Influence of Connection Types and Implant Number on the Biomechanical Behavior of Mandibular Full-Arch Rehabilitation

Ravel M. Sousa, DDS, MS[1]/Paulo Cézar Simamoto-Junior, DDS, MS, PhD[2]/
Alfredo Júlio Fernandes-Neto, DDS, MS, PhD[3]/Jos Vander Sloten, MS, PhD[4]/
Siegfried V. N. Jaecques, MS, PhD[5]/Roberto S. Pessoa, DDS, MS, PhD[6]

Purpose: *To evaluate the influence of different implant numbers and connection types on the biomechanical behavior of mandibular full-arch implant-supported rehabilitation.* ***Materials and Methods:*** *Computed tomography–based finite element models comprising a totally edentulous mandible and 3.8 × 13-mm-diameter implants, abutments, abutment screws, bar retaining screw, and bar were constructed. Different implant numbers (three, four, and five implants) and loading conditions (symmetrical/balanced, unilateral, and posterior with diverse loading magnitudes) were simulated for both external hex and Morse-taper connections. The peak equivalent strain (EQV strain) in the bone and the peak of von Mises stress (EQV stress) in the abutment screw and bar retaining screw were evaluated.* ***Results:*** *Lower strain values were observed for a symmetrical loading distribution. Considering the same loading conditions, significantly higher bone strain levels were observed for external hex, compared with the Morse-taper connection. The number of implants had no significant influence on strain levels in bone, irrespective of the connection types. Compared with the external hex connection, the Morse-taper connection type presented significantly lower EQV stress values in abutment screws, but significantly higher stress in the bar retaining screw. Increasing the number of implants significantly reduced the EQV stress in the abutment screw and bar retaining screw.* ***Conclusion:*** *The Morse-taper connection type significantly decreased the strain levels in peri-implant bone, while increasing the stress in bar retaining screws. A smaller number of implants in an inferior full-arch rehabilitation slightly increased the stress in the abutment and bar retaining screws. Balanced adjustments of the loading improve the biomechanics of a mandibular full-arch rehabilitation.* Int J Oral Maxillofac Implants 2016;31:750–760. doi: 10.11607/jomi.4785

Keywords: *biomechanics, dental implants, edentulous mandible, finite element analysis*

Implant-supported fixed prostheses are considered to be a consistent treatment option for edentulous patients. Long-term studies have demonstrated that the edentulous arch can be successfully restored with implants supporting a full-arch mandibular rehabilitation.[1–8]

Four to six implants have been traditionally considered to be an adequate number to support mandibular full-arch prostheses.[5] However, the additional implants and components increase the treatment cost and lead to more invasive surgical procedures.[6] In addition, in some clinical situations, the limited bone quantity may

[1]PhD Student, Department of Fixed Prostheses, Occlusion and Dental Materials, School of Dentistry, Federal University of Uberlândia, Uberlândia, Brazil.
[2]Assistant Professor, Department of Fixed Prostheses, Occlusion and Dental Materials, School of Dentistry, Federal University of Uberlândia, Uberlândia, Brazil.
[3]Professor, Department of Fixed Prostheses, Occlusion and Dental Materials, School of Dentistry, Federal University of Uberlândia, Uberlândia, Brazil.
[4]Professor, Biomechanical Section, Catholic University of Leuven (K.U.Leuven), Leuven, Belgium.
[5]Researcher, Leuven Medical Technology Centre (L-MTC) and Biomechanical Section, Catholic University of Leuven (K.U.Leuven), Leuven, Belgium.
[6]Assistant Professor, CPBio – Research Center for Biomechanics, Biomaterials and Cell Biology, Department of Fixed Prostheses, Occlusion and Dental Materials, School of Dentistry, Federal University of Uberlândia, Uberlândia, Brazil; Biomechanical Section, Catholic University of Leuven (K.U.Leuven), Leuven, Belgium.

