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Effect of Arm Design and Chemical Polishing on Retentive Force of Cast Titanium Alloy Clasps

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Keywords
Retentive force; titanium; Akers clasp; accuracy; clasp designs.

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Abstract

Purpose: Titanium dentures have recently increased in popularity. A decrease in the retentive force of the titanium clasps has frequently been observed. This study investigated the effect of retentive arm design and chemical polishing of titanium clasps.

Materials and Methods: Four Akers clasps with 0.25-mm and 0.5-mm undercuts were designed so that the retentive arms were placed at 1/2 and 2/3 of the undercut area. Wax patterns were fabricated and invested in phosphate-bonded Al_2O_3/LiAl SiO_6. They were cast with CP Ti Grade 2, CP Ti Grade 3, and Ti-6Al-7Nb using an argon gas-pressure casting unit (Autocast HC-III). After sandblasting, the castings were chemically treated with an HNO_3/HF solution. The retentive force (N) was measured up to 10,000 insertion/removal times. The results ($n = 5$) were analyzed by ANOVA/Tukey's test ($\alpha = 0.05$).

Results: The retentive forces significantly decreased with increasing immersion time in the HNO_3/HF solution ($p < 0.05$). The retentive force of the 2/3 undercut was significantly greater than that of the 1/2 undercut ($p < 0.05$). Excluding the initial insertion/removal period of the 2/3 undercut, there were no significant differences among all the titanium metals tested ($p > 0.05$). As the number of insertion/removal times increased, the retentive force of the 2/3 undercut greatly decreased. The retentive force of the Ti-6Al-7Nb clasps for the 1/2 undercut decreased the least after 10,000 insertion/removal times.

Conclusions: Chemical treatment for titanium clasps should be performed for 1–5 minutes due to the following factors: accuracy, surface roughness, surface structure, initial retention, and stability of retention. To maintain appropriate long-term retentive force, the retentive arms should be placed in the 1/2 undercut area of the abutment tooth.

The use of a single metal for all restorations in the mouth is important, because it protects against metal corrosion caused by the contact of different metals.[1-4] For this "one metal rehabilitation" concept, titanium should be used for restorations and prostheses on the assumption that implant treatment will be carried out after a tooth is lost. Commercially pure (CP) titanium has appropriate mechanical properties, light weight (low density) compared to conventional dental alloys, and outstanding biocompatibility that avoids metal allergic reactions.[5,6] Cast titanium prostheses began to be delivered to fully and partially edentulous patients approximately 15 years ago. In clinical practice, practitioners discovered several practical problems, such as debonding of the denture base resin from the titanium framework, discoloration of the titanium surface, and severe wear of CP titanium prostheses. Additionally, permanent deformation of titanium clasps has frequently been observed.[7-10] Recently, implant abutments and frameworks have been fabricated with titanium using computer-aided design and computer-aided manufacturing (CAD/CAM).[11] Because an oxide reaction layer is always attached to the surface of cast titanium, CAD/CAM fabrication is ideal compared to conventional casting; however, the frameworks of removable partial dentures (RPDs) are very complicated and are made up of many component parts, such as major connectors, minor connectors, retainers, clasp assembly, and bracing and stabilizing components. In particular, the clasp arm with an undercut cannot be milled with CAM burs. Therefore, titanium frameworks for RPDs must continue to be made by casting. The details are still unclear regarding cast titanium clasps. Titanium is purported to be more flexible than Co–Cr alloy.[12] If a retentive force similar to that of Co–Cr clasps is necessary, the clasps should be designed with deeper undercuts or thicker and wider arms or the retentive arms can be placed at 2/3 of the undercut area. The fatigue resistance of titanium is reported to be significantly lower than that of Co–Cr and Type IV gold alloys.[13,14] Vallittu and Kokkonen[15] suggested that this property may cause loss of retention of the RPDs and clasp failure. In addition, titanium clasps designed to achieve great retention may promote clasp failure.

