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Effect of Loading Frequency on Cyclic Fatigue Lifetime of a Standard-Diameter Implant with an Internal Abutment Connection

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Abstract

Objectives: To investigate the effect of loading frequency on the fatigue lifetime of one standard-diameter titanium dental implant system.

Methods: Thirty-six titanium dental implant specimens (Bone Level RC, Straumann) were assembled following manufacturer's instructions and torqued into cylindrical holder blocks following the apparatus specified by the ISO 14801 test standard. Stainless steel loading hemispheres were bonded on the abutments with a moment arm of 11 mm. The holder blocks had layers of differing stiffness to simulate human jaw bone. Constant-stress fatigue lifetime testing was conducted at two frequencies (2 Hz and 15 Hz) with a stress ratio of 0.1 until fracture in deionized water at 37 °C on servo-hydraulic load frames (MTS). The fractured specimens were retrieved and examined using fractographic technique to determine the failure mode. The lifetime data were fit to a general log-linear regression model.

Results: The coefficient for the load amplitude term of the regression model indicated that increasing load amplitude had a statistically significant negative effect on the fatigue lifetime. The coefficients for the cyclic frequency term and the load-frequency interaction term were not significantly different from zero, which indicated that increasing loading frequency did not have an influence on the number of cycles to failure. Fractographic analysis showed that all specimens exhibited an identical combined fracture of abutment and abutment screw adjacent to the bone level.

Significance: Higher loading frequency at least up to 15 Hz may be used for future studies of some implant systems to improve the efficiency of fatigue testing.

1. Introduction

Over the past decade, root-form titanium dental implants have been widely used to restore missing teeth due to improvements in implant designs and understanding of their proper indications [1]. However, biological and technical complications of dental implants

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commonly occur, and this sometimes significantly compromises the clinical outcome of dental implant therapy and the quality of life for patients. Technical complications of dental implants include fracture or loosening of the abutment and abutment screw, implant body fracture, and superstructure fractures. The total long-term incidence rates of these technical complications have been reported to vary from 21% to 31% and is significantly affected by the implant design [1–5]. One systematic review [6] reported that the annual rate of total number of technical complications ranged from 3.5% to 15%, translating into a 5-year complication rate ranging from 16% to 53% over the last decade. While these complication rates can be considered generally high, it is also important to note that the follow-up terms of clinical studies are usually shorter than the expected service lifetime of dental implants, therefore the prevalence of these technical complications is likely to be underestimated by the existing literature.

In order to investigate the mechanism of these technical complications, researchers examined *in vivo* failed implants and their components using advanced microscopy techniques such as confocal laser scanning microscopy [7] and scanning electron microscopy (SEM) [8–11]. They determined that these implant failures had a metal fatigue mechanism, which was verified by the existence of fatigue striations in the fractured component. The rate of fatigue crack growth is usually quite slow under physiologic loads and normal chewing pattern, until one crack reaches critical size and consequently exhibits rapid catastrophic propagation. So far it is almost impossible to detect or monitor fatigue crack growth by routine clinical follow-up methods. This makes it important to conduct efficient and clinically relevant *in vitro* fatigue studies to evaluate the long-term fatigue behavior of dental implants, predict the fatigue lifetime under physiologic loading with less testing cost, and eventually improve the designs of implants and make them more fatigue-resistant.

To standardize cyclic fatigue testing of dental implants, the International Organization for Standardization (ISO) published standard 14801, which describes a method of cyclic fatigue testing for single post endosseous dental implants [12]. This standard requires conducting the test at a frequency no more than 2 Hz in liquid media or no more than 15 Hz in dry condition. Accelerated usage rate testing, such as cycling at 15 Hz, can significantly reduce the overall testing time and reduce cost, but there is a possibility that it may introduce other fatigue mechanisms that do not exist in the clinically relevant low-frequency situation. Unfortunately, there have been few studies about the effect of loading frequency on dental implant fatigue testing. This is likely due to the high experimental cost of performing fatigue tests multiple times at different frequencies. Two studies by the same research group [13, 14] compared cyclic fatigue testing at 2 Hz and 30 Hz for a commercially available dental implant system. They found that fatigue failure was the predominant crack propagation mode at 2 Hz but not at 30 Hz, which led to the conclusion that high frequencies such as 30 Hz may not provide clinically relevant data, since 2 Hz is thought to be similar to the normal human chewing frequency. However, ISO 14801 defined 15 Hz as the maximum frequency for fatigue testing, and it still remained unclear whether 15 Hz would have a significant influence on the fatigue lifetime similar to 30 Hz. If a frequency of 15 Hz can be used, then it would still accelerate tests by 7.5 times compared to testing at 2 Hz.

