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REVIEW ARTICLE

The influence of the connection, length and diameter of an implant on bone biomechanics

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Abstract

Background. Regardless of the multiple options of connections, diameters and heights for dental implants, the clinician should know the biomechanical behavior of the bone to plan the treatment according to the biological and anatomical conditions of each patient, without risk to the long-term treatment success. ***Review.*** The following review attempts to summarize the relevant literature to establish guidelines for clinicians based on the scientific evidence regarding the influence by the implant's connection, diameter and length on the bone biomechanics. ***Conclusions.*** The length, diameter and connection of each implant have a degree of influence in bone biomechanics. Despite the influence of different implant connections, diameters and lengths on peri-implant bone stress and strain, these characteristics should remain within the physiological limits to avoid a pathological overload, bone resorption and consequent risk to the long-term success of implant-prosthetic treatment.

Key Words: *biomechanics, bone, dental implants, implant connection*

Introduction

In the last decades, major technological and scientific advances have occurred in dental implantology. The esthetic requirements of clinicians and high patient expectations continue to increase, as do advances in implant design and clinical techniques [1].

Many types of implants are available, which have different external designs, surfaces, platforms, connections, diameters and lengths [2]. This extensive range of options should be analyzed biomechanically by the clinician, according to the biological and anatomical conditions of each patient. The clinical success of rehabilitation by implant prostheses is predominantly related to the way that mechanical stresses are transferred from the implant to the surrounding bone without generating tensions that could endanger the longevity of the implants and prostheses [3,4].

The following review attempts to summarize the relevant literature to establish guidelines for clinicians based on scientific evidence regarding the factors that influence the biomechanical behavior of implant prosthesis. The objective of this review was to focus on studies that report the influence of the connection, diameter and length on an implant on bone biomechanics.

Biomechanics in implantology

The term *biomechanics* is related to the application of mechanical engineering to solve biological problems [5]. This field is primarily important in dentistry because the teeth, temporomandibular joint, maxilla and mandible undergo biomechanical activities during chewing and functioning [5].

The prognosis and long-term success of dental implant treatment are greatly influenced by the biomechanical environment in which they are exposed [6], in combination with the physical and geometric properties of each implant component [7]; the clinician's poor knowledge of biomechanical concepts is related to the failure of implant restorations [8]. Clinicians should know that the process and

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consequences of stress transmission to the surrounding bone depends on the nature of the applied load (the amplitude, direction and frequency), the implant design, the biological and biomechanical properties of the bone–implant interface and the bone tissue reaction to the mechanical environment created by the load over the implant [9].

The balance of forces and bending moments in bone do not allow teeth and implants to move freely in the maxilla or mandible. An external force produces an internal stress with reactions in the bone anchorage, that are of equal value but with opposite direction [10]. The interface between the bone and implant is mechanically the most vulnerable region of the prosthesis–implant system [11], predominantly because bone losses could be caused by complex biomechanical interactions between the implant and bone [12].

When a load is applied over an implant, the load is partially transferred to the bone, concentrating greater stresses at the cervical region of the implant [13]. This phenomenon is derived from an engineering principle, which states that, when a load is applied between two materials, the stresses are highest at the starting point [14]. In bone, the compressive forces tend to maintain the integrity of the implant–bone interface, whereas the tension tends to separate the integrity [15]. The tensile strength yield is 30% lower than that of the compressive forces in bone [11]. The cortical bone is more resistant to compression and less to shearing [11].

One of the most important factors to be considered in bone is its accelerated resorption by an excess of stress above its limit [4,16–18]. Although the level of bone is maintained by stress within a physiological range [19], an excessive increase in stress might create an imbalance in the function of osteoclasts and promote a significant bone loss [20–22]. Mullender et al. [23] have described the bone re-modeling process as a self-organizing biological control in which osteocytes act as sensors of mechanical signals, with each sensor producing a stimulus for regulating the nearby bone mass.

Osteocytes need mechanical stimulation to maintain their viability and survival and, if the stimulation does not occur, an apoptotic process might be induced [24]. When stress is maintained below the balance limit, it could activate the re-modeling process to reduce the strength of the entire bone, removing the cancellous and endocortical bone parts. When strain and stress exceed the limit, micro damage and microscopic cracks occur in the bone matrix, which lead to bone resorption [25]. A biomechanics signal change might start the re-modeling process in which bone tries to adapt to a new mechanical environment [26].

Frost [27] suggests a mechano-statistical theory, which provides that bone re-modeling remains inactive for loads that cause deformations in a range between 200–400 micro strains ($\mu\varepsilon$). If higher strains exist, an increase of bone density of between 400–3000 $\mu\varepsilon$, approximately, might occur, identifying an accumulation of micro damage in values over 4000–25 000 $\mu\varepsilon$. Frost [28] reported 4000 $\mu\varepsilon$ as the possible limit strain value to achieve a pathological bone overload. Duyck et al. [29], through a finite element analysis and based on computerized tomography scans, estimated that over 4.200 $\mu\varepsilon$ is the limit associated with a possible induction of bone resorption. Bone tends to gradually change in morphology in an attempt to adapt to a new external load [30].

Cehreli et al. [31] stated that the threshold values of physiological strains were not quantified in the jaws and, therefore, the clinical relevance of these absolute values of bone deformations are only speculative. In a study with monkeys, Melsen and Lang [32] evaluated the biological reactions of the alveolar bone during load application to the implants, verifying bone apposition in most cases in which strains varied between 3400–6600 $\mu\varepsilon$. Values above 6700 $\mu\varepsilon$ showed bone resorption and imbalance in the apposition–resorption process [32].