Correspondence to: Dr Roberto S. Pessoa, R. Olegário Maciel, 818, 1o. Andar, CENTRO, Uberlândia – MG CEP: 38400-084, Brazil. Email: rp@inpes.com.br

impair the correct distribution of a high number of implants in a favorable interimplant distance for cleaning and maintenance of the prosthetic rehabilitation.[7,8] In this way, efforts have been made to restore edentulous mandibles using a limited number of implants.[9–12]

However, biomechanical evaluations of mandibular full-arch rehabilitation have shown that, among other factors, the number of supporting implants has a decisive effect on load-sharing and consequent peri-implant stress/strain magnitudes.[13–15] The stress concentration can exceed bone's tolerance level, causing microdamage accumulation and inducing bone resorption.[16–18] Under certain conditions, this excessive occlusal loading may cause implant failure, even in well-osseointegrated implants.[19,20]

Nevertheless, recent developments in implant systems have contributed to a more favorable biomechanical environment in peri-implant bone. Finite element analysis (FEA) evaluating Morse-taper implant-abutment connections has shown lower concentration and better distribution of stresses/strains in peri-implant bone, compared with external hex connections.[21,22] The authors suggested that Morse-taper implants could better maintain the physiologic levels of strain in bone, principally in challenging clinical situations, such as implants in esthetic areas and in overloading chewing.[21,22] These assumptions have been corroborated by clinical assessments that have shown a smaller peri-implant bone loss for Morse-taper implants compared with other connection types.[23–26] One might speculate whether the use of such optimized implant designs could be an alternative to increase the predictability of the reduced number of implants in total-arch rehabilitations. However, it is unclear how different implant numbers and connection types could influence the biomechanical behavior of fixed implant-supported complete dentures.

On the other hand, adverse forces over the implant-supported prostheses could not only jeopardize osseointegration,[16,17,19,20] but could also cause abutment screw loosening and mechanical failures.[27,28] Although some biomechanical studies provided valuable information concerning the biomechanical behavior of total-arch rehabilitation, with regard to the number of supporting implants, measurements were limited to the peri-implant bone, and the implant components have not been modeled in detail.[13–15]

Therefore, the aim of this study was to evaluate by nonlinear three-dimensional (3D) FEA the influence of different numbers of implants and connection types on the biomechanical behavior of mandibular full-arch implant-supported rehabilitation.

MATERIALS AND METHODS

The computed tomography (CT) images of a totally edentulous dry mandible, from the Department of Anatomy of the Federal University of Uberlândia, were taken by a helical scanner CT Brightspeed Elite Select Multislice (GE Healthcare, NYSE: GE) with a gantry tilt of 0 degrees, at 120-kV acceleration voltage and 200-mA current. The projection data were exported using the Digital Imaging and Communication in Medicine (DICOM) file format. The data set had a voxel size of $0.35 \times 0.35 \times 0.625$ mm and consisted of contiguous slices with respect to the Z-axis.

Bone segmentation and reconstruction of mandible geometry were accomplished by thresholding within an image processing software (Mimics 15.1, Materialise). The 3D computer-aided design (CAD) solid models of 13-mm conical implants, with a 3.8-mm shoulder diameter, abutments, and abutment screws were obtained by reverse-engineering to resemble the commercially available 3.8×13 mm-diameter SIN UNITITE (SIN Sistema de Implante), with external hex and Morse-taper connections. The Morse-taper abutment and abutment screw are one single piece.

The implants were imported in Mimics (Materialise) and positioned within the mandibular bone between the mental foramens, with the shoulder at the bone level. Primarily, five implants were arranged: two of them 4 mm distance from the mental foramen on both sides, two in the canine region, and one in central position. Three implant configurations (three, four, and five implants) were then implemented, by removing two implants in the canine region (three implants), by removing the central implant (four implants), and maintaining all implants (five implants). These configurations were investigated for both external hex and Morse-taper connections.