The objective of this study was to investigate the effect of two frequencies (2 Hz and 15 Hz) on the fatigue behavior of one commonly-used standard-sized titanium dental implant in deionized water. We hypothesized that there would be a significant difference in fatigue lifetime between specimens tested by 2 Hz and 15 Hz constant-stress tests. 15 Hz was chosen because it is the maximum value defined in ISO 14801. Deionized water of body temperature was used as testing medium for all the tests in this study to simulate the wet environment of oral cavity.

2. Materials and methods

2.1. Specimen preparation:

Thirty-six standard-diameter ($\varnothing = 4.1$ mm) titanium dental implants and matching anatomic internal abutments (Bone Level RC, Straumann, Switzerland) were assembled following the manufacturer's instructions, and they were placed into cylindrical holder blocks following the apparatus specified by ISO 14801. The insertion torque was 35 Non, which is the value recommended by the manufacturer. Stainless steel loading hemispheres (acorn nuts, Ace Hardware, Jackson, MS, USA) were bonded on top of the abutments using a cold-cure acrylic (QuickSet Acrylic, Allied High Tech Products, Rancho Dominguez, CA, USA) with a moment arm of 11 mm, and the implants were positioned in the simulated bone holder such that they were clamped 3.0 mm apical to the nominal bone level to simulate the worst case scenario with the respect to bone resorption. A custom designed specimen holder was used for this study to better simulate the mechanical properties and layered structure of human jaw bone. The holder blocks were composed of two layers with differing elastic moduli (shown in Figure 1). The block was made of a glass-fiber reinforced composite resin material (G10, Piedmont Plastics, NC, USA) with similar elastic modulus (16 GPa) to that of human cortical bone [15]. A cylindrical inner cavity was prepared using a drill press from the bottom of the holder block with 3 mm of thickness remaining on the top to simulate the cortical bone. Once the implant was placed into the holder, this cavity was fully filled with cold-cure acrylic (Quick Set Acrylic, Allied High Tech Products), which has a lower elastic modulus of 6 GPa and was considered to be similar to the value of human cancellous bone [15]. The elastic moduli for these materials and method of measurement were previously reported [16, 17]. Although they have differing elastic moduli, it should be noted that both materials used for this holder block met the requirement of ISO 14801, which is minimally 3 GPa for Young's modulus of elasticity.

2.2. Cyclic fatigue testing:

The specimens were then placed in custom fixtures to obtain a 30-degree angle of inclination between the direction of load application and the long axis of the implant. Constant-load amplitude fatigue lifetime testing was conducted at two frequencies (2 Hz and 15 Hz) with a stress ratio of 0.1 and a sine waveform until fracture. A more complex method of assigning specimens to load amplitudes was used in this study compared to that specified by ISO 14801. Although the ISO protocol is sufficient to simply determine the fatigue limit of an implant system, this study went beyond that goal and was aimed at determining the possible interaction of cyclic frequency with the load amplitude. After each specimen was fractured, the lifetime was added to the regression model, and the load amplitude for the next specimen

was assigned with the aim of achieving lifetimes ranging from 20,000 to 1,000,000 cycles to provide a broad range from which to more precisely estimate population parameters. This resulted in different proportion of specimens assigned to each load amplitude, as well as one different load amplitude, for the two cyclic frequencies. The load amplitudes chosen were 410 N (n=4), 450 N (n=6), and 480 N (n=8) for 2 Hz and 370 N (n=1), 410 N (n=9), and 480 N (n=8) for 15 Hz. All of the experiments were conducted in deionized water at 37 °C on servo-hydraulic load frames controlled by two digital controllers (FIextest 60 and Flextest GT, MTS, Eden Prairie, MN, USA). Figures illustrating the apparatus were previously published [Duan *et al.* Dent Mater (last article)]. The failure detector was set at 5 mm of downward displacement in the axial direction. The fractured specimens were carefully collected, cleaned and examined by fractographic technique using a high-resolution digital microscope (VHX-1000, Keyence, Japan) and field emission SEM (Supra 40, Carl Zeiss, Germany) to determine the failure mode.

