Published literature summary

Selected scientific literature concerning LASAK products
LASAK Ltd. is a research-oriented medical technology company, established in 1991 in Prague, Czech Republic. The mission of the company is the development, production and sale of medical devices. Currently LASAK is one of the leading Czech companies in the field of dental implantology and bone tissue regeneration.

Since 1992, the research and development of bioactive materials for bone-tissue replacement has been carried out systematically at LASAK in cooperation with universities, research institutes and major clinics. Results of the research have then been used in the development of new products, which successfully entered the market. These products are now widely used in clinical practice, especially in the fields of dental implantology, maxillofacial surgery, orthopedics, neurosurgery, as well as in other fields. An important product line consists of resorbable and nonresorbable ceramic materials based on calcium phosphates (PORESORB®-TCP) and on hydroxyapatite (OssaBase®-HA). Significant progress has been achieved by the development of surface-treated bioactive titanium, which exhibits unique properties enabling faster implant healing. This new biomaterial has been successfully used in the production of dental (IMPLADENT®) and spinal (IMPLASPIN) titanium implants.

The development results and clinical evaluations are widely documented in a number of research and clinical studies published in professional journals, conference proceedings and monographs. This publication presents a collection of selected studies published in the field of dental implantology, bone tissue regeneration and spinal surgery in connection with LASAK products.
**List of selected publications**

### Dental implants

#### Clinical studies


#### Experimental studies


**Modelling of stress and deformation distribution around endosteal implants:** Konvičková Š., Káčovský A., Strnad Z.: Quintessenz, Vol. 8, No. 12, 1999.
Bone regeneration materials

Clinical Studies


Experimental studies


Spinal surgery


Two years of experiences with the biotitanium replacement (Implaspin) used in treatment of degenerative lumbar spine diseases; Mrůzek M., Filip M., Veselský P., Paleček T., Strnad Z.: Chirurgia narządów ruchu i Ortopedia Polska 75 (2), 2010, p. 131–135.


Orthopaedics


Development of Implant Stability During Early Healing of Immediately Loaded Implants

A. Šimůnek, D. Kopecká, T. Brázdá, J. Strnad, L. Čapek, R. Slezák


Purpose: To monitor the development of stability of immediately loaded implants during early healing. Materials and Methods: A total of 90 interforaminally placed implants with an alkali-treated surface were considered. The stability of each implant was examined at placement and 1, 2, 3, 4, 5, 6, 8, and 10 weeks after the surgery using resonance frequency analysis (RFA) and damping capacity measurement. The development of implant stability focusing on the decrease in stability (as measured by implant stability quotient (ISQ)) and the interplay of primary (ISQ0) and secondary implant stability was evaluated. The implants were divided into three groups based on primary stability: group L (ISQ0 < 68), group M (ISQ0 68 to 72), and group H (ISQ0 > 72). Stability curves for each group were created and analyzed statistically. Implant stability measurement results gained with RFA and damping capacity were compared employing the Wilcoxon paired test, correlation coefficients, and regression analysis. The threshold for statistical significance was set at P < .05. Results: The most pronounced decrease in ISQ values occurred 1 week after implant placement (mean decrease of 2.2 ISQ). During the 10-week experiment, mean ISQ rose by 5.5 in group L and by 1.3 in group M and dropped by 1.8 in group H (P < .001). The coefficient of determination R² = 0.06 showed a weak dependence of RFA on the damping capacity (P < .001). Conclusions: Implants with low primary stability showed a significant increase in stability during healing. In contrast, implants with high primary stability lost some stability over time.

Implant stability is considered one of the most important parameters in implant dentistry. It affects the healing and successful osseointegration of implants. Its importance is further increased when employing modern treatment protocols, i.e., accelerated treatments such as immediate loading. Implant stability (total stability) is usually divided into two stages: primary stability (implant stability reached during implant placement) and secondary stability (implant stability after healing). Primary implant stability has been proven to be a mechanical phenomenon, whereas secondary stability is a result of biologic events (osseointegration). The proportion of biologic and mechanical components varies during the healing period. At the time of implant placement, implant stability is based solely on the mechanical component. During the healing period, mechanical stability decreases, whereas biologic stability increases. Finally, for an osseointegrated implant, stability relies entirely on the biologic component. This implies that an implant that has been loaded after a healing period resists masticatory forces by means of biologic stability, whereas an immediately loaded implant is immobile immediately after insertion only as a result of mechanical stability.

According to conventional opinions, overall implant stability increases during the healing process. However, this appears to be a rather simplified view of the complex healing process. More precisely, implant stability increases during healing only in implants with low primary stability, whereas in implants with high primary stability, a decrease in stability is observed. Therefore, the primary stability affects the development of stability during the healing process. However, this pattern of implant stability development was demonstrated with a delayed-loading protocol. For immediately loaded implants, the stability curve may follow a different course as a result of the functional loading of implants during the healing period. There is a need for larger studies to confirm this finding.

Several methods have been proposed to determine implant stability non-invasively in clinical practice, but only two of them—measurement of the damping capacity and resonance frequency analysis (RFA)—have been considered sufficiently valid. The only commercially available device utilizing damping capacity measurement is the Periotest device (Siemens). The determination of implant stability with this device is determined by tapping on a rod abutment and recording its contact time, which is considered to be a function of implant mobility. The result is displayed using numeric values ranging from -8 to +50, referred to as Periotest values (PTVs). The lower the value, the greater the stability.

The most reliable noninvasive method to measure implant stability is RFA. This method was introduced by Meredith et al in 1996. The RFA principle is predominantly used in devices of the Osstell series, with the wireless Osstell being its most recent modification. A magnet on an aluminum metal rod (SmartPeg) is screwed into the implant. After it receives a signal from the device, vibrations in two perpendicular directions are produced. Because the resonance frequency is directional, the highest and the lowest values are presented simultaneously. If the numeric difference between the values is greater than three units, both values are displayed. Higher resonance frequencies correspond to higher implant stability. The resonance frequencies are transformed into implant stability quotients (ISQs), which range from 0 to 100.

Numerous studies have investigated the development of implant stability during the healing period. Some have recorded implant stability only at placement and compared it with the stability obtained after healing is complete. However, this does not provide an accurate record of how implant stability is established. Longitudinal monitoring of implant stability has provided data indicating that implant stability is not established in a linear fashion. In the case of a slow increase in biologic stability and a rapid decrease in mechanical stability, a transient decrease in overall stability during healing occurs. This phenomenon has been termed a “dip” (or “drop” or “gap”) in stability. In principle, it is caused by the loss of mechanical stability when not sufficiently compensated by the growing biologic stability. The existence and pattern of the stability dip are probably influenced by a variety of factors, such as the quality of bone, final insertion torque, and implant design, especially its surface. In some studies, no dip was reported, while other studies have reported differences in its timing, duration, and depth. Understanding this issue is crucial for accelerated loading protocols.

The aims of the present prospective clinical study were, in light of previous research, (1) to monitor the development of implant stability in immediately loaded implants during the initial healing period, (2) to investigate how primary stability affects stability post-healing, (3) to compare measurements of implant stability obtained with RFA and damping capacity, and (4) to determine mutual relationships between selected insertion parameters (type of bone, final insertion torque, and primary stability). The experiment was conducted using implants with an alkali-treated surface, a surface that shows signs of bioactivity.
Measurement of Implant Stability and Marginal Bone Loss

The stability of each implant was measured at baseline and 1, 2, 3, 4, 5, 6, 8, and 10 weeks after the surgery using the Osstell and Periotest devices. All measurements were performed by an experienced surgeon (TB). At each follow-up visit, the Osstell device was used initially. The provisional prostheses and abutments were removed and the SmartPeg was screwed to each implant and tightened to approximately 35 Ncm. The transducer probe was aimed at the small magnet on top of the SmartPeg at a distance of 2 to 3 mm and held stable during the pulsing time until the instrument beeped and displayed the ISQ value. If two ISQ values were displayed simultaneously, the mean value was recorded. Measurements were taken twice in the buccolingual direction as well as in the mesiodistal direction. The mean of all measurements was rounded to the nearest whole number and was regarded as representative of the ISQ. The abutments were then screwed back on the implants and tightened to 35 Ncm; thereafter, measurements using the Periotest device were performed. The stylus was positioned perpendicular to the abutment in a buccolingual direction as apically as possible. Measurements were repeated until the same value was obtained twice in succession. This value was recorded. Then the provisional prosthesis was reinserted.

Statistical Analysis

Statistical analysis was carried out using Statistica software. The Wilcoxon paired test, correlation coefficients, and regression analysis were employed. The statistical significance of all tests was defined as P < .05. A dip in stability was defined as a significant drop in implant stability in one or several consecutive weeks, followed by a significant increase. Statistical analysis of the entire cohort of implants was accomplished and the development of stability (as compared with ISQ0) was evaluated. For this purpose, the implants were divided into three groups: those with low primary stability (ISQ0 < 68, group L); those with moderate primary stability (ISQ0 68 to 72, group M); and those with high primary stability (ISQ0 > 72, group H). Stability curves were created for each group and evaluated statistically.

Results

One implant (1.1%) was removed after 8 weeks because of mobility. Ten weeks postplacement, all remaining implants were classified as successful. With regard to bone type, 33.3% of implants were placed in type 1 bone, 45.6% were placed in type 2 bone, and 21.1% were placed in type 3 bone. The mean (± standard deviation [SD]) final insertion torque was 61.3 ± 11.5 Ncm (6 implants with 31 to 40 Ncm, 12 implants with 41 to 50 Ncm, 26 implants with 51 to 60 Ncm, 33 implants with 61 to 70 Ncm, and 13 implants with 71 to 80 Ncm). The mean (± SD) primary stability, as measured with RFA, was 72.5 ± 5.5 ISQ (2 implants with 51 to 60 ISQ, 6 implants with 61 to 65 ISQ, 25 implants with 66 to 70 ISQ, 31 implants with 71 to 75 ISQ, 19 implants with 76 to 80 ISQ, and 7 implants with 81 to 85 ISQ). The mean primary stability (± SD) as measured via damping capacity was -4.6 ± 1.3 PTV (1 implant with 0 to -1 PTV, 14 implants with -2 to -3 PTV, 51 implants with -4 to -5 PTV, and 24 implants with -6 to -7 PTV). A correlation between the final insertion torque and bone type was confirmed (r = -0.63, P < .001). A correlation between bone type and ISQ0 was not confirmed (r = -0.07, P > .05), and there was no correlation between final insertion torque and ISQ0 (r = 0.02, P > .05). On the other hand, correlations between bone type and PTV0 (r = 0.43, P < .001) and between final insertion torque and PTV0 (r = 0.40, P < .001) were confirmed.

Stability curves created from the ISQs and PTVs are presented in Fig 1, and stability values are presented in Table 1. While a dip in stability was not detected by damping capacity, it was obvious and statistically significant as measured with RFA. The most pronounced dip in ISQs occurred 1 week after implant placement, with a mean decrease of 2.2 (range, -7 to +4, P < .001). The ISQs rose significantly every week until the fourth week (P < .001, P < .05, P < .001, and P < .001, respectively). The closest value to ISQ0 was reached at 5 weeks postplacement, and the ISQs then
rose continuously but insignificantly until the end of the experiment. While ISQ10 was not significantly different from ISQ0, PTV10 was significantly lower than PTVO (mean difference, 0.96 PTV, \( P < .001 \)).

The coefficient of determination describing the dependence of ISQ on PTV values was \( R^2 = 0.06 \) (\( P < .001 \)). Thus, the strength of this relationship was rather low, but it remained statistically significant (Fig. 2).

A multiple regression model was used in which ISQ10 was a dependent variable and ISQ0, density, and torque were independent variables (\( R^2 = 0.637 \)). According to this analysis, only ISQ10 was dependent on ISQ0 (\( P < .001 \)), while ISQ0 was not related to density or torque (\( P > .05 \)).

Fifteen implants were included in group L, 29 implants in group M, and 46 implants in group H. Stability curves for each of these groups are shown in Fig. 3, and the respective stability values are shown in Table 2. The stability dip was most significant at 1 week postimplantation in all groups, when it reached 3.5 ISQ in group L, 1.8 ISQ in group M, and 2.0 ISQ in group H (\( P < .001 \)). The stability dip was greater in group L than in groups M and H (\( P < .01 \)). During the 10-week experiment, implant stability rose by 5.5 ISQ in group L and by 1.3 ISQ in group M and dropped by 1.8 ISQ in group H (\( P < .001 \)). Differences between the three groups were highly significant (\( P < .001 \)).

The progression of ISQs and PTVs in the only failed implant is shown in Fig. 4. The implant was inserted in type 3 bone with an insertion torque of 60 Ncm. During the first 6 weeks after placement, this implant did not show any signs of failure and all performed measurements were free of pain. However, 8 weeks after placement it was not possible to remove the abutment without causing pain, and the implant was removed from its bone bed, with the site anesthetized locally.

### Discussion

The present study was designed to accomplish, as much as possible, standardized experimental conditions. The implants were inserted in the interforaminal area of the mandible, where compact bone prevails. All implants featured the same length, diameter, and surface properties. In the present study, high values for insertion torque were achieved as a result of the bone quality in the anterior mandible, the implant design, and the omission of tapping. A high final insertion torque may be useful for immediate implant loading and can contribute to safe manipulation with the implant during the healing period. Insertion torque can be measured easily and is considered to be an indirect indicator of primary implant stability. However, the present study observed a correlation of insertion torque with PTV only; a correlation between final insertion torque and ISQ could not be confirmed.

Any dip in implant stability has fundamental clinical importance for immediately loaded implants. If the stability decreases below a critical level during the healing process, a functionally loaded implant cannot withstand masticatory forces, becomes mobile, and fails. There is a considerable lack of agreement regarding the parameters of the dip in postinsertion stability. Studies that have demonstrated this dip in stability have usually observed it between the second and eighth weeks following implant placement. The maximum stability drop was detected during the third or fourth week postplacement and ranged from 2 to 12 ISQs below the baseline ISQ. However, some studies did not observe this decrease in stability. These differences in results may be related to variations in the design of implants employed, especially variations in surface properties. A time dependence of implant stability, without the initial decline, has been observed in association with rapid increases in bone-implant contact. This feature is typical for fluoride-treated or alkali-treated, ie, potentially bioactive, surfaces. Accelerated formation of bone-to-implant contact contributes to a faster increase in biologic stability. This biologic process compensates for any decrease of mechanical stability and ensures consistency in stability over time, without the drop during the healing period. Geckili et al measured the stability of titanium grit-blasted dental implants with and without fluoride treatment longitudinally in a comparative study. Implants were inserted interforaminaly in the mandible and were followed for 24 weeks. Implants without fluoride treatment showed a statistically significant drop in ISQ (mean, 4.9 units) in the first 2 weeks after implant placement. This change was statistically insignificant in the second group of implants, suggesting that fluoride modification of the implant surfaces may enhance the osseointegration process. Similar trends were observed with fluoride-treated implants in other studies.

The results are ambivalent for the widely used SLA (sandblasted, large-grit, acid-etched; Straumann) and SLActive surfaces, despite the unquestionably positive effect of both surfaces on osseointegration. Sim and Lang detected a continuous increase in ISQs without signs of a dip in implants with an SLA surface during a 12-week period after implant placement. Valderrama et al came to similar conclusions. On the other hand, Schätzle et al detected a dip in stability in palatal implants with SLA and SLActive surfaces. However, features of the stability curve were different for both surfaces, suggesting a tendency for the SLActive surface to contribute to a decreased healing time. Han et al also detected a period of reduced stability, but without any difference between SLA and SLActive surfaces. ISQs decreased by 3 to 4 units during healing and reached the lowest values in the third week. Following this, the ISQs increased steadily up to the 12th week. Lai et al detected a dip in stability between weeks 2 and 6 for implants with the SLA surface. Depression of the curve was significant and reached 12 ISQs.

Fig. 1: Mean PTVs during the study period

<table>
<thead>
<tr>
<th>Value</th>
<th>0</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
<th>5</th>
<th>6</th>
<th>7</th>
<th>8</th>
<th>9</th>
<th>10</th>
</tr>
</thead>
<tbody>
<tr>
<td>ISQ</td>
<td>72.5 ± 5.5</td>
<td>70.3 ± 6.2</td>
<td>70.7 ± 5.8</td>
<td>71.7 ± 5.4</td>
<td>72.3 ± 4.9</td>
<td>72.4 ± 4.4</td>
<td>72.6 ± 4.1</td>
<td>72.7 ± 3.6</td>
<td>72.9 ± 3.1</td>
<td></td>
<td></td>
</tr>
<tr>
<td>PTV</td>
<td>-4.6 ± 1.3</td>
<td>-5.0 ± 1.1</td>
<td>-5.1 ± 1.3</td>
<td>-5.1 ± 1.5</td>
<td>-5.0 ± 1.3</td>
<td>-5.5 ± 1.6</td>
<td>-5.4 ± 1.2</td>
<td>-5.7 ± 1.1</td>
<td>-5.6 ± 1.1</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 1: Mean ISQs and PTVs (± SDs) During the Study Period
Dental implants

Fig. 2: Regression of ISQs on PTVs. The correlation was weak ($R^2 = 0.06$) but significant ($P < .001$).

Fig. 3: Development of implant stability (in ISQ) in relation to primary stability. Group H: ISQ0 > 72; Group M: ISQ0 68 to 72; Group L: ISQ0 < 68.

Fig. 4: Development of stability in a failed implant, which was removed 8 weeks postplacement.

The present study confirmed most previously published results. When ISQ0 and ISQ10 values were compared, group L, with the lowest primary stability, presented a significant increase in ISQs (average increase of 5.5 ISQ, $P < .001$). On the other hand, group H implants, with high primary stability, demonstrated a significant drop in stability (on average, 1.8 ISQ, $P < .001$). Group M implants, with moderate primary stability, presented a mild but a significant increase in stability (on average, 1.3 ISQ, $P < .001$), in contrast to an earlier study conducted by the same research group. This minor discrepancy may be a result of

Table 2: Mean ISQs (± SDs) in Each Group During the Study Period

<table>
<thead>
<tr>
<th>Group</th>
<th>0</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
<th>5</th>
<th>6</th>
<th>8</th>
<th>10</th>
</tr>
</thead>
<tbody>
<tr>
<td>Group H</td>
<td>76.6 ± 3.1</td>
<td>74.7 ± 3.5</td>
<td>74.8 ± 3.1</td>
<td>75.6 ± 3.0</td>
<td>75.7 ± 2.6</td>
<td>75.5 ± 2.3</td>
<td>75.4 ± 2.3</td>
<td>75.2 ± 2.6</td>
<td>74.8 ± 2.7</td>
</tr>
<tr>
<td>Group M</td>
<td>70.2 ± 1.5</td>
<td>68.4 ± 2.7</td>
<td>68.9 ± 2.9</td>
<td>69.6 ± 2.5</td>
<td>70.3 ± 3.0</td>
<td>70.4 ± 3.2</td>
<td>70.6 ± 3.3</td>
<td>70.8 ± 2.5</td>
<td>71.5 ± 1.3</td>
</tr>
<tr>
<td>Group L</td>
<td>64.1 ± 3.8</td>
<td>60.7 ± 4.3</td>
<td>61.7 ± 4.5</td>
<td>63.7 ± 4.2</td>
<td>65.6 ± 4.3</td>
<td>66.7 ± 3.5</td>
<td>68.0 ± 3.5</td>
<td>68.9 ± 2.9</td>
<td>69.7 ± 2.8</td>
</tr>
</tbody>
</table>
the arbitrary classification of the primary stability into three intervals. When analyzing the dip in implant stability with respect to primary stability, the observed dip in stability was significantly more pronounced in group L than in the other two groups. Group L had less bone-to-implant contact initially, such that the same pattern of remodeling could reduce the percentage of bone-to-implant contact more significantly than for group H. Very similar results were observed in implants with the SLA surface by Bereswill et al.14 Implants in type 4 bone had significantly lower primary stability and showed a significantly greater dip in stability than implants in types 1, 2, or 3 bone. The development of implant stability in all three groups was characterized by an initial drop in stability, which was most pronounced at 1 week after implant placement. An increase in implant stability followed. It could be hypothesized that this phenomenon is caused by an unusually fast decrease in stability as a result of the implant design or the above-average bone mineralization in the anterior mandible, combined with a rapid onset of osseointegration of the acid- and alkali-treated implant surface. To confirm this hypothesis, further studies are necessary. Further developments, consisting of a stability increase in groups L and M over the initial values and a drop in implant stability in group H to below the initial value, could be explained by an overall trend of achieving a common level of secondary implant stability.4 This trend could be influenced by the biologic response of variously mineralized bone to the immediate loading.

Contradictory results have been reported on the comparability of RFA and damping capacity measurements. Cehreli et al performed a meta-analysis and reached the conclusion that there is no correlation between both methods.39 Determination of the damping capacity is considered to be more susceptible to clinical variables influencing the measurements and less precise.34,35 On the other hand, several animal experiments, human cadaver studies, and clinical studies showed moderate to strong correlation of both methods.34,35 Some in vitro experiments with the Osstell and Periotest devices found a linear association between measurements, with high statistical correlation coefficients of -0.9 and -0.8.36,39 However, in the present study, dependence of ISQ values with the Osstell and Periotest devices found a linear association between both methods.

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References

5. Zhou W, Han C, Li Y, Li D, Song Y, Zhao Y. Is the osseointegration of immediately and delayed loaded implants the same? Comparison of the implant.


Changes in Stability After Healing of Immediately Loaded Dental Implants

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Purpose: To investigate the parameters that affect primary stability of dental implants, to determine how primary stability influences posthealing stability, and to ascertain the effect of primary stability and insertion parameters on marginal bone loss. Materials and Methods: A total of 940 immediately loaded implants were considered. Using resonance frequency analysis, primary stability (primary implant stability quotient [pISQ]) and stability after 4 months (tISQ) were recorded. When the differences between pISQ and tISQ exceeded 5 units, marginal bone loss was measured. The implants were placed into three groups based on their primary stability: high (pISQ > 72), moderate, and low (pISQ < 68). Changes in stability after 4 months of loading were evaluated. The relationships between pISQ, insertion parameters, ΔISQ (ie, tISQ – pISQ), and marginal bone loss were analyzed. The Student t test, one-way analysis of variance, and Spearman nonparametric correlation coefficient were employed for statistical evaluation. Results: Of the 940 implants, tISQ was recorded in 526 implants and marginal bone loss was measured in 76 implants. There was no statistical relationship between pISQ and insertion torque. Primary stability was influenced by implant diameter but not by implant length. There was a significant relationship between implant insertion torque and bone type. The low primary stability group showed a significant increase in stability during healing. However, high primary stability implants demonstrated a significant reduction in their stability. The linear regression analysis demonstrated that at a pISQ of 69.2, tISQ value would equal pISQ value. Correlations between marginal bone loss and final insertion torque and between marginal bone loss and ΔISQ values were observed. Conclusions: Stability of immediately loaded implants with high pISQ decreased significantly during the initial 4 months of healing. However, stability of implants with low primary stability increased significantly. ΔISQ and insertion torque showed correlation with marginal bone loss. Int J Oral Maxillofac Implants 2010;25:1085–1092.

Dental implant stability is a measure of the anchorage quality of an implant in the alveolar bone and is considered to be the consequential parameter in implant dentistry. Implant stability has been confirmed to affect the process of osseointegration, the pattern of implant loading, and, finally, the success of an implant.1 Stability of an implant can be classified into that measured immediately after implantation (ie, primary stability) and that seen posthealing (ie, secondary stability). Primary implant stability has been proven to be a mechanical phenomenon.2 On the other hand, secondary stability occurs through a cascade of biologic events, such as bone regeneration and remodeling at the bone-implant interface.2 It is influenced by many factors, including implant surface topography, bone quality, and patient behavior.3 Earlier investigation showed that during the healing process, mechanical anchorage of the implant in the bone weakens; conversely, biologic stability of the implant increases.4 Several methods have been proposed to determine implant stability in clinical practice. Among these, resonance frequency analysis (RFA) has been found to be the most accurate.5 Meredith et al introduced RFA into implant dentistry in the 1990s. Since then, it has become a widely accepted and used technique.6 The only commercially available device based on RFA is Osstell (Integration Diagnostics). It has lately been modified and upgraded in the Osstell Mentor device. The function of this instrument is to measure resonance frequency, which is automatically transformed into an implant stability quotient (ISQ) ranging from 0 to 100.7

The absolute RFA values are not completely dependent on the quality of osseointegration. There are three important factors that can directly influence RFA: the stiffness of the implant-bone interface, the stiffness of the bone itself, and the stiffness of the implant components.6–8 Consequently, the clinically measurable characteristic of implant stability can be compared in the follow-up of each individual implant, but caution should be exerted in comparing these values among different implants, either in the same patient or interindividually.9

Previous longitudinal studies have indicated that implant stability changes during the process of osseointegration. Typically, implant stability is anticipated to decrease during the early weeks of healing; this is followed by an increase in stability.10,11 This is related to the biologic reaction of the bone to surgical trauma. During the initial bone remodeling phase, bone and necrotic material are resorbed by osteoclastic activity, which is reflected by a reduction in the ISQ value. This process is followed by new bone apposition initiated by osteoblastic activity, ie, adaptive bone remodeling around the implant.3,11 There is a lack of agreement among investigators regarding the exact timing of the greatest decrease in postinsertion stability of an implant; the recorded data range between the third and eighth weeks following implant placement.12–18 Some studies did not observe any decline in stability during the healing phase.17,18 The reason for these differences in results may have to do with variations in the designs of the implants employed, especially variations in surface properties.8,11 Time dependence of implant stability without the initial decline was observed in association with fast increases in bone-implant contact, which is typical for fluoride-treated or alkali-treated (and thus potentially bioactive) surfaces.11,12 An accelerated formation of bone-to-implant contact contributes to a faster increase in secondary stability. This biologic process eliminates the decrease in primary stability and ensures consistency of stability over time (without the drop during the healing period).18 There is a limited amount of documentation about the relationship between primary and secondary stability. Sennerby and Meredith11 confirmed that implants of many types would, over time, approach a similar level of secondary stability. He also denoted that consistent attainment of an ISQ value of 65 to 75 seems to correspond to Bränemark implants and an ISQ value of 55 to 65 was seen for Straumann implants.11

Hence, it was the intent of this retrospective clinical study to further elucidate some aspects related to the stability of immediately loaded implants under relatively uniform clinical conditions in the interforaminal region of the mandible. The objectives of the present study were (1) to investigate how primary stability influences posthealing stability, (2) to determine the parameters that can affect primary stability of dental implants, and (3) to ascertain the effect of primary stability and insertion parameters on marginal bone loss.
Materials and methods

Surgical Procedure and Measurement of Implant Stability

In this study, consecutively placed implants in the interforaminal region of the mandible, which were designed for the immediate loading concept „Teeth in 6 Hours,” were considered. This concept is based on the insertion of five implants in the region between the first premolars, which are then immediately loaded by a provisional cantilevered prosthesis fabricated from an existing mandibular removable complete denture and attached to the abutments by means of titanium impression copings. All surgical procedures were performed between October 2004 and January 2008 at The Center for Dental Implantology, University Hospital, Hradec Kralove, Czech Republic. The local ethical committee officially approved the design of this study. All the patients were informed about the nature of the study, and their participation and written consent were obtained according to the Helsinki Declaration of 1994.

All included patients needed a fixed full-arch prosthesis supported by dental implants for their edentulous mandible. Patients were included based upon a current stable medical condition and the ability to withstand the stress of dental implant surgery. Patients with metabolic bone disease, unstable systemic conditions (eg, uncontrolled diabetes or untreated hypothyroidism), and smokers of more than five cigarettes a day were excluded. All surgical procedures were performed under local anesthesia in a sterile hospital setting. Amoxicillin clavulanate (1 g orally twice per day) was prescribed for 6 days; an initial dose (2 g) was administered 1 hour before surgery. All implants were inserted into healed extraction sockets. After a mucoperiosteal flap was raised, both mental nerves were isolated and the alveolar crest was contoured as required. Then five self-drilling, screw-form implants with sandblasted, acid-treated, and alkali-treated surfaces (STI-Bio-C, Lasak) were inserted at regular intervals into the interforaminal region according to the manufacturer’s protocol. The final insertion torque of the implants was measured using a torque wrench. The type of bone was classified using Lekholm and Zarb criteria on the basis of the subjective evaluation of the surgeon.11

Primary stability (\(pISQ\)) of each implant was measured using an Osstell device. Two experienced surgeons conducted measurements independently. The transducer was secured at the implant level perpendicular to the long axis of the alveolar bone. Measurements were repeated until the same value was measured twice in succession, and this value was recorded. Following this, the abutments were attached and the wound was sutured. Tiaprofenic acid (300 mg) was recommended three times a day for pain relief. An abutment-level impression was immediately made with additional silicone material (Aquasil Rigid and Aquasil Ultra LV, Dentsply Caulk) with a modified preformed plastic impression tray. A cantilevered fixed screw-retained provisional prosthesis was fabricated that extended to the second premolars. The prosthesis was delivered and fully functionally loaded. Oral hygiene instructions were given and the patients were scheduled for regular recall appointments.

The impression for the definitive prosthesis was made after 4 months of healing. For those patients for whom an impression was made in September 2006 or later, the measurement with the Osstell device was repeated for each implant. The measured values were recorded and denoted as \(tISQ\). After confirming passive fit of the construction and correcting the occlusion, a definitive cantilevered prosthesis that extended to both first molars was fabricated and delivered within 2 weeks after taking the impression. A digital panoramic radiograph (Planmeca ProMax) was obtained immediately after fixation of the definitive prosthesis according to the manufacturer’s recommendation and was done by the same technician.

Measurement of Marginal Bone Loss

Marginal bone loss was determined at all implants at which the absolute value of the difference between \(pISQ\) and \(tISQ\) exceeded 5 units. The measurement of bone loss was conducted independently by the two surgeons on the patient's digital panoramic radiographs. On the radiograph, bone levels were measured from the implantabutment interface to the first visible bone-implant contact. The implantabutment interface was used as a reference point, because the implants were normally placed with the implant-abutment connection at the level of the alveolar crest (Fig. 1). The distance between the thread peaks (1.0 mm) served as a known standard to calculate the exact bone loss on the mesial and distal sides of the implants. These measurements were rounded to the nearest 0.1 mm. With these data, the mean marginal bone resorption was determined for each implant. However, if the radiograph did not clearly reproduce the exact bone level, the implant was excluded from the cohort.

Statistical Analysis

The difference between posthealing stability (after 4 months of loading) and primary stability (\(tISQ – pISQ\)) was denoted as \(\Delta ISQ\). The linear regression line, calculated from the plot of \(\Delta ISQ\) versus \(pISQ\), was used to determine a \(pISQ\) value at which \(\Delta ISQ\) attains a value of zero. With respect to this value and to the unpublished results of a previous investigation, the implants were further divided into three study groups: those with low primary stability (\(pISQ < 68\)), those with moderate primary stability (\(pISQ 68 to 72\)), and those with high primary stability (\(pISQ > 72\)). Statistical analysis was employed to verify the main working hypothesis that the...
immediately loaded implants with higher primary stability would lose some of their stability during healing, whereas the implants with lower primary stability would gain stability during healing.

In addition, the dependence of marginal bone loss on pISQ, on final torque, and on ΔISQ was evaluated. Additional working hypotheses were that pISQ, similar to the final torque, is positively correlated with marginal bone loss, while ΔISQ is negatively correlated with marginal bone loss. Statistical analysis was carried out using Statistica software (Statsoft Inc.). The Student t test, one-way analysis of variance, and Spearman nonparametric correlation coefficient were employed to test the hypotheses. The statistical significance of all tests was defined as P < .05.

Results

A total of 940 dental implants placed in 188 patients (84 men and 104 women; mean age 54.3 ± 9.4 years) was initially considered for this investigation. However, Osstell measurements showed an invalid Bode diagram for 22 implants, and these implants were therefore excluded from the cohort. Thus, the remaining 918 implants were considered for statistical analysis. The majority (97.2 %) of implants were 3.7 mm in diameter, whereas only 2.8 % of implants were 5.0 mm in diameter. A large majority of implants (82.2 %) were 16 mm long; 9.3 % were 14 mm long, 3.8 % were 12 mm long, 3.8 % were 18 mm long, and 0.9 % were 10 mm long. With regard to bone type, 37.5 % of implants were placed in type 1 bone, 40.4 % were placed in type 2 bone, 21.8 % were placed in type

Fig. 3: Primary stability (in ISQ) of the implants.

Fig. 4: Implant stability over time according to the level of primary stability (see text). The decrease in ISQ for implants with high primary stability and the increase in ISQ for implants with low primary stability were highly significant (P < .001).

Fig. 5: Marginal bone loss versus final insertion torque (R = 0.27; P < .05).

Fig. 6: Marginal bone loss versus change in stability (ΔISQ) (R = −0.27; P < .05).

Fig. 7: Change in stability (ΔISQ) versus primary stability (pISQ) (R = −0.47; P < .01). The regression curve indicates that ΔISQ attains zero value at a pISQ of 69.2.
3 bone, and 0.3 % were placed in type 4 bone. Of the total number of implants placed, six implants (with pISQ 61 to 79) failed to osseointegrate (two implants in one patient and one implant each in four other patients). The osseointegration rate was therefore 99.3 %.

The distribution of the final torque of implants is shown in Fig. 2. The mean final insertion torque for the implants was 60.2 ± 12.0 Ncm (65.7 ± 7.2 Ncm for type 1 bone, 61.9 ± 10.3 Ncm for type 2 bone, 52.3 ± 14.6 Ncm for type 3 bone, and 30.0 ± 0.0 Ncm for type 4 bone). Statistical comparison of implant insertion torque versus bone type at the site of implantation showed a highly significant relationship (P < .001; between type 3 and type 4 bone, P < .05). No significant correlation between the final torque and implant diameter was found (61.4 ± 11.4 Ncm for 3.7-mm-diameter implants, 60.8 ± 11.9 Ncm for 5.0-mm-diameter implants). Figure 3 shows the distribution of pISQ among the surveyed implants. The recorded mean pISQ value was 72.2 ± 5.0. The mean pISQ values for each bone type were type 1 bone, 71.8 ± 4.9 for type 2 bone, 72.7 ± 5.1 for type 3 bone, and 71.3 ± 2.5 for type 4 bone. There was no significant difference among groups, except for a marginally significant difference between type 2 and type 3 bone (P = .03). Furthermore, statistical analysis disproved the dependence of pISQ on implant length; mean values for pISQ were 70.8 ± 6.1, 73.4 ± 5.3, 72.0 ± 4.4, 72.2 ± 5.0, and 71.9 ± 4.6 for implants with lengths of 10, 12, 14, 16, and 18 mm, respectively. However, the primary stability of 5.0-mm-diameter implants was significantly higher than that of 3.7-mm-diameter implants (pISQ 75.1 ± 5.2 versus 72.1 ± 4.9, respectively; P < .01). No significant correlation was found between pISQ and final torque (pISQ of 69.0 ± 5.9, 72.3 ± 3.7, 72.1 ± 5.1, 71.9 ± 4.9, and 72.5 ± 5.0 for a final torque of < 15.1, 16.3 to 35.6, 45.9 to 50, and > 50 Ncm, respectively; P > .05). In this current study, the tISQ value was measured for 526 implants. Among these, 100 belonged to the low primary stability group, 189 to the moderate primary stability group, and 237 to the high primary stability group. An increase in stability was seen during the healing period in the low primary stability group (from 64.2 ± 2.8 ISQ to 66.8 ± 5.6; P < .001) (Fig. 4). The moderate primary stability group did not exhibit any significant change in stability (from 70.3 ± 1.4 to 70.0 ± 5.4) (Fig. 4). However, the high primary stability group showed a decrease in stability during the healing phase (from 75.9 ± 2.6 to 72.0 ± 5.0; P < .001) (Fig. 4).

Marginal bone loss was measured for 76 implants and had a mean of 0.9 ± 0.7 mm (range, –1.0 to 2.6 mm). The values measured on the mesial side of the implant (0.9 ± 0.7 mm; range –0.9 to 2.9 mm) and on the distal side (0.9 ± 0.7 mm; range –1.1 to 2.9 mm) did not differ significantly. A statistically significant relationship between primary stability and bone loss was not confirmed (Spearman nonparametric correlation coefficient, R = 0.07). However, a positive correlation was found between final torque and bone loss (R = 0.27; P < .05) (Fig. 5). In addition, negative correlations were also found between ΔpISQ and bone loss (R = –0.27; P < .05) (Fig. 3) and between pISQ and ΔpISQ (R = –0.47; P < .01) (Fig. 7). Linear regression analysis indicated that ΔpISQ attains a value of zero at a pISQ of 69.2 (Fig. 7).