The abutment and abutment screw models were subsequently aligned to the implants following the instructions from the implant manufacturer. The framework beam (ie, bar) was designed as a geometric solid, 6 mm high and 5 mm thick, in a horseshoe configuration following the shape of the mandible. Cantilevers measured 13 mm from the most distal implants, on both sides. Abutment components were afterward aligned over the abutments and glued to the bar by means of Boolean addition. Finally, the bar retaining screws were positioned in the abutment screws (Fig 1). No simplifications were made regarding the implant system macrogeometry (ie, truly spiral threads and implant and abutment internal geometries). The smallest elements in the constructed tetrahedral meshes were approximately 50 μm in size. Different levels of mesh refinement were used for feature recognizing (eg, at the threads). In addition, the bone mesh was refined

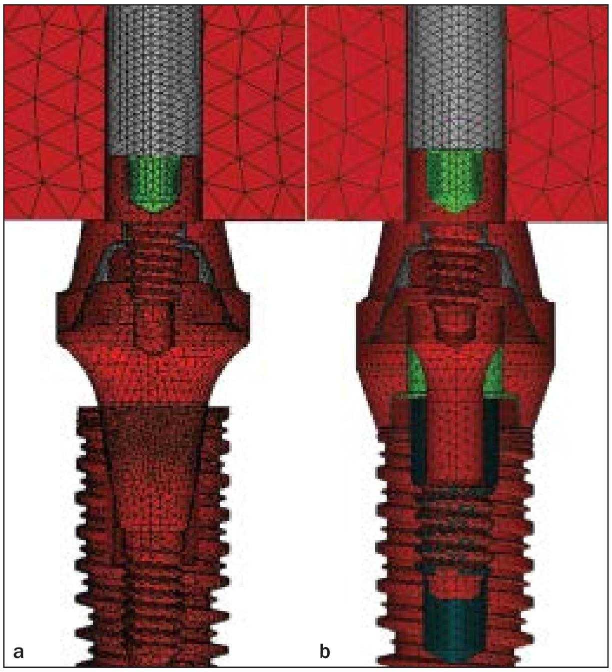

Fig 1 Alignment of bar, bar retaining screw, abutment screw, and abutment and implant. *(a)* Morse-taper implant system. *(b)* External hex implant system. Note that 3D finite element models are shown as sectional views.

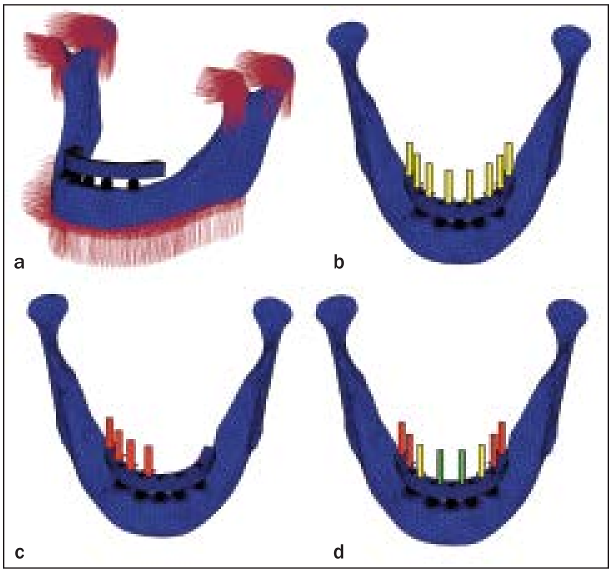

Fig 2 *(a)* Displacement constrained in all directions, at nodes on inferior border, condyles, and coronoid processes. *(b)* Symmetrical loading distribution. *(c)* Unilateral loading distribution. *(d)* Posterior to anterior decreasing loading distribution.

Table 1 Mechanical Properties of Bone, Implant, and Prosthetic Materials

	Materials		
Properties	Titanium	Cortical bone	Trabecular bone
Young's modulus (E) – [MPa]	110,000	13,700	1,370
Poisson ratio (ν) – [-]	0.33	0.30	0.30

at the bone-implant interface. All the abutments were 2 mm height from the implant shoulder. The implant insertion hole in the mandibular solid model was obtained by Boolean subtraction, between the bone and implant solids. Second-order effects resulting from tightening of the abutment and the preload in the abutment screw or bar retaining screws were not considered in the present study.

Bone, implants, abutments, abutment screws, bars, and bar retaining screw models were meshed separately in MSC.Patran 2010r2 (MSC.Software). Whenever necessary, adjustments in model meshes were made in 3Matic 7.0 (Materialise).

