The effect of low temperature aging on the mechanical property & phase stability of Y-TZP ceramics

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STATEMENT OF PROBLEM. Recently Yttrium-stabilized tetragonal zirconia polycrystal (Y-TZP) has been introduced due to superior flexural strength and fracture toughness compared to other dental ceramic systems. Although zirconia has outstanding mechanical properties, the phenomenon of decrease in the life-time of zirconia resulted from degradation in flexural strength after low temperature aging has been reported. **PURPOSE.** The objective of this study was to investigate degradation of flexural strength of Y-TZP ceramics after various low temperature aging treatments and to evaluate the phase stability and micro-structural change after aging by using X-ray diffraction analysis and a scanning electron microscope (SEM). **MATERIAL AND METHODS.** Y-TZP blocks of Vita In-Ceram YZ (Vita Zahnfabrik, Bad Säckingen, Germany) were prepared in 40 mm (length) x 4 mm (width) x 3 mm (height) samples. Specimens were artificially aged in distilled water by heat-treatment at a temperature of 75, 100, 125, 150, 175, 200, and 225℃ for 10 hours, in order to induce the phase transformation at the surface. To measure the mechanical property, the specimens were subjected to a four-point bending test using a universal testing machine (Instron model 3365; Instron, Canton, Mass, USA). In addition, X-ray diffraction analysis (DMAX 2500; Rigaku, Tokyo, Japan) and SEM (Hitachi s4700; Jeol Ltd, Tokyo, Japan) were performed to estimate the phase transformation. The statistical analysis was done using SAS 9.1.3 (SAS institute, USA). The flexural strength data of the experimental groups were analyzed by one-way analysis of variance and to detect statistically significant differences (α = .05). **RESULTS.** The mean flexural strength of sintered Vita In-Ceram YZ without autoclaving was 798 MPa. When applied aging temperature at below 125℃ for 10 hours, the flexural strength of Vita In-Ceram YZ increased up to 1,161 MPa. However, at above 150℃, the flexural strength started to decrease. Although low temperature aging caused the tetragonal-to-monoclinic phase transformation related to temperature, the minimum flexural strength was above 700 MPa. **CONCLUSION.** The monoclinic phase started to appear after aging treatment above 100℃. With the higher aging temperature, the fraction of monoclinic phase increased. The ratio of monoclinic/tetragonal + monoclinic phase reached a plateau value, circa 75% above 175℃. The point of monoclinic concentration at which the flexural strength begins to decrease was between 12% and 54%. **KEY WORDS.** Low temperature aging, Y-TZP ceramic, Phase transformation, Mechanical property **[J Adv Prosthodont 2009;1:113-7]**

INTRODUCTION

Recently, all ceramic restorations are widely used not only for anterior and posterior single crowns, but also for anterior three or four-unit fixed partial dentures. Among several dental ceramics, Yittrium-stabilized tetragonal zirconia polycrystal (Y-TZP) was accepted by dentists and patients due to its superior flexural strength and fracture toughness compared to other dental ceramic systems.[1]

Zirconia has three different crystal phases at different temperatures: monoclinic, tetragonal and cubic. At room temperature zirconia exists in monoclinic phase.[2] Above 1070℃, the monoclinic phase transforms into a tetragonal phase, which is stable between 1170℃ and 2370℃.[2] The cubic phase is stable only at very high temperature. Zirconia can maintain the tetragonal phase at room temperature by adding the stabilizers such as yttria, magnesia and ceria etc. The martensitic tetragonal-to-monoclinic phase transformation may be initiated by stress (such as wear, machining, etc) and/or water. The phase transformation from tetragonal to monoclinic exhibits a volume expansion of 3 - 4%.[3] The crack propagation is inhibited by the volume expansion at the crack tip. This phase transformation toughening phenomenon results in high strength and fracture toughness of the zirconia.

Although zirconia has outstanding mechanical properties, the phenomenon that decreases the life-time of zirconia has been reported for more than twenty years. Kobayashi *et al.* were the first to observe the degradation concerning the aging phenomenon at 150 - 400℃.[4] Since then, numerous studies have focused on the low temperature aging phenomenon associated with the mechanical properties.[5-9]

The objectives of this study were i) to investigate the degradation of flexural strength of Y-TZP ceramics after various low temperatures aging treatment and ii) to evaluate the phase stability and micro-structural change after aging by using a scanning electron microscope and X-ray diffraction analysis.

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MATERIAL AND METHODS

Preparation of specimens

Y-TZP blocks of Vita In-Ceram YZ (Vita Zahnfabrik, Bad Säckingen, Germany) were fully sintered according to manufacturer's recommendations. The fully sintered blocks were cut into the size of 41 (length) × 5 (width) × 4 mm (height) approximately. Then the blocks were divided into 8 groups, 10 specimens for each. All specimens were polished with 6 μm diamond paste, followed by 1 μm for 10 minutes at 3 kg load, then annealed in a porcelain furnace at a temperature of 1200℃. The purposes of the polishing were to remove all surface residual stresses, to round chipped corners of the specimens, and to eliminate any cracks from inherent material defects. The annealing process can remove any monoclinic phase on the surface of specimens during specimen preparation procedures. Final size of the specimens was 40 ± 0.2 (length) × 4 ± 0.1 (width) × 3 ± 0.1 mm (height).

Low temperature aging

For a constant boiling of the specimens, the Autoclave reactor (ECO solution, Seoul, Korea) was used. Specimens were artificially aged in distilled water by heat-treatment at a temperature of 75, 100, 125, 150, 175, 200, and 225℃ for 10 hours in order to induce the phase transformation at the surface.

Flexural strength

To measure the mechanical property, the specimens were subjected to a four-point bending test using a universal testing machine (Instron model 3365, Instron, Canton, Mass, USA) at a crosshead speed of 0.5 mm/min. The load to failure of the specimens was recorded in Newton (N), and the flexural strength was calculated in mega Pascal (MPa). To estimate the reliability and variability of strength, the Weibull modulus (m) was calculated. It was obtained from the slope of the curve generated by plotting ln[ln 1/(1-Pf)] against lnσ.

The following equation was used to calculate the flexural strength.[10]

$$\sigma(4-point) = \frac{6P}{2bd^2}\left(\frac{L_1 - L_2}{2}\right)$$

Where P is the load at failure, L_1 and L_2 are the outer and inner span length respectively. b is the specimen's breadth, and d is the specimen's height. In this study, L_1: 30 mm, L_2: 10 mm, b: 4 mm, and d: 3 mm.

X-ray diffraction analysis

To measure the crystalline phase fraction, X-ray diffraction (DMAX 2500; Rigaku, Tokyo, Japan) was carried out. The specimens were placed in the holder of a diffractometer and scanned by using Cu-Kα radiation (λ = 1.54056 Å) at a diffraction angle change from 25° to 40°. Step size of 0.02° and scan speed of 1 degree per minute were used to determine the peak position and composition. Peaks from the XRD output were compared with library data and the monoclinic/tetragonal phase ratio X_m was calculated using the following equation, as described by Garvie and Nicholson.[11]

$$X_m = \frac{I_{m(111)} + I_{m(11\bar{1})}}{I_{m(111)} + I_{m(11\bar{1})} + I_{t(111)}}$$

Where Ij{*h k l*} is the area of the {*h k l*} peak of the phase j measured by XRD. The X-ray penetration depth is around 5 μm.[12]

Scanning electron microscopy

The image of surface grains of each subgroup was analyzed by a scanning electron microscope (Hitachi s4700, JEOL, Tokyo, Japan). Photographs were taken at magnifications of × 15.0 k, × 30.0 k. The specimens were coated with a 15 nm gold layer prior to observation in order to make their surface conductive.

Statistics

The statistical analysis was done using SAS 9.1.3 (SAS institute, Cary, USA). The flexural strength data of the experimental groups were analyzed by one-way analysis of variance to detect statistically significant differences (α = .05).

RESULTS

Flexural strength

Mean 4-point flexural strength value, standard deviation, and Weibull modulus (m) of Vita In-Ceram YZ are listed in Table I. The mean flexural strength of sintered Vita In-Ceram YZ without autoclaving was 798 MPa. The specimens autoclaved at 125℃ for 10 hours had the highest flexural strength of 1161 MPa.

Weibull statistics indicated that the Weibull modulus (m) was ranged from 6 to 33 (Table I).

Fig. 1. showed the porous microstructure and grain size of Vita In-Ceram YZ. The average grain size of 0.5 μm was measured using the linear intercept method. The gap among the grains became definite in the groups autoclaved at 150 and 225℃ for 10 hours, compared to the specimen without autoclaving.

X-ray diffraction analysis

The peaks from the X-ray diffraction analysis were compared with the library data, which is the X-ray diffraction standards file (17-0923) on zirconium oxide. The results from X-ray diffraction analysis of the specimens without autoclaving revealed only the tetragonal phase. However, the X-ray diffraction analysis for the other groups indicated that they had various ratio of tetragonal and monoclinic phase related to the temperature applied. The major peaks of the tetragonal phase

Table I. Flexural strength and Weibull modulus (m) of Vita In-Ceram YZ

Group	Mean 4-point flexural strength in MPa (SD)	Weibull modulus (m)
A Vita In-Ceram YZ (room temperature, 20° C)	798 (96)	9
B Vita In-Ceram YZ (75° C 10 h)	1025 (134)	8
C Vita In-Ceram YZ (100° C 10 h)	1047 (110)	10
D Vita In-Ceram YZ (125° C 10 h)	1161 (196)	6
E Vita In-Ceram YZ (150° C 10 h)	932 (93)	11
F Vita In-Ceram YZ (175° C 10 h)	880 (22)	33
G Vita In-Ceram YZ (200° C 10 h)	824 (60)	14
H Vita In-Ceram YZ (225° C 10 h)	722 (38)	21

Scanning electron microscopy (SEM) analysis

of Vita In-Ceram YZ in the range from 25 degree to 40 degree 2θ were 30.18, 34.62, and 35.18 degree. Also, those of the monoclinic phase were detected at 28.16 and 31.4 degree. The dominant peak was the tetragonal phase at 30.18 degree. The relative intensities of the tetragonal and monoclinic phase changed with various materials and autoclaving temperature. The monoclinic phase concentration (by % weight) was calculated from the relative intensity of the tetragonal and monoclinic peaks using the equation proposed by Garvie and Nicholson.

Fig. 2. shows the relationship between the ratio of monoclinic / tetragonal + monoclinic phase and various temperatures. And, Fig. 3. reveals the relationship between the flexural strength and monoclinic phase concentration of Vita In-Ceram YZ. The monoclinic phase started to appear after aging treatment above 100℃. The fraction of monoclinic phase increased as the aging temperature became higher. The ratio of monoclinic / tetragonal + monoclinic phase reached a plateau value, circa 75% above 175℃.

DISCUSSION

The data of flexural strength, monoclinic / tetragonal + monoclinic ratio, and microstructure of fractured surface indicated that the mechanical property of Y-TZP ceramic varied after low temperature aging.

Flexural strength increased from room temperature to 125℃, and at a temperature of 125℃, the flexural strength was the highest. After that, it started to decrease. This result was different from that of the previous studies. In the previous studies, low temperature aging did not reduce the flexural strength of zirconia or the degradation of strength was dependent on the aging temperature and time.[5-9,13,14] The temperature applied to low temperature aging varied with the studies. Cales *et al.*[13]

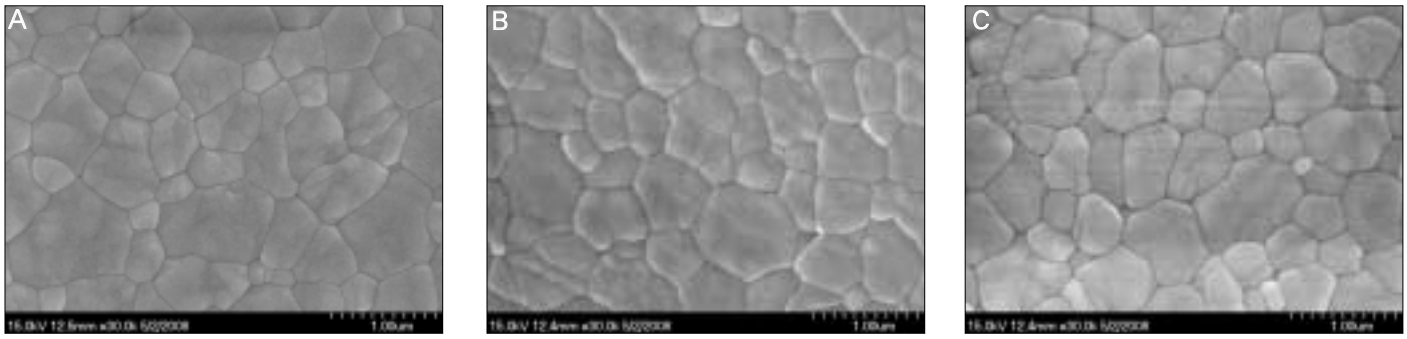

Fig. 1. The images taken by scanning electron microscope.
(A) Vita In-Ceram YZ without autoclave aging, original magnification × 30.0 k (B) Vita In-Ceram YZ with autoclave aging at 150℃ for 10 hours, original magnification × 30.0 k (C) Vita In-Ceram YZ with autoclave aging at 225℃ for 10 hours, original magnification × 30.0 k

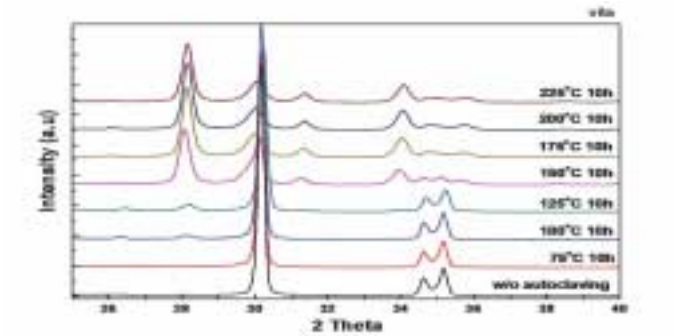

Fig. 2. X-ray diffraction pattern of Vita In-Ceram YZ after aging process at different temperatures.

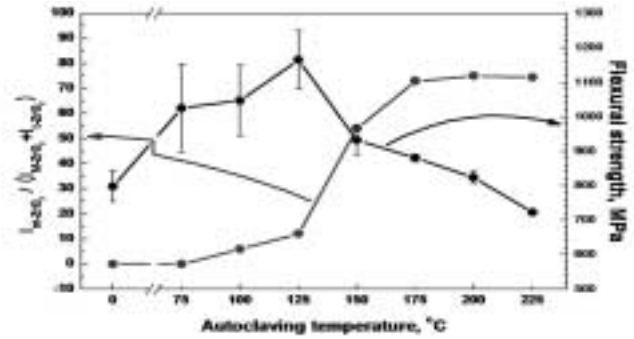

Fig. 3. Graph of the flexural strength and monoclinic phase concentration of Vita In-Ceram YZ.

applied 37℃, and Shimizu *et al.*[14] used above 100℃ for aging temperature. In the study that low temperature aging treatment did not reduce the flexural strength, the temperature applied to the aging treatment was relatively low. In fact, 37℃ is a lower temperature than that applied in the other studies and can not cause degradation of Y-TZP in a short time.[15]

Chevalier *et al.*[16] investigated the time-dependent aging changes on Y-TZP. The authors reported that the higher the temperature and dwell duration, the higher the transformation rate and it could be related to the overall grain size. As the size of tetragonal grain increased, the stability began to diminish. In this study, the specimens were aged at various temperatures for 10 hours. Aging at 134℃ for 1 hour is theoretically similar to 3 - 4 years *in vivo*.[5,17,18] So, the experimental condition may approximately correspond to a long period of time. Chevalier *et al.*[5] found that aging at 130℃ after 7 hours, for a monoclinic content higher than 30%, the nucleation rate reached a saturation level. In the present study, the flexural strength was higher in the group autoclaved at 125℃ for 10 hours than the group without aging. These results revealed that monoclinic concentration to some extent provides the increase of flexural strength. At first, the tetragonal-to-monoclinic phase transformation is initiated at the surface of Y-TZP. And, when the amount of monoclinic phase was small, the change of grain size and physical property was insignificant.[15] On the contrary, the residual stress diminishes and the volume expansion by the phase transformation inhibits the crack propagation.[2] This phenomenon may be related to the increase of flexural strength. However, when the saturation in monoclinic phase is reached at the surface, the phase transformation then proceeds into the bulk of the Y-TZP ceramic.[9] Under this condition, internal flaws can be critical, and the degradation of the flexural strength occurs. Nevertheless, further reduction of the flexural strength was not observed below 700 MPa. The strengths were still higher than those of other dental ceramic systems, and those were above the value of occlusal force reported by the previous studies.[19] Therefore, the strengths of Vita In-Ceram may be sufficient for allowing reliable use for crown and fixed partial denture in the posterior region.

Ceramics have a wide variability of strength due to the flaws and micro cracks inside. The Weibull modulus (m) is related to the flaw-size distribution.[20] In order to get the reliability of experimental data, it is important to investigate the mean flexural strength and to estimate the Weibull modulus (m). Thus, in the study concerning ceramics, the Weibull modulus was calculated. While low m value corresponds to a wide flaw-size distribution and large error range, high m value corresponds to a narrow flaw-size distribution and outstanding reliability.[20,21] Most ceramics are reported to have 'm' values in the range of 5 to 15.[22] In the present study, Vita In-Ceram has 'm' values in the range of 6 to 33. These facts indicate that they have great structural reliability.

Three-point bending test is the standard method for estimating the flexural strength of dental ceramics. However, in this study, 4-point bending test was performed to reflect the flaws and defects in specimens thoroughly. The 4-point bending test develops lower level of shear force in the specimen compared to the 3-point bending test.[23]

X-ray diffraction analysis indicated that the flexural strength was increased until up to 12% of monoclinic concentration. The flexural strength decreases remarkably above 125℃ aging temperature, At the same time, monoclinic concentration increases sharply from 12% to 75%. Further study is needed to determine the critical amount of monoclinic phase at which the flexural strength begins to fall.

It was reported that the tetragonal-to-monoclinic transformation proceeds from the external to the internal of Y-TZP ceramic.[9] Therefore, it can be assumed that tetragonal-to-monoclinic phase transformation begins to proceed from the external to the internal when the amount of monoclinic increases sharply. Images taken by SEM showed that the transformation from tetragonal to monoclinic phase during low temperature aging procedure begins to make rapid progress between at 125℃ and at 150℃. Low temperature aging caused the fractured surface of Vita In-Ceram YZ to reveal an "orange peel" like texture.[24] This means that each grain was pushed out of the surface and the roughness of surface increased due to low temperature aging. This change of surface may lead to the degradation of mechanical property of Y-TZP ceramics. Previous low temperature aging studies have also attributed decreased strength to phase transformation in Y-TZP ceramics.[5,6,8] Due to the difficulty of performing the X-ray diffraction analysis on depth deeper than the first 5 μm, another method such as AFM is necessary for more precise analysis.[5]

In clinical situations, fractures of veneer and/or interface between veneer and core happen more frequently than those of including core generally.[25] Therefore, flexural strength exceeds 700 MPa even after low temperature aging may not guarantee that this result would be effective in the clinical situation. In order to make the condition similar to clinical situation, it is strongly recommended to design an improved experimental model loaded repeatedly vertical and lateral with the force applied to the veneer/core specimens after low temperature aging.

