ORIGINAL ARTICLE

Clinical study on the primary stability of two dental implant systems with resonance frequency analysis

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Abstract Primary stability has a major impact on the longterm success of dental implants. The aim of this study was to investigate the correlation of resonance frequency analysis (RFA) and insertion torque of self-tapping and non-self-tapping implants and their respective differences in primary stability. A group of 263 patients were treated with a total of 602 conically formed dental implants: 408 nonself-tapping Ankylos[®] and 194 self-tapping Camlog[®]. The maximum insertion torque during implant placement was recorded. Resonance frequency, measured as the implant stability quotient (ISQ), was assessed once immediately after insertion and twice 3 months later. Torque values of the non-self-tapping implants were significantly higher than those in the self-tapping group (p=0.023). RFA did not show differences between the 2 groups (p=0.956), but a correlation between ISQ values after implantation and 3 months after implant placement was measured (r=0.712). Within the implant systems, no correlation between insertion torque and resonance frequency values could be determined (r=0.305). Our study indicates that the ISQ values obtained from different implant systems are not comparable. The RFA does not appear suitable for the evaluation of implant stability when used as a single

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S. G. Köhler Meoclinic, Quartier 206, Friedrichstraße 71, 10117 Berlin, Germany method. Higher insertion torque of the non-self-tapping implants appeared to confirm higher clinical primary stability.

Keywords Resonance frequency analysis \cdot Insertion torque \cdot Primary stability \cdot Self-tapping implants \cdot Non-self-tapping implants

Introduction

Primary stability, defined as the biometric stability immediately after implant insertion, is an important prognostic marker for the success of dental implants. It is principally determined by the contact between the implant and bone. Primary stability is a major requisite both for direct bone deposit onto the surface of the implant and for its mineral tissue integration [26, 39]. The greater the primary stability, the smaller the micro-motions are between the surface of the implant and surrounding bone. This allows for uninhibited healing and osseointegration [12, 21]. Primary stability is influenced by various factors, including both material and local tissue dependent variables. These comprise the length and diameter of the implant, its design, the micro-morphology of the implant surface, the insertion technique and the congruity between the implant and the surrounding bone. Further important determinants are the quality and quantity of the bone [26, 40]. The greatest primary stability of dental implants can be reached with simple drilling; the use of additional thread cutters and bone condensers has been shown to lessen primary stability significantly [9].

Until now, the quantitative measurement of primary stability has been limited to invasive methods such as pull out and push out attempts and the calculation of removal torque. These biometric and destructive tests, common in ex vivo animal experiments, may also be used to determine the level of osseointegration after a period of healing and giving an indication of interfacial strength. They are, however, not suited to clinical use [3]. A scientifically established method of evaluating primary stability is the measurement of insertion torque as an invasive, single use technique [14, 15]. The maximum insertion torque is necessary to sink the implant into the prepared bone cavity [22]. In general, measured values lie between 5 and 50 Ncm. At present, it is unknown how much torque is necessary to achieve sufficient primary stability for individual implant systems [3]. Current opinions suggest that a minimum of 30 Ncm should be used. The higher the insertion torque, the higher the bone density. Studies have demonstrated a clear correlation between implant insertion torque or thread cutting torque and bone mineral density (BMD), which can be determined using micro-radiography [14, 15] or with the help of the dental quantitative computed tomography (DQCT) [6, 17]. For this reason, the measurement of insertion torque to quantify primary stability in vivo has become widely established. Indeed, insertion torque is integrated into the documentation systems of commercially available surgery motors [3].

