FATIGUE OF COMMERCIALLY PURE TITANIUM DENTAL IMPLANTS

Timotius Pasang[1], Maziar Ramezani[1*], Matthew Prygoski[2], Russell Wanhill[3],

Rohan Byrnes[4], Osamu Kamiya[5], Kiyoshi Tanaka[6]

[1]Dept. of Mechanical Engineering, Auckland University of Technology, New Zealand

[2]Dept. of Aerospace & Mechanical Engineering, University of Notre Dame, Indiana, USA

[3]National Aerospace Laboratory NLR, Emmeloord, The Netherlands

[4]Airframes & Engine Division, Platform Sciences Laboratory, Defence Science & Technology Organisation (DSTO), Melbourne, Australia

[5]Dept. of Mechanical Engineering, Akita University, Japan

[6]Dentistry Clinic, Akita University Hospital, Akita, Japan

*e-mail: maziar.ramezani@aut.ac.nz

Abstract. Failure of two commercially pure titanium dental implants was analysed. The implants fractured at the abutments after only three years use. One abutment failed at the junction of the threaded and unthreaded areas, the other failed within the threaded area. Both failures were by reversed bending fatigue, whereby fatigue cracks began from relatively coarse machining marks on the abutments. The final fracture areas were less than 10 % of the total fracture surfaces, indicating that the abutments had not been abnormally loaded. However, bone resorption would have increased the abutment bending stresses by causing an increased moment arm of the acting forces and misalignment of the abutments in the maxilla. The effect of misalignment was investigated by finite element modelling, which showed that even a slight misalignment could double the bending stresses. This unanticipated large effect, combined with the coarse machining marks, explains the premature failure of the implants.

1. Introduction

Titanium and titanium alloys have many applications as implant materials in the human body. These include dental implants (fixtures and abutments), bone plates and their screws, hip and knee joints. Titanium has gained widespread use in dentistry since the discovery of osseointegration by Brånemark [1, 2]. Commercially pure titanium (CP Ti) is one of the most often used materials for oral implants owing to its biocompatibility, moderate strength, and low density [3].

A single dental implant normally consists of (i) a fixture, (ii) an abutment, (iii) a gold crown or other superstructures, and (iv) a bridge screw, see Fig. 1. Other designs also have a gold cylinder and gold screw as well as the bridge screw. In the present case, described in section III of this paper, there were two implants, with a gold connecting bar attached to the two abutments, see Fig. 2. A full upper denture was attached to the resulting construction.

The various types of implant construction enable transferring forces from occlusion (biting, chewing, or clenching) to the bone/jaw [6]. The fixtures and abutments are normally made of CP titanium, the bridge screws are made of gold, and the superstructures are commonly made of platinum or gold alloys. The implants are intended for long-term use, but their design often

takes into account the possibility of failure, which then should occur in the bridge/gold screws, allowing easy retrieval of the remaining parts [7]. Other types of failure, especially in the abutments or fixtures, are to be avoided, since retrieval of the remaining parts from the jaw would require relatively lengthy surgical procedures.

Fig. 1. Photograph showing typical single dental implant components. More details of dental implant systems are summarised by Mollersten et al. [4] and Binon [5].
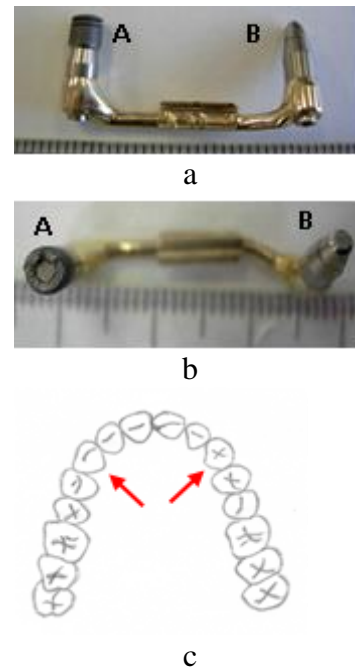
Fig. 2. Side (a) and top (b) views of the failed implants and (c) a schematic of the implant locations at the upper canines (cuspids). A full upper denture was attached to the implants. Note that part of the fixture for abutment A was present as well.