CONCLUSION

1. Low temperature aging produced the positive and negative effects on the mechanical properties of Y-TZP ceramics depending on the temperature applied.
2. When applied aging temperature at below 125℃ for 10 hours, the flexural strength of Y-TZP ceramics increased. However, at above 150℃, the flexural strength started to decrease.

3. Low temperature aging caused the tetragonal-to-monoclinic phase transformation related to temperature. The monoclinic phase started to appear after aging treatment above 100℃. The fraction of monoclinic phase increased with the higher aging temperature. The ratio of monoclinic/tetragonal + monoclinic phase reached a plateau value, *circa* 75% above 175℃.
4. The point of monoclinic concentration at which the flexural strength begins to decrease is between 12% and 54%.
5. After aging process of various temperatures, minimum flexural strength of 700 MPa and minimum Weibull modulus of 6 was maintained.

REFERENCES

1. Kosmac T, Oblak C, Jevnikar P, Funduk N, Marion L. The effect of surface grinding and sandblasting on flexural strength and reliability of Y-TZP zirconia ceramic. Dent Mater 1999;15:426-33.
2. Piconi C, Maccauro G. Zirconia as a ceramic biomaterial. Biomaterials 1999;20:1-25.
3. Subbarao EC, Maiti HS, Srivastava KK. Martensitic transformation in zirconia. Phys Status Solidi Series A 1974;21:9-40.
4. Kobayashi K, Kuwajima H, Masaki T. Phase change and mechanical properties of ZrO_2-Y_2O_3 solid electrolyte after ageing. Solid State Ionics 1981;3/4:489-93.
5. Chevalier J, Cales B, Drouin JM. Low-temperature aging of Y-TZP ceramics. J Am Ceram Soc 1999;82:2150-204.
6. Lilley E. Review of low temperature degradation in Y-TZPs. In: Tressler RE, McNallan M, editors. Ceramic transaction: corrosion and corrosive degradation of ceramics. Westerville: American Ceramics Society, 1990:387-407.
7. Lawson S. Environmental degradation of zirconia ceramics. J Eur Ceram Soc 1995;15:485-502.
8. Deville S, Chevalier J, Fantozzi G, Bartolome JF, Requena J, Moya JS. Low-temperature ageing of zirconia-toughened alumina ceramics and its implication in biomedical implants. J Eur Ceram Soc 2003;23:2975-82.
9. Yoshimura M. Phase stability of zirconia. Am Ceram Sot Bull 1988;67:1950-5.
10. Lube T, Manner M, Danzer R. The miniaturization of the 4-point bend test. Fatigue Fract. Engng Mater Struct 1997;20:1605-16.
11. Garvie RC, Nicholson PS. Phase analysis in zirconia systems. J Am Ceram Soc 1972;55:303-5.
12. Emsley J, The Elements (2nd ed.). Oxford University Press, Oxford, UK, 1991.
13. Cales B, Stefani Y, Lilley E. Long-term *in vivo* and *in vitro* aging of a zirconia ceramic used in orthopaedy. J Biomed Mater Res 1994;28:619-24.
14. Shimizu K, Oka M, Kumar P, Kotoura Y, Yamamuro T, Makinouchi K, Nakamura T. Time-dependent changes in the mechanical properties of zirconia ceramic. J Biomed Mater Res 1993;27:729-34.
15. Pittayachawan P, McDonald A, Petrie A, Knowles JC. The biaxial flexural strength and fatigue property of Lava Y-TZP dental ceramic. Dent Mater 2007;23:1018-29.
16. Chevalier J, Deville S, Münch E, Jullian R, Lair F. Critical effect of cubic phase on aging in 3mol% yttria-stabilized zirconia ceramics for hip replacement prosthesis. Biomaterials 2004;25:5539-45.
17. Cales B. Zirconia as a sliding material: histologic, laboratory, and clinical data. Clin Orthop Relat Res 2000;379:94-112.
18. Deville S, Gremillard L, Chevalier J, Fantozzi G. A critical comparison of methods for the determination of the aging sensitivity in biomedical grade yttria-stabilized zirconia. J Biomed Mater Res B Appl Biomater 2005;72:239-45.
19. Suárez MJ, Lozano JF, Paz Salido M, Martínez F. Three-year clinical evaluation of In-Ceram Zirconia posterior FPDs. Int J Prosthodont 2004;17:35-8.
20. Oblak C, Jevnikar P, Kosmac T, Funduk N, Marion L. Fracture resistance and reliability of new zirconia posts. J Prosthet Dent 2004;91:342-8.
21. Roebben G, Basu B, Vleugels J, Van der Biest O. Transformation-induced damping behaviour of Y-TZP zirconia ceramics. J Eur Ceram Soc 2003;23:481-9.
22. Guazzato M, Quach L, Albakry M, Swain MV. Influence of surface and heat treatments on the flexural strength of Y-TZP dental ceramic. J Dent 2005;33:9-18.
23. Wen MY, Mueller HJ, Chai J, Wozniak WT. Comparative mechanical property characterization of 3 all-ceramic core materials. Int J Prosthodont 1999;12:534-41.
24. Ardlin BI. Transformation-toughened zirconia for dental inlays, crowns and bridges: chemical stability and effect of low-temperature aging on flexural strength and surface structure. Dent Mater 2002;18:590-5.
25. Aboushelib MN, de Jager N, Kleverlaan CJ, Feilzer AJ. Microtensile bond strength of different components of core veneered all-ceramic restorations. Dent Mater 2005;21:984-91.

Evaluation of proximal contact strength by postural changes

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STATEMENT OF PROBLEM. Proper proximal contact is important for maintaining and stabilizing the dental arch. However, the proximal contact strength (PCS) is not a constant value and can be affected by a variety of factors. **PURPOSE.** This study examined the influences of postural changes on the posterior PCS. **MATERIAL AND METHODS.** Twelve adults with a normal occlusion and had not undergone prosthetic treatment or proximal restoration were participated in this study. A metal strip was inserted into the proximal surface and removed at a constant velocity. The contact strength was measured in every contact point between canine to second molar in both arches. The PCSs were obtained initially in the upright position, secondly in the supine position and finally in the upright position again. All measurements were repeated after a 2 hour period. Statistical analysis was carried out using the Friedman test ($P < .05$). **RESULTS.** Generally, a decrease in PCS occurred when the posture was changed from the initial upright to supine position, while it increased when the posture was changed from the supine to upright position. A significant change was observed in all areas except for between the canine-first premolar in the maxilla and between the first molar-second molar in the mandible areas. **CONCLUSION.** The posterior PCS, which dentists generally believe to be a static feature of occlusion, is affected significantly by posture. **KEY WORDS.** Proximal contact strength, Upright position, Supine position **[J Adv Prosthodont 2009;1:118-23]**

INTRODUCTION

Proximal contact is the area of a tooth that is in close association, connection or in touch with an adjacent tooth in the same arch.[1] The tooth is stabilized by contact with the adjacent teeth as well as by occlusal contact with the opposite tooth.[2] Proper proximal contact plays an important role in maintaining and stabilizing the dental arch.[3] However, weak or slightly opened proximal contact causes food impaction, dental caries, periodontal disease, failure of occlusion and an undesirable drift of the teeth. On the other hand, too tight contact can damage the periodontal tissue or cause improper tooth movement or interfere with the physiological displacement of the teeth. Therefore, maintaining the proper proximal contact in natural dentition and in tooth restorations is important.[4-11]

The proximal contact strength (PCS) during tooth restoration is generally determined by the floss that passes through the contact point. Dentists note the proper contact in clinical treatment as the entry of floss with a snap.[12] This method is simple but it is difficult to detect the detailed changes in the PCS.[10] The PCS is considered to be too tight if the floss cannot pass through the contact area or tear out during entry, but too weak if the floss passes the contact area too easily.[13]

Therefore, it is important to investigate the proper PCS. Osborn[14] initially reported the concept of a PCS measuring machine using the frictional force that occurs when an inserted thin metal strip escapes from the proximal area. When a metal strip is inserted between the nearby teeth, each tooth is displaced and they produce a force to resist the displacement. The PCS is defined as the force of the teeth resisting the mesio-distal displacement, and the technique of measuring the resistance is based on the concept of frictional force. Many studies have evaluated the PCS by using the frictional force. To measure the PCS, Dörfer *et al.*[15] used a 0.05 mm thick metal strip and Oh *et al.*[16] invented equipment to remove a 0.03 mm thick metal strip parallel to the proximal surface.

Southard *et al.*[17] reported that the anterior portion of an occlusal force made from the posterior molar and the degree of irregularity in the anterior teeth were both related to the contact strength of unrestored posterior teeth. Dörfer *et al.*[15] Suggested that the PCS is not a constant value but can be influenced by a variety of factors. Although the importance of proper contact strength is widely accepted by the dental community, the physiological factors that influence its magnitude have not been clearly defined. Therefore, the purpose of this study was to examine the influence of a postural

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change on the posterior PCS.

MATERIAL AND METHODS

Details of the PCS measuring equipment used in this study are reported elsewhere.[18] Briefly, the apparatus consists of a sensor part, motor part, body part and measuring part. The sensor transforms the output voltage from the strain gauge sensor to a digital signal, and the measurement limit is a maximum of 98 N. The motor part is a driving motor. In the measuring part, a metal strip, 2 mm in width and 0.03 mm in thickness, is fixed with a screw and inserted into the proximal surface. When the start button is pushed, the metal strip is removed at a velocity of 8 mm / s. The distal part of the measuring section, which is bent in a right angle, enables easier measurements at the posterior teeth.

Twelve young and healthy adults (male: 8, female: 4), aged between 23 - 33 years (mean age: 25.3 years old), with a class I, normal occlusion consented to participate in the study. All subjects had healthy periodontal tissue and complete dentition from the second molars forward. No subject had a history of prosthetic treatment, proximal restoration or orthodontic treatment over last 1 year. In addition, they did not show any signs or symptoms of food impaction or TMJ disorders.

The subjects were required to remain in the upright position for at least 1 hour before the start of the measurements, and asked to refrain from eating at least 1 hour before the experiment until the end of the experiment. However, occlusal contact or loading that occurs during general swallowing or clenching was allowed. After the subject sat in the upright position, they were asked to remain in a comfortable, muscle relaxed state. After turning on the equipment, the zero point setting was carried out. Before each experiment, the proximal surface was dried with an air syringe and a metal strip was then inserted into the proximal surface (Fig. 1).

The initial PCSs were obtained at the maxillary and mandibular canine - first premolar, first premolar - second premolar, second premolar - first molar and first molar - second molar proximal contact points. After pushing the start button, the metal strip was removed at a constant velocity. The highest value obtained during removal was regarded as the PCS. Each measurement was repeated 4 times. The representative PCS of each space was determined from an average of the 2 intermediate measurements, i.e. the highest and lowest values were excluded. During the measurement, all subjects were restricted not to make an occlusion, and a 2 minute - interval was allowed between each measurement. All experiments were carried out at approximately 7 pm and the subject was allowed sufficient rest after dinner.

After measuring the initial PCS, the subjects were asked to next assume the supine position. After 2 hours had passed, all the measurements were repeated. The subjects then returned to the upright position for an additional 2 hours and the same procedures were repeated.

Statistical analysis of the data was carried out using SPSS ver.12.1 (SPSS, Inc., Chicago, IL, USA). A Friedman test was used to evaluate the changes in the PCS in each proximal surface according to the postural change; initial upright, supine and final upright position. A value of $P < .05$ was considered significant.

RESULTS

Generally, the contact strength decreased when the posture was changed from an initial upright to a supine position. The contact strength then increased when the position was changed from the supine to upright position. In the maxilla, every other area except between the canine - first premolar showed significant changes (Tables I - IV). However, in the mandible, the only area showing a significant change was between the first molar - second molar (Tables V - VIII).

When a postural change was made from an initial upright position to a supine position, the maximum decrease was observed between the second premolar - first molar in the maxilla, and the minimum decrease was observed at the canine - first premolar in the mandible. After returning to the upright position, the maximum increase was observed at the second premolar - first premolar in the maxilla, and the minimum increase occurred at the canine - first premolar in the mandible (Figs 2 and 3).

After changing position from initial upright position to supine position, every area except between mandibular canine and first premolar showed statistically significant decrease and after coming back to final upright position, every area except between upper and lower canine and first premolar and between mandibular second premolar and first molar showed significant increase (Tables IX and X).

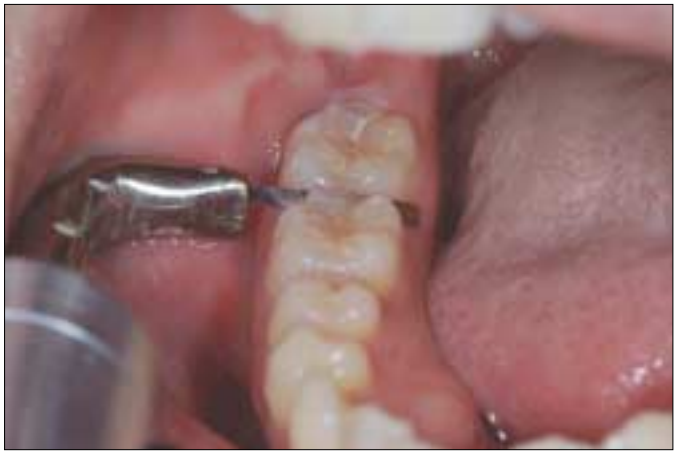

Fig. 1. Measurement of the proximal contact strength between the first molar and second molar in the right side of the mandible.

Table I. Proximal contact strength (N) between the canine and first premolar in the maxilla according to the posture of the subjects

Posture / Subject	Initial upright	Supine	Final upright	*P* value
1	0.66	0.67		
2	0.90	0.82	1.03	
3	0.54	0.41	0.60	
4	2.09	1.23	1.78	
5	0.69	0.55	0.69	
6	0.91	0.86	0.85	.14
7	0.98	1.07	1.12	
8	0.94	0.90	0.82	
9	0.78	0.72	0.72	
10	1.16	1.11	1.10	
11	0.83	0.70	1.03	
12	1.25	0.88	0.87	

Table II. Proximal contact strength (N) between the first premolar and second premolar in the maxilla according to the posture of the subjects

Posture / Subject	Initial upright	Supine	Final upright	*P* value
1	0.91	0.78	0.99	
2	1.08	0.86	1.17	
3	0.76	0.72	0.71	
4	1.81	1.15	1.76	
5	0.87	0.67	0.81	
6	0.91	0.82	0.84	.01
7	0.93	0.94	1.19	
8	1.07	0.88	0.9	
9	1.01	0.85	0.94	
10	1.25	1.35	1.40	
11	1.00	0.91	0.99	
12	1.36	1.12	0.98	

Table III. Proximal contact strength (N) between the second premolar and first molar in the maxilla according to the posture of the subjects

Posture / Subject	Initial upright	Supine	Final upright	*P* value
1	1.35	1.20	0.53	
2	1.14	1.05	1.17	
3	0.93	0.81	0.82	
4	2.75	1.73	3.08	
5	0.89	0.73	0.89	
6	1.07	0.82	0.87	
7	1.00	1.05	1.27	
8	1.06	0.88	1.09	
9	1.12	1.02	0.95	
10	1.92	1.72	1.72	
11	1.05	0.88	0.97	
12	1.68	0.97	1.42	

Table IV. Proximal contact strength (N) between the first molar and second molar in the maxilla according to the posture of the subjects

Posture / Subject	Initial upright	Supine	Final upright	*P* value
1	0.88	0.71	1.04	
2	0.98	0.90	0.98	
3	0.77	0.69	0.85	
4	1.85	1.46	1.52	
5	0.74	0.61	0.72	
6	1.41	1.13	1.20	.00
7	1.00	1.04	1.35	
8	1.19	1.12	1.29	
9	1.02	0.73	0.8	
10	2.63	2.20	2.68	
11	1.07	0.71	0.78	
12	1.55	1.14	1.29	

Table V. Proximal contact strength (N) between the canine and first premolar in the mandible according to the posture of the subjects

Posture / Subject	Initial upright	Supine	Final upright	*P* value
1	0.78	0.72	0.97	
2	0.57	0.66	0.79	
3	0.51	0.56	0.43	
4	1.17	0.94	1.16	
5	1.07	0.89	0.93	
6	0.98	0.83	0.98	.67
7	1.12	0.91	1.07	
8	0.79	0.80	0.73	
9	0.67	0.71	0.69	
10	0.37	0.47	0.42	
11	0.76	0.4	0.72	
12	0.99	0.91	0.85	

Table VI. Proximal contact strength (N) between the first premolar and second premolar in the mandible according to the posture of the subject

Posture / Subject	Initial upright	Supine	Final upright	*P* value
1	0.87	0.74	1.00	
2	0.70	0.67	0.84	
3	0.86	0.91	0.84	
4	1.73	1.44	1.80	
5	0.81	0.61	0.79	
6	1.12	1.06	1.13	.05
7	1.11	0.87	1.16	
8	1.10	0.95	1.09	
9	1.01	0.91	0.89	
10	1.21	1.41	1.58	
11	1.08	0.80	0.98	
12	1.35	1.13	1.06	

Table VII. Proximal contact strength (N) between the second premolar and first molar in the mandible according to the posture of the subjects

Posture / Subject	Initial upright	Supine	Final upright	*P* value
1	1.10	0.93	1.18	
2	1.39	1.32	1.44	
3	0.86	0.91	0.86	
4	2.29	1.93	2.41	
5	0.90	0.81	0.89	
6	0.99	0.98	0.94	.17
7	1.24	1.16	1.30	
8	1.48	1.21	1.19	
9	1.19	1.01	1.13	
10	2.27	1.96	2.45	
11	1.00	0.99	0.96	
12	1.12	1.24	1.11	

Table VIII. Proximal contact strength (N) between the first molar and second molar in the mandible according to the posture of the subjects

Posture / Subject	Initial upright	Supine	Final upright	*P* value
1	1.11	1.04	1.52	
2	1.09	1.08	1.18	
3	0.85	0.92	0.96	
4	1.59	1.30	1.39	
5	1.12	0.87	1.06	
6	1.42	1.26	1.31	.00
7	1.49	1.20	1.62	
8	1.30	1.20	1.27	
9	1.33	1.04	1.25	
10	3.30	2.58	3.15	
11	0.94	0.68	0.84	
12	1.42	1.33	1.17	

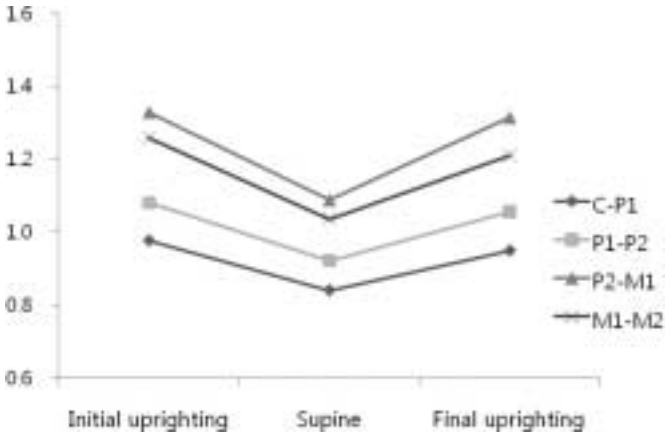

Fig. 2. Diagrammatic presentation of the changes in proximal contact strength (N) in each region of the maxilla according to the postural change.