Non-invasive resonance frequency analysis (RFA) is another, easily practicable method of measuring quantitative stability [20, 23, 24, 27] that can be used repeatedly in the intra-operative and post-operative settings [26]. The resultant Hertz waves are converted into a numeric value, the so-called implant stability quotient (ISO), to deliver values, which can be compared independently of the implant system used. The scale used ranges from 1 to 100 with the correlation to the resonance frequency value being almost linear. The higher the ISQ value, the more securely the implant is presumed to be anchored in the bone. Literature describes ISQ values for successfully integrated implants from 57 to 82 with an average of 69 ISQ after 1 year of use [5]. ISQ values of less than 50 should be viewed critically [3]. Another non-invasive technique for assessing the stability of an implant is the well-established Periotest[®] (Siemens, Germany) evaluation. This electronic instrument was designed to measure the damping characteristics of the periodontal ligament surrounding a tooth, and thus establishing a value for its mobility. As such, it has been widely used to measure implant mobility. There are, however, a number of variables that may influence Periotest[®] values: the vertical measuring point on the implant abutment, the horizontal distance of the handpiece from the implant and its angulation [26]. The use of this instrument is limited by its lack of resolution, poor sensitivity and susceptibility to operator variables. In comparison, resonance frequency analysis seems to be the more precise technique to measure implant stability [24].

Typically, tapered implants have a higher insertion torque than cylindric implants due to a greater frictional surface. The torque values for continual machine inserted implants are measured at 15-20% less than for corresponding values for discontinual manual insertion [38]. This can be explained by the intermittent adhesional friction component of discontinual manual insertion by wrench. The influence of implant geometry on torque has not been fully investigated until now, as is also the case for the target torque values for various implant systems. It is in part unknown which torque should be used to reach sufficient primary stability or to guarantee that immediate functional loading is possible, according to the implant system used [3].

The aim of this study was to investigate whether primary stability measured by RFA correlates with torque values both for self-tapping and non-self-tapping dental implant systems. Further investigation considered whether a relation exists between resonance frequency values immediately after implant insertion and after 3 months of submerged implant healing. Another aim of the study was the investigation of both implant systems of different morphologies with regards to their primary stability. For this purpose, torque maxima, RFA and the course and shape of insertion curves were compared. Most studies to date have focussed on implant torque maxima. Curve analysis is rare in existing literature and so, care was taken to include this aspect in the current study.

Materials and methods

Implants

A total of 263 patients were enrolled in this study. Of these patients, 165 were women and 98 were men. The age of the respondents ranged from 20 to 91 years at the time of the study (median 55.8 years). In total, 602 conical formed implants were inserted and 408 non-selftapping Ankylos® implants (Friadent, Mannheim, Germany) (Fig. 1) were used. The endosseous part of the implant features a special progressive thread design with the thread depth increasing towards the apex. This results in a reduction of functional stresses at the cervical section and crestal bone. A sharp edge and rough surface at the thread portion of the implant is produced by grit blasting. The very precise fit of the conical connector eliminates the microgap. The bacteria-proof seal and the lack of micro-movements of this rotationally stable connection provides excellent biological and mechanical stability [30]. The tapered implants are supplied in 3 diameters (3.5, 4.5 and 5.5 mm) and in 5 different lengths (8.0, 9.5, 11.0, 14.0 and

Fig. 1 Conically formed, nonself-tapping Ankylos[®] implant (Dentsply, Mannheim, Germany)



17.0 mm). All of these 15 types of implants were employed in this study. Furthermore, a total of 194 self-tapping Camlog[®] Root-Line-Type screw implants (Camlog Biotechnologies, Wimsheim, Germany) (Fig. 2) were inserted. This root-shaped implant system with its self-cutting thread is offered with diameters of 3.8, 4.3, 5.0 and 6.0 mm. Camlog[®] implants are available in lengths of 9.0, 11.0, 13.0 and

Fig. 2 Conically formed, self-tapping Camlog[®] root-line implant (Camlog, Wimsheim, Germany)