Stress distribution in surrounding bone might be influenced by various factors such as the implant position and angulation, implant–abutment connection and magnitude of the occlusal load [33,34]. The process and result of load transmission through the implant to the surrounding bone depend on the nature of the applied load (the direction, amplitude and frequency), implant design, biological and biomechanical properties of the bone–implant interface, bone quantity and quality and reaction of the bone tissue to the mechanical environment created by the implant load [9,35].

On the other hand, in cases of immediate implants subjected to functional loads, there might be an increased risk of biomechanical failure due to the excessive micromotion [36], considering that the most critical factors for implant's success and maintenance of the mechanical environment are good primary stability and minimal micromotion (50–150 μm) [37–39]. The bone–implant interface in the cases of immediate implants is considered even more critical because the implants are loaded into prepared alveoli and there is no complete contact of perimplantar bone at the ultrastructural level, in such a way that the primary stability is maintained only by the initial mechanical pre-load of the implant [39]. Occlusal contacts avoidance seems to be mandatory in single or even partial cases in immediate implants.

Stress and strain reduction in the peri-implant bone remains an important and complex objective for different implant designs [40]. Because of the wide variety of applications that use implants and the continued evolution of the clinical indications and protocols for prosthetic treatment, it is necessary to analyze stress and strain in the bone–implant system [17]. The optimization of an implant design may favor the mechanical environment for bone preservation [12].

Implant design

New implant designs and prosthetic components, as well as new clinical techniques, have been developed to prevent or reduce the alveolar ridge loss [1]. With the increase on the demands and clinical applications of implants, new designs have been studied and there are more than 20 geometric variations on the market [41]. This has lead manufacturers to develop different types of retaining screws that support higher torque values, change types of the implant material and create new designs for the implant–abutment interface [42].

Some implants might have common technical failures including prosthetic screw loosening, implant or prosthetic fractures and micro-displacement of crowns and/or crown-implant connection [33,43,44]. These failures generally do not lead to implant loss, yet they represent a significant problem for clinicians and patients and result in additional costs. To attenuate these difficulties, implants and abutments have been subjected to changes; there are a great variety of implants, connections and abutment geometries with different shapes, length, diameters, thread patterns and surface topographies [45].

The key factor in biomechanical implant systems is the connection, which is the interface between the implant and the abutment that keeps them together by force applied to the locking screw [46]. Bozkaya et al. [33] stated that implant design, characterized by the geometry and type of abutment connection, is an important factor in establishing the performance and maintenance of the osseointegration of implant-supported prostheses, which should determine the load transmission in the peri-implant bone as the implant–abutment interface. Assunção et al. [47] reported that high stress levels observed in the apical area of an implant might be related to the design, which, if modified, could greatly improve the distribution of stress over the trabecular bone. Lin et al. [48] showed that the influence of the connection type in the stress values over the bone was ~5%, which was less important than the position or load condition of the implant.

Implant designs might influence bone stress and could contribute to bone ridge loss [49]. Also, implant geometry plays an important role in improving the primary stability of immediate implants. For immediate implants, contact and friction are critical for their biomechanical behavior and subsequent success [50]. Biomechanically, the connections should reduce the tension on the prosthetic components and at the bone–implant interface, ensuring an adequate stability for the prosthesis [51]. The most commonly used implantology connection types are external hexagon, internal hexagon and Morse taper.

External hexagon

This connection type is formed by an implant with a hexaedrical platform in the upper region and a separate abutment screw; the unique structure of the hexagon does not allow platform rotation. The hexagon has a height of 0.7 mm that allows for an adjustment of the implant placement only during surgery [52].

The external hexagon is the most widely used design in implantology [53]; however, the external connection has been gradually modified and incorporated a mechanism to prevent prosthetic rotation [54,55]. One of the problems of this connection type is loosening of the abutment screw, which has a direct relationship to the rotational misfit at the abutment/implant interface that is related to higher rotational freedom with greater loosening of the screw [54,55]. The external hexagonal configuration determines a rotational position; however, it had no positive locking system [56]. Subsequently, a technique was developed by Binon and McHugh [54,55] to improve the problem of rotational misfit.

This system presents more micromotion, predominantly by the smallest settlement area between the implant–abutment junctions, which might cause clinical complications [42,57]. Most of the complications of external hexagon implants are related to the loss or fracture of the abutment screw, especially in a single prosthesis [52,58]. The external rotation center promotes a lower resistance under lateral loads, creating a possible gap in the implant–abutment interface, which might lead to bone resorption [59]. Weinberg [60] shows that this type of connection is particularly vulnerable because of its limited junction with its external components in combination with the presence of a small fulcrum point when oblique loads are applied to the implant.

Some factors responsible for abutment screw loosening in the external hexagon system are the vertical overload, the lateral load on the side of the non-working side or the structures with misfits [53]. The hexagonal platform limitations became more visible when its application was extended to partially edentulous arches or single-unit prostheses and the use of anti-rotational internal connections became more appropriate to such cases [53]. This platform resists the vertical loads; however, the abutment screw is the only one that absorbs lateral loads [6,56]. It was designed primarily to receive axial forces, which do not occur during chewing. During lateral loads, stress depends only on the contact angle between the hexagonal abutment and the implant [54,55], providing a lower distribution area compared to an internal connection system [52,61]. The region of the external hexagonal platform appears to determine a rotational position more than it absorbs lateral forces [56,62].

In an external hexagon connection, the abutment screw is uniquely responsible for maintaining the abutment during functional loads [31,51]. In this context, Kim et al. [44] found that a hexagonal system has smaller seating in the platform junction compared to internal hexagon systems. Pessoa et al. [6] compared the stress of the abutment screw connections between the external and internal hexagon, concluding that the stress was considerably higher in implants with an external hexagon. Similar results have been reported in studies [41,56,63] that identified more stress on the implant screw as well as on the implant when both types of hexagonal connections were compared.