2. Implant Failures

When failures do occur, they are generally attributed to one, or a combination, of several factors including (i) poor bone quality (bone resorption/bone loss), (ii) insufficient bone volume, (iii) local infection, (iv) implant instability and unsatisfactory position of the implant, (v) unfavourable implant loading (excessive bending, overloading, clenching, etc), (vi) dimensions of the implants, and (vii) smoking habits [8-13]. These factors can be strongly interrelated.

2.1. Case Histories. Some illustrative case histories are listed in Table 1 and discussed below. Morgan et al. [14] examined five CP Ti fixtures that failed after 1-5 years of clinical service. All the failures occurred at the same heights as the ends of the abutment screws. These locations have less bending resistance because there are transitions from solid to annular cross-sections. The fixture fracture surfaces showed fatigue striations with 0.1-1.0 μm spacings, and laboratory fatigue tests on new fixtures showed very similar striations. As indicated in Table 1, the clinical fixture failures were caused by bone resorption and a stress concentration owing to the sharp root of one of the screw threads. Bone resorption was responsible for two effects. Firstly, the moment arms of any applied forces were lengthened; secondly, the annular cross-sections of the fixtures became exposed and unsupported. These effects increased the bending stresses during service.

In a similar study, Piatelli et al. [8] determined that fatigue resulted in failure of four CP Ti fixtures after only 1-3 years of clinical service. Three of the fixtures came from one patient,

whereby they had been placed in a straight line in the jaw. This arrangement is known to increase the bending moments on fixtures [6]. Additional causes of these three failures, and also the main causes of the fourth failure, were coronal bone resorption and bruxism. Piatelli et al. [8] pointed out that coronal bone resorption is particularly detrimental since it not only reduces the overall structural support but often exposes the annular cross-section of the fixtures where the abutment screws end. As pointed out above, these locations have less bending resistance.

Table 1. Some case histories of implant failures.

Implant component	Failure mechanism	Combinations of failure causes	References
Fixtures	bending fatigue	• bone resorption • sharp root of screw thread	[14, 23]*
		• multiple in-line implants • coronal bone resorption • bruxism	[8, 9, 15]
abutment screws and /or bridge screws	screw loosening or fractured	• inadequate pre-torque/pre-load • overload/excessive loading • bruxism • increased torsional loads due to non-optimum superstructure	[10, 13, 15, 16, 17, 23]
		• inadequate prosthesis & screw design • poor component fit • surface micro-roughness	[18, 19]
		• fistulas	[20]
		• micro-motion due to machining tolerances	[21]
		• fatigue • unfavourable orientation	[22]
		• hydrogen embrittlement • machining marks on screw thread roots	[24]

Hernandez et al. [9] conducted experimental and FE analyses for a case study failure in a Ti-6Al-4V dental implant. Based on their results, failure was promoted by bone re-sorption leading a cantilever condition resulting an overload system with cyclic high level stresses. This condition along with the rough surface finish found in the screw and the concentration factor by geometry change, caused a crack which was propagated until the failure of the component.

Shemtov-Yona and Rittel [10] developed a protocol for analyzing fractured implants and implant's parts. They mentioned in their protocol that the main failure mechanism for dental implants is metal fatigue. The overall fracture mechanisms that were identified on the retrieved Ti–6Al–4V and of CP-Ti dental implants were identical to those found on fatigue fracture surfaces of the specimens' fractured in lab conditions.

Although fatigue failures are important, Table 1 shows that there are other failure mechanisms. Abutment screw loosening is common, seen in up to 25 % of patients, and can also lead to premature failure of the screw, e.g. after only 9 months of implantation [13]. Loosening can be caused by inadequate pre-torque, overloading, bruxism, and any superstructure that unnecessarily magnifies torsional loads on the abutment [13].

Abutment screw loosening also depends on the implant design. Conical abutment connections actually resist loosening, since the moment needed to cause loosening is 10-20% higher than the tightening moment. The opposite trend is seen with other designs [21]. Many implant systems are designed with anti-rotational features to prevent loosening, but micro-motion due to machining tolerances can still compromise the pre-torque [21].

