J Mater Sci: Mater Med (2007) 18:2025–2031
DOI 10.1007/s10856-007-3159-6

Load bearing capacity of bone anchored fiber-reinforced composite device

Ahmed Mansour Ballo · Lippo V. Lassila · Pekka K. Vallittu · Timo O. Närhi

Received: 13 May 2006 / Accepted: 21 August 2006 / Published online: 9 June 2007

Abstract The purpose of this study was to evaluate the push-out load-bearing capacity of threaded fiber-reinforced composite (FRC) devices for use as bone-anchored devices. The purpose was also to evaluate the possibility to use bioactive glass (BAG) granules on the experimental FRC devices in terms the mechanical behavior.

Three experimental FRC devices (n = 15) were fabricated for the study: (a) threaded device with smooth surface; (b) threaded device with BAG granules (S53P4, Vivoxid Ltd, Turku, Finland) and supplementary retention grooves, and (c) unthreaded device with BAG granules. Threaded titanium devices were used as controls. The FRC devices were prepared from a light-polymerized dimethacrylate resin reinforced with preimpregnated unidirectional and bidirectional E-glass fibers (EverStick, StickTech Ltd, Turku, Finland). Experimental and control devices were embedded into dental plaster to simulate bone before the mechanical push-out test was carried out. ANOVA and Weibull analysis were used for the statistical evaluation. Threaded FRC devices had significantly higher push-out strength than the threaded titanium device ($p < .001$). The push-out forces exceeding 2,500 N were measured for threaded FRC devices with supplementary grooves and BAG coating. No thread failures were observed in any FRC devices. The unthreaded FRC devices with BAG lost 70% of glass particles during the test, while no BAG particles were lost from threaded FRC devices. It can be concluded that threaded FRC devices can withstand high push-out forces in the dental plaster without a risk of thread failure under physiological load.

A. M. Ballo (✉) · L. V. Lassila · P. K. Vallittu · T. O. Närhi
Department of Prosthetic Dentistry and Biomaterials Science, Institute of Dentistry, University of Turku, LemminkÃñisenkatu 2, Turku 20520, Finland
e-mail: ahmbal@utu.fi

Introduction

Dental implants have been used for teeth replacement successfully over the last three decades [1–3]. Many endosseous dental implant systems with various designs are currently on the market [4]. Titanium is the traditional material in all commercially available dental implants. Surface topography and implant configuration have a great influence on the initial stability of an implant. Good primary stability is mandatory for successful osseointegration [5]. Inadequate initial stability accounts for 32% of implant failures [6]. Poor initial stability results in the formation of a so-called ''peri-implant membrane'', which corresponds to non-mineralized connective tissue layers on bone-to-implant interfaces [7].

Commonly used implants are screw-shaped, designed to maximize the potential area for osseointegration and provide good initial stability. Higher failure rates after loading have been reported for implants with relatively smooth surfaces [8, 9], in comparison with rough-surfaced implants [10, 11]. Many methods to modify implant surface topography have been introduced and evaluated in both in vitro [12, 13] and in vivo studies [14–16]. Incorporation of different design features, such as pattern of implant threads and roughness configuration can optimize initial stability and maximize the crestal cortical bone preservation [17].

The surface's roughness can be created directly on the implant surface by sandblasting or acid etching, or by

utilizing various coating techniques (plasma-spray, magnetron sputtering or laser treatment). Apart from the roughness, implant coatings may accelerate the osseointegration process by facilitating chemical (bonding osteogenesis) bonding between implant and bone [18–22].

Until now, none of the commercially available implants is able to attach to bone tissue with a periodontal ligament-like structure that might reduce the impact of the occlusal loads transmitted to the bone [23]. In poor bone conditions, the mismatch of stiffness between bone and metallic implant may lead to implant failure [24]. This occurs when the tensile or compressive load exceeds the physiological limit of bone tolerance and causes microfracture at the bone-to-implant interface, or initiates bone resorption [25].

Fiber-reinforced composites (FRC) are strong materials with a lower elastic modulus than metals [26]. Adding E-glass fiber to the polymer significantly improves the material's mechanical properties. Fiber reinforcement can even be utilized in devices that require high strength [27]. This improvement in mechanical properties is well documented [28, 29]. There has been growing interest in using FRC for applications involving some degree of structural performance in load-bearing applications such as dental crowns, fixed partial dentures, and implant-supported prostheses [30–32].