The implant system used in each insertion (i.e. selftapping and non-self-tapping) was randomly assigned. In all cases, the implants of both systems were inserted with simple drilling without the use of additional bone condensers, following the manufacturer's instructions. The lastused drill in each implant system, corresponding to the respective implant size, created a cylindrical hole of the desired width and depth. The last step in the preparation of the implant site for non-self-tapping Ankylos® implants involves reaming the bone to provide tapered implant site and tapping the threads. The bone quality at each placement was assessed by the operator during drilling of the implant bed, based on the hand-felt perception of the drilling resistance and classified according to the system described by Lekholm and Zarb in 1985 [25]. One hundred ninetyeight Ankylos® implants were placed in the maxilla and 210 in the mandible. With respect to the bone quality, 60 Ankylos[®] implants were inserted in bone type 1 and 7 implants in bone type 4. On the basis of the study by Trisi and Rao in 1999 [41], which demonstrated that hand feeling allows to distinguish bone type 1 and 4 but failed between the 2 intermediate classes, the remaining 341 Ankylos® implants were combined in 1 class of bone quality 2 and 3. One hundred thirty Camlog[®] implants were used for the maxilla and 64 implants in the mandible, comprising of 16 implants in bone type 1, 2 implants in



Fig. 3 Topographic distribution of all inserted implants with the FDI 2-digit notation (n=602)

bone type 4 and the remaining 176 implants in the combined class of bone type 2 and 3.

The study involved non-augmented delayed implant placements with the second-stage surgery occurring 3 months after submerged healing. Fixed restorations were installed after the second-stage surgery so there was no possibility for further measurements. The clinical observation period lasted for 12 months.

Insertion torque

During machine implant insertion, the necessary insertion torque for each implant was recorded with respect to time and the maximum end value was calculated in Ncm. The surgical drilling unit used, Frios® Unit E (Friadent, Mannheim, Germany), included an integrated measurement and data recording function. During the documentation of the torque values, the first 0.1 s was used to calibrate filters and to adjust the software accordingly. Subsequently, measurement values were recorded at 0.25 s intervals. Machine insertion was concluded as soon as the implant reached its end position and as soon as its rotation stopped due to friction with the periimplant bone tissue. The torque data gathered were saved on DOC-Chipcards. A card reader was used to transfer the data via the ImpDat® 2.1-Software (Kea Software, Pöcking, Germany) to a computer for evaluation.

Resonance frequency analysis

For 85 of the total 602 implants, resonance frequency analysis (RFA) was performed using the Osstell[®] (Integration Diagnostics, Göteborg, Sweden). Measurements were made immediately after implant insertion and compared to the insertion torque values. In a different group including 63 implants, a second measurement at re-entry procedure was recorded for comparable resonance frequency values.

To achieve this, a system-adapted transducer was attached to the internal screw threads after the removal of the cover screw using a hand screwdriver, which was connected to the measuring device. The transducer used

Table 1 Means and standard deviations (SD) of both implant systemsin terms of maximum insertion torque and resonance frequencyanalysis (RFA) (p < 0.05; t-test)

	Implant system	п	Mean	SD	Significance (<i>p</i>)
Insertion torque (Ncm) RFA (ISQ)	Ankylos [®] Camlog [®] Ankylos [®] Camlog [®]	408 194 27 58	28.8 25.9 66.5 66.6	15.2 14.7 8.4 7.3	0.023 0.956





implant system

Fig. 4 Box-and-whisker plots of the maximum torque values grouped by implant type

was a Type F 23 for the Ankylos[®] system and three transducer types F 17, F 18 and F 19 for the Camlog[®] system. The starting torque was supposed to reach a value ≥ 10 Ncm, although greater torque does not influence the measurements [23, 27]. Graphic representation of the results appeared on the device monitor. Three measurements were made per implant and the mean was calculated. Data transfer was performed via an infrared link from the measuring device to a computer via the Software Osstell[®] Data Manager, Version 3.0.