On the other hand, the breakage of cast titanium clasps may also occur because of the internal porosity in the castings. A high incidence of casting porosity in clasp assemblies has been reported in radiographic studies of titanium frameworks, which are more prone to porosity than frameworks made of conventional dental casting alloys.[16] Thus, many casting factors, such as sprue designs, investments, and casting pressure, have been investigated to decrease the porosity.[9,17,18] Of the sprue designs, it seems that porosity in the clasps may be reduced through the use of curved sprues and thicker auxiliary sprues. Low casting pressure and attachment of air vents might also aid in decreasing porosity.[18,19]

Cracking and defects on the internal surfaces of titanium clasps are also frequently observed and may contribute to permanent deformation or a reduction in retentive force.[12] Most of the external surface layer of titanium castings is usually removed to get rid of the contamination layer;[20–23] however, the internal surfaces of the clasp arms should not be ground or cut using rotational burs, because they must fit the abutment tooth accurately. After sandblasting, chemical polishing of the titanium clasps is performed with hydrofluosilicic acid to obtain a smooth internal surface and to remove the oxide reaction layer from the surfaces.[24] Only a few studies have focused on the chemical polishing of titanium clasps, and little is known about the correlation between clasp designs and retentive force.

The purpose of this study was to clarify the relationship of the length of time of chemical polishing of the clasps using hydrofluosilicic acid, the fitness of the clasp to the abutment tooth, and the retentive force. The effect of retentive arm design of the titanium clasps on maintaining appropriate retentive force was also investigated.

Materials and methods

Specimen preparation

An 18-8 stainless steel die (10.0-mm diameter, 8.0-mm high, 7.5-mm radius of curvature) simulating the first molar was prepared (Fig 1). After an impression of the die was made using silicone impression material (Duplicone, Shofu, Tokyo, Japan), a master cast of the die was fabricated using extra-hard stone plaster (Fuji-rock, GC Corp., Japan). Surveying was performed, and the clasp marking was engraved by marking the off-pin on the master cast. After an impression of the master cast was made, a duplicate cast was made with phosphate-bonded Al_2O_3/LiAl SiO_6 investment material (T-invest, GC). As shown in Figure 2, four Akers clasp assemblies were designed so that the retentive arms were placed at 1/2 (Design A) and 2/3 (Design B) of the undercut area, and the tips were engaged at the 0.25-mm undercut and at 1/2 (Design C) and 2/3 (Design D) of the 0.5-mm undercut. Wax patterns (RKG, Dentaurum, Pforzheim, Germany) were fabricated and invested in a phosphate-bonded investment material (T-invest). Designs A–D were cast with CP Ti Grade 2 (CP2; T-alloy M, GC), CP Ti Grade 3 (CP3; T-alloy H, GC), and Ti-6Al-7Nb (6–7; T-alloy Tough, GC) using an argon gas-pressure casting machine (Autocast HC-III, GC) according to the manufacturer's instructions

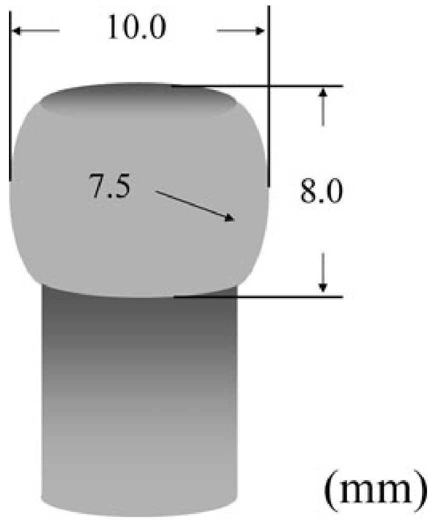

Figure 1 Illustration of the die simulating the first molar.

(Table 1). As controls for Design A, cobalt-chromium (Co–Cr) alloy (Crutanium, Krupp, Germany) and Type IV gold (Au–Pt) alloy (PGA 3, Ishifuku, Japan) were cast conventionally. All titanium castings were sandblasted with 50 μm Al_2O_3 particles under 500 kPa for air-abrasion pressure and chemically treated with an HNO_3/HF solution (Chemipolish, Shofu) for 1 minute. To investigate the effect of the chemical treatment, only the CP3 Design A clasp was immersed in the solution for 0 (non-immersed), 0.5, 1, 5, 10, and 15 minutes. To ensure uniformity, polishing was not performed; only the nodules and burs were carefully removed. Five clasps were fabricated for each factor, namely, 20 CP2, 45 CP3, 20 6–7, 5 Co–Cr, and 5 Au–Pt clasps; a total of 95 clasps were prepared.

