago Research Committee for providing Postgraduate Publishing Bursary. This study has been partially funded by Nobel Biocare with research grant 2009-859. The authors would also like to thank Peter Fleury for electrical connection of the strain gauges.

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