Statistical analysis

The torque and resonance frequency values collected were first tested for significant differences using the *t*-test with a significance level of 5%. The correlation analysis between both parameters and within the RFA for submerged implant

Table 2 Means of the resonance frequency analysis (RFA), maximum insertion torque and Pearson product–moment correlation coefficient (PMCC) (*r*) (level of significance $r \ge 0.5$ and $r \le -0.5$)

	п	Mean insertion torque (Ncm)	Mean RFA (ISQ)	PMCC (r)
Total	85	30.8	66.6	0.305
Ankylos®	27	34.8	66.5	0.621
Camlog®	58	29.0	66.6	0.095

healing was performed using the Pearson product-moment correlation coefficient (r). In this regard, correlations of $r \ge$ 0.5 and $r \le -0.5$ were considered significant. Statistical analysis was performed with the help of the SPSS software version 14.0 (SPSS, Munich, Germany). The diagrams of the insertion curves and the box-and-whisker plots were used only for descriptive purposes.

Results

The topographical distribution of the inserted implants is shown in Fig. 3. In general, it can be seen that the canine positions in the lower jaw and the region of the lower first molar were the most common implant locations. Quantitatively however, the maxilla was preferred. Of the 602 inserted implants, 9 (1.5%) had been lost after 1 year at the study's end. These were 8 (2.0%) of 408 Ankylos® implants and 1 (0.5%) of 194 Camlog[®] implants. This may be explained with reference to the corresponding implant sites. Five of the nine lost implants were placed in the posterior region of the maxilla. However, none of them was inserted in bone type 4, and only one of the nine lost implants was used for bone type 1.

The comparison of the insertion torque maxima of all 602 implants of both systems (Table 1) shows a higher median value for the non-self-tapping Ankylos® system (28.82 Ncm) than for the self-tapping Camlog[®] system



insertion torque [Ncm]

implant system

Fig. 5 Box-and-whisker plots of the maximum torque values grouped by implant type



implant system

Fig. 6 Box-and-whisker plots of the resonance frequency analysis (RFA) grouped by implant type

(median 25.86 Ncm). This difference was statistically significant (p=0.023). The greatest distribution with the highest values was seen for the Ankylos® system (Fig. 4). The difference in the resonance frequency values for both implant systems was not statistically significant (p=0.956).

Comparisons of the maximal insertion torque with the values measured by resonance frequency analysis directly after insertion (Table 2) in a second study group of 85 implants (27 Ankylos[®] and 58 Camlog[®] implants) by calculation of the Pearson product-moment correlation coefficient (r) yielded no correlation (r=0.305). A significant correlation could be measured between insertion torque and RFA (r=0.621) only in the case of the Ankylos[®] system. The greatest median value for maximal torque was measured at 34.8 Ncm, also for the Ankylos® system

Table 3 Means of the resonance frequency analysis (RFA) immediately after insertion and after 3 months and the Pearson productmoment correlation coefficient (PMCC) (r) (level of significance $r \ge$ 0.5 and $r \le -0.5$)

	п	Mean RFA (ISQ) (immediately post-op)	Mean RFA (ISQ) (3 months post-op)	PMCC (r)
Total	63	66.5	66.8	0.712
Ankylos®	37	67.9	66.5	0.667
Camlog®	26	64.4	67.3	0.828

(Camlog[®] 29.0 Ncm). The median resonance frequency values of both systems are practically identical, 66.5 ISQ for the Ankylos® system and 66.6 ISQ for the Camlog® system. The distribution of the values was greatest for the Ankylos[®] system, both with respect to the insertion torque and RFA (Figs. 5 and 6). None of the differences were statistically significant. With regards to the bone density and according to the study of Friberg et al. [13], we divided the implant sites into three groups (i.e. soft, medium and dense bone) by means of the insertion torque values. The corresponding resonance frequency values at implant insertion were plotted against these groups. Statistical analysis showed a significant difference between the two implant systems only in soft bone (p=0.002). No significant differences were detected in the other groups (medium bone p=0.097, dense bone p=0.39).

Comparisons of resonance frequency values for individual implant systems in the immediate post-operative period and after 3 months of submerged healing (Table 3) yield a correlation within the RFA (r=0.712).