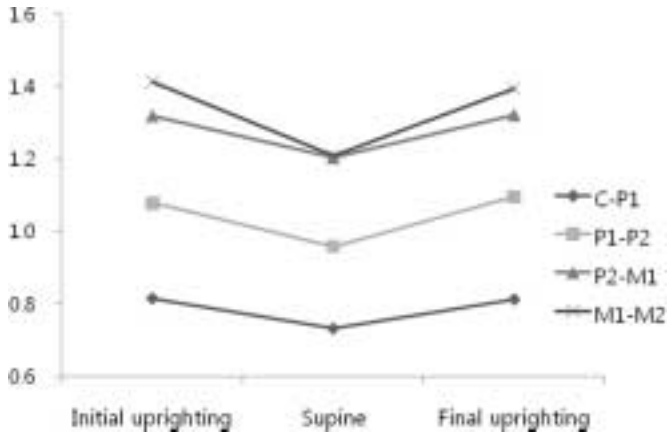

Fig. 3. Diagrammatic presentation of the changes in proximal contact strength (N) in each region of the mandible according to the postural change.

Table IX. *P* value in each site measured when changing from initial upright position to supine position

	Canine - First premolar	First premolar - Second premolar	Second premolar - First premolar	First molar - Second molar
Maxilla	.011	.007	.001	.001
Mandible	.129	.021	.021	.002

Table X. *P* value in each site measured when changing from supine position to final upright position

	Canine - First premolar	First premolar - Second premolar	Second premolar - First premolar	First molar - Second molar
Maxilla	.104	.013	.005	.000
Mandible	.134	.009	.105	.007

DISCUSSION

The equipment used in this study measured the PCS by evaluating the frictional force occurred while pulling a thin metal strip by an electric motor after inserting it into the proximal surface. Southard *et al.*[19] could not measure the PCS between the first and second molars due to the limitation of the measuring equipment. However, the equipment used in the present study enabled measurement during occlusion as well as at between the first and second molars by minimizing the volume of the intraoral part and by bending the tip in a right angle. The frictional force is independent from the surface contact area and velocity at a low velocity, particularly in range of 0.83 - 8.33 mm / s.[20,21] Therefore, the metal strip was removed at a constant

velocity of 8 mm/s to rule out the effects of the removal velocity on the contact strength.

As the posture was changed from an initial upright to a supine position, the PCS generally decreased in both dental arches and then increased after returning to the final upright position from the supine position. When the posture changed from an initial upright to supine position, there was a significant decrease in all areas except between the canine and first molar area in the mandible, which corresponds to the results reported by Southard *et al*..[19] After returning to the final upright position, significant increases were observed in the remaining area except between canine - first premolar in both arches and second premolar - first molar in the mandible. In contrast, Southard *et al*.[19] reported significant increases only between the second premolar - first molar in the maxilla and mandible. However, they could not measure the value between the first and second molar due to the limitations of their measuring equipment. Dentists generally consider the proximal tooth contact to be a static feature of occlusion. However, this assumption is actually incorrect because the PCS changes with posture.

It appears that the PCS is influenced by a variety of factors, including the place and shape of the teeth, masticatory action, amount of mouth opening and postural changes, such as various positions of the head etc..[15] In several studies, general increases in PCS from the anterior to posterior teeth were reported and significant differences were observed between the anterior and posterior teeth.[9,15,22] Southard *et al.*[23] insisted that the nearby teeth make contact with slight pressure with each other, and Kasahara *et al.*[11] observed a 3 - 21 μm space between the adjacent teeth using a CCD (Charge Coupled Device) microscope.

Slight contact of the space between the adjacent teeth can be tightened or disappeared by tooth intrusion caused by the vertical portion of the power during its function, and the mesial displacement of the teeth caused by the horizontal portion of the power. These changes increase the PCS.[17] The masticatory habits also have effects, and people with a unilateral masticatory habit show a larger increase in the working side after mastication.[9] In order to minimize the effects of function, the subjects were asked to refrain from eating from 1 hour before the measurement until all measurements had been taken. However, occlusal contact or occlusal loading that occurred during functions, such as swallowing or speaking, were allowed.

The amount of mouth opening also affects the PCS due to mandibular deflection resulting from activation of the inferior head of the lateral pterygoid muscle.[14] As the equipment used in this study removed a metal strip in the parallel direction, there was no need for excessive mouth opening. This is in contrast to other equipment, which removes the metal strip forward to the occlusal plane.[15] Hence, the opening range of the subjects was limited to approximately 20 mm. In order to minimize the effect of mouth opening on the PCS, the subjects were asked to be in a comfortable state so every muscle would be relaxed as much as possible. In addition, the frictional force could be changed by the condition of the contact surface. Therefore, the measuring site was dried before each measurement.[20]

Although the difference is slight, the PCS changed with time. The PCS in the stable state increases in the morning and then decreases in the afternoon.[15] This was explained by fatigue and the mucoelastic feature of the periodontal ligament. Because the highest level of masticatory muscle activation occurs during the eating periods in the daytime,[24] every measurement was carried out at 7 pm, which is after dinner and sufficient rest. An interval of 2 hours was allowed between each measurement because the changes in contact strength were not observed immediately and we wished to obtain full expression of any effect from the posture.

A change in head position alters the blood flow of the periodontal ligament and gravity acting on the teeth. When a person moves between the upright and supine position, the musculo-skeletal system, neural system and circulatory system cooperate to compensate for the change in blood flow. In the supine position, an increase in cardiac output due to increased venous return up-regulates the diastolic arterial pressure and average arterial pressure.[25,26] Furthermore, the increased pressure of the blood vessels produces a force sufficiently large to move the teeth under the physiological state.[27] Several studies have reported that teeth undergo pulsatile movements corresponding to the arterial pulse.[28,29] Changing posture from an upright to a supine position can increase the level of blood congestion or blood pressure in the periodontal ligament. The fact that these alterations can result in slight extrusion of the teeth from the alveolar socket seems to be quite reasonable. This extrusion can cause a slight decrease in the alveolar bone support for the teeth, and this decreased resistance can decrease the PCS. Opposite actions occurred when returning to the final upright position from the supine position.

There is a change in the direction of gravity functioning on the teeth as the subjects move from the upright to supine position. When one is in the upright position, gravity functions along the long axis of the teeth. However, the gravity changes direction toward the back in the supine position.[19] Hence, every tooth experiences dorsal angulation in the bony alveolar socket. Consequently, the more posterior teeth will show a higher level of distal angulation caused by a postural change. The interrelationship between the posterior position of the teeth and the gradually increasing distal angulation would weaken the PCS in the posterior teeth. This is clear from the results of the present study. An opposite reaction was observed when the subject returned to the final upright position from the supine position.

The effects of posture on the PCS should be considered when placing fixed prostheses or restoration in the proxi-

mal surface.[12,30] The placement of a posterior restoration and subsequent judgment of the contact strength by dentist after the patient is initially seated upright in the dental chair will reflect most accurately the PCS when the patient is functioning during the day. On the other hand, a restored contact that is judged to be of suitable tightness after a long appointment with the patient in the supine position may be excessively tight resulting in undesirable tooth movement.

In this study, the PCS from some subjects did not follow the general changes. An evaluation of the effect of factors other than posture will be needed because the PCS is a physiological feature, influenced by many factors such as position and shape of the teeth, mastication, amount of mouth opening and passage of time.

CONCLUSION

The effects of posture on the PCS of the posterior teeth were examined by measuring the PCS of 12 adults in the initial upright, supine and final upright position. Generally, there was a decrease in strength at all posterior proximal contacts in the maxilla and mandible when the subjects assumed a supine posture. On the other hand, there was an increase in contact strength after returning to the upright position from the supine position. Overall, the PCS of the posterior teeth, which dentists generally consider to be a static feature of occlusion, was altered significantly by changes in posture.

REFERENCES

1. The academy of prosthodontics. The glossary of prosthodontic terms. 8th ed. J Prosthet Dent 2005;94:37,57.
2. Andrews LF. The six keys to normal occlusion. Am J Orthod 1972;62:296-309.
3. Wheeler RC. An atlas of tooth form. 4th ed. Philadelphia, W.B. Saunders Co; 1969:12.
4. Sluder TB. Clinical dental anatomy, histology, physiology and occlusion. In: Studevant CM, ed. The art and science of operative dentistry. 2nd ed. New York: McGraw-Hill; 1985:21.
5. Hirschfeld I. Food impaction. J Am Dent Assoc 1930;17:1504-28.
6. Hancock EB, Mayo CV, Schwab RR, Wirthlin MR. Influence of interdental contacts on periodontal status. J Periodontol 1980;51:445-9.
7. Nielsen IM, Glavind L, Karring T. Interproximal periodontal intrabony defects. Prevalence, localization and etiological factors. J Clin Periodontol 1980;7:187-98.
8. Jernberg GR, Bakdash MB, Keenan KM. Relationship between proximal tooth open contacts and periodontal disease. J Periodontol 1983;54:529-33.
9. Vardimon AD, Matsaev E, Lieberman M, Brosh T. Tightness of dental contact points in spaced and non-spaced permanent dentitions. Eur J Orthod 2001;23:305-14.
10. Prakki A, Cilli R, Saad JO, Rodrigues JR. Clinical evaluation of proximal contacts of Class II esthetic direct restorations. Quintessence Int 2004;35:785-9.
11. Kasahara K, Miura H, Kuriyama M, Kato H, Hasegawa S. Observations of interproximal contact relations during clenching. Int J Prosthodont 2000;13:289-94.
12. Campagni WV. The final touch in the delivery of a fixed prosthesis. CDA J 1984;12:21-9.
13. Sturdevant JR, Sturdevant CM. Gold inlay and gold onlay restoration for Class II cavity preparations. In: Sturdevant C, Barton R, Sockwell C, Strickland W, eds. The Art and Science of Operative Dentistry. St. Louis, MO, USA: C.V. Mosby Co. 1985:490.
14. Osborn JW. An investigation into the interdental forces occurring between the teeth of the same arch during clenching the jaws. Arch Oral Biol 1961;5:202-11.
15. Dörfer CE, von Bethlenfalvy ER, Staehle HJ, Pioch T. Factors influencing proximal dental contact strengths. Eur J Oral Sci 2000;108:368-77.
16. Oh SH, Nakano M, Bando E, Keisuke N, Shigemoto S, Jeong JH, Kang DW. Relationship between occlusal tooth contact patterns and tightness of proximal tooth contact. J Oral Rehabil 2006;33:749-53.
17. Southard TE, Behrents RG, Tolley EA. The anterior component of occlusal force. Part 1. Measurement and distribution. Am J Orthod Dentofacial Orthop 1989;96:493-500.
18. Choi WJ, Kim KH, Kim JA, Kang DW, Oh SH. Evaluation and development of digital device for measuring proximal tooth contact tightness. J Korean Acad Prosthodont 2007;45:687-95.
19. Southard TE, Southard KA, Tolley EA. Variation of approximal tooth contact tightness with postural change. J Dent Res 1990;69:1776-9.
20. Shames IH. Engineering Mechanics, Vol. 1, Englewood Cliffs, NJ: Prentice-Hall, 1966;170-4.
21. Fuhrmann R, Grave C, Diedrich P. *In vitro* evaluation of a measurement method to analyze the interdental, mesially directed force. J Orofac Orthop 1998;59:362-70.
22. Kim KH. Evaluation of proximal tooth contact tightness in permanent dentitions. M.S. Thesis. In: Korea, Chosun University School of Dentistry. 2007.
23. Southard TE, Southard KA, Tolley EA. Periodontal force: a potential cause of relapse. Am J Orthod Dentofacial Orthop 1992;101:221-7.
24. Miyamoto K, Yamada K, Ishizuka Y, Morimoto N, Tanne K. Masseter muscle activity during the whole day in young adults. Am J Orthod Dentofacial Orthop 1996;110:394-8.
25. Smith JJ, Kampine JP. Circulatory physiology- the essential. Baltimore, MD: Waverly Press, Inc., 1984:235-7.
26. Rowell LB, Blackmon JR. Reflex and local control of the splanchnic circulation in humans. In: Hunyor S, Ludbrook J, Shaw J, McGrath M, eds. The peripheral circulation. New York: Excerpta Medica, 1984:114.
27. Moxham BJ. The role of the periodontal vasculature in tooth eruption. In: Davidovitch Z, ed. The biological mechanisms of tooth eruption and root resorption. Birmingham: EBSCO Media, 1988:208.
28. Körber KH. Periodontal pulsation. J Periodontol 1970;41:382-90.
29. Ng GC, Walker TW, Zingg W, Burke PS. Effects of tooth loading on the periodontal vasculature of the mandibular fourth premolar in dogs. Arch Oral Biol 1981;26:189-95.
30. Boice PA, Niles SM, Dubois LM. Evaluation of proximal contacts with shim stock. J Oral Rehabil 1987;14:91-4.

A comparative study on the accuracy of the devices for measuring the implant stability

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STATEMENT OF PROBLEM. How the ISQ values measured by Osstell™ and Osstell™ Mentor are related, and whether the ISQ values acquired from the two machines changes in accordance with changes in implant stability are not yet fully understood. **PURPOSE.** The aim of this study was to find out correlation between the ISQ values acquired from Osstell™ and Osstell™ Mentor, and to evaluate the clinical effectiveness and accuracy of two devices. **MATERIAL AND METHODS.** Sixty two implants were inserted into 47 patients, and their ISQ values were measured using Osstell™ and Osstell™ Mentor. In the first stage surgery, the ISQ values of forty four implants inserted into thirty five patients were measured. In the second stage surgery, the values of fifty implants inserted into thirty seven patients were measured. The values were analyzed to determine the difference between the mean ISQ values of Osstell™ and Osstell™ Mentor. In addition, the correlation between implants used in the first and second stage of surgery with regard to their types and areas of insertion were analyzed. The difference between the ISQ values of 32 implants in each patient during the first and second stage was analyzed. The statistical assessment was carried out using SPSS V. 12.0 for Win. (SPSS Inc., Chicago, USA). The Pearson correlation coefficient was used to examine the correlation between Osstell™ and Osstell™ Mentor in the first and second stages of surgery, whereas the difference between their ISQ values was evaluated using a paired *t*-test. **RESULTS.** In the first stage, the mean ISQ value for Osstell™ and Osstell™ Mentor was 70.84 and 75.09, respectively, showing a significant difference ($P < .01$). In the second stage, the mean ISQ value of Osstell™ and Osstell™ Mentor was 71.76 and 75.94, respectively, also showing a significant difference ($P < .01$). The difference between the ISQ values in patients in the first and the second stages was significant with both instruments. **CONCLUSION.** The significant difference in the values obtained using the Osstell™ and Osstell™ Mentor between the first and second stages of implant surgery indicates that these values can be a convenient and precise way for evaluating the implant stability in clinical practice. **KEY WORDS.** Osstell, Osstell Mentor, ISQ, RFA, Stability **[J Adv Prosthodont 2009;1:124-8]**

INTRODUCTION

Osseointegration, which was defined by Brånemark[1] in 1985, is essential for the clinical success of the implant-driven restoration.[2] Therefore, measuring the implant stability is an important method for evaluating the success of an implant.[3] The implant stability can be classified into two categories: primary stability, which can be acquired while inserting the implant, and secondary stability, which is obtained during healing and remodeling of the surrounding bone. The primary stability is affected by the quantity and quality of bone that the implant is inserted into, surgical procedure, length, diameter, and form of the implant.[4,5] The secondary stability is the one developed from regeneration and remodeling of the bone and tissue around the implant after insertion but is affected by the primary stability, bone formation and remodeling, etc.[6] Many clinical and experimental methods of measuring osseointegration and implant stability has been developed. Histomorphologic research and removal torque test are classified as destructive methods. Nondestructive methods include percussion test, radiography, cutting torque test while placing implants, Periotest® (Siemens AG, Benshein, Germany), and resonance frequency analysis (RFA).

Although the removal torque test[7,8] is useful for measuring the degree of osseointegration of an implant, its use is limited by the direct tension placed on the interface of an implant and surrounding bone, resulting in possible failure of the implant. A percussion test measures the stability of an implant by simple percussion with the handle of a dental instrument on the implant abutment. However, this method is rather subjective and lacks precision.[9] Radiography provides a useful method for evaluating the quantity and quality of bone in the area for an implant to be inserted before placing the fixture, as well as the quantity and quality of the adjacent marginal bone, suitability of an abutment for prosthodontic treatment and height of the peri-implant bone.[10,11] However, uniform resolution and standardized taking of X-rays are difficult to achieve. In addition, it is difficult to perceive changes in the bone structures and morphology of the implant-bone interface. While Chai *et al.*[12] reported that Periotest® could be an objective and

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reproducible method of evaluating implant stability, Derhami *et al.*[13] concluded that its result lacks consistency since it is susceptible to the height of the abutment, measurement angle, and the distance between the implant and hand piece. Meredith *et al.*[4,14] showed resonance frequency analysis as a means of measuring implant stability. In this method, an L-shaped transducer composed of two piezo-ceramic components is connected to an implant and as the frequency increases from 5k Hz to 15k Hz, the first curved resonance frequency detected is considered as the peak. This can be a useful way of evaluating implant stability because of its objectivity and availability to continuously observe the changes in stability that follows the healing process.[6,15]

Osstell™ (Integration Diagnostics Ltd., Göteborg, Sweden) (Fig. 1) and Osstell™ Mentor (Integration Diagnostics Ltd., Göteborg, Sweden) (Fig. 2) are devices developed for the clinical application of resonance frequency analysis. Osstell™ (electronic method), which was developed first, measures the resonant frequency by connecting the implant to a transducer, whereas Osstell™ Mentor (magnetic method) measures the resonant frequency using the magnetic frequencies by connecting the implant to a Smartpeg™ (Integration Diagnostics Ltd., Göteborg, Sweden). The RFA value is converted into an ISQ (implant stability quotient) that is scaled from 1 to 100.[14,16] The ISQ increases in proportion to the stiffness of the bone-implant interface or with that of the peri-implant bone.[4,6,14]

Currently how the ISQ values measured by Osstell™ and Osstell™ Mentor are related, and whether the ISQ values acquired from the two machines changes in accordance with changes in implant stability are not yet fully understood.

This study examined the correlation between the ISQ values measured by Osstell™ and Osstell™ Mentor as well as the coincidence of the two instruments, and evaluated their utility in practice by investigating the changes in implant stability as a function of time.

Fig. 1. Osstell™ (Integration Diagnostics Ltd., Göteborg, Sweden).