Discussion

A 99.3% success rate of the implants confirmed that immediate loading of splined implants in the interforaminal region is a viable treatment alternative. Primary stability of the failed implants did not differ significantly from that of implants that osseointegrated successfully. Measurement of RFA at the time of implant placement is therefore incapable of predicting implants with a prognosis of nonosseointegration, as described elsewhere.22 The topic of primary stability is currently the subject of intense scientific interest. Several authors have investigated the relationship between pISQ and other parameters, particularly the final insertion torque. Although a positive correlation between pISQ and insertion torque may initially seem probable, many authors have not found a significant relationship.23–27 The results of the present study are in agreement with the aforementioned studies and do not support the sporadic findings of contradictory results.28–31 However, it cannot be excluded that the discrepancies in the results were affected by the design of the implant or by local bone quality. As was also confirmed by the present study, primary stability is influenced mainly by implant diameter,32–36 and not by implant length.23,30–32 This study demonstrated a highly significant relationship between final torque and bone type. However, no statistical relationship was found between final torque and implant diameter. Several authors have studied the effect of primary stability on the development of stability during healing. The recent investigation by Karl et al assumed a general increase in stability during healing as a common phenomenon. This appears to be the somewhat confused view of the authors.34 A few groups of authors have indicated that changes in stability during healing were mainly dependent on the initial stability level of an implant. In their 12-week clinical study, Nedir et al35 found that 1TI implants with a pISQ < 60 exhibited an increase in stability, whereas implants with a pISQ of 60 to 69 exhibited decreased stability after 8 weeks. At the end of the 12th week, the implants had returned to their initial stability values. Implants with pISQ values > 69 exhibited decreased stability during the first 4 weeks, after which they maintained consistent stability.22 In a similar longitudinal study, West and Oates36 employed the same type of implants and divided them into two groups (pISQ < 56 and pISQ > 56). During the first 16 weeks, implants from the first group continuously maintained a lower ISQ versus implants from the second group. Thereafter, differences were statistically insignificant. The stability of both groups remained at a value of 61 ISQ. Similarly, in 1999, Friberg et al stated that the stability of implants placed in softer bone would „catch up” over time to implants placed in denser bone.36 Balshi et al and Olsson et al37 came to the conclusion that implants with high primary stability lose part of their stability during healing, whereas implants with low primary stability have a tendency to increase their stability. The results of the present study support this theory. A significant increase in stability was recorded for the implants with low primary stability (pISQ < 68), whereas the implants with high primary stability (pISQ > 72) lost some of their stability over time. This confirms the main working hypothesis of the current investigation.

It could be further hypothesized that, in clinical practice, it may be optimal to achieve a primary stability that corresponds to the final stability value of the osseointegrated implant (ie, pISQ = tISQ; ΔISQ = 0). In this study, the pISQ value was determined by linear regression analysis to equal 69.2. This hypothetical value is probably not generally valid but is more likely specific to a particular implant system, surgical protocol, or bone type.8 An intraoral standardized radiograph is frequently used for exact measurements of the amount of marginal bone loss. However, unfavorable anatomical conditions frequently prevent the use of this radiographic technique in the interforaminal region of the mandible, especially in patients with an atrophied edentulous arch.33 Consequently, panoramic radiography has been used in similar studies by other authors as an alternative.33–35,39,40 In the present study, a noteworthy finding of bone gain was frequently encountered at 4 months after implant placement. Similar findings have been reported elsewhere.41 An explanation may be found in the neck of the implant, which has a miniature thread and a chemically modified surface with signs of bioactivity. Functional stimulation of the bone by immediate loading may also play a role.42,43 Marginal bone loss was measured immediately after fixation of the definitive prosthesis, approximately 5 months after insertion of the implant. Bone loss could be caused by several factors, for example, surgical trauma, inadequate fit of the provisional restoration, or overloading of the implants. A detailed analysis, however, exceeds the scope of this study. The detected value of 0.9 ± 0.7 mm is acceptable.45 No relationship was found in this study between primary stability and marginal bone loss; thus, the first of the three additional working hypotheses was not confirmed.

However, correlations were confirmed between final torque and bone loss and between ΔpISQ and bone loss. Thus, the remaining two additional working hypotheses were supported. Taking the essence of these results and considering the fact that the final insertion torque did not correlate to primary stability, it is possible to infer further that the use of extremely high insertion torque should be avoided. On the other hand, it is necessary to emphasize that the regression analysis indicated a weak dependence between the variables. Therefore it can be concluded that these relationships are affected by additional factors that were not ex-
Within the limitations of the present study, it was seen that:

1. Primary stability was influenced by implant diameter and not by implant length. There was no significant relationship between primary stability and healing period.

2. Implants with low primary stability showed a significant increase in stability, while implants with high primary stability showed a significantly decreased stability over time. On the basis of linear regression analysis, the change in implant stability quotient attains a value of zero at a primary implant stability quotient of 69.2.

3. A statistically insignificant relationship was found between primary stability and marginal bone loss. A positive correlation was confirmed between the final insertion torque and marginal bone loss. A negative correlation was confirmed between primary stability and change in stability during healing and between change in stability during healing and marginal bone loss.

References

Extention of alveolar ridge without raising the mucoperiosteal flap using minimally-invasive dental implant surgery – a new step in effective implantology

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Introduction

It is not surprising that minimally-invasive flapless implant surgery has become a hit in the past years. With this method a dental implant is inserted without raising the mucoperiosteal flap. With regard to aesthetics, the benefits of flapless implant surgery are: reduced post-operative pain, less surgical and overall treatment time, faster healing and lower costs.1 This method has its origin some 5,500 years ago2 and it has been re-appearing regularly ever since, being particularly progressive at the end of the 19th and start of the 20th centuries before being overshadowed by the two-phase implant insertion, which was considered at the time to be safer.3 At the present time, since the 1990s, the method of minimally-invasive flapless procedure has been dramatically expanding – as it has been shown that its results are comparable, providing that the correct protocol is followed, with the two-phase method. In such cases where bone augmentation is not required – either to ensure implant stability or to achieve a good aesthetic result – the method of flapless implant surgery has multiple advantages for the patient as well as for the surgeon. Its many benefits are:

- a/ morphological – as the anatomy of the alveolar ridge is not disturbed, b/ nutritive – as the periosteum is left intact and the blood supply to the site maintained, healing is faster and the likelihood of crestal bone resorption and soft tissue inflammation minimized, c/ a reduced risk in medically-disadvantaged patients, i.e. patients with diabetes and reduced immunity (thanks to the reduced trauma and smaller wound size), d/ a reduced need for the use of analgetics – the reduced risk of inflammation and damage to the nerves brings about less pain, e/ psychological – the single-phase operation and the reduced overall treatment time makes it more acceptable to patients, f/ financial – a less complex protocol reduces the operation time and makes the treatment cheaper, thus accessible to a wider range of patients.

There are, nonetheless, disadvantages as well as advantages associated with this method:

- a/ Despite being a seemingly simple operation – it nevertheless requires a highly-experienced dental surgeon as the implant has to be inserted into the bone without direct visual checking of its position in the bone. The flapless procedure thus demands a surgeon who has considerable previous experience with flap surgery.4,5 b/ In the case of an unsatisfactory anatomy of the alveolar ridge (with regard to aesthetics or stability), conventional methods of alveolar ridge augmentation need to be employed.4,6

With the minimally-invasive procedure, implants are inserted without raising the periosteal flap, by employing one of the following procedures:

- a/ Immediate implantation – placed somewhere immediately between a flapless and an augmentation procedure – the implant is inserted immediately after tooth extraction, b/ punch incision – a cylindrical punch hole is made using trephine, the bone is then examined and the implant inserted, c/ transmucosal – implants are inserted directly through the mucoperiosteum.

When inserting the implant surgeons choose from the following procedures to guide them when inserting the implant:

- a/ clinical examination – implants are inserted without any further instrumentation other than using tactile orientation; this method can only be used in cases of abundant tissue, especially soon after tooth extraction when orientation is easier thanks to the structure of the alveolar socket, b/ drilling templates – derived from laboratory modelling of the future denture and the reconstruction of the real bone shape using invasive instruments (caliper, endodontic instruments) for mapping the bone width, and tomography (CT scan), or possibly by reconstruction of the probable bone shape on a cast model – this is the most commonly-used method, c/ navigation using a CAD/CAM method which includes CT-based implant planning and fabrication of a CT-scan-designed surgical template or possibly an individual implant abutment – this is, without doubt, the safest method but requires more preparation time and the cost is higher (CT-scan, fabrication of surgical template)6,9. Also not all dentists have access to a CT-scan.

Our indications for minimally-invasive implantology and results in its use

The use of the minimally-invasive implant procedure is limited in particular by the alveolar width. To comply with the widely-accepted recommendation, an implant should be surrounded by a minimum of 1 mm of bone. The method is used and is most suitable for alveolar ridges, which are of a generous height and width.1,10,11

Our dental surgery has offered minimally-invasive dental surgery since 1995 and with our increasing experience we have set up two indication groups based on alveolar width and implant diameter:

- a/ safe indications – the diameter of the alveolar ridge is in all directions and along its whole length larger by 4 mm than the diameter of the used implant. Our criterion is based upon the widely accepted requirement of 1 mm of intact bone surrounding the implant plus 2 mm as a cover for possible surgical inaccuracies (i.e., a 5 mm STI-BIO-C implant, LASAK Ltd., Czech Republic, would be inserted into an alveolar ridge of 9 mm width or more). A criterion of 4 mm is an arbitrary one and much depends on the surgeon’s skill. For a trainee surgeon just beginning (practical dentists)
we would suggest to increase that level to 5 mm, whilst for an experienced surgeon (specialised dental surgeon) 2-3 mm of alveolar width over the implant diameter might suffice. More than 50% of patients in general clinical practice, (in our case, more than 60% as we are specializing in more complicated cases), are not suitable for this type of implant procedure. Under this indication, between 1994–2006, we have inserted 486 implants, all of them having been loaded for a minimum of six months up until today. From this total number, 11 implants were lost (four of which were in one patient who lost all implants inserted due to a suspected systemic cause), and 8 implants had an extended healing period (in all these cases the implants had been inserted too close to one of the alveolar walls which resulted in bone fracture and exclusion of bone sequestrum). Out of this total number, we inserted 234 Impladent implants (LASAK Ltd., Czech Republic) and 252 Straumann implants (Straumann AG, Switzerland). We did not find significant differences in treatment success between these two implant systems; however, an exact comparison is not possible, as the Straumann implants, thanks to the conical shape of the TE system, were used in all cases of immediate implantation and inner sinus-lift operations, whilst the Impladent system was used in all other indications. The total success rate was 97.7% with 1.7% cases having healing complications. Due to the fact that many patients with medical contraindications (such as diabetes, moderate to heavy smoking, immunodeficiencies, liver disorders) were included, we consider these results comparable to the conventional treatment.

b/ Risky indications – when the diameter of the alveolar ridge is smaller than the safe indication and the augmentation of the alveolar ridge indicated was not performed (refused by the patient, due to medical complications, financial contraindication). In all such cases, patients were always informed about the advantage of having augmentation done. Since 1999, we have been using osteoplasty as developed by Tatum. It involves implant insertion without the use of rotating instruments; the bone is opened with the use of a bone scalpel and subsequently the alveolar bone along with the mucoperiosteum was widened using bone spreaders to obtain the shape and diameter required for a given implant (it is necessary for the mucoperiosteum to remain intact). Despite the fact that the system was developed by Hilt O. Tatum and for his own special implants, thanks to the calibration of bone instruments it can be used with other implant systems too. Osteoplasty is accompanied by bone condensation: bone not being removed but rather prepared for implant insertion. This allows for a predictable placement of a 5 mm Impladent implant into 2 mm bone. The weakness of this method is that it is difficult to reproduce and, thus, results are highly dependent on the skills of the surgeon employing it. It has been observed that some surgeons cannot achieve predictable results using this method, whilst others can in specific indications (such as narrow but high alveolus) achieve even better results than when using the conventional bone-augmentation method.

New methods of closed alveoplasty

In recent years, several new systems have evolved that enable highly reproducible alveoplasty not only in the horizontal direction but also in the vertical direction. The most complex system that we have used is the Bone Management System (Meisinger, Germany), that has a number of sub-systems, enabling both vertical as well as horizontal bone augmentation. Despite the fact that this system has been recommended by the manufacturers to be used for open alveoplasty, i.e. smoothing out the bone after the alveolar ridge has been exposed, we employed it, using our previous experience, for a minimally-invasive procedure using the systems of Horizontal-Control (authors Fuchs, Cierny) and Split-Control (authors Streckbein, Hassenpflug). This enabled us to perform even a closed, minimally-invasive osteoplasty. Later on, we modified the system in such a way as to enable the immediate loading of inserted abutments. The Horizontal-Control System consists of a calibrated series of conical instruments of increasing diameter that, when being inserted, themselves model the alveolar bone. The Split-Control System achieves a similar effect using spreaders with increasing diameters. The use of the Split-Control System leads to good bone condensation but it is not so good at bone modelling, whilst the Horizontal-Control System is just the opposite, so used together there can be a suitable control of both systems.

In spite of the Bone Management System being designed for use with the Meisinger system, it also works well with Impladent implants too (Lasak Ltd., Czech Republic). Only in complicated indications – e.g. in a maxillary alveolar ridge of low height in the distal reach which required an inner sinus-lift to be performed - did we use Straumann implants (Straumann AG, Switzerland), TE system; these show, with sinus-lifts, a higher stability due to their conical shape which prevents the screws entering the jaw cavity.

To ensure good primary stability, we left a minimum of 4 mm and 6 mm, respectively, of intact bone in the vertical direction, when using implants of 5 mm and 3.7 mm in diameter, for preparation of the bone cavity with the use of calibrated drills with incomplete preparation (i.e. the final drill was inserted coronally by 2 mm in order to enhance primary stability). The aim was to gain primary stability in a deeper and wider reach of the bone (corpus mandibulae) using a conventional method, whilst the coronal implant part (processus alveolaris) would lean against the bone widened by alveoplasty. The criteria for the bone length in mm were set based on our own experience and literature sources. The necessary lengths for different preparation types will certainly be a subject of further research. In our study of primary loading following alveoplasty, solely Impladent implants (Lasak, Czech Republic) were used. Twenty patients with a total number of 58 implants were treated; 29 alveoplasties were performed on one side of stereodefected jaws, employing the Bone Management System without raising the mucoperiosteal flap. Only patients with a suitable alveolar bone, i.e. with a minimum height of 10 mm and width of at least 4 mm on the side were included in the study. In experiments conducted outside this presented study, we were successful even with a bone width of only 2 mm. The Horizontal-Control System enables, after the preparatory drilling, a series of conical angle modulators with increasing diameter to be subsequently used to widen the alveolar ridge to obtain a suitable diameter and shape. In our case, we always used modulators of a diameter that was less than the implant diameter, i.e. in the case of a 3.7 mm implant we created a cavity of 3.1 mm in diameter, and in the case of a 5 mm implant a cavity of up to
Dental implants

4 mm in diameter. The final alveolar shape was created by bone spreading when the self-tapping screw was being inserted. Implants were inserted in the above-described procedure and immediately loaded. If all implants on one side of the jaw could be loaded, we preferred to use a fixed bridge made from resin and placed on temporary abutments from the manufacturer. If only one implant at the side was available, we used a temporary, screw-retained prosthesis fixed on a ball attachment from the manufacturer. The requirement of a torque of at least 35 Ncm was achieved for all implants in the study.

On the opposite side, which was used as a control, implants were inserted either by a two-phase bone augmentation procedure, where necessary, or by using a minimally-invasive procedure without immediate loading. The objective and subjective results were then evaluated. As for the objective results, all 30 implants on the research and control side were successfully loaded. On the research side an average loss in bone height of 0.3 mm ± 0.8 mm was measured whilst on the control side, where either augmentation or a minimally invasive-procedure without immediate loading was performed, a bone gain of 1.1 mm ± 1.8 mm was measured. Thus, augmentation proved to bring a bone gain in many cases; the results, however, are highly variable.

The disadvantage of the pilot study was that it was not homogeneous. Nevertheless, it could be said that a minimally-invasive procedure implemented along with alveoplasty and subsequent immediate loading does not cause a higher risk of implant loss. Its big advantage is its much shorter treatment time. As far as evaluation of the treatment by patients goes: all patients preferred to have minimally-invasive treatment without alveoplasty but such a treatment could only have been offered to four patients (with sufficient alveolar bone width) out of twenty involved in the study. Minimally-invasive treatment with alveoplasty (although some patients initially feared the widening of the implant cavity) was the preferred choice over a two-phase implant insertion procedure with bone augmentation. As reported in the literature, the latter was found to be more painful - causing swelling - and had a longer healing time.

To conclude, the Bone Management System of Meisinger and Implantdent implants (Lasak, Czech Republic) can be successfully used in flapless alveoplasty - even for cases that would have been contraindicated in the past. In addition, in cases of sufficient bone availability (height of 10 mm and width of 4 mm) and satisfactory primary stability, the immediate loading of implants can be recommended.

**Literature**

11. Chen ST, Wilson TG Jr, Hammerer CH. Immediate or early placement of implants following tooth extraction: review of biologic basis, clinical procedures, and outcomes.
Reconstruction of cleft palate using implants – Case Report

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Objective: Early prosthodontic therapy (usually at around 18 years of age) often leads to early loss of teeth and in extreme cases to complete loss of dentition at between 40 and 50 years of age.

Patients: This report describes the treatment of two middle ages cleft patients. Edentulous maxilla with cleft defect was treated with screw-retained prosthesis supported by 6 implants (STI-BIO-C, IMPLADENT, LASAK, Ltd., Czech Republic).

Results: Treatment of the whole dental arch on the basis of implants is currently frequently used as it provides a possibility of thorough functional and aesthetic therapy to a patient. The biomechanics of the reconstruction enables individual adjustment of the shape of the dental arch.

Conclusions: The problem in cleft patients involves altered relations in the dental arch caused by the defect alone or also by affecting of the growth of the maxillary segment by surgery. A potentially removable framework is therefore the main method of choice because the position of the implants must be prosthetically modified.

Is lateral sinus lift an effective and safe technique?
Contemplations after the performance of one thousand surgeries

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Introduction

The authors assess the importance of lateral sinus lift in current implantology. They prefer a variant of the surgical protocol that minimizes the surgical and financial burdens for the patient. In addition they provide comments on contraindications and describe the most frequently occurring complications. After comparing it with alternative procedures, they concluded that Lateral Sinus Lift, despite having some disadvantages, is the most effective method for implantation into dorsal parts of the maxilla.

Dental implantology at the beginning of the Third Millennium can replace tooth defects almost always if they occur in an adult individual who is willing to cooperate and to provide a financial contribution thereto. Contraindications for implants are increasingly being reduced. Thanks to the augmentation procedure, we know of virtually no situations that the implant could not be implanted due to insufficient quantity of the alveolar bone. Lateral Sinus Lift is one of the most widely used augmentation procedures. It enables to make an implant in the dorsal parts of the maxilla, where the bone often has poor quality and is reduced by the extended maxillary sinus. When considering that the minimum safe length of the implant is 10 mm, the bone at the site of the first premolar is very low in one-fourth (25%) of patients. The bone is insufficient in more than half of patients at the level of the second premolar, and in 80 to 90% of patients at the level of molars.
Lateral Sinus Lift is usually carried out under general or local anesthesia, or under analgesia. After lifting the mucoperiosteum from the front wall of the maxilla, we should first use a ball drill to create a window in the thin bone demarcating the maxillary sinus (Fig. 1). Antral mucosa must remain undamaged. Then we should lift the mucosa away from the bone using a special raspatorium to the extent of the alveolar recession and dislocate the same cranially (Fig. 2). A space will be created at the base of the maxillary sinus (Fig. 3) that will be filled with an appropriate augmentation material.2 When an inadvertent perforation of the antral mucosa occurs during the preparation, such defect is most often covered by a resorbable barrier membrane, or sometimes collagen tape, a plate from autologous or lyophilized bone, or it can be closed by fibrin glue or by a fine suture.1,4 The scientific literature typically says that when the residual alveolar bone is at least 3 to 5 mm high, dental implants should be introduced during the Sinus Lift, as this would ensure their sufficient primary stability (single-step surgery) (Fig. 4).2 When the bone is lower, the implant should not be inserted before partial consolidation of the augmentation material (two-step surgery) (Fig. 5, 6).2 The time of healing depends on the augmentation material used. If non-autologous, implants in the single-step Sinus Lift should not be exposed to a functional load until 9 months. When using the two-step variant, the implants should be applied after 6 to 9 months and exposed to a functional load after another 9 to 6 months. Use of autologous bone can reduce the length of treatment to one third.2

The objective of the work is to present our own experience with sinus lift surgery, describe its risks, most frequent complications and to evaluate the effectiveness of the procedure. This contemplation is based on the evaluation of results from one thousand lateral sinus lift surgeries that were performed from October 1995 until April 2007 at the authors’ facility. A total of 2056 Implant implants (Lasax, Ltd., Czech Republic)2 and 232 Replace Select Tapered implants (Nobel Biocare, Sweden) were inserted into the augmented boned.

### Surgical Protocol

Lateral Sinus Lift can be performed in many variants, characterized for example by type of anesthesia, method of bone preparation, selection of augmentation material, number of surgery phases, surface of implants or length of healing period.2 These factors have an influence on the rate of treatment, surgical load for patients, length of convalescence period, frequency of complications, and price of procedure. We consider the following surgical protocol to be optimal in terms of usual clinical practice. The protocol is adjusted to minimize the invasiveness of the procedure, risk of complications and financial burden for the patients. The said surgical protocol is characterized by the following parameters:

<table>
<thead>
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<th>Regimen:</th>
<th>outpatient</th>
</tr>
</thead>
<tbody>
<tr>
<td>Anesthesia:</td>
<td>local</td>
</tr>
<tr>
<td>Bone preparation:</td>
<td>bone cutter, diamond ball cutter</td>
</tr>
<tr>
<td>Augmentation material:</td>
<td>non-autologous</td>
</tr>
<tr>
<td>Type of surgery:</td>
<td>single step</td>
</tr>
<tr>
<td>Dental implants:</td>
<td>hydroxyapatite coated</td>
</tr>
<tr>
<td>Bone window coverage:</td>
<td>without barrier membrane</td>
</tr>
<tr>
<td>Healing period:</td>
<td>6 to 9 months</td>
</tr>
</tbody>
</table>

Simplifying the procedure is the priority. Therefore, local anesthesia is fully sufficient when the surgery procedure is performed quickly. The use of a high-performance bone cutter with a diameter of 3 mm (40000 revolutions per minute), which should be replaced at the end with a gentle, albeit less effective diamond ball of the identical diameter, reduces the time needed for the creation of the bone window to 1 to 3 minutes (Fig. 7, 8). The surgery as a whole including the implantation takes 30 to 40 minutes and is very well tolerated by patients. An unquestionable advantage is if the surgery is performed by a maxillofacial surgeon, as the procedure falls under the discipline of facial surgery rather than dental medicine. Use of non-autologous augmentation material only substantially reduces the operative burden for the patient, although sometimes at the expense of longer in-growth period. Deproteinized bovine bone or beta-tricalcium phosphate are preferred materials.6 Addition of a small quantity of autologous bone (such as from tuber maxillae) has no positive influence on the augmentation effect.1,4 According to the authors, limited indications of one-step Sinus Lift for alveoli higher than 3 to 5 mm is not justified, as it is contradictory to the commonly published opinions of experts.2 Publications that prove this statement are currently under preparation.

One-step variant of the Sinus Lift is more beneficial in all aspects (Fig. 9). The standard healing period is 9 months and can be reduced to 6 months depending on the height of the alveolus. According to the experience of the authors, the rate of non-osseointegration for hydroxyapatite-coated implants is significantly lower compared to the implants with textured titanium surfaces.1,2 Use of a barrier membrane to cover the bone window is unnecessary. It is beneficial to apply medicines against the postoperative

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**Fig. 7:** A larger part of the preparation can be performed by a high-performance bone cutter.
Contraindications

Disorders and conditions that contraindicate the Sinus Lift have not been fully defined yet. We would like to add the following remarks on the generally known and recognized rules:

1. Purulent exudate in the maxillary sinus is the most frequently occurring contraindication of Sinus Lift. Empyema, whether asymptomatic or not, is an absolute, though temporary, contraindication.
2. S/P Caldwell-Luc operation usually makes the Sinus Lift highly difficult or impossible. Scar tissue cannot be treated as physiological mucosal lining.
3. If the patient reports a history of acute sinusitis and the cause thereof has not been eliminated, the augmentation may increase the propensity to further attacks of inflammation. The patient must be informed to this respect.
4. Chronic sinusitis does not complicate the Sinus Lift. On the contrary, hyperplastic antral mucosa is increased mechanical resistance, which facilitates the preparation.
5. Mild osteoporosis is not considered to be contraindication, while moderate forms of this disease require prolongation of the healing period up to 12 months. We never performed surgery in case of severe osteoporosis.
6. Concurrent treatment with anti-aggregation drugs causes no life-threatening bleeding. Nevertheless it is recommended to discontinue such treatment subject to an agreement with the treating physician. Dose reduction is required in case of concurrent treatment with anticoagulants (the borderline level is INR 1.8). If not realistic, the patient should be transferred to low-molecular heparin.
7. Inhalation or superficial application of corticoids has no influence on the effects of surgery, as the absorbed dose of the medication is low.
8. Age itself is not a contraindication.
9. Controlled diabetes mellitus is not considered to be a contraindication, independently of the type of treatment.
10. We don’t agree with the frequently occurring statement that heavy smokers have a thin mucous lining of the maxillary sinus, that is highly prone to perforation during the surgery.

Complications

The following list contains notes on the most significant complication of Sinus Lift. Serious complications are very rare, while the occurrence of the other complications corresponds to the character of the procedure and is acceptable for both the patient and the surgeon.

1. By far the most frequently occurring complication is perforation of mucosa of the maxillary sinus during the surgery. If not closed spontaneously, we should use oxycellulose mesh (Surgicel, Johnson & Johnson) for coverage. This original procedure is fast, cheap and reliable, and was repeatedly published by the authors (Figures 10, 11). In emergency, the mesh can be used to reconstruct the entire ceiling of the augmented space.
2. Acute sinusitis is the most serious complication. It is most frequently caused by infection of the augmentation material during the surgery. It has dramatic manifestations and requires revision surgery of the maxillary sinus under general anesthesia with removal of all extraneous objects. It is a quite rarely occurring episode, that had the occurrence of 0.1% in the presented group of patients.
3. Mild purulent exudate from a dehiscent mucosal wound accompanied by swelling, pain and subfebrile conditions, is not a big threat. It can be usually managed by irrigations and antibiotic therapy.
4. From time to time, we observe second intention healing, which represents no big risk for the effectiveness of the procedure. If the bone window is situated too close to the mucosal incision, or if the augmentation material is too much compressed, the augmentation material can be liberated from the wound. In this case, it is recommended to use antibiotic treatment and try to apply a secondary suture.
5. Postoperative hematoma is observed mostly in older females (Fig. 12). It has annoying effects in esthetic terms but usually re-absorbs within two weeks.
6. Primary failure (non-osseointegration) of the implant remains a very rare event in hydroxyapatite-coated fixtures (1.4% in our group vs. 0.6% in implants into non-augmented bone). Long-term success is not significantly different from that of usual implantations.

Evaluation of the Method

Sinus Lift is not an ideal technique in terms of advanced implantology that is characterized by efforts for easier and more rapid treatment. Relatively high invasiveness compared to plain dental implants, need for great erudition of the surgeon, impossibility to correct vertical atrophy of the alveolar crest in the oral direction, financial burden, and especially prolonged healing period, are obvious handicaps.
However, alternative methods intended for implantation into dorsal parts of the maxilla also have some disadvantages. A limited effectiveness of the closed Sinus Lift, significantly lower reliability of tuberol or pterygoidal implants and surgical burden of zygoma implants are also significant shortcomings. Onlay augmentation by an extraoral bone graft is barely applicable common practice as it is difficult and invasive. A controlled bone regeneration that elevates the alveolar process is accompanied by a disproportionately high number of complications and failures. Special short, large diameter implants cannot be always used and their long-term success rate has not been sufficiently confirmed. Management of a shortened dental arch with long distal dens pendens is beyond the lege artis procedure. Sinus Lift is the only technique that enables the use of sufficiently long and optimally localized implants. Therefore we prefer Sinus Lift over any other techniques.

Conclusion

Lateral sinus lift, despite having some disadvantages, such as in particular high demands on both surgeon and the patient and longer healing period, is in most cases the best available solution for insufficient quantity of the alveolar bone during the implantation into the dorsal parts of the maxilla. Its role in current dental implantology is still non-replaceable. The invasiveness of the procedure can be substantially reduced when performed by an experienced surgeon using the presented surgical protocol. The risk of complications remains low.

Literature

Teeth in six hours

A. Šimůnek, T. Vosáhlo, D. Kopecká, T. Brázda, M. Sobotka, D. Dušková

Implantologie Journal 8/2006

1 Centre of dental implantology at the Clinic of Stomatology, Teaching Hospital and Faculty of Medicine, Charles University, Hradec Králové, Czech Republic
2 Private Dental Clinic, Hradec Králové, Czech Republic

The authors describe their own modification to a fixed mandibular prosthesis using Impladent implants. A temporary fixed bridge is made from the lower total prosthesis and fixed into the mouth using impression abutment coping. Patients are able to eat immediately. After six weeks the provisional bridge is replaced by the classical Brånemark bridge. Over a 30-month period, 92 patients were treated in this way - not a single implant was lost, giving a success rate of 100% for this supraconstruction.

Introduction

The restoration of a mandibular occlusion using a conventional prosthesis is rarely successful. When the alveolar ridge is highly atrophied it becomes extremely difficult to fit a denture. In such cases implant surgery can offer an effective solution. Two options are available: a hybrid prosthesis or a fixed bridge. In the first case, the prosthesis is connected to two or four implants via attachments; ball, bar-clip and magnet attachments are the most commonly used. The limited long-term success of these implants, the frequent technical problems with attachments, and the frequent occurrence of other complications, has led us to abandon this type of treatment. A fixed bridge offers the patient higher comfort, has a better performance for chewing and, in comparison with a hybrid prosthesis, is far more reliable. Its disadvantages are the higher cost, the extensive surgery and more demanding hygienic care required. Thus the need to develop a new procedure for the replacement of a mandibular occlusion that would be spared some or all of the above constraints and satisfy the following requirements:
1. Acceptable price
2. Suitable for a highly-atrophied alveolar ridge
3. Short treatment time
4. Easy hygiene care
5. Minimally-invasive treatment
6. Long-term success

Most of the above is offered by the bridge first presented by P-I. Brånemark in 1965 and commonly known as the Brånemark bridge. Traditionally, this bridge was fixed to implants that were already healed in, thus not complying with the above requirement for a short treatment time. However, there are published cases where immediate loading had been extremely difficult to fit a denture. In such cases implant surgery can offer an effective solution. Two options are available: a hybrid prosthesis or a fixed bridge. In the first case, the prosthesis is connected to two or four implants via attachments; ball, bar-clip and magnet attachments are the most commonly used. The limited long-term success of these implants, the frequent technical problems with attachments, and the frequent occurrence of other complications, has led us to abandon this type of treatment. A fixed bridge offers the patient higher comfort, has a better performance for chewing and, in comparison with a hybrid prosthesis, is far more reliable. Its disadvantages are the higher cost, the extensive surgery and more demanding hygienic care required. Thus the need to develop a new procedure for the replacement of a mandibular occlusion that would be spared some or all of the above constraints and satisfy the following requirements:
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Most of the above is offered by the bridge first presented by P-I. Brånemark in 1965 and commonly known as the Brånemark bridge. Traditionally, this bridge was fixed to implants that were already healed in, thus not complying with the above requirement for a short treatment time. However, there are published cases where immediate loading had been performed - but unfortunately then the cost of the treatment went up. In March 2004, the Implantology Department of the Dental Clinic in Hradec Králové (T. Vosáhlo) conceived the idea and developed a procedure that satisfied all the above stated criteria. The preparation of the provisional bridge followed that of the earlier published methodology but this time included the use of the IMPLADENT implant system, allowing a reduction in the number of necessary components used. This approach proved to be decisive in satisfying the requirement of acceptable cost. This concept has been commercially presented as ‘Teeth in six hours’. The aim of this paper is to provide a description of the treatment protocol and an evaluation some two and half years after its first use.

Basic characteristics of the concept ‘Teeth in six hours’

The concept ‘Teeth in six hours’ works with the Impladent implant system (Lasak Ltd.). It is based on inserting five Impladent STI-Bio-C implants (implants having a bioactive surface – Bio surface) with high primary stability in the interforaminal area. Immediately the abutments for the screw-retained prosthesis are then attached. Using standard titanium impression copings that ensures not only a high mechanical strength but also the necessary passive accuracy, the provisional bridge is made from an existing total screw-retained prosthesis. Both the surgical and prosthetic phase need take no more than six hours. The provisional bridge allows for food to be taken immediately. In patients with a partial denture, the remaining teeth are extracted directly prior to implant surgery and the supraconstruction made by completion of the partial prosthesis. After implant osseointegration, the provisional bridge is replaced by a classical Brånemark bridge. However its appearance is a unique and specific one, unlike any found among conventional prosthetics. It consists of a metal framework with long cantilevers leading to the first molar region. The metal framework is covered by pink resin and holds twelve prefabricated resin teeth.

Step-by-step procedure

1. Application of antibiotics, local anaesthesia.
2. If necessary, extraction of existing mandibular teeth, careful excochlea-
tion of walls and bottom of the extraction site.
3. Raising of mucoperiost of the alveolar ridge to the extent of 35 – 45.
4. Detection of both foramina mentalia.
5. Smoothening of alveolar ridge.
6. Localisation of suitable sites for implant placement; maintaining a regular interforaminal distance so that the gap between the end implant and foramen mentale is at least 3 mm.
7. Preparation of five bone beds for fixing Impladent STI-Bio-C implants 3.7 mm in diameter and usually 16 mm in length (the implant length depends on the bone but the minimum length should be 10 mm).
8. Implant insertion ensuring high primary stability (torque ≥ 45 Ncm, ISO ≥ 60) (Fig. 1).
9. Attachment of abutments for screw-retained supraconstruction (usually 4 mm long, depending on the width of soft tissue), torque 35 Ncm.
10. Suture using resorbable material (Fig. 2).
11. Fixing of impression copings equipped with blocks of self-curing acrylic resin (PMMA).
12. Impression copings equipped with blocks of self-curing acrylic resin (PMMA) are connected together using the same resin to form a rigid block (Fig. 3).
13. An impression is taken using A-silicon impression material and titanium impression copings; the recommended method is an open-tray impression technique but instead of an impression tray a modified lower dental prosthesis is used (Fig. 4).
14. A plaster working model with implant replicas is made.
15. The working model prosthesis is extended using self-curing acrylic resin; impression copings become part of the prosthesis (Fig. 5).
16. Transformation of the prosthesis into a fixed bridge by reducing it to 35 – 45 and by reducing the saddle.
17. Final preparation of the bridge (Fig. 6).
18. Fixing of the provisional bridge to the implants using screws (15 Ncm) (Fig. 7).
Fig. 1: Five Impladent STI-Bio-C implants inserted in the interforaminal area

Fig. 2: The final surgical phase, abutments are ready for taking the impression

Fig. 3: Impression coping fixed with fastening screw and kept in blocks by self-polymerising resin

Fig. 4: Impression taken. A modified lower dental prosthesis is used instead of impression tray; impression copings become part of the impression

Fig. 5: Working model with impression coping and prepared prosthesis

Fig. 6: Temporary bridge made of prosthesis and impression coping ensuring passive fit

Fig. 7: Temporary bridge fixed after six hours since the start of operation

Fig. 8: Brånemark bridge
Results

The concept ‘Teeth in six hours’ was used by the authors in 92 patients between March 2004 and August 2006, 57 of which were male and 35 female. The average age of the patients was 62 years (range from 27 to 79 years of age). Contra-indications that might have an impact on the treatment success rate included diabetes mellitus in 13 patients (in four cases compensated by diet, in six cases by per oral application of antidiabetics, and in three cases by the application of insulin), and one patient suffering from kidney polycystos that in a few weeks after treatment resulted in kidney failure. In a twenty-seven-year-old patient teeth loss was caused by ectodermal dysplasia. Altogether 460 Impladent implants (Lasak Ltd., Prague) with bioactive modified implants and 50 were of a newer type STI-Bio. The implant dimensions are given in Table 1. The height of the gingival part of abutments for screw-retained supraconstruction is given in Table 2.

<table>
<thead>
<tr>
<th>Diameter/length [mm]</th>
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<th>16</th>
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Table 1: Implant dimensions

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<td>0</td>
<td>1</td>
<td>445</td>
<td>12</td>
</tr>
</tbody>
</table>

Table 2: Height of the gingival part

All surgeries were performed in out-patient care, in a dental chair, under local anaesthesia using Ubistesin F, and a prophylactic course of doxycyclin (Deoxyymyokin tbl. 1x100 mg p.o.), or, in case of allergy of doxycyclin, clindamycin (Dalacin C cps. 3x300 mg p.o.). The application of antibiotics started one day before surgery and finished seven days after. The implants were surveyed over a period of 0 to 30 months, on average over 12.5 months: all implants and superconstructions were functional, i.e. survival rate was 100%.

Discussion

The cost for the patient has been reduced by limiting the use of prosthetic components and by making a provisional bridge from the patient’s existing total prosthesis and by manufacturing a Brånemark bridge with resin teeth as the final bridge10. The given data set, that includes all the implant treatments of the type performed at the authors’ workplace, shows an extremely high success rate. It is expected that such a high success rate has been favoured by the rather high number of implant treatments included in this study compared to other documented procedures.7, 8 A statistical evaluation of the data set was performed only at the mid-term period. Considering that we have more than ten years experience and excellent results with conventional Brånemark bridges, we expect that the use of the same bridge type and the concept ‘Teeth in six hours’ will have similarly good results in the long term. This hypothesis, of course, will have to be confirmed by a long-term study.

A typical characteristic of the implants used was their Bio surface. It is a chemically-treated titanium surface that speeds up and enhances the process of osseointegration compared to other surface types.12, 13, 14, 15 The latest type of implant, STI-Bio-C, differs from its predecessor STI-Bio in having a cervical micro-thread, new thread design and a more-pronounced narrowing apex. The concept of the Brånemark bridge that is characterised by its concentration of implants in the interforaminal area avoids the problem of the lack of alveolar bone which in the interforaminal area is almost always abundant and of high quality1. The Brånemark bridge is particularly destined for older patients, often having lowered motor skills and thus having difficulties to maintain good oral hygiene. The design of the Brånemark bridge makes oral hygiene easier.