Analysis of the insertion curves of both systems shows that both lack an initial insertion curve. For the Ankylos® system, both the insertion time and the maximal insertion torque fall with increasing diameter. The shape of the curve is concave with a clear increase in torque shortly before reaching the correct insertion depth. A peak (a short and very high rise) in torque of 178.5 Ncm could only be seen in the insertion curves for the smallest implant diameter. The insertion curves for the Camlog® system show no shortening of the insertion time with decreasing implant diameter. A lessening of the maximal torque dependent on the diameter or length of the implant could not be measured either. The curves tend to be linear with no terminal torque increase. A peak of up to 175.8 Ncm could be measured for all implant sizes. Figures 7 and 8 show examples of the insertion curves of both systems with respect to a diameter/ length ratio.



Fig. 7 Insertion curve of the Ankylos[®] implants (diameter of 3.5 mm/ length of 11 mm) (n=124)



Fig. 8 Insertion curve of the Camlog[®] implants (diameter of 3.8 mm/ length of 11 mm) (n=25)

Discussion

The primary aim of this study was to evaluate the resonance frequency values to quantify the primary stability of selftapping and non-self-tapping dental implants using invasive torque measurements as a comparison. Within the limitations of this study, it has been shown that higher torque values were reached using the non-self-tapping Ankylos® implant system, a fact that indicates higher primary stability. Successfully inserted implants were associated with torque values greater than 32 Ncm [36] or greater than 35 Ncm [31]. Neither the Ankylos[®] system with a mean insertion torque of 28.8 Ncm nor Camlog® implants with a mean insertion torque of 25.9 Ncm achieved the suggested minimum of 30 Ncm insertion torque for a sufficient primary stability [3]. However, the 98.5% survival rate of both systems seems to be comparable with reported values in the literature of successful implant treatment [12].

Both systems show nearly identical ISQ values. This conformity between the resonance frequency values, despite the significantly higher torque values of the two systems, has been shown in another study [3]. However, the study also describes higher median torque values for a self-tapping system in comparison with a non-self-tapping system. Other studies also demonstrate higher insertion torque values and significantly higher ISQ values for the self-tapping system [34]. Both primary stability and torque appear to be dependent on the system design [10]. A tapered implant design brings higher primary stability than straight cylindric implant geometry [37], although no negative bone tissue reactions were observed [35]. It should be noted that, from a biological perspective, an increase in primary stability does not dictate better implant success [3].

As in this study, many others have failed to demonstrate a correlation between RFA and insertion torque or removal torque [1–3]. A comparison of both torque values did confirm a significant correlation [2] whereby lower removal torque values were measured compared to the insertion torque values [1]. In one study, strong correlations were observed between insertion torque values, resonance frequency values and bone density using computerised tomography. It was concluded that initial implant stability could be predicted from pre-surgical CT diagnosis [42].

In this study, the non-self-tapping system demonstrated a minimally higher primary stability compared with the selftapping system, as measured with RFA. ISQ values of different implant systems do not, however, appear to be comparable [3, 11]. Moreover, there is as yet no evidence for a universally "critical" ISQ value, under which implants are not primarily stable [3]. This value should be individually determined for each implant system [11]. In light of these facts, it was possible to make an early identification of implants in incomplete stages of healing, using RFA to look for decreases in the ISQ value [16]. Another study, however, failed to confirm RFA as a dependable diagnostic instrument for the identification of mobile implants [29]. Overall, RFA measurements only have predictive value for implant stability when used repeatedly over a longer time period, whilst single measurements are meaningless [43].

A correlation of resonance frequency values for a single implant system, measured as the ISQ values immediately after implant insertion compared with those measured 3 months later after second-stage surgery, has also been confirmed in other studies [7]. However, a study failed to demonstrate a relation between the RFA after implant placement and 6 months later. This was neither the case for immediate functional loading nor in cases involving submerged healing [32]. A very interesting investigation demonstrated significant differences in resonance frequency at implant insertion between implants placed in soft, medium and dense bone, based on cutting torque values. One year later, no differences were detected, suggesting that implant stability in soft bone sites seems to "catch up" over time [13].