Fig. 2. Osstell™ Mentor (Integration Diagnostics Ltd., Göteborg, Sweden).

MATERIAL AND METHODS

Specimen

Sixty two implants inserted into 47 patients who received implant treatment in the Department of Prosthodontics, College of Dentistry, Dankook University were examined. The patients' age ranged from 23 to 78 years with a mean age of 51. Only two-stage implants were used in this study. In the first stage of surgery, 44 implants placed in 35 patients (28 implants in the mandibles and 16 in the maxillas) were measured. In the second stage of surgery, 50 implants were inserted into 37 patients: 36 implants in the mandible and 14 in the maxilla. Twenty five patients with 32 implants underwent the measurements in both stages. Twelve implants in 10 patients were measured in the first stage only, whereas 18 implants in 12 patients were measured in the second stage only (Table I).

Implants

The implants used for insertion were Replace™ Select Tapered TiUnite (NobleBiocare™ AB, Sweden), Brånemark System® MKIII TiUnite (NobleBiocare™ AB, Sweden), Osseotite® (3i Corp, USA), US II (Osstem, Korea) (Table II).

Thirty implants from Replace™ Select Tapered TiUnite, 11 from Brånemark System® MKIII TiUnite, 2 from US II and 1 from Osseotite® were inserted in the first stage of surgery. Their lengths and diameters are shown in Tables III and IV, respectively.

Table I. Number of patients and implants used in this study

	Number of patients	Number of implants
1st surgery	35 (10)	44 (12)
2nd surgery	37 <12>	50 <18>

( ): Number of patients and implants measured only in the 1st surgery
< >: Number of patients and implants measured only in the 2nd surgery

Table II. Classification of implants used in this study

Implant system	R/S	Brå MKIII	US II	Osseotite®	Sum
N	38	21	2	1	62

Table III. Diameter of the implants used in the 1st surgery

Diameter	NP	RP	WP	Sum
N	2	22	20	44

NP: Narrow Platform, RP: Regular Platform, WP: Wide Flat Platform

Table IV. Length of the implants used in the 1st surgery

Length (mm)	8	10	11.5	13	16	Sum
N	1	19	7	16	1	44

Table V. Diameter of the implants used in the 2nd surgery

Diameter	NP	RP	WP	Sum
N	3	28	19	50

Table VI. Length of the implants used in the 2nd surgery

Length (mm)	10	11.5	13	16	Sum
N	16	10	22	2	50

In the second stage of surgery, 29 implants from R/S, 18 from Brå MKIII, 2 from US II and 1 from Osseotite® were inserted. Tables V and VI, respectively list their lengths and diameters.

Methods of measurement

Using Osstell™ and Osstell™ Mentor, each ISQ value was measured after inserting the implant in the first stage and after removing the cover screw in the second stage surgery. While using the Osstell™, implant system and the transducer which was modified for different diameters were put on to the implant without any contact with surrounding soft tissue.

With Osstell™ Mentor, a Smartpeg™ was connected to the implant in accordance with the diameter, and the measurements were taken from the buccal and mesial sides. Both instruments required three identical values and stored them in the computer.

Analysis

Forty four implants from the first stage surgery and 50 from the second stage of surgery were analyzed to determine the difference between the mean ISQ values of Osstell™ and Osstell™ Mentor. In addition, the correlation between implants used in the first and second stage of surgery with regard to their types and areas of insertion were analyzed.

The difference between the ISQ values of 32 implants in each patient during the first and second stage surgery was analyzed.

The statistical assessment was carried out using SPSS V. 12.0 for Window (SPSS Inc., Chicago, IL, USA). The Pearson correlation coefficient was used to examine the correlation between Osstell™ and Osstell™ Mentor in the first and second stages of surgery, whereas the difference between their ISQ values was evaluated using a paired *t*-test. A paired *t*-test was used to examine the difference between the ISQ values in each surgery in an identical patient.

RESULTS

Table VII shows the ISQ values of the 44 implants obtained using Osstell™ and Osstell™ Mentor.

The measurements from the two instruments were significantly different ($P < .01$) and showed a significant correlation ($r = 0.61$; $P < .01$) (Table VIII). The scatter diagram is shown in Fig. 3.

There was a significant correlation in ISQ values from Osstell™ and Osstell™ Mentor ($P < .01$) for the implants placed in mandible, whereas there was no correlation for those inserted in the maxilla.

As for the correlation with regard to the implant types, there was a significant correlation with Replace™ Select Tapered TiUnite ($P < .01$), whereas there was no correlation with the Brånemark System® MKIII TiUnite.

Table IX lists the ISQ values of the 50 implants obtained with Osstell™ and Osstell™ Mentor.

The ISQ values obtained from Osstell™ and Osstell™ Mentor showed a significant difference ($P < .01$) as well as a significant correlation ($r = 0.65$; $P < .01$), (Table X). Fig. 4 shows the scatter diagram.

Table VII. Mean and SD of the ISQ values measured using Osstell™ and Osstell™ Mentor in the 1st surgery

	Osstell™	Osstell™ Mentor
Range	60 - 83	64 - 88
Mean / SD	70.84 ± 6.13	75.09 ± 6.08

Table VIII. Pearson's correlation coefficient between Osstell™ ISQ and Osstell™ Mentor ISQ in the 1st surgery

	OT	OTM
OT		
OTM	*	

OT : Osstell™ , OTM : Osstell™ Mentor

*denotes a pair of groups significantly different at the 0.01 level

Table IX. Mean and SD of the ISQ measured with Osstell™ and Osstell™ Mentor in 2nd surgery

	Osstell™	Osstell™ Mentor
Range	55 - 84	55 - 88
Mean / SD	74.14 ± 6.64	78.90 ± 7.21

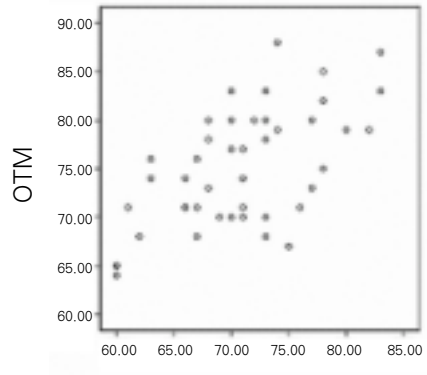

Fig. 3. Scatter diagram of Osstell™ ISQ and Osstell™ Mentor ISQ in the 1st surgery.

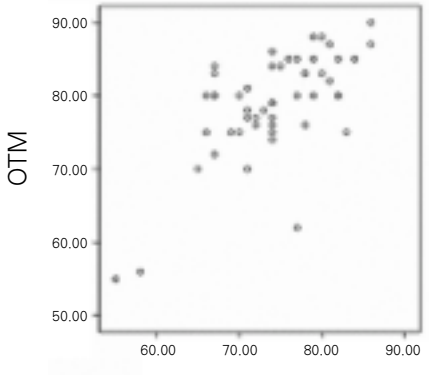

Fig. 4. Scatter diagram between Osstell™ ISQ Osstell™ Mentor ISQ in 2nd surgery.

Table X. Pearson's correlation coefficient between Osstell™ ISQ and Osstell™ Mentor ISQ in the 2nd surgery

	OT	OTM
OT		
OTM	*	

*denotes pair of groups significantly different at the 0.01 level

Table XI. ISQ values of the patients measured in 1st and 2nd surgery

	OT	OTM
1st surgery	71.28 ± 6.22	75.44 ± 6.42
2nd surgery	75.00 ± 5.61	80.06 ± 5.42

Table XII. Results of the paired *T*-test between the 1st surgery ISQ and 2nd surgery ISQ in Osstell™ and Osstell™ Mentor

	OT1	OTM1	OT2	OTM2
OT1				
OTM1				
OT2	*			
OTM2		*		

OT1 : ISQ measured with Osstell™ in the 1st surgery
OTM1 : ISQ measured with Osstell™ Mentor in the 1st surgery
OT2 : ISQ measured with Osstell™ in the 2nd surgery
OTM2 : ISQ measured with Osstell™ Mentor in the 2nd surgery
*denotes pair of groups significantly different at the 0.01 level

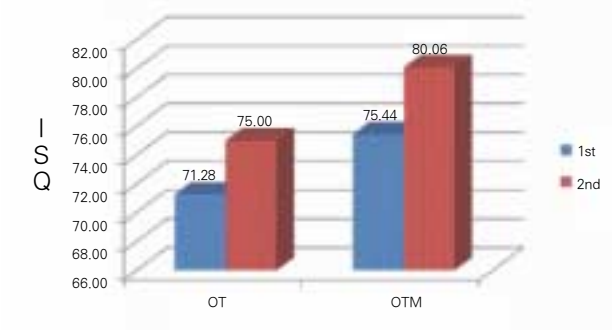

Fig. 5. ISQ value measured with Osstell™ and Osstell™ Mentor in the 1st and 2nd surgery.

The ISQ values of Osstell™ and Osstell™ Mentor exhibited a significant correlation ($P < .01$) regardless of the location of insertion and types of implants inserted.

Table XI lists the ISQ values of the 32 implants inserted into 25 patients obtained with Osstell™ and Osstell Mentor™ during the first and second stage surgery.

The changes in the ISQ values of both Osstell™ and Osstell Mentor™ showed a significant difference ($P < .01$) (Table XII) (Fig. 5).

DISCUSSION

Since the first report by Brånemark in 1969, many clinical and experimental studies have shown that the insertion of an osseointegrated implant is a good treatment for either fully or partially edentulous patients.[17] Clinically, an implant shows an excellent long-term success rate of approximately 90%.[18,19] However, failure can occur due to the unsuitable quantity and quality of bone, infection during the healing process, and an excess load while functioning, etc.[14]

Although clinically severe mobility and obvious bone absorption observed in radiography can guarantee failure of an implant, it is difficult to confirm this without such evidence.[20] RFA is a non-destructive and objective way of assessing the bone-implant interface.[20] It can also measure the change in implant stability as a function of time, and it is useful for determining the critical time for making prosthetics.[15]

According to Nkenke *et al.*[21], RFA is not affected by the PTV (Perio test value) or bone quality, but by bone-to-implant contact. Huang *et al.*[22] also suggested in their histomorphologic study that the bone-to-implant contact and RFA were related. While Friberg *et al.*[23] mentioned that RFA was related to the cutting force when inserting an implant, Huang *et al.*[24] used RFA to demonstrate that the implant stability increases during the healing period and reported it to be a precise and reliable device. Valderrama *et al.*[20] stated that in cases of a low ISQ value, an adequate healing period would be necessary before loading, whereas Friberg *et al.*[25] reported that a low ISQ value indicated failure of an implant weeks before radiographic evidence could be obtained. Glauser *et al.*[16] suggested that RFA could be used to diagnose the possibility of implant failure before the presence of clinical evidence.

The stability of 44 implants were tested, and the results showed that Osstell™ displayed ISQ values ranging from 60 to 83 with an average of 70.84, while those for Osstell™ Mentor ranged from 64 to 88 with a mean value of 75.09. The difference between the two devices was 4.25, which was statistically significant. Regarding the 50 implants tested during their second stage surgery, the ISQ values for Osstell™ and Osstell™ Mentor showed a mean value of 74.14 and 78.90, respectively. The difference was 4.76, which was also significant.

During the second stage surgery, the locations in jaws or types of implants did not have a significant impact on the results of the two devices as observed for the first stage surgery, whereby the implants were inserted in the maxillas with the exception of the Brånemark System® MKIII TiUnite. In first and second stages of surgery, Osstell™ Mentor was likely to give 4 - 5 higher ISQ values than Osstell™ . With the relatively consistent difference between these two devices, Osstell™ and Osstell™ Mentor are objective and can measure the implant stability confidently. These results are somewhat different from Valderrama's report, ISQ values for Osstell™ Mentor were 8 to 12 points higher than those for Osstell™. This appears to be because Valderrama *et al.* used one-stage implants, whereas two-stage implants were used in this study. While a one-stage implant is measured at 2.8 mm above the bone level with Osstell™, Osstell™ Mentor involves connecting the implant to a

SmartPeg™, which causes a larger difference between the two devices. Recognizing this difference in one-stage implants would help in their clinical application.

The changes in ISQ values according to the healing process were recorded in the same patient, while he or she underwent each stage of surgery with the two different devices. The ISQ values measured in first and second stage surgery using the Osstell™ ranged from 60 to 83 with the mean value of 71.28. In the second stage surgery, they ranged from 65 to 85 with the mean value of 75.00, this change was statistically significant. With the Osstell™ mentor, the values ranged from 64 to 85 in the first surgery with an average of 75.44, whereas they ranged from 62 to 88 with an average of 80.06 in the second stage surgery. This difference was statistically significant. This shows that the ISQ value measured in the second stage surgery showed a significant increase compared to that of the first surgery, which is in agreement with Meredith *et al*.[26] and Rasmusson *et al*.[27] This also suggests that observing the ISQ values constantly after placing an implant in the case of a one-stage implant or after the second procedure in the case of a second-stage implant, would be very useful for determining the critical time for a final prosthetic setting.

CONCLUSION

The ISQ values obtained using Osstell™ and Osstell™ mentor were examined. Both instruments showed a significant difference in the ISQ values (4 to 5 points) with a significant correlation ($P < .05$). The healing process increased the ISQ values significantly for both instruments ($P < .05$). Overall, measuring the implant stability with Osstell™ and Osstell™ mentor is objective and reliable, and can be considered a useful method in practice.

REFERENCES

1. Brånemark P, Zarb G, Albrektsson T. Tissue-integrated prostheses. Chicago: Quitessence Publishing; 1985:p11-43.
2. Choi JW, Heo SJ, Chang IT, Koak JY, Han JH, Kim YS, Lee SH, Yim SH. Resonance frequency analysis of implants with anodized surface oxides. J Korean Acad Prosthodont 2004;42:294-300.
3. Zarb GA, Schmitt A. Osseointegration and the edentulous predicament. The 10-year-old Toronto study. Br Dent J 1991;170:439-44.
4. Meredith N. Assessment of implant stability as a prognostic determinant. Int J Prosthodont 1998;11:491-501.
5. Albrektsson T, Zarb G, Worthington P, Eriksson AR. The long-term efficacy of currently used dental implants: a review and proposed criteria of success. Int J Oral Maxillofac Implants 1986;1:11-25.
6. Sennerby L, Meredith N. Resonance frequency analysis: measuring implant stability and osseointegration. Compend Contin Educ Dent 1998;19:493-8, 500, 502; quiz 504.
7. Johansson CB, Albrektsson T. A removal torque and histomorphometric study of commercially pure niobium and titanium implants in rabbit bone. Clin Oral Implants Res 1991;2:24-9.
8. Wennerberg A, Albrektsson T, Andersson B, Krol JJ. A histomorphometric and removal torque study of screw-shaped titanium implants with three different surface topographies. Clin Oral Implants Res 1995;6:24-30.
9. Misch CE. Contemporary Implant Dentistry. St. Louis: CV Mosby;1993,p327-54.
10. Brånemark PI, Zarb G, Albrektsson T. Tissue-integrated prostheses. osseointegration in clinical dentistry. Chicago: Quintessence publishing; 1985,p199-210.
11. Park C, Lim JH, Cho IH, Lim HS. A study of the measurement of the implant stability using resonance frequency analysis. J Korean Acad Prosthodont 2003;41:181-206.
12. Chai JY, Yamada J, Pang IC. *In vitro* consistency of the Periotest instrument. J Prosthodont 1993;2:9-12.
13. Derhami K, Wolfaardt JF, Faulkner G, Grace M. Assessment of the periotest device in baseline mobility measurements of craniofacial implants. Int J Oral Maxillofac Implants 1995;10:221-9.
14. Meredith N, Alleyne D, Cawley P. Quantitative determination of the stability of the implant-tissue interface using resonance frequency analysis. Clin Oral Implants Res 1996;7:261-7.
15. Meredith N, Shagaldi F, Alleyne D, Sennerby L, Cawley P. The application of resonance frequency measurements to study the stability of titanium implants during healing in the rabbit tibia. Clin Oral Implants Res 1997;8:234-43.
16. Glauser R, Sennerby L, Meredith N, Rée A, Lundgren A, Gottlow J, Hämmerle CH. Resonance frequency analysis of implants subjected to immediate or early functional occlusal loading. Successful vs. failing implants. Clin Oral Implants Res 2004;15:428-34.
17. Adell R, Lekholm U, Rockler B, Brånemark PI. A 15-year study of osseointegrated implants in the treatment of the edentulous jaw. Int J Oral Surg 1981;10:387-416.
18. Lekholm U, Gunne J, Henry P, Higuchi K, Lindén U, Bergström C, van Steenberghe D. Survival of the Brånemark implant in partially edentulous jaws: a 10-year prospective multicenter study. Int J Oral Maxillofac Implants 1999;14:639-45.
19. Buser D, Mericske-Stern R, Bernard JP, Behneke A, Behneke N, Hirt HP, Belser UC, Lang NP. Long-term evaluation of non-submerged ITI implants. Part 1: 8-year life table analysis of a prospective multi-center study with 2359 implants. Clin Oral Implants Res 1997;8:161-72.
20. Valderrama P, Oates TW, Jones AA, Simpson J, Schoolfield JD, Cochran DL. Evaluation of two different resonance frequency devices to detect implant stability: a clinical trial. J Periodontol 2007;78:262-72.
21. Nkenke E, Hahn M, Weinzierl K, Radespiel-Tröger M, Neukam FW, Engelke K. Implant stability and histomorphometry: a correlation study in human cadavers using stepped cylinder implants. Clin Oral Implants Res 2003;14:601-9.
22. Huang HM, Pan LC, Lee SY, Chiu CL, Fan KH, Ho KN. Assessing the implant/bone interface by using natural frequency analysis. Oral Surg Oral Med Oral Pathol Oral Radiol Endod 2000;90:285-91.
23. Friberg B, Sennerby L, Meredith N, Lekholm U. A comparison between cutting torque and resonance frequency measurements of maxillary implants. A 20-month clinical study. Int J Oral Maxillofac Surg 1999;28:297-303.
24. Huang HM, Chiu CL, Yeh CY, Lin CT, Lin LH, Lee SY. Early detection of implant healing process using resonance frequency analysis. Clin Oral Implants Res 2003;14:437-43.
25. Friberg B, Sennerby L, Linden B, Gröndahl K, Lekholm U. Stability measurements of one-stage Brånemark implants during healing in mandibles. A clinical resonance frequency analysis study. Int J Oral Maxillofac Surg 1999;28:266-72.
26. Meredith N, Book K, Friberg B, Jemt T, Sennerby L. Resonance frequency measurements of implant stability *in vivo*. A cross-sectional and longitudinal study of resonance frequency measurements on implants in the edentulous and partially dentate maxilla. Clin Oral Implants Res 1997;8:226-33.
27. Rasmusson L, Meredith N, Sennerby L. Measurements of stability changes of titanium implants with exposed threads subjected to barrier membrane induced bone augmentation. An experimental study in the rabbit tibia. Clin Oral Implants Res 1997;8:316-22.