The treatment period is reduced to a minimum. The method of immediate loading is normally used in the case of the remaining teeth, immediate reconstruction after teeth extraction and immediate implant loading being performed10. From the patient’s psychological point of view, it is also very important that the surgery brings results within a few hours rather than several months. In this way post-operative difficulties are easier to overcome too. The surgery is performed in an easily accessible area and is completed within 90 minutes of anaesthesia and with minimum trauma for the soft tissues. Neither sedation nor hospitalisation are necessary.

Conclusions

‘Teeth in six hours’ represents the immediate replacement of the mandibular arch, using the Brånemark bridge with all its advantages. Clinical experience has shown that its success rate is very high. In our opinion its cost-effectiveness is the highest amongst the whole of implantology. The treatment is extremely fast and simple and the costs relatively low. Teeth replacement that can be completed within one day has a positive psychological impact on the patient, making the method highly attractive and, in addition, also raising the image of dental surgery.

Literature

28 | Dental implants

Early loading (4 weeks) of dental implants Impladent in maxilla and mandible - monitoring of the healing process using resonance frequency analysis

A. Štěpánek, J. Strnad, Z. Strnad

Quintessenz, Vol. 14, No. 9, 2005

Introduction

There are certain conditions that enable the early, or even immediate, loading of implants as an alternative to the more ‘classical’ two-phase implantation process. The classical implantation process involves a healing period of no-direct-loading, lasting six months for implants inserted into the maxilla and three months for mandible implants, using a titanium screw implant with a machined surface. This ensures the needed immobility of the implant at the beginning of the healing period: necessary for the development of secondary stability of the implant, which results in the long-life of the fully-loaded implant. A shorter healing period, or its complete elimination, brings with it new demands on both the primary and secondary stability of the implant. Primary implant stability is mainly dependent on the mechanical characteristics of the bone (its local quality and quantity), the type of implant used (its geometry, diameter, length and surface) and the method of insertion. Secondary stability represents an enhancement of stability as a result of new bone formation and its ‘remodelling’ at the contact surface of bone/implant and within the implant’s surroundings. The use of a shorter healing period has to be compensated by an early and sufficiently fast increase in secondary stability that can withstand the expected demands of implant loading. Exceeding the limits of implant immobility might result in unwanted fibrous encapsulation of the implant and its subsequent failure.

In order to achieve faster healing, the macro-morphology of the titanium surfaces of implants may be modified by sand-blasting, or the micro-roughness modified using acid-etching or anodic oxidation in mineral acids. Currently, a new generation of implant surfaces is appearing (e.g. STI-Bio, Osseospeed), initiated through the development of some chemical modification of the surface in order to obtain a specific surface reactivity - called bioactivity. This bioactivity of the surface stimulates the formation of calcium phosphates on the implant’s surface immediately after implantation, i.e. at a time when the synthesis of bone minerals by osteogenic cells is not yet possible. Bioactive surfaces are characterized by their perfect hydrophilic character (wetting properties), high surface area and high levels of hydration. Implants STI-BIO with their bioactive surface initiate osteointegration faster and thus hasten the needed development of an implant’s secondary stability. The shorter healing period of Implant STI-BIO implants achieved through their bioactive surface (STI BIO) – six weeks for a mandibular implant and twelve weeks for the maxillar implant – has already been documented by a clinical study. The present study evaluates the possibility of a further reduction of the healing period to four weeks in both the maxilla and mandible, using Implant STI-BIO-surface implants. Resonance frequency analysis (RFA) was used to assess primary implant stability and the factors that influence it, and also to check the development of implant stability during the periods of healing and full-use.

The aim of the study was to assess the impact of a reduced (4-week long) healing period following the application of an implant - Implant STI BIO with a bioactive surface - in the maxilla and mandible with a statistical evaluation of the success rate. The statistical analysis included an evaluation of primary stability of implants and of the factors that influence primary stability, as well as concurrent monitoring of the time-dependent development of implant stability during the healing and fully-functional periods using resonance frequency analysis.

Material and methods

Between October and December 2004, altogether 90 implants of Implant STI BIO (LASAK, Praha, Czech Republic) were implanted. This involved 34 patients, 22 men and 12 women, aged between 26-71 years (the average age was 51.5 years). None of the patients were diagnosed with any medical contraindication for implantation. Out of the total number of 90, 53 (58.9%) implants were implanted in the maxilla and 37 (41.1%) in the mandible. Indications are summarised in Table 1. A two-stage procedure was used in all cases. One week after implantation healing cylinders were introduced, and four weeks (at the latest) after implantation the implants were loaded using provisional restoration fixed to temporary abutments (Fig. 1). Thirteen weeks after implantation, at the latest, the provisional restoration was replaced by the final supraconstruction (Table 2). Bone quality, classified as D1, D2, D3 or D4 according to Lekholm-Zarb, was assessed during the bone bed preparation using resistance as a subjective measure. The primary stability of implants was evaluated in two ways: by measuring the insertion torque and with the use of resonance frequency analysis (RFA) by Ostet (Integration Diagnostics AB, Göteborg, Sweden). Insertion torque was measured using a ratchet torque-control adapter (Lasak) at the final position of the implant in the bone bed (ITf). With resonance frequency analysis, the range of the dynamic resonance frequency of the whole complex - transducer/implant/bone - (3,500 Hz – 8,500 Hz) is divided into 100 ‘intervals’ and expressed as ISQ (Implant Stability Quotient) from 0 to 100. Implant stability values, ISQ(t), were always measured by RFA at weeks one, four and thirteen after implantation (time t being measured in weeks). Change in implant stability after t weeks of healing, dISQ(t), was expressed as the difference between the stability in a given week of healing ISQ(t) and primary implant stability (ISQp).

The success rate was assessed following Albrektsson. Implants considered as successful were every loaded implant that: was clinically not moving; did not cause any chronological discomfort (pain or other); showed no observation of repeated infections or radiolucent bone in the implant surroundings; and did not show progressive marginal bone loss. For the evaluation of success, life-table analysis was used. Data obtained on implant stability underwent statistical analysis (mean value and standard error). Differences of experimental data were tested using a t-test (Type 2), being statistically significant if p<0.05.

The aim of this paper is to provide a description of the treatment protocol and an evaluation some two and half years after its first use.
Fig. 1: Making temporary prosthesis (c,d) from acrylate on temporary abutments Impladent (a,b) and final prosthesis with a screw-retained metal-ceramic bridge (e,f). The implant in localization 33 showing a remarkable reduction in stability during the first week after implantation (ΔISQ(1) = -11) was not loaded by a temporary prosthesis (c,d) and was later explanted. When healed, reimplantation in the same localization was done and the implant was loaded with final prosthesis (e,f).

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**Table 1: Indication**

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**Table 3: Dimensions of implants used**

<table>
<thead>
<tr>
<th>Bone quality</th>
<th>Maxilla</th>
<th>Mandible</th>
</tr>
</thead>
<tbody>
<tr>
<td>D1</td>
<td>9</td>
<td>13</td>
</tr>
<tr>
<td>D2</td>
<td>41</td>
<td>36</td>
</tr>
<tr>
<td>D3</td>
<td>3</td>
<td>4</td>
</tr>
<tr>
<td>D4</td>
<td>0</td>
<td>0</td>
</tr>
</tbody>
</table>

**Table 4: Implant frequency per bone quality**
Results

Primary stability

Mean primary stability (ISQp) of all introduced implants was 60.2±7.7 (n=90). Altogether six parameters that could influence primary stability were considered: implant localization (maxilla and mandible), bone quality, diameter and length of implant, and the means of implant insertion. The primary stability of mandible implants was statistically somewhat higher 63.6±7.8 (n=37) in comparison to maxilla implants 57.8±6.6 (n=53), (p=0.0003). In the mandible, primary stability (ISQp) was higher for all teeth positions compared to those in the maxilla. For the position 3 (canine) and 4, 5 (premolars), however, the difference was not statistically significant (Fig. 2).

The data obtained on stability were further analyzed to assess the implant’s impact on the bone. With decreasing bone quality (D1>D2>D3), the decrease in primary stability of implants was statistically significant (Fig. 3). In turn, bone quality had an impact on the insertion torque, which at its final position may indicate the primary stability of the implant19. Mean values of final insertion torques of the implant groups, divided according to bone quality, became lower with lowering bone quality (Fig. 4) in a similar way to that of ISQp (Fig. 3). Analysing the data for individual implants (n=90) showed that the final insertion torque (IT) and primary stability (ISQp) were directly related: [IT] = 0.5 Ncm, [ISQp] = -6.3; though the correlation coefficient was low (R²=0.1892).

Another studied factor that has an influence on primary implant stability was the selected surgery procedure. Groups of implants with high (ISQp>60), medium (ISQp=50-60) and low (ISQ<50) primary stability were assessed. Within the group ISQp>60, a statistically-significant higher mean primary stability (ISQp) was found in implants where the bone bed had been prepared using a threadformer rather than the drill alone. Within the group with lower primary stability (ISQp=50-60 and ISQ<50), higher ISQp was observed in self-tapping implants but this was not statistically significant (Fig. 5).

Loaded implant stability 13 weeks after implantation

The mean value of stability of loaded implants 13 weeks after implantation (ISQ13) was 59.1±5.2 (n=88). For the mandible the value of ISQ13 reached 62.8±5.3 (n=36) and for the maxilla ISQ13 56.4±3.7 (n=52). Bone quality did not influence implant stability measured 13 weeks after implantation. Mean values of implant stability ISQ13 did not show any statistically significant differences between implant groups of varying bone quality (Fig. 6). In contrast, the impact of implant diameter was statistically significant. Implants of 5.0 mm in diameter showed higher stability ISQ13 61.6±6.3 (n=30) than implants of 3.7 mm in diameter (ISQ13 = 57.4±4.2; n=58).
Development of implant stability during the periods of healing and functional loading

Primary stability of implants had an essential impact on the development of secondary stability during the healing and loading periods. Figure 7 shows the development of implant stability for high (ISQp > 60), medium (ISQp = 50-60) and low (ISQp < 50) primary stability, the mean values being ISQp 66.3 ± 3.8 (n=45), 56.7 ± 2.0 (n=34) and 45.8 ± 3.9 (n=11), respectively; differences of mean ISQp values are statistically significant (p < 0.05). In the group ISQp > 60, implant stability initially declined but after the first week remained stable. The group of implants with medium ISQp did not show any statistically significant changes either during the healing period or within the thirteen weeks of functional loading. The group of implants with low primary stability showed an increasing enhancement of stability since the first week of the healing period and through the period of functional loading studied.

Discussion

A healing period reduced to four weeks offers less time for osseointegration of an implant prior to its loading, which increases the risk of micromovements of the implant in the bone bed that can result in fibrous membrane formation when loaded. In this case, a higher frequency of implant loss can be expected. The effect, therefore, of a reduced healing period was evaluated against a control group of implants STI BIO where a longer healing period (twelve weeks for maxilla and six weeks for mandible) was used and the results evaluated after twelve months of functional loading (Table 6).

The statistical evaluation of success rate of the experimental and control implant sets did not show any significant difference during the healing phase or the nine-week loading phase tested. We can say, therefore, that the reduced healing period does not increase the risk of implant loss compared to the original twelve-week healing period for the maxilla and six-week period for the mandible.

Resonance frequency analysis allows for an objective assessment of the implant stability with sufficient exactness and reproducibility of results (±0.5 ISQ units). The method, however, does cover an aggregate of implant stability, based on the dynamic resonance frequency of the whole complex: transducer/implant/bone, including the stiffness of their connections. The higher the frequency, the higher is the stiffness of the measured complex. When interpreting data, it has to be borne in mind that the value of ISQ combines a number of factors.

Amongst the six parameters assessed, the highest impact on implant stability was found to be the location in maxilla or mandible, bone quality, the method of implant insertion and implant diameter. The higher primary stability found for the mandible ensures implant immobility and a higher success rate when compared with the maxilla. The main difference can be attributed to bone density and to some extent also tooth location. The use of a threadformer to prepare the bone bed is characterized by its greater exactness and is less traumatic, enabling a higher implant stability to be achieved and is in accordance with the Impladent protocol. Some authors recommend the use of implants of larger diameter to deal

<table>
<thead>
<tr>
<th>Time period</th>
<th>Number of implants</th>
<th>Number of implant loss</th>
<th>Lost from evidence</th>
<th>Success rate (%)</th>
<th>Cumulative success rate (%)</th>
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</thead>
<tbody>
<tr>
<td>Healing period (0-4 weeks)</td>
<td>90</td>
<td>2</td>
<td>0</td>
<td>97.8</td>
<td>97.8</td>
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<tr>
<td>Functional loading (4-13 weeks)</td>
<td>88</td>
<td>0</td>
<td>0</td>
<td>100</td>
<td>97.8</td>
</tr>
</tbody>
</table>

Table 6: Success rate of STI BIO implants for experimental and control set

<table>
<thead>
<tr>
<th>Time period</th>
<th>Success rate per interval (%)</th>
<th>Statistical significance of difference between sets compared</th>
</tr>
</thead>
<tbody>
<tr>
<td>Healing phase</td>
<td>97.8 (n=90)</td>
<td>p=0.3998 NS</td>
</tr>
<tr>
<td>Mandible</td>
<td>97.3 (n=37)</td>
<td>p=0.2678 NS</td>
</tr>
<tr>
<td>Maxilla</td>
<td>98.1 (n=53)</td>
<td>p=0.8877 NS</td>
</tr>
<tr>
<td>Functional loading</td>
<td>100 (n=88)</td>
<td>p=0.5012 NS</td>
</tr>
</tbody>
</table>

NS = statistically unsignificant difference

Table 5: Life-table analysis
Dental implants

with compromised clinical situations, with the expectation of enhancing the implant’s primary stability. Our results, that show a significantly higher primary stability for implants of 5 mm in diameter compared to those of 3.7 mm, confirm the above clinical assumption.

The development of implant stability with time during the first week after implantation is characterised either by a decrease or increase in stability, depending on the primary implant stability (Fig. 7). During the first of week after implantation, the stability decreases in cases where the primary stability was high (ISQ>60), being characterized by high insertion torque and high frequencies of dense bone. In this case, osseointegration can be seen as compensation for low mechanical fixation caused by relaxation processes and biological changes accompanying the early healing stage. The resulting stability (measured by RFA) after the first, fourth and thirteenth week following implantation was not decreasing further as the decrease in primary stability was fully compensated by increasing secondary stability due to the fast interaction between the implant’s bioactive surface and the bone. This situation makes it suitable for early implant loading and is typical for bone of high density (Fig. 9). The rate at which a stable boundary between the implant and the bone bed is formed, besides the primary stability, is given by the osseosconductivity characteristics of the implant surface. When bio-inert surfaces (such as titanium with a machined surface) are used, secondary stability starts to grow later and thus, also, compensation of any marked decrease in primary stability occurs later. As an example can be given the loss of primary stability measured by a reverse torque method on machined surfaces of titanium implants when used in rats, where a decreasing reverse torque (reducing from 24 Ncm to 19 Ncm) was measured still four weeks following implantation and clinical measurements using RFA which indicated decreasing implant stability for twenty weeks with machined titanium surfaces and for eight weeks with the oxidized surface TiUnite. A sand-blasted and acid-treated implant surface (SLA) tested on sheep using the reverse-torque method showed a decrease in stability from 100 Ncm to 88 Ncm after two weeks of implantation and statistical tests on RFA data revealed decreasing stability up until the sixth week for implants of primary stability higher than 60 ISQ.

Implants of low primary stability, ISQp<50, showed increasing stability from the first week of the healing period (Fig. 7). These implants have a lower insertion torque (Fig. 8) and higher frequency of soft, spongy bone.

A less traumatic preparation of the bone bed (in spongy bone with abundant blood supply), along with the bioactive properties of the STI BIO implant surface, are the main stimulators of a remarkable increase in secondary stability during the healing period of unloaded implants. In this case, the increased rate in implant stability is considered to be a critical parameter which decides when the implant can be loaded. This is especially true for bones of a lower quality. In bones of D3 quality, already after four weeks implants reached 98.7% of their final stability as measured thirteen weeks following implantation (Fig. 9).

The value of primary stability of the implant, assessed by resonance frequency analysis does not unambiguously predetermine implantation success. The study presented did not find a difference in primary stability between implants that were lost during the healing phase and those that were successful and clinically stable when fully loaded thirteen weeks after implantation. However, a marked decrease in stability often indicates unsuccessful implantation and calls for a specific treatment procedure (Fig. 10).

Conclusions

The statistical evaluation of data shows that, for Impladent STI BIO implants, a healing period reduced to four weeks for both maxilla and mandible does not reduce the rate of the implants’ healing success, nor does it increase the frequency of lost implants during the nine week period of full loading. The reduced healing period thus does not pose any enhanced risk on implantation and maintains the prediction of a similar success rate. Primary implant stability is influenced by the implant localization, bone quality, selected insertion method, and implant diameter. Higher values of primary stability were found for mandibular implants ISQp 63.6±7.8 rather than maxillary implants ISQp 57.8±6.6. Bone quality mainly influences the primary stability of implants. ISQp within the higher D1 bone quality (65.3±8.8) was reduced by 10% within the group with D2 bone quality (59.1±6.2), and by 20% within the D3 bone quality group (52.4±8.8). Preparation of the bone bed in dense bone with the use of a threadformer, in accordance with the protocol for the Impladent system, enabled a higher primary stability to be achieved. The implant diameter was found to have a significant impact on the implant’s primary stability: implants with a diameter of 5.0 mm showed a higher primary stability 62.4±7.8 compared to implants with 3.7 mm in diameter, for which the primary stability was 59.0±7.4. The differences in stability for various implant lengths were not statistically significant.
The level of primary stability of the implant, assessed by resonance frequency analysis, does not determine unambiguously implantation success. It does, however, have an impact on stability development during the healing period and full loading.

**Acknowledgement**

The study was funded by the grant MPO ČR FT-TA/087

**Literature**

12. Strnad J. Kinetika tvorby hydroxyapatitu na anorganických materiálech, Disertační práce, VŠCHT, Praha .2004
Reduced healing time of Impladent Implants with bioactive surface

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Quintessenz, No. 3, Volume 13, 2004

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2 Lasak Ltd., Papírenská 25, Prague, Czech Republic

Dental implantology is a popular and generally recognised part of stomatology. There is no doubt about the long-term functioning and aesthetic benefits of implants. The increasing requirements of medical doctors and their patients are leading to faster and simpler treatment using implants. The greatest efforts are being directed towards attempts to shorten or completely eliminate the healing time of the implant.

Introduction

The length of the healing period, during which the implant is not subjected to loading, has been determined for a number of decades by the Bråne-mark treatment protocol14,18. In the upper jaw, the implant is left nonfunctional for six months and, in the lower jaw, usually for three months1,28. This unpleasantly long period of time is no longer acceptable in contemporary implantology and is being shortened. If the time reserved for healing the implant is reduced by at least one half, this is termed early loading. If this time is shortened to less than 48 hours, this corresponds to immediate loading.

The difference between early and immediate loading is not only quantitative, but also qualitative. For early loading, the special surface of the implant accelerates the formation of osteointegration, i.e. secondary stability, so that the implant heals sooner1,2. The implantologist is usually able to verify the healing (tapping, torque wrench, Periotest, resonance frequency analysis, etc.) and only then functionally load the implant13. Immediate loading requires not only the same surface quality, but also high primary stability of the implant1,4. It ensures sufficient resistance to loading until primary stability is replaced by secondary stability8,30. Thus, the incompletely healed implant is brought into function. Immediate loading is a promising method; however, not all its disadvantages have been evaluated at the present time.

Both therapeutic approaches are connected with a special implant surface accelerating the occurrence of osteointegration. Various modifications of highly structured surfaces have been developed by practically all leading international manufacturers. The significance of these changes is of key importance in the field and greatly affects the nature of dental implantology. The sur-
face of the implant is modified macroscopically, i.e. on a scale of hundreds of micrometres and, for some surfaces, also microscopically. Macro-roughness is most frequently created by sand-blasting with an abrasive medium (e.g. the TiOblast surface of the Astra Tech company) or plasma spraying of titanium powder (TPS from the Straumann company). This increases the surface of the implant and bone trabecules can grow into it in later phases of the healing. Micro-roughness is created, for example, by anodic oxidation-creating a porous titanium oxide layer on the surface of the implant (TiUnite from the Nobel Biocare company). In other cases, it is machined, the smooth titanium being etched with mineral acids (e.g. Osseotite from the 3i company). In some cases titanium is etched in a similar manner prior to sand-blasted acid etching (SLA from the Straumann company).

Implant implants (Lasak s.r.o.) have also been given a new surface, called Bio because of its bioactive properties. This was developed in 1999 and a year later was used for titanium screw implants STI-Bio with a diameter of 3.7 mm and later 5.0 mm.

The production of the Bio surface occurs in three phases. The titanium is first sand-blasted to obtain macroroughness; then it is etched with a mineral acid to obtain micro-roughness (Fig. 1). The third phase is most important; here, the implant is exposed to an alkaline medium. A submicroscopic gradient times, and simultaneously increases its hydration by an order of magnitude. This change in the treatment protocol was manifested in the success rate of the treatment.

Material and methods

All STI-Bio implants between March 2002 and December 2003 at the Clinic of Stomatology, Hradec Králové were introduced into natural, i.e. not augmented bone, where the healing period was reduced to one half compared to the Brånemark protocol, were included in a retrospective study.

None of the patients in the study suffered from a disease that would be an absolute „contra-indication“ for implantation, underwent radiation or cytostatic therapy, or were treated with corticoids or anticoagulants. Eight patients had compensated diabetes mellitus, of which four were treated only by modified diet, three by peroral antibiotics and one by insulin.

The implantation was carried out by the two-phase technique, a maximum of twelve weeks in the maxilla and six weeks in the mandible expired between the first and second surgical phases of the implantation. With the exception of the healing time, the operation protocol specified by the manufacturer was followed. In the second surgical phase, all the clinically stable implants were tested by a 35 N.cm torque wrench acting in the clockwise direction. If they resisted the torque, they were evaluated as successfully osteointegrated.

Production of a superstructure was commenced two weeks after the second surgical phase. After this the patients were instructed by a dental hygienist and included in the follow-up care program. They were invited for controls after three months and then after a further nine months (Fig. 5). Following one-year loading, those implants which were clinically stable, did not cause chronic subjective discomfort, were not the cause of repeated peri-implant infections, did not exhibit progressive loss of marginal bone and tissue, and whose surroundings were not radiolucent, were considered to be successful.

An x-ray was taken immediately after affixing the abutment and then one year later, using a Planmeca ProMax digital orthopantomograph or Gendex Visualix radiovisigraphic instrument with XCP-DS grooves ensuring parallel projection, i.e. pathway of the central beam perpendicular to the plane of the implant and the recorder. Resorption of the marginal bone was read off an enlarged radiovisigraph or orthopantomogram with a precision of 0.5 mm, on the mesial and distal sides of the implant.

The statistical treatment included implants that completed the healing period by January 2004. Life-table analysis, the log-rank test and two-choice t-test were employed to evaluate the success of the implantation and resorption of the marginal bone.

Results

During the monitored period a total of 1092 implants with Bio surface, meeting the above criteria, were used with 420 patients; 202 men and 218 women between the ages 15 and 76 years (average 45.2 years). Of these, 1013 implants, 431 in the maxilla and 582 in the mandible completed the healing period (Tab. 1).

<table>
<thead>
<tr>
<th>Diameter/length (mm)</th>
<th>8</th>
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<th>12</th>
<th>14</th>
<th>16</th>
<th>18</th>
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<tr>
<td>3.7</td>
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<td>86</td>
<td>181</td>
<td>235</td>
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<tr>
<td>5.0</td>
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<td>20</td>
<td>77</td>
<td>47</td>
<td>80</td>
<td>0</td>
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</tbody>
</table>

Table 1: Dimensions of the implants employed

Osteointegration did not occur for twelve implants, of which seven were in the maxilla and five in the mandible. Thus 98.8 % success was achieved in the healing phase (98.4 % in the maxilla and 99.1 % in the mandible, p>0.05). A superstructure was fitted on all the healed implants. 176 implants bore a single crown, 663 implants bore a fixed bridge and 162 implants bore a hybrid replacement.
During the first year of functional loading, one implant was explanted because of loss of stability. At the end of the first year, 770 implants were examined; the patients with remaining implants did not accept the follow-up care. 767 implants met the criterion of success. The interval success rate for the first year of loading equaled 99.5%. The cumulative success rate for the entire monitored period was 98.3%. These data, treated in the form of life-table analysis, are summarised in Table 2. Resorption of the marginal bone (±5D) at the end of the healing period equaled 1.1 ± 0.4 mm and, after the first year of loading, increased by 0.7 ± 0.4 mm.

Discussion

The minimal invasiveness of the operation, the almost hundred-percent probability of osteointegration, a high success rate of the implants in the long term, and almost perfect aesthetic effect of the superstructures are all attributes which have received maximum attention from implantologists. Reducing the interval between introduction of the implant and bringing it into use, or between removing a tooth and introducing an implant into use is a requirement that cannot be met without revision of the Brånemark protocol. The development of a new generation of titanium implant surfaces is a key precondition for this step to be revealed by evaluation of the success rate for the healing phase, supported by the short-term statistical success rate of the functional phase. It is not probable that early loading of implants would have negative manifestations only in the long-term. The probability of the formation of osteointegration of 97.6-98.8% was found from formerly published values of sand-blasted and hydroxyapatite-coated Impladent implants with unreduced healing periods. The success of the implantation after one year was from 96.8 to 98.3%. The results of the described study (98.8 and 98.3%) correspond with these results. The loss of marginal bone for sand-blasted and hydroxyapatite-coated Impladent implants measured at the end of the healing period and after the first year of functional loading have already been published and equaled 1.80 and 0.15 mm, resp. The values found here (1.10 mm and 0.70 mm) do not indicate faster resorption for implants with Bio surfaces.

Conclusions

It follows from the above statistical findings that STI-Bio Impladent implants can be loaded following twelve months of healing in the upper jaw and after six weeks in the lower jaw without risk of the implant not healing, acceleration of resorption of the marginal bone, or an increase in the frequency of failure of the implants during the first year of functional loading. In this sense, we recommend changing the treatment protocol laid down by the manufacturer of the implant system. It is probable that even in shortening the healing time to one half, the potential of the Bio surface is not exhausted. It is not clear though how far the healing time can be shortened. It follows from clinical experience that a six-week interval for implants in the mandible is usually acceptable for patients, while twice this time for the maxilla is not satisfactory. Further studies will be carried out to evaluate the potential for shortening the healing time to six weeks in both jaws and to establish criteria for immediate loading. This study was supported by grant of IGA MH CR 77113-03.

<table>
<thead>
<tr>
<th>Time interval</th>
<th>Number of implants</th>
<th>Number of failures</th>
<th>Lost from records</th>
<th>Interval success rate (%)</th>
<th>Cumulative success rate (%)</th>
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<td>healing period</td>
<td>1013</td>
<td>12</td>
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<td>98.8</td>
<td>98.8</td>
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<td>first year of loading</td>
<td>1001</td>
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<td>230</td>
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<td>98.3</td>
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</table>

Table 2: Life-table analysis

Literature

Peri-implantitis, problems and solutions – a 2 year study

Z. Novák

Quintessenz, Vol.13, No.6, 2004

Peri-implantitis is defined as a progressive loss of bone near an implant accompanied by inflammation of the soft tissue. As is apparent from long-term monitoring, peri-implantitis is considered to be the main reason for implant failure and, if not promptly treated, it will result in implant loss. Several possibilities of halting peri-implantitis or even to restore and return the anatomy of supportive tissues to a normal state are discussed. The need for prevention is stressed and the results of short-term observations on the stability of the bone level around two-phase implants are presented.

Peri-implantitis is an eminent complication in implant therapy. It greatly affects its success; in the case of bone loss the ratio between the extra- and intra-alveolar parts of the implant changes, thus reducing the occlusal force which the implant is able to transfer on the bone in the frame of physiological load. The therapy becomes more complicated and modification of the treatment procedure used for periodontal therapy of natural teeth is required. Two types of marginal peri-implantitis are known: peri-implant mucositis, i.e. an inflammatory process that is limited to the peri-implant soft tissue (occurring at the implant neck) without resorption of the alveolar bone, and real peri-implantitis, i.e. inflammation of soft tissues and bone loss detectable by x-ray. Also known from literature is apical peri-implantitis, which is not directly connected to the implant neck and is therefore not discussed in this paper.

Evaluation of two stage implant system Implantad

STI-BIO with respect to marginal bone resorption

The criteria for implant success have been set up with reference to the Bränemark Implant System. The major criterion is the annual measurement of marginal bone resorption and, in subsequent years, no more than 1.5 mm. No implant was lost during the period monitored. Substantial bone loss (2.0–3.5 mm) was observed in patients with mucosal dehiscence above the implant during the healing phase.

Conclusions: Results of the short-term STI-BIO implant study are fully comparable with more renowned systems; larger bone loss at the implant neck was explained in all cases. Peri-implantitis may occur with all implant systems presently in use. The most important factors causing peri-implantitis are: bacterial infection (colonisation of the implant surface and micro-gaps between the implant components), biomechanical overloading, chronic traumatisation of soft tissues, and acute trauma of hard tissues (preparatory thermal or mechanical). The prevention of peri-implantitis includes careful surgical procedure, exact prosthetic treatment, and especially oral hygiene (individual as well as professional). Based on clinical experience, prevention of peri-implantitis is being highly stressed as the most reliable measure ensuring the long-term success of implant function.
Stability assessment of immediately loaded alkali-etched implants

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² Lasak Ltd., Papířenská 25, Prague, Czech Republic

To determine changes in stability of immediately loaded implants with an alkali-etched surface.

18 STI-BIO Impladent dental implants were surgically placed in the mandible of patients ranging in age from 52 to 74 years. All implants were loaded immediately with ball attachments. Of the 18 inserted implants none failed over a subsequent 2 years period. The Periotest™ (Medizintechnik Gulden) was used for implant stability measurement. Periotest values (PTVs) were taken at weeks by 0, 2, 4, 8, 12, and 26 by a single observer. Each measurement was repeated three times and average value was used. Evaluation of bone quality according to Lekholm and Zarb’s classification was made during surgery for each site.

Across bone qualities 1, 2 and 3; 8, 8 and 2 implants were inserted respectively. Initial PTVs were in the range of -3 to +1 with a mean value of -1.54 (SD = 1.37, n = 18). Final PTVs at 26 weeks were in the range from -6 to -1 with mean value -3 (SD = 1.35, n = 18). Student t-test analysis showed a statistically significant decrease of implant stability after 2 weeks (p=0.01), as well as significant subsequent increase over the following six weeks (p=0.001). The maximum of the PTV values were detected 2 weeks after insertion (Fig. 1). It is assumed that the maximum represents a transition in the degree of stability from the time of primary bone contact to the development of early secondary bone contact during healing. After this time, secondary stability increased and stabilized at about 8 weeks. The results indicated that alkali-etched implants exhibit a greater rate of stability increase than machined implants (Ti-Machined) and an rate implant stability increase comparable to acid etched and grit blasted implants.

Based on the limited number of patient included, the implants with alkali-etched surface offer predictable and reliable results for immediate loading procedure in the edentulous mandible.

Alkali treatment - new concept of titanium implant surface modification

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² Lasak Ltd., Papířenská 25, Prague, Czech Republic

This study presents properties of recently developed alkali-treated titanium implant surface and evaluates the effect of early-loading protocol on the success rate of the alkali-treated implants.

Four titanium implant surfaces – alkali-treated, machined, sandblasted, and acid-etched - were tested. They were analyzed using electron microscopy and x-ray photoelectron spectroscopy. The wettability was determined using dynamic contact angle measurement and the real surface area was measured using krypton adsorption isotherm. The level of surface hydration was evaluated by infrared spectroscopy. In the 34-month clinical study the success rate of 1013 alkali-treated implants (Impladent STI-Bio, Lasak Ltd. Prague, Czech Republic) was evaluated. The healing time of these implants was 50% shorter than the conventional period. The alkali treatment of the titanium created a porous, hydrated, and reactive titanium oxide surface. The contact angle of the alkali-etched surface significantly decreased (Q = 29.9°) compared to that of the acid-etched (Q = 119.7°) and sandblasted (Q = 79.9°) surfaces. The level of surface hydration was increased 14 times compared to the acid-etched surface. The relative surface area of the machined, acid-etched, and alkali-etched titanium was 1.4, 7.2, and 137.9 respectively. In the clinical study no difference between the success rates of early loading and delayed loading protocol was found.

The alkali-treated surface proved to possess more favorable properties compared with other tested surfaces. The 50% reduction of the standard healing period had no significant effect on the implant success rate.
Internal sinus augmentation using porous resorbable calcium phosphate ceramic material

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² Lasak Ltd., Papířenská 25, Prague, Czech Republic
³ Institute of Molecular genetics, AS, Prague, Czech Republic

Sinus lift allows predictable implant placement in severely deficient posterior maxilla. In the present study internal sinus lift was used as an alternative to the most commonly used lateral window approach. As an augmentation material porous, resorbable calcium phosphate ceramics (PORESORB-TCP) with intrinsic osseoinductivity were used. The osseoinductive properties of the material are demonstrated by in vitro cell cultures and in vivo tests. Owing to the good X-ray contrast of the material the process of resorption and replacement by newly formed bone could be observed. Over a time period of 2 years the internal sinus lift procedure was used with 22 patients who received 48 hydroxyapatite coated implants (IMPLADENT, Lasak Ltd, Czech Republic). Clinically, all implants were successfully integrated, as supported by radiographic evaluation. The stability of implants was checked by a Periotest measurement. All implants are prosthetically loaded and fully functional. No complications arising from injury of the maxillary sinus mucous tissue were observed. We consider the internal sinus lift method as minimally invasive and broadening the indication of dental implants in the posterior maxilla region. It was proven that calcium phosphate material can be successfully used as augmentation material for sinus lift without any addition of autogenous graft.

Introduction

Sinuslift is a relatively recent method, which enabled applications of enossal implants to be greatly expanded, because it allows the bone to be augmented at the place where it is most deficient. The classical sinuslift is an operation, which must be performed very carefully to avoid damaging the epithelium of the maxillary sinus. Indeed, damaging the mucous membrane principally deteriorates the healing prognosis, as well as the function of the implant. The primary sinuslift, which is understandably preferred, means that the operation of the sinuslift is performed at the same time as the introduction of the implant. This procedure requires the primary stability of the implant to be well secured by a certain layer of alveolar bone. If this layer is 4 – 8 mm thick, applying the internal sinuslift is considered, the whole operation being performed from the place, where the implant is to be introduced without opening the maxillary sinus by antrotomy. Using this procedure, the operation is much less intrusive for the patient.

Materials and methods

Material characterisation (PORESORB®-TCP)
The structure of the material exhibits two types of porosity. Macropores of approx. 100μm in size and micropores ranging from 1 to 5μm. Micropores enable fast penetration of blood into the material and provide microrough surface favourable for cell attachment. The intergranular gaps of the packing from spherical granulates form a macro structure guiding the blood vessels, which supply the newly formed bone structures immediately with the necessary biochemical components. The resorption of calcium phosphate ceramics should take place at the same rate as that at new bone formation. PORESORB-TCP is usually completely resorbed with in 6 to 12 months and replaced by newly formed vital bone.

Surgical procedure
Between 2000 and 2002, applying the internal sinuslift, we implanted a total of 48 IMPLADENT SHA-O implants, diameter 5 mm, in 22 patients (14 male and 8 female) and 18 Ankylos B implants, diameter 4.5 mm. We combined the IMPLADENT implants with PORESORB-TCP material (LASAK Ltd., Prague, Czech Republic).
The preparatory work with drills of the same diameter as the future implant must be terminated 1 – 2 mm above the sinus floor. We then start working with bone condensers with a specially shaped head. The maxillary sinus is penetrated bluntly by careful impacts, the cortical bone layer of the sinus floor being lifted together with its mucous membrane. The extraction of the condenser may only be accompanied by bleeding, not air bubbles, the presence of which indicates the creation of an oro-antral connection with damage to the mucous membrane. After sufficient separation of the membrane, when the condenser penetrates the sinus to a depth of 6 – 10 mm, the created volume is filled with PORESORB-TCP bioactive ceramics. Having filled it, we introduce the implant of a given, largest possible diameter and length (longer than 12 mm). The introduction of the implant increases the pressure in the sinus and removes more of the sinus membrane. In most cases the primary stability of the implant is very good. It is secured by a bone layer more than 4 mm thick. Since implants of larger diameters (5 mm) are used exclusively in this procedure, it is nearly always necessary to commence the implantation with bone spreading. This spreads the spongy bone of the alveolar ridge of the upper jaw to the diameter of the implant. This spreading of the bone is simultaneously a very important factor for improving the primary stability of the implant. The primary stability enables the implant to be loaded gradually already after 3 – 4 months. The gradual loading is a process of getting the bone accustomed to the implant,
Clinical results

All implants are prosthetically loaded and the patients have no subjective or objective complaints. In the course of implant healing, an inflammatory complication in the region of the alveolar ridge of the upper jaw was observed in two patients. One patient complained of neuralgia of the 2nd branch of the trigeminus, most probably caused by irritation due to too robust an implant.

Complications associated with injury to the membrane of the maxillary sinus were not observed in our group of patients. The Periotest values ranged from -3 to 0. Postoperation X-rays show the filled area in the implant neighbourhood thanks to X-ray contrast. The contrast decreases after 2 years due to the resorption of the material and its replacement by the newly formed bone tissue.