2.3. Statistical analysis:

The lifetime data (λ) were fit to a general log-linear regression by using the following equation in accelerated lifetime testing and analysis software ALTA PRO (Reliasoft, Tucson, AZ, USA).

$$\lambda = \alpha_0 + \alpha_1 f + \alpha_2 \ln(\sigma) + \alpha_3 f \cdot \ln(\sigma) \quad (1)$$

Where f is cyclic frequency, σ is the load amplitude, and α_0 is the regression constant, α_1 , α_2 , and α_3 are regression coefficients which respectively represent the influence of frequency on lifetime, influence of load amplitude on lifetime, and the interactional relationship between frequency and load amplitude. 95% confidence intervals (95% CI) were also calculated for all the model coefficients. Plots of the number of cycles to failure versus load amplitude and their 95% CI lines were also graphed for each of the two cyclic frequency groups in ALTA PRO software (Reliasoft) after fitting a Weibull distribution and an inverse power law lifetime-load amplitude relationship (Equation 2) individually to the 2 Hz and 15 Hz data [18].

$$\ln(1 - P_f) = - (K\sigma^n \lambda)^\beta \quad (2)$$

where P_f is the probability of failure by a given lifetime, and K , n , and β are regression parameters. The β parameter represents the Weibull modulus.

3. Results

The parameters of best fit for the general log-linear model with 95% CI are listed in Table 1. Both coefficients α_1 and α_3 were not significantly different from zero because their 95% CI included zero. This indicated that increasing loading frequency did not have an influence on the number of cycles to failure. However, coefficient α_2 was significantly less than zero. This indicated that increasing load amplitude had a negative influence on the fatigue lifetime of tested implants. The plot of lifetime corresponding to 63.2% probability of failure versus

load amplitude for the two frequency groups is shown in Figure 2. Blue lines indicate the characteristic lifetime line (eta line) and 95% CI bounds of the 2 Hz group, while red lines indicate the eta line and 95% CI bounds of the 15 Hz group. The regression parameters for these plots are given in Table 2. The plot shows that the lifetime of implants decreased rapidly with an increase of load amplitude. The 95% CI lines for the two frequency groups overlap, which indicates that there is no significant difference between the fatigue lifetimes of these two groups (no significant effect of frequency). This result also agrees with the regression result of the general log-linear model.

Fractographic analysis showed that all specimens of both groups had a similar combined fracture of abutment and abutment screw, while the implant body remained intact. The location of the fracture was found to be close to the bone level. Figure 3 shows the optical microscope photos of a typical fracture surface. One side of the fracture surface has a relatively smooth and flat appearance, while the opposite side has a rougher look and forms a curved shape, which was identified as compression curl. The side opposite of the compression curl was the location of the fracture origin because the curl is formed due to the deflection of crack under compressive stress, and failure typically originates in the location of greatest tensile stress on the opposite side of the neutral axis from the compressive stress. Figure 4 shows the SEM images of one representative specimen. The failure was found to be initiated near the prefabricated chamfer on the wall of abutment. Fine ridged structures were observed on the fracture surface of the abutment close to the fracture origin. These structures were identified as typical fatigue striations for metal materials, which usually indicate the advancement of fatigue crack growth and are caused by the repeated iterations of cold working, brittle fracture, and ductile blunting of the crack tip [19]. In the fractured abutment screw and the compressive side of the fractured abutment, the surface showed a more porous appearance with many dimples. These structures form by microvoid coalescence and typically occur at higher crack growth rates. They are considered to be the signs of rapid catastrophic fracture in metallic materials [19].

4. Discussion

The results showed that the load amplitude had a negative influence on the lifetime as we expected. This also verified that an inverse power law can be used for modeling the relationship between the load amplitude and lifetime for titanium dental implants. Our results showed that there is no significant difference in fatigue lifetime between 2 Hz and 15 Hz loading under a wet testing condition, therefore the hypothesis was rejected. In a previous study, Lee *et al.* [13] tested four groups of commercially available dental implants at 2 Hz or 30 Hz at room temperature in air or saline (n=6). They found that fatigue failure was the predominant crack propagation mode at 2 Hz but not at 30 Hz. A subsequent study from the same research group [14] also investigated the effect of frequency (2 Hz and 30 Hz) and elastic modulus of testing block materials on implant failure at room temperature in air. The second study also found that fatigue failure was the predominant crack propagation mode at 2 Hz but not at 30 Hz. The reason for seemingly contradictory conclusions between these studies and our results could be explained by the effect of strain rate. It has been found that strain rates increase with the loading frequency over a large range for dental implant testing; for example, the same load amplitude produced 3047 microstrain/s at 2 Hz, 11224

microstrain/s at 15 Hz, and 20378 microstrain/s at 30 Hz [14]. Higher strain rate during implant testing can possibly introduce a different mechanism of the fatigue damage. For example, Chen *et al.* [20] did a study to investigate the effects of strain rate on the cyclic fatigue behavior of AZ31B magnesium alloy processed using surface mechanical attrition treatment. It was found that fatigue life increases with increasing strain rate, possibly owing to the fact that high strain rate favors twinning activity and restricts dislocation slip, thus enhancing the fatigue crack resistance and prolonging the fatigue life. This might explain the possible mechanism of this frequency effect at 30 Hz, which doubled the strain rate compared to 15 Hz. Further investigations would be beneficial to explore the mechanism of this effect.