Polymer-based bone cements are generally accepted as reliable biomaterials in orthopedic applications [33–35]. The good mechanical properties achieved with FRC have raised the idea of using fiber-reinforced composites in bone-anchored devices. Bioactive glasses (BAG) are ceramic materials, which chemically adhere directly to the surrounding bone. BAG can also be used in implant coatings to improve healing and properties of bone around the coated implants [36].

The objective of this study was to evaluate the load-bearing capacity of three threaded experimental FRC devices using the push-out test in a simulated bone structure. As control material, conventional titanium with experimental thread design and commercially available titanium implants were used.

Materials and methods

The materials used for the fabrication of the specimens for this study are listed in Table 1. The specimens were combined with five fiber-reinforcement bundles each consisting of 4,000 continuous unidirectional E-glass fibers (diameter ca. 15 μm). The fiber bundle was impregnated manually in light-polymerizable bisGMA-TEGDMA resin. The composition of fibers was: 55% SiO_2, 15% Al_2O_3, 22% CaO, 6% B_2O_3 and 0.5% MgO, > 1.0% Fe + Na + K. The group with fiber-reinforced thread was manufactured by adding bidirectional weave around the threads. Porous PMMA-weave reinforcements (StickNet) was pre-impregnated for 24 h in light polymerizable resin to dissolve PMMA, and to form a semi-IPN polymer network. The fiber reinforcements were inserted into a mold along the long axis of the specimens (Fig. 1).

Three different kinds of FRC specimens were made (Fig. 2): (I) an unthreaded FRC device with BAG coating, (II) a threaded FRC device and (III) a FRC device FRC reinforced threads was coated with BAG and extra retention was provided by supplementary retention grooves. SEM image of threaded FRC devices are shown in (Figs. 3, 4). Fifteen specimens were prepared for each group. The test FRC specimens were polymerized in a light-curing oven at 60 °C in a vacuum (Visio Beta Vario 3M/ESPE, Seefeld, Germany) for 15 min to eliminate the oxygen inhibition layer on resin. Subsequently, the polymerization was completed in a light curing oven (LicuLite, Dentsply De Trey GmbH, Dreieich, Germany) for 1 h, in which the temperature was increased to 80 °C (Kerr-Have., I, USA). To optimize the degree of monomer conversion (DC%) the

Table 1 Materials used in the study

Composition	Lot no.	Manufacture	Description	Product
BisGMA-[a] TEGDMA[b]	54031672	Stick Tech, Turku, Finland	Light curing resin	Stick Resin
E-glass,[c] PMMA	2050523-W 0053	Stick Tech, Turku, Finland	Polymer preimpregnated bidirectional weave fiber	Stick net fiber
E-glass	11372313	Ahlstrom, Karhula, Finland	Unidirectional fiber	E-glass fiber
BAG[d] SiO_2 53%, Na_2O 23%, CaO 20% and P_2O_5 4%	ABM S53-8-01	Vivoxid Ltd, Turku, Finland	Average particle size, 90–315 μm	BAG granule

[a] Bis-GMA, bisphenol A-glycidyl dimethacrylate

[b] TEGDMA, triethylenglycoldimethacrylate

[c] E-glass, electrical glass

[d] BAG, bioactive glass

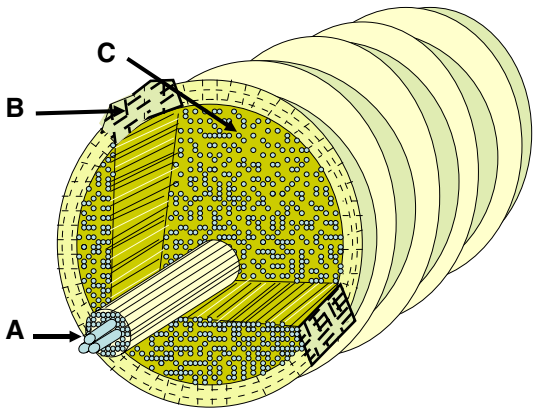

Fig. 1 Schematic and simplified picture of structural design of the threaded fiber-reinforced composite device (**A**) Unidirectional (E-glass) fibers, (**B**) Bidirectional weaves fibers, (**C**) Light-cured resin matrix