A third common failure location is the bridge/gold/occlusal screw joint. These failures are often due to overload, and are easier to remedy. However, Yokoyama et al. [24] found one case of a CP Ti occlusal screw that failed after only 3 years owing primarily to hydrogen embrittlement. Machining marks on the screw thread roots could have contributed to the failure by providing an increased stress concentration [24].

2.2. Bone Quality. Bone quality has also been implicated in implant failures. Adell et al. [25] proposed that the inadequate bone anatomy of the upper jaw and inferior maxillary bone density contribute to higher failure rates in maxillary implants than in mandibular implants. However, Hutton et al. [26] suggested that the poorer quality and quantity of maxillary bone was probably the main cause.

2.3. Remedial Measures. A few ways to reduce mechanical failure rates of dental implants are (i) extra care during placement, especially for maxillary implants, (ii) regular monitoring, (iii) correct tightening torque, (iv) increased implant diameter, (v) modifying the screw joint design, (vi) changing the materials, and (vii) surface finish modification [4, 7, 8, 12, 13, 23, 27]. In particular, smaller diameter implants tend to fracture earlier than larger ones [11, 29]. Hence, it has been suggested that implants should have the largest possible diameter (limited by the bone quantity) and be as long as possible to obtain deep purchase in the bone [5, 8]. These factors will increase resistance to failure from bending moments and occlusal forces [6, 11, 28, 29].

3. The Present Implant Failure: Background and Scope of the Investigation

Figure 2 shows the parts of the failed implants made available for investigation and their locations with respect to the upper jaw and a full denture attached to them. Both failures were in the abutments and occurred after about 3 years of service. Abutment A failed within the threaded area, abutment B failed at the junction of the threaded and unthreaded areas. Another difference was that abutment B was longer than abutment A by about 1.5mm, and therefore its fixture was implanted deeper into the jaw.

The fixtures and abutments were made of CP Ti grade 4 with chemical composition Ti-0.05C-0.05Fe-0.03N-0.01H-0.4O. The abutments were connected by a gold bar. The dental technician who removed the implants noticed bone resorption around them, and this had resulted in the implants being slightly skewed from the correct vertical position.

The failed abutments and the fixture remnant were examined in detail by Scanning Electron Microscopy (SEM) and optical metallography. SEM was used to (a) compare surface finishes, including that of an unused dental implant from the same manufacturer and with the same abutment diameter (3.5 mm), and (b) fractographically determine the failure mechanisms. Prior to the detailed examinations, the used implant parts were cleaned with acetone in an ultrasonic cleaner for approximately 30 minutes. They had originally been decontaminated and cleaned soon after they were removed from the patient. Ultrasonic cleaning has been conducted at low frequency (20 kHz) to minimize scratching and gouging to the implant. By comparing the implants before and after cleaning, it was found that ultrasound cleaning did not damage the implants.

Besides the physical examinations, see section IV of this paper, Finite Element Modelling (FEM) was used to determine the effect of implant misalignment on abutment service stresses. This is because the failed implants were slightly skewed from vertical when removed. The FEM modelling and results are given in section V.

These sections are followed by a discussion summarising the results, section VI, and conclusions and recommendations for avoiding similar failures.

4. Examination Results

4.1. Surface Finishes. Figure 3 shows (a) the fixture and abutment from the *unused* implant, (b) the fixture thread surface, indicative of high-intensity shot peening, and (c) a detail of the abutment thread surface showing coarse machining grooves. The broken fixture in Fig. 2 had also been shot blasted; the remaining thread on the broken abutment A also showed coarse machining grooves; and abutment B had coarse machining on its surface, see Figure 6. Therefore it is reasonable to assume that both the fixtures from the failed implants had been shot peened and both the broken abutments (A and B) had screw threads with coarse machining grooves.
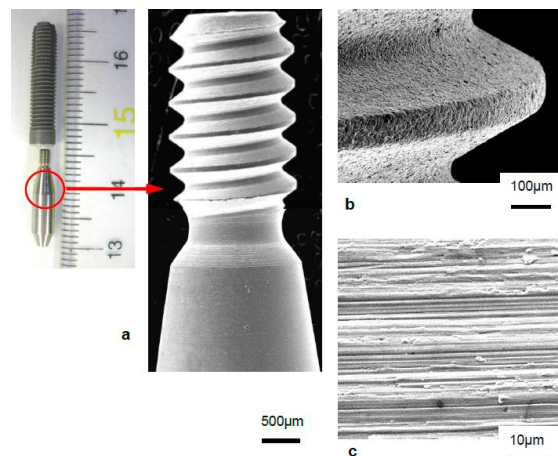
Fig. 3. Macroscopic and microscopic images showing (a) the unused fixture and abutment, (b) the shot-peened finish on the fixture, and (c) coarse machining grooves on the abutment screw threads.