Measurements of accuracy

To measure the gap distance between the clasp and the cask-forming die, the silicone film method was used as in a previous study.[25] White high-viscosity silicone impression material (Fit-checker, GC) was mixed at a base:catalyst ratio of 6:1, amply applied on the internal surface of the clasp, and placed on the die. After the clasp was held for 3 minutes under a constant load of 3 kg, it was removed from the die, and then the black silicone material (Bite-checker, GC) was mixed (base:catalyst = 6:1) and poured inside the white silicone material of each clasp. After 3 minutes, the two combined silicone materials were removed from the clasp. The silicone block materials from each clasp were sectioned with a razor blade buccolingually at 1 mm from the end of the clasp tip. To check the accuracy of the clasp, the white silicone layer at the sections was measured using a profile projector (V-16E, Nikon, Tokyo, Japan) at a magnification of $\times 50$.

Observations of the surface structure

After the clasp arms were cut at the shoulder, they were placed on the noncontact 3D surface roughness profilometer (NH-4H, Mitaka, Japan). The surface roughness (Ra) of five random points on the internal surface of each clasp was measured with

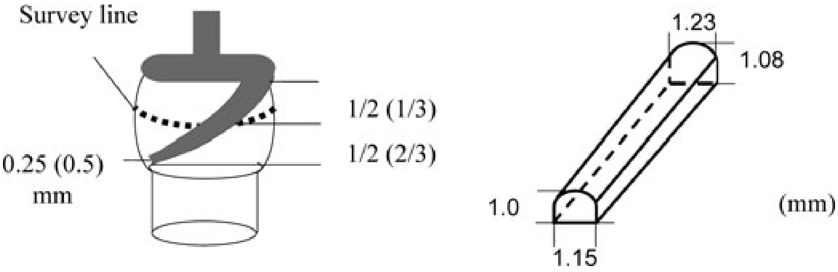

Figure 2 Akers clasp assemblies and size of retentive arm.

Table 1 Four designs, Akers clasp, and metals used

	Undercut		
Designs	Amount (mm)	Area	Metals
A	0.25	1/2	CP2, CP3, 6–7, Co–Cr, Au–Pt
B	0.25	2/3	CP2, CP3, 6–7
C	0.5	1/2	CP2, CP3, 6–7
D	0.5	2/3	CP2, CP3, 6–7

a cut-off (the specified value correlates to the peak-to-valley height) of 0.8 λc and measurement area of 100 μm × 100 μm. The surface structure of the chemically-treated clasp was observed with a scanning electron microscope (SEM, JSM 6300, JEOL, Tokyo, Japan) using an accelerating voltage of 15 kV. To observe the oxidation reaction layer, the clasp arms were embedded in epoxy resin (Epo-Mix, Epoxide, Buehler, Lake Bluff, IL) and sectioned perpendicularly to the surface.

Measurements of retentive force

Before tensile testing, all the titanium clasps were examined radiographically to detect possible casting defects that would eliminate their use from the retention testing. An X-ray generator (Shima-vision, Shimazu, Tokyo, Japan) was set to 70 kVp and 100 mA for a 32-second exposure time at 40 cm between the force and the film. An insertion/removal testing apparatus designed at the Tsurumi University School of Dental Medicine was used to measure the retentive force (Fig 3). Altogether, there were five tensile-compressive mechanical moving parts for retentive force measurement of the five clasps. After each clasp was mounted in the apparatus, it was inserted in the terminal position of the die under a 1-kg load and subsequently removed from the die to simulate the placement and removal of RPDs. These actions constituted one cycle. The retention test was performed by repeated tensile and compressive motions (crosshead speed: 950 mm/min) up to 10,000 cycles in distilled water at 37°C. After 500 cycles, the retentive force (N) of each clasp was evaluated as the tensile force obtained when the clasp was separated from the die at a crosshead speed of 50 mm/min. This measurement was repeated for 10 cycles, that is, 1–10 cycles, 501–510 cycles, and 1001–1010 cycles. The 10 sets of data were averaged, and the data were analyzed with the SPSS statistical package (Version 12.0, SPSS, Inc., Tokyo, Japan) by one-way analysis of variance (ANOVA) and the Tukey's multiple comparison test at a significance level of $\alpha = 0.05$.

Results

Accuracy

The gap distances between the clasp arm and the die are indicated in Figure 4. As the immersion time increased, the gaps also apparently increased. The titanium clasps without chemical treatment and with treatment for 30 seconds had 20–30 μm gaps. The gaps were significantly smaller than for 1 to 15-minute treatment groups ($p < 0.05$).