Internal hexagon

This system consists of an implant with an internal junction at the neck level and a hexaedrical abutment with a separate screw. The abutment lateral wall facilitates dissipation of the lateral loads and the abutment screw protects it from excessive stress [6]. This system emerged as an evolution of the external hexagonal connection, which has advantages such as lower occurrence of abutment screw loosening and fracture [61], greater load absorption and better settlement at the abutment–implant interface [44]. This type of connection promotes homogeneous stress distribution around the implant, reducing the stress on the bone [64]. The above could be explained by the greater depth within the hexagonal junction of the implant, reducing the lever arm and changing the prosthesis–implant fulcrum to the middle region of implant [46]. This connection has a mechanism that provides a lateral support for external oblique loads, completely restricting the abutment and screw and increasing the resistance to screw loosening [44]. This mechanism produces better stress distribution on the bone [59] and promotes greater stability for the prosthesis retaining screw [61], reducing the risk of screw fracture and consequent prosthesis failure [51,65].

Compared to external connections in laboratory tests, internal connection systems have shown better results in stress distribution and greater anti-rotational stability [66]. Under oblique loads, the lateral abutment walls contribute to stress distribution, which is favorable for its use in a one-step surgical procedure and with single implants in areas with limited inter-occlusal space [59]. Tsuge and Hagiwara [53] stated that there is no evidence that an internal connection is better than an external connection. Gurgel-Juarez et al. [66] concluded that internal connections produce ~60.4% less stress on the trabecular bone compared to external hexagon systems. Baggi et al. [17] reported similar results in relation to the lower level of stress on the cancellous bone of the internal hexagonal system compared to that of the external hexagon. Other authors [6,59,67] have reported lower stress on the bone in internal hexagon systems. Pessoa et al. [6] demonstrated that the internal connection showed lower shear stress in the cervical region and, according to Stanford [68], the abutment of the internal connection is biomechanically more advantageous for the long-term success of prosthetics.

There are several types of internal connections on the market, including different internal hexagon types such as hexagonal, triangular and 'cloverleaf' designs [1].

Morse taper

This type of connection consists of an implant and a long, tapered screwed or press-fit abutment that precisely adapts to the inner part of the implant. The conical and deep interface between the implant and the abutment walls give resistance to lateral loads [6]. There is a convergence angle of 4–16° in the Morse taper, which increases the flexural strength of the abutment [42]. Geometric lock and taper friction are the basic principles of the abutment protection mechanism of an excessive functional load [56].

The Morse taper system emerged as an alternative to the external hexagon because of its better stability and smaller magnitude and quantity of micromotions [69]. In pure Morse Taper conception, the abutment is attached to the implant through an internal conical connection with no screw; the union is predominantly achieved by mechanical friction between the inner implant wall and the external abutment [70]. The mechanism of the pressure fitting the tapered abutment, combined with a screw at the end, appears to be responsible for the abutment protection during excessive loads [9]. The tapered geometry facilitates distribution of the loads, which are resisted primarily by its walls, allowing a retention by friction forces and preventing abutment motion or tilting [56,59].

The Morse taper system is favored by the seal between the abutment and implant, automatically inducing settlement in the axial direction [71]. The Morse taper connection virtually eliminated the abutment loss or fracture [72]. Biomechanical complications such as abutment screw loss or prosthetic component fractures occur with lower frequency in tapered connections, reducing the micromotion of the implant–abutment system [1]. Some authors have reported that the Morse taper system is a more stable system than the external hexagon [56,73,74]. Norton [73,74] reported that a tapered interface between an implant and abutment significantly improves the ability to resist bending moments. Compared to that of the external hexagon, tapered abutment improves the stress distribution in the peri-implant bone, reduces the bending effects and allows a better settlement with a locking and friction system [73,74], favoring its stability [56]. Morse taper implants present greater stability and less rotation because of the wide connection area as well as an increased resistance to lateral loads through the lateral wall of the abutment, which facilitates and promotes better stress distribution and sealing, avoiding micro-gaps [59].

Biomechanically, the internal region of the Morse taper system performs as a single body [75], with less internal micro-motion [1,69], acting as a stiffer body and concentrating more stress in the fulcrum point between the abutment and the implant [6].

The Morse taper abutment, located frequently at the marginal bone level, appears to decrease substantially the peak stress, improving the stress distribution in the surrounding bone. This effect could be explained by this system having a tapered interface, allowing shear stress to be located more apically, reducing stress and marginal bone resorption [76]. An increase in the depth of the abutment–implant connection promotes better stress distribution in the bone, primarily by increasing the contact area [33,40,41]. The taper degree of abutment does not affect the stress generated in the bone when subjected to vertical or oblique loads [40].

The tapered abutment–implant interface provides a decrease in the peak shear stress at the bone–implant interface compared to other connections [56,77]. The Morse taper system has better compliance with the Saint-Venant physical principle, which states that the local effects produced by any load acting on a body dissipates better in those regions that are more distant from the location of the load [78]. For 8 years, Doring et al. [79] clinically studied the complications associated with prosthetic components in Morse taper connection implants and found no mechanical complications such as screw loss or fractures.

Concerning the biomechanics over the surrounding bone, some authors [6,17,37,71] have observed that the Morse taper system had the lowest stress concentration, compared with the external or internal hexagon implant types, in conjunction with greater abutment stability and lower concentration stress over the internal hexagon system [6]. Other authors revealed in photo elastic studies with multiple conventional fixed prostheses that Morse taper implants showed higher stress values than external hexagon connections [80,81]. Streckbein et al. [82] reported that there is not a statistically significant difference in the stress distribution over the surrounding bone between the external hexagon, internal hexagon and Morse taper connections.