During meshing of the bone solid model, the entire volume that is contained within the outer bone surface was meshed. This means that the mesh consists of tetrahedral elements located in either cortical or trabecular bone. To discriminate between both tissues, different elastic properties were assigned, based on the grey values in the CT images.[29,30]

The values of the Young's modulus and Poisson ratio for the materials used in the present study were adopted from the relevant literature[30] and are summarized in Table 1.

For simulating the contact between the implant-abutment components, nonlinear frictional contact elements (Coulomb frictional interface) were used. Between the implant, abutment, abutment screw, bar, and bar retaining screw regions in contact, a frictional coefficient of 0.5 was assumed.[22,31,32] Frictional contact configuration allows minor displacements between all components of the model without interpenetration. Under these conditions, the contact zones transfer pressure and tangential forces (ie, friction), but no tension. For simulating the implant osseointegration, the bone-implant interface was assumed to be a bonded contact. In this configuration, no relative motion could occur at the bone-implant interface. The interface conditions remained the same, regardless of the FEA model.

Models were fully constrained in all directions at the nodes on the inferior borders of the model, as on the condyle and coronoid process of both sides (Fig 2a). Three different loading distributions were simulated.[33] In the first, total loading of 320, 400, and 480 N was symmetrically (ie, balanced loading, Sym) applied in eight nodes (ie, 40, 50, and 60 N at each point, respectively), over the entire extension of the bar (Fig 2b). In the second situation, total loading of 320, 400, and 480 N was unilaterally applied (Unilat), in four nodes (ie, 80, 100, and 120 N at each point, respectively), at the right side of the bar (Fig 2c). Finally, in the third situation, total loading of 480 N was distributed in eight points over the bar, in such a way that the loading magnitudes were decreasing from distal to medial (Post): four distal point loads of 80 N each (ie, two point loads at each side of the bar), two intermediate point loads of 60 N each (ie, one point load at each side of the bar), and two medial point loads of 20 N each (ie, one point load at each side of the bar; Fig 2d).

A total of 42 models were constructed by varying numbers of implants (three, four, and five implants), connection type (external hex and Morse-taper), and loading condition (Sym 320 N, Sym 400 N, Sym 480 N, Unilat 320 N, Unilat 400 N, Unilat 480 N, and Post 480 N). The analysis and postprocessing were performed for each model by MSC.MARC/Mentat 2010r3 software (MSC.Software).

The data for peak equivalent strain (EQV strain) in the bone, peak equivalent von Mises stress (EQV stress) in the abutment screw, and bar retaining screw were assessed. The average EQV strain in the bone (ie, the average of the sum of the peak EQV strain in each implant of each model) and the mean EQV stress in the implant system (ie, the average of the sum of the peak EQV stresses in the abutment and bar screws of each model) were analyzed using a general linear model analysis of variance (ANOVA, SAS/STAT statistical software, version 9.1, SAS Institute). This procedure allowed calculation of the percentage contribution of each of the evaluated parameters (number of implants, connection type, loading situation) and their interactions on the assessed results.[34]

RESULTS

Table 2 shows the results for the peak equivalent strain (EQV strain) in the bone, peak equivalent von Mises stress (EQV stress) in abutment screws and in bar screws for the two connection types, three implant numbers (three, four, and five implants), and seven loading conditions. The results of the analysis of variance on the relative contribution of each evaluated parameter (ie, mismatch size, connection type, loading magnitude, and clinical situation) are shown in Tables 3 and 4.

Considering the same loading condition, significantly higher bone strain levels were observed for the external hex connection type, compared with the Morse-taper connection (Figs 3 and 4). Nevertheless, in all the simulated loading conditions, the values of bone strains were below 4,200 $\mu\varepsilon$,[16,35] regardless of the connection type, implant number, and loading condition. Also, connection type presented the highest relative contribution (67.6%) for the differences found in bone strain levels. Implant number had little effect (6.2% of contribution) on strain magnitudes in bone, irrespective of connection types. However, loading magnitude and distribution greatly influenced the peri-implant strain concentration (21.9% of contribution). Lower strain values and a better strain distribution among the implants were observed for a balanced loading distribution (Sym 320 N, Sym 400 N, and Sym 480 N loading designs). Distal implants presented higher bone strain values, in comparison with medial implants, for all loading conditions. For the unilateral loading condition, the distal implant at the side of the loading application showed the highest peri-implant bone strain levels (Fig 5).