Shear bond strength of veneering porcelain to zirconia and metal cores

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STATEMENT OF PROBLEM. Zirconia-based restorations have the common technical complication of delamination, or porcelain chipping, from the zirconia core. Thus the shear bond strength between the zirconia core and the veneering porcelain requires investigation in order to facilitate the material's clinical use. **PURPOSE.** The purpose of this study was to evaluate the bonding strength of the porcelain veneer to the zirconia core and to other various metal alloys (high noble metal alloy and base metal alloy). **MATERIAL AND METHODS.** 15 rectangular (4x4x9mm) specimens each of zirconia (Cercon), base metal alloy (Tillite), high noble metal alloy (Degudent H) were fabricated for the shear bond strength test. The veneering porcelain recommended by the manufacturer for each type of material was fired to the core in thickness of 3mm. After firing, the specimens were embedded in the PTFE mold, placed on a mounting jig, and subjected to shear force in a universal testing machine. Load was applied at a crosshead speed of 0.5mm/min until fracture. The average shear strength (MPa) was analyzed with the one-way ANOVA and the Tukey's test (α = .05). The fractured specimens were examined using SEM and EDX to determine the failure pattern. **RESULTS.** The mean shear strength (± SD) in MPa was 25.43 (± 3.12) in the zirconia group, 35.87 (± 4.23) in the base metal group, 38.00 (± 5.23) in the high noble metal group. The ANOVA showed a significant difference among groups, and the Tukey's test presented a significant difference between the zirconia group and the metal group. Microscopic examination showed that the failure primarily occurred near the interface with the residual veneering porcelain remaining on the core. **CONCLUSION.** There was a significant difference between the metal ceramic and zirconia ceramic group in shear bond strength. There was no significant difference between the base metal alloy and the high noble metal alloy.
KEY WORDS. Zirconia ceramic, Delamination, Core-veneer ceramic, Shear bond strength, Failure mode **[J Adv Prosthodont 2009;1:129-35]**

INTRODUCTION

For the past 40 years the porcelain-fused-to-metal systems have been extensively used in fixed partial dentures (FPDs) and still represents the gold standard.[1] The advantages of the PFM systems are to combine the fracture resistance of the metal substructure with the esthetic property of the porcelain. However, recently the increasing demand for esthetic restorations as well as the questionable biocompatibility of some dental metal alloys has accelerated the development and improvement of metal-free restorations.[2]

In the early 1990s, yttrium oxide partially stabilized tetragonal zirconia polycrystal (Y-TZP) was introduced to dentistry as a core material for all-ceramic restoration and has been applied to clinical use through the CAD/CAM technique. Due to the transformation toughening mechanism, Y-TZP has been shown to have superior mechanical properties compared to other all-ceramic systems. (flexural strength of 900 - 1200 MPa, and fracture toughness of 9 - 10 MPa · $m^{1/2}$)[3] Due to its mechanical property, zirconia has enough strength to withstand the high occlusal stress.[2,4] Therefore, it can be used in extensive all-ceramic FPDs having more than 4 units.[5]

According to clinical studies, the Y-TZP core ceramic exhibited high stability as a framework material. No fractures of the zirconia framework have been reported so far.[6-8] However, delamination or minor chip-off fracture of veneering porcelain was described as the most frequent reason for the failures of zirconia FPDs. The incidence of veneer fractures in zirconia FPDs was significantly higher compared with those in metal-ceramic FPDs.[6] Therefore, the bond between core and veneer or the veneer material itself is one of the weaknesses in layered zirconia based restorations and plays a significant role in their long-term success.[9]

The adhesion mechanism between metal and porcelain is believed to be the micro-mechanical bond, compatible coefficient of thermal expansion (CTE) match, van der Waals force, and mainly the suitable oxidation of metal and interdiffusion of ions between the metal and porcelain.[10] Data presented in literature has shown the bond strength of ceramic or resin to metal substrates to be in the range of 54 - 71 MPa[11], and a sufficient bond for metal-ceramic has been accepted when the fracture stress is greater than 25 MPa.[12,13]

However, the bonding mechanisms for veneering ceramic to the zirconia are up to now unclear. According to investigations

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on the wettability of the zirconia core with the veneering ceramic, micromechanical interactions were merely regarded. Moreover, there are less information available on the bond strength values between the all-ceramic core and veneering materials, and there exists no accurate test method for obtaining information on core/veneer adhesion in bi-layered all-ceramic materials in dentistry.

Many variables may affect the zirconia core-veneer bond strength; such as the surface finish of the core, which can affect mechanical retention; residual stress generated by mismatch in coefficient of thermal expansion (CTE); development of flaws and structure defects at core-veneer interface; and wetting properties and volumetric shrinkage of the veneer.[14] The cause of core-veneer bond failure may be related to multiple factor.

The purpose of this study was to evaluate the shear bond strength of zirconia and metal alloys with their corresponding veneering porcelains. Scanning electron microscopy (SEM) was used to classify the failure pattern, and the interface chemistry was evaluated using energy dispersive X-ray microanalysis (EDX).

MATERIAL AND METHODS

The materials tested for this study were listed in Table I.

Three types of core-veneer combinations (N = 45, n = 15/group) were fabricated by one dental technician according to the manufacturer's instructions. The corresponding porcelains for each core were veneered to zirconia (Group I), base metal alloy (Group II), and high noble metal alloy (Group III).

1. Preparation of the zirconia core-veneer specimens (Group I)

Fully sintered Cercon®blocks (Degudent, Hanau, Germany) (23 × 15 × 9 mm) were used for this study. The Cercon® Base blocks were sandblasted with 110 μm Al_2O_3 particles at 2.5 bar pressure according to the manufacturer's pre-treatment recommendation. The bars were steam-cleaned and air-dried. After a thin liner (Cercon® Ceram Kiss Liner, Degudent Hanau, Germany) layer was fired, the veneering ceramic (Cercon® ceram kiss, Degudent, Hanau, Germany) was built up to the final dimension (thickness of 3 mm) according to the firing program of the manufacturer (Austromat 3001, Dekema Dental-Keramiköfen GmbH & Co, Freilassing, Germany). Due to the shrinkage of porcelain, three separate firings were required to establish the correct dimension.

The blocks were cut in a sawing machine with diamond wheels to 15 bars (4 × 4 × 12 mm: 9 mm core / 3 mm veneer) (Fig. 1). After surface examinations with a magnifying glass, the intact specimens were selected.

2. Preparation of the metal core-veneer specimens (Group II, III)

The bars (4 × 4 × 9 mm) were cast in Ni-Cr base metal ceramic alloy (Tillite, Talladium Inc., LA,USA), high noble metal ceramic alloy (Degudent H, Degudent, Hanau, Germany), according to the manufacturer's instructions. The veneering ceramic (Vita VM13, VitaZahnfabrik, BadSäckingen, Germany) was built up to thickness of 3mm after degassing, second layer of opaque firing procedures. All specimens were examined and excess porcelain was removed using a high speed diamond bur with a low-speed handpiece. The final dimensions of bars of group II and III were identical to those of group I.

3. Shear bond strength test (SBS test)

3.1 Mounting

Each bar was embedded in the customized polytetrafluoroethylene (PTFE) mold using PMMA resin. Every effort was made to place the core-veneer interface on the same level as the upper plane of the mold (Fig. 2).

The core-veneer interface of the specimen was placed on the same level as the upper plane of the mold using discs for horizontal plane adjustment.

3.2. Shear bond strength test (SBS test)

The PTFE molds holding the specimen were first inserted into a custom-made shear test jig (Instron, Canton, MA, USA), and the jig was secured in a bench vice. Then, the specimens

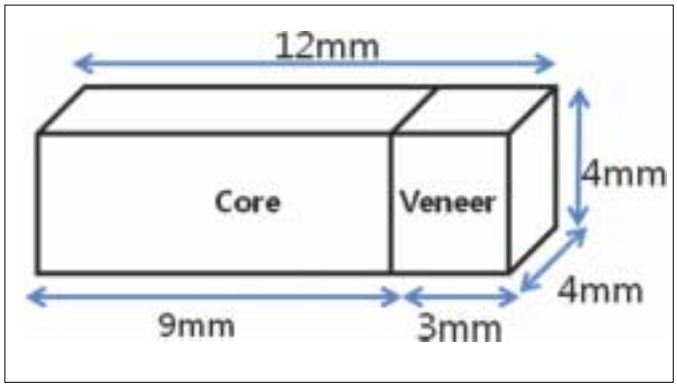

Fig. 1. Final dimensions of specimens.

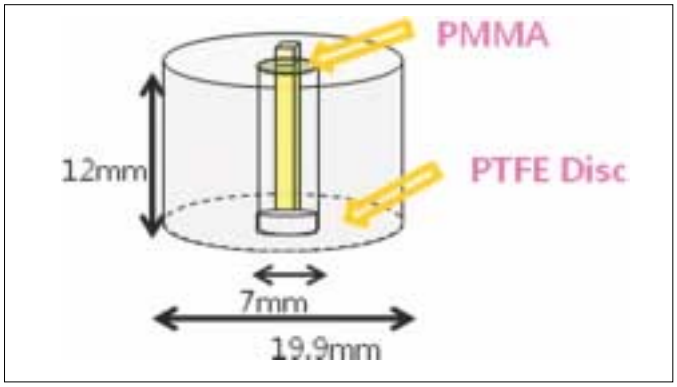

Fig. 2. Schematic diagram of specimen embedded in PTFE molds.

were stressed in shear at a constant crosshead speed of 0.5 mm / min15 until failure occurred using an Instron Universal Testing machine (Model 3345, Instron, Canton, MA, USA). The test was carried out at room temperature. Force was applied to the specimen so that shear load was exerted adjacent to and directly to the bonding interface (Fig. 3).

Load deflection curves and ultimate load to failure were recorded automatically and displayed by the computer software of the testing machine (Bluehill® Lite software, Instron Canton MA, USA). Shear bond force was recorded in Newtons, and the average shear bond strength (MPa) was calculated through dividing the load (N) at which failure occurred by the bonding area (mm^2).

$$\text{Shear stress (MPa)} = \text{Load (N)} \div \text{Area } (mm^2)$$

4. SEM and EDX analyses

To determine the mode of failure, the broken specimens were examined under scanning electron microscopy (SEM) (s-4700, Hitachi, Japan) under × 30 to × 1000 magnifications.

The definition for failure modes are presented in Table II. And the chemical composition at the fractured core was analyzed using energy dispersive X-ray microanalysis (EDX).

5. Statistical analysis

Statistical analysis was carried out using statistical software (SPSS 14.0, SPSS, Inc., Chicago, IL, USA). The data was analyzed using the one-way analysis of variance test (ANOVA) and the Tukey' s multiple comparison test. The test was performed at a level of significance of 0.05.

RESULTS

Table III shows the mean shear strength of the core-veneer interface of 3 groups. The highest mean shear strength was recorded for group III (38.00 ± 5.23 MPa) followed by group II (35.87 ± 4.23 MPa) and group I (25.43 ± 3.12 MPa).

The one-way ANOVA showed a significant difference for the shear bond strength among the materials tested at the significant level of 0.05 (Table IV). The Tukey' s multiple comparisons of the test were computed to make all pair-wise comparisons

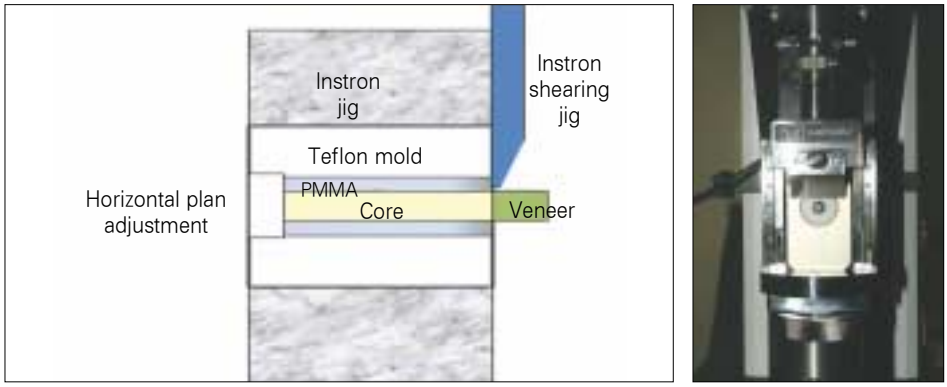

Fig. 3. Schematic diagram of SBS test.

Table I. Brand, composition & manufacturers of materials selected for this study

Brand	Composition	Manufacturer
Cercon® base	ZrO_2 92 vol %, Y_2O_3 5 vol %, HfO_2 2 vol %, Al_2O_3 and silica < 1 vol %	DeguDent, Hanau, Germany
Cercon® Ceram Kiss Liner	Selenium, Feldspathic porcelain	DeguDent, Hanau, Germany
Cercon® Ceram kiss dentin / enamel	Feldspathic veneering ceramic (SiO_2 60.0-70.0; Al_2O_3 7.5-12.5; K_2O 7.5-12.5; Na_2O 7.5-12.5)	
Tillite alloy	Predominantly based alloy(Ni-Cr)	Talladium Inc. ,CA, USA
Degudent H	Au 84.4%, Pt 8.0%, Pd 5.0%, Ag 2.5%, In 2.5%, others < 1%	DeguDent, Hanau, Germany
Vita VM13 Opaque	Feldspathic veneering ceramic	Vita Zahnfabrik, Bad Säckingen, Germany
Vita VM13 Dentin	Opaque: SiO_2 40-44, Al_2O_3 11-14, K_2O 4-6, CeO_2 13-16 Dentin: $SiO_2$59-63, $Al_2O_3$711-14, K_2O 9-11, Na_2O 4-6	

Table II. Definitions of different failure modes

Failure type	Definition
Adhesive failure	Complete delamination of veneering porcelain from core material
Cohesive failure	Fracture occurs completely and only within veneering porcelain or within core material
Mixed adhesive / cohesive failure	Fractured surfaces are within veneering porcelain with areas of core materials exposed indicating localized adhesive failure

Table III. Mean shear bond strength (MPa) with standard deviations in parentheses of the different materials

Group	N	SBS mean (MPa)	95%-CI	Max	Min	Sign*
I	15	25.43 (3.12)	23.70 - 27.15	30.45	18.08	a (.000)
II	15	35.87 (4.23)	33.53 - 38.21	45.14	30.39	b (.404)
III	15	38.00 (5.23)	35.10 - 40.90	45.03	29.48	b (.404)

*Values with the same letter are not statistically different using Tukey' s test at $P < .05$

Table IV. One-way ANOVA data of shear strength of groups

Source	Sum of squares	Mean squares	Df	F-ratio	*P* value
Inter-group	1358.596	679.298	2	37.065	.000
Intra-group	769.739	18.327	42		
Sum	2128.335		44		

among the 3 groups in this study. These comparisons are listed in the last column of Table III. The P values of the different comparisons did not show significant difference between the metal groups (II, III). But the zirconia group (group I) had significantly lower values than Group II and Group III ($P < .05$).

Upon examination under the SEM (× 30), the zirconia group exhibited mixed cohesive / adhesive failures with only small remnants of porcelain attached to the core material. A SEM images of the zirconia group under high magnification showed many small pores in the veneering porcelain, where fracture originated and propagated in the veneering ceramics. Careful examination exhibited a thin layer of veneering porcelain covering the fracture surface(Fig. 4A, 4B, and 4C). Also, EDX results revealed that fractured zirconia surfaces were mainly covered by liner or veneer material and some zirconia crystals were exposed (Fig. 7A).

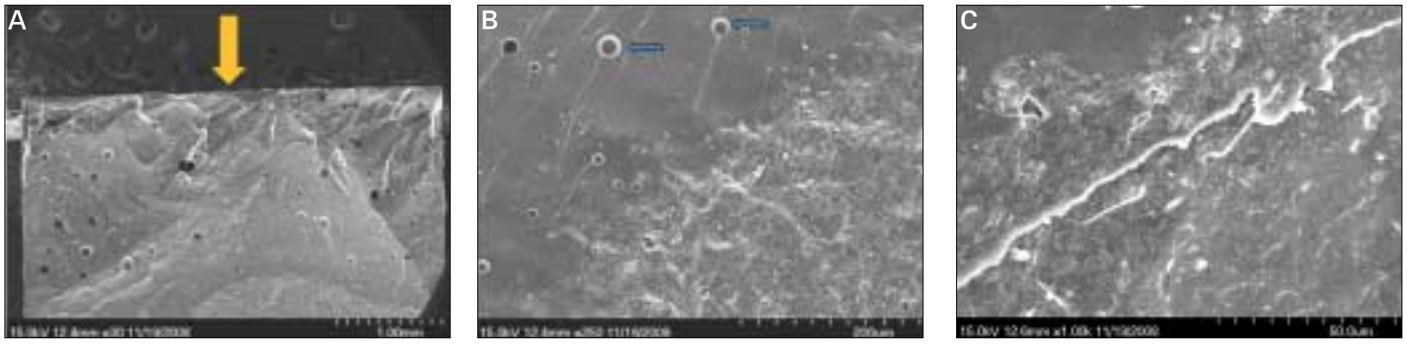

Fig. 4. SEM image of zirconia-veneer group (Group I).
(A) The arrow indicates the direction of load. The loaded side demonstrates cohesive failure within the veneering porcelain (original magnification × 30), (B) Note many pores within veneering `porcelain (arrow), where fracture originated. The fractured Cercon Ceram kiss veneer demonstrates multiple cracks extending in a vertical direction (Hackle patterns) (original magnification × 250), (C) High magnification SEM image exhibited a very thin layer of porcelain covering zirconia grains (original magnification × 1000).

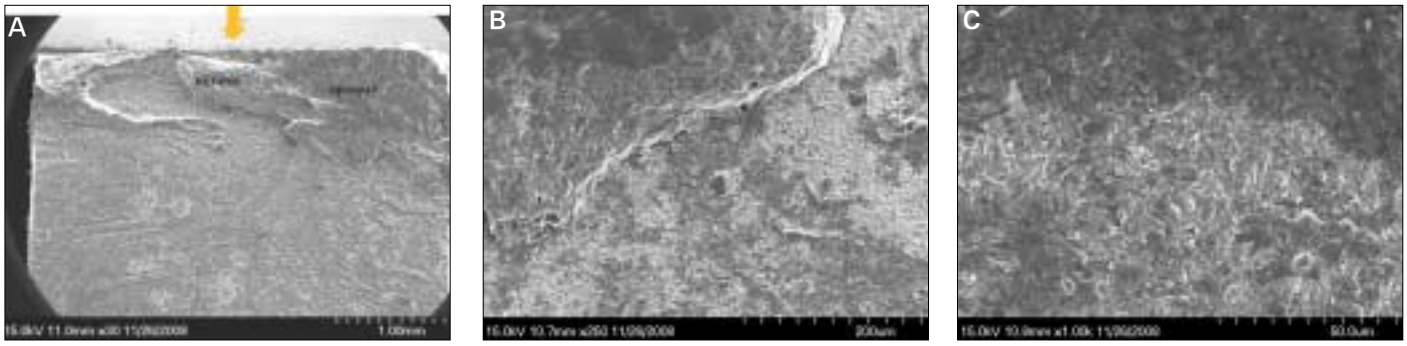

Fig. 5. SEM image of base metal alloy-veneer group (Group II).
(A) The arrow indicates the direction of load. The loaded side demonstrates cohesive failure within the veneering porcelain (original magnification × 30), (B) Interface of the veneering porcelain and the metal core (original magnification × 250), (C) High magnification SEM image exhibited an opaque layer and an oxide layer (original magnification × 1000).