Discussion

Our present experience indicates that the internal sinuslift is a considerate method, which can also be performed by a dentalalveolar surgeon. Thanks to a certain layer of alveolar bone, the stability of the implant is very good, which enables beginning loading after only 4 months. The loading is, of course, gradual by provisional bridges and requires certain restrictions on foodstuffs which are eaten. If the alveolar ridge is narrow, it must be widened at the beginning of the operation by expanders using the bone spreading method. The resorbable PORESORB-TCP augmentation material proved to be suitable for this application. We consider the greater X-ray contrast of PORESORB-TCP to be an advantage. Thanks to it the gradual replacement of this material by newly formed bone tissue can be monitored by X-ray. This is not possible, for example, with bovine hydroxyapatite. The periotest values do not vary greatly during the healing of the implant and, in our opinion, are determined by the height of the alveolar ridge, the quality of the bone and length of the implant.


Chicken embryo bone marrow from long bones derived cells were cultivated in vitro with pieces of TCP size of 0.3 mm. After initial cultivation
for 4 days in standard culture medium (M) the ALP activity was measured and found to be at the same level in all selected combinations of treatment (Fig. 1). Standard medium was then changed for Target media (D or V) and 3 days later ALP activity was found high in combinations of two factors: TCP-D or TCP-V, while in single treatments: TCP-M and all controls without TCP be it M, D or V no particular raise of ALP activity was observed. This trend continued until day 16, and then declined as cultures grew old and started dying. On day 27 the remaining cultures were stained for intracellular ALP, which confirmed the previous findings of extracellular ALP activity. In Fig. 2 (picture width 300 mm) ALP positive cells around a piece of TCP in medium V are shown. Similarly positive cells were observed in bright field around TCP in medium D in Fig. 3a in bright field and in 3b in phase contrast. 

Explanations:
TCP – synthetic, ß-tricalcium phosphate (PORESORB–TCP)

Conclusions
It was shown that the internal sinus lift method is minimally invasive and broadens the indication of dental implants in the posterior maxilla region. It was proven that beta tri-calcium phosphate (PORESORB-TCP) material can be successfully used as an augmentation material for sinus lift without any addition of autogenous graft. Thanks to the resorbability and good X-ray contrast of the material the process of new bone formation can be well monitored.

Results of the in vitro experiments confirmed osseoinductive properties of TCP providing that there are three cooperating components of the assay system. The first component was the committed cells in this case derived from chicken bone marrow. The second component was the osteogenic medium in either of the two forms used here, and the last indispensable ingredient was the TCP material. Functionality of such a system depends largely on the conditions of the cells to be committed to osteogenesis as in some earlier experiments these cells could start production of the ALP spontaneously. From this point of view the composed system described here apparently manifests higher potential for directing to osteogenesis even cells that are only weakly committed to osteogenic lineage.

Acknowledgements
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Literature
Bioactive titanium implants for shorter healing period

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A new type of surface treatment for titanium implants has been developed. It combines grit-blasting and acid- and alkali-etching procedures to create three levels of roughness on the macro, micro and sub-micro scale. It has been previously shown that this Bio surface exhibits bioactive and osseoadhesive properties. The effect of the Bio surface on cell differentiation was studied using in vitro cultivated cells isolated from chicken embryos. The first results indicated that the chemically treated titanium surface had the ability to influence differentiation of the surrounding cells towards osteoblastic phenotypic expression. The possibility of reducing the healing period by using implants with a Bio surface has been the subject of a clinical study. Over a period of 27 months, 485 implants were studied. The healing time was reduced to periods between 90 % and to 30 % of the standard healing period. The success rate of the osseointegration was 99.4 %, and the overall success rate of the implants was 98.6 %. The statistical data processing reveals that it is possible to reduce the healing period to one half of the conventional time. A prospective clinical study will investigate the possibility of shortening the healing period even further.

Replacement of individual teeth with IMPLADENT implants - a 5 year study

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Dental implantology is one of the fastest developing stomatological disciplines in the Czech Republic. Thanks to the constantly evolving practice of domestic implantology, dental implants are becoming generally accepted as a part of modern stomatology. In the early 1990’s there were several implantation systems developed in the Czech Republic, exemplifying further remarkable evolution11. Thanks to the wide offer of accessories, domestic implantation systems match the quality of foreign competitors, whilst still being available for reasonable prices. These implants however lack long-term clinical success rate evaluation, published in professional periodicals.

Introduction

We have so far devoted a series of articles to the IMPLADENT implant system, offered since 1992 by the Prague-based company Lasak s.r.o. (Ltd.)11. This has focussed on the shortened dental arch 11, and edentulous jaws11, presenting a three-year multicentric statistical analysis of a collection of 1264 implants11. We now aim to consider the potential of this system for the single tooth replacement.

We replace frontal teeth with dental implants far more frequently in the maxilla than in the mandible. This area is typical for high aesthetic demands concerning the quality of supraconstructions, generally limited bone availability, due to alveolar atrophy and extra-axial load on the implant. In lateral areas, we use implants in both jaws, in the area of molar teeth, mandible is predominant because the volume of alveolar bone available in maxilla is often considerably limited due to the maxillary antrum. The aesthetic aspect is much less important. The greatest problem is the risk of overloading the implant. Sometimes, it is recommended to use two implants under one crown, however, this is often impossible due to the gap length.

Material and methodology

The analysis incorporates all IMPLADENT implants, applied from July 1993 to February 1998, concerning the indication ‘loss of one tooth’, which were assigned for a solo crown. The data on implantations were acquired retrospectively from medical documentation. We followed the commonly specified contra-indications11. Before applying each implant, orthopantomogram was carried out for each patient, sometimes completed with intraoral x-ray images. In order to find out the thickness of alveolus, we made use of dental CT analysis or mapping the gingiva, using a hypodermic needle with a rubber disc. Most patients were instructed on dental hygiene. Recall examinations were carried out according to the previously published scheme.

The IMPLADENT system offers seven types of fixtures. Four of them are made of titanium alloy, Ti6Al4V (ISO 5832-3), with a hydroxyapatite coating, either cylindrical (VHA) or screw-shaped (SHA). They may contain antirotary elements which ensure that abutments do not rotate in the implant (SHA-O, VHA-O). The remaining three implants are made of pure titanium (ISO 5832-2) with grit-beasted surface. The first one is screw-type (STI), complemented with a modification comprising an antirotary element (STI-O). The diameter of all the above said implants is 3.6 mm and the length ranges from 8 to 14 mm. In addition, there is a titanium self-tapping screw with an antirotary element (STI-S), diameter 2.9 mm and length 10 to 15 mm. The implants SHA, VHA and STI were available from the very beginning of the monitored period. The SHA-O modification was introduced in October 1995 and we subsequently ceased to use all previous types for the replacement of one tooth. The STI-O and STI-S modifications were made available from October and December 1997, respectively and...
Dental implants

The VHA-O modification in February 1998. We used abutments with a diameter 3.6 or 4.8 mm, either direct or angled at 15 or 25 degrees. The supraconstructions were produced using the appropriate impression and laboratory tools.

Implants were left free of load and isolated from the mouth cavity for a period of at least three months (mandible) and six months (maxilla). Upon the conclusion of the healing period, the second surgery phase followed, with a two-week application of healing cylinders. We then fitted a new abutment, produced a solo crown and fixed it with cement.

We modified the success criteria according to Olsson et al. An implant was classified as a failure in the case of occurrence of at least one of the following: elimination from alveolus, poor stability, signs of chronic infection, pain or any other undesired subjective feelings, or considerable mechanical damage. Non-osseo-integration during the healing period was classified as primary failure, later malfunction as secondary failure. Implants of patients who did not accept the follow-up care were excluded from our statistical analysis. We evaluated the success rate using the “input-output” method and the life-table was determined according to the Kaplan-Meier method. The second evaluation method is based on the time interval elapsed from the application of each particular implant. It gives less optimistic results, however, it is a truer expression of reality.

Results

We inserted 1212 IMPLADENT implants between June 1993 and February 1998. We used IMPLADENT implants with 148 (12.2 %) patients (75 male and 73 female patients). One hundred and twenty-four patients had one tooth replaced, nine patients had two teeth replaced and in two cases, we replaced three teeth. Our group therefore comprised 135 patients. For further basic information, see Table 1, Diagram 1 details the age structure of the group.

Almost all of the implantations we carried out were delayed at least one year following the extraction of the tooth. In three cases, we carried out immediate or delayed immediate implantation, always in the frontal area of the maxilla, which in all cases turned out to be a success. In one case we replaced a tooth # 27 with two SHA 10 mm implants, bearing a single crown. We decided upon this option in order to avoid excessive load on the short implants in a poor-quality bone density D4.

For the type and length of implants used, see Diagrams 2 and 3. Ninety-eight (66.2 %) implants were made in the maxilla and fifty (33.8 %) in the mandible; for the exact location. For frequency, see Diagram 4. Twenty-three patients are still in the healing phase, which was completed amongst concerning the remaining 125 implants. We encountered primary failure in three cases, in locations 24, 33 and 36. The extractions were carried out not later than during the second surgery phase. The remaining 97.6% of implants were successful. Secondary failure was reported in one implant, location 11, which was extracted nineteen months after the application of the crown. The subsequent our-patient programme was not accepted by 29 patients (19.6%), with 19 tooth replacements in the maxilla and 10 replacements in the mandible. Other implants were monitored for an average period of 25.4 months from the implantation (ranging from 3 to 53 months) and 19.3 months after the application of the crown (from 1 to 50 months). Based on the „input-output” analysis, a total of 95.0% of implantations were successful. The result was identical for both jaws. For the life-table, see Diagram 5. Towards the end of the first and second year, the success rate reached 96.8%, dropping down to 96.8% during the third year. It subsequently remained unchanged.

Discussion

Comparison of statistical results between various publications often reveals considerable discrepancies. We therefore consider the most reliable a comparison with the previously published three-year multicentric study of 1264 IMPLADENT implants, used for all implantological indications. The success of the healing phase (97.7 %) was very close to our result – 97.6 %. According to the „input-output” analysis, 97.7 % of all implants were classified as successful after a period of thirty-six months. In our statistics, the success rate reached 96.8%, dropping down to 96.8% during the third year. It subsequently remained unchanged.

Diagram 1: Age structure of patients
Diagram 2: Types of implants applied
Diagram 3: Length of implants applied
Diagram 4: Implantation frequency

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result was 96.8% during the first and second year, and 94.8% after the third and fourth years. We compared literature results by Malevez et al. in order to analyse the topic of a one-tooth replacement. Their evaluation focussed on 84 Brånemark® implants, reporting 2.4% failure within a 5-year period, always during the first year after the implantation itself. Becker et al. evaluated 24 Brånemark® implants with the same indication. After a period of one year, they achieved a success rate of 95.7%. Klemke et al. indicate a 90.0% success rate of one-tooth replacements after 10 years, based on an analysis of 236 implants.

Conclusion

The study focuses on the topic of single tooth replacements using IMPLADENT dental implants. It references previous publications dealing with this implantation system. The authors draw our attention to specific problems of this particular procedure and present solutions based on the IMPLADENT system. Within the statistical section, the study presents 148 implants inserted over a period of 56 months. Implants were monitored during an average period of 25.4 months. Osseointegration was successful for 97.6% of all implants, based on the final „input-output” evaluation, 95.0% of all implants were successful. There was no considerable statistical difference between implants applied in the maxilla and the mandible. According to the Kaplan-Meier’s analysis, the success rate reached 96.8% after the first and second year, dropping down to 94.8% during the third year, without any further changes. The results are comparable to similar reports of foreign authors involved in the application of prestigious implantation systems. The authors conclude that replacement of single teeth using dental implants is one of the most complex and difficult tasks in stomatology. The IMPLADENT system exhibited adequate reliability over a medium time horizon, and convenient for the patient under consideration.

Literature

6. Impladent manual, Protetika, Lasak s.r.o., Prague 1996
Impladent was among the first systems introduced on the market in 1992 in CR and since that time has been subject to several analyses presented in professional periodicals. Our objective is to consolidate these studies and present a longitudinal clinical study of a representative collection of Impladent implants, inserted over a three-year period.

Impladent is a system of two-stage implants, of either of a cylindrical or screw type. The implants are available with a sand-blasted or hydroxyapatite coated surface. All these modifications are available with a diameter of 3.6 mm and lengths of 8, 10, 12 and 14 mm. In addition, there is a titanium self-tapping screw with a diameter of 2.9 mm and lengths of 12, 14, 16 and 18 mm. There is a wide range of abutments supplied: direct, with a 15 or 25 grade angulations, with antirotary elements, ball attachments, for temporary supraconstructions, etc. We recorded basic information on the patients (age, sex, diseases) and the implants applied (type, length), the indication and localisation, dates of first and second implantation phase, type of supraconstruction and follow-up data. For our statistical analysis, we applied the two-selection t-test and the relative frequency balance test. We then determined the success curve (life table) applying the Kaplan-Meier’ method. Over a period of three years, 529 patients received a total of 1264 implants. In total, 698 implants were applied in the maxilla and 566 implants were applied in the mandible. At the conclusion of the monitoring, the healing phase of 983 implants (77.8%) was successfully completed. Osseointegration was unsuccessful in 23 cases (primary failure), the success rate of the healing phase therefore being 97.7%. Fifteen of the osseointegrated implants were left dormant because the patient failed to appear for further treatment. Supraconstructions were applied to the remaining implants. The longest period from the application of an implant until the final check-up was 1110 days (437 days on average) and the longest period from the surgery phase to the final check-up took 915 days (264 days on average). We carried out final evaluation with respect to 950 implants, because 13 patients, comprising a total of 33 implants, failed to accept the follow-up care. Consequently, a total of 96.6% of implants were successful. We classified additional 32 implants (3.4%) as a failure, of which 23 were primary failures and 9 were secondary failures.

The success rate was also expressed on the basis of a “life table” analysis. The results are comparable with results exhibited by prestigious foreign implants. The statistical analyses of coated implants did not reveal any significant difference between the maxilla and mandible. Nor was there any significant difference in the number of primary and secondary failures. During the first twelve months from implantation, the risk of failure was not higher than during the subsequent period. Authors attribute these positive properties to the quality of hydroxyapatite implant coating. No event of delamination or dissolution of the coating has been reported. The authors believe that coated implants are particularly appropriate for use in areas of poorer bone quality. This hypothesis shall be examined in further research.
Use of Impladent dental implants in edentulous jaws

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Purpose of this 2 year study was to evaluate the efficacy of the Impladent system in the treatment of edentulous jaw with overdentures supported by ball attachments, screw-retained bars and magnetic attachments.

During the period from September 1993 to April 1996, we applied the Impladent implantation system to 54 patients with an indication „edentulous jaw“ (28 male and 26 female patients), aged from 35 to 74 years. The average age was 55.6 years. We replaced the upper dental arch of 17 patients and the lower dental arch of 30 patients, seven patients had both arches replaced. We had at our disposal ball attachments with Ceka nylon matrix and a metal adaptor, Dyna system of magnetic fixation (magnets identified as 500g) and the Preci-Horix bar with plastic sliders. After the delivery of a prosthetic replacement, all patients were included in an out-patient programme, including examinations carried out after one, three, six and nine months and then annually. Each year an additional x-ray examination was carried out.

We applied a total of 180 screw or cylindrical implants with hydroxyapatite coatings Impladent, 85 in the maxilla and 95 in the mandible, on average 3.3 implants per patient. We monitored implants for a period of 1 – 32 months after their insertion (19.0 months on average) and 1 – 27 months after the completion of the healing phase (13.6 months on average). Osseointegration failure only occurred in two implants, the success rate therefore being 98.8%. The healing phase was successful for 98.7% of cases in the maxilla, and 98.8% of cases in the mandible (considering only implants with the healing phase completed). After this period, no other implants were extracted.

All patients with ball attachments, clips and temporary bridges reported satisfaction with the outcome. Concerning the group with magnetic attachments, 12 of them reported being satisfied (48.0%), 4 of them were quite satisfied (16.0%) and 9 of them were not satisfied (36.0%) at all. The reason for their discontent was an inadequate retention of the replacement in eight cases (magnetic implants were replaced with other type of attachments). One patient was not satisfied with the aesthetic properties. The remaining 4 replacement were not evaluated, having lost contact with those out-patients.

Discussion: We are particularly satisfied with the successful osseo-integration rate. The success rate of 98.9% places Impladent among the world’s top-quality products in comparison with the results of other studies. Without any doubt, one of the greatest factors in this success is the hydroxyapatite coating. Together with other authors, we found the vague and poorly demonstrated claims questioning the quality of hydroxypapatite implant coatings to be poorly justified. Time will prove whether these objections are valid or whether these claims are simply due to commercial considerations.
Secondary Stability Assessment of Titanium Implants with an Alkali-Etched Surface: A Resonance Frequency Analysis Study in Beagle Dogs

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Purpose: This study was carried out to quantify the effect of an alkali-modified surface on implant stability during healing using an animal model. Materials and Methods: A total of 24 screw-shaped, self-tapping, commercially pure titanium dental implants, divided into a test group (implants with an alkali-modified surface or “biosurface”) and a control group (implants with a turned, machined surface) were inserted without pretapping in the tibiae of 3 beagle dogs. The resonance frequency analysis method was used to measure the implant stability quotient (ISQ) 0, 1, 3, 9, and 12 weeks after implantation. The animals were sacrificed after 2, 5, and 12 weeks, and the bone-implant contact (BIC%) was evaluated histomorphometrically. Results: The difference in the osseointegration rates (ΔISQ/Δhealing time) between the implants with alkali-modified surface (biosurface) and those with a turned, machined surface was evaluated as a mean of 0.843 ISQ/week within the first 9 weeks of healing. The mean increase in the secondary implant stability was found to be proportional to the mean increase in the BIC at healing period earlier than 5 weeks. Discussion: The characteristics that differed between the implant surfaces, ie, specific surface area, contact angle, and hydroxylation/hydration, may represent factors that influence the rate of osseointegration and the secondary implant stability. Conclusion: The alkali-treated surface enhances the secondary stability in the early stages of healing compared to the turned, machined surface, as a consequence of faster BIC formation.

In modern dental implantology, advanced treatment protocols (eg, early or immediate loading) are frequently used to reduce treatment time. Shortening the healing period entails new demands on both the primary and secondary stability of the implant. Primary implant stability is mainly dependent on the mechanical characteristics of the original bone (its local quality and quantity), the type of implant used (its geometry, diameter, length, and surface), and the surgical techniques employed. Secondary stability represents enhancement of the stability as a result of peri-implant bone formation through gradual bone remodeling and osteoconduction, with the possibility of new bone formation at the implant-bone interface.1 Contemporary knowledge indicates that the degree of micromotion at the bone-implant interface (primary stability) during initial healing is of utmost importance in achieving good secondary stability.14 However, several experimental and clinical studies have shown that secondary stability is also strongly influenced by the implant surface characteristics.5,6

To enhance secondary stability and accelerate the formation of stable and functional bone-implant interfaces, a number of implant surfaces have been developed. Surface roughness is the most frequently studied property affecting secondary stability. A summary of a number of studies1–4 providing analysis of bone-implant contact and removal torque values using animal models, indicates that roughened titanium surfaces generally exhibit greater contact with the bone and/or higher removal torque values than smoother implant surfaces, such as turned, machined or polished titanium surfaces. Some authors suggested that surfaces with mean roughness of 1.0 to 1.5 μm exhibit stronger bone response than smoother or rougher implant surfaces.8–12 These suggestions do not take into consideration chemical composition changes and physical chemical property variations introduced to the compared surfaces by the roughening procedures. It has been shown that the optimal roughness value varies according to the chemical composition and physical chemical properties of the tested surfaces and that, in some cases, the bone response can be more strongly affected by these parameters than by the roughness alone.13–15

Currently, surface modification of titanium by acid etching of turned or sandblasted titanium surfaces to create a micro-rough texture is being introduced to support and accelerate the healing and bone formation processes around the implant.16–18 Some experimental studies have indicated that acid-etched surfaces with minimal roughness (Ra = 0.62 μm)19 or greater roughness (Ra = 2.15 μm)17 exhibit a stronger bone response compared to sandblasted, moderately rough surfaces (Ra = 1.26 μm)14 or to mechanically turned, machined low-roughness surfaces (Ra = 0.86 μm).17 Some acid-etched implants with a micro-rough texture can accelerate the osseointegration processes despite their minimal roughness compared to the moderately roughened sand-blasted surfaces.

Surface chemistry of titanium implants is also thought to affect the secondary stability independently of the surface topography although it is difficult to separate the effects of these 2 factors. There is extensive experimental evidence in the literature showing that biomaterials with different chemical compositions trigger different biologic responses.22–26 A specific biologic response is elicited at the interface with bioactive materials with high surface reactivities,27 resulting in the formation of a bond between the tissue and the material surface.22 Bioactive materials with special chemical compositions, including bioactive glasses, bioactive glass ceramics, hydrated silica, and titanium gel oxides or hydroxyapatite have the ability to very rapidly form a stable interface with bone tissue following the formation of calcium phosphate deposition on their surfaces as a consequence of chemical interaction with body fluids.22 For instance, 50% bone-implant contact is achieved within 7 days for bioactive glasses.22 The mechanical strength of this interface usually exceeds the strength of the bone tissue to which the bioactive material is bonded.27,28 However, because their poor mechanical properties prevent their use as solid implants under high load-bearing conditions, bioactive materials (especially hydroxyapatite) are generally used as a coating applied on the surface of titanium implants to achieve faster and more reliable bonding with bone tissue. In spite of the success achieved in accelerating the bone healing process and in exhibiting greater tolerance to low primary stability in the early phases of healing,27–29 hydroxyapatite plasma-coated coatings have often been the subject of controversy regarding their stability30,31 and low long-term success rate.34,35 On the other hand, some medium-term and long-term clinical studies demonstrate the high success rate of hydroxyapatite-coated implants.36–38 The literature tends to indicate the imperfect adhesion of the hydroxyapatite surface layers and their chemical and mechanical instability rather than their inability to substantially improve osseointegration, especially in the initial stages of healing.
Various attempts have been made to modify the titanium surface to make it bioactive without the use of a thick coating of another bioactive material. The most successful methods of titanium bioactivation have been, for example, alkali or fluoride treatment.\textsuperscript{39–41} Recently, an alkali-modified surface for titanium dental and spinal implants was clinically introduced under the brand name Bio-surface as a potentially bioactive surface.\textsuperscript{42–47} The present study was carried out to quantify the effect of this surface modification on the implant stability during healing, using resonance frequency analysis and histologic examination on an animal model. This study also describes the characteristic surface properties of this surface and compares them with the turned, machined titanium surface.

Materials and methods

Implant Materials
A total of 24 screw-shaped, self-tapping, (c.p.) titanium dental implants divided into a control group (12 implants with a turned, machined surface) and a test group (12 implants [Lasak, Prague, Czech Republic]) with sandblasted, acid-, and alkali-treated surface [Bio-surface; Lasak]) were inserted without pretapping into the tibiae of 3 beagle dogs. Identically shaped implants with a diameter of 3.7 mm and a length of 10 mm were used for the test and control groups.

Surface Characterization
The overall surface morphology of the employed implants was characterized using a scanning electron microscope (SEM, Hitachi, Tokyo, Japan) with an accelerating voltage of 15 to 30 kV (Fig. 1). Surface roughness measurement was carried out using a scanning surface topography instrument, a Talyssurf CLI 1000 with a confocal CLA gauge (Taylor Hobson, Leicester, United Kingdom), which provides highly accurate non-contact 3-dimensional measurements. The following were measured for 3 implants in each group: 3 thread tops, 3 thread valleys, and 3 thread flanks. Four 3-dimensional parameters (amplitude, spacing, and hybrid) were calculated (Table 1). The original unfiltered measurements and measurements made with a filter size of 50 X 50 μm were evaluated.

Dynamic contact angle measurement was performed by the Wilhelmy plate method using a Tensiometer K15 instrument (Knuss, Hamburg, Germany). The wetting angle values in water were determined from the dependence of the wetting force on the immersion depth. The mean values of the wetting angle were calculated from 4 repeated measurements (Table 2). The specific surface area was determined by the BET krypton gas adsorption method. The surface area is expressed in relation to unit geometric surface area of the implant as the mean value of 4 repeated measurements (Table 2). The determination was performed using an ASAP 2010 M instrument (Micromeritics, Norcross, GA). Diffuse reflectance infrared Fourier transform (DRIFT) spectroscopy was used to determine the degree of surface hydration. The measurement was performed on a Nicolet 740 instrument.

Fig. 1: SEM images of (a to c) the turned, machined surface and (d to f) the biosurface (original magnification a and d X30; b and e X200, c X1500; f X4000).
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Hydroxyl group density was expressed as the radiation absorption in Kubelka-Munk units (KMU) evaluating a band at 3,400 cm⁻¹ with resolution of 4 cm⁻¹ (Fig. 2).

Surgical Procedure and Implant Placement

Three beagle dogs (mean weight, 16 ± 2 kg; mean age, 2 years) were used in this study, which was approved by the Ethics Committee for Work with Experimental Animals at the Teaching Hospital, Charles University, Hradec Králové, Czech Republic. Anesthesia was carried out using a dose of 15 mg/kg of 5% Narkamon (Spofa, Prague, Czech Republic) and 2 mg/kg of 2% Rometar (Spofa). The animals were premedicated with a dose of 0.05 mg of Atropin (Biotica, Slovenská Lupca, Slovakia) 30 minutes prior to surgery.

Under total anesthesia, in a supine position, following the usual preparation of the operation field and toweling, a surgical cut with a length of approximately 7 cm was made on the anteromedial surface of the tibia. A sharp cut was made in the fascia and then in the periosteum, which was widened to the side with a raspatory. Following uncovering of the surface of the tibia, the positions for drilling the holes for implanting the tested implants were marked. The implant sites were prepared with 1.5-mm, 2.0-mm pilot, and 2.2-mm final drill. The implants were inserted to a depth of 10 mm with a torque of 35 Ncm. The bone-implant contact was evaluated using a histometric analysis.
3.0-mm drills at 800 rpm with simultaneous cooling with a physiologic solution. A countersink drill was then used. The implants were screwed in with an insertion torque of approximately 40 Ncm. The biosurfaces and machined implants were alternately inserted in the tibia side by side (eg, biosurfaced, machined, biosurfaced, and so on), as depicted in Fig. 3. The order of the implants was the opposite in the other leg of the same animal (eg, machined, biosurfaced, machined, and so on). The first animal received 12 implants (6 from each group). Each of the other 2 animals received 6 implants (3 from each group). In the first animal, implantation was followed by measuring the stability of the implants using resonance frequency analysis. A probe was screwed into the cortical part of the implant, and the soft tissues were pulled to one side with a hook so that they did not affect the probe. The measurement was carried out twice for each implant. This was followed by rinsing of the operation incision. The implants were closed with cover screws, and, following control of hemostasis, the operation incisions were closed in layers and were finally covered with a sterile bandage. Implant stability measurement was repeated for each implant in the 1st, 3rd, 9th, and 12th weeks using the same procedure. Following completion of the experiment, the dogs were sacrificed after 2, 5, and 12 weeks by an overdose of thiopental (ICN, Roztoky u Prahy, Czech Republic), and the tibiae were removed and fixed in 10% formaldehyde prior to histologic evaluation.

Implant Stability Measurement

The resonance frequency analysis (RFA) method was used to measure implant stability. The measurement was performed using an Oststell instrument (Integration Diagnostics, Göteborg, Sweden) with a commercially available transducer (type F37 L5). The perpendicular orientation of the transducer along the long axis of the bone was always maintained. The transducer, which is fixed to the implant in the bone, contains a wave source and an analyzer. The wave source vibrates with gradually increasing frequency. The analyzer records the frequency of the source, causing the resonance in the transducer-implant-bone system, including the bone-implant interface. The recorded frequency value in Hz is converted to an implant stability quotient (ISQ) value, which varies in a range from 0 to 100, with 100 indicating maximum stability. The method allows measurement with a precision of ± 1 ISQ.

Histologic Preparation and Analysis

The tibiae were dissected and blocks of 7 mm thickness containing 1 implant each were prepared. Non-decalcified (ground) sections were processed according to the method of Donath and Breuner.26 Thin sections with a thickness of 30 to 50 μm were stained with toluidine blue and examined using an optical microscope (Olympus BX-60, Tokyo, Japan) equipped with an image system (Quick PHOTO Industrial 2.0; Olympus, Prague, Czech Republic). Bone-implant interfaces at the threaded part were histometrically analyzed by evaluating the percentage of bone-implant contact (BIC). The length of the bone tissue in direct contact with the implant (BC) and the total interface length (II) were measured (Fig. 4). The percentage of BIC is given by the ratio of the direct contact length to the total interface length multiplied by 100. The presented mean values were calculated from the 3 sections as an average for each implant for both types of implant surface.

Statistical Analysis

Implant stability quotient (ISQ) and BIC% data are reported as mean values with standard deviations (SD). Nonparametric Friedman’s analysis of variance (ANOVA) was performed to analyze variations of ISQ and BIC% during the 12-week follow-up period. The analysis was followed by nonparametric Wilcoxon tests to determine differences within groups between particular time intervals. To investigate the statistical significance of the implant group differences, the data were subjected to the nonparametric Mann-Whitney tests. A difference was considered significant when P < .05.

To determine the difference in the osseointegration rate between the test and control groups, the time-dependence of the differences (test ISQ – control ISQ and test BIC% – control BIC%) were evaluated by the method of linear regressions and the parameters of the straight lines (slopes and intercepts) were determined. These parameters were calculated together with the P values, reliability limits, and correlation coefficients. The values of the physico-chemical properties of surfaces are presented as mean values and standard deviations (SD).

Results

Primary Implant Stability and Variation of Implant Stability During Healing

The mean ISQs for the machined and biosurfaced groups (± SD) at baseline (primary stability) and at 1, 3, 9, and 12 weeks are presented in Fig. 5. The RFA measurements indicate similar mean primary stabilities for both groups of monitored implants (74.0 ± 2.45 for biosurfaced and 74.5 ± 2.99 for machined); no significant difference was observed between the 2 groups (P = .871; Mann-Whitney test). Friedman ANOVA revealed no statistically significant differences in biosurface-group ISQ at any of the measured time points (P = 0.482). In contrast, significant differences (P < .001) were revealed during the 12-week follow-up for the machined group. There was a statistically significant decrease in ISQ between baseline and 3 (P = .028) and 9 (P = .028) weeks (Wilcoxon matched pairs test). A statistically significant difference between the test and control groups was observed after 3 (P = .0035) and 9 (P = .0035) weeks (Mann-Whitney test; Fig. 5).

The time dependence of the differences (biosurface ISQ – machined ISQ) in the first 9 weeks of follow-up was evaluated by the method of linear regressions, and the parameters of the straight line (slope and intercept) were determined as follows: biosurface ISQ – machined ISQ = 0.843 X time (weeks) + 1.0111. The correlation coefficient (r = 0.756) indicated a moderately strong relationship between variables; the slope of the regression line was positive and statistically significant (P < .001).

Fig. 5: Mean implant stability (ISQ ± SD) of biosurfaced implants (B) and machined implants (M) at placement (baseline) and after 1, 3, 9, and 12 weeks. * indicates significant difference (P < .05).

Fig. 6: Time dependence of implant stability difference (biosurface ISQ – machined ISQ). Regression straight line: biosurface ISQ – machined ISQ = 0.843 X time (weeks) + 1.0111. Boundaries of reliability are shown (dashed lines). Correlation coefficient r = 0.756.
Histologic Examination (Histologic Observations)

Histologic examination of the bone-implant interface was performed for both types of tested implant surfaces 2, 5, and 12 weeks after implantation. Three implants from each group were evaluated for each time interval. The cervical part of the implants was mostly surrounded by cortical bone, while the thread part was surrounded by trabecular bone. Close BIC was frequently observed at the cervical part of the implants with the biosurface. In contrast, the specimens with turned, machined surfaces showed patchy BIC, and intermediate soft tissue was indicated in some cases (Fig. 7). The time development of the bone-implant contact (BIC%) at the thread part of the implant was evaluated histometrically (Figs. 8 and 9). The results of the histometric analysis are presented in Fig. 10.

Friedman ANOVA revealed statistically significant differences in BIC% for the test group (P = .029) as well as the control group (P = .032) at all measured time points. BIC of the biosurfaced group increased sharply during the first 2 weeks in contrast to the machined group, which exhibited a gradual increase starting at a later timepoint (Fig. 11). Using the Mann-Whitney test, statistically significant differences between the test and control groups were observed after 2 weeks (P = .046), 5 weeks (P = .049), and 12 weeks (P = .049). The time dependence of the differences in BIC between the groups in the first 5 weeks was evaluated by the method of linear regressions, and the parameters of the straight line (slope and intercept) were determined as follows: test BIC − control BIC = 7.94 * time (weeks) + 10.92. The positive slope of the regression line was found to have a correlation coefficient of r = 0.7561 and to be statistically significant (P = .013).

Relationship Between Implant Stability and BIC

Evaluation of the correlation between the contribution of the biosurface to the implant stability (test ISQ − control ISQ) and the BIC (test BIC% − control BIC%) could not be performed by the standard method of correlation coefficients because of the experimental arrangement. Nevertheless, it was shown that test BIC% − control BIC% was proportional to time (P = .013) as was test ISQ − control ISQ (P < .001). This supports the hypothesis that during the first 5 weeks of the healing period, these 2 quantities are proportional to each other.

Discussion

This study presents the results of measurement of changes in the stability and BIC during healing of implants with biosurfaces and turned, machined surfaces.
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If the osseointegration rate is defined as proportional to the secondary stability rate ($\Delta ISQ/\Delta t$), the difference in the osseointegration rate between the biosurfaces and turned, machined surfaces could be estimated as a mean value at 0.843 ISQ/week within the first 9 weeks of healing.

Referring to the evaluation of the correlation between the contributions of the biosurface to ISQ and BIC, it may also be concluded that the variations in the secondary stabilities of the biosurfaced implants were proportional to the changes in the BIC within the first 5 weeks of follow-up. These findings demonstrated that the biosurface enhances secondary stability compared with the turned, machined surface as a consequence of more rapid formation of BIC in the early stages of healing.

It can be speculated that the differences in the rates of osseointegration in the initial stages of healing for the biosurfaces and the machined surfaces could be related to different surface reactivity due to the different surface material properties, eg, specific surface area, surface wettability, surface contact angle, and surface hydroxylation/hydration. In general, surface reactivity, which is a common characteristic of bioactive materials, increases with increasing specific surface area. Therefore, the 3-dimensional macro-, micro-, and nano-structured biosurface which exhibits a surface area almost 100 times larger than the turned, machined surface, may significantly enhance the surface reactivity with the surrounding ions, amino acids, and proteins, which modulate the initial cellular events at the cell-material interface. In addition, the easily wettable hydrophilic biosurface enables establishment of good contact between the body (specifically, blood) and the rough and porous structure of the implant, and thus contributes to cellular and biomolecular migration and adhesion. This biosurface, which is rich in hydroxyl groups, exhibits a hydrophilic character, a low wetting angle, and high surface free energy, which is a common characteristic of bioactive materials, increases with increasing specific surface area. Therefore, the 3-dimensional macro-, micro-, and nano-structured biosurface which exhibits a surface area almost 100 times larger than the turned, machined surface, may significantly enhance the surface reactivity with the surrounding ions, amino acids, and proteins, which modulate the initial cellular events at the cell-material interface.

Conclusions

The test surface (biosurface) enhances the secondary stability at an early stage of healing compared with the turned, machined surface, as a consequence of more rapid bone-implant contact formation. In contrast to the hydrophobic turned, machined surface, the biosurface, which is rich in hydroxyl groups, exhibits a hydrophilic character, a low wetting angle, and a high specific surface area.

Acknowledgments

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Effect of chemically modified titanium surfaces on protein adsorption and osteoblast precursor cell behavior

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Purpose: To investigate the effects of different chemically treated titanium on protein adsorption and the osteoblastic differentiation of human embryonic palatal mesenchyme cells (HEPM). Materials and Methods: In this study, a total of 3 different surfaces were evaluated. The first had a machined surface used as controls for the study (Ti-M). The second sample type (Ti-AE) was acid etched. The third type of altered titanium (Ti-AAE) samples was prepared by exposing the Ti-AE samples to NaOH solution. The surface characteristics of chemically modified Ti were investigated by means of scanning electron microscopy (SEM), Fourier transform infrared spectroscopy (FTIR), and profilometry. To evaluate the production of biomarkers, commercial kits were utilized. Results: This study shows that surface composition and morphology affected the kinetics of protein adsorption. Ti-AE surfaces manifested a higher affinity for fibronectin adsorption when compared to Ti-M and Ti-AAE surfaces. It was observed that Ti-AE and Ti-AAE surfaces promoted a significantly higher cell attachment when compared to Ti-M surfaces. Statistically significant differences were also observed in the expression of ALP activity, osteocalcin, and osteopontin on all three titanium surfaces. Discussion: Alkaline phosphatase (ALP) activity and osteocalcin production up to day 12 suggested that differentiation of the cells into osteoblasts had occurred and were expressing a bone-forming phenotype. Conclusions: It was thus concluded from this study that surface morphology and composition play a critical role in enhancing HEPM cell proliferation and its differentiation into osteoblast cells.