Surface structure

Figure 5 shows the internal surface roughness of the chemically-treated titanium clasps. As the treatment time increased, the surface roughness decreased. A significant difference was found between 0 minutes (9.2 Ra) and 1–15 minutes (1.9–4.0 Ra), and between 0.5 minutes (6.3 Ra) and 1–15 minutes ($p < 0.05$).

SEM pictures of the internal surfaces and sections of the chemically-treated titanium clasp arms are shown in Figures 6 and 7, respectively. The appearance of the chemically-treated surface at six immersion times (0–15 minutes) correlated with the surface roughness values (Ra). The surfaces of the

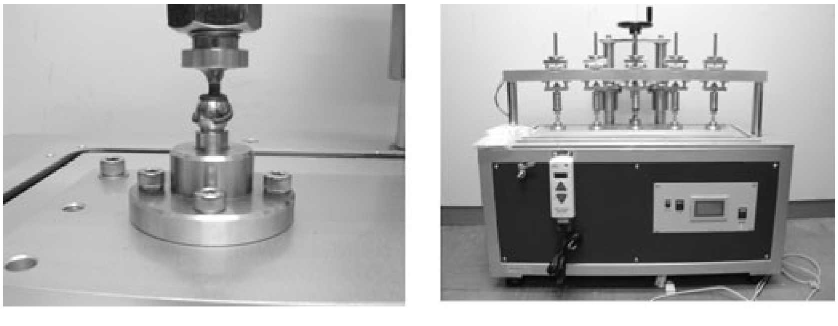

Figure 3 Insertion/removal testing apparatus for measurement of retentive force up to 10,000 cycles.

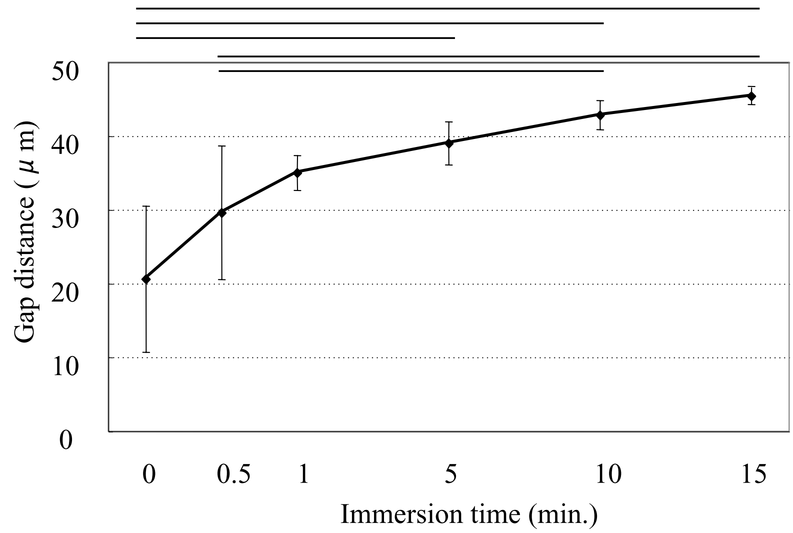

Figure 4 Gap distance between clasp arm and die after certain immersion times for chemical treatment. Horizontal bars indicate significant difference ($p < 0.05$).

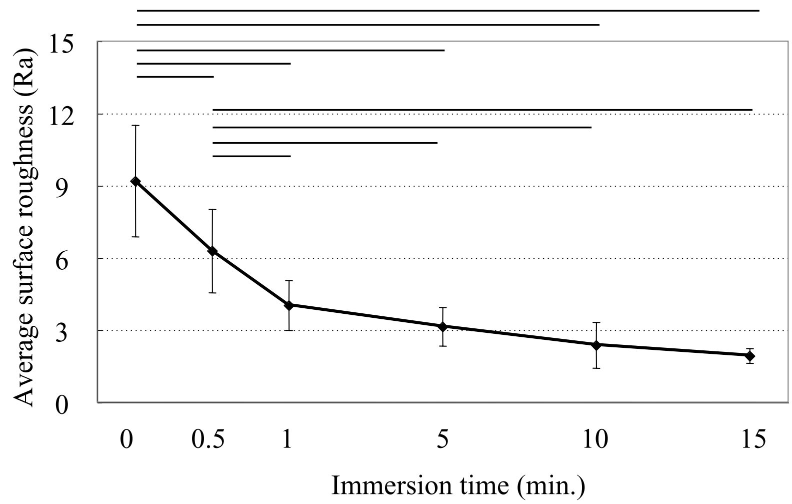

Figure 5 Internal surface roughness of chemically-treated titanium clasps. Horizontal bars indicate significant difference ($p < 0.05$).