Some base metal alloy specimens revealed mixed cohesive / adhesive failures with porcelain attached to the loaded side of the core. But many specimens presented a cohesive failure between the metal oxide and alloy, with a thin metal oxide layer covering areas of the fractured surface. Under high magnification evaluation, there were many pores in veneering porcelain and internal pores acted as the fracture origin (Fig. 5A, 5B and 5C). Fractographic analysis by SEM showed that the fracture origin in the veneering porcelain was mostly on the loaded surface. EDX results revealed that the fractured core surface mainly failed in oxide layer or at oxide layer / metal interface (Fig. 7B).

In high noble metal alloy group (III), SEM image and EDX analysis of the fractured surface showed that cohesive failures within the veneering porcelain were predominant (Fig. 6).

DISCUSSION

The bond strength measurement of metal ceramic system was standardized by the Organization of Standardization through the Schwickerath crack initiation test (three point bending test), and the mean debonding strength / crack initiation strength should be greater than 25 MPa to meet the ISO requirement.[13] Due to the brittleness of all-ceramic core materials this test set-up cannot be applied to all-ceramic multilayered system.[16] To date an adequate standardized test set-up and a minimum required bond strength for bi-layered all-ceramic materials has not been determined.[17]

In a survey of the literature, few articles utilized various bond strength test methods for all-ceramic core and veneering ceramic, such as the shear bond strength test[17-20], three and four point loading test[21], biaxial flexure strength test[14], and the

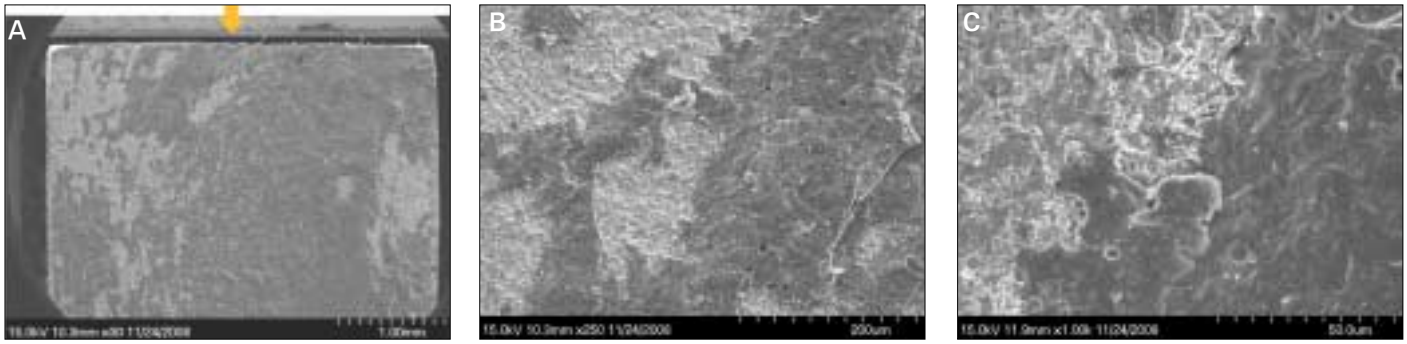

Fig. 6. SEM image of high noble metal alloy-veneer group (Group III)
(A) The arrow indicates the direction of load (original magnification × 30), (B) Predominance of cohesive failure (original magnification × 250), (C) High magnification SEM image exhibited an opaque layer and oxide layer (original magnification × 1000).

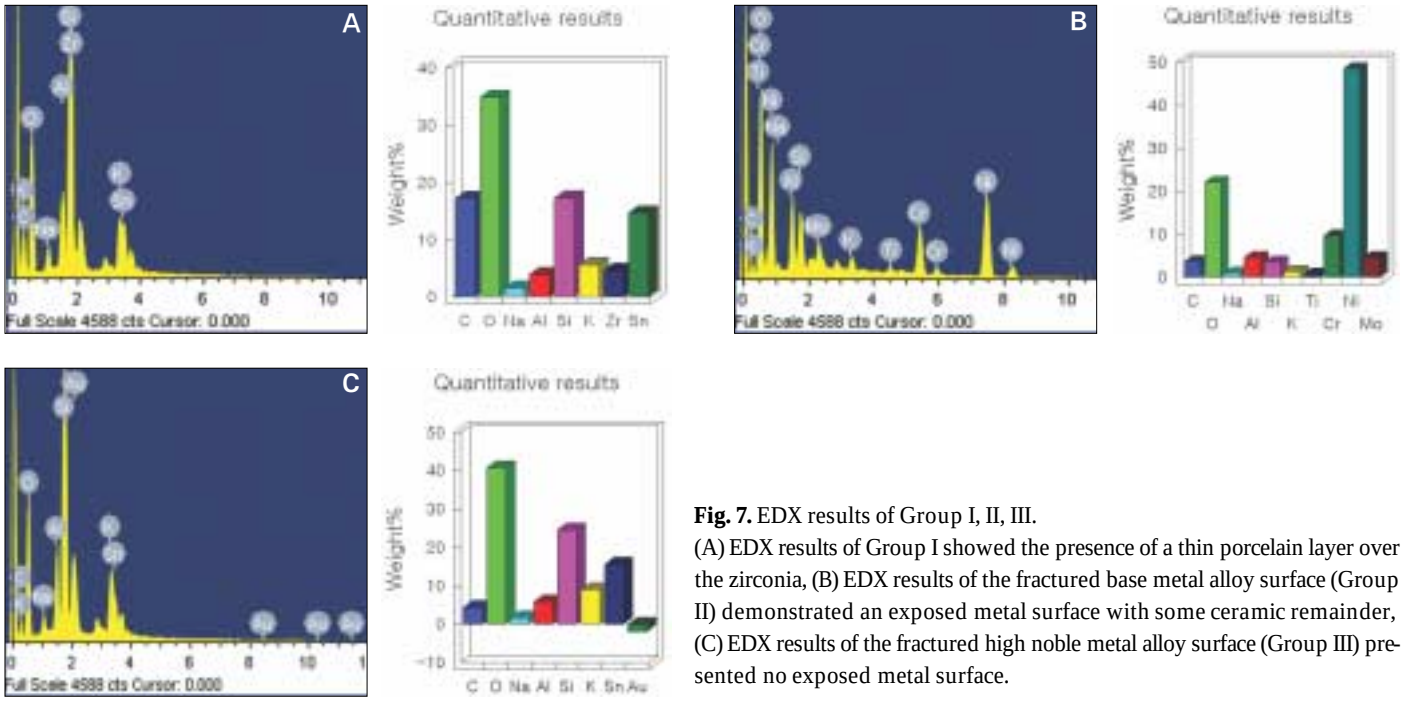

Fig. 7. EDX results of Group I, II, III.
(A) EDX results of Group I showed the presence of a thin porcelain layer over the zirconia, (B) EDX results of the fractured base metal alloy surface (Group II) demonstrated an exposed metal surface with some ceramic remainder, (C) EDX results of the fractured high noble metal alloy surface (Group III) presented no exposed metal surface.

microtensile bond strength test.[19,22,23,28] However, each test has a common limitation which is the difficulty in determining the core-veneer bond strength from applied force at failure on the sample in the specific test setup. In this study, the shear bond strength test method was selected because of its simplicity, such as the ease of specimen preparation, simple test protocol and the ability to rank different products according to bond strength values. But the SBS test has some disadvantages such as high standard deviations, occurrence of non-uniform interfacial stresses, and the influence from specimen geometry. Therefore, the standardization of specimen preparation, cross-sectional surface area, rate of loading application are important for improving the clinical usefulness of SBS test.

The specimens tested in this study were fabricated in rectangular forms so as to standardize the cross-sectional area easily. But there are some limitations in the methodology of this study. The first problem lies in the fixation of the test specimens by embedding in the customized PTFE mold using PMMA resin. When the strength of PMMA resin is weaker than the metal-veneer bond strength (especially the high noble metal-veneer group), failures occurred within the PMMA resin. Thus, improved method for fixation of specimens should be required. And the other limitation is that the specimens had to be custom fabricated and subjected to grinding, which may have produced some flaws or cutting defects in the specimens.

In previous studies, Dundar *et al.*[18] reported shear bond strength in the range of 23 - 41 MPa and Al-Dohan[17] reported shear bond strength in range of 22 - 31 MPa for commercially available core-veneer all-ceramic systems. In this study, the SBS value of veneering ceramic to zirconia core was 25.43 MPa, confirming the findings of previous studies. However unlike in Al-Dohan' s study[17], our study' s results indicate a significant difference in mean SBS values between the zirconia group and metal groups. This difference in findings could be attributed to many factors, such as study design, methodology, skill and experience of the operator, and different properties of different materials.

Although there were no significant differences in mean SBS values between base metal alloy and high noble metal alloy to the corresponding veneer, the value of the high noble alloy group(38 MPa) was higher than that of the base metal group (35. 87 MPa).

The longevity of metal ceramic restorations depends on reliable bonding between metal and ceramic, primarily produced by the oxide layer.[10] If the oxide layer is absent or thin, it would be completely eliminated during ceramic sintering, resulting in poor bonding. However, a heavy oxide layer should be avoided because it has poor cohesive strength[24] it would obstruct the mechanical bond. According to the manufacturer, thicker oxide layers occur in nickel- and cobalt- based alloy because they contain elements that easily oxidate during the initial step of oxidation. Therefore, the initial oxidation step is not recommended. In this study, initial oxidation step was performed on the both base metal alloy and high noble metal alloy, which may cause the low values of SBS for base metal alloy group.

The predominance of cohesive failures of ceramic in the high noble metal group (Group III) suggests that the adhesive zone (opaque ceramic / oxide layer interface, oxide layer and oxide layer / alloy interface) had higher strength than that of the ceramic. In contrast, for the base metal alloys, a predominance of interface failures was noted, suggesting that the oxide layer was weaker than that of the veneering ceramic.

Understanding the failure mechanics of dental ceramic is one of the key points in developing stronger ceramic materials. Fractography has always been a powerful tool in understanding the failure mechanics of brittle materials such as dental ceramics. Identifying location, size, and types of crack initiation illustrates how cracks start, propagate, and extend to a macroscopic level, ending ultimately in fracture restoration.[25]

SEM evaluation revealed that the fracture originated in veneering porcelain in both the zirconia and metal ceramic groups. The failure modes of the specimens from the metal ceramic and zirconia groups suggest the importance of the mechanical properties of the veneering porcelain, since cracks initiated in the veneering porcelain. It is possible that pores and internal defects of veneer lead to the initiation of fracture. Thus, the fabrication technique has critical procedures, such as layering, firing, surface finishing, and polishing of veneering porcelain.[26] In addition, the strength of the veneering porcelain, which is also related to the degree of crystallinity of the veneering porcelain, is paramount to the longevity of the restorations.[27]

Recently, a new generation of ceramic has been introduced for the veneering zirconia framework adopting the pressing technology. The advantages of this system are simplicity, quickness, and defect-free structures. Also the higher tensile strength of these press-on veneers, in addition to their superior interface quality and higher bond strength with zirconia, makes them optimal materials for application.[28] Thus further studies comparing different veneering techniques, such as the conventional layering technique and pressed-on veneering technique, are needed.

The results of this study are in agreement with the observed clinical behavior of zirconia based system. In some clinical studies, the mechanical failures of zirconia- based systems were in the form of minor cohesive porcelain chipping. Improving the bonding strength of zirconia core-veneer and the strength of veneering porcelain may reduce the failure due to chipping or delamination of veneering porcelain.

As the limitation of this study was the fact that a static test was performed in a dry environment. Water would be constantly

present in the actual oral environment, which would undergo repeated temperature and pH changes. According to most studies on the bond strength, the actual bond strength would be lower than expected since the bond strength would decrease further with thermocycling or artificial aging.[29] Therefore thermocycling or artificial aging procedures should be included in subsequent studies.

CONCLUSION

Within the limitations of this study, the following conclusions can be drawn.

1. There was no significant difference among metal-ceramic groups (Group II vs Group III) in the shear bond strength. ($P < .05$)
2. There was a significant difference between the metal ceramic groups and zirconia group in the shear bond strength ($P < .05$).
3. Surface analysis of failure modes showed combined failure modes: cohesive failures in the veneer (loaded side) and adhesive failures at the core veneer interface (unloaded side). SEM evaluation showed that the fractures originated in the veneering porcelain in both the zirconia and metal groups and the fracture origin in the veneering porcelain was mostly on the loaded surface. In the case of interfacial fractures, a thin layer of the veneering ceramic or oxide layer remained on the core materials.

REFERENCES

1. Tan K, Pjetursson BE, Lang NP, Chan ES. A systemic review of the survival and complication rates of fixed partial dentures (FPDs) after an observation period of at least 5 years. Clin Oral Implants Res 2004;15:654-66.
2. Raigrodski AJ. Contemporary materials and technologies for all-ceramic fixed partial dentures: a review of the literature. J Prosthet Dent 2004;92:557-62.
3. Guazzato M, Albakry M, Ringer SP, Swain MV. Strength, fracture toughness and microstructure of a selection of all-ceramic materials. Part II. Zirconia based dental ceramics. Dent Mater 2004;20:449-56.
4. Tinschert J, Natt G, Mohrbotter N, Spiekermann H, Schulze KA. Lifetime of alumina- and zircoina ceramics used for crown and bridge restorations. J Biomed Mater Res B Appl Biomater 2007;80:317-21.
5. Sundh A, Sjogren G. A comparison of fracture strength of yttrium-oxide-partially-stabilized zirconia ceramic crowns with varying core thickness, shapes and veneer ceramics. J Oral Rehabil 2004;31:682-8.
6. Sailer I, Pjetursson BE, Zwahlen M, Hammerle CHF. A Systematic review of the survival and complication rates of all-ceramic and metal-ceramic reconstructions after an observation period of at least 3 years. Part II: Fixed partial prostheses. Clin Oral Implants Res 2007;18:86-96.
7. Sailer I, Feher A, Filser F. Prospective clinical study of zirconia posterior fixed partial dentures: 3-year follow-up. Quintessence Int 2006;37:685-93.
8. Raigrodski AJ, Chiche GJ, Potiket N. The efficacy of posterior three-unit zirconium-oxide-based ceramic fixed partial denture prostheses: a prospective clinical pilot study. J Prosthet Dent 2006;96:237-44.
9. Guazzato M, Proos K, Sara G, Swain MV. Strength, reliability, and mode of fracture of bilayered porcelain/core ceramics. Int J Prosthodont 2004;17:142-9.
10. Anusavice KJ. Phillips' science of dental materials. 11th Ed. Philadelphia:W.B. Saunders; 2003, p621-54.
11. Ozcan M, Niedermeier W. Clinical study on the reasons and location of the failures of metal-ceramic restorations and survival of repairs. Int J Prosthodont 2002;15:299-302.
12. Craig RG, Powers JM. Restorative dental materials 11th ed. Mosby; 2002.
13. ISO 9693 Metal-ceramic bond characterization (Schwickerath crack initiation test) Geneva, Switzerland: International Organization for Standardization; 1999.
14. Isgro G, Pallav P, van der Zel JM, Feilzer AJ. The influence of the veneering porcelain and different surface treatments on the biaxial flexural strength of a heat-pressed ceramic. J Prosthet Dent 2003;90:465-73.
15. Hara AT, Pimenta LA, Rodrigues AL Jr. Influence of cross-head speed on resin-dentin shear bond strength. Dent Mater 2001;17:165-9.
16. Albakry M, Guazzato M, Swain MV. Fracture toughness and hardness evaluation of three pressable all-ceramic dental materials. J Dent 2003;31:181-8.
17. Al-Dohan HM, Yaman P, Dennison JB, Razzoog ME, Lang BR. Shear strength of core-veneer interface in bi-layered ceramics. J Prosthet Dent 2004;91:349-55.
18. Dündar M, Ozcan M, Cömlekoglu E, Güngör MA, Artunç C. Bond strengths of veneering ceramics to reinforced ceramic core materials. Int J Prosthodont 2005;18:71-2.
19. Dündar M, Ozcan M, Gökçe B, Cömlekoglu E, Leite F, Valandro LF. Comparison of two bond strength testing methodologies for bilayered all-ceramics. Dent Mater 2007;23:630-6.
20. Guess PC, Kulis A, Witkowski S, Wolkewitz M, Zhang Y, Strub JR. Shear bond strengths between different zirconia cores and veneering ceramics and their susceptibility to thermocycling. Dent Mater 2008;24:1556-67.
21. White SN, Miklus VG, McLaren EA, Lang LA, Caputo AA. Flexural strength of a layered zirconia and porcelain dental all-ceramic system. J Prosthet Dent 2005;94:125-31.
22. Aboushelib MN, de Jager N, Kleverlaan CJ, Feilzer AJ. Microtensile bond strength of different components of core veneered all-ceramic restorations. Dent Mater 2005;21:984-91.
23. Aboushelib MN, Kleverlaan CJ, Feiler AJ. Microtensile bond strength of different components of core veneered all-ceramic restorations. Part II: zirconia veneering ceramics. Dent Mater 2006;22:857-63.
24. de Melo RM, Travassos AC, Neisser MP. Shear bond strengths of a ceramic system to alternative metal alloys. J Prosthet Dent 2005;93:64-9.
25. Mecholsky J. Fractography: Determining the sites of fracture initiation. Dent Mater 1995;11:113-6.
26. Kelly JR, Giordano R, Pober R, Cima MJ. Fracture surface analysis of dental ceramics: clinically failed restorations. Int J Prosthodont 1990;3:430-40.
27. Quinn JB, Sundar V, Lloyd IK. Influence of microstructure and chemistry on the fracture toughness of dental ceramics. Dent Mater 2003;19:603-11.
28. Aboushelib MN, Kleverlaan CJ, Feilzer AJ. Microtenisile bond strength of different componentsof core veneered all-ceramic restorations. Part 3: double veneer tenchnique. J Prosthodont 2008;17:9-13.
29. Sobrinho LC, Cattell MJ, Glover RH, Knowles JC. Investigation of the dry and wet fatique properties of three all-ceramic crown systems. Int J Prosthodont 1998;11:255-62.