4.2. Metallography. Figure 4 gives etched metallographic cross-sections of abutment A and its fixture. The abutment microstructure consisted primarily of equiaxed grains with sizes ~ 10 µm. There were, however, numerous constituent particles, which is unusual for titanium alloys, and occasional deformation twins. The fixture had a relatively coarse grain size, 30 – 50 µm, and there were numerous deformation twins, especially near the external surface, but also deep within the fixture. This twin distribution is consistent with high-intensity shot peening of the external surface.
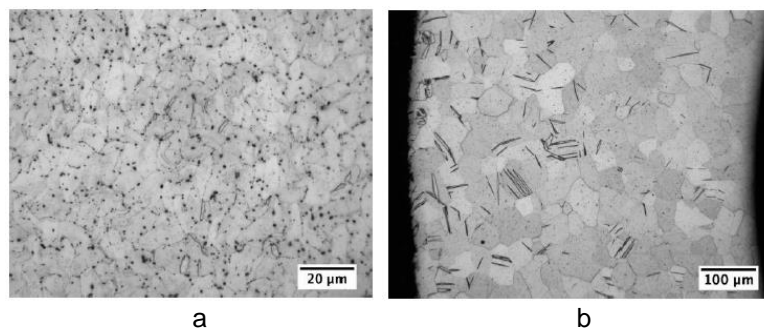
Fig. 4. Microstructures of (a) abutment A and (b) its fixture: Kroll's etch.

4.3. Fractography. Figure 5 gives overviews of the abutment fracture surfaces. Abutment A fractured at the second thread while abutment B fractured at the junction of the threaded and unthreaded areas. In both cases fatigue began on opposite sides of the abutments.

This indicates fatigue by reversed bending, although the relative amounts of fatigue-cracked areas and the positions of the final fracture areas show that the bending stresses must have been much higher on the sides from which the larger areas of fatigue commenced. Even so, the final fracture areas were less than 10 % of the total fracture surfaces, indicating that the abutments had not been abnormally loaded.

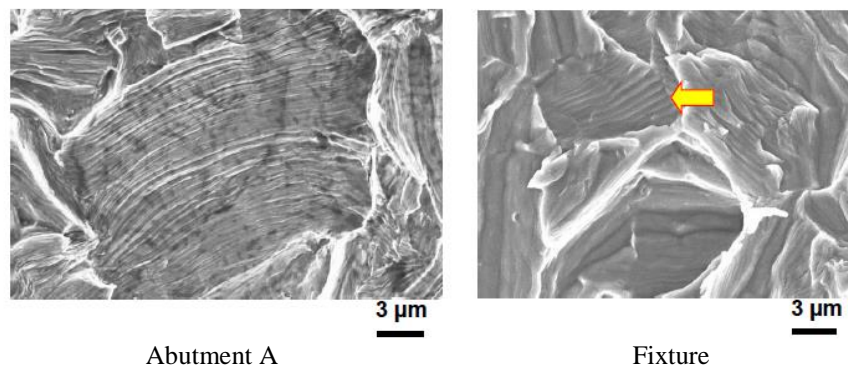Abutment A   Abutment B
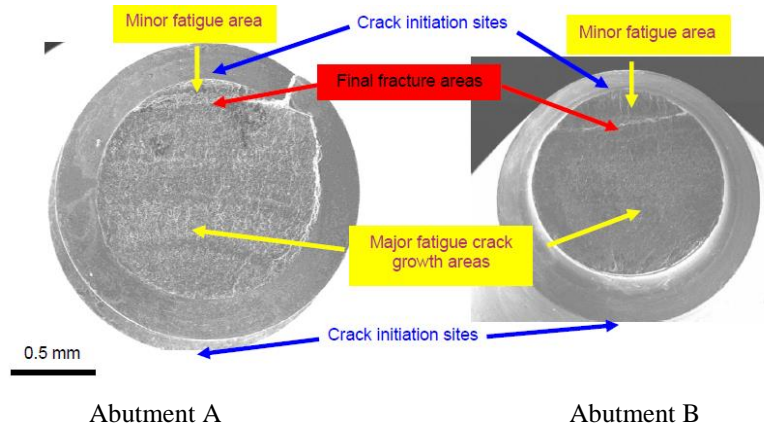
Fig. 5. SEM overviews of the abutment fracture surfaces.