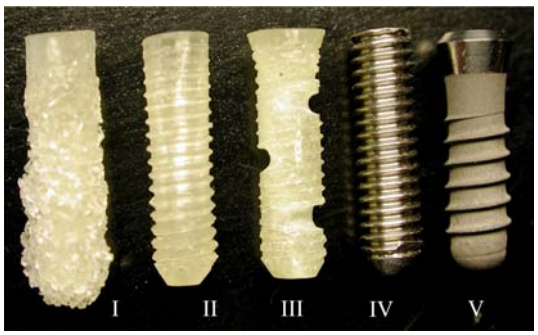

Fig. 2 Picture of the experimental specimens (I) Unthreaded FRC device with bioactive glass coating, (II) Threaded FRC device, (III) Threaded BAG-coated FRC device with supplementary grooves (IV) Custom-made titanium screw-shaped device (V) Commercial Straumann dental implant

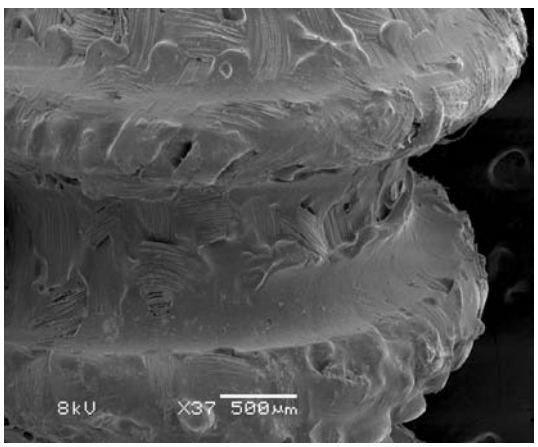

Fig. 3 SEM micrograph illustrating the fiber-reinforced thread structure of the fiber-reinforced composite device

specimens were post-cured in an oven for 24 h at 120 °C, which is close to the glass transition temperature (T_g) of pBisGMA-pTEGDMA-copolymer. The high DC%

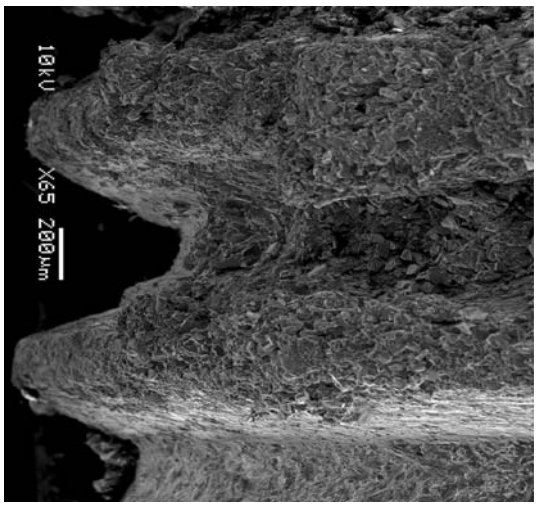

Fig. 4 SEM micrograph illustrates the surface of threads with fiber reinforcement covered with BAG coating. (Original magnification ×65)

improves the material's biocompatibility. Commercially available BAG (S53P4, Vivoxid Ltd. Turku, Finland) granules were used in the preparation of specimens containing BAG. Two different sized BAG granules were used: (a) <45 μm and (b) 90–315 μm. The non-threaded specimens dipped in the resin before disseminating with BAG granules. The granules were then pressed onto the specimen's surface and polymerized with an Optilux 501 (Kerr-Have., I, USA) hand light-curing unit for 40s.

After polymerization, the specimens were wet ground with 500 grit (FEPA) silicon carbide paper from the ending. The length of each FRC specimen was 10 mm, and the diameter 4.0 mm. After polymerization, the size of each specimen was confirmed by measuring its dimensions from three different areas. The size of specimens was based on the assumption that, if used as oral implants, their load-bearing capacity should exceed average maximum occlusal forces within the physiological strain limit of bone [37].

The specimens were conditioned in air at room temperature for 2 days before being used for mechanical testing. Custom-made titanium screw-shaped specimens of the same size were made by manual milling and used as positive controls.

Commercially available Straumann implants with SLA surface (Sand-blasted, Large-grit, Acid-etched) were also used as a reference device.

The specimens were embedded into gypsum plaster using the powder–liquid ratio of 100 g/20 ml as recommended by the manufacturer. The gypsum/specimen blocks were fixed in the testing device and loaded by downwards vertical force with a Lloyd material testing machine (model LRX, Lloyd Instruments Ltd., Fareham, England) at a cross-head speed of 1.0 mm/min until failure of the spec-

imen-gypsum interface (Fig. 5). The peak force of failure was recorded with PC computer Software (Nexygen, Lloyd Instruments Ltd.).