Results of immediate loading for implant restoration in partially edentulous patients: a 6-month preliminary prospective study using SinusQuick™ EB implant system

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STATEMENT OF PROBLEM. Many dental clinicians are concerned about immediate loading of inserted implants. However, there have been few clinical studies surveying the success rates of immediate loading, based on Korean implant systems. **PURPOSE.** The aim of this study was to evaluate the outcome of immediate functional loading of the implant (SinusQuick™ EB, Neobiotech Co., Seoul, Korea) in partially edentulous maxilla or mandible. **MATERIAL AND METHODS.** Total 15 implants were placed. Within 2 weeks after implant insertion, provisional implant-supported fixed partial dentures were delivered to the patients. Quantitatively, marginal bone loss was measured at the time of immediate loading, after 3-months of continued loading and at the last follow-up. The mean follow-up period was 4.8 months. **RESULTS.** Mean marginal bone loss from implant surgery to early loading, 3-months follow-up and last follow-up was 0.03 $\pm$ 0.07 mm, 0.16 $\pm$ 0.17 mm and 0.29 $\pm$ 0.19 mm. No implant failed up to 6 months after insertion, resulting in a 100% survival rate. **CONCLUSION.** Immediate loading exhibited high success rate in partial edentulism for up to 6 months. Well-controlled long term clinical studies with large sample size are necessary to confirm this finding. **KEY WORDS.** Immediate loading, Partially edentulous, Dental implant, Prospective clinical study, Marginal bone loss **[J Adv Prosthodont 2009;1:136-9]**

INTRODUCTION

Traditionally, implant treatment is based on a 2-stage protocol with a healing period of 3-6 months during which the implants are submerged to achieve osseointegration.[1] Recently, this clinical suggestion has been challenged. Numerous practitioners now advocate immediate or early loading of implants.[2] The advantages of immediately loaded implants are clear: they require shorter treatment periods and allow immediate recovery of function and esthetics.[3]

High success rate of immediately loaded implants in humans was first documented in the middle 1980s. The 88% cumulative success rate on 1739 immediately loading implants was suggested.[4] The clinical performance and prognosis of the single-stage surgical protocol are known to be comparable to the traditional 2-stage method.[5] There are some articles reporting a cumulative survival rate of 95%, which investigated immediately loaded single implants.[6,7] Results from these studies suggest that immediate loading could achieve equal success rates as those found in delayed loading.

It is also known as a common claim that treatment with immediate loading improved patient satisfaction and was cost effective although no scientific evidence was presented to support.[8] However, advantages of early or immediate loading as mentioned may be offset by an increased risk of implant failure. It was reported that immediately loaded implants were approximately 3 times more likely to fail within 1 year of placement.[3] Furthermore, there have been few clinical studies investigating the success or failure rates of immediate loading based on Korean implant systems.

The aim of this preliminary prospective study was to evaluate the outcome of immediate functional loading in partial edentulism, using SinusQuick™ EB (Neobiotech Co., Seoul, Korea) implant system.

MATERIAL AND METHODS

Four subjects (2 smokers and 2 non-smokers) recruited from a population of patients under routine care at Seoul National Bundang Hospital were enrolled in the study. The patients were selected according to the following inclusion criteria: they were in normal general health with sufficient bone to allow the placement of implants at least 7 mm length. Patients with high masticatory or parafunctional forces were

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excluded. The mean age of the subjects was 50.3 years (range from 39 to 65 years), with a gender distribution of 100% men. Informed written consent was obtained from all subjects following approved institutional review board guidelines for clinical research.

Surgery was performed under local anesthesia (1 : 100,000 Epinephrine) or conscious intravenous sedation with 1% Propofol solution and Midazolam. After a crestal incision, a mucoperiosteal flap was elevated. Implants were inserted according to the procedures recommended by manufacturers. During the period from April to October 2009 (range from 2 to 6 months), 15 SinusQuick™ EB implants were installed in patients' jaws. Immediate loading was applied to the implants showing implant stability quotient (ISQ) and insertion torque values that were more than 60 and 35 Ncm, respectively. Immediate loading is defined as provisional or final implant-supported restoration delivered within 2 weeks.[9] Therefore, provisional implant-supported fixed partial dentures were delivered within 2 weeks (Fig. 1). The patients were instructed in soft diet and thorough oral hygiene care. The definite restoration was performed approximately 12 weeks after implant insertion.

Periapical radiographs were taken using commercially available film holders and a paralleling imaging technique during the investigating period. In each patient, peri-implant marginal bone level was evaluated by IMPAX® (Agfa Co., Mortsel, Belgium) system of periapical radiographs. Measurements were recorded at the time of surgery, immediate loading, after 3-months of continued loading, and at the last follow-up (Fig. 2). Marginal bone height was determined on these images by measuring the distance from a reference point, defined as the platform of the implant (Fig. 2), to the most coronal point of bone-to-implant contact on both the mesial and distal sides of the implant. A single value for marginal bone height was then calculated by obtaining the mean of these two measurements for each implant.

The definition of implant success was based on the following clinical and radiologic criteria: 1) absence of clinically

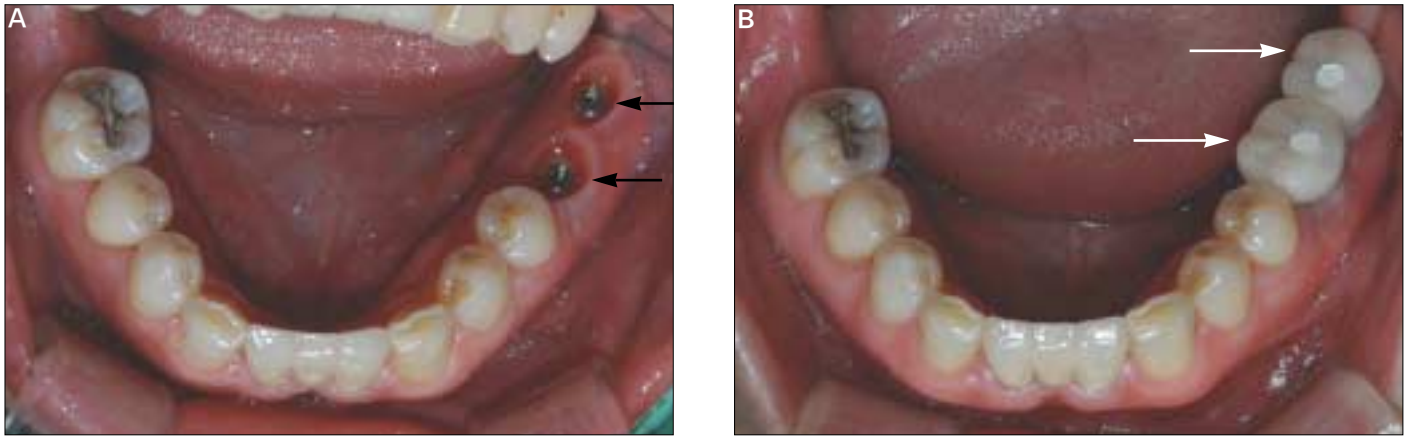

Fig. 1. (A) Two implants (SinusQuick™ EB, Neobiotech Co., Seoul, Korea) were inserted at #36 and #37 area (black arrows). (B) Provsional restoration (white arrows) was delivered 14 days after implant placement.

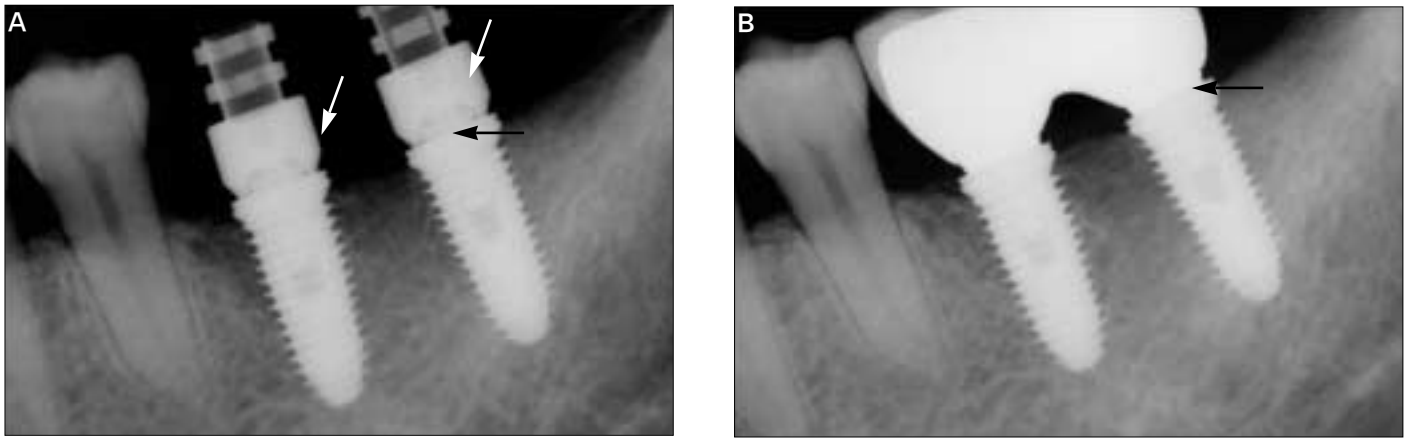

Fig. 2. Periapical radiograph was taken at the time of (A) immediate loading and (B) 3-months after continued loading. The platform (black arrows) was a reference point to measure marginal bone loss. Provsional resin restoration was made by polymethylmethacrylate that is radiolucent. Therefore, only temporary cylinders are seen (white arrows).

detectable implant mobility, 2) absence of pain or any subjective sensation, 3) absence of recurrent peri-implant infection, and 4) absence of continuous radiolucency around the implant.[10]

RESULTS

Total 15 implants were placed and were loaded immediately. Table I shows the details of distribution of inserted implants. Marked variability was noted in the implant sizes selected for placement, although implants 11.5 mm length and 5.0 mm diameter were most commonly used. The mean follow-up period was 4.8 months (range, 2 to 6 months). Mean marginal bone loss from implant surgery to immediate loading, 3-months follow-up and last follow-up was found to be 0.03 mm, 0.16 mm and 0.29 mm respectively (Table II). No implant failed up to 6 months after insertion, resulting in a 100% survival rate.

Table I. Distribution of implant dimensions

Diameter (mm)	Length (mm)					No. of Implants
	7	8.5	10.0	11.5	13	
3.5	0	0	1	3	0	4
4.0	1	0	0	0	0	1
5.0	3	1	0	2	4	10
No. of implants	4	1	1	5	4	15

Table II. Marginal bone loss at early loading, 3-months follow-up and last follow-up

Time	No. of implants	Marginal bone loss (Mean ± SD) (mm)
Early loading	15	0.03 ± 0.07
3-months follow-up	15	0.16 ± 0.17
Last follow-up	15	0.29 ± 0.19

DISCUSSION

All the inserted implants showed successful integration and stable peri-implant condition up to six months. Primary stability was reported to be the most important determining factor on immediate implant loading.[6] Micromovements of more than 100 μm were sufficient to jeopardize healing with direct bone-to-implant contact.[6] Szmukler-Moncler *et al.* indicated that micromotions at the bone-implant interface beyond 150 μm resulted in fibrous encapsulation instead of osseointegration.[11] If the primary implant stability could not be achieved or was questionable, it was strongly recommended to follow a conventional treatment protocol.[6] Most agreed that an insertion torque of at least 32 Nm and a resonance frequency analysis of at least 60 ISQ was required to achieve a high level of stability.[12] In this study, mean ISQ of 15 early loaded implants was 64.9 ± 4.9.

Generally, clinicians agreed that the quality of bone was significant for success in immediate loading. The initial stability of the implant reduces in the first 3-6 weeks after placement due to remodeling and an increased ratio of woven to lamellar bone.[12] Barewal *et al.* indicated that implants placed in areas of high bone quality are relatively stable over the early healing periods.[13] However, we reported that both maxillary and mandibular arches showed no failure of implants although the sample size was too small to analyze the data. Horiuchi *et al.* also reported about no difference in the success rate between arches in immediate loading.[14] Further studies are required about the relationship between bone quality and the success rate of immediate loading.

It has been established that there are no absolute contraindications to implant placement although a number of conditions exist, which are associated with an increased risk of failure.[12] Tobacco was reported to be only a risk factor for the implant failure.[3] However, the results of this investigation showed there was no implant failure in the participating patients who were smokers. Long-term studies about relevance of smoking to early loading are necessary.

There were some limitations associated with this study. The number of investigated implants was insufficient to analyze the data using proper statistics. The follow-up period was also short. Therefore, we could not assess the long-term outcome of immediate loading. Further controlled clinical studies are needed to evaluate the long-term success of early loaded implants.

CONCLUSION

Within the limitation of this clinical study the preliminary results indicate that immediate loading of the implants in partial edentulism, based on SinusQuick™ EB implant system, may be successful for short period up to six months. Well-controlled long term clinical studies with large sample size are necessary.

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REFERENCES

1. Adell R, Lekholm U, Rockler B, Brånemark P. A 15-year study of osseointegrated implants in the treatment of the edentulous jaw. Int J Oral Surg 1981;10:387-416.
2. Chiapasco M, Gatti C, Rossi E, Haefliger W, Markwalder TH. Implant-retained mandibular overdentures with immediate loading. A retrospective multicenter study on 226 consecutive cases. Clin Oral Implants Res 1997;8:48-57.
3. Susarla SM, Chuang SK, DodsonTB. Delayed versus immediate loading of implants: survival analysis and risk factors for dental

implant failure. J Oral Maxillofac Surg 2008;66:251-5.
4. Babbush CA, Kent JN, Misiek DJ. Titanium plasma-sprayed (TPS) screw implants for the reconstruction of the edentulous mandible. J Oral Maxillofac Surg 1986;44:274-82.
5. Henry P, Rosenberg I. Single-stage surgery for rehabilitation of the edentulous mandible: preliminary results. Pract Periodontics Aesthet Dent 1994;6:15-22.
6. Calandriello R, Tomatis M. Immediate Occlusal Loading of Single Lower Molars Using Brånemark System(R) Wide Platform TiUnite Implants: A 5-Year Follow-Up Report of a Prospective Clinical Multicenter Study. Clin Implant Dent Relat Res 2009. [Epub ahead of print].
7. MalóP, Rangert B, Dvärsäter L. Immediate function of Brånemark implants in the esthetic zone: a retrospective clinical study with 6 months to 4 years of follow-up. Clin Implant Dent Relat Res 2000;2:138-46.
8. Attard NJ, Zarb GA. Immediate and early implant loading protocols: a literature review of clinical studies. J Prosthet Dent 2005;94:242-58.
9. Misch CE, Wang HL, Misch CM, Sharawy M, Lemons J, Judy KW. Rationale for the application of immediated load in implant dentistry: Part I. Implant Dent 2004;13:207-17.
10. Hobo S, Ichida E, Garcia LT. Osseointegration and occlusal rehabilitation. Tokyo: Quintessence Publishing Co., Ltd.; 1989.
11. Szmukler-Moncler S, Salama H, Reingewirtz Y, Dubruille JH. Timing of loading and effect of micromotion on bone-dental implant interface: review of experimental literature. J Biomed Mater Res 1998;43:192-203.
12. Henry PJ, Liddelow GJ. Immediate loading of dental implants. Aust Dent J 2008;53:S69-81.
13. Barewal RM, Oates TW, Meredith N, Cochran DL. Resonance frequency measurement of implant stability *in vivo* on implants with a sandblasted and acid-etched surface. Int J Oral Maxillofac Implants 2003;18:641-51.
14. Horiuchi K, Uchida H, Yamamoto K, Sugimura M. Immediate loading of Brånemark system implants following placement in edentulous patients: a clinical report. Int J Oral Maxillofac Implants 2000;15:824-30.

The level of buccal gingival margin around single and two adjacent implant restorations: a preliminary result

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STATEMENT OF PROBLEM. Little information is available about the buccal gingival level of multiple implant restorations. **PURPOSE.** This study was aimed to evaluate the relationship between width and height of buccal soft tissue around single and 2 adjacent implant restorations. **MATERIAL AND METHODS.** Four implant restoration groups (first and second molars, single second molars, posterior single restorations between teeth, and anterior single restorations between teeth) were randomly chosen from one dental institute. Each group comprised of 6 patients. After 6 months of function, silicone impressions were taken and stone models were fabricated for each restoration group. The stone models were cut in bucco-lingual direction at the most apical point of buccal gingival margin. The height and width of buccal supra-implant soft tissue were measured. One way ANOVA and Tukey HSD post hoc tests were performed to analyze the data obtained ($P < .05$). **RESULTS.** The most unfavorable width-height ratio was noted for the group, which was comprised of the second molar in the multiple adjacent (first and second molar) implant-supported restorations. The group also resulted in the shorter height of buccal supra-implant mucosa rather than that of anterior single implant restorations between natural teeth. **CONCLUSION.** To achieve a favorable level of buccal gingival margin, greater thickness of buccal supra-implant mucosa is required for the implant restorations without a neighboring natural tooth compared to the implant restorations next to a natural tooth. **KEY WORDS.** Implant esthetics, Buccal gingival level, Single implant, Multiple adjacent implants, Width-height ratio **[J Adv Prosthodont 2009;1:140-4]**

INTRODUCTION

Soft tissue esthetics is regarded as a prerequisite for successful implant restorations.[1,2] To achieve an esthetic outcome, the level of peri-implant soft tissue should be in harmony with that of adjacent natural teeth as well as the contralateral tooth. Chang and colleagues compared the esthetic outcomes of single implant restoration to those of contralateral natural tooth, and reported that the crowns supported by implants were on the average 1 mm longer than the clinical crown of contralateral natural teeth.[3] Fürhauser *et al.* assessed 7 esthetic variables, which were mesial papilla, distal papilla, soft-tissue level, soft-tissue contour, alveolar process deficiency, soft-tissue color and texture, to evaluate the esthetic outcome of peri-implant soft tissue. The study showed that the soft-tissue level and soft-tissue color variables resulted in poorer esthetic outcome compared to other variables.[4]

The level of facial gingiva is influenced by several factors such as thickness of labial bony wall, the position of implant shoulder, the orientation of implant, the diameter of implants and the gingival biotypes.[1,5-7] The stability of peri-implant soft tissue is also a determinant factor for the level of gingival margin. Grunder investigated the stability of the mucosal topography around single tooth implants. The 1-year results revealed that the soft tissue shrinkage on the buccal side of the implant crown was 0.6 mm on average.[8] Small and Tarnow reported that approximately 1 mm of recession occurred on the midbuccal gingiva at 1 year following abutment connection surgery. The major changes took place within the first 3 months.[9] Other studies reported that apical displacement of the facial soft tissue margin mainly took place during the first 6 months following insertion of final restorations, with relatively little change afterwards.[10,11]

Recently, Nozawa and colleagues noticed the width of buccal supra-implant mucosa as a determining factor of the level of buccal marginal gingiva.[12] They measured the height and width of the buccal supra-implant mucosa in 14 single implant restorations, and reported that the average biologic height to width ratio was 1 : 1.58. When the width of buccal supra-implant mucosa was less than 1.5 times of the height of mucosa, they recommended horizontal tissue graft to prevent the decrease of mucosal height.

Until now, most of studies about peri-implant mucosal level investigated the level of single implant restorations. No attempt was made to measure the height of buccal supra-implant soft tissue around multiple adjacent implant restorations.

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The purpose of current study was to evaluate the width and height of buccal supra-implant soft tissue around single and multiple adjacent implant restorations. The null hypothesis tested was that the number of implants (single or multiple adjacent), as well as the location of the implant (anterior or posterior segment, implant between teeth or the terminal implant) does not influence the width, height and width to height ratio of buccal supra-implant soft tissue.