Dry versus wet condition is another important factor for implant fatigue testing, but current reports are controversial regarding the influence of the environment on the cyclic fatigue testing. Lee *et al.* [13] reported that there was no influence of water on the failure probability of dental implants. However, Shemtov-Yona *et al.* [21] investigated the influence of aqueous environment mimicking intraoral conditions on the fatigue performance of standard-diameter implants. Their results showed that environmental conditions did adversely affect the fatigue performance. Lee *et al.* [13] also concluded that saline may enable chemically assisted crack growth during the stage of fatigue crack growth, although it did not influence the failure probability. Since there is no consistent conclusion regarding the effect of water on implant fatigue lifetime, it would be more clinically relevant to test the implants in water environment, thus this is the condition that we used in our study.

The fractographic features recorded the process of fatigue failure. The crack was initiated near the prefabricated chamfers on the wall of the abutment and slowly propagated at first. These structures are part of the self-guiding connection design of this implant system so that greater prosthetic flexibility and accurate abutment insertion can be obtained. But they also serve as stress raisers in this case, which emphasizes that structural integrity, stability, and clinical needs should be well-balanced in designing the connector geometry of implants. The advancement of fatigue cracks during numerous load cycles generated a similar sized region of fatigue striations in each specimen. This subcritical stage of crack propagation accounts for the greatest portion of the fatigue lifetime and can easily take millions of cycles without any damage being externally visible. Once the cracks reached a critical point, catastrophic failures suddenly occurred and fractured the central screw and the rest of the abutment. It is also worthwhile to mention that only upper structure fractures occurred for the tested implants. Surgical procedures are usually needed if the osseo-integrated implant body fractures. Conversely, an upper structure failure, such as the abutment and/or central screw failure in this case, is usually easier and less traumatic to correct by replacing the broken parts with new abutments and screws. Therefore, the fatigue failure pattern could also have a significant influence on the clinical outcome in addition to the fatigue lifetime of implants and should be given careful consideration during product design.

5. Conclusion

There was no significant difference in lifetime and failure mode between 2 Hz and 15 Hz groups for this implant system. The results suggested that a frequency of 15 Hz under wet

testing condition is appropriate to be used to improve the efficiency of implant fatigue testing.

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Highlights:

- There was no significant difference in lifetime and failure mode between 2 Hz and 15 Hz testing.
- A frequency of 15 Hz under wet testing condition is appropriate to improve the efficiency of testing.
- Only upper structure fractures (abutment and connector screw) occurred for the tested implants.

Figure 1.

Schematic design of the specimen holder block with a layered structure to simulate human jawbone. The outer layer is made of G10 composite material, and the inner layer is made of acrylic.

Figure 2.

Plots of number of cycles to failure versus load amplitude for a Weibull-inverse power law regression model individually fit to 2 Hz (blue) and 15 Hz (red) data. The straight interior lines represent the load amplitude corresponding to a 63.2% failure probability. The curved exterior lines are the 95% confidence intervals.

Figure 3.
Optical microscopy images of a fractured implant specimen. All specimens exhibited this same failure mode.

Figure 4. SEM images of a fractured implant specimen to show fatigue striations (top) and fast fracture microvoids (bottom) on the fracture surface. All specimens exhibited this same failure mode.

Table 1.

General log-linear model parameters with 95% confidence intervals. These parameters are required for Equation 1 to best fit the combined 2 Hz and 15 Hz implant lifetime data.

Parameter	Estimate
α_0	168 (67, 268)
α_1	-2.49 (-10.30, 5.32)
α_2	-25.4 (-41.8, -9.0)
α_3	0.39 (-0.88, 1.67)

Table 2.

Weibull-inverse power law model parameters with 95% confidence intervals. These parameters are required for Equation 2 to best fit the individual 2 Hz and 15 Hz implant lifetime data.

	2 Hz	15 Hz
β	0.732 (0.500, 1.07)	0.687 (0.468, 1.01)
$-\log K$	70.5 (33.7, 107)	56.9 (33.0, 80.7)
n	24.6 (10.8, 38.4)	19.6 (10.6, 28.7)