Push-out load values for all groups were analyzed with analysis of variance (ANOVA) (SPSS 11.0, SPSS Inc., Chicago, IL, USA), followed by Tukey's post hoc analysis using a significance level of $p < 0.05$.

To evaluate reliability of the devices, Weibull analysis was carried out using Weibull++ software (Reliasoft Corporation, Tucson, AR USA) using the following formula:

$$P_f = 1 - \exp\left\{-\left(\frac{s - s_u}{s_o}\right)^m\right\}$$

where m = Weibull modulus, s_o = characteristic push-out load and s_u = theoretical failure load (= 0).

Results

Results of the push-out test are shown in (Figs. 6, 7). The push-out force of Group IV (experimentally threaded titanium) was 778N (141N), which did not differ statistically ($p > 0.05$) from the push-out force of Group I (unthreaded BAG-coated), 895 (122) N. However, 70% of the BAG coating was lost during the test. The push-out force of Group II (threaded FRC) was 1277 (137) N, which was statistically ($p > 0.05$) similar to the force of Group V (Strauman SLA implant), 1395 (183) N.

The highest push-out force, 2302 (265) N, was recorded for Group III (threaded FRC/BAG specimens with supplementary grooves). In all FRC devices, the screw threads could sustain the push-out load, and no thread failures were observed.

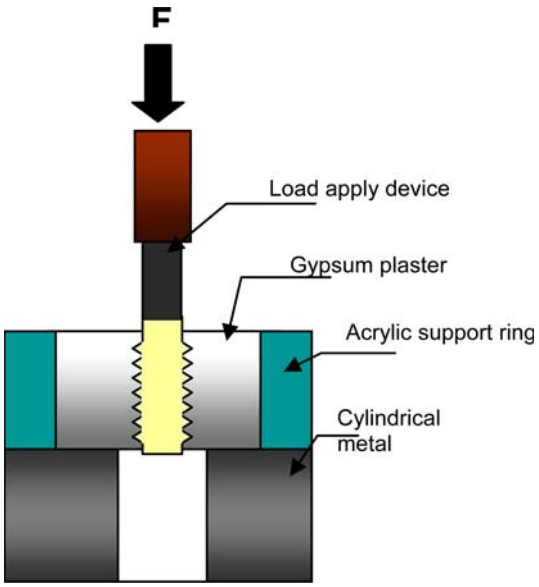

Fig. 5 Push-out test of the specimen

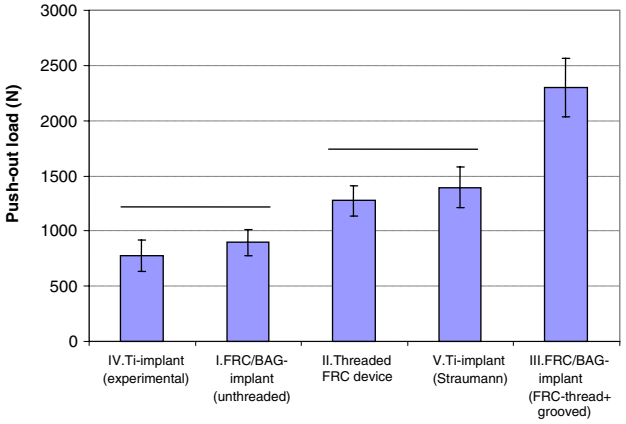

Fig. 6 Results of mean values of push-out test. Horizontal line above bars represents homogeneous subsets (Tukey-post hoc test)

Weibull analysis revealed the lowest Weibull modulus, 5.95, and a characteristic push-out force of 836 N for Group IV (experimental titanium), which shows the lowest reliability within the studied groups. The Weibull modulus of FRC devices varied between 7.75 and 11.94, which suggests improved reliability.

Discussion

The moderately rough titanium surface is generally considered preferable in bone-anchored devices like orthopedic and dental implants [38]. However, poor bone quality still exhibits lower implant survival rates than a dense bone structure [39]. The reason for implant failure often remains unclear, but differences in the elastic properties between the bone and the metallic implant are obviously one of the main reasons for biomechanical failures [24]. Thus, a heavy load on an implant may induce microfractures of the thin bony trabecles. The modulus of elasticity of FRC materials may be tailored to match closely that of cortical bone. This is an important aspect as bone requires physiological stimulus from mechanical stress to prevent bone resorption and to maintain its structure. This study was conducted to determine the load-bearing capacity of bone-anchored FRC devices using the push-out test before their evaluation in a living bone environment.