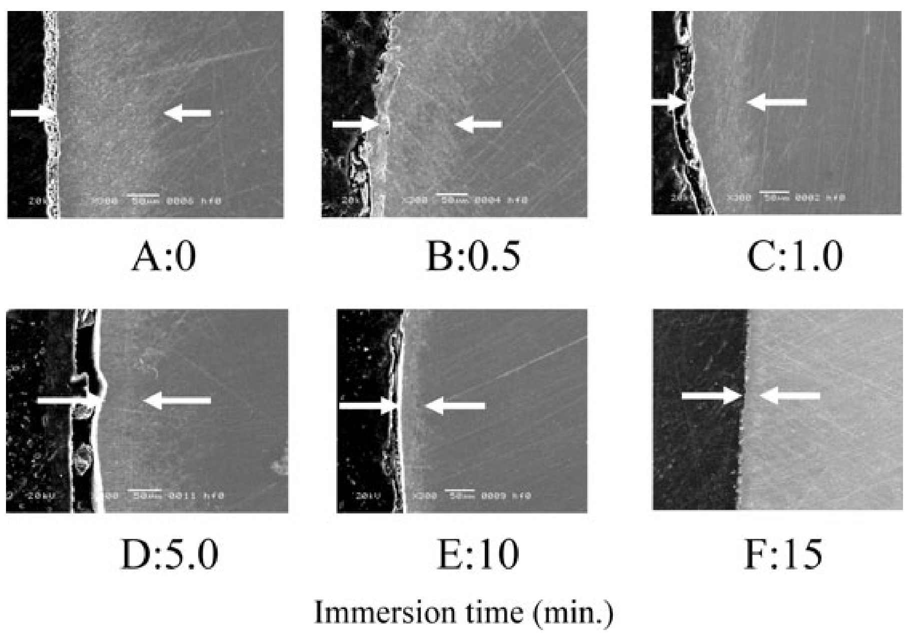

Figure 6 SEM observation of sections of chemically-treated titanium clasps (×300) (the space between the two white arrows indicated α-case).

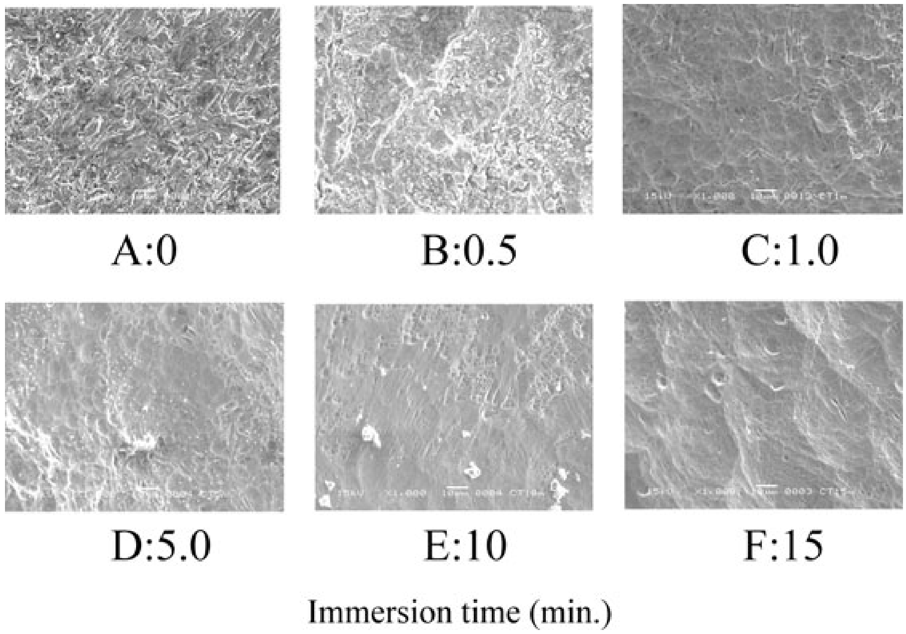

Figure 7 SEM observation of internal surface of chemically-treated titanium clasps (×1000): (A) 0 minutes, (B) 0.5 minute, (C) 1 minute, (D) 5 minutes, (E) 10 minutes, (F) 15 minutes.

nontreated and 30-second-treated clasps were rougher and had many sharp edges (burs), compared with the clasps treated for more than 1 minute. Similarly, the oxidation layers were thinner as the immersion times increased, that is, 170 μm (0 minutes) to 35 μm (15 minutes).