The implant connection is one of the factors that might influence the biomechanical behavior of prostheses over implants and the importance of the implant length and diameter are also frequently cited as important factors.

Implant length

Many implant models are available for specific clinical applications, with varying lengths and diameters. These designs are predominantly standard, long, short, wide or narrow types of implants [2]. Depending on the manufacturer, these implants are available in different shapes and configurations, with various thread depths [35].

The implant length appears to be a critical issue, with two contrary positions in the literature. Some authors [83,84] have claimed that one factor related to implant failure is the length, with the risk of failure being higher when the implant is smaller; other researchers [2,8,12,52,85,86] have stated that short implants produce higher stress and strain over the bone because of the smaller contact area at the implant–bone interface in combination with a less homogenous stress distribution [35]. Rubo and Capello Souza [87] showed that the stress decreased ~14% when the implant length increased from 10 mm to 13 mm. Tada et al. [85] recommended the use of long implants in patients with low bone density. Petrie and Williams [88] reported that an increase in the implant length might facilitate a decrease in bone deformation and maximum stress compared with short implants [86].

According to Petrie and Williams [88], a 1.65-fold increase in length reduces the stress on the bone ridge in narrow and tapered implants; however, if the increase in length exceeds 60% of the height of the cancellous available bone, the reduction is not significant. Many authors [13,89,90] declared that, under axial loads, stress variation is not significant between different lengths.

Pierrisnard et al. [90], in a finite element analysis, found that the shear stress during oblique loads in implants is concentrated on the first 7 mm and observed no differences related to the implant length. Renouard and Nisand [91] stated that failures could not be attributed only to the implant length. It appears that, whatever the implant length is in the cancellous bone, stress is always located at the implant neck level and has a slight influence on the peri-implant bone stress values [90]. According to Geng et al. [92], if the bone is of good quality (bone type I), the implant length does not influence the treatment success because the degree of inequality in the stress distribution over the interface depends on the difference in the elastic modulus of both structures; with better bone quality, there is less difference in the stiffness of these two structures.

For immediate implant cases, the literature reports contradictory statements about the influence of implant length on bone biomechanics based in finite element analyses. In a clinical follow-up study of Schnitman et al. [93], the authors conclude that 50% failures of immediate loaded external hexagon implants during 10 years were related with utilization of implants shorter than 10 mm. Miyamoto et al. [94] stated that length is a weak factor to primary stability in immediate implants cases, however longer implants, with more than 10 mm may result in a considerable strain reduction [95]. Kong et al. [50] concluded that greater implant length can reduce the cortical and cancellous bone stress and the implant displacement during axial loads in immediate implant cases.

In the literature, from a biomechanical point of view, there is no consensus that greater length is synonymous with better clinical performance [96].

Implant diameter

The appropriate selection of the implant diameter is generally based on the amount of available bone [97]. When the cortical bone thickness is not sufficient, the implant diameter might influence the success of treatment. When the height is limited, the use of larger implants might increase the bone–implant interface [25].

In some cases, surgeons try to use narrow implants, which are advantageous in treating bone defects and avoiding anatomical risk areas [1]. In the use of narrow diameter implants, the similarity of the survival rates to those of regular implants does not extend to the mechanical behavior [98,99]. Biomechanically, the implant diameter appears to influence the stress concentration at the implant as well as at the surrounding bone and, hence, affects the success rate [8,100]. An increase of the implant diameter, comparing the same load, produces a decrease in stress in the implant and the peri-implant bone [17,25,30,53,101,102]. Lower stress values in the bone and the implant have been observed in implants with larger diameter, which is justified as a better distribution by the largest contact area between the implant and peri-implant bone [12,25,40]. According to Oliveira [11], when the implant diameter increases by 0.25 mm, the contact surface is augmented by ~10%.

An increase in the implant diameter produces significant stress reduction, particularly in the cortical bone [25], whereas the length of the implant has a certain influence on the stress patterns in the trabecular bone implant interface [17,33]. Iplikcioglu and Akça [89] demonstrated that the implant length does not influence stress reduction over the peri-implant bone, in contrast to the implant diameter.

Petrie and Williams [88] concluded that the increase of the implant diameter decreases the stress in the alveolar crest 3.5-times, with a better effect for short and tapered implants because the diameter predominates over the length and taper. Kong et al. [103] reported that the narrowing of the implant neck also induces a more favorable stress distribution in the bone.

Kong et al. [50] reported that greater diameter can also reduce the cortical bone stress and implant displacement under lateral loads in immediate implant cases. In addition, the same researchers [50] concluded in a finite element analysis that the best combination is implants with a diameter higher than 4 mm and larger than 11 mm to achieve optimal biomechanics properties for immediate load cases. Other authors agree that implant diameter may reduce the bone strain and stress in immediate loaded implants [104,105], but it may not influence the implant micro-motion [105].

The diameter of the implant could be considered the most effective factor when compared with the length [102,106,107]. It has been suggested that increasing the implant diameter appears to compensate for the implant length [104,106].

Conclusions

In planning dental treatment with implants, the laws of biomechanics must be considered to prevent overload risks on the surrounding bone. Despite different implant connections, the diameter and length appear to influence peri-implant bone stress and strain; these characteristics should remain between the physiological limits to avoid a pathological overload, bone resorption and risk to the long-term success of implant–prosthetic treatment.