Various techniques have been used to examine the mechanical strength of medical devices. The choice of these tests depends on the clinically most significant failure mode of implants. The strength of a FRC composite device is limited by the strength of its weakest component. However, the load-bearing capacity of threads in screw-

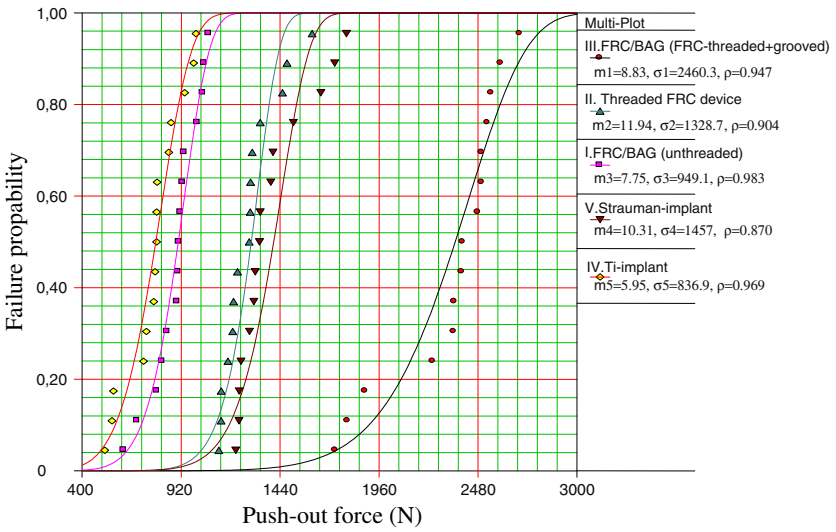

Fig. 7 Weibull plot of the investigated bone-anchoring devices

shaped FRC devices under vertical load has not been known. In this study, we chose to embed the experimental devices into dental plaster instead of bone. A similar method has been used by Mattila et al. [40] to evaluate the effect of surface texture as a retention property of the implant. With this method, it is possible to test the load-bearing capacity of devices in an equivalent environment, without having the problem of standardizing the porous bone structure.

The result of the push-out tests conducted on FRC specimens with FRC threads revealed a load approaching 1,300 N. These values were almost double those achieved with identitical threaded titanium (IV, control) devices. Fracture always occurred in the plaster, and no thread failures were observed. The reason for the higher push-out force of FRC specimens is obviously related to the low modules of elasticity of FRC (40 GPa) compared to titanium (120 GPa) [41]. In the vertical force of the load-bearing test, the FRC reduced the compressive radial stress at the plaster-device interface and around the threads of the FRC specimens. The stress was distributed more evenly throughout the FRC structure and to the surrounding area compared to the titanium device.

The threaded FRC devices aim to achieve fixation through mechanical interlocking with the surrounding bone (Figs. 8, 9), while the bioactive coatings are used to achieve a biologic bond with bone. However, a certain degree of surface roughness is required to obtain mechanical interlocking also for the coated devices. The cylindrical non-threaded FRC devices with BAG coating (Group I) had the most irregular surface with noticeable pits and fissures, while the threaded FRC specimens with BAG coating (Group III) showed a rough surface with very small pits. The threaded FRC devices (Group II) without any surface modification and the control titanium specimens had smooth surfaces. The Straumann dental implants, which were also used as a reference had a modified surface (SLActive surface). The SLA surface has been demonstrated to have better bone bonding properties than the previously used titanium plasma-sprayed (TPS) surfaces [42]. In terms of the push-out force, the threaded FRC devices used in this study performed as well as the Strauman SLA implants.