Retentive force

Some titanium clasps (approximately 5%) had small porosities and were rejected after nondestructive X-ray examination. Others showed no definite defects and were used for the retention test. The changes in the retentive force up to 10,000 cycles and the relationship with the chemical treatment times are found in Figure 8. The initial retentive forces ranged from 6.2–14.7 N. There were statistical differences between 0 minutes and 1–15 minutes, 0.5 minute and 10–15 minutes ($p < 0.05$). The retentive force for 0 and 0.5 minutes rapidly decreased, particularly for 500 cycles. Although the retentive force at 0 and 0.5 minutes after 10,000 cycles decreased to 37.3% and 38.4%, respectively, of its original value, there were smaller changes (approximately 50% decrease) at 1–15 minutes. There were no statistical differences in the 0 to 15-minute treatment times after 10,000 insertion/removal cycles ($p > 0.05$).

The changes in retentive force of all the metals tested, including the controls for Design A (1/2 and 0.25 undercut), are shown in Figure 9. The initial retentive forces (N) of the different metals from highest to lowest were: Co–Cr (23.12) > CP2 (10.39) > 6–7 (9.99) > CP3 (9.89) > Au–Pt (9.4). There was

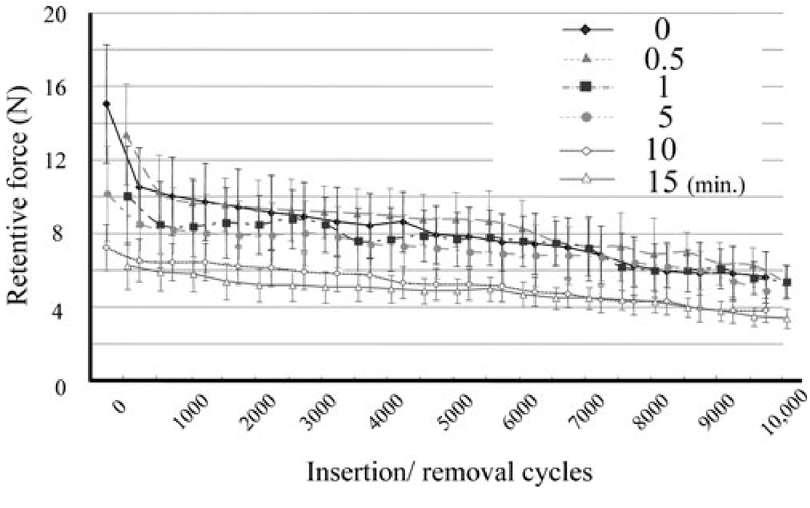

Figure 8 Changes in retentive force up to 10,000 cycles affected by the chemical treatment time.

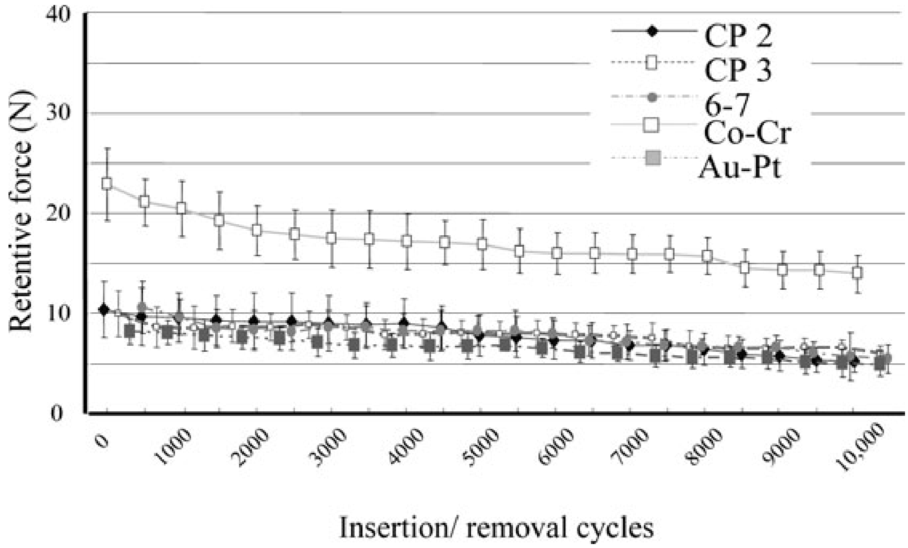

Figure 9 Changes in retentive force of all metals tested including controls for Design A.

a statistical difference between the Co–Cr and the others ($P < 0.05$). Similarly, fewer changes (as % of original force) in the retentive force occurred in the following order: Au–Pt (65.6) > Co–Cr (61.6) > 6–7 (58.8) > CP3 (53.5) > CP2 (50.1). Among the different designs, fewer changes (as % of original force) were found for Designs A (54.4) > C (40.4) > B (28.4) > D (28.3).