Considering the mean peak EQV stress in the implant system, the connection type (43%) and loading condition (35.2%) presented the highest contribution, respectively. However, the number of implants also significantly affected the peak EQV stress (11.4%).

Compared with the EQV stress in the abutment screw for different connection types, the Morse-taper connection type presented significantly lower stress values. For both connections, distal abutment screws showed the highest stress levels, in comparison with medial abutment screws, regardless of the implant numbers and loading designs (Fig 6). Implant number had no significant influence on EQV stress values in the abutment screw for Morse-taper connections. However, a decrease in EQV stress values in external hex abutment screws could be noted with the increase in the number of implants. Loading magnitude and distribution greatly influenced the EQV stress values for both connection types. Unilat and Post loading conditions induced significantly higher stress in abutment screws, compared with the balanced (ie, Sym) loading condition.

With respect to the peak equivalent von Mises stress (EQV stress) in bar retaining screws, significantly higher values of EQV stress were found for Morse-taper connection models, compared with the external hex connection, regardless of the loading design and implant number (Fig 7). For Morse-taper connections, distal bar retaining screws presented higher stress magnitudes, in comparison with medial bar screws, for the majority of the loading conditions, except for the Post condition. An inverse situation could be observed for external hex groups, in which the medial bar retaining screws showed higher stress values. In addition, EQV stress values in bar retaining screws were inversely proportional to EQV stress in abutment screws, for the external hex connection and MT connection in the Post 480 N loading condition. The number of implants also influenced the stress magnitudes in bar retaining screws. A slightly higher stress concentration in bar screws could be noted for the three-implant configuration, in comparison with the four- and five-implant

Table 2 Results for EQV Strain in Bone and EQV Stress in Abutment Screw and Bar Screw for All Simulated Models

Connection type	Loading condition	No. of implants	Peak EQV strain (με): Bone 1	2	3	4	5	Mean peak EQV strain (με): Bone
Morse-taper	Sym 320 N	3	519.9	–	280.9	–	565.2	455.3
		4	575.9	199.1	–	276.5	580.9	408.1
		5	532.2	290.5	233.1	284.4	465.8	361.2
	Sym 400 N	3	653.5	–	375.0	–	708.3	578.9
		4	716.8	318.7	–	376.0	720.2	532.9
		5	662.6	300.6	277.4	356.1	595.3	438.4
	Sym 480 N	3	782.9	–	421.7	–	858.7	687.8
		4	861.9	343.2	–	463.2	852.1	630.1
		5	799.9	369.0	367.0	413.8	708.2	531.6
	Unilat 320 N	3	909.0	–	255.2	–	170.0	444.7
		4	988.0	365.0	–	160.0	294.5	451.9
		5	933.4	379.0	221.6	175.0	255.4	392.9
	Unilat 400 N	3	1145.6	–	346.1	–	255.6	582.4
		4	1232.5	414.5	–	190.7	265.2	525.7
		5	1170.9	403.5	275.0	214.6	236.8	460.2
	Unilat 480 N	3	1372.1	–	403.0	–	243.0	672.7
		4	1476.0	507.0	–	229.4	312.2	631.2
		5	1402.0	492.8	348.0	260.0	270.5	554.7
	Post 480 N	3	1278.4	–	697.8	–	1401.0	1125.7
		4	1284.0	467.6	–	684.1	1283.0	929.7
		5	1136.0	406.8	523.5	636.0	1048.5	750.2
External hex	Sym 320 N	3	1455.8	–	497.6	–	1472.5	1184.9
		4	1653.0	221.0	–	296.5	1646.1	954.2
		5	1796.1	343.6	277.9	381.3	1583.2	876.4
	Sym 400 N	3	1820.1	–	625.2	–	1853.6	1432.9
		4	2052.7	286.9	–	449.9	2075.0	1216.1
		5	2250.0	425.7	337.0	455.3	2000.9	1093.6
	Sym 480 N	3	2184.1	–	753.5	–	2247.0	1728.2
		4	2460.2	391.8	–	539.7	2488.6	1470.1
		5	2719.0	495.5	398.7	534.9	2381.3	1305.9
	Unilat 320 N	3	2690.0	–	469.1	–	303.2	1154.1
		4	2920.9	417.6	–	183.0	404.0	981.4
		5	2634.0	493.6	353.1	178.0	256.7	783.1
	Unilat 400 N	3	3359.1	–	611.5	–	412.3	1461.0
		4	3647.8	562.3	–	250.2	529.1	1247.3
		5	3526.7	554.7	443.8	541.8	774.5	1168.3
	Unilat 480 N	3	4032.4	–	744.0	–	473.9	1750.1
		4	4380.0	710.5	–	281.6	646.0	1504.5
		5	3962.0	753.5	540.0	308.3	417.0	1196.2
	Post 480 N	3	2964.7	–	211.6	–	3019.0	2065.1
		4	3318.3	394.6	–	286.0	3196.2	1798.8
		5	3359.2	346.4	163.0	483.0	3108.5	1492.0