Fig. 1: FN adsorption kinetics of 0.5 mg/mL FN over 3-hour period. Values represent means standard deviation (n = 5)

Fig. 2: Osteoblast precursor cell attachment on Ti-M, Ti-AE and Ti-AAE. Values represent means standard deviation (n = 5)

Fig. 3: Osteoblast precursor cell proliferation after 1, 6, 12, 18 and 24 days of culture on Ti-M, Ti-AE and Ti-AAE. Values represent means ± standard deviation; each bar represents 5 disks * denotes a statistically significant difference (P < .05)

Fig. 4: SEM images of osteoblast precursor cells plated on (a) Ti-M, (b) Ti-AE and (c) Ti-AAE surfaces after 6 hours of cell culture
Chemically treated titanium: early surface activity detected in vitro

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This study examined the early interaction of the acid and alkali-etched titanium surfaces with the simulated body fluid (SBF) and cultured cells. Machined commercially pure (CP) titanium was used as a reference material. After exposure to SBF samples were analyzed by X-ray photoelectron spectroscopy, X-ray diffraction and electron microanalysis. The results proved presence of adsorbed calcium and phosphate ions on the chemically treated (CT) surface after 2.5 minutes exposure to SBF. The adsorbate later transformed into crystalline apatite. The surface changes were in accord with those in the solution, showing an immediate decrease of calcium and phosphate concentrations. The rapid formation of the calcium phosphate on the surface could cause the bioactive properties of the (CT) surface. The significance of this finding was tested in an interaction with living cells in vitro. Machined CP titanium did not stimulate long lasting cells, ALP negative, did not discriminate between CT Ti or CP Ti or steel within first 2 days. Where as over a period of 2 weeks it was proven that haptotaxis was a property of only the CT Ti. Chicken embryonic cells with an osteoblastic potential during 3 weeks in contact with the CT Ti surface attached themselves to it and showed increased alkaline phosphatase (ALP) activity. This therefore indicated its conceivable osteoinductive potential. This relationship between surface chemistry and bioactivity was revealed by an in vitro bioassay with living cells.

Effect of plasma sprayed hydroxyapatite coating on osteoconductivity of commercially pure titanium implants

Z. Strnad1, J. Strnad2, C. Povýšil3, K. Urban4


This study examines the formation of a calcium phosphate layer on surfaces of plasma sprayed hydroxyapatite (PSHA) and sand-blasted commercially pure (CP) titanium in simulated body fluid (SBF) with ion concentrations similar to those of human blood plasma. The surface of PSHA induced the formation of calcium phosphate surface layers, while the precipitation of calcium phosphate on sand-blasted CP titanium has not been detected. Histologic evaluation of in vivo tests has demonstrated that implants with a PSHA coating enabled the growth of the bone tissue into gaps with a depth of up to 1 mm without significant formation of intermediate fibrous tissue. In comparison to sand-blasted CP titanium, implants with PSHA coating exhibited greater tolerance to unfavorable conditions during healing, such as gaps at the interface or primary instability of the implant. In the case of good primary stability of the implant gap-filling with fibrous tissue was observed for sand-blasted CP titanium implants over the greater part of the surface of gaps with a depth of 0.3 mm. A direct contact of CP titanium implants with bone was achieved only when press-fit implantation model was used.

Introduction

It is known that, in addition to their ability to form a direct bond with living bone tissue, bioactive materials (bioglass, A-W glassceramic, Hydroxyapatite) also exhibit osteoconductive properties in contrast to bioinert materials such as CP titanium or the titanium alloy, Ti6Al4V. Osteoconductive properties are understood to consist of the ability of the material to act as a lattice for the osteoblast in the interconnection of defects (gaps) during the gradual formation of new bone. It can be expected that this specific property of bioactive materials is a consequence of their ability to form a thin calcium phosphate layer on the surface of the implant during a period of minutes to days, depending on the type of material, as a consequence of reactions with body fluids. The chemical and crystallographic properties of this calcium phosphate phase are almost identical with bone apatite. It can be assumed that bioactive implants with osteoconductive ability will thus exhibit greater tolerance to unfavorable conditions during implant healing, such as micro-movements or gaps between the implant and the bone matrix. It was recently demonstrated in the work of Clemens et al.
that there is a maximum gap size which still permits bone apposition on the surface of an implant with plasma sprayed hydroxyapatite (PSHA) coating during healing, without the gap filling with soft tissue. The authors demonstrated on an animal model that this limit lies between 1-2 mm. Gaps of 1 mm between an implant with PSHA coating and the bone were filled with new bone without being filled by soft tissue. This is in contrast to a CP titanium implant, where the bone was separated by fibrous tissue under the same conditions. The same results were achieved both 6 weeks and 6 months following the implantation. It can be assumed that a maximal size of the gap for which gap-filling with soft tissue will not occur will also exist for titanium implants, with a value of less than 1 mm.\textsuperscript{10,11,12}

<table>
<thead>
<tr>
<th>$R_s$ (um)</th>
<th>S (um)</th>
<th>$R_k$</th>
</tr>
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<tbody>
<tr>
<td>sand-blasted</td>
<td>0.9</td>
<td>12.9</td>
</tr>
<tr>
<td>PSHA coated</td>
<td>8.3</td>
<td>47.0</td>
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Table 1: Roughness parameters of sand-blasted and plasma sprayed hydroxyapatite (PSHA) coated surfaces.

Thus, models with gaps of 0 to 1 mm were selected for the present study to investigate the growth of bone tissue. Samples of uncoated CP titanium implants and implants with PSHA coating were implanted into the tibiae of dogs in model arrangements with different primary stability and primary contact of bones to the surface of the implant, for histologic evaluation. The interactions of implants with PSHA coating and CP titanium implants with simulated body fluid (SBF) were studied by tests in vitro. These tests involved analysis of the surface of the implants and examination of changes in the concentrations of ions in SBF solutions dependent upon the exposure period.

## Methods and materials

### Sample preparation

Samples in simulated body fluid (in vitro test) were prepared using commercially pure (CP) titanium (Austenal Dental material AB, Malmö, Sweden) grade 3, in the form of discs with a diameter of 10 mm and thickness of 1 mm. Non-coated samples were roughened by sand-blasting with alumina powder (grain size 100μm). The samples were washed in ethanol in an ultrasonic cleaner and dried at 120°C. Coated samples were prepared by plasmatic deposition of hydroxyapatite on the surface of the abovementioned sand-blasted CP titanium discs. Thickness of the coating was 50μm. Surface roughness parameters of both sand-blasted and plasma sprayed hydroxyapatite (PSHA) coated samples were evaluated (Tab.2): $R_s$ - arithmetic mean of the profile departures from the mean line, S – mean spacing of the adjacent local peaks, $R_k$ – profile sharpness. The profilometer TALYSURF 6 (Taylor Hobson, Leicester, U.K.) was used. Test implantation was undertaken using implants with the above described sand-blasted and PSHA coated surfaces. Both kinds of implants had the shape of cylinders (or cylinders with a 0.3 mm deep groove, Fig. 2), with a diameter of 3.7 mm and height of 10 mm. Before plasma spraying the diameter of implants was machined to 3.6 mm so that an identical final diameter (3.6 mm±0.02 mm) of both PSHA coated and sand-blasted implants was ensured.

### Exposure of samples in simulated body fluid (SBF) and analytical methods

The samples were exposed in SBF (Fig. 1) with a composition similar to that of the inorganic part of blood plasma (Table 2). The pH value was adjusted to 7.40 at 36.5°C. The samples were immersed in 100 ml of SBF-solution, where the ratio of the surface of the sample (S) to the volume of the solution (V) had a value of $S/V = 0.02cm^{-1}$. Concentration changes in sample extracts were determined spectrophotometrically (UV-1201, Shimadzu Europe, Ltd.), by atomic absorption spectroscopy (VARIAN-Spectr AA300) and using a pH meter (WTW-526). Following exposure in SBF, the surface of the samples was studied using a scanning electron fitted with an energy dispersion analyzer -SEM-EDS (Jeol, U.S.A., Inc.) and thin film X-ray diffraction (TF-XRD) using a Seifert XRD 3000P diffractometer.

### Implantation of the tested materials and histological evaluation

The Ethical Board of the Orthopedic Clinic of the Faculty Hospital in Hradec Králové, Czech Republic, approved all experimental procedures used in the study. Prior to the operation, the tested materials were sterilized with saturated water vapor at a temperature of 125°C and pressure of 140 kPa for a period of 15 minutes. Implantation was carried out on 2 dogs of different sex of an unknown breed with a weight of 12 kg (± 2 kg). They were premedicated with Dolzin (Biotech a.s., Slovakia), 10 mg/kg of weight, half an hour prior to the operation. Anesthesia was carried out continuously by an i.v. infusion of a 2% solution of Thiopental (Sofia a.s.). Following dissection of the operation area and drying, an incision was made above the upper edge of the tibia. Following passage through the soft tissue and moving the periosteum to one side, holes were drilled in the corticais using a bit with a diameter of 3.7 or 4.7 mm.

Three implantation models were used. Implants with a diameter of 3.7 mm were introduced into the holes with the same diameter, i.e. were placed inside the bone with good primary stability - press-fit (model A, Fig. 2a). The second implantation model differed from the previous one in that the implants had a groove around their circumference with a depth of 0.3 mm.
pairs of implants in the right tibia -models A, C. The other animal received one pair -model C in the left and one pair -model B in the right tibia. The total of 12 implants were used in the study.

In the post-operative period, antibiotics were not administered. Three months after the operation, the animals were put down an overdose of Thiopental. The tibiae were removed, immersed in 10% formaldehyde and blocks containing 1 implant each were prepared. The blocks were then dehydrated using graded methanols (70-100%) and embedded in methylmethacrylate. Samples were processed undecalcified. From each implant two longitudinal sections with thicknesses of 5-50μm were made on a saw (Struers Accutom-2, Struers, Copenhagen, Denmark). Sections were affixed onto a glass slides and if necessary hand-ground to a thickness of 5-10μm. Sections were stained with toluidine blue. Photomicrographs for histologic analyses were taken using an OLYMPUS BX-60 microscope fitted with JAI 2040 CCD camera (JAI Corp., Yokohama, Japan). The digitalized image analysis was performed using LUCIA 4.1 software (Laboratory imaging, LIM, Ltd., Czech republic).

The length of the bone tissue in direct contact with implant (BC) and the total interface length (IL) were measured. The percentage of bone-implant contact is given by the ratio of the direct contact length to the total interface length (BC/IL). The presented mean values were calculated from the four results (4 sections) available for all types of the particular implantation model.

Results

Immersion of the PSHA coated implants in the simulated body fluid was accompanied by removal of calcium and phosphate ions from solution (Fig. 3a, b) and deposition of spherulitic crystals of calcium phosphate on the surface, as depicted in the electron microscope image (Fig. 4a, b). During the first few days, there was a clear decrease of the Ca2+ and PO4 3- concentration in the solution and the process stabilized after 10-12 days. Following more than 10 days of exposure, the surface of the sample was covered with a continuous layer of calcium phosphate agglomerates (Fig. 5). Analysis of the surface layer with an electron microprobe demonstrated that the molar Ca/P ratio of 1.50 was close to that of hydroxyapatite (1.67) (Fig. 6). The crystalline nature of the precipitated layer was also confirmed by thin film X-ray diffraction analysis (Fig. 7). The X-ray diffraction pattern of the calcium phosphate layer (Fig. 7b) indicates its diffussion character, similar to that of bone apatite (Fig. 7c), in contrast to the x-ray diffraction pattern of the original PSHA coated sample (Fig. 7a), which indicates larger crystals of the apatite phase and higher crystallinity.

Sand-blasted CP titanium samples did not exhibit a decrease in the concentration of calcium and phosphate ions in the leaching solution following exposure in SBF, even after 50 days, as indicated in the time dependence in Fig. 3. The changes in the concentrations of Ca2+ and PO4 3- ions in the solution lay within the range of the changes in concentration observed in a parallel control sample. In addition, analysis of the surface of samples using an electron microprobe did not indicate the presence of calcium or phosphorus. Microscopic examination of implants introduced by the press-fit method (model A) showed a high percentage of bone tissue apposition to the surface of the implant three months following the implantation, both for sand-blasted CP titanium implants (BC/IL=54%) and for implants with PSHA coating (BC/IL=79%).

(Figs. 8 a, b). The effect of the different osteoconductive ability of the tested materials was marked for implants with a defined gap between the implant and the bone with relatively good primary stability of the implant (model B). Three months after the implantation of the sand-blasted CP titanium implant, the gap (0.3 mm) was filled with new bone tissue; however, over its entire surface it was separated from the titanium surface by fibrous tissue with a thickness of about 50 μm (BC/IL=2%), in contrast to the gap with the PSHA coated surface, where the newly formed bone was immediately adjacent to the surface of the implant without significant gap-filling by fibrous tissue (BC/IL=88%) (Figs. 8 c, d). Histological evaluation of the non-press-fit implants in the bone matrix with a gap of 0 - 1 mm with an average value of 0.5 mm (model C) once again demonstrated the effect of PSHA coating. The surface of the sand-blasted CP titanium implants was almost entirely coated with a layer of fibrous tissue with negligible area in direct contact with bone (BC/IL=5%). The implant coated with hydroxyapatite was in direct contact with the newly formed bone tissue over 72% of the total surface of the implant (BC/IL=72%) (Figs. 8 e, f).

Discussion

A plasma sprayed hydroxyapatite coating exposed to a solution of simulated body fluid (SBF) induced the formation of a calcium phosphate surface layer, which was chemically and crystallographically similar to bone apatite. This was in contrast to sand-blasted CP titanium, where a calcium phosphate layer was not formed. The results indicate that the formation of calcium phosphate layers on the surface of PSHA coated implants may contribute to their osteoconductive properties. Unfavorable conditions for the healing of implants such as gaps between the bone and implant and primary instability of the implant at the time of placement were overcome better by implants with PSHA coating than sand-blasted CP titanium implants. The surface of implants with PSHA coating permitted the growth of bone tissue into gaps with a depth of 1 mm without significant gap filling by fibrous tissue. The formation of direct contact between the surface of the implant and the bone was affected less by primary instability of the implant than in the case of sand-blasted CP titanium surface. The sand-blasted CP titanium surface enables only a small degree of

![Fig. 3: Time dependence of changes in the concentration of calcium and phosphate ions in a simulated body fluid solution (SBF) for exposure of samples of pure titanium (△) and plasma sprayed hydroxyapatite (○). Values depicted as (●) correspond to the SBF blank.](image-url)
gap healing without the formation of intermedial fibrous tissue. Gap filling by soft tissue over a major part of the area of the gap was still observed for gaps with a depth of 0.3 mm, with relatively good primary stability of the implant. Only the press-fit implantation model (model A) permitted osseo-integration of implants with sand-blasted CP titanium surface. It is reasonable to assume that the observed difference in osteoconductivity of the tested surfaces is not associated with any difference in roughness, since the average roughness (Ra) of the sand-blasted surface was close to the interval of about 1-1.4 um, which is usually considered the most suitable for good metal-bone fixation. Unfortunately, the small number of implants used in the study was unsuitable for statistical analysis, but clinically relevant conclusions may nevertheless be drawn from the results. It should be pointed out that these conclusions were based on unloaded implants histologically examined using light microscopy.

Fig. 4: Electron microscope images (a) of the surface of a sample with plasma sprayed hydroxyapatite coating and (b) the surface of a sample with plasma sprayed hydroxyapatite coating covered with bone-like apatite spherulitic crystals after 7 days of soaking in SBF. Bar = 100 um

Fig. 5: Electron microscope image of a sample with plasma sprayed hydroxyapatite coating after 14 days of soaking in SBF showing a continuous layer of calcium phosphate agglomerates. Bar = 100 um

Fig. 6: SEM-EDS analysis of layers formed on the surface of a sample with plasma sprayed hydroxyapatite coating after 7 days of soaking in SBF

Fig. 7: Thin-film-X-ray diffraction patterns (a) of plasma sprayed hydroxyapatite. (b) plasma sprayed hydroxyapatite exposed to SBF for 14 days. (c) bone apatite

(a) a sand-blasted CP titanium implant (b) a CP titanium implant with plasma sprayed hydroxyapatite coating - implanted by the press-fit procedure - model A (magnification 320x; toluidine blue stain) (c) a sand-blasted CP titanium implant
Conclusion

This study demonstrated that plasma sprayed hydroxyapatite (PSHA) coated implants exhibit greater tolerance than sand-blasted CP titanium implants to unfavorable conditions during healing such as gaps at the interface or the primary instability of the implant. PSHA coated implants showed high percentage of bone contact if gaps were smaller than 1mm even when non press-fit insertion was used. In the case of sand-blasted CP titanium implants gap-filling with fibrous tissue was observed over the greater part of the surface of gaps with a depth of 0,3mm. Direct contact of sand-blasted CP titanium implants with bone was achieved only when press-fit implantation was used. It can be concluded that more precise surgical placement is needed for CP-titanium implants than for PSHA coated ones.

Acknowledgement

The study was supported by the Grant Agency of the Czech Republic under project No: 106/97/0985 and by the Internal Grant Agency of Ministry of Health under the No.: NK4745-3

Literature

Bone-like apatite formation on titanium and silica glass

J. Strnad¹, A. Helebrant², J. Hamáčková³


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If a material induces apatite formation in SBF (simulated body fluid) it is likely to exhibit bioactive behavior and form a stable interface with bone tissue. Apatite formation kinetics can help to estimate bioactivity of material and also to describe the formation mechanism. In present study, apatite formation on the surface of titanium and silica glass was investigated using microscope image analysis and by weighing the precipitated apatite. The titanium surface was chemically treated in sodium hydroxide before the interaction with SBF.

A microscope image analysis of the sample surface was used to describe apatite formation kinetics during the initial stages of precipitation, when isolated apatite spherulites were formed. These are parameters evaluated by the image analysis: mean and maximum diameter of apatite spherulites, their number per unit area and the percentage of apatite covered surface. The time dependence of maximum diameter was the most reproducible one and it could be used to determine the apatite growth rate and to estimate the induction times for different supersaturations of the SBF solution.

It was found that not immediate but continuous nuclei formation occurs. It was shown that it is possible to characterize the initial stages of apatite precipitation by the induction times evaluated using extrapolation of the linear time dependence of spherulite maximum diameter. This method was used to characterize apatite formation on chemically treated titanium and silica glass.

The induction times could be used to determine the apparent interfacial energy \( \Sigma_{CL} \) for nucleation according to modified Gibbs-Thomson equation:

\[
\ln \left( \frac{1}{t} \right) = \ln(A) - \frac{16 \pi \Sigma_{CL} v^2}{3kT} \left( \frac{1}{(\ln S)^2} \right)
\]

where \( t \) is induction time, \( v \) -molecular volume, \( k \) –Boltzman constant, \( T \) –absolute temperature and \( S \) –supersaturation ratio.

The obtained apparent interfacial energy value for apatite precipitation on chemically treated titanium was 3.7 mJ/m². This value could serve as a measure of a substrate’s ability to induce apatite formation and thus it could help to estimate the bioactivity of a material on the basis of in vitro tests.

Because during the interaction of titanium with SBF no significant release of any component occurs, the apatite growth rate can be evaluated by weighing the precipitate. It is possible to describe the growth rate using equation \( R = k(S-1)^n \), in which \( S \) is the supersaturation and \( n \) is the effective order of reaction indicating the rate controlling mechanism. The calculated value of \( n=2.8 \) indicates that the rate controlling process of apatite formation on the surface of chemically treated titanium is polynucleation.

The apatite precipitation on the surface of silica glass occurred only in SBF solution with increased supersaturation ratio.

Modelling of stress and deformation distribution around endosteal implants

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Quintesenz, Vol. 8, No. 12, 1999

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The study uses a finite elements analysis to determine of the stress and deformation distribution in bone around dental implant. Two types of implant shape were considered in the calculation – screw and cylinder, 12mm in length and 3,6mm in diameter. For each implant type three models of bone-implant interface were considered – 1. osseointegrated bioactive implant; 2. direct contact of a bioinert material with bone without bonding; 3. nonosseointegrated bioinert implant without a direct contact to bone (fibrous tissue encapsulation). For modelling we used a jaw segment corresponding to the position of a second molar. The implant was loaded by systematic forces. By calculation, equivalent stress and deformation values in the bone around the cervical section of the implant were determined. The model of osseointegrated bioactive implant (1) exhibited significantly lower maxima compared to that in the other two model types. Regarding the shape of the implant, a screw-like design proved to be more suitable than a cylindrical one.
Guided bone regeneration in the pre-implantation phase

P. Poleník


Bone regeneration is a complex procedure that plays an important role in many fields of medicine, including dentistry. It is mainly connected to some kind of defect of the bone, of variable origin, and is a component of the complex regeneration of periodontal tissues. Intracranial bone regeneration is important in planning in-bone implant insertion.

For successful bone regeneration procedures need to follow various basic requirements. There are several factors that might, under certain conditions and in certain situations, have a negative impact on bone regeneration. For example, tissue regeneration within an area of periodontal disease is likely to be affected by bacterial infection. To promote bone regeneration in the area not in direct contact with the periodontal disease, the bone regeneration site can be isolated. On the other hand, this limits the scope of cells that can contribute to the bone regeneration process only to those cells located in adjacent bone marrow, whereas in a periodontal site multipotent mesenchymal cells of periodontal ligaments are also available.

The regeneration procedure requires a suitable matrix to provide the skeleton for future bone tissue and a suitable environment for cells involved in the regeneration process. The structure resembling a labyrinth needs to be of the size of adhering cells and enable their interaction. The same material should prevent unwanted cells from entering the site. The optimal pore size should be between 200 – 400 μm, as this size reflects the average size of osteon in humans (which is ca. 230 μm). The surface topography of the material can also affect the proliferation activity of osteoblasts. Such a mechanism is especially important in resorbable materials in which the characteristics of the material undergoes continuous change.

Another vital characteristic of the matrix is its ability to induce vascularization of the regeneration site. The material should be resorbable to be able to subsequently give space to the newly growing bone. The material components themselves may also participate in creating the structure of the newly growing bone. An important role in the bone regeneration process is also played by hormonal factors - these perform an irreplaceable role in governing all ongoing reactions and are especially responsible for their coordination in time. The cascade process of bone regeneration is mainly run by transforming and growth factors. The transforming factors, in fact, are responsible for a reincapulation of the cell transformation events that occur during embryonic development, the growth factors being engaged in further cell production and differentiation. In connection with bone regeneration are the functions of the transforming growth factor-β (TGF-β), bone morphogenetic proteins (BMP), fibroblast growth factor (FGF), platelet-derived growth factor (PDGF), epidermal growth factor (EGF), insulin-like growth factor (IGF) and vascular endothelial growth factor (VEGF).

Transforming growth factors play an irreplaceable role in the proliferation and differentiation of osteoprogenitor cells and osteoblasts. The bone morphogenetic proteins induce the transformation of mesenchymal cells into cells forming bone tissue. Platelets are a suitable source of growth factors – after degranulation they release a number of the growth factors named above. The procedure is not technically complicated and, given certain preconditions, can be performed in any dental surgery.

The presented study aimed at verifying the function of a matrix consisting of porous β-tricalciumphosphate and platelet-derived growth factors in the reconstruction of alveolar ridge defects prior to implant placement into the bone.

Material and methods

Twelve patients (32-48 years) were treated for horizontal defects of the alveolar ridge caused either as a result of trauma or periodontal disease. In all patients implants were subsequently inserted into the bone. At the time of implant insertion, some bone material was obtained in order for an histological survey to be undertaken to evaluate the efficacy of the regenerative system used.

Bone defects were treated using porous β-tricalciumphosphate (PORESORB–TCP Lasak Ltd., Czech Republic). The structure of PORESORB–TCP is similar to the structure of spongiform bone with a macroporosity of ca. 100 μm and a microporosity of ca. 5 μm. Macroporosity is important for the settlement of regenerative cells, whilst microporosity is important for the adherence of protein mediators and for regenerating vascularization. Prior to surgery, from each patient 20 ml of anticoagulated venous blood was taken; this was centrifuged in two steps to obtain platelet-rich plasma (PRP) that was further treated by adding calcium chloride and thrombin. In this way, platelet alpha-granules are degraded and growth factors released that are then ready to be used for the reparation and regeneration of integral tissue. The gel that is produced is a suitable carrier for the powdered PORESORB–TCP and when both are mixed together a material that is easily applied is obtained for use in bone defects.
A fully-mobilised mucoperiosteal flap was raised at the site of the defective alveolar ridge under local anaesthesia. The surface of the defected bone was then perforated to allow better contact between the augmentation site and bone marrow. The gel-like material with added PORESORB–TCP granules was used to repair the bone defect, i.e. to reconstruct the bone shape reaching a slightly larger volume. A collagen membrane was employed over the augmentation site to exclude unwanted cells from the defect area. Subsequently the mucoperiosteal flap was sutured back. The operated site was protected against mechanical damage and during the healing period an intensive antibacterial regime was ensured. Nine months later, an X-ray was taken and implants placed into the bone. Samples of the newly-formed alveolar bone were obtained for histological evaluation using a 2 mm drill. The bone samples were stained using hematoxylin-eosin and Goldner trichrome.

Results

The healing of soft tissue following augmentation was without complications and the augmented site fully integrated. Radiographic evaluation nine months later showed that the structure of the newly-formed bone was identical to standard bone and also the torque applied to the bone tissue during preparation for implant placement was equivalent (Fig. 1). The histological assessment of bone specimen material showed, to a complete extent, the formation of lamellar bone with remains of PORESORB–TCP granules (Fig. 2). Also vascularization of the bone tissue was normal. Figure 3 shows radiograph of the operated site with a healed-in implant.

Summary

The presented results show that porous β-tricalciumphosphate together with platelet-derived growth factors is able to ensure the regeneration of good-quality bone tissue. The use of platelet-rich plasma (PRP), and thus platelet-derived growth factors, has been described in many papers with varying results. Some papers present a positive impact of PDGF on tissue regeneration, others no effect at all. The most likely explanation for such variable findings is a differing procedure for the separation of PDGFs and also the characteristics of the materials used as a carrier for the cellular and humoral components. The most important characteristic is the ability of the material to adsorb proteins. In an ideal situation the adsorbed proteins should be subsequently gradually released; in this way the dynamics of the growth factors and their participation in individual reactions is regulated. Despite the fact that for therapeutic treatment the platelet-derived growth factors are supplied in quantities highly in excess of the physiological need, it is important to ensure that they are present for a sufficient time at a given site, as otherwise their effect might be negligible. PORESORB–TCP, the material that we have used in this study, is capable of ensuring proper PDGF functioning and thus contributes to new bone formation. In addition, PORESORB–TCP is a material that is resorbable and nine months following its application was almost undetectable in the histological evaluation. In combination with the membrane technique, unwanted cells did not interfere with the healing process. As the placed implants have been functional for several years now, the mechanical bone quality can also be considered to be satisfactory.

Porous β-TCP and platelet rich plasma (PRP) in treatment of periodontal defects

P. Poleník

J. Int. Acad. Periodontol., Vol. 6, 2004, No. 2

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The aim of the present investigation was to evaluate the clinical and histological results obtained in the therapy of periodontal defects by composite graft formed by calcium sulphate, β-TCP (Poresorb) and platelet rich plasma (PRP). In10 patients with advanced periodontitis, ten 3-wall periodontal defects were treated. Autologous PRP was prepared immediately before periodontal surgery. Probing dept (PD), clinical attachment level (CAL), bleeding on probing (BOP) and X-ray examinations were recorded before surgery and 6 and 24 months after treatment. After 6 months the histological examination of regenerated bone was performed. After 6 months, reduction of pocket depth from 8.1 to 3.8 mm was observed. Nearly the same value was measured after 24 months. The attachment gain after 6 months reached the average value of 4.2 mm without some substantial differences after 24 months. Radiologically evaluated bone fill was 4.6 mm after 6 months and 4.4 mm after 24 months. Bleeding in probing (BOP) was 54% at baseline and 36% and 34% after 6 months and 24 months respectively. Histological examination after 6 months showed complete resorption of calcium sulphate particles, advanced resorption of β-TCP and formation of regular alveolar bone. The clinical results also demonstrate that calcium sulphate and β-TCP in combination with PRP acts as a very good osteoconductive material for osteoinductive mechanisms and they are suitable materials for this type of GTR procedures.
β-TCP as a scaffold for tissue engineered bone

P. Poleník

J. Clin. Periodontol., Vol. 30, 2003, suppl. 4, s. 51

Clinic of Dental Medicine, Medical Faculty, Charles University, Czech Republic

The use of platelet rich plasma (PRP) can enhance tissue regeneration. Macro and micro-porous beta-tricalcium phosphate (PORESORB-TCP, Lasak, Czech Rep.) was used as a scaffold for periodontal bone defects treated by PRP. The surface of this material was formed in order to enable the best attachment of proteins. 27 patients with adult periodontitis and alveolar bone defects were treated by combination of PORESORB and PRP. Plaque index, pocket depth, attachment level and extent of bone fill were recorded before treatment and after six and twelve months. Six specimens from the area of regeneration were taken and histologically prepared by HE and Goldner trichrome. After 6 months reduction of pocket depth from 7.6 to 3.2mm was observed. Nearly the same value was recorded after 12 months. The attachment gain after 6 months reached the average value of 3.2 mm without any substantial difference after 12 months. Radiologically evaluated bone fill revealed the average value of 4.1 mm after 6 months and 4.0 mm after 12 months. The level of hygienic conditions was did not vary during the experimental period with values of PlI 0.72 before treatment and 0.65 and 0.62 after 6 and 12 months respectively. Histological examinations revealed very good integration of material inside the periodontal and defect, and also osteogenic activity in the central part of its volume. This study indicated that PORESORB acts as a very good scaffold for tissue engineered bone.
In vivo behaviour of the synthetic porous hydroxyapatite prepared by low temperature processing and comparison with deproteinized bovine bone

M. Strnadová, Z. Strnad, P. Sponeť, J. Jirošová, J. Strnad


Abstract

The aim of this study was to evaluate the osteoconductive properties of synthetic porous hydroxyapatite prepared by low-temperature microwave processing OssaBase®-HA (SPHA) in comparison with biological apatite, non-sintered deproteinized bovine bone Bio-Oss® (DBB). The materials were implanted into the bone sockets of the tibia of Beagle dogs for 3 and 6 months. The bone response to granules of the materials of similar sizes was compared. Histological analysis of the specimens with histomorphometry was performed at different times after in vivo implantation. Based on the histological analysis, the level of bone formation in the spaces between the implanted granules and through the interconnected pores of both implanted materials within a cortical region was significantly higher (bone area ingrowth 72–85%) than within a cancellous bone site (bone area ingrowth 16–28%) at three and six months after implantation. According to our study, the bioactive and osteoconductive properties (bone implant contact and bone area ingrown) of the synthetic porous hydroxyapatite are very high and comparable with the biological apatite, non-sintered deproteinized bovine bone. The favourable influence of the high specific surface area and carbonate content of the synthetic, porous hydroxyapatite on bone formation was confirmed.

Introduction

The bone-graft substitutes include a spectrum of products that have various effects on bone healing. Calcium phosphate biomaterials are frequently used for bone defect reconstructions, because of their bone-like chemical composition. They are bioactive and highly osteoconductive. These properties are dependent on the chemical composition of the material, the technique of fabrication, the level of crystallinity, the morphological and structural parameters that have a direct or indirect effect on the cells’ migration, and the adhesion, proliferation and differentiation with subsequent new bone formation. Hydroxyapatite (HA) seems to be one of the most important calcium phosphate materials for bone regeneration. In contrast to stoichiometric HA $\text{Ca}_{10}(\text{PO}_4)_6(\text{OH})_2$ with a Ca/P molar ratio of 1.67, biological hydroxyapatites, which contain minor substituents in their structure ($\text{CO}_3^{2-}, \text{Cl}^-, \text{Mg}^{2+}, \text{K}^+, \text{Na}^+$), are usually calcium-deficient with a Ca/P molar ratio lower than 1.67 and $\text{CO}_3^{2-}$ substitutes primarily for $\text{PO}_4^{3-}$ groups (B-type substitution). The synthetic porous carbonated HA prepared by low-temperature synthesis resembles the structure and chemical composition of the biological bone HA and can successfully substitute for non-sintered bovine apatite prepared by the deproteinisation of bovine bone with benefit of absolute elimination of the risk caused by residual antigenic proteins of xenogenic bone.

Materials and Methods

The synthetic nanostructured hydroxyapatite powder was prepared by the reaction of calcium hydroxide aqueous solution and phosphoric acid using microwave technology. The powders were structured during drying without using any porogenic additives. The porous granules (SPHA) OssaBase®-HA (LASAK Ltd., Czech Republic) of the tested materials measuring 0.6–1.0 mm in diameter were implanted into tibia defects of 3.7 mm diameter and 8 mm depth. The deproteinized bovine bone (DBB) commercially available as Bio-Oss® (Geistlich Pharma AG, Wolhusen, Switzerland) was used as control material.

X-ray diffraction analysis (XRD 3000P, Seifert, Germany), chemical analysis, differential thermal analysis, SEM- scanning electron microscopy (Vega II-LSU, Tescan, CR) and gas adsorption (ASAP 2010 M Micromeritics, USA), mercury porosimetry (Autopore III Micromeritics, USA), optical microscopy (Olympus BX60) were used for the characterisation of the physical-chemical materials properties. Carbonate substitution in the structure of samples was distinguished by infrared spectroscopy.

An in vitro cytotoxicity test and the preclinical evaluation of this material were performed. In vivo tests were carried out in the tibiae of beagle dogs. In the cancellous bone sites, the bone implant contact (BIC%) was determined using histomorphometry after three and six months postoperatively. In the cortical bone region, the bone ingrowth in the spaces between granules (BAI%-bone area ingrowth) was evaluated using the quantification of the material/new bone/soft tissue amount after three and six months postoperatively.

Statistical analysis

The values were presented as mean ± standard deviation (SD). The Mann-Whitney U test was used to determine differences between the implanted materials. A P value of less than 0.05 was considered significant.

Results and Discussion

Carbonate substitution in the structure of samples was distinguished by infrared spectroscopy. A-type carbonate substitution was detected by the presence of a band observed at 1545 [cm$^{-1}$] and B-type at wave number 1415 [cm$^{-1}$]. Infrared spectra were recorded on a Nicolet 7600 FTIR spectrophotometer (Thermo Nicolet Instruments Co., Madison, USA). A high structural and morphological resemblance of the synthetic hydroxyapatite (SPHA) to nanocrystalline non-sintered deproteinized bovine bone (DBB) was found (Table 1).
The X-ray diffraction patterns of the synthetic hydroxyapatite (SPHA) and deproteinized bovine bone (DBB) showed apatitic patterns (PDF database card 9-342) with low crystallinity. The X-ray diffraction pattern of deproteinized bovine bone (DBB) has been almost identical to that of synthetic HA (SPHA). The diffusion character of the X-ray diffraction lines indicates the small size of crystals. The differences of the crystallinity between low-temperature and high-temperature hydroxyapatites were determined (Fig. 1).

The porous structure of the samples was studied by SEM - scanning electron microscopy. The size of the macro pores of the synthetic HA (PSHA) and deproteinized bovine bone (DBB) was evaluated. The macro pore size was more than 100 μm for both tested materials (Fig. 2). The micro pores and high specific surface of the tested materials allows fast penetration of body fluids and the adhesion of proteins.

Bone healing in the cortical bone sites—histomorphometry

In the cortical bone sites, the amount of new bone formation, remaining material volume (OssaBase (SPHA), BioOss (DBB), and amount of soft tissue (marrow spaces) were determined using histomorphometry after three and six months postoperatively (Fig. 4). No evidence of giant cell foreign body reaction was found in any biopsy. The bone formation in the spaces between the implanted granules within the cortical region (BAI % -Bone Area Ingrowth) at 6 months had reached 80.49 %, and 85.90 % in OssaBase®-HA and Bio-Oss®, respectively. However, there was no significant difference in BAI % (Fig. 3) The new bone tissue was viable, organised into Haversian systems and completely mineralized at 6 months.

Table 1: Physical-chemical characteristics of the implanted materials

<table>
<thead>
<tr>
<th></th>
<th>(SPHA) (OssaBase®-HA)</th>
<th>(DBB) Bio-Oss®</th>
</tr>
</thead>
<tbody>
<tr>
<td>Size of granules [mm]</td>
<td>0.6-1</td>
<td>0.6-1</td>
</tr>
<tr>
<td>Size of macropores [μm]</td>
<td>&gt;100</td>
<td>&gt;100</td>
</tr>
<tr>
<td>Mean size of micropores [nm]</td>
<td>5</td>
<td>12.4</td>
</tr>
<tr>
<td>Specific surface area [m²/g]</td>
<td>78.3 ±0.34</td>
<td>74.4± 0.27</td>
</tr>
<tr>
<td>Porosity [%]</td>
<td>83</td>
<td>54.8</td>
</tr>
<tr>
<td>Crystal size [nm]</td>
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<td>30</td>
</tr>
<tr>
<td>Ca/P molar ratio</td>
<td>1.65</td>
<td>1.57</td>
</tr>
<tr>
<td>Rel CO₃⁻⁻ 'A' (11545/11041) *</td>
<td>1.10⁶</td>
<td>1.9.10⁶</td>
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<tr>
<td>Rel CO₃⁻⁻ 'B' (11420/11041) *</td>
<td>0.023</td>
<td>0.051</td>
</tr>
</tbody>
</table>

* Values expressing a relative carbonate content using phosphate band at 1041 cm⁻¹ as standard

Bone regeneration materials

![Bone regener.png](attachment://Bone regeneration materials.png)
Bone healing in the cancellous bone sites—histomorphometry
The bone implant contact of the tested and control materials SPHA and DBB was >70% after three months and increased to >80% after six months independently of the granules’ size (Fig. 5). These histomorphometrical findings indicate that the in vivo osteoconductivity of SPHA was similar to the DBB at both three and six months. The values were presented as mean ± SD. The Mann-Whitney U test was used to determine differences between the implanted materials. A P value of less than 0.05 was considered significant.

The thickness of the new bone layer formed by osteoconduction on the surface of granules was relatively unified and ranged from 100 to 300 [μm]. The bone bridging up to 2000 [μm] in size was observed but the spaces between granules were not completely filled by bone tissue (Fig. 6). The bone formation in the spaces between the implanted granules within the cancellous region (bone area ingrowth as BAI%) had reached 22 ± 6% without significant differences between different implanted materials and time points.