Implant diameter appears to have the greatest influence on implant stability [33], whereas implant length, localisation or bone level shows no adverse impact [18]. Other studies contradict these results, finding no correlation between implant stability and length or diameter of an implant [5, 8]. In general, RFA is influenced by various factors, such as the height, width and density of periimplant bone. Striking differences can be observed in cases involving lower boundary density and greater boundary thickness [19]. No correlation could be demonstrated between bone density, height or localisation using dental computed tomography studies [6]. Overall, the thickness of the corticalis appears to be the decisive determinant for primary stability [28]. Furthermore, the procedure of under-dimensioned drilling seemed to increase primary stability [37].

The insertion curves of the conical formed, non-selftapping Ankylos® system tend to be concave with a clear rise shortly before the end of the insertion. This also suggests that it has greater primary stability compared to the root-shaped Camlog[®] system, which has a nearly linear curve without an abrupt increase in torque. The course of the insertion curve depends on implant geometry. The curves measured for cylindric implants show an initial insertion curve followed by a nearly linear increase in torque with a clear peak shortly before the correct insertion depth is reached. The effective cutting surface remains constant during the entire insertion process where the greatest part of the torque results from the increased resistance to thread cutting within the corticalis. The constant increase in torque can be attributed to the increase in frictional surface. With tapered implants by contrast, no initial insertion curve can be demonstrated. In these cases, the increasing cutting torque results from an increase in effective surface which is to be cut, which, by analogue to the cylindric design, causes a constantly increasing frictional torque [38]. The increase in torque shortly before reaching the correct insertion depth (seen with the Ankylos[®] system) is caused by large frictional forces, which can be caused by the conical fit of the implant neck or by the apical point contact of the implant tip with the bottom of the pre-drilled opening. Deep insertion arising from high torque carries a risk of reduced circulation in the bone tissue and resultant necrosis in the implant carrier with connective tissue implant incision. The process of pre-cutting the threads also involves the danger of an expanded, slightly conical opening in cases where the axis of the thread cutter does not remain absolutely stable. The contact between implant and bone can be reduced from tip to neck in this event. When using cylindric implants, the complete thread length must be passed during insertion, which lengthens the insertion time greatly. Tapered implants, on the other hand, have the advantage of being able to be placed a certain depth into the pre-drilled bone cavity before the turning of the implant for insertion begins. This minimises the need for vertical space at the beginning of insertion and also reduces the danger of the implant being placed into the opening at a crooked angle, allowing the threads to be cut more easily. When cylindric implants are turned further than the depth of the cavity, bone threading is destroyed; a fact that is advantageous for the protection of neighbouring structures [38].

The manufacturer's recommended maximum torque of up to 70 Ncm was often exceeded in this study. Multiple peaks occurred with measurements of 178.5 Ncm. The peaks occurred during the insertion process and not, as is often the case, when the end position was reached or at the end of transmission. This extreme torque increase constitutes a risk of destroying the bone threads or the implant insertion interface or of causing an implant fracture [38]. It should be noted that in contrast to the experience with cortical screws, cases involving endosseous dental screw type implants primarily show incidents where bone threads are damaged. This can be explained by the fact that such implants are larger than other systems [4].

In summary, it can be concluded that the ISQ values of the RFA show no correlation to the measurable insertion torque maxima of the implant systems examined. Whilst significantly different median values for torque were measured, mean resonance frequency values were almost identical. It can be assumed from this discrepancy that the ISO values of different systems are not comparable. Within a single system, however, resonance frequency values do appear to show a correlation over time, although multiple measurements over a longer period are necessary to make a relevant judgement of implant stability. In light of these facts, RFA cannot be recommended as a single indicator for quantifying implant stability. It is surprising to note that the non-self-tapping Ankylos® implant system appears to have greater primary stability than the self-tapping Camlog® system. This can be seen in the system's greater torque values and in minimally greater ISQ values in RFA measurements immediately after implant placement and with respect to the insertion curves.

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