The fatigue cracks most probably began directly from coarse machining grooves on the abutment screw threads (abutment A) and just at the beginning of the screw threads (abutment B). Figure 6 shows coarse machining grooves directly associated with the beginning of the major fatigue crack in abutment B.
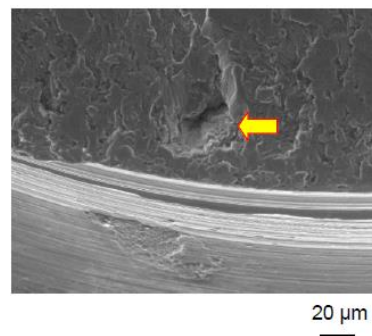
Fig. 6. SEM detail of the major fatigue crack initiation site in abutment B, showing coarse machining grooves on the abutment surface. The arrow points to a dent (incidental post-fracture damage). This dent is also visible in Fig. 5, just above the blue arrowhead.

Abutment A   Fixture

Fig. 7. SEM details of the fatigue fracture surfaces of abutment A and its fixture. The arrow points to an isolated patch of coarse striations.

Figure 7 shows details of the fatigue fracture surface of abutment A, and also evidence of fatigue on the fracture surface of its fixture. The abutment had large areas of fatigue striations, whose spacings were typically 0.3-0.5μm. However, striations on the fatigue fracture surface of the fixture were coarser and in isolated patches between large microvoids. This shows that the fixture failed under (very) high stresses after, and owing to, complete failure of the abutment.

Figure 8 identifies the small region of final overload failure bounded by the two areas of fatigue crack growth in abutment A. The detail in figure 8b shows irregular microvoid coalescence characteristic of overload failure.
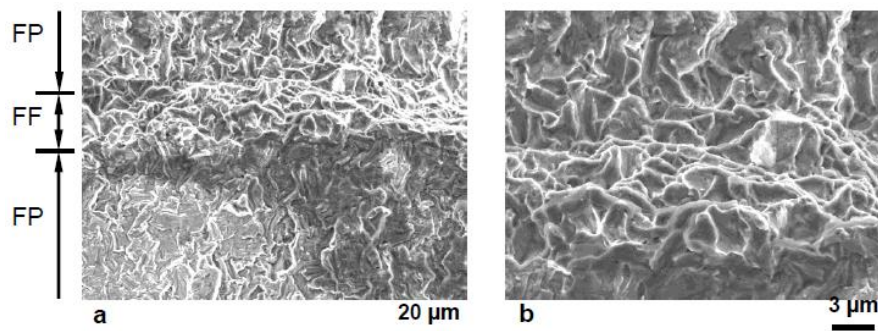
Fig. 8. SEM fractographs identifying final overload failure in abutment A: (a) the transition area between fatigue propagation and final fracture area bounded by the two fatigue areas; (b) typical microvoid coalescence characteristic of overload. FP = fatigue propagation; FF = final fracture.

5. Numerical Stress Analysis

Besides the physical examinations, described in section IV of this paper, Finite Element Modelling (FEM) was used to determine the effect of implant misalignment on service stresses experienced by the abutments. As mentioned in section III, the failed abutments were found to be slightly skewed from the correct vertical position owing to bone resorption. Although it is unknown when the implants became skewed, it is likely that this misalignment would have amplified the stresses in the abutments and fixtures and contributed to earlier fatigue failure.