Overload is also a factor to be considered when planning implant prostheses. Despite the fact that the values of strength and load produced by human occlusion and mastication rarely surpass the numbers considered scientifically dangerous, care and good judgment must be used in order to submit implants and prosthetic junctions to normal moderate long-term forces.

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References

[1] Rodriguez AM, Rosenstiel SF. Esthetic considerations related to bone and soft tissue maintenance and development around dental implants: report of the Committee on Research in Fixed Prosthodontics of the American Academy of Fixed Prosthodontics. J Prosthet Dent 2012;108:259–67.
[2] Hasan I, Heinemann F, Aitlahrach M, Bourauel C. Biomechanical finite element analysis of small diameter and short dental implant. Biomed Tech (Berl) 2010;55:341–50.
[3] Van Oosterwyck H, Duyck J, Vander Sloten J, Van der Perre G, De Cooman M, Lievens S. The influence of bone mechanical properties and implant fixation upon bone loading around oral implants. Clin Oral Implants Res 1998;9:407–18.
[4] Kayabaşi O, Yüzbasioğlu E, Erzincanli F. Static, dynamic and fatigue behaviors of dental implant using finite element method. Adv Eng Softw 2006;37:649–58.

[5] Brunski JB. Biomaterials and biomechanics in dental implant design. Int J Oral Maxillofac Implants 1998;3:85–97.
[6] Pessoa RS, Muraru L, Júnior EM, Vaz LG, Sloten JV, Duyck J, et al. Influence of implant connection type on the biomechanical environment of immediately placed implants – CT-based nonlinear, three-dimensional finite element analysis. Clin Implant Dent Relat Res 2010;12:219–34.
[7] Meriç G, Erkmen E, Kurt A, Vaz LG, Sloten JV, Duyck J. Biomechanical effects of two different collar implant structures on stress distribution under cantilever fixed partial dentures. Acta Odontol Scand 2011;69:374–84.
[8] Akça K, Iplikçioğlu H. Finite element stress analysis of the influence of staggered versus straight placement of dental implants. Int J Oral Maxillofac Implants 2001;16:722–30.
[9] Cehreli M, Sahin S, Akça K. Role of mechanical environment and implant design on bone tissue differentiation: current knowledge and future contexts. J Dent 2004;32:123–32.
[10] Richter EJ. Basic biomechanics of dental implants in prosthetic dentistry. J Prosthet Dent 1989;61:602–9.
[11] Oliveira EJ. Princípios de bioengenharia em implantes oseointegrados. 1st ed. Rio de Janeiro: Pedro Primeiro Ltda; 1997.
[12] Qian L, Todo M, Matsushita Y, Koyano K. Effects of implant diameter, insertion depth, and loading angle on stress/strain fields in implant/jawbone systems: finite element analysis. Int J Oral Maxillofac Implants 2009;24:877–86.
[13] Lin CH, Kuo YC, Lin TS. Effects of dental implant length and bone quality on biomechanical response in bone around implants: a 3-D non-linear finite element analysis. Biomed Eng Appl Basis Comm 2005;17:44–9.
[14] Isidor F. Influence of forces on peri-implant bone. Clin Oral Implants Res 2006;17:8–18.
[15] Bidez MW, Misch CE. Force transfer in implant dentistry: basic concepts and principles. J Oral Implantol 1992;18:264–74.
[16] Geng JP, Tan KB, Liu GR. Application of finite element analysis in implant dentistry: a review of the literature. J Prosthet Dent 2001;85:585–98.
[17] Baggi L, Cappelloni I, Di Girolamo M, Maceri F, Vairo G. The influence of implant diameter and length on stress distribution of osseointegrated implants related to crestal bone geometry: a three-dimensional finite element analysis. J Prosthet Dent 2008;100:422–31.
[18] Lin CL, Lin YH, Chang SH. Multi-factorial analysis of variables influencing the bone loss of an implant placed in the maxilla: prediction using FEA and SED bone remodeling algorithm. J Biomech 2010;43:644–51.
[19] Frost HM. A 2003 update of bone physiology and Wolff's Law for clinicians. Angle Orthod 2004;74:3–15.
[20] Koyama Y, Mitsui N, Suzuki N, Yanagisawa M, Sanuki R, Isokawa K, et al. Effect of compressive force on the expression of inflammatory cytokines and their receptors in osteoblastic Saos-2 cells. Arch Oral Biol 2008;53:488–96.