The study also showed that in all threaded FRC devices, the measured push-out forces exceeded the reported maximum bite force [43], but were lower than the strength of

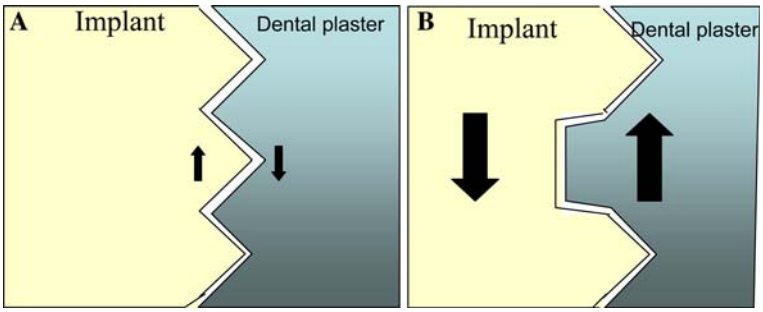

Fig. 8 Schematic and simplified picture of the interface shear strength of FRC specimens cast in plaster: (**A**) threaded FRC specimen, (**B**) BAG-coated threaded FRC specimen with supplementary grooves

Fig. 9 SEM micrograph of fracture surface of polymeric threaded FRC device after push-out test. Fracture occurred through outer layer of FRC threads, suggesting that main reason for failure was low cohesive shear strength of plaster. (Original magnification ×35)

the weakest part of the threaded FRC devices. Some failures occurred in the threads when the push-out force was higher than 2,500 N. Such high forces were registered only with FRC devices additionally coated with BAG, and when extra retention was provided by supplementary retention grooves (Group III). In the unthreaded specimens, failure occurred due to detachment of the BAG granules from the surface of the FRC device. Hence, the interface between the BAG granules and the FRC structure was the weakest link in the unthreaded specimens.

Thus, although a bond between the FRC device and bone may be improved through the use of a bioactive material at the device surface, a significant limitation remains on the interfacial bond between the bioactive particles and the FRC device. In the unthreaded devices, the FRC/BAG interface was assumed to be able to resist compressive stress but not shear stress. However, in the threaded FRC specimens with BAG coating, the BAG granules were partly in compression with respect to the FRC surface, which provided a better environment for the BAG granules.

It may be assumed that the interface between bone and FRC device can tolerate higher loads than those described in this study since the bioactive glass particles will form an apatite surface layer on the implant-to-bone interface, which can lead to biologic bone bonding [22].

Although the mechanical properties of FRC anchoring devices have been found to be comparable to those of titanium implants, further studies are needed to explore the biologic behavior of bone-anchored FRC devices.

Conclusion

Within the limitations of this study, it can be concluded that:

The threads of FRC bone-anchoring devices can withstand static load values comparable to human maximal bite forces without fracture.

The push-out force from dental plaster is higher for threaded FRC devices than for a similar titanium device.

References

1. R. ADELL, U. LEKHOLM, B. POCKLER and P. I. BRÅNEMARK, *Int. J. Oral. Surg*. **6** (1981) 387
2. J. A. EKELUND, L. W. LINDQUIST, G. E. CARLSSON and T. JEMT, *Int. J. Prosthodont*. **16** (2003) 602
3. T. JEMT, U. LEKHOLM and R. ADELL, *Int. J. Oral Maxillofac. Implants* **4** (1980) 211
4. J. E. LEMONS, *Implant Dent.* **7** (1998) 351
5. T. ALBREKTSSON, P. I. BRÅNEMARK, H. A. HANSSON and J. A. LINDSTOM, *Acta Orthop. Scand.* **52** (1981) 155
6. B. FRIBERG, T. JEMT and U. LEKHOLM, *Int. J. Oral Maxillofac. Implants*. **6** (1991)142
7. S. YAMADA, S. SEKIYA, K. YAMANOUCHI, H. KITAMURA, M. OHSHIMA and T. SATO, *Bull. Tokyo Dent. Coll*. **30** (1989)187
8. O. BAHAT, *Int. J. Oral Maxillofac. Implants*. **7** (1992) 459
9. M. NEVINS and B. LANGER, *Int. J. Oral Maxillofac. Implants*. **8** (1993) 428
10. R. J. LAZZARA, S. S. PORTER, T. TESTORI, J. GALANTE and L. ZETTERQVIST, *J. Esthet. Dent*. **10** (1998) 280
11. U. GRUNDER, N. BOITEL, M. IMOBERDORF, K. MEYENBERG, T. MEIER and C. ANDREONI, *Compend Contin. Educ. Dent*. **20** (1999) 628
12. B. CHEHROUDI, T. R. GOULD and D. M. BRUNETTE, *J. Biomed. Mater. Res*. **26** (1992) 493
13. B. CHEHROUDI, D. MCDONNELL and D. M. BRUNETTE, *J. Biomed. Mater. Res*. **34** (1997) 279
14. C. HALLGREN, H. REIMERS, J. GOLD and A. WENNERBERG, *J. Biomed. Mater. Res*. **57** (2001) 485
15. J. LI, H. LIAO, B. FARTASH, L. HERMANSSON and T. JOHNSSON, *Biomaterials*. **18** (1997) 691
16. Y. T. SUL, C. B. JOHANSSON, Y. JEONG, K. ROSER, A. WENNERBERG and T. ALBREKTSSON, *J. Mater. Sci. Mater. Med*. **12** (2001) 1025
17. L. VIDYASAGAR and P. APSE, *Baltic Dent. Maxillofac. J*. **6** (2004) 51
18. J. T. KRAUSER, C. BONER and N. BONER, *Cah Prothese*. **71** (1990) 56
19. Y. L. CHANG, D. LEW, J. B. PARK and J. C. KELLER, *J. Oral Maxillofac. Surg*. **57** (1999) 1096
20. C. K. CHANG, J. S. WU, D. L. MAO and C. X. DING, *J. Biomed. Mater. Res*. **56** (2001) 17
21. J. SCHROOTEN and J. A. HELSEN, *Biomaterials*. **21** (2000) 1461
22. N. MORITZ, S. ROSSI, E. VEDEL, T. TIRRI, H. YLANEN, H. ARO and T. NÄRHI, *J. Mater. Sci. Mater. Med.* **15** (2004) 795
23. C. E. MISCH, Z. QU and M. W. BIDEZ, *J. Oral Maxillofac. Surg*. **57** (1999) 700
24. J. E. LEMONS, *Implant Dent.* **7** (1998) 351
25. J. B. BRUNSKI, *Adv. Dent. Res.* **13** (1999) 99