Figure 10 indicates the changes in the retentive force up to 10,000 cycles of the four designs of the clasps. During the initial insertion/removal cycles, the retentive force of the 2/3 undercut (Designs B and D) was significantly greater (approximately 1.5–2 times) than that of the 1/2 undercut (Designs A and C) ($p < 0.05$). For each design, there were no significant differences among all the titanium tested ($p > 0.05$). As the number of insertion/removal cycles increased, the retentive force of the 2/3 and 0.5-mm undercut (Design D) greatly decreased (CP2: 28.9%, CP3: 27.3%, 6–7: 29.2%).

Discussion

The surface structures of titanium castings are greatly affected by casting methods. In commercial titanium casting systems, the greatest oxidation layer may be produced in the GC Autocast system, because phosphate-bonded investment material (Al_2O_3/ $LiAlSi_2O_6$) is used. As there is Si in the investment, the cast surface becomes rougher from the burning out of patterns.[17,18,27] Therefore, the limitation of this study was that the results were from only one casting machine (GC Auto-cast) and one investment material (T-invest). As for accuracy, titanium frameworks tend to be difficult to place precisely in the original positions; consequently,[28] they tend to be less accurate, because the thermal expansion coefficient of titanium is less than for conventional dental alloys. However, the gaps of the titanium clasps in this study were 20–30 μm, which were nearly the same as the values for the Au–Pt alloys in a previous study.[26] Other reports recommended casting with titanium and then connecting the clasps by laser-welding to maintain accuracy.[29-31] In the case of the Au–Pt alloy, the in vitro gaps (21.7 $\pm$ 3.9 μm) between the clasp and tooth were larger than the in vivo gaps (15.5 $\pm$ 3.5 μm).[26] Therefore, the clinical accuracy of titanium clasps might be better than the results of this study suggest.

Although the oxidation layer on the internal surface of the clasp arm causes deformation and failure, the fitness between the clasp and tooth would be worse if the oxidation layer were removed.[8] Hence, the time of chemical polishing should be clarified to achieve both removal of some degree of the oxidation layer and accuracy. The surface roughness and the gap distance rapidly decreased up to 1 minute of chemical treatment, and these generally decreased from 1 minute. Appropriate retentive force up to 10,000 cycles was obtained from a 1 to 5-minute treatment. Therefore, the above factors might be satisfied with an approximately 1 to 5-minute chemical treatment.

Of Design A, three titanium clasps and the Type IV gold alloy clasp had similar retentive force (9.4–10.39 N), because these four alloys have similar elastic moduli (CP2: 120 GPa; CP3: 126 GPa; 6–7: 123 GPa; Au–Pt: 100 GPa).[32] Although the elastic modulus of Co–Cr is approximately two times (200 GPa)[32] higher than the others, the retentive force was about three times greater than theoretically predicted. In contrast, Designs C and D had approximately two times greater theoretical retentive force than did Designs A and B; however, a 2/3 undercut (Designs C and D) is generally used for flexible clasps such as wire clasps. Although CP titanium is a flexible metal,[5,12] the use of a 2/3 undercut for the cast clasps with CP titanium and titanium alloy may be inadequate, because the initial retention is too great, and the retention remarkably decreased after 10,000 insertion/removal cycles. The stability of retention of CP titanium (51.0–58.3% decrease) was less than for the conventional alloys (Co–Cr: 61.6%, Au–Pt: 65.6%). Oshima[33] reported on