[21] Flanagan D, Ilies H, Lasko B, Stack J. Force and movement of non-osseointegrated implants: an in vitro study. J Oral Implantol 2009;35:270–6.
[22] Hu K, Wang C, Zhang X. High pressure may inhibit periprosthetic osteogenesis. J Bone Miner Metab 2010;28:289–98.
[23] Mullender MG, Huiskes R, Weinans H. A physiological approach to the simulation of bone remodeling as a self-organizational control process. J Biomech 1994;27:1389–94.
[24] Galli C, Passeri G, Macaluso GM. Osteocytes and WNT: the mechanical control of bone formation. J Dent Res 2010;89:331–43.
[25] Okumura N, Stegaroiu R, Kitamura E, Kurokawa K, Nomura S. Influence of maxillary cortical bone thickness, implant design and implant diameter on stress around implants: a three-dimensional finite element analysis. J Prosthodont Res 2010;54:133–42.
[26] Xinghua Z, He G, Dong Z, Bingzhao G. A study of the effect of non-linearities in the equation of bone remodeling. J Biomech 2002;35:951–60.
[27] Frost HM. Bone's mechanostat: a 2003 update. Anat Rec A Discov Mol Cell Evol Biol 2003;275:1081–101.
[28] Frost HM. Perspectives: bone's mechanical usage windows. Bone Miner 1992;19:257–71.
[29] Duyck J, Ronold HJ, Van Oosterwyck H, Naert I, Vander Sloten J, Ellingsen JE. The influence of static and dynamic loading on marginal bone reactions around osseointegrated implants: an animal experimental study. Clin Oral Implants Res 2001;12:207–18.
[30] Lin D, Li Q, Li W, Swain M. Dental implant induced bone remodeling and associated algorithms. J Mech Behav Biomed Mater 2009;2:410–32.
[31] Cehreli M, Duyck J, De Cooman M, Puers R, Naert I. Implant design and interface force transfer. A photoelastic and strain-gauge analysis. Clin Oral Implants Res 2004;15:249–57.
[32] Melsen B, Lang NP. Biological reactions of alveolar bone to orthodontic loading of oral implants. Clin Oral Implants Res 2001;12:144–52.
[33] Bozkaya D, Muftu S, Muftu A. Evaluation of load transfer characteristics of five different implants in compact bone at different load levels by finite element analysis. J Prosthet Dent 2004;92:523–30.
[34] Kozlovsky A, Tal H, Laufer BZ, Leshem R, Rohrer MD, Weinreb M. Impact of implant overloading on the peri-implant bone in inflamed and non-inflamed peri-implant mucosa. Clin Oral Implants Res 2007;18:601–10.
[35] Bourauel C, Aitlahrach M, Heinemann F, Hasan I. Biomechanical finite element analysis of small diameter and short dental implants: extensive study of commercial implants. Biomed Tech (Berl) 2012;57:21–32.
[36] Chou H-Y, Müftu S. Simulation of peri-implant bone healing due to immediate loading in dental implant treatments. J Biomech 2013;46:871–8.
[37] Szmukler-Moncler S, Salama H, Reingewirtz Y, Dubruille JH. Timing of loading and effect of micromotion on bone-dental implant interface: review of experimental literature. J Biomed Mater Res 1998;43:192–203.
[38] Javed F, Romanos GE. The role of primary stability for successful immediate loading of dental implants. A literature review. J Dent 2010;28:612–20.
[39] Eser A, Tonuk E, Akça K, Cehreli MC. Predicting time-dependent remodelling of bone around immediately loaded dental implants with different designs. Med Eng Phys 2010;32:22–31.
[40] Chu CM, Huang HL, Hsu JT, Fuh LJ. Influences of internal tapered abutment designs on bone stresses around a dental implant: three-dimensional finite element method with statistical evaluation. J Periodontol 2012;83:111–18.
[41] Tang CB, Liu SY, Zhou GX, Yu JH, Zhang GD, Bao YD. Nonlinear finite element analysis of three implant-abutment interface designs. Int J Oral Sci 2012;15:101–8.
[42] Pita MS, Anchieta RB, Barão VA, Garcia IR Jr, Pedrazzi V, Assunção WG. Prosthetic platforms in implant dentistry. J Craniofac Surg 2011;22:2327–31.
[43] Geramy A, Morgano SM. Finite element analysis of three designs of an implant-supported molar crown. J Prosthet Dent 2004;92:434–40.
[44] Kim KS, Lim YJ, Kim MJ, Kwon HB, Yang JH, Lee JB. Variation in the total lengths of abutment/implant assemblies generated with a function of applied tightening torque in external and internal implant-abutment connection. Clin Oral Implants Res 2011;22:834–9.
[45] Quaresma SE, Cury PR, Sendyk WR, Sendyk C. A finite element analysis of two different dental implants: stress