Summary
The new porous synthetic bone regeneration low-temperature material based on hydroxyapatite with micro and macro porous interconnected structure indicates high osteoconductive properties comparable with non-sintered deproteinized bovine bone (DBB). This material is very perspective material suitable for the bone regeneration due to its high structural and morphological resemblance to biological apatite.

This synthetic material seems to be a successful bone replacement material in bone surgery with benefit of absolute elimination of the risk caused by residual antigenic proteins of xenogenic bone.

Acknowledgement
This work was supported by the Department of Industry and Trade of the Czech Republic under the Project No. FR-TI3/587.
In vivo behaviour of low-temperature calcium-deficient hydroxyapatite: comparison with deproteinised bovine bone

P. Šponer, M. Strnadová, K. Urban

International Orthopaedics, 2010

Abstract

This study aims to evaluate in detail the biological osteoconductive properties of the low-temperature synthetic porous calcium-deficient hydroxyapatite and to compare it with the biological apatite. Bone reactions to granules of similar sizes of the low-temperature hydroxyapatite and commercially available non-sintered deproteinized bovine bone were compared. Two different temperatures were used to fabricate two batches of newly developed porous hydroxyapatite with different carbonate groups content and specific surface area. The histological analysis of specimens with histomorphometry was performed at different time after in vivo implantation. Based on histological analysis, the level of bone formation in the spaces between the implanted granules and through the interconnected pores of all implanted materials within a cortical region (bone area ingrowth 16–28 %) at three and six months after implantation. Within the cancellous bone site, bone coverage of the implanted material at six months was significantly higher in hydroxyapatite material formed using low-temperature synthesis and subsequent processing at 150°C than in hydroxyapatite scaffold developed using low-temperature synthesis with subsequent processing at 700°C or deproteinized bovine bone. According to our study, the bioactive properties of the low-temperature calcium-deficient hydroxyapatite are comparable with the biological apatite. The favourable influence of a high specific surface area of a low-temperature calcium-deficient hydroxyapatite on in vivo bone formation was emphasized.

Introduction

There is a need for treatment of bone defects throughout the whole spectrum of orthopaedic surgery—to replace bone substance which has been lost due to congenital disorders, traumatic and inflammatory events, for reconstructive surgical procedures in hip arthroplasty revision surgery and surgery for bone tumours and tumour-like lesions. Bone remodelling allows the skeleton to achieve the optimal architecture dependent on applied load. Bone tissue regeneration and repair processes can be effective in small defects (under 60 cm² of a low-temperature calcium-deficient hydroxyapatite on in vivo applied load. Bone tissue regeneration and repair processes can be effective in small defects (under 60 cm² of a low-temperature calcium-deficient hydroxyapatite). The porous scaffolds allowed the cells to migrate, and the adhesion, proliferation and differentiation with subsequent new bone formation. Hydroxyapatite (HA) seems to be one of the most important calcium phosphate materials for bone substitution. In contrast to stoichiometric calcium HA \( \text{Ca}_3(\text{PO}_4)_2(\text{OH})_2 \) with a Ca/P molar ratio of 1.67, biological hydroxyapatites contain minor substituents in their structure (\( \text{CO}_3^2-, \text{Cl}^-, \text{Mg}^{2+}, \text{K}^+, \text{Na}^+ \)), are usually calcium-deficient with a Ca/P molar ratio lower than 1.67 and \( \text{CO}_3^2 \) - substitutes primarily for \( \text{PO}_4^{3-} \) groups (B-type substitution). The synthetic porous B-type carbonated HA prepared by low-temperature synthesis (precipitation) resembles the structure and chemical composition of the biological bone HA and can successfully substitute non-sintered bovine apatite prepared by deproteinisation of bovine bone with absolute elimination of risk caused by residual antigenic proteins of xenogeneic bone.

To date there have been no reports which compare the in vivo bioactivity of the synthetic porous carbonated HA with non-sintered bovine apatite after implantation into defects of long bones. In this study, we investigated two different types of scaffolds; bone reaction to granules of similar sizes of the new low-temperature calcium-deficient HA and commercially available deproteinised bovine bone (DBB) were compared. Two different temperatures were used to create two batches of newly developed porous HA with different carbonate groups content and specific surface area. The aim of this study was to evaluate the biological osteoconductive properties of the low-temperature synthetic porous calcium-deficient HA and to compare it with biological apatite—non-sintered deproteinised bovine bone.

Materials and methods

Implant materials

The porous HA used in this study was prepared using low-temperature synthesis and subsequent processing at 150°C (HAp1) or 700°C (HAp2) with hydrogen peroxide as a pore-creating additive (Fig. 1). Both synthetic low-temperature HA bioceramics had different material properties and were produced by Lasak Praha (Prague, Czech Republic) (Table 1). The third implanted biomaterial was DBB Bio-Oss® (Geistlich Pharma AG, Wolhusen, Switzerland). The porous granules of the tested materials measuring 0.6–1.0 mm in diameter were implanted into tibial defects of 3.7 mm diameter and 8 mm depth.
Material characteristics

The phases present were identified by X-ray diffraction analysis (XRD 3000P, Seifert, Germany). The X-ray diffraction patterns of the synthetic samples (HAp1, HAp2) and DBB showed apatitic patterns (PDF database card 9-342) with low crystallinity. The diffuse characteristic of X-ray diffraction peaks increases with increasing carbonate content. The X-ray diffraction pattern of DBB has been almost identical to that of HAp1 (Fig. 2).

Carbonate substitution in the structure of samples was distinguished by infrared spectroscopy. A-type carbonate substitution was detected by the presence of a band observed at 1,545 cm⁻¹ and B-type at wave number 1,415 cm⁻¹. Infrared spectra were recorded on a Nicolet 7600 FTIR spectrophotometer (Thermo Nicolet Instruments Co., Madison, USA). The specific surface area of the implanted materials was measured using BET method (ASAP 2010 M, Micromeritics, USA). The porosity was determined by mercury intrusion porosimetry (Autopore III Micromeritics, USA), standard light microscopy.

<table>
<thead>
<tr>
<th>Characteristics</th>
<th>HAp 1</th>
<th>HAp 2</th>
<th>DBB</th>
</tr>
</thead>
<tbody>
<tr>
<td>Size of granules (mm)</td>
<td>0.6–1</td>
<td>0.6–1</td>
<td>0.6–1</td>
</tr>
<tr>
<td>Size of macropores (μm)</td>
<td>&gt;100</td>
<td>&gt;100</td>
<td>&gt;100</td>
</tr>
<tr>
<td>Mean size of micropores (nm)</td>
<td>5</td>
<td>20</td>
<td>12.4</td>
</tr>
<tr>
<td>Specific surface area (m²/g)</td>
<td>78.3 (±0.34)</td>
<td>25.4 (±0.11)</td>
<td>74.4 (±0.27)</td>
</tr>
<tr>
<td>Porosity (%)</td>
<td>83</td>
<td>74</td>
<td>54.8</td>
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<tr>
<td>Crystal size (nm)</td>
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<td>Ca/P molar ratio</td>
<td>1.65</td>
<td>1.66</td>
<td>1.57</td>
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<tr>
<td>Rel CO₃⁻ &quot;A&quot; (I₁₅₄₅/I₁₀₄₁) x</td>
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<td>2.₁₀⁻⁶</td>
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</tr>
<tr>
<td>Rel CO₃⁻ &quot;B&quot; (I₁₄₂₀/I₁₀₄₁) x</td>
<td>0.₀₂₃</td>
<td>0.₀₀₆</td>
<td>0.₀₅₁</td>
</tr>
</tbody>
</table>

Table 1: Physical and chemical characteristics of the implanted materials

* Values expressing a relative carbonate content using phosphate band at 1,041 cm⁻¹ as standard

Fig. 1: Scanning electron micrograph of the tested material HAp1 showing its macro- (a) and nanostructure (b)

Fig. 2: The X-ray diffraction patterns of DBB, HAp1 and HAp2
Bone regeneration materials

(605x808, Japan) with image analysis and scanning electron microscopy (Vega II-LSU, Tescan, Czech Republic). The phosphorous content was analysed colorimetrically. The complexation titration method was used for the calcium determination. High structural and morphological resemblance of the synthetic biomaterials (HAp1, HAp2) to nanocrystalline non-sintered biological hydroxyapatite (DBB) was found (Table 1).

Implantation procedure

Twenty-seven implantations were performed on two-year-old Beagle dogs weighing between 12 and 16 kg. The tested materials were implanted into the tibia of the hind legs of the dogs. Each material composition (HAp1, HAp2, DBB) was applied in three implantations for a period of three months and in six implantations for a period of six months. The study was performed in agreement and with the approval of the competent ethical committee.

The animals were anaesthetised with Narkamon 5% (15 mg/kg) and Rometa 2% (2 mg/kg) and maintained with a mixture of oxygen, nitrous oxide and halothane after administration of Atropin by way of premedication. The left and right hind limb were clipped, scrubbed and draped for aseptic surgery. An incision was made adjacent to the right tibial diaphysis and the soft tissue was dissected to reveal the underlying bone. Defects, 3.7 mm in diameter and 8 mm in depth, were created within the bone using drill bits and a battery drill. The defect sites were subsequently irrigated with sterile saline solution and the granules of the tested materials were implanted into the defects. After implantation, the incisions were closed using interrupted suturing. The process was repeated on the opposite hind leg with granules of a different composition. Analgesics (Novalin 16 mg/kg/day) were administered for three days by intramuscular injections; no antibiotics were given.

Histological evaluation and histomorphometry

The dogs were put to sleep at three and six months after surgery by an intravenous overdose of barbiturates and both tibiae were dissected. Radiographs of the tibiae were obtained perpendicular to the direction in which the implants were placed. The trimmed tibiae containing the implants were placed in 10% w/v of paraformaldehyde solution for fixation. Tibial bone defects filled with the tested materials were separated as tissue blocks. The samples were dehydrated using increasing concentrations of alcohol and embedded in an acrylate resin. The EXACT technology was used for preparation of the longitudinal undecalcified thin cut sections. A total of three sections were cut perpendicular to the longitudinal axis of the tibia per implant area. The sections with a thickness of 30–50 μm were stained at 60°C for one hour with 0.9% w/v toluidine blue solution for histology and histomorphometry. The amount of newly-formed bone was assessed using an Olympus BX 60 optical microscope and quantified using the image analysis software Image-Pro Plus 5.1. In the cortical bone region, the bone ingrowth in the spaces between granules (BAI%- bone area ingrowth) for all implant area.

Image-Pro Plus 5.1. In the cortical bone region, the bone ingrowth in the spaces between granules (BAI%- bone area ingrowth) for all implant area.

Bone healing in the cortical bone sites—histomorphometry

In the cortical bone sites, the amount of new bone formation, remaining material volume (HAp1, HAp2, DBB), and amount of soft tissue (marrow spaces) were determined using histomorphometry after three and six months postoperatively. Histological evaluation indicated that bone bridging had occurred between the new bone and the surface of the implanted granules with no evidence of fibrous encapsulation for all implant compositions and time points (Fig. 3). No evidence of giant cell foreign body reaction was found in any biopsy.

The newly-formed bone tissue in the cortical sites of the HAp1 implants had reached 36.07% (±8.04) of the total cortical defect after three months in vivo; in HAp2 and DBB the newly-formed bone was 39.99% (±8.70) and 40.08% (±7.35) of the total cortical defect, respectively (Fig. 4a).

Bone formation in the spaces between implanted granules of HAp1 within the bone ingrowth in the spaces between granules (BAI%- bone area ingrowth) for all implant area/new bone/soft tissue amount. In the cancellous bone region, the osteoconductive properties of materials were assessed using the coverage of the material surface (bone implant contact as BIC%).

Statistical analysis

The values were presented as mean ± standard deviation (SD). The Mann-Whitney U test was used to determine differences between the implanted materials. A P value of less than 0.05 was considered significant.
the cortical region (BAI % -Bone Area Ingrowth) had reached 73.2% (± 11.6); in HAp2 and DBB the newly-formed bone filled 79.2% (± 15.8) and 79.9% (± 7.3), respectively. At three months in vivo, the newly-formed bone between the implanted particles connected them into a mass of mostly mineralised tissue. The amount of remaining material (HAp1, HAp2, and DBB) within the cortical sites of implantation decreased in the period between three and six months after the operation (a relative decrease of 3.2%, 5.4%, and 12.5%, respectively). At the same time the amount of new bone formation increased (a relative increase of 6.2%, 0.5%, and 13.6%, respectively).

The bone formation in the spaces between the implanted granules within the cortical region (BAI % -Bone Area Ingrowth) had reached 80.49%, 72.46%, and 85.90% in HAp1, HAp2, and DBB, respectively. However, there was no significant difference in BAI %. The new bone tissue was viable, organised into Haversian systems and completely mineralised at six months.

Bone healing in the cancellous bone sites—histomorphometry

In the cancellous bone sites, the bone implant contact (BIC%) was determined using histomorphometry after three and six months postoperatively. The bone implant contact of the tested materials (HAp1, HAp2 a DBK) was >70% after three months and increased to >80% after six months independently of the granules’ size (Fig. 4b). These histomorphometrical findings indicate that the in vivo osteoconductivity of HAp1 material was greater than that for HAp2 and DBB at both three and six months.

The thickness of the new bone layer formed by osteoconduction on the surface of granules was relatively uniform and ranged from 100 to 300 μm. The interconnections which had a size above 100 μm were partially or completely filled with vital bone. The bone bridging up to 2000 μm in size was observed but the spaces between granules were not completely filled by bone tissue (Fig. 5). The bone formation in the spaces between the implanted granules within the cancellous region (bone area ingrowth as BAI%) had reached 22 ± 6% without significant differences between different implanted materials and time points.

Discussion

It remains a great challenge to design the ideal bone graft substitute that emulates nature’s bone structure and function. Consequently, the first generation of biomaterials consisted of bioinert materials which reduced the immune response and foreign body reaction to a minimum. Between 1980 and 2000 the field of development began to shift to bioactive and resorbable biomaterials. Bioactive components could interact with the biological environment to enhance the tissue/surface bonding. Resorbable materials undergo a progressive degradation while new tissue regenerates and heals. The third generation of biomaterials are able to stimulate specific cellular response at the molecular level, and the bioactivity and biodegradability concepts are combined. A temporary three-dimensional porous structure that stimulates the cells’ invasion, adhesion, proliferation, differentiation and extracellular matrix production is required for clinical applications. DBB, which was used in our study as a control, is calcium-deficient carbonate apatite, chemically and physically identical to human bone with high osteoconductive properties. The granules provide an ideal scaffold for new bone formation. DBB is a xenogeneic material from which all organic components have been removed. The testing of materials of bovine origin was recommended despite a complete absence of proteins to guarantee that they are free of infectious prion particles. Xenograft bone substitutes have been successfully applied in stomatology and traumatology. Histological evaluations have shown that DBB is a highly osteoconductive material with no adverse effects, such as an inflammatory cell infiltrate or foreign body response even at long-term follow-up. The presence of DBB particles perfectly integrated into the bone architecture but without complete resorption could be explained by the influence of a microenvironment at the bone–xenograft interface on the recipient osteoclastic activity. The chemical composition, solubility, and surface morphology were reported as significantly important factors regulating osteoclast activity. Integrins play a critical role as mediators of osteoclastic adhesion to the surface of the implanted material with subsequent acid secretion into the resorption lacuna. The erosion of the material surface via a mineral release causes a high concentration of Ca ions in this area that could inhibit osteoclastic activity. HA (Ca₃(PO₄)₂(OH)), β-tricalcium phosphate (TCP) (Ca₅(PO₄)₃), their derivatives and their combinations are the most commonly used ceramic materials in bone surgery. The requirements for an ideal synthetic bone substitute appear deceptively simple. It should be a porous matrix with interconnecting porosity that promotes rapid bone ingrowth. The microstructure of an implant should allow a formation of environment optimal for cellular settlement, growth and function. At the same time, the ideal matrix should possess a sufficient strength to prevent its crushing under physiological loads during integration and healing. The pore size and interconnection pathway play a crucial role in ceramic biomaterials. A minimal pore size of ~100 μm appropriate for bone growth into ceramic has been described. The porosity must be interconnected to allow the protein adhesion, ingrowth of cells and formation of effective blood supply that is necessary for new bone formation and subsequent bone remodelling. The ideal material should form a secure bond with the surrounding tissues by encouraging new cells to grow and penetrate. The implantation of HA ceramics with porosities range from 35% to 75%, and pore sizes between 50 μm and 400 μm have been reported in the literature. There are few reports about quantitative analysis of bone formation with respect to different particle sizes of implanted bioceramics. Kuroda reported that the percentages of bone in contact with the HA surfaces did not differ significantly after implantation of granules with three different sizes into osseous tissue.

In recent years, considerable attention has been paid to the development of fabrication methods to prepare porous ceramic scaffolds suitable for bone tissue regeneration. From the presented histological analysis, it is evident that the level of bone formation in the spaces between implanted granules and through the interconnected pores of implanted materials (HAp1, HAp2, DBB) within a cortical region (BAI% 72–85%) was several-fold higher than within a cance-
Bone regeneration materials

Bone regeneration materials

Table 2: The index of bioactivity of the implanted materials

<table>
<thead>
<tr>
<th>Implanted material</th>
<th>( t_{0.5_{bb}} ) (days)</th>
<th>( b )</th>
<th>( I_b = \frac{100}{t_{0.5_{bb}} \text{ (days}^{-1})} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>HAp1</td>
<td>12</td>
<td>8,3</td>
<td></td>
</tr>
<tr>
<td>HAp2</td>
<td>17</td>
<td>5,9</td>
<td></td>
</tr>
<tr>
<td>DBB</td>
<td>18</td>
<td>5,6</td>
<td></td>
</tr>
<tr>
<td>HA*</td>
<td>32</td>
<td>3,1</td>
<td></td>
</tr>
</tbody>
</table>

Values expressing parameters of the synthetic sintered hydroxyapatite produced at high temperature according to [19].

References


Conflict of interest statement None of the authors have a conflict of interest relating to the content of this manuscript.
Physical and chemical characterization of bone regeneration materials based on TCP

Strnadová M., Skrčená A., Nathanský Z.


The purpose of this study was to determine basic physical and chemical properties including phase purity of bone regeneration materials based on β-tricalcium phosphate. Three commercially available materials were investigated: BioResorb (B) (Oraltronics, Germany), Cerasorb (C) (Curasan, Germany) and Poresorb-TCP (P) (Lasak, Czech Republic). X-ray diffraction was used to determine the phase composition. The chemical composition was analyzed by X-ray fluorescence analysis. The microstructure was observed by scanning electron microscope. Specific surface area and porosity was determined by krypton adsorption (BET) and mercury porosimetry. Changes of pH values during exposure in a static physiological solution at 37±0.5°C were evaluated.

X-rays diffraction detected β-tricalcium phosphate as a single crystalline phase in samples C and P. Two phases – α and β-tricalcium phosphate were detected in the sample B. Based on the chemical composition analysis the theoretical phase purity was calculated to 99.4% and 99.0% in the C and P sample. Specific surface area of samples B, P and C was 0.78, 0.18 and 0.17 [m²g⁻¹] respectively. Total porosity was 60% (B), 35% (P) and 30% (C). The change in pH value during the interaction with solution was 1.4 and 1.2 in case of P and C samples respectively and significantly higher pH increase (2.9) was detected in sample B. Single crystalline phase was detected in two of three samples evaluated. The presence of more reactive α-TCP phase in the third sample could cause the higher pH increase after the exposure in solution.

Role of the glass phase in bioactive glass-ceramics

Z. Strnad


Glass-ceramics, or composites with a glass-ceramic matrix prepared by controlled crystallization, almost invariably contain a residual glass phase. A suitable composition for the residual glass phase of bioactive glass-ceramics can be found approximately and controlled on the basis of calculation of a structural parameter Y, which in the simplified concept of the glass structure corresponds to the mean number of bridging oxygens per polyhedron in the glass lattice. The structural parameter Y can be used as a tool for tailoring bioactive glass-ceramics, and composites with a glass-ceramic matrix, which exhibit a combination of the additional required properties.
Development and Clinical Evaluation of Bioactive Implant for Interbody Fusion in the Treatment of Degenerative Lumbar Spine Disease

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Introduction

Due to new information about the pathophysiology and biomechanics of degenerative lumbar spine disease, the surgical treatment of this disease has undergone a significant increase over the past forty years. Novel diagnostic approaches and the development of new materials provided the impetus to produce new types of instrumentation, and these instruments have led to the modernization of interbody fusion including PLIF, TLIF and ALIF methods. These interventions are performed in either an open mini-invasive or endoscopic manner. The open interventions are indicated in cases where the spinal canal stenosis is caused by severe degenerative lesions affecting the motion of intervertebral discs, joints, ligaments, or vertebral arch. Despite the development of other surgical techniques (e.g., functional disc substitutes, dynamic stabilization), the posterior interbody fusion represents a powerful approach in the surgical treatment of degenerative stenosis of the spinal canal.

The PLIF method was first applied in the 1940s by Briggs and Milligan who inserted crushed bone grafts into the intervertebral space, and the bone grafts insertion technique was further developed by Cloward (Cloward, 1953). Due to complications associated with autografts (i.e., pain at the sampling site, procedure prolongation, etc.), the PLIF surgical technique was improved in the 1980s, and new implants constructed of various materials were developed (Bienik and Svicek, 1991; Brantgan et al, 1994; Khoo et al, 2002; Šrámek et al, 2010). Likewise, novel diagnostic tools have been developed including MRI, 3D CT, SPECT-CT (Crock, 1976, Modic et al, 1988; Blumenthal et al, 1988), and new materials (e.g., ceramic, titanium, PEEK) have yielded new types of implants leading to the modernization of the interbody fusion via PLIF techniques (Alexander et al, 2002; Besse et al, 1997; Brayan et al, 2002; Ciappetta et al, 1997; Kokubo, 1990; Yamamoto, 1995; Hashimoto et al, 2002; Thalgott et al, 2002; Sandhu, 2003). Currently, the majority of implants for PLIF consist of two separate components, including the solid cage shape and osseoconductive material (i.e., TCP, BMP) that ensures osteoelastic activity and the interbody fusion formation. To date, no material with both suitable mechanic properties and high grade bioactivity is currently available. For instance, solid materials (e.g., medical steel, titanium, PEEK) lack bioactivity that is able to support the osseoconduction (Carlson et al, 1988; Williams and McNamara, 1987; Zdeblick and Philips, 2003). Likewise, bioactive or resorbable materials (e.g., glass-ceramic, hydroxyapatite, polysacharides) do not meet the mechanical requirements for fusion implants of intervertebral discs (Hench et al, 1971; Filip et al, 1996; Sobale et al, 1990; McAfee, 1986).

Currently, the majority of implants consist of cages that form various shapes. The perimeter is constructed from a solid material that ensures the structural strength. The centre of the cage is hollow and is filled by bone grafts or osseoconductive material (e.g., b- or tri-calcium phosphate) to promote bone fusion in this part of the implant. The optimal implant for an interbody fusion should imitate the properties of the bone tissue by combining sufficient mechanical strength as well as bioactive surface. Therefore, the mechanical strength and the shape of the implant should ensure the primary stability of the segment of the lumbar spine following the operation. Furthermore, the bioactive surface should allow stimulation of osteoblast proliferation at the interface of the implant and bone, and should promote activation of their migration along the implant surface. The bioactive surface should also act as a conductor for osteoblast migration to the fixed vertebral bodies to form the fusion. This quality would prevent the requirement for additional filling of the implant by osseoconductive material. The aim of our work was to create an implant with optimal strength and bioactivity in an attempt to replace the use of autografts and two-compartment implants for PLIF.

Research

Advantages and disadvantages of PLIF surgical technique using autografts

When the conservative treatment fails, patients from categories LS syndrome and FBS syndrome (Failed Back Surgery syndrome) are often referred to surgical management via posterior interbody fusion (Benzel et al, 2003; Cloward, 1953; Crock, 1976; Daniaux, 1986; Dove, 1990; Gurr et al, 1999; Cho et al, 2002). The indication for this surgery is based on neurological finding. In general, the patient predominantly either suffers from back pain associated with progression of root lesion in a lower extremity or with neurogenic claudications in the lower limbs, and shows no reaction to the full conservative therapy algorithm (Anderson, 2000; Brinnckman et al, 1989; Brodke et al, 1997; Cloward, 1953, Hrabálek et al, 2009; Paleček et al, 1994; Fischgrund et al, 1997). Furthermore, the disease is supported by graphic images of compression of the neural structures caused by degenerative lesions (Knudson, 1944; Crock, 1978; Modick et al, 1983; Sonntag and Theodore, 2000). The desired clinical effect can be achieved by the decompression of neural structures together with spondylodesis of the affected spine segment using PLIF (Steefee, 1988; Hashimoto et al, 2002; Dick, 1987; Wang et al, 2005). When surgical treatment is necessary, no acceptable scientific long-term evidence of efficacy exists for any type of surgical treatment of the degenerative lumbar spine disease (Brodke et al, 1997; Benzel, 2003; Sonntag and Theodore, 2000; Paleček et al, 1994).

We performed PLIF using autografts that were developed in the 1980s. An autograft (mostly iliac crest bone grafts) stripped of connective tissue was inserted under compression into the intervertebral space. The best stability was achieved by transpedicular fixation of the operated segment necessary for osteointegration of the grafts via their remodeling and PLIF formation (Bauer and Muschler, 2000). The advantage of a recently collected autograft material is the presence of live bone cells with mineralized extracellular matrix. The biological activity, structure and proteins of bone morphogenesis are important prerequisites of the fusion. In addition, clinical experiences from the first half of the 20th century have proved better surgical outcomes with autograft grafts in comparison to simple decompression (Cloward, 1953; Dawson et al, 1981; Dick, 1987; Carlson et al, 1988).

Autologous bone grafts in this form have been the gold standard for PLIF in the majority of spondylodesis clinics through the end of the 20th century. Despite improving surgical outcomes with a growing number of operated patients, new complications still exist regarding this otherwise successful surgical technique (Kurz et al, 1989).
The most common complications associated with this surgery include problems with bone graft sampling in that limitations are present in bone size and structure that may be safely collected from a live patient in cases of extensive intervention. Furthermore, patients can suffer from unpleasant reactions including debilitating postoperative pain, infection, sepsis, cosmetic defects, nerve injury, hip fractures, vessel injury and blood loss. These adverse reactions can occur in 10 to 30% of cases (Arrington et al., 1996; Banwart et al., 1994; Banwart et al., 1995). Therefore, these reactions and other problems have led to search for artificial materials for PLIF. The optimal material for PLIF substituting bone grafts should ideally have the following characteristics. First, the material should show solid structural support (load resistance immediately after implantation). Second, the material should display osseococonductivity and bioactivity or the ability to bind with a bone, fusion support without any other additional material (e.g., bone, BCP etc.). Third, the material should provide the possibility for a radiographical assessment of the bone fusion process. Finally, the material should show biomechanic properties (elasticity modulus similar to bone).

Development of a new implant for PLIF

As described above we considered using an implant made from a synthetic material for PLIF in the early 1990s to eliminate the disadvantages of autografts (Madawi et al., 1996). The most available implants were constructed of medical steel (Bagby, 1988). However, these implants did not meet our notion of sufficient strength accompanied by bioactivity. Spondylosurgeons in Charkov (Professor Gruntovskij) have successfully used corundum implants in combination with hydroxyapatite for PLIF in the surgical treatment of degenerative lumbar spine disease in the 1980s. According to results of this clinic, the success of this implant resulted from its prism shape with projections firmly anchored in the intervertebral space that helped the implant to fixate the segment with or without transpedicular fixation following operation. Due to its bioinertion, hydroxypatite was added, and this soft material was placed around the corundum (Rowlings, 1993; Gogolewski et al., 1993). Therefore, this implant stimulated formation of osteoblasts, and served as a conductor for their migration between adjacent surfaces of adjoining vertebral bodies. In the early 1990s, another type of prosthesis produced from bioactive glass-ceramic was developed by Electric Nippon Glass, and was used by Japanese orthopedists for PLIF (Yamamuro, 1995, Kokubo, 1990). While transpedicular fixation was added to PLIF due to its fragility, the bioactivity of the implant surface allowed fusion due to migration of bone cells along its surface without addition of any supporting material (e.g., bone, hydroxyapatite) (Sobale, 1990; Yamamuro, 1995). Based on these experiences, we began searching for a material for PLIF implant that would combine the advantages of both the shape and the strength of corundum and the bioactivity of glass-ceramic used in the early 1990s. Thereby, the combination of these two properties would allow strong anchoring of this material in the intervertebral space, the restoration of anatomy in the operated segment, the stabilization of unstable segment, and the formation of interbody fusion associated with osseoconductive properties without addition of another material and without the risk of migration.

Experimental development of glass-ceramic implant (BAS-O)

Unlike bioinert or biotolerant materials, bioactive glass-ceramic material BAS-O, forms a strong chemical bond with live bone tissue (Fadley et al., 1979; Urban, 1992). Material BAS-O is prepared by progressive steps, such as sintering, controlled crystallization and other. The controlled crystallization allows control of processes that determine the bioactive ability of the final material including material transformation, the control of chemical structure, and the structure of the glass phase (Strnad, 1992). The ability of this material to form a strong bond with bone tissue results from the formation of an apatite layer on the material surface resulting in the connection of the bioactive material with body fluid. Crystallographic chemical characteristics of apatite released on the material surface is similar to the organic part of the bone tissue. Thereby, the stability of the operated segment without micromovements and the tight contact of the material without microgaps are necessary for perfect chemical bond BAS-O / live tissue. Otherwise, a risk of connective tissue penetration exists that can prevent the chemical bond on the bone / implant interface (Kokubo, 1990; Urban, 1992).

The most important finding for the planned use of the lumbar implant necessitated that the biochemical and mechanical properties of the glass-ceramic BAS-O mimic the cortical bone tissue. According to the Young model, the shape of their implant exceeded twice the strength of the vertical load, and was close to its flexural strength. Therefore, we based our implant shape on our previous experiences and according to the models that we observed during our study visits. Together with size and shape development, we also created application instrumentation used for the intervertebral space as well as the operation procedure. At this time, the fragility of the ceramic in the contact with steel represented our only disadvantage in that this fragility could cause problems with insertion using metal application instrumentation. The application instrumentation was coated by Teflon in order to prevent damage to the implant. A rectangular prism-shaped implant (25 mm long, 8 mm high and 10 mm wide) was progressively developed after repeated experiments with cadavers from 1991 to 1993 (Filip et al., 1995). “Winglets” have been placed on the opposite sides of the prism (Figure 1). The winglets cut into the adjacent vertebral bodies after its rotation by 90 degrees, and the implant was firmly attached within the space without a risk of migration into the spinal canal. Due to its bioactivity, the implant should stimulate migration of bone cells along its surface to form interbody fusion. The application technique for the glass-ceramic implant was the same as with other implants for PLIF. During experimental application in cadavers, the implants were well-anchored in the vertebral bodies without compression of dural sac in the spinal canal, and this placement was confirmed by imaging techniques (X-ray and CT).

We also planned to apply such PLIF methods alone in both low-grade instabilities and to add transpedicular fixation in high-grade instabilities.

![Fig. 1: Outline of the implant and its position in the interbody space (1992).](image)
The implant was inserted using a specially developed instrumentation into the interbody space (Figure 2).

Implant made of material BAS-0 in clinical practice

Implant BAS-O was registered by the Ministry of Health of the Czech Republic with registration number 89/492/98-IIB in 1992. After registration of the implant and based on experimental results, we introduced the implant into clinical practice in 1994. Based on the advantages from the experimental studies (e.g., stability in the operated space, restoration of anatomy, elimination of the risk of bone grafts sampling, etc.), we expected that these results would be confirmed. From 1994 to 1999, we used this technique in 65 patients observing the indication criteria and the surgical procedure described in the previous chapters. We assessed clinic and graphic postoperative findings in 25 patients out of this population during follow-ups confirmed due to the shape and elimination of risk associated with bone graft sampling. However, the limiting factor for the universal use, in particular for application stand alone technique, was the mechanical resistance to bending at the ultimate load as well as the probable incongruence of the implant and the bone bed, especially related to shorter implants (under 10 mm). This finding was reflected by implant damage on X-ray in the clinical practice (see Table 2). Due to the relatively good results previously observed (Filip et al., 1996), we extended the stand alone technique to higher grades of translation. However, the lower mechanical resistance of the glass-ceramic to bending was observed, and damage of the implant was detected several months after the operation. This problem affected five patients following operation using this technique during twelve months. Despite implant damage, no migration towards the spinal canal or across vertebral bodies occurred resulting from the construction with retention winglets and the chemical bond (Figure 5).

Additionally, PLIF was performed using a pair of implants by the stand alone technique in ten patients (Figure 3), and PLIF was conducted using one or two implants with additional transpedicular fixation by various companies in fifteen patients (Synthes, Stryker etc.). We assessed our results three, six and twelve months after the operation using the ODI score (Oswestry Disability Index; see Table 1). The Oswestry Disability Index (ODI) has become one of the principal condition-specific outcome measures used in the management of spinal disorders. We also used imaging techniques that were available at the time (i.e., X-ray, CT, rarely MRI), and we assessed the change of the implant position in the operated space (i.e., damage, dislocation) using the postoperative imaging techniques (X-ray; see Table 2).

<table>
<thead>
<tr>
<th>ODI score of our population [%]</th>
<th>Mean [n] 25</th>
<th>Primary instability [n] 8</th>
<th>Degenerative listhesis grade I-II [n] 11</th>
<th>Isthmic listhesis up to grade II [n] 6</th>
</tr>
</thead>
<tbody>
<tr>
<td>Before operation</td>
<td>60</td>
<td>55</td>
<td>67</td>
<td>58</td>
</tr>
<tr>
<td>3 months after operation</td>
<td>39</td>
<td>36</td>
<td>44</td>
<td>38</td>
</tr>
<tr>
<td>6 months after operation</td>
<td>40</td>
<td>39</td>
<td>42</td>
<td>38</td>
</tr>
<tr>
<td>12 months after operation</td>
<td>42</td>
<td>40</td>
<td>46</td>
<td>41</td>
</tr>
</tbody>
</table>

Table 1: Clinical assessment according to ODI score.

<table>
<thead>
<tr>
<th>BAS-O implant position on X-ray</th>
<th>Assessment Month 3</th>
<th>Assessment Month 6</th>
<th>Assessment Month 12</th>
</tr>
</thead>
<tbody>
<tr>
<td>No damage, no migration</td>
<td>25</td>
<td>23</td>
<td>20</td>
</tr>
<tr>
<td>Damage, no migration</td>
<td>2</td>
<td>3</td>
<td>5 (20%)</td>
</tr>
</tbody>
</table>

Table 2: Assessment of position change using X-ray.

Our results indicated that we achieved a mean improvement of 1 grade in the aforementioned population according to ODI assessment during twelve months [60% (severe invalidity) and 42% (moderate invalidity) with mild progression in long-term; 38% at month three and 42% at month twelve]. Using X-ray, we diagnosed implant damage without fragment(s) migration towards the spinal canal or in the prevertebral direction in five patients (20%) after twelve months. Initially, we utilized the sole interbody fixation (stand alone technique) mainly in patients with low grade instability. We found that the clinical condition stabilized in these patients, and the postoperative imaging investigation showed good fixation of the operated segment without prosthetic damage and with adequate postoperative changes around the nerve structures (Figure 4).

In case of 1st or 2nd grade translation as previously defined by Meyerding, we added posterior transpedicular fixation of the whole segment to the implant application. The assessments of the population showed advantages and disadvantages associated with the glass-ceramic implant. For example, the operation time was shortened, and firm anchoring in the interbody space was confirmed due to the shape and elimination of risk associated with bone graft sampling. However, the limiting factor for the universal use, in particular for application stand alone technique, was the mechanical resistance to bending at the ultimate load as well as the probable incongruence of the implant and the bone bed, especially related to shorter implants (under 10 mm). This finding was reflected by implant damage on X-ray in the clinical practice (see Table 2). Due to the relatively good results previously observed (Filip et al., 1996), we extended the stand alone technique to higher grades of translation. However, the lower mechanical resistance of the glass-ceramic to bending was observed, and damage of the implant was detected several months after the operation. This problem affected five patients following operation using this technique during twelve months. Despite implant damage, no migration towards the spinal canal or across vertebral bodies occurred resulting from the construction with retention winglets and the chemical bond (Figure 5).
We also added transpedicular fixation of the affected segment in case a patient experienced clinical impairment(s) due to implant damage and instability progression. We did not observe implant damage in the fixed segments in this study; however, we stopped using the stand alone technique with the glass-ceramic implant for PLIF after this experience. Unfortunately, we failed to directly demonstrate osseoconductive properties of the glass-ceramic implant BAS-0 for PLIF that was associated with fusion of the adjacent vertebral bodies by migrating bone tissue from 1994 to 1999.

Experimental development of bioactive titanium in forms by LASAK

The LASAK Company developed bioactive titanium with original surface modification at the end of the 1990s. Due to the limitations of the glass-ceramic implant mentioned above, we have been developing a new type of implant combining bioactive properties and higher mechanical resistance in cooperation with LASAK Company since 1998. Characteristics of this material (higher strength, bioactivity) have provided optimal implant characteristics for PLIF (Yan, 1997; Strnad, 2010). The material used for this implant is technically pure titanium (grade 3) which is dedicated for surgical implants (Regulation ISO 5832-2: 1993(E); Implants for surgery, ISO 5835-2). To ensure bioactivity of this material, the implant surface is chemically modified by LASAK technology (Adjudication on Permission to Use a Medical Device No. 82/125/00-IIB by State Institute for Drugs Control of the Ministry of Health of the Czech Republic). Mechanical properties of this material are identical to pure titanium, and its strength and fracture persistence are several times better than characteristics of the bone tissue and the glass-ceramic material (see Table 3).