Forces and bending moments in dental implants have been discussed in detail by Patterson et al. [28] and Morgan & James [30]. They concurred that bending moments might be more important for implant failures than previously thought. The study by Morgan & James [30] is particularly relevant to the present case because they considered a similar implant geometry, of multiple implants supporting a full prosthesis. They used an analytical model to predict the bending and torsional stresses in the implants, and found that the bending moment due to the vertical component of the applied load can cause stresses an order of magnitude higher than the direct axial stresses on the implants.

5.1. Choice of FEM for the Stress Analyses. Owing to complicating factors in this case study, a numerical (FEM) approach was chosen to predict the stresses in the abutments and fixtures under normal chewing and biting loads. These factors are:

(1) The two implants were joined by the connecting bar (gold alloy) as well as the resin-based full denture, see Figs. 2 and 9. These connecting members have different mechanical properties, which are difficult, if not impossible, to model otherwise. In particular, the denture's compliance (lack of rigidity) is important, since this increases the loads transferred to the implants [6].

(2) The misalignment of the implants should be accounted for.

To fully capture the effects of these factors a finite element model was developed using the commercial finite elements software ABAQUS. The undeformed model is shown in Fig. 9.
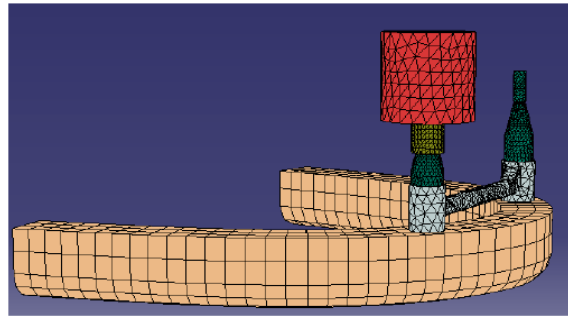
Fig. 9. The undeformed finite element model. The model components include the denture, connecting bar, abutments, fixtures, and bone. The shorter abutment A is shown in the foreground. The fixture and bone have been removed schematically from abutment B to highlight the notch between the tapered and threaded sections.

5.2. Material Property Considerations for ABAQUS. Since the implants failed by fatigue the deformations were assumed to be primarily elastic. Material properties typical of CP Ti were assigned to the fixtures and abutments with tensile strength of 440 MPa, yield strength of 380 MPa, elastic modulus of 114 GPa and density of 4510 kg/m^3. For the connecting bar properties were derived from a typical dental gold alloy with tensile strength of 460MPa, yield strength of 350 MPa, elastic modulus of 95 GPa and density of 1580 kg/m^3 [31, 32].

The elastic properties of the denture resin, ceramic and jawbone were taken from Daas et al. [33]. For the jawbone this meant homogenizing the properties of cortical and cancellous bone, and in the present context it was deemed acceptable to ignore the difference in quality of maxillary and mandibular bone.

Similarly, the densities of the denture resin and ceramic were estimated from those of commercially available materials [32]. The densities of the cortical and cancellous bone were estimated from various sources [34-36].

5.3. Implant Dimensions, Positions and Modelling Details. The dimensions of the fixtures, abutments, connecting bar, and denture were estimated from detailed photographs of the failed implants, examples of which are given in Figs. 2a, 2b, and 3a. These estimates took account of the different lengths of the two abutments.

Apart from the notches where the tapered parts of the abutments transitioned to screw threads, see Fig. 3a, the threaded sections were modelled as uniform cylinders. This simplification allowed more realistic surface interactions while still enabling comparison of the bending stresses for correctly positioned and misaligned implants. Two implant positions were considered: one perfectly vertical and the other rotated 10^0 about the axis of the connecting bar.

The threaded connections between the abutments and fixtures were modelled by tying the two FEM meshed surfaces together. Contact interactions were allowed between the abutments and fixtures in the tapered regions. The mesh interface between the bone and implant was also tied to simulate effective bone ingrowth. Finally, the fixtures were allowed to extend slightly above the bone surfaces to capture the effects of bone resorption: the amount of resorption was estimated based on the assumption that fracture occurred at the bone-mucosa interface.

5.4. Example FEM Results. As a test case, a simple vertical load was applied to the denture where the upper front teeth would be. This simulates a simple biting action. The applied force of 30 N was well below the maximum biting forces of 147-284 N achieved by adults with a full upper denture [37, 38].