distribution in the prosthesis, abutment, implant, and supporting bone. J Oral Implantol 2008;34:1–6.
[46] Yousef H, Luke A, Ricci J, Weiner S. Analysis of changes in implant screws subject to occlusal loading: a preliminary results. Implant Dent 2005;14:378–82.
[47] Assunção WG, Tabata LF, Barão VA, Rocha EP. Comparison of stress distribution between complete denture and implant-retained overdenture-2D FEA. J Oral Rehabil 2008;35: 766–74.
[48] Lin CL, Chang SH, Chang WJ, Kuo YC. Factorial analysis of variables influencing mechanical characteristics of a single tooth implant placed in the maxilla using finite element analysis and the statistics-based Taguchi method. Eur J Oral Sci 2007;115:408–16.
[49] Huang HL, Hsu JT, Fuh LJ, Tu MG, Ko CC, Shen YW. Bone stress and interfacial sliding analysis of implant designs on an immediately loaded maxillary implant: a non-linear finite element study. J Dent 2008;36:409–17.
[50] Kong L, Gu Z, Li T, Wu J, Hu K, Liu Y, et al. Biomechanical optimization of implant diameter and length for immediate loading: a nonlinear finite element analysis. Int J Prosthodont 2009;22:607–15.
[51] Nishioka RS, De Vasconcellos LG, De Melo Nishioka LN. External hexagon and internal hexagon in straight and offset implant placement: strain gauge analysis. Implant Dent 2009; 18:512–20.
[52] Pellizzer EP, Verri FR, De Moraes SL, Falcón-Antenucci RM, de Carvalho PS, Noritomi PY. Influence of the implant diameter with different sizes of hexagon. Analysis by FEM-3D. J Oral Implantol 2013;39:425–31.
[53] Tsuge T, Hagiwara Y. Influence of lateral-oblique cyclic loading on abutment screw loosening of internal and external hexagon implants. Dent Mater J 2009;28:373–81.
[54] Binon PP. The effect of implant/abutment hexagonal misfit on screw joint stability. Int J Prosthodont 1996;9:149–60.
[55] Binon PP, McHugh MJ. The effect of eliminating implant/abutment rotational misfit on screw joint stability. Int J Prosthodont 1996;9:511–19.
[56] Merz BR, Hunenbart S, Belser UC. Mechanics of the implant-abutment connection: an 8-degree taper compared to a butt joint connection. Int J Oral Maxillofac Implants 2000;15:519–26.
[57] Jemt T, Carlsson L, Boss A, Jörneús L. In vivo load measurements on osseointegrated implants supporting fixed or removable prostheses: a comparative pilot study. Int J Oral Maxillofac Implants 1991;6:413–17.
[58] Eckert SE, Wollan PC. Retrospective review of 1170 endosseous implants placed in partially edentulous jaws. J Prosthet Dent 1998;79:415–21.
[59] Maeda Y, Satoh T, Sogo M. In vitro differences of stress concentrations for internal and external hex implant-abutment connections: a short communication. J Oral Rehabil 2006;33:75–8.
[60] Weinberg LA. The biomechanics of force distribution in implant-supported prostheses. Int J Oral Maxillofac Implants 1993;8:19–31.
[61] Kitagawa T, Tanimoto Y, Odaki M. Influence of implant/abutment joint designs on abutment screw loosening in a dental implant system. J Biomed Mater Res B Appl Biomater 2005;75:457–63.
[62] Cibirka RM, Nelson SK, Lang BR, Rueggeberg FA. Examination of the implant-abutment interface after fatigue testing. J Prosthet Dent 2001;85:268–75.
[63] Freitas-Junior AC, Rocha EP, Bonfante EA, Almeida EO, Anchieta RB, Martini EP, et al. Biomechanical evaluation of internal and external hexagon platform switched implant-abutment connections: an in vitro laboratory and three-dimensional finite element analysis. Dent Mater 2012;28: e218–28.
[64] Bernardes SR, De Araujo CA, Fernandes-Neto AJ. Photoelastic analysis of abutment union to external and internal hexagons implants. Implant News 2006;3:355–9.
[65] Silva EF, Pellizzer EP, Villa LMR, Mazaro JVQ, Vedovatto E, Verri FR. Influência do tipo de hexágono e do diâmetro do implante osseointegrado na distribuição do estresse. Implant News 2007;4:549–54.
[66] Gurgel-Juarez NC, De Almeida EO, Rocha EP, Freitas AC Jr, Anchieta RB, de Vargas LC, et al. Regular and platform switching: bone stress analysis varying implant type. J Prosthodont 2012;21:160–6.
[67] Bernardes SR, De Araujo CA, Neto AJ, Simamoto Junior P, das Neves FD. Photoelastic analysis of stress patterns from different implant-abutment interfaces. Int J Oral Maxillofac Implants 2009;24:781–9.
[68] Stanford CM. Achieving and maintaining predictable implants esthetics through the maintenance of bone around dental implants. Compend Contin Educ Dent 2002;23: 13–20.
[69] Jaworski ME, Melo AC, Picheth CM, Sartori IA. Analysis of the bacterial seal at the implant-abutment interface in external-hexagon and morse taper-connection implants: an in vitro study using a new methodology. Int J Oral Maxillofac Implants 2012;27:1091–5.
[70] Salvi GE, Lang NP. Changing paradigms in implant dentistry. Crit Rev Oral Biol Med 2001;12:262–72.
[71] Dailey B, Jordan L, Blind O, Tavernier B. Axial displacement of abutments into implants and implant replicas, with the tapered cone-screw internal connection, as a function of tightening torque. Int J Oral Maxillofac Implants 2009;24: 251–6.
[72] Schwarz MS. Mechanical complications of dental implants. Clin Oral Implants Res 2000;11:S156–8.
[73] Norton MR. An in vitro evaluation of the strength of an internal conical interface compared to a butt joint interface in implant design. Clin Oral Implants Res 1997;8:290–8.
[74] Norton MR. Assessment of cold welding properties of the internal conical interface of two commercially available implant systems. J Prosthet Dent 1999;81:159–66.
[75] Akça K, Cehreli MC. A photoelastic and strain-gauge analysis of interface force transmission of internal-cone implants. Int J Periodontics Restorative Dent 2008;28:391–9.
[76] Hansson S. A conical implant-abutment interface at the level of the marginal bone improves the distribution of stresses in the supporting bone. An axisymmetric finite element analysis. Clin Oral Implants Res 2003;14:286–93.
[77] Hansson S. Implant abutment interface: biomechanical study of flat top versus conical. Clin Implant Dent Relat Res 2000;2: 33–41.
[78] Hibbeler RC. Mecánica de materiales. 2nd ed. Ciudad de México: Prentice Hall Hispanoamérica S.A; 1998. p 70–88.
[79] Doring K, Eisenmann E, Stiller M. Functional and esthetic considerations for single-tooth Ankylos implant-crowns: 8 years of clinical performance. J Oral Implantol 2004;30: 198–209.
[80] Tonella BP, Pellizzer EP, Falcon-Antenucci RM, Ferraço R, de Faria Almeida DA. Photoelastic analysis of biomechanical behavior of single and multiple fixed partial prostheses with different prosthetic connections. J Craniofac Surg 2011;22: 2060–3.
[81] Coelho Goiato M, Pesqueira AA, Falcon-Antenucci RM, Ferraço R, de Faria Almeida DA. Stress distribution in implant-supported prosthesis with external and internal implant-abutment connections. Acta Odontol Scand 2013;71: 283–8.
[82] Streckbein P, Streckbein RG, Wilbrand JF, Malik CY, Schaaf H, Howaldt HP. Non-linear 3-D evaluation of different oral implant-abutment connections. J Dent Res 2012;91: 1184–9.