26. A. M. Ballo, L. Lassila, T. Närhi and P. K. Vallittu, *J. Contemp Dent Pract.* 2007 (In press)
27. P. K. Vallittu and J. Assila, Oral Rehabil. **19** (1992) 225
28. P. K. VALLITTU, *J. Oral Rehabil*. **25** (1998) 100
29. K. CHUNG, T. LIN, and F. WANG, *J. Oral Rehabil*. **25** (1998) 214
30. M. A. FREILICH, J. P. DUNCAN, E. K. ALARCON, and K. A. ECKROTE, *J. Prosthet. Dent*. **88** (2002) 449
31. M. BEHR, M. ROSENTRITT, R. LANG, and G. HANDEL, *Clin. Oral. Implants Res*. **12** (2001) 174
32. J. P. DUNCAN, M. A. FREILICH and C. J. LATVIS, *J. Prosthet. Dent*. **84** (2000) 200
33. S. L. EVANS, C. M. HUNT and S. AHUJA, *J. Mater. Sci-Mater. Med*. **13** (2002) 1143
34. J. PALUSSIERE. et al., *Eur spine J.* **14** (2005) 982
35. G. S. ANDREASSEN, P. R. HOINESS, I. SKRAAMM, O. GRANLUND and L. ENGEBRETSEN, *Arch. Orthop. Trauma Surg*. **124** (2004) 161
36. N. MORITZ, S. ROSSI, E. VEDEL, T. TIRRIT, H. YLANEN, H. ARO and T. NÄRHI, *J. Mater. Sci. Mater. Med*. **15** (2004) 795
37. L. B. LUM and J. F. OSIER, *J. Oral Implantol*. **18** (1992) 349
38. T. ALBREKTSSON, A. WENNERBERG, *Int. J. Prosthodont*. **17** (2004) 536
39. A. PIATTELLI, A. SCARANO, L. FAVERO, G. IEZZI, G. PETRONE, G. A. FAVERO, *J. Periodontol.* **74** (2003) 385
40. R. MATTILA, M. PUSKA, L. LASSIILA and P. K. VALLITTU, *J. Mater. Sci.: Mater. Med.* **41** (2006) 4321
41. H. Yuehuei, ''Mechanical properties of Bone. Mechanical Testing of Bone and the bone-Implant Interface'' (CRC Press; 2000) p.43
42. D. BUSER, T. NYDEGGER, T. OXLAND, D. L. COCHRAN, R. K. SCHENK, H. P. HIRT, D. SNETIVY and L. P. NOLTE, *J. Biomed. Mater. Res*. **45** (1999) 75
43. T. M. Van EIJDEN, *Arch. Oral Biol.* **36** (1991) 535