Using this simple loading case, the maximum stresses on the surfaces of the abutments were calculated for both the perfectly aligned and misaligned implants. These maximum

stresses were near the notches where fatigue actually initiated. For the perfectly vertical case the maximum von Mises stress was approximately 80 MPa on the notch surface of abutment A. For the misaligned case the maximum von Mises stress was approximately 172 MPa in a very similar location, i.e. a misalignment of only about 10º more than doubled the maximum stress. This is shown in Fig. 10.
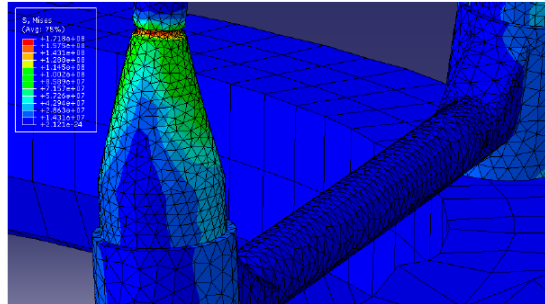
Fig. 10. The stress concentration on abutment A for the misaligned implant case. The maximum stress on the abutment surface is about 172 MPa. Note that the fixture has been hidden in this image.

6. Discussion

Failure of dental implants, particularly in the abutment/fixture threaded area, is highly undesirable. As mentioned in section II, mechanical failure can be attributed to a number of factors. Typical failure mechanisms are fatigue and overload. The stresses that cause fatigue failure come from repeated biting or chewing that operates at a normal load, or clenching that possibly operates at a relatively high load.

The present case concerned two fatigue-dominated abutment failures. The fatigue cracks most probably began directly from coarse machining grooves on the abutment screw threads (abutment A) and just at the beginning of the screw threads (abutment B). This conclusion is consistent with the findings of Yokoyama et al. [24]. In general, a coarser machined surface promotes fatigue crack initiation, since machining marks, especially grooves normal to the principal stresses, act as stress concentrations. Also, titanium and its alloys are considered to be notch-sensitive [3].

A corrective measure could be as simple as shot peening the surfaces of the abutments, especially near the areas of highest stress. Shot peening tends to flatten and smooth out the machining marks and introduces beneficial compressive residual stresses at and near the surfaces. These compressive stresses inhibit or delay fatigue cracking. Shot peening is commonly used for dental implants, especially the fixtures, and has been shown to improve the fatigue resistance of dental titanium [39].

For a simple vertical loading case the finite element model showed semi-quantitatively that a slight (10º from vertical) misalignment of the implants more than doubled the maximum stress at the stress concentration of an abutment. The main reason for this large effect is a significant increase of the moment arm of the vertical force. Some additional contribution could be due to stresses induced by interaction between the two coupled implants. It is unknown whether the misalignment in the present case was present initially, or that it occurred mainly owing to bone resorption, which seems more likely. If bone resorption was the main culprit, then it could have been self-perpetuating. In other words, the higher loads and stresses on the implants, and hence the surrounding bone, could have caused an increased resorption rate [11].

Irrespective of the cause of the misalignment, the much higher bending stresses on the abutments would greatly shorten the fatigue lives. Combined with the locally poor surface quality (coarse machining grooves) on the abutments, it is not surprising that the implants failed after only three years in service.

7. Conclusions and Recommendations

The CP Ti dental implants failed owing to reversed bending fatigue of the abutments after about three years in service. The final fracture areas were less than 10% of the total fracture surfaces, indicating that the abutments had not been abnormally loaded. The fatigue cracks most probably began directly from coarse machining grooves on the abutment screw threads (abutment A) and just at the beginning of the screw threads (abutment B).

A finite element model using a simple but common loading scenario showed that a slight misalignment of the implants contributed to much higher stress levels at critical locations in the abutments. This indicates that great care be taken in aligning the implants during insertion, and that any misalignment due to bone resorption can be problematic. Whether or not implant misalignments occur during service, it is also recommended to pay special attention to the surface finish of the abutments. In the present case their fatigue lives could have been extended by providing a (much) better surface quality, e.g. by finishing with shot peening. This would have smoothed out the coarse machining grooves and introduced beneficial compressive residual stresses at and near the abutment surfaces.

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