MATERIAL AND METHODS

Four groups of implant restorations, each group comprised of 6 patients, were randomly chosen from the patients who were treated in Gangnam Severance Dental Hospital, Department of Prosthodontics, from Mar 2005 to Dec 2007. Patients who have any third molar were excluded from this study. All patients were fully understood the treatment procedures and an informed consent was made for each patient.

The 4 groups of patients were as follows.

- Group 1: Two adjacent implant restorations, which restored maxillary or mandibular first and second molars. Group 1 was divided into 2 subgroups.
- Group 1a: tooth side implant restorations, which replaced the first molars
- Group 1b: the terminal implant restorations, which replaced the second molars
- Group 2: single implant restorations, which replaced the terminal teeth (second molars)
- Group 3: posterior single implant restorations, which were placed between natural teeth.
- Group 4: maxillary anterior single implant restorations, which were placed between natural teeth.

Each group was comprised of six implant restorations, therefore, a total of 30 implants were included for the current study. The implant used in current study was Inplant™ (Warantec, Seoul, Korea), which had internal friction fit implant/abutment joint. The diameter of implant platform was 4.3 mm whereas that of implant/abutment junction was 3 mm. The diameter of abutment/restoration junction was 4.5 mm, which was 1.5 mm greater than that of implant/abutment junction. Therefore, the transmucosal abutment had concave profile, which allowed thicker peri-implant soft tissue rather than a divergent abutment (Fig. 1). Two piece ready-made abutments (Top abutment, Warantec, Seoul, Korea) with 2 mm gingival collar height were utilized for every final restoration. To measure the width and height of buccal supra-implant soft tissue, the following procedures were performed. The impression of an implant restoration and surrounding soft tissue was taken at 6-month follow up appointment with medium/low viscosity silicone impression material (Aquasil, Dentsply International, York, PA, USA). The impression was removed from the patient's mouth after its complete set. The implant restoration and its corresponding abutment were retrieved from the patient's mouth and connected to a fixture replica. In case the implant restoration was cement-retained, small amount of temporary cement (Temp-Bond NE, Kerr Dental, Orange, CA, USA) was used to secure the position of the restoration from its abutment. The restoration/abutment/replica assembly was inserted into the impression. Each restoration/abutment/replica assembly was very stable in its corresponding impression material. Type IV dental stone (Crystal Rock, Maruishi Plaster Co. Ltd, Japan) was mixed following manufacturer's instruction and poured into the impression body. After the stone set, the impression was removed from

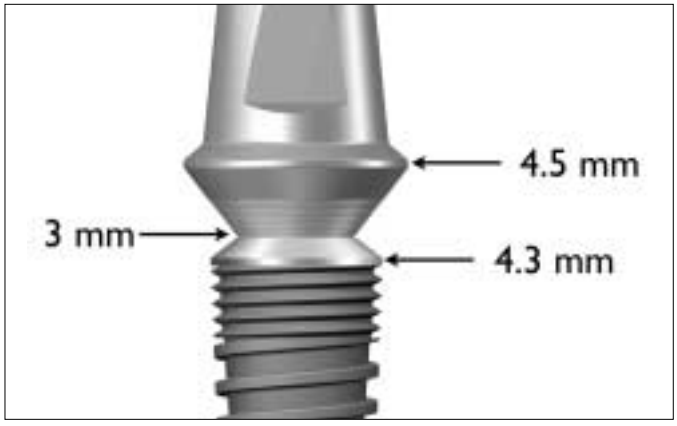

Fig. 1. Close view of implant/abutment junction. The diameter of implant/abutment junction is 1.3 mm narrower than that of implant. The diameter of abutment/restoration junction is 4.5 mm, which is 1.5 mm greater than that of implant/abutment junction resulting transmucosal abutment with inherent concave profile.

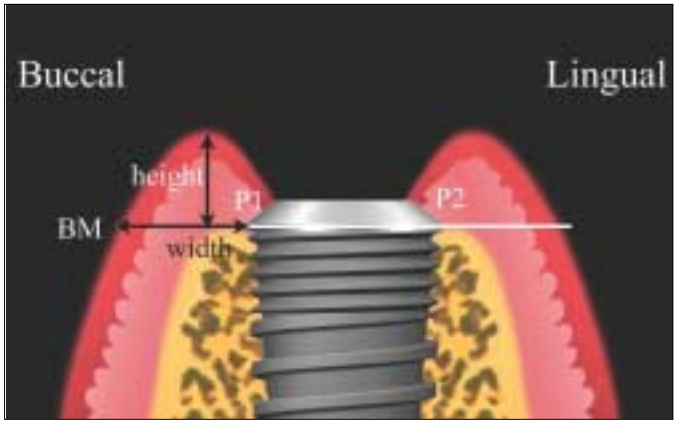

Fig. 2. Schematic drawing of cut area. The first line, which passes buccal (P1) and lingual platform (P2) of implant replica, was drawn. BM is the point, where the line passes the outer contour of buccal soft tissue. The width of buccal-supra implant soft tissue is the distance between BM and P1. A second line, which is perpendicular to the first line and passes the buccal gingival margin (BGM), was drawn and the distance from BGM to the first line is the height of buccal supra-implant soft tissue.

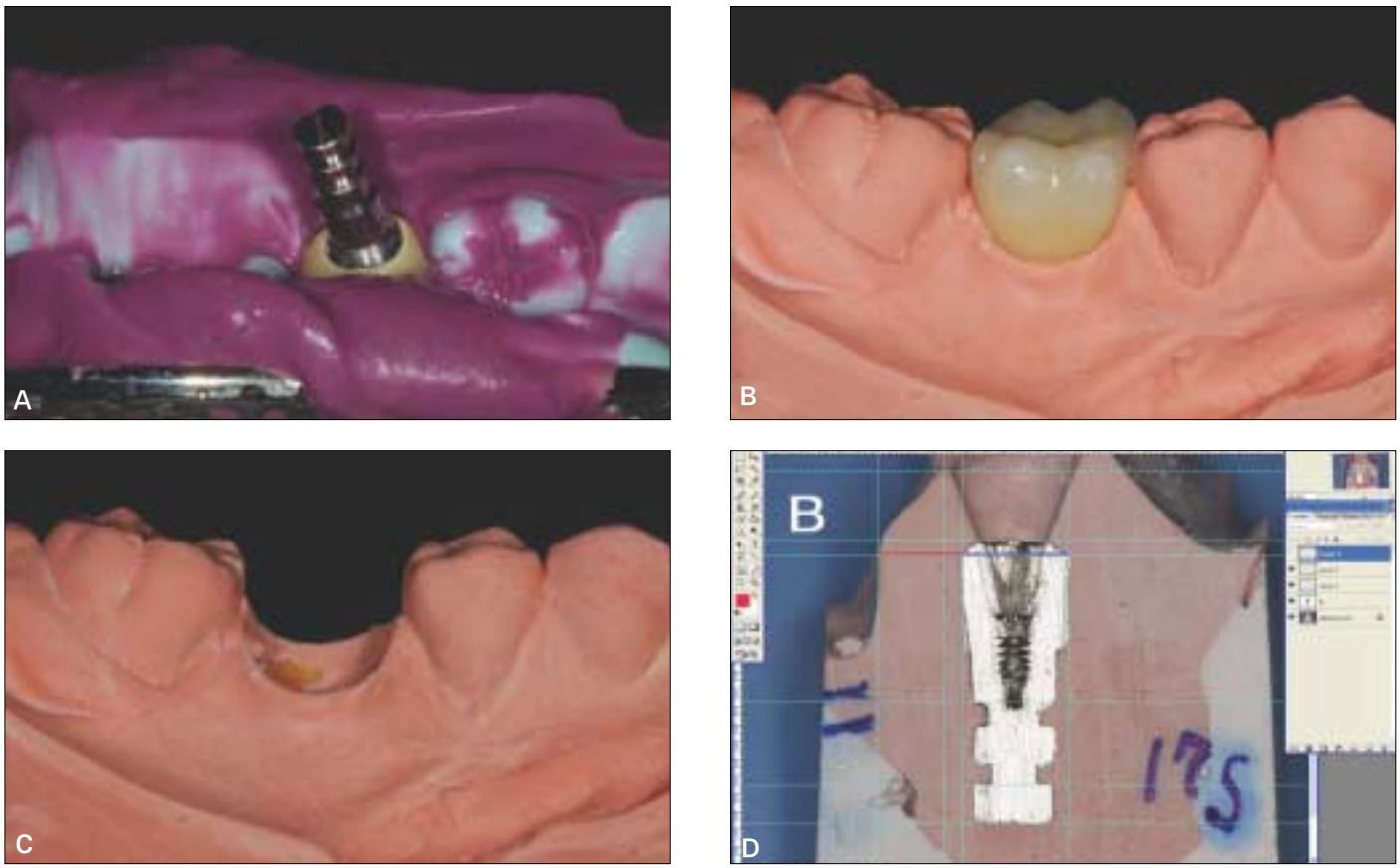

Fig. 3. (A) Implant restoration / abutment / replica assembly was inserted into PVS impression. (B) A stone model was fabricated with Type IV dental stone. (C) implant restoration and abutment were retrieved from the stone model. (D) Measurements of height and width of buccal supra-implant soft tissue by use of a digital image software.

the model. The implant restoration and the abutment were retrieved from the model. With use of a dental model trimmer and #1200 carbon abrasive papers, each model was cut in bucco-lingual direction at the most apical point of buccal marginal gingiva. Digital photograph of the sections were taken with a DSLR camera (D40, Nikon, Japan). Using a digital imaging software (Adobe Photoshop, Adobe, CA), a horizontal line which passes from the buccal to lingual platform of fixture replica and another, perpendicular to horizontal line, were drawn (Fig. 2). The width and height of buccal supra-implant soft tissue were measured for 30 implant sites. The overall procedures were described from Fig 3a to Fig 3d.

Statistical Analysis

The means and standard deviations of the width, the height, and the width-height ratio were statistically evaluated by statistics software (SPSS 14.0, SPSS, Chicago, Il, USA) One-way Analysis of Variance was utilized to test the differences in means of width, height, and width / height ratio ($P < .05$). Tukey HSD post hoc test was performed when ANOVA test indicated statistical significance.

RESULTS

Table I represents means and standard deviations of the width, height and width / height of each group.

There was no significant difference of width among the groups evaluated. The average height of buccal supra-implant soft tissue in Group 4 was greater than that of Group 1b. The width-height ratio of Group 1b was significantly greater than those of Groups 1a, 3, and 4. In addition, Group 1b showed a strong tendency of greater width-height ratio rather than Group 2 ($P = 0.051$).

Table I. Means and standard deviations of evaluated parameters around buccal supra-implant soft tissue

	Width (mm)	Height (mm)	Width / Height
Group 1a	2.85 ± 0.73^{a}	$3.02 \pm 0.96^{a,b}$	$0.98 \pm 0.22^{a,b}$
Group 1b	3.89 ± 1.36^{a}	2.70 ± 0.73^{a}	1.42 ± 0.26^{c}
Group 2	2.99 ± 0.85^{a}	$2.81 \pm 0.78^{a,b}$	$1.09 \pm 0.24^{b,c}$
Group 3	3.51 ± 1.05^{a}	$3.46 \pm 0.96^{a,b}$	$1.01 \pm 0.13^{a,b}$
Group 4	3.17 ± 0.93^{a}	4.27 ± 1.04^{b}	0.74 ± 0.10^{a}
P value	$P = 0.406$	$P = 0.037$	$P = 0.000$

Within the same column, means with the same superscript letters are not statistically different.

DISCUSSION

Esthetic outcomes of implant restorations are influenced by the peri-implant soft tissue as well as the restoration itself.[4,13-15] The esthetics of peri-implant soft tissue depends on the level of facial gingiva and the height of interproximal papilla.[1] The height of interproximal papilla around implant supported restorations were well documented.[13,16] The height of inter-implant papilla is on average 1.5 mm shorter than that of interproximal papilla between implant and natural tooth.[17] Until now, less information is available about the characteristics of the facial gingival level. Most of previous studies investigated the change of gingival level around single implant restorations.[9-11] No study was performed to measure the level of facial gingiva around multiple adjacent implant restorations.

In natural teeth, the direction of tooth movement and the bucco-lingual thickness of the gingiva are known as the influencing factors determining the position of facial gingival margin.[18] Wennstrom reported that there is a ratio of about 1 : 1.5 between its thickness at the most coronal fiber attachment to the root and its height. As like in natural teeth, the position of facial gingival margin of an implant restoration is also influenced by the thickness of buccal supra-implant mucosa. Compared to natural teeth, the average biologic width-height ratio was 1.58 : 1.12 However, the results were measured from single implant restorations with external flat top implant/abutment joints. Rompen and co-workers suggested that abutments with divergent transmucosal profile could create centrifugal pressure at the internal side of the soft tissues and the pressure could result in tendency for recession.[19] They evaluated the vertical stability of soft tissue at buccal aspect of implants to which experimental concave, inwardly narrowed transmucosal abutments were connected and concluded that the use of abutments with concave transmucosal profile seems to allow for better and more predictable soft tissue stability than those with divergent profiles.

In the current study, the width, the height and the width-height ratio of buccal supra-implant soft tissue were evaluated not only in single implant restorations but also in multiple adjacent implant restorations. The groups were divided by the number and the position of implant restorations. The results of this study showed that there was no significant difference in the width of buccal supra-implant mucosa between groups whereas the height and the width to height ratio resulted in significant difference between the groups. The width to height ratio of Group 1b was significantly greater than those of Groups 1a, 3 and 4. In addition, Group 1b showed a strong tendency of greater width-height ratio rather than Group 2 ($P = 0.051$). Compared to other groups, Group 1b comprised of the implant restorations without adjacent natural tooth. Therefore, it is assumed that the width-height ratio of buccal supra-implant soft tissue was influenced by the presence of a natural tooth next to the implant restoration. Based on this result, it is inferred that favorable width-height ratio could result in every single implant restorations whereas unfavorable width-height ratio might be expected in the multiple implant restorations without neighboring natural tooth.

Nozawa and colleagues reported that every single implant restorations showed greater width of buccal supra-implant soft tissue rather than the height.[12] However, in the current study, 15 out of 30 implant restorations showed greater height than its corresponding width. Furthermore, the height was always greater than the width of implant restorations in the maxillary anterior segment. These differences could be due to the configuration of transmucosal abutment. Nozawa measured the width to height ratio around transmucosal abutments with divergent profile whereas concave transmucosal abutments were used in the current study.

Impressions were taken 6 months after the delivery of final restorations. It was reported that the majority of the recession occurred within the first 3 months after implant placement surgery for 1 stage implants or abutment connection for 2 stage implants.[9] Therefore, 6 months of function was chosen for minimum functional period for the current study.

In this study, soft tissue parameters around single and 2 adjacent implant restorations were evaluated. As three or more adjacent implant restorations were not included in this study, it could not be performed to assess the width/height ratio of buccal supra-implant soft tissue around an implant restoration between the implants. Each group comprised of only 6 implant restorations. Larger sample size and inclusion of more than 3 adjacent implant restorations would be beneficial in gathering more information about the characteristics of peri-implant soft tissue.

CONCLUSION

1. Unfavorable width-height ratio was noted for the second molar in the first and second molar implant-supported restorations.
2. To achieve an esthetic outcome, greater thickness of buccal supra-implant mucosa is required for the implant-supported restorations without any neighboring natural tooth rather than implant-supported restorations next to a natural tooth.

REFERENCES

1. Buser D, Martin W, Belser UC. Optimizing esthetics for implant restorations in the anterior maxilla: anatomic and surgical considerations. Int J Oral Maxillofac Implants 2004;19:43-61.
2. Jemt T. Restoring the gingival contour by means of provisional resin crowns after single-implant treatment. Int J Periodontics Restorative Dent 1999;19:20-9.
3. Chang M, Wennström JL, Odman P, Andersson B. Implant supported single-tooth replacements compared to contralateral

natural teeth. Crown and soft tissue dimensions. Clin Oral Implants Res 1999;10:185-94.
4. Fürhauser R, Florescu D, Benesch T, Haas R, Mailath G, Watzek G. Evaluation of soft tissue around single-tooth implant crowns: the pink esthetic score. Clin Oral Implants Res 2005;16:639-44.
5. Kois JC. Predictable single tooth peri-implant esthetics: five diagnostic keys. Compend Contin Educ Dent 2001;22:199-206; quiz 208.
6. Evans CD, Chen ST. Esthetic outcomes of immediate implant placements. Clin Oral Implants Res 2008;19:73-80.
7. Saadoun AP, Touati B. Soft tissue recession around implants: is it still unavoidable?-Part I. Pract Proced Aesthet Dent 2007;19:55-62; quiz 64.
8. Grunder U. Stability of the mucosal topography around single-tooth implants and adjacent teeth: 1-year results. Int J Periodontics Restorative Dent 2000;20:11-7.
9. Small PN, Tarnow DP. Gingival recession around implants: a 1-year longitudinal prospective study. Int J Oral Maxillofac Implants 2000;15:527-32.
10. Bengazi F, Wennström JL, Lekholm U. Recession of the soft tissue margin at oral implants. A 2-year longitudinal prospective study. Clin Oral Implants Res 1996;7:303-10.
11. Oates TW, West J, Jones J, Kaiser D, Cochran DL. Long-term changes in soft tissue height on the facial surface of dental implants. Implant Dent 2002;11:272-9.
12. Nozawa T, Enomoto H, Tsurumaki S, Ito K. Biologic height-width ratio of the buccal supra-implant mucosa. Eur J Esthet Dent 2006;1:208-14.
13. Choquet V, Hermans M, Adriaenssens P, Daelemans P, Tarnow DP, Malevez C. Clinical and radiographic evaluation of the papilla level adjacent to single-tooth dental implants. A retrospective study in the maxillary anterior region. J Periodontol 2001;72:1364-71.
14. Cornelini R, Cangini F, Covani U, Wilson TG Jr. Immediate restoration of implants placed into fresh extraction sockets for single-tooth replacement: a prospective clinical study. Int J Periodontics Restorative Dent 2005;25:439-47.
15. Juodzbalys G, Wang HL. Soft and hard tissue assessment of immediate implant placement: a case series. Clin Oral Implants Res 2007;18:237-43.
16. Tarnow D, Elian N, Fletcher P, Froum S, Magner A, Cho SC, Salama M, Salama H, Garber DA. Vertical distance from the crest of bone to the height of the interproximal papilla between adjacent implants. J Periodontol 2003;74:1785-8.
17. Tarnow DP, Cho SC, Wallace SS. The effect of inter-implant distance on the height of inter-implant bone crest. J Periodontol 2000;71:546-9.
18. Wennström JL. Mucogingival considerations in orthodontic treatment. Semin Orthod 1996;2:46-54.
19. Rompen E, Raepsaet N, Domken O, Touati B, Van Dooren E. Soft tissue stability at the facial aspect of gingivally converging abutments in the esthetic zone: a pilot clinical study. J Prosthet Dent 2007;97:S119-25.