Table 3: Comparison of mechanical properties of titanium, bone and glass-ceramic BAS-0.

<table>
<thead>
<tr>
<th></th>
<th>Titanium</th>
<th>Bone</th>
<th>Glass-ceramic</th>
</tr>
</thead>
<tbody>
<tr>
<td>Compressive strength (MPa)</td>
<td>100-230</td>
<td>1080</td>
<td></td>
</tr>
<tr>
<td>Tensile strength, flexural strength* (MPa)</td>
<td>240–680</td>
<td>200*</td>
<td>170–218*</td>
</tr>
<tr>
<td>Elasticity modulus (GPa)</td>
<td>100–120</td>
<td>25</td>
<td>220</td>
</tr>
<tr>
<td>Fracture persistence (MPa \cdot m^{1/2})</td>
<td>~40–100</td>
<td>2-12</td>
<td>2</td>
</tr>
</tbody>
</table>

The mechanically and chemically modified surface of the bioactive titanium by LASAK technology is able to induce the production of calcium-phosphate (apatite), and this compound arises from the interaction between the surface of the material and body fluid within hours to days. The chemical and crystallographical properties of this mineral are nearly identical with the bone apatite. Experimental studies with bioactive and bioinert titanium demonstrated that titanium with a bioactive surface better tolerates unfavourable conditions for osseointegration, as gaps between the implant and the bone (Strnad et al, 2003). Bioinert titanium allows penetration of fibrous tissue into the interface implant/bone, and promotes instability or migration of implants towards the spinal canal in conditions requiring spondylodesisurgery.

Based on these mathematical analyses, we maintained the basic shape of the implant with the above mentioned parameters. The higher strength of the material allowed us to design simplified application instrumentation. We used a thread in the implant body instead of the Teflon-coated fork.
used in the implant BAS-O. Due to its strength, no opposite space dilatation was necessary before the application as a result of the bioactive titanium implant, and no risk of damage of the implant shape by metal loaders was detected. Therefore, the handling of the implant during an intervention is easy and safe. The shape of the implant ensured good restoration of the anatomy of the operated area (restoration of the interbody space and its stability) with minimal risk of implant plunge into the adjacent vertebral bodies, as demonstrated by imaging investigations. Other benefits of the new implant included higher strength and shape variability.

We removed the whole motion segment with implants from cadavers after experimental operations, and we assessed their localization and the degree of their damage by X-ray and CT scans (Filip et al, 2001). Both investigations showed proper localization of the implants in the intervertebral spaces without any contact with the spinal canal or perforation of the winglets into the adjacent vertebral bodies (Figure 7).

Additionally, their shape and surface were not damaged by the new type of instrumentation. Therefore, we assumed that these findings would transfer from experimental studies into clinical practice. However, we were not able to verify the osseoconductive properties of the implant surface in the cadavers. A perfect contact was observed between the surrounding bone tissue resulting from the simple application in cadavers, which was a good precondition for supporting osseointegration in the interbody space via osteoblasts’ migration along its surface. Thus, we verified the osseoconductive properties of the BIO surface of the implant in an animal model (Strnad, 2008). The implant surface in the direct contact with newly produced bone tissue yielded the following values [BIC (%) = 48.5 ± 2.9, 66.0 ± 7.4 and 90.6 ± 7.0, respectively, two, five and twelve weeks after implantation].

**Implaspin in clinical practice**

Encouraged by these experimental results, we began to use this type of implant in clinical practice in indications for PLIF instead of the glass-ceramic implant since 2002 (Figure 9). The operation technique PLIF was identical to the operation technique used in cadavers (Filip et al, 2010). For example, we decompressed the nervous structures through posterior median line approach, and we then radically removed the degenerated intervertebral disc under the control of the operation microscope. Afterwards, we removed the surfaces of the adjacent vertebral bodies, and we then inserted the bioactive titanium implant using...
the innovated instrumentation (Figure 10). Finally, we added transpedicular fixation of the whole segment (Synthes, Signus, Easy spine, etc.) (Figure 11).

To date, we have not observed any complications associated with the implant application into the interbody space. According to the postoperative scans, the implant was always placed in the correct position with winglet penetration into the spongy tissue of the adjacent vertebral bodies. We have selected the size empirically according to the extent of osteochondrosis of the affected disc and the degenerative lesions of the surrounding tissues on scans (X-ray, CT, MRI) during the intervention. In the majority of cases, we used implants (8 or 10 mm high) with angle 4% to maintain lordosis in the lumbar area (Figure 12).

According to the experimental studies, tight contact with the surrounding bone tissue was necessary to activate the bioactivity of the surface. This contact was ensured by the shape of the implant and the winglets that penetrated into the spongious bone tissue of the adjacent vertebral bodies, and was the precondition for migration of the osteoblasts along the implant body resulting in the formation of a junction of the adjacent vertebral bodies by bone tissue without the need to sample bone grafts or to add supporting synthetic materials inside or around the implants.

In 2002 to 2007, operations were performed on 57 patients using the bioactive implant Implaspin in the Neurosurgery Clinic of the Faculty Hospital in Ostrava and in the Neurosurgery Department of Tomáš Bata’s Regional Hospital in Zlín. We assessed a population of 25 patients with follow-up examinations conducted two or more years following surgery, according to the clinical condition. The follow-ups were also based on the generally used score system ODI and imaging methods (X-ray, CT, MRI) that occurred three, six, twelve and 24 months after surgery. During the follow-ups, we examined the patients for signs of implant damage, instability of the operated segment, and signs of supposed osteoblastic activity of the bioactive surface of the implant on the scans. Results of the ODI questionnaire showed that with Implaspin, our success rate improved by 1 degree (59%–40%), or we stabilized the clinical condition of the majority of the patients long-term (2 and more years), which corresponds to results of other clinics using other implant types (Bessho et al., 1997; Brantigan et al. 1993; Brayan et al., 2002; Bienik and Swiecki, 1991; Cappetta et al., 1997; see Table 4).

The assessment of the implant position on scans (X-ray, CT, MRI) at postoperative visits demonstrated no signs of implant damage or implant migration out of the intersomatic space. These investigations have not yet shown any signs of instability of the operated segment (i.e., formation of new osteophytes, progression of hypertrophy of the articular facets, and migration of the implant at the site of application). We observed one severe complication in the population which was caused by an inaccurate application of the transpedicular screws. The wound healed in this patient, and the neurological findings stabilized after removal of the screws. The stabilization of the condition may be supported by the implant shape and the winglets which prevented instability even after the removal of the transpedicular screws. This finding was confirmed by the imaging investigations. Based on the clinical condition and the absence of instability signs on imaging investigations, we concluded that the formation of bone fusion was due to osteoblasts’ migration along the bioactive of Implaspin surface.

**Assessment of bioactivity of the implant using SPECT-CT**

During our investigations, we attempted to demonstrate the migration of bone cells along the surfaces of the glass-ceramic or bioactive implants using imaging investigations. Unfortunately, standard CT or MRI were not able to provide this precise information. The CT scans were limited by screw artefacts, and the MRI scans were generally unable to detect changes in bone. In an attempt to resolve these problems, we utilized SPECT-CT, a method that provides up-to-date computed tomography (CT) and gamma camera (SPECT), to detect the activity of the osteoblasts on the body of the titanium implant applied into the interbody space. The computed tomography (CT) can precisely display the anatomic structure of the investigated tissue, and the gamma camera investigation (SPECT) can yield a functional view of the metabolic process in the patient’s body, but without its precise localization or other anatomical details. Thereby, the combination of these investigations provided more complete information on the precise place of the metabolic process as well as its dynamics. In our study, the metabolic process included the activity of the osteoblasts on the surface of the bioactive implant, as applied by the PLIF method.

<table>
<thead>
<tr>
<th>ODI score in our population</th>
<th>Mean [%]</th>
<th>FBSS [n]</th>
<th>IS [n]</th>
<th>DI [n]</th>
</tr>
</thead>
</table>

*Table 4: Mean Oswestry score values before surgery and at regular visits (FBSS – failed back surgery syndrome; IS – isthmic spondylosis; DI – degenerative instability).*
In 2009, we performed this type of investigation in four patients after surgery for the primary instability of the lower lumbar spine segment using the PLIF operation technique with Implaspin. The study was conducted before the surgery as well as two and six months after the intervention, and we assessed the anatomical changes and metabolic activity at the location where the implants were applied by using the combined scans. The investigation provided preoperative signs of instability localized to the affected space in the area of the disc in all four patients. We detected a hyperintense signal at the operated segment two to three months after the surgery, which was a sign of osteoblast activity on the surface of the implant. We also observed a decrease of this activity (hypointense signal) six months after the surgery as well as a change on the surface of the implant using the combined CT scans. According to our method, hypointensity signified the completion of the osteoblastic activity. The changes on the CT scans were completed by conducting a measurement using Hounsfield’s units (metal – about 2000 HU; bone tissue 100–300 HU), which provided evidence that the implant was overgrown by bone tissue. This kind of image detects the primary successful binding of the implant via activation of the osteoblasts by its specially adapted bioactive surface (figure 13 and 14). Using this combined imaging technique in all four patients, we demonstrated the migration of bone cells along Implaspin wall and the formation of fusion without the addition of another material, such as autografts or TCP, six months after the surgery. Therefore, the successful fusion was indirectly confirmed using the SPECT-CT improving the postoperative clinical findings.

Conclusion

The development of both the material and the shapes of implants continues to progress. Currently, the primary focus of this development is to produce an implant that forms a firm fusion as soon as possible and to ensure the formation of new bone due to its material composition. The current implants for PLIF combine two separate components, including a solid cage shape and osseoconductive material (i.e., TCP, BMP) that ensures the activity of osteoblasts and the formation of the interbody fusion. To date, none of the materials for PLIF available on the market optimally meet both characteristics (see Table 5).

During this investigation, our goal was to develop an implant that would combine both of these components in one unit, ultimately maintaining the strength and bioactive properties present in two-component implants. At the end of the 1990s, we were close to the development of such material due to the implant BAS-0. However, the resistance of the glass-ceramic at the ultimate load provided a limitation that negatively influenced the shape and the application process, as described in chapters above. However, due to these experiences, we and other technicians successfully designed an implant that meets our original conception. This implant is currently used in clinical practice, and experimental studies have confirmed its supposed properties. The combination of the implant’s strength and shape with bioactivity enables the smooth application and restoration of anatomy, thereby providing a perfect fixation of the operated segment and stimulating growth of osteoblasts and their migration along its surface. Our original implant Implaspin combines the osteoconductive and osteoplastic properties of the glass-ceramic with the strength of titanium, which was the aim of our research. Thanks to these properties, this implant represents a quality alternative to implants constructed from other materials dedicated to PLIF (see Table 5).

Acknowledgement

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Table 5: Parameters of the implant according to the type of material.
References

Bioactive titan cage Implaspin in treatment of degenerative disease of the cervical spine - the results from 2007 till 2008

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Chirurgia Narządów Ruchu i Ortopedia Polska, 75 (1), 2010

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2 LASAK s.r.o (Ltd.)- Glass and Ceramics Laboratory, Praha 6, Czech Republic

Summary

The authors present results of surgical treatment of cervical spine degenerative disease via Implaspin biotitanium replacement. Surgery was indicated for a group of 24 patients with symptoms of cervical spondylogenic myelopathy or the irritation decay root syndrome non-reacting to conservative treatment. Pre-surgery X-ray and MRI examinations showed spinal canal stenosis caused by the intervertebral disk osteochondrosis combined with prolapse or dorsal osteophytes. Clinical problems of the group of patients were evaluated through the JOA classification before surgery and during the 2nd, the 6th and month 12th after surgery. The surgery rate of success was evaluated in percentages during post-surgery examinations that took place in the 12th month. Based on the JOA classification, that rate of success falls into the good surgery results zone. The post-surgery X-ray examinations showed two sink replacements by 1/3 of its height into the surrounding vertebral bodies. In these cases we performed the control MRI. No signs of the new spinal compression were found and the spinal canal was free in the operated site. Based on our short-term experiences, the Implaspin bioactive replacement seems to be a suitable alternative to other types of replacements designed for intervertebral fusion in the lower cervical spine area.

Introduction

Between 1955 and 1958, Robinson, Smith and Cloward1,2 published gradually their first experience with a decompression of the spinal canal in the cervical part from the anterior approach. Due to this knowledge the surgical therapy of the degenerative disease of the cervical spine achieved a great development in the recent 40 years. The various methods, modifications of decompression and fixation of the cervical spine from the anterior approach gradually appear. From the simple decompression with or without the interbody fusion using the bone graft or various cages with or without the split to the functional replacement of the intervertebral disk.3,14 Between 2007 and 2008, we operated 61 patients with a degenerative disease of the cervical spine at the neurosurgery department of the KNTB Zlín. In 24 patients we performed a decompression with the fusion from the anterior approach by means of the biotitanium cage Implaspin. The clinical condition of the patients was evaluated using the JOA classification before and 2, 6 and 12 months after the surgery. The position of the implants in the operated field was evaluated in the same time period using the X-ray images. The presence of the fusion was evaluated using the CT between the 6. and 12. month after the surgery. According to the JOA classification we achieved the efficacy of 50% which lies in the range of the good surgical results. According to the post-operative X-ray images no impairment or dislocation of the cage occurred. The CT examination in four patients performed between the 6. and 12. month after the surgery found a development of the osseous fusion.

Material and methods

In 2005, based on good practice with the bioactive titanium cages of the lumbar spine, we developed a cage for the cervical spine from the same material, as well. Similarly as in the case of the lumbar spine cage the osteococonductive properties of the material remained the same and the rigidity was significantly improved. The cage has a shape of the skewed prism narrowed by 1 grade towards the spinal canal of 13-15 mm in length and a graduated height of 8-5 mm and 13 mm in width. It is produced from the technically pure titanium with chemically modified surface that ensures its bioactive properties. The bioactive surface enables to create a rigid binding with the osseous tissue and has osteococonductive properties.15,16 The material is grey-black with a density of 4500 kg.m-3 and it is rigid in traction to a maximum of 450 MPa. The replacement is intended for a replacement of the intervertebral disk to prevent the instability of the affected locomotive segment of the cervical spine. On the opposite parts of the prism which are attached to the vertebral bodies the replacement has sharp wings of 0.5 mm in height and the angle of 30 grades. They ensure the primary stability for an undisturbed healing of the implant into the surrounding vital bone tissue. The shape and size were supported by the biomechanical studies. The patients with persisting or progressive irritation extinction radicular syndrome and/or spondylogenic cervical myelopathy after unsuccessful conservative therapy were indicated for the anterior decompression of the spinal canal with interbody fusion in the area of the lower part of the cervical spine.3 According to the JOA classification the motor function is rated from 0 to 4 points on the upper limb, 0 to 4 points on the lower limb, 0 to 2 points for sensitive disorders and 0 to 3 points in the urinary bladder function. The less rate the more severe is the affection of a patient.

<table>
<thead>
<tr>
<th>Percent of the successfulness according to the JOA formula</th>
<th>Evaluation</th>
</tr>
</thead>
<tbody>
<tr>
<td>100-75%</td>
<td>excellent result</td>
</tr>
<tr>
<td>75-50%</td>
<td>good result</td>
</tr>
<tr>
<td>50-25%</td>
<td>weak result</td>
</tr>
<tr>
<td>25-0%</td>
<td>bad result</td>
</tr>
</tbody>
</table>

Table 1: Evaluation of the successfulness of the surgery according to the JOA classification

The successfullness of the surgical therapy according to the JOA classification was in our set of 24 patients evaluated in percents by comparison of the points in the group before the surgery and during the preoperative check-ups on the 2., 6. and 12. month after the surgery and by using the formula:

\[
\text{successfullness} \% = \left(\frac{A - B}{C - B}\right) \times 100
\]

A - Postoperative JOA score in the group
B - Preoperative mean JOA score in the group
C - JOA score in a healthy human17

Evaluation of the successfulness of the surgery is displayed in Table 1.
The imaging methods (X-ray and MRI) found degenerative changes in all patients. These changes narrowed the spinal canal in the anterior part, most often by the dorsal osteophytes with or without the prolapse of the intervertebral disk (Fig. 1a and 1b).

The own surgical technique via the anterior approach is performed according to Smith and Robinson including the Caspar’s instruments and the surgical microscope. From the anterior approach under the control of the surgical microscope we remove the degenerated disk including the dorsal osteophytes and remove the residua of both parts of the posterior longitudinal ligament up to the dural sac to the neuronal radices. Then we remove the covering disks of the adjacent vertebral bodies using the high-speed rotational cutter. Using the Caspar’s instruments we distract the operated space and insert the biotitanium cage to fit the size of the bed. Then we cancel the distraction and the vertebral bodies are going to lie on the implant with their sanguinous side which is, due to the compression, wings and the bioactive properties firmly fixed in the space on to the surrounding osseous tissue. Due to this mechanism it fixes the whole operated segment of the spine. The operated segment does not have to be fixed by the splint (Fig. 2a and 2b).

Within 48 hours from the surgery the chemical rigid binding should be established on the implant-osseous tissue interface and the osseous fusion should develop within 6 months due to the bioactive properties of the implant.

**Results**

In our set of 24 patients who were operated between 2007 and 2008 at the neurosurgery department of the KNTB Zlin the localization and type of the affection of the impaired segments according to the preoperative X-ray and MRI examinations is described in Tables 2 and 3.

**Table 2: Localization of the affected space**

<table>
<thead>
<tr>
<th>Space</th>
<th>Number of patients 2007 – 2008</th>
</tr>
</thead>
<tbody>
<tr>
<td>C3/4</td>
<td>2</td>
</tr>
<tr>
<td>C4/5</td>
<td>4</td>
</tr>
<tr>
<td>C5/6</td>
<td>10</td>
</tr>
<tr>
<td>C6/7</td>
<td>8</td>
</tr>
<tr>
<td>Combination of 2 spaces</td>
<td>6</td>
</tr>
</tbody>
</table>

**Table 3: Type of the degenerative disease**

<table>
<thead>
<tr>
<th>Type of the disease</th>
<th>Number of patients 2007 – 2008</th>
</tr>
</thead>
<tbody>
<tr>
<td>Simple prolapse</td>
<td>1</td>
</tr>
<tr>
<td>Prolapse and dorsal osteophytes</td>
<td>16</td>
</tr>
<tr>
<td>Osteochondrosis of the disk +     osteophytes</td>
<td>7</td>
</tr>
</tbody>
</table>

Preoperative JOA score in our group was evaluated to 12.5 points. Two months after the surgery the group was evaluated according to the JOA score to 15 points and in the 6. month to 16 points. During the check-up 12 and more months after the surgery the JOA score was again 15 points. The successfulness of the surgical therapy was expressed in percents after applying all the values into the formula according to the JOA. It was 50% twelve and more months after the surgery. The succe-
Spinal surgery

Satisfaction in percents ranges within the good results at all check-ups – see Table 1.

During the clinical check-up we performed the X-ray control of the operated part as well with the focus on the position of the implant – see Table 4 (Fig. 3a and 3b).

In our group, we found two sink replacements by 1/3 of its height into the surrounding vertebral bodies. In these cases we performed the control MRI. No signs of the new spinal compression were found and the spinal canal was free in the operated site.

Table 4: Position of the Implaspin replacement on the X-ray images after 12 months

<table>
<thead>
<tr>
<th>Control</th>
<th>12. month</th>
</tr>
</thead>
<tbody>
<tr>
<td>No changes, preserved lordosis, without propagation into the spinal canal, without any new osteophytes</td>
<td>22 patients</td>
</tr>
<tr>
<td>Sinking into the bodies of the surrounding vertebra, without propagation into the spinal canal, without any new osteophytes</td>
<td>2 patients</td>
</tr>
</tbody>
</table>

Between the 6. and 12. month after the surgery we tried to find out the direct signs of the osseous fusion using the ultrathin cuts by means of the CT (Fig. 4a and 4b) in four patients. The fusion was developed due to the osseocoductive properties of the replacement (a growth of the osteoblasts on the walls of the implant as it is seen in the CT cuts).

Discussion

The interbody fusion via the anterior approach remains the verified standard therapeutic procedure in the subaxial part of the cervical spine in mono and bisegmental severe degenerative stenoses caused by the posterior osteophytes and/or by the osteophytes in combination with the prolapse of the intervertebral disk.6-9

Using of the various materials for the purpose of the interbody fusions was initiated worldwide approximately in the second half of 80’s. We compared the properties of the Implaspin cage with the similar implants manufactured by Medtronic, Biomed, Synthes, Aesculap, and Johnson which we use concurrently with this implant or we used them at the previous departments in the same indications.4, 8-12, 16

The surgical procedure and the own surgical technique was similar in all the implants. We compared the shape, function of the wings at the opposite sides and the bioactivity of its surface with other implants. Implaspin has a shape of a skewed prism in various sizes. It provides better prerequisites for maintaining the lordosis of the cervical spine in the postoperative period compared to some types of the cages. They have a shape of a prism or oval but without skewing.

The wings of the Implaspin cage pointing at the surrounding vertebral bodies. Together with the compression they increase a stability and ensure a direct contact of the implant with the osseous tissue. The wings help to mechanically fix the operated part in the first hours until the chemical binding bone-implant develops. Some implants available on the market such as Ti-bone of Biomed do not have these wings and the mechanical stability of the operated part is provided only by the compression of the surrounding vertebra.

Due to the bioactivity of the surface of the Implaspin implant a chemical bin-
The chemical binding and a subsequent interbody fusion occurs only at this place. The bioactive material is not homogenous which may cause retardation of growing of the osteoblasts during the fusion. There is also a higher risk of sinking of the implant into the surrounding vertebral bodies due to a hollow center of the implant. This risk is minimized in case of the Implaspin replacement as confirmed by the postoperative X-ray. A sinking of the cervical implant into the surrounding vertebral bodies with out any clinical finding occurred only in two patients. We assume that the sinking occurred due to impairment of the spongious tissue during milling of the bed for the implant.

During the application of the Implaspin replacement there is no need to fill the implant with the bone grafts or hydroxyapatite as in other replacements. It is not necessary to take the bone graft from the patient. Therefore the time of the own surgical procedure is reduced and the patient is not traumatized by the collection of the bone graft from the hip bone. Therefore, we consider the shape, wings and the bioactive surface as an advantage compared to other replacements.

**Conclusion**

Based on the clinical results and imaging examinations it appears that we use the Implaspin cage separately for the fusion in serious preoperative clinical findings (pre operative JOA score of 12.5 point) caused by the compression of the nervous structures most often due to prolapse of the disk with the dorsal osteophytes based on the imaging methods (X-ray, MRI). The postoperative results OOA, X-ray, CT show the Implaspin is a good alternative to other cages intended for the anterior interbody fusion of the cervical spine. It successfully competes with the foreign products as for its quality and price.

**Literature**

Two years of experiences with the biotitanium replacement (Implaspin) used in treatment of degenerative lumbar spine diseases

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Chirurgia narządów ruchu i Ortopedia Polska 75 (2), 2010, p. 131–135

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2 LASAK Ltd, Papírenská 25, Prague 6, Czech Republic

Summary

A group of 25 patients operated on at the neurosurgical clinic of the University Hospital in Ostrava from May 2001 to December 2003 due to their lumbar spine monosegmental stenosis with progressing neurological symptomatology and serious patient limitation in accordance with the Oswestry score / pain scale (59% / poor) were continuously evaluated after surgery every 3, 6, 12, and 24 months by means of the Oswestry protocol and the pain scale supplemented by projection methods (X-ray image – anteroposterior and side projection, functional X-ray images, and CT scans). Spinal canal stenosis was diagnosed by means of X-ray, permyelo – CT, and MRI. Sixteen patients from that group were operated on due to their primary degenerative problems (instability associated with osteochondrosis and osteoarthrosis, and isthmic spondylolithesis grade I, II.). Nine patients from that group were operated on due to their secondary degenerative problems caused by previous surgeries (FBSS – Failed Back Surgery). Upon decompression of nerve structures, the whole group was subject to posteriori lumbar interbody fusion (PLIF) by means of the new Implaspin bioactive replacement (LASAK, Ltd) and transpedicular fixation (Stryker, Miami-Moss, Click). During the monitoring period, it was possible to decrease Oswestry scores of the operated patients to 42 - 40 % without any significant impact on the pain scale (good-poor) by the last check in the 24th month (no change of the instrumentarium position, based on visual examinations). Only one complication associated with fixation was identified in our group. It was a case of surgical wound infection observed on the 6th day after the surgery and caused by an improperly transpedicularly applied screw. We removed the transpedicular fixation and used only interosmotic fusion based on Implaspin replacements. The patient did not report growing problems and the replacement positions did not change (X-ray).

Introduction

During the past twenty years, surgical treatment of the lumbar spine degenerative disease significantly progressed thanks to the new findings in the area of the lumbar spine degenerative disease pathophysiology. Those findings were confirmed by modern imaging methods like 3D CT or myelo – MRI. New possibilities of diagnostics and development of new materials triggered development of new types of instrumentarium for surgical treatment of stenosis caused by the lumbar spine degenerative disease. Interbody fusion gets modernized via the PLIF, TLIF, and ALIF1, 2, 3, 4, 5, 6, 7 techniques. Those operations may currently be conducted in the open, mini-invasive or endoscopic manner. Open surgeries with PLIF are indicated in those cases when spinal canal stenosis is caused by degenerative changes affecting the whole motor segment (intervertebral disk, joints, ligaments, disk arch). From the end of the 90th. surgical techniques like intervertebral disk functional replacements or dynamic stabilizations have been used again. Those surgeries are generally indicated in connection with degenerative diseases affecting mainly the intervertebral disk1 with minor affection of the motor segment rear elements. Despite the rebirth of those techniques, rear decompression of nerve structures supplemented by PLIF with transpedicular fixation conducted during surgical treatment of serious degenerative steno-

Material and methodology

The Implaspin replacement is a rectangular prism with a length of 20 mm. The implant front part height is from 8 to 10 mm. The angle towards the spinal canal is 4 degrees and its width is 8 mm. The shape that could already serve as an intervertebral disk replacement features two pairs of wings (height of 2 mm, sharp edges) located on the opposite sides of the tapered prism. The strength of the wings depends on the used titanium type. Upon turning in the interbody area, the wings penetrate spongiosis of adjacent vertebrae bodies and secure implant mechanical stability in the operated section before creation of the bone fusion itself. Material mechanical strength allows anchoring of the application instrumentarium in the replacement. As a result of that, handling in the surgical field during its application is made easier and better organized. Thanks to the application of bioactive material by means of etching to the whole titanium replacement surface, there is no risk of damaging the bioactive layer during application in the interbody area like in the case of those replacements whose bioactive surface is applied by means of plasma or enameling. Thanks to those properties, within 48 hours the bone-implant chemical bond is created over its whole surface that is in contact with adjacent vertebrae bodies. The chemical bond increases the strength of the mechanical fixation created by means of the wings. The mechanical fixation and the chemical bond stabilize the operated section and allow, therefore, creation of bond fusion within 12 weeks from surgery thanks to material osteoconductivity. Bioactive surface osteoconductove properties expedite the growth of osteoblasts on the replacement walls and production of bone interosmotic fusion of the operated section.

Methodology

The interosmotic fusion via PLIF was indicated upon completion of conservative therapy, based on the patient’s problems evaluated through neurological examination, the Oswestry questionnaire, and the pain scale (poor, fair, good, and excellent). Five patients reported mainly LS spine pain, 8 patients suffered irritation and destructive root lesion, and 12 patients were diagnosed with intermittent claudications (see Table 2). The pre-surgery limitation value of the group patients obtained via the Oswestry questionnaire was 59 %. That means serious problems up to paralysis and pain perception at the fair level, based on the pain scale. That means intense all-day pains.
Imaging methods supplemented the pre-surgery diagnostics. Plain and functional X-ray images were produced and pereybloo CT / static MRI comple-
ted. The imaging examinations evaluated both the type and the dynamics of the lumbar spine examined section affection.15, 16 The patients diagnosed with FBSS were examined by a psychologist.15, 16 In order to obtain objective data on root lesion, electrophysiological examinations were completed, as necessary. In the case of 2 patients, indication for PLIF was completed via diagnostiscs through external fixator due to nondescript clinical finding during seri-
ous post-surgery changes identified via imaging methods.17, 18 Imaging examinations indicated the following patients for PLIF: 10 patients with de-
generative instability, 6 patients with isthmic spondylolisthesis, and 9 patients with FBSS (see Table 1).

Surgical technique

The surgery itself was conducted under total anaesthesia. The patient was in his/her prone position, under control of the C shoulder. Skeletization of paravertebral muscles and clarification of the scope of decompression were followed by laminectomy conducted above the affected segment and re-
lease of nerve structures from hypertrophied facets, arcs or scars caused by the previous surgeries completed by means of a surgical microscope.19,
20 When protection of the released nerve structures was established rear disectomy was completed. Cover surfaces of the adjacent vertebrae were covered with blood by means of sharp curettes. This way a bed for inter-
body application of replacements was created from both sides. Measuring of that bed was followed by application of the Implaspin replacement of the right size by means of application instrumentarium. Transpedicular fixation of the operated section was applied as well. The transpedicular fixation was completed by means of the Miammi-Moss instrumentarium (8x), Click X (4x), and Stryker XIA (13x) (see Table 4).

After surgery, the patients were treated at the ICU for 24 hours and then at a standard ward when their overall condition stabilized. Verticalization of the patients in corsets was initiated on the 3rd – 5th day after surgery when their post-surgery pains subsided. During the first 2 days, those pains were suppressed by opiates and then by a combination of peroral and injection
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cation to the replacement surface through etching, the risks of

Discussion

In connection with surgical indications we have to constantly remember that there is no acceptable scientific evidence of effectiveness of any type of surgical treatment of lumbar spine degenerative disease treatment on a long-term basis. That was also confirmed by the results of our work during evaluations of the Oswestry questionnaire and the pain scale. From a long-
term perspective, there was a percentage improvement by approximately one grade (59 % – 40 %) and the level of pain was not significantly affected.

The use of materials with bioactive properties expediting PLIF started in the world approximately in the second half of the 1980s.21, 22 In the previous period, PLIF was conducted by means of separate bone grafts or replace-
ments filled with bone grafts. During surgery, bone grafts are obtained for example from the patient’s iliac bone or cadaverous tissue. The patient faces either an additional surgical load with all its risks or the risk of transfer of potential infections from the cadaver. Until the year 1993, the Neuro-
surgical Clinic of the University Hospital in Ostrava would complete PLIF via separate bone grafts and in the year 1993 it started surgical treatment of lumbar spine degenerative disease based on application of replacements of bioactive glass ceramics.8, 9, 10 Thanks to their shape and bioactivity, they successfully replaced bone grafts during PLIF. In the years 1993 – 1998, we accumulated plenty of experiences and also discovered the disadvantages and limitations of glass ceramic replacements deriving mainly from the me-
chanical properties of glass ceramics. The limiting factor of application of glass ceramic replacements in the case of significant degenerative changes of the lumbar spine rest in the implant strength that depends on its size. Another limiting factor is glass ceramic’s fragility, which does not allow any contact with metal tools during surgery. It is also not possible to attach the instrumentarium to the glass ceramic replacement – the replacement must be inserted into the instrumentarium. That means the surgeon must com-
pete a large-scale decompression whose scope sometimes exceeds the scope of degenerative changes. If a replacement is not big enough, there is a risk of its damage when subjected to flexural load. There were recor-
ded cases of post-surgical complications like fracture and dislocation of the replacement fragment in the interbody area. The desire to eliminate the undesirable mechanical properties of glass ceramic replacements triggered development of new material – bioactive titanium developed by the LA-
SAM company.14, 27 That material combines the benefits of glass ceramics bioactivity with mechanical properties of titanium. Thanks to the bioactive layer application to the replacement surface through etching, the risks of the replacement bioactivity damage associated with the replacement appli-
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Results

The clinical condition of the group of 25 patients operated on was contin-
uously evaluated via the limitation level specified in % by means of the Oswestry score (0-20 % – mild limitation, 20-40 % – medium-heavy limi-
tation, 40-60 % – heavy limitation, 60-80 % – disability, 80 % and more – inability to conduct a normal way of life) and the pain scale expressed in points (excellent 0-2, good 3-5, poor 5-7, bad 7-10). The average age was 51.6 years (the age range was 36-65 years). Out of that, there were 13 women and 12 men. There were two surgeries of the L3/L4 section, 14 surgeries of the L4/L5 section, and 9 surgeries of the L5/S1 section (see Ta-
ble 3). During post-surgery check-ups conducted 3, 6, 12, and 24 months after surgery, the patients were checked through the Oswestry questionnaire, the pain scale, and imaging methods. The Oswestry score / pain scale were evaluated in the 3rd month (42 % / good), in the 6th month (40 % / good), in the 12th month (41 % / good), and in the 24th month (40 % / good). During the whole monitored period of 24 months, based on the Oswestry seri-
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generative instability, 6 patients with isthmic spondylolisthesis, and 9 patients with FBSS (see Table 1).
Implaspin replacements and changed the way of attaching the application instrumentarium. The work results showed that the new Implaspin replacement eliminated the glass ceramic replacement disadvantages associated with application and the material strength. During surgeries, it was not necessary to conduct any excessive decompression thanks to the size of the application instrumentarium and the way of its attachment in the replacement. Post-surgery check examinations of all the 25 patients of the group did not identify and implant damage or position changes. One case of incorrect transpedicular screw application did not relate to the implant application.

The surgery technique and the mechanical and bioactive properties of the Implaspin replacement were compared with similar implants we have been using over the past 7 years (produced by Stryker, SpineTech, Comesa, and Miami-Moss). Both the type of surgery and the surgical technique itself were similar to the foreign implants. The scope of removal of bone and ligament structures of rear elements conducted during application of the Implaspin replacement was comparable with the other types of implants. In terms of all strength-related parameters and creation of intersomatic fusion, the mechanical and bioactive properties of our implant match the properties of similar products made by foreign companies. Based on functional images, our group contained no signs of instability. No CT scans showed direct signs of any osteoblastic activity.

We see the benefits of the Implaspin replacement, compared to other types of implants, in the shape of the replacement with its wings and its osteointegration through a chemical bond. The wings allow firm fixing of a replacement in the operated intersomatic area. This mechanical stability represents a tight contact of bone marrow with the implant surface and it is the first precondition of creation of the future bone fusion. Osteointegration via chemical bond increases the strength after application and enables osteoblastic activity of bone cells at the replacement walls. That should allow creation of bone fusion in the operated area within 20 weeks. This statement cannot be backed by any direct signs of osseointegration due to a low-quality CT image. X-ray images showed a fusion through indirect signs in the form of unchanged positions of fixation equipment and during all the X-ray checks no signs of instability were detected in all the 25 patients. Thanks to those properties, Implaspin does not have to be filled with bone grafts or hydroxyapatite like the other replacements.

Bioactive titanium of the LASAK company guarantees good X-ray contrast during surgery and after surgery at an X-ray amplifier or X-ray images and a replacement in the operated intersomatic area. This mechanical stability enables osteoblastic activity of bone cells at the replacement walls. That should allow creation of bone fusion in the operated area within 20 weeks. This statement cannot be backed by any direct signs of osseointegration due to a low-quality CT image. X-ray images showed a fusion through indirect signs in the form of unchanged positions of fixation equipment and during all the X-ray checks no signs of instability were detected in all the 25 patients. Thanks to those properties, Implaspin does not have to be filled with bone grafts or hydroxyapatite like the other replacements.

Bioactive titanium of the LASAK company guarantees good X-ray contrast during surgery and after surgery at an X-ray amplifier or X-ray images and quality legibility of findings in CT/MRI images of post-surgery conditions.

Conclusion

The Implaspin lumbar replacement perfectly integrates the osseoconductive properties of glass ceramics and the strength of titanium. During our evaluations of the group we registered a decreased level of the patients’ problems, based on the Oswestry questionnaire, and no deterioration on the pain scale during the 24-month post-surgery period. Our imaging methods revealed no signs of damage, replacement progression or instability of the operated section. It is clear from the results that the Implaspin replacement of the LASAK company represents a perfect alternative to other lumbar spine replacements in the case of intersomatic fusion indications. Thanks to its quality and reasonable price, it successfully competes with foreign products.

Literature

Degenerative instabilities 10
Degenerative and isthmic spondylolistheses 6
FBSS 9

Table 1: Indications based on paraclinical examinations

Miammi Moss 8x
Click X 4x
XIA Stryker 13x

Table 4: Applied instrumentaria

LS spine pains 5
Irritation and destr. root lesions 8
Intermittent claudication 12

Table 2: Indications based on clinical discomforts

Oswestry score before surgery 59 %
Oswestry score 3 months after surgery 42 %
Oswestry score 6 months after surgery 40 %
Oswestry score 12 months after surgery 41 %

Table 5: Oswestry score

L3/L4 2x
L4/L5 14x
L5/S1 9x

Table 3: Heights of the operated segments

Pain scale:

0--------5--------10
Legend: 0 – smallest pain
10 – maximum pain

The patient will mark his/her level of pain on the scale.
The Implaspin biotitanium intervertebral disk replacement used in treatment of cervical spine degenerative diseases – the first experiences

M. Filip, P. Veselský, M. Mrůzek, T. Paleček, Z. Strnad, J. Strnad

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1 Neurosurgical clinic FNsP Ostrava-Poruba, Czech Republic
2 LASAK Ltd, Papířeřská 25, Prague 6, Czech Republic

Summary

The authors present their short-term results of surgical treatment of cervical spine degenerative disease via Implaspin biotitanium replacement. Surgery was indicated for a group of 12 patients with symptoms of cervical spondylotic myelopathy or the irritation decay root syndrome non-reacting to conservative treatment. Pre-surgery X-ray and MRI examinations showed spinal canal stenosis caused by the intervertebral disk osteochondrosis combined with prolapse or dorsal osteophytes. Clinical problems of the group of patients were evaluated through the JOA classification before surgery (11.6 points) and during the 2nd and the 6th month after surgery (15.5 and 15.0 points). The surgery rate of success was evaluated in percentages during post-surgery examinations that took place in the 2nd and the 6th month (72.2 % and 62.0 %). Based on the JOA classification, that rate of success falls into the good surgery results zone. The post-surgery X-ray examinations showed no replacement damage or dislocation. Based on our short-term experiences, the Implaspin bioactive titanium replacement seems to be a suitable alternative to the other types of replacements designed for intervertebral fusion in the lower cervical spine area.