[83] Goodacre CJ, Kan JY, Rungcharassaeng K. Clinical complications of osseointegrated implants. J Prosthet Dent 1999;81: 537–52.
[84] Sennerby L, Roos J. Surgical determinants of clinical success of osseointegrated oral implants: a review of the literature. Int J Prosthodont 1998;11:408–20.
[85] Tada S, Stegaroiu R, Kitamura E, Miyakawa O, Kusakari H. Influence of implant design and bone quality on stress/strain distribution in bone around implants: a 3-dimensional finite element analysis. Int J Oral Maxillofac Implants 2003;18:357–68.
[86] De Carvalho NA, De Almeida EO, Rocha EP, Freitas AC Jr, Anchieta RB, Kina S. Short implant to support maxillary restorations: bone stress analysis using regular and switching platform. J Craniofac Surg 2012;23:678–81.
[87] Rubo JH, Capello Souza EA. Finite-element analysis of stress on dental implant prosthesis. Clin Implant Dent Relat Res 2010;12:105–13.
[88] Petrie CS, Williams JL. Comparative evaluation of implant designs: influence of diameter, length, and taper on strains in the alveolar crest. A three-dimensional finite-element analysis. Clin Oral Implants Res 2005;16:486–94.
[89] Iplikcioglu H, Akca K. Comparative evaluation of the effect of diameter, length and number of implants supporting three-unit fixed partial prostheses on stress distribution in the bone. J Dent 2002;30:41–6.
[90] Pierrisnard L, Renouard F, Renault T, Barquins M. Influence of implant length and bicortical anchorage on implant stress distribution. Clin Implant Dent Relat Res 2003;5:254–62.
[91] Renouard F, Nisand D. Impact of implant length and diameter on survival rates. Clin Oral Implants Res 2006;17:35–51.
[92] Geng JP, Xu DW, Tan KB, Liu GR. Finite element analysis of an osseointegrated stepped screw dental implant. J Oral Implantol 2004;30:223–33.
[93] Schnitman PA, Whörle PS, Rubenstein JE, DaSilva JD, Wang NH. Ten-years results for Brånemark implants immediately loaded with fixed prostheses at implant placement. Int J Oral Maxillofac Implants 1997;12:495–503.
[94] Miyamoto I, Tsuboi Y, Wada E, Suwa H, Iizuka T. Influence of cortical bone thickness and implant length on implant stability at the time of surgery-clinical, prospective, biomechanical, and imaging study. Bone 2005;37:776–80.
[95] Georgiopoulos B, Kalioras K, Provatidis C, Manda M, Koidis P. The effects of implant length and diameter prior to and after osseointegration: a 2–D finite element analysis. J Oral Implantol 2007;33:244–56.
[96] Ortega-Lopes R, Olate S, Noia CF. Influencia del diámetro y longitud de implantes en la pérdida tardía de implantes dentales. Int J Odontostomat 2012;6:77–80.
[97] Bayraktar M, Gultekin BA, Yalcin S, Mijiritsky E. Effect of crown to implant ratio and implants dimensions on periimplant stress of splinted implant-supported crowns: a finite element analysis. Implant Dent 2013;22:406–13.
[98] Andersen E, Saxegaard E, Knutsen BM, Haanaes HR. A prospective clinical study evaluating the safety and effectiveness of narrow-diameter threaded implants in the anterior region of the maxilla. Int J Oral Maxillofac Implants 2001;16: 217–24.
[99] Romeo E, Lops D, Amorfini L, Chiapasco M, Ghisolfi M, Vogel G. Clinical and radiographic evaluation of small-diameter (3.3-mm) implants followed for 1-7 years: a longitudinal study. Clin Oral Implants Res 2006;17:139–48.
[100] Anitua E, Orive G. Finite element analysis of the influence of the offset placement of an implant-supported prosthesis on bone stress distribution. J Biomed Mater Res B Appl Biomater 2009;89:275–81.
[101] Ding X, Zhu XH, Liao SH, Zhang XH, Chen H. Implant-bone interface stress distribution in immediately loaded implants of different diameters: a three-dimensional finite element analysis. J Prosthodont 2009;18:393–402.
[102] Santiago-Junior JF, Pellizzer EP, Verri FR, de Carvalho PS. Stress analysis in bone tissue around single implants with different diameters and veneering materials: a 3-D finite element study. Mater Sci Eng C Mater Biol Appl 2013; 33:4700–14.
[103] Kong L, Hu K, Li D, Song Y, Yang J, Wu Z. Evaluation of the cylinder implant thread height and width: a 3-dimensional finite element analysis. Int J Oral Maxillofac Implants 2008;23:65–74.
[104] Ding X, Liao SH, Zhu XH, Zhang XH, Zhang L. Effects of diameter and length on stress distribution of the alveolar crest around immediate loading implants. Clin Implant Dent Relat Res 2009;11:279–87.
[105] Hsu JT, Fuh LJ, Lin DJ, Shen YW, Huang HL. Bone strain and interfacial sliding analyses of platform switching and implant diameter on an immediately loaded implant: experimental and three-dimensional finite element analyses. J Periodontol 2009;80:1125–32.
[106] Lee JS, Lim YJ. Three-dimensional numerical simulation of stress induced by different lengths of osseointegrated implants in the anterior maxilla. Comput Methods Biomech Biomed Engin 2013;16:1143–9.
[107] Vairo G, Sannino G. Comparative evaluation of osseointegrated dental implants based on platform-switching concept: influence of diameter, length, thread shape, and in-bone positioning depth on stress-based performance. Comput Math Methods Med 2013;2013:250929.