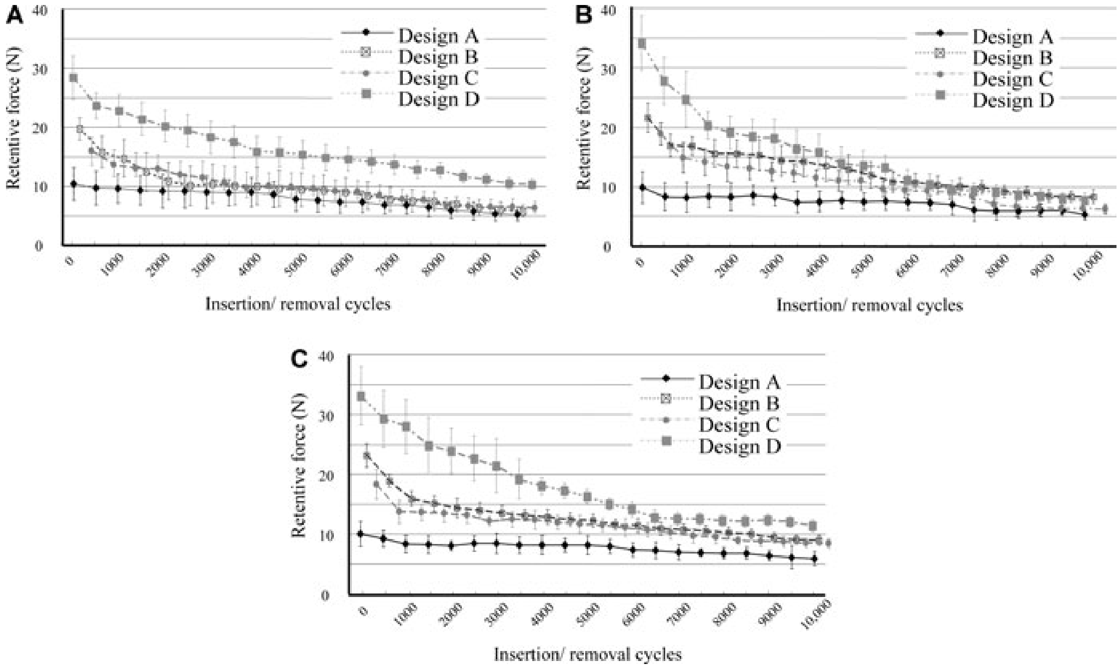

Figure 10 Changes in retentive force up to 10,000 cycles of the four clasp designs: (A) CP2, (B) CP3, (C) 6–7.

the retentive force of titanium special form clasps (occlusal rest with buccolingual undercut margins). As in this study, CP titanium exhibited less stable retention compared to Co–Cr and Au–Pt because of the wear of the internal surface of the clasp and permanent deformation; however, permanent deformation has not yet been scientifically explained in previous studies or the present study. The poor tribological characteristics of titanium are well known.[7] In their review of titanium alloys for orthopedic applications, Long and Rack[34] presented the complications of the wear phenomenon of titanium and indicated that overall alloy composition, which controls the surface oxide composition and subsurface deformation behavior, is a critical factor in titanium wear. Moreover, severe wear of CP titanium prosthetic teeth has frequently been observed, and gold alloy teeth have exhibited better wear resistance.[7] The wear resistance of titanium should be improved by surface modification, or a new titanium alloy with a different microstructure should be developed for use in areas likely to undergo friction and wear.

Appropriate retentive force of a clasp must be determined by many factors, namely, the pattern of partial edentulousness, tooth condition, loss of periodontal tissue, retention required for one RPD, number of retainers, etc. Judging from many clinicians' opinions, approximately 20 N per RPD would be adequate to hold it in place when sticky foods are chewed and still allow patients to easily remove the RPD. For clasps, appropriate retention would be 5–10 N so that harmful forces are not applied to abutment teeth.[35] From these viewpoints, titanium clasps should be accepted for clinical use despite the existence of the surface oxidation layer, and they should be ideally designed as Design A (1/2 and 0.25-mm undercut) to maintain appropriate retention.

Conclusions

Within the limitations of this study, the following conclusions were obtained:

1. Chemical treatment for titanium clasps should be performed for 1–5 minutes because of the following factors: accuracy, surface roughness, surface structure, initial retention, and stability of retention.
2. CP titanium and titanium alloy had similar initial retentive force to Type IV gold alloy, and significantly less retentive force than Co–Cr alloy.
3. Among all the metals tested, the stability of retention was ranked as Type IV gold > Co–Cr > titanium alloy > CP titanium.
4. The stability of retention from highest to lowest for the different designs fell in the order of Designs A > C > B > D.

To summarize, titanium clasps should be designed so the retentive arms and the tips are placed in the 1/2 undercut area and 0.25-mm undercut, respectively, of the abutment tooth.

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