Introduction

In the year 1934, Mixter and Bar completed the first laminectomy for the lumbar intervertebral disk prolapse. This method of spinal canal decompression subsequently became a standard procedure used during spinal canal obturation due to various reasons. In the years 1955-1958 Robinson, Smith, and Cloward gradually published their first experiences with spinal canal decompression completed in the cervical section from the front. Thanks to those findings, the degenerative cervical spine disease treatment significantly advanced during the past 40 years. Various methods of decompression modification and fixation of the cervical spine from the front became available. They include plain decompression without or with interbody fusion by means of bone grafts or various replacements with or without splints, including the intervertebral disk functional replacement. We have been dealing with those issues at our clinic for already 30 years. In June 2003, we started completing interbody fusions through the biotitanium cervical replacement Implaspin made by LASAK, Ltd., Czech Republic. During one year, we provided this implant to 12 patients. We evaluated our results via JOA classification-based examinations and X-ray images.

Material and methodology

In the year 2003, based on our positive experiences with the lumbar spine bioactive titanium replacement that gradually replaced the glass ceramic replacement we also developed a replacement for the cervical spine of the same material. The material osseoadhesive properties applied in the lumbar spine replacement were preserved and the strength significantly increased. Thanks to that we were also able to use smaller sizes than in the case of the glass ceramic replacement. The implant’s basic shape is a tape-red prism narrowed by 1 degree towards the spinal canal with a length of 13-15 mm and a graded height of 8-5 mm, and a width of 13 mm. On the opposite sides of the prism attached to vertebra bodies upon application, the implant features sharp wings with a height of 0.5 mm. They secure primary stability necessary for undisturbed implant healing in the surrounding vital bone tissue. The implant is made of commercially pure titanium with chemically treated surface securing its bioactive properties (bio surface). The implant bioactive surface allows creating of firm bonds with bone tissue and features osseoadhesive properties.

The material is black-gray and its density is 4,500 kg/m³. Its tensile strength is at least 450 MPa. The implant is designed for intervertebral replacements preventing instability of the affected cervical spine motor segment. The anterior decompression of spinal canal with interbody fusion in the lower section of the cervical spine was completed in patients with permanent or prograding irritation decay root syndrome or spondylotic cervical myelopathy after unsuccessful conservative therapy caused by dorsal osteophytes, osteochondrosis or cervical spine intervertebral disk prolapse.

Clinical findings were evaluated through the JOA (Japanese Orthopaedic Association) classification. JOA uses 0-4 points for evaluation of the upper limb motor function and the lower limb motor function, 0-2 points for sensitive failures, and 0-3 points for bladder functions. The smaller the number of points, the bigger the problems faced by the patient. The surgical treatment rate of success evaluated via the JOA classification was expressed as a percentage based on the comparison of points available before surgery and after during post-surgery examinations in line with the following formula:

\[ \text{Success rate (\%)} = \frac{(A - B)}{(C - B)} \times 100 \]

A – post-surgery JOA score of the group, B – pre-surgery average JOA score of the group, and C – JOA score of a healthy individual.

The surgery success rate evaluations are available in Table 1. Imaging methods (X-ray and MRI) applied to all the patients showed degenerative changes that narrowed the spinal canal anterior part, most often via dorsal osteophytes with or without intervertebral disk prolapse. Imaging methods (X-ray and MRI) applied to all the patients showed degenerative changes that narrowed the spinal canal anterior part, most often via dorsal osteophytes with or without intervertebral disk prolapse. Imaging methods (X-ray and MRI) applied to all the patients showed degenerative changes that narrowed the spinal canal anterior part, most often via dorsal osteophytes with or without intervertebral disk prolapse.

Surgery technique

The surgery technique based on the anterior access was applied in accordance with Smith and Robinson and it was supplemented by Caspar’s instrumentarium and surgery microscope. Under the surgery microscope a degenerated disk and dorsal osteophytes were removed from the anterior section, including remainders of both parts of the rear longitudinal ligament up to the dura matter to nerve roots. After that, cover disks of adjacent vertebra bodies were removed by a high-speed milling machine. The operated area was dilated by Caspar’s instrumentarium and a biotitanium replacement of the right size was inserted. After that, the dilution ended and the blood-covered surfaces of vertebra bodies were attached to the implant. Thanks to compression, the wings, and the bioactive properties,
Spinal surgery

During recent treatments of cervical spine degenerative diseases renaissance of functional replacements of the intervertebral disk takes place. It generally applies that those replacements are applied in those cases where degenerative changes are associated mainly with affected intervertebral disks without any changes of their surrounding bone skeleton. Despite the first optimistic references, we still have to wait for long-term results. Interbody fusion based on anterior access remains a verified standard method of treatment applied to the distal section of cervical spine in connection with mono- and bisegmental serious degenerative stenoses caused by osteophytes or osteoarthrosis in combination with intervertebral disk prolapse. Approximately in the second half of the 1980s the world started to use the interbody fusion materials. Our clinic began using bioactive glass ceramic replacements in treatment of lumbar spine degenerative diseases in the year 1993 and in the year 1995 also in treatment of cervical spine degenerative diseases. Growing experiences with glass ceramic replacements contributed to development of precise indications for their applications. The disadvantages and limitations deriving from mechanical.

**Table 1:** Evaluation of surgery success rate according to the JOA classification

<table>
<thead>
<tr>
<th>Success rate percentage based on the JOA formula</th>
<th>Evaluation</th>
</tr>
</thead>
<tbody>
<tr>
<td>100 – 75 %</td>
<td>Excellent result</td>
</tr>
<tr>
<td>75 – 50 %</td>
<td>Good result</td>
</tr>
<tr>
<td>50 – 25 %</td>
<td>Poor result</td>
</tr>
<tr>
<td>25 – 0 %</td>
<td>Bad result</td>
</tr>
</tbody>
</table>

**Table 2:** Type of degenerative disease, as revealed by X-ray and MRI examinations

<table>
<thead>
<tr>
<th>Area</th>
<th>Number of patients</th>
</tr>
</thead>
<tbody>
<tr>
<td>C4/5</td>
<td>3</td>
</tr>
<tr>
<td>C5/6</td>
<td>7</td>
</tr>
<tr>
<td>C6/7</td>
<td>2</td>
</tr>
<tr>
<td>Combination of 2 areas</td>
<td>2</td>
</tr>
</tbody>
</table>

**Table 3:** Type of degenerative disease, as revealed by X-ray and MRI examinations

<table>
<thead>
<tr>
<th>Area</th>
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</tr>
</thead>
<tbody>
<tr>
<td>Plain prolapse</td>
<td>2</td>
</tr>
<tr>
<td>Prolapse and dorsal osteophytes</td>
<td>8</td>
</tr>
<tr>
<td>Disk osteochondrosis</td>
<td>2</td>
</tr>
</tbody>
</table>

**Table 4:** Implaspin replacement positions in X-ray images obtained during the 2nd and the 6th month after the surgery

<table>
<thead>
<tr>
<th>Biotitanium replacement position in X-ray images</th>
<th>Examination completed in the 2nd month</th>
<th>Examination completed in the 6th month</th>
</tr>
</thead>
<tbody>
<tr>
<td>No changes, wellpreserved lordosis without propagation into the spinal canal</td>
<td>12 patients</td>
<td>12 patients</td>
</tr>
</tbody>
</table>

**Table 1:** Evaluation of surgery success rate according to the JOA classification

**Results**

The group comprised 12 patients – 4 women and 8 men with an average age of 48 years. Tables 2 and 3 show the numbers of patients with medical problems in their intervertebral areas and types of diseases, as indicated by pre-surgery X-ray and MRI examinations.

The pre-surgery JOA score of our group was 11.6 points. During an examination conducted in the 2nd month the group received its JOA score of 15.5 points and 15.0 points in the 6th month. The success rate of surgery treatment based on the Implaspin replacement expressed as a percentage in accordance with the JOA classification was 72.2 % in the 2nd month and 62.4 % in the 6th month after the surgery. The success rate expressed as a percentage falls in the good result zone (both examinations) – see Table 1. Our clinical examinations also included X-ray examinations of operated areas with a focus on implant positions – see Table 4. Examinations of no patient revealed implant dislocation or damage – Fig. 2 and 3.
properties of glass ceramics became clear as well. The limiting factor concerning the use of glass ceramic replacements in the cervical spine area, like in the case of the lumbar spine, remains the implant strength linked to its size. Another limiting factor deriving from mechanical properties of glass ceramics became clear as well. The limiting factor concerning the use of glass ceramic replacements in the cervical spine area, like in the case of the lumbar spine, remains the implant strength linked to its size. Another limiting factor is the fragility of glass ceramics that requires elimination of contact with metal tools during surgeries. The application instrumentarium may not be attached to the glass ceramic replacement. When the replacement size was respected due to the significantly narrowed intervertebral area it was necessary to remove not only the degenerated disk and its cover vertebra, but also its adjacent cancellous tissue. That situation represents a risk of replacement propagation into the adjacent vertebrae bodies and creation of lateral stenosis. When the replacement size was insufficient it was damaged in the intervertebral area without fragment dislocation into the spinal canal. For the time being, similar complications were not observed among the group of patients operated by means of the Implaspin replacement produced from bioactive titanium. It seems that the properties of bioactive titanium of the LASAK, Ltd. minimize complications associated with the size and mechanical strength. Mechanical properties of the Implaspin replacement were compared to properties of the Ti-bone replacement made by the Biomed company. The Ti-bone replacement has been used in our clinical practice together with the Implaspin replacement since the year 2002. The surgery techniques and approaches of both implants are similar. The Ti-bone advantage rests in the fact that during surgery its size is determined through a model before its application into the interbody area. The Implaspin replacement does not allow that. Unlike the Ti-bone replacement, our replacement should feature benefits in the form of its shape, the wings on the opposite sides, and its surface treatment.

Implaspin is produced in the form of tapered prisims of various sizes. It provides better conditions for preservation of cervical spine lordosis in the post-surgery period, compared to the Ti-bone replacement shape. It looks like a prism but it is not tapered. The Implaspin replacement wings aiming towards adjacent vertebrae bodies and the compression increase stability and secure direct contact of the implant with bone tissue. As a result of that, the wings help fix the mechanically operated section during the first hours before the bone-implant chemical bond is produced. The Ti-bone replacement does not have those wings and mechanical stability of the operated section only depends on compression of neighboring vertebrae.

Thanks to the bioactive material application to the whole surface of the titanium replacement, a bone-implant chemical bond covering the whole surface that is in contact with adjacent vertebral bodies is produced within 24 hours. That chemical bond increases fusion strength. Osteoconductive properties of the bioactive surface facilitate the growth of osteoblasts on the replacement walls and they allow creation of bone intersomatic fusion of the operated section within a few weeks. The Ti-bone replacement contains a cavity filled with bioactive material. The surrounding supporting titanium features no bioactive properties. The chemical bond and the subsequent interbody fusion occur only at that location. The bioactive material is not homogeneous and that could slow down intergrowthing of osteoblasts during the creation of fusion. That subsequently increases the risk of the operated area’s instability. The Implaspin replacement does not have to be filled with bone grafts or hydroxyapatite in order to create fusion like in the case of the other replacements. The patient does not have to provide any graft. That shortens the time of the surgery itself and the patient is spared of another risk associated with obtaining a bone graft (pain, cosmetic effect).

These expected benefits of the Implaspin replacement, compared to the Ti-bone replacement, must be confirmed by long-term clinical results that are not available so far.

## Conclusion

Based on the first short-term results, the Implaspin cervical replacement perfectly combines the osteoconductive properties of glass ceramics with titanium strength. During a six-month monitoring the group of 12 operated patients demonstrated no complications associated with replacement damage, its dislocation into the canal or the prevetebral area. X-ray images showed positive surgery results and the JOA score was good as well. We have to wait for a final evaluation of results for at least 2 years. Now, it seems that the Implaspin replacement made by the LASAK, Ltd. applied in connection with indications of interbody fusion in the cervical spine area could be a perfect alternative to the other replacements.

## Literature

Surgical technique verification with the use of new bioactive titan cage in the treatment of degenerative disease of the lumbar spine – experimental study

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The authors performed experimental PLIF operations on cadavers to verify the appropriate shape and size of a cage to be made from bioactive titanium material. The suitable approach for application in the intervertebral region of the lumbar spine and appearance on X-ray and CT scans was also investigated.

In experiment on cadavers five implantations of interbody cages were performed, two of them in interbody space L3/L4 and three of them in L4/L5. Subtotal discectomy and end plates preparation was undertaken before cage insertion. The cage is made from bioactive titan material and fastened using an application instrument developed specifically for this purpose.

To verify the operational approach we used midline incision and opened the spinal canal through laminectomy, partial hemilaminectomy or partial medical facetectomy. This phase was followed by discectomy and end plates preparation. Stability of inserted cages in intervertebral space is ensured by rotation and anchorage of the cage wings in the end plates. Operated motion segment was removed and evaluated by X–rays and CT scans.

The operational approach through partial hemilaminectomy and partial medial facetectomy was fully sufficient for cage application with respect to the operated segment. All implantations were successful and position of cages satisfactory. Due to the use of a less robust gripping instrument, the approach was more efficient compared to a glass-ceramics cage and comparable with commercially produced cages with which we have experience (Spine Tech, Stryker, cage Comesa). Due to cage being radio opaque, proper position of cages is easy to control on X-ray and CT scans and allows good per-operative monitoring by X-ray magnifier.

Use of bioactive materials in spinal surgery began with glass ceramics. Based on our experience we determined suitable indications. To eliminate some of the disadvantages of glass ceramics (mainly mechanical properties), a cage made from bioactive material was developed. This type of cage does not require bone harvesting to fill the cage. This fact results in for better comfort for the patient in postoperative and eliminates complications from bone graft harvesting. The bioactive surface of the cage with osseointegrative features creates prerequisites for solid fusion without bone grafts. Titanium material guarantees mechanical strength and makes it possible to produce a wide range of shapes and sizes.

The strength of the material enables more secure gripping of the cage with the application instrument. The cage is easily visible in X-ray and in MRI scans artefacts are considerably reduced. The operational approach and technique are similar to those employed for other commercially produced cages and the extent of destabilization is limited to a minimum. Experimentally we repeatedly verified the operational approach, and the suitable shape and applicability of a bioactive titanium cage into intervertebral space. For clinical use bioactive titanium could be a possible method for replacement of bone grafts. For spinal surgeons it represents a chemically and mechanically stable material capable of interaction with environment in which it is implanted. Even in difficult conditions the level of osseo-integration is high.
Healing of cavitary bone defects

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Introduction

Cavitary bone defects and their management are common in orthopaedic surgery and may present after curettage of benign tumours (most often enchondromas) or as tumour-like lesions including solitary bone cysts, non-ossifying fibromas or aneurysmal bone cysts. These defects should be encountered during revision arthroplasty procedures, as well as in cases of compressive metaphysical fractures. Autologous bone grafts, allografts, or bone graft substitutes can be used for filling up of these defects. Autologous bone grafts remain the gold standard. However, there is an increasing popularity of bone graft substitutes, especially of resorbable materials like β-tricalcium phosphate (β-TCP). Cavitary defects differ in size, localization, availability of mesenchymal stem cells and growth factors from surrounding tissues. Patients with such defects differ in gender, age, health status and long-term used medication. The same factors have impact on the quality of bone defects healing. We reviewed the healing of bone defects and tried to find indications for β-TCP application.

Materials and methods

As already mentioned, materials used and further more compared in this study included autologous bone grafts, β-TCP and the mixture of autologous bone grafts with β-TCP. We included patients treated in our clinic in the time interval of 6 years, from 2003 to 2009, and main inclusion criteria to the study were bone defects in the upper or lower extremity, due to benign tumour or tumour-like lesion, and follow-up of the patients for at least 12 months after operation. Excluded from the study were patients with malignant tumours, immunosuppressed, users of corticosteroids and cytostatics, smokers. Pre-operatively in consideration were taken age, gender of patients, diagnoses, location of defects, dominant type of bone in the region of the defects (cancellous, corticocancellous, cortical bone) and volume of defects. The volume of the defect was calculated in cm³ as the rectangular parallelepiped of the anteroposterior, transverse and cephalocaudal dimensions of a lesion measured on two perpendicular radiographs. The largest dimension in each plane was evaluated according to Anker et al. 1. Post-operatively we considered type of filling material used, graft-related complications, possible occurrence of β-TCP in soft tissues. Patients were followed-up clinically and radiographically at 6 weeks, 3 months, 6 months and 1 year time. Radiographs were taken in standard projections and in same conditions. Two independent reviewers assessed each radiograph, by means of level of resorption of the filling material and trabeculation of the defect, by absolute numbers in cm³ and by relative values in percents (including deminishing of radiolucency around the filling material, growth of trabeculae and their joining with filling material used). The data of the two reviewers were then checked and averaged for each observation therefore resulting in a single value. Weight-bearing of lower extremity or loading of upper extremity was allowed at 6 weeks or 3 months time, depending on the size and location of the defect and patients complaints. Using a statistical analysis, we assessed a dependence of the healing of bone defects on their size, on patients’ age and gender, by means of regressive analysis, as well as the marginal volume for successful healing (we accepted an average value of 80% trabeculation as a rate of successful healing, by the end of the first year). We evaluated trabeculation of defects during follow-up in each group (autologous grafts, β-TCP and mixture), healing of defects depending on weight-bearing of the lower extremity and the type of the surrounding bone (cancellous, corticocancellous, cortical).

Results

In the time period from 2003 to 2009 selected patients included 49 men and 38 women of average age of 26 years (range 3–77 years, median 17 years). Primary diagnosis included bone cysts (44), enchondromas (24), non-ossifying fibromas (9), fibrous dysplasia (6), intraosseous ganglions (2), intraosseous lipoma (1) and chondroblastoma (1). Lesions were located in epiphysis or metaphysis.

Background: It has been a retrospective review of healing of bone defects, due to benign tumours or tumour-like lesions. We evaluated the therapeutical outcome in patients after use of either autologous bone grafts, or β-tricalcium phosphate (β-TCP), or mixture of both components. We tried to find indications for β-TCP application. Methods: The initial volume of defects, as well as their volume in the follow-up period resorption of the filling materials and trabeculation of the defects were measured using radiographs. By means of statistical analysis, we sought a dependence of bone healing on the initial volume of the defect, type of bone involved, patient’s age, gender and weight-bearing. Results: During the period of 6 years from 2003 to 2009, 87 patients with bone defects (average volume 15 cm³) were included. Significant factor for bone healing remains the volume of the defect, and in case of large defects, there was a restricted progress of trabeculation into the central part. Conclusions: Autologous bone grafts remain the gold standard in the healing of bone defects. β-TCP may be indicated in cases where there is insufficient amount of autologous bone grafts or in cases of bone defects up to 4 cm³ particularly in epiphysis or metaphysis.

Fig. 1: Course of trabeculation of filled defects depending on time

[98x598]3

[101x0]
in the femur (14), tibia (18), fibula (3), talus (3), calcaneus (9), humerus (12), radius (2), metacarpals (8) and phalanges (18). Affected types of bone were cancellous (45), corticocancellous (26) and cortical (16). The average volume of the defects was 15 cm$^3$ (range, 0.4–144 cm$^3$, median 9 cm$^3$). Twenty-eight defects were filled with autologous bone grafts, 44 with ß-TCP and 15 with a mixture of both components. The average follow-up period was 4 years (1–6 years).

We did not observe any significant complications after harvesting autologous bone grafts from iliac crest. There was a mild resorption of autologous bone grafts in the first 6 weeks followed by a rapid phase of trabeculation until the end of the first year after operation that continued with a slow phase process. Where ß-TCP was used, we noticed a slow resorption during the years and two phases of trabeculation—faster until first year after operation and slower thereafter. However, the trabeculation in case of ß-TCP was less intensive than in autologous bone grafts. Similar situation was in cases where mixture of ß-TCP with autologous bone grafts was used. In comparison with use of pure ß-TCP, we noticed a faster resorption and trabeculation in the first 6 months after operation (Fig. 1). In 11 patients, ß-TCP was identified within the soft tissues but was completely resorbed between 6 weeks and 3 months after operation. Within the end of the first year after operation, when the rapid phase of trabeculation was complete, we were able to evaluate trabeculation in relation to the original size of the defect. Correlation coefficient (c.c.) was assessed. We found out a strong dependence between the size of the defect in absolute values and trabeculation in case of autologous bone grafts (c.c. 0.98), in case of mixture of autologous bone grafts with ß-TCP (c.c. 0.95) and strong dependence in case of pure ß-TCP (c.c. 0.82). However, when we assessed dependence between trabeculation and size of defect in relative merits, there was no dependence (c.c. 0.48 for autologous bone grafts, c.c. 0.6 for pure ß-TCP, c.c. 0.14 for mixture of the both components) (Fig. 2). The process of resorption and trabeculation proceeded from the periphery to the central part of the defect. In large defects, there was a considerable resorption of the material and trabeculation in the periphery (absolute value of trabeculation volume), but the defect had reduced about 10% (relative merit). This was due to restricted progress of trabeculation into the central part of the defects. Simultaneously there was no dependence of trabeculation, in the end of the first year, on gender and age of patients (c.c. 0.1 for gender, c.c. 0.12 for age). Average trabeculation in the first year when autologous bone grafts were used exceeded 70% (Fig. 1). Therefore, we accepted an average value of 80% trabeculation as a rate of a successful healing, by the end of the first year. Defects with volume up to 4 cm$^3$ had the same successful rate of healing both for autologous bone grafts and ß-TCP. Only 40% of defects greater than 4 cm$^3$ just filled up with autologous bone grafts were completely healed (Fig. 3). ß-TCP was successfully used in defects with volume up to 4 cm$^3$ (Fig. 4). Trabeculation depended on type of surrounding bone. We observed slower resorption and trabeculation in cortical bones, especially when the defects were filled up with ß-TCP (Fig. 5). We noted two fractures in cortical bone in bone defects filled up with ß-TCP (1 year after operation in the first patient, 2 years after filling up of the defect and 1 year after osteosynthetic material removal in the second patient). Follow-up of the dependence of trabeculation on weight-bearing in case of lower extremity defects with additional osteosynthesis (10 patients) (Fig. 6), showed a rapid acceleration of trabeculation after full weight-bearing between 3 and 6 months after operation and the second mild acceleration of trabeculation, after the first year, when osteosynthetic material was removed (Fig. 7). During the removal of the osteosynthetic material, there was an opportunity to harvest a sample of tissue from the original defect for histologic assessment (Fig. 8).
Discussion

Healing of bone defects after curettage of present benign tumours or tumour-like lesions may be a suitable for understanding of healing process of the other bone voids, i.e., cases of loosening arthroplasties. The advantage of these lesions is their easy follow-up and calculation of their approximate volume and structural changes on two plane perpendicular radiographs according to Anker et al. Bone defect volume plays a significant role in the healing process. Resorption of bio-material and trabeculation processes start from periphery towards the central part of the defect and they are strongly dependent on oxygen diffusion. Due to central hypoxia, there could be formed fibrous tissue especially in large defects regardless of type of bio-material used. In the present time, there is an effort to reinforce the angiogenesis during the bone defect healing by means of tissue engineering—special matrices and gene therapy. However, there are differences among studies in vitro, animal models and in humans, and it is difficult to develop an experimental model with realistic clinical settings. We may notice rapid resorption of ß-TCP in experimental models with small defects up to 4 cm³. Despite the fact that the concentration of osteoinductive factors and osteoprogenitor cells is different, ß-TCP is a preferable bone graft substitute. ß-TCP is a well-tolerated bio-material in histologic samples. ß-TCP is well tolerated, though there has been reported eventual risk of macrophage reaction to damaged ceramic during implantation. ß-TCP added to autologous bone grafts replaces missing grafts. This composite material appears with osteoconductive, osteoinductive and osteogenic properties, despite the fact that the concentration of osteoinductive factors and osteoprogenitor cells is different and has positive impact on bone healing especially in the first 6 months after the operation. In case of the ratio of autologous bone grafts to ß-TCP in our study (1:2), the long-lasting results were similar like in pure ß-TCP. However, the results may differ in case of another value of rate. Considering, on one hand, that resorption and trabeculation of ß-TCP were significantly slower than in case autologous bone grafts were used, and on the other hand, it is quite pure mechanical properties, we may conclude that ß-TCP is preferable used in the location without bone bending overloading. Even though there is slower bone defect healing in regions where prevales cortical bone, ß-TCP is widely used in defects of metaphysis and epiphysis. In cases of large defects, additional osteosynthesis might be performed, to prevent possible pathological fracture. However, additional osteosynthesis may slow down the trabeculation in comparison with healing of defects without it (in case of ß-TCP average trabeculation in 1 year after operation—50% in cases without osteosynthesis, 35% in cases with additional osteosynthesis). Positive impact of weight-bearing and real loading on bone remodelling was obvious even after removing of osteosynthetic material.

Conclusion

Healing of bone defects was quickest when autologous bone grafts were used, thus it proved to be the best material for filling. Quality of healing depends on their size and thus corresponding nutrients supply from periphery to the central part of the defect. Reinforcing angiogenesis during healing could appear essential especially in case of large defects. It will be further task of our research. ß-TCP (Poresorb-ß-TCP®, Lasak Ltd, Prague, Czech Republic) is a reliable bone graft substitute and we suppose that it could be indicated to fill up defects in patients with insufficient amount of autologous bone grafts, or in cases where bone defects measure up to 4 cm³ particularly in epiphysis or metaphysis.

Conflict of interest None.
References


Fig. 8: Histologic assessment of tissue harvested from the original region of the defect filled by β-TCP during the removing of the plate 1 year after primary operation. Marked new lamellar bone formation
The use of interconnected β-tricalcium phosphate as bone substitute after curettage of benign bone tumours

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Objective: The purpose of this study was to analyse the clinical and radiological outcome in patients after implantation of β-tricalcium phosphate as a bone graft substitute to fill the defects after curettage of benign bone tumours and tumour-like lesions. Method: A total of 21 male and 26 female patients underwent the process of curettage of the tumour and filling of the bone defect with interconnected β-tricalcium phosphate in granule form. In 39 patients, β-tricalcium phosphate was exclusively used; in contrast, in 8 patients, it was combined with a cancellous autograft. The mass of implanted β-tricalcium phosphate ranged from 1.5 to 66 g (mean = 12.5 g). The clinical examination and radiographs were performed 24–96 months (50 months on average) after curettage of the tumour and implantation of the bioactive ceramics. Results: No patient complained of local pain, and all patients were satisfied with their limb function. Periodic radiographic assessments revealed that the material was incorporated in the surrounding bone without significant difference between implantation of β-tricalcium phosphate only and implantation of β-tricalcium phosphate mixed with autografts. Gradual resorption has started on the periphery and progressed centrally in both groups. Signs of the implanted β-tricalcium phosphate still remained radiographically in all 8 cases after implantation of synthetic material mixed with bone grafts and 27 of 39 cases after implantation of synthetic material only. The resorption was dependent on the mass of implanted β-tricalcium phosphate. In small defects with the mass of implanted material ≤3.5 g, we observed complete resorption of the material. The larger lesions with the mass of implanted material ≥3.5 g have healed more slowly, and β-tricalcium phosphate granules have been gradually resorbed but still remained radiographically distinct. Conclusion: According to our study, interconnected β-tricalcium phosphate is a safe and successful bone graft substitute for the treatment of benign bone tumours and tumour-like lesions because of its biocompatibility and bioresorbability.

Introduction

A variety of synthetic bone graft substitutes have been developed to fill bone defects in an effort to overcome the limitations of autografts and allografts. An ideal synthetic bone substitute should be a porous matrix with interconnecting porosity that promotes rapid bone ingrowth, and at the same time, it should possess a sufficient strength to prevent its crushing under physiological loads during integration and healing. Hydroxyapatite (Ca10(PO4)6(OH)2), β-tricalcium phosphate (Ca3(PO4)2), their derivatives and combinations are the most commonly used ceramic materials in bone surgery. While hydroxyapatite materials provide an osteoconductive matrix for bone ingrowth and ongrowth, slow in-vivo resorption profiles can potentially limit their clinical applications. Although β-tricalcium phosphate has been studied extensively in animal models and its biocompatibility, osseointegration and cell-mediated resorbability have been reported, there have been only limited data regarding long-term outcome of its clinical use in surgery for bone tumours. The information regarding biological responses such as bone bonding and resorption of ceramics is very important in clinical applications.

The purpose of this study was to analyse the clinical and radiological outcome in patients after implantation of β-tricalcium phosphate as a bone graft substitute to fill the defects caused by curettage for benign bone tumours and tumour-like lesions.

Materials and methods

There were 47 consecutive patients who fulfilled the following inclusion criteria: (1) histologically confirmed benign bone tumour or tumour-like lesion, (2) treatment by curettage of the lesion and implantation of beta-tricalcium phosphate, (3) follow-up after at least 24 months to confirm a static radiographic outcome without recurrence of benign bone lesion. A total of 21 were men and 26 were women. Their age at the time of surgery was 5–65 years with an average age of 19 years. The benign bone tumours and tumour-like lesions were located in the humerus (11 patients), femur (10), tibia (10), phalanx (7), metacarpal bone (3), calcaneus (3) and fibula (3). Histological examinations revealed that 18 were unicameral bone cysts, 11 were enchondromas, 8 nonossifying fibromas, 5 fibrous dysplasias, 4 aneurysmal bone cysts and 1 bone cyst in neurofibromatosis.

All patients underwent the process of curettage of the tumour and filling of the bone defect with β-tricalcium phosphate (Poreosorb®, Lasak Ltd., Prague, Czech Republic) in granule form with particle size of 0.6–2 mm. The porosity of the interconnected β-tricalcium phosphate scaffold was 35 ± 5%, the average macropore size was 100 μm in diameter, the size of micropores was 1–5 μm and the sintering temperature was 1180°C. The mass of implanted β-tricalcium phosphate ranged from 1.5 to 66 g (mean = 12.5 g). In 39 patients, β-tricalcium phosphate was exclusively used; in contrast, in 8 patients, it was combined with a cancellous autograft harvested from iliac crest. Internal fixation was employed in 8 patients because a pathological fracture occurred in 4 cases and an impending fracture was in 4 patients. Splints or bandages were used postoperatively for patients judged to be at risk of pathologic fracture. Full weight bearing was allowed after 8–12 weeks. The patients were scheduled for follow-up evaluations which included clinical and radiographic examinations at 4- to 6-week, 10- to 14-week, 6- and 12-month intervals. Thereafter, they were seen yearly. All patients were followed up 24–96 months (50 months on average) after curettage of the tumour and implantation of the bioactive ceramics. The clinical findings were evaluated according to the criteria of Enneking et al. The radiographs were taken in standard projections and assessed independently by three investigators (two orthopaedic surgeons and one independent radiologist). The data were then checked for interobserver agreement. In the case of disagreement, the patient’s radiograph was re-evaluated by all three observers together. To determine changes in the radiolucent zone surrounding the bioactive ceramics and to evaluate the resorption of β-tricalcium phosphate, a radiographic analysis was performed. We also investigated whether a difference of incorporation after implantation of β-tricalcium phosphate only and β-tricalcium phosphate mixed with cancellous autografts exists.
The values were presented as mean ± SD. The Mann–Whitney U test was used to determine differences between the periods necessary for disappearance of radiolucent zones in β-tricalcium phosphate only and β-tricalcium phosphate mixed with autografts. Comparison of implanted synthetic material mass in both groups was performed using the Kolmogorov–Smirnov test. The Aspin–Welch unequal variance test was applied to determine differences between the amounts of implanted synthetic material for completely and partially resorbed cases. A P value of less than 0.05 was considered significant.

**Results**

Neither postoperative infection nor adverse reaction due to the material was observed. No patient complained of local pain at final examination. All patients were satisfied with their limb function; the average limb function was 100%. Radiographs obtained immediately after surgery demonstrated radiolucent zones between the implanted ceramics and the surrounding bone. Over time, radiolucent zones faded and new bone developed in all 47 patients. The mean period necessary for disappearance of these zones was 9 weeks (range 5–13 weeks) without significant difference between implantation of β-tricalcium phosphate only and implantation of β-tricalcium phosphate mixed with cancellous autografts. Periodic radiographic assessments showed decreased radiographic density of β-tricalcium phosphate and replacement of β-tricalcium phosphate granules by newly formed bone trabeculae. These processes appear to have started on the periphery and progressed centrally, and remnants of the implanted material still remained distinct 3 years later.

Postoperative fractures occurred in two patients with a unicameral bone cyst in the humerus. One boy fell 3 weeks after surgery; the fracture was treated conservatively. The other patient has had a car accident 20 months after operation; the displaced diaphyseal fracture was treated with open reduction and plate osteosynthesis. In two young patients, growth arrest or deformity were seen before curettage of the lesion and implantation of β-tricalcium phosphate. Premature closure of the physseal plate with resulting shortening of the arm was found as a complication after pathological fracture of proximal humerus in one boy with unicameral bone cyst. Mild coxa vara was observed as a complication after repeated pathological fracture of proximal femur in one girl with fibrous dysplasia. Recurrences of the lesion were seen in only 4 cases (two unicameral bone cysts, two fibrous dysplasias). All patients had further curettage of recurrence in the area surrounding an incorporated ceramic material. In the skeletally mature male patient with a unicameral bone cyst, the cavity was filled with cancellous autografts; in other patients, β-tricalcium phosphate was added to the cavity at the time of repeat curettage.

**Discussion**

The synthetic bone graft substitutes were fabricated from a variety of materials, including calcium phosphates, calcium sulphates, bioactive glasses and glass-ceramics which appear to be the ideal substances for use as matrices because the inorganic component of bone is composed of calcium hydroxyapatite. Bioceramics can be divided into three categories: bioinert ceramics (alumina), surface-bioactive ceramics (sintered hydroxyapatite, bioactive glasses and apatite-wollastonite glass-ceramics) and resorbable bioactive ceramics (low-crystalline hydroxyapatite and tricalcium phosphate). The ideal bio-degradable substitute materials should fulfill some requirements such as biocompatibility, adequate initial strength and stiffness and retention of mechanical properties throughout sufficient time to assure its biofunctionality and nontoxicity of the degradation by-products.

The continuous degradation of a resorbable implant causes a gradual load transfer to the healing tissue, preventing stress-shielding phenomenon, and stimulates the healing and remodelling of bones. The surgeon should be concerned with the mechanical and biological properties of the bone substitute material as well as the handling and ability to assess healing of the grafted site. The small granules of the implanted β-tricalcium phosphate allow tight packing of irregularly shaped defects. This material can be only used in regions with intrinsic skeletal stability and the postoperative immobilization may be required due...
to its low mechanical properties because the synthetic filling contributes no significant structural support. Aside from chemical composition, the microstructure (the volume, density and size of pores and interconnections, the specific surface) acts on the bone ingrowth of porous materials. An increase in porosity would make bone ingrowth inside materials easier but would decrease their biomechanical properties. The interconnections in a porous biomaterial act only as pathways for nutritional elements, vascularization and cells between the pores that are the sites for bone tissue growth proceeding from the outside to the inside. Therefore, pore size should be larger than interconnection size 23. Macroporosity (pore size \( \geq 50 \mu m \)), microporosity (pore size \( \leq 10 \mu m \)) and pore wall roughness play a critical role in new bone formation 16, 24–29. Thus, the larger surface area can contribute to higher bone inducing protein adsorption, ion exchange and bone-like apatite formation by dissolution and reprecipitation, while the surface roughness enhances attachment, proliferation and differentiation of bone-forming cells 20. These data suggest that more extensive dissolution and reprecipitation of low-crystalline calcium phosphates can cause more osteoconductive and cell-mediated degradation characteristics 20.

Resorption is an important characteristic of biomaterials and can be divided in two mechanisms: solution-mediated dissolution processes and cell-mediated (phagocytic) processes. The degradation characteristics of calcium phosphates are dependent on the chemical composition, crystal structure, crystal and grain size, microporosity, neck geometry and neck dissolution rates of the materials 20. Pore density and interconnection density that expresses the quantity of connections between pores of porous materials play a more important role than their size that is modified during degradation of resorbable bioceramics, whereas the sizes and the densities are equally important in unresorbable biomaterials 23. This study presents the appropriate new bone formation with incorporation of the implanted \( \beta \)-tricalcium phosphate. In agreement with other authors, resorption was dependent on the defect size 9, 11, 31. In small defects with the mass of implanted \( \beta \)-tricalcium phosphate \( \leq 3.5 \) g, we observed complete resorption of the material. The larger lesions with the mass of implanted \( \beta \)-tricalcium phosphate \( \geq 5.5 \) g appear to heal more slowly, and the replacement of \( \beta \)-tricalcium phosphate granules by newly formed bone appears to have begun on the periphery and progressed centrally. Although the material was incorporated in the surrounding bone and gradually resorbed, signs of the implanted \( \beta \)-tricalcium phosphate still remained radiographically in 35 cases (74%). In our opinion, the implanted mass of biomaterial was higher in our patients compared to other authors 3. More experimental and clinical studies will be required in order to resolve the healing problems of large bone defects using osteoinductive factors and cell cultures. According to our study, interconnected \( \beta \)-tricalcium phosphate is a safe and successful bone.
References


Conflict of interest

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