EQV strain = peak equivalent strain; EQV stress = peak von Mises stress; Sym = symmetrically, ie, balanced loading;

Table 3 Analysis of Variance for the Mean Peak Equivalent Strain in the Bone

Parameter	DF	SS	MS	*P* value	Contribution (%)
Connection type	1	5,882,199.5	5,882,199.5	< .0001*	67.6
Loading condition	6	1,907,315.2	317,885.9	< .0001*	21.9
Implant number	2	548,870.2	274,435.1	< .0001*	6.3
Connection type x Loading condition	6	162,320.4	27,053.4	< .0001*	1.9
Connection type x Implant number	2	119,373.7	59,686.9	< .0001*	1.4
Loading condition x Implant number	12	63,649.7	5,304.1	.0153*	0.7

*Statistically significant; $P < .05$. DF = degrees of freedom; SS = sum of squares; MS = mean square.

Peak EQV stress (MPa): Abutment screw					Peak EQV stress (MPa): Bar screw					Mean peak EQV stress (MPa)
1	2	3	4	5	1	2	3	4	5	Screws
38.3	–	15.6	–	38.3	148.6	–	82.8	–	119.8	73.9
40.3	15.7	–	14.4	40.5	120.5	116.2	–	128.5	145.0	77.6
37.7	14.1	15.8	11.9	35.5	131.0	117.1	64.5	137.0	118.7	68.3
48.6	–	23.3	–	51.4	171.4	–	77.9	–	185.7	93.1
52.2	20.3	–	24.2	56.6	153.9	112.3	–	109.5	160.9	86.2
57.9	17.1	17.6	16.9	52.3	102.9	114.6	119.4	112.0	137.0	74.8
75.9	–	38.1	–	67.0	185.6	–	100.5	–	183.1	108.4
62.8	32.4	–	38.2	71.0	149.6	112.8	–	104.7	176.7	93.5
69.8	23.0	23.7	20.8	55.2	129.0	115.4	101.9	126.3	145.0	81.0
71.7	–	15.5	–	10.7	155.7	–	79.4	–	151.0	80.7
78.3	35.5	–	9.1	10.4	153.2	115.0	–	76.5	143.2	77.6
94.5	23.6	14.7	7.6	9.7	136.0	143.4	87.0	75.0	127.0	71.8
88.0	–	37.6	–	13.0	220.0	–	130.1	–	79.3	94.7
92.1	34.9	–	11.9	16.9	178.1	157.9	–	117.8	166.3	97.0
90.0	38.5	18.2	9.6	13.9	167.0	176.4	110.5	79.5	159.1	86.3
102.3	–	24.0	–	20.0	179.2	–	109.3	–	149.2	97.3
108.6	35.8	–	15.5	19.3	176.0	168.0	–	115.0	148.0	98.3
104.4	40.1	35.0	11.7	13.0	184.1	159.0	110.0	73.0	145.0	87.5
114.8	–	70.3	–	124.1	157.2	–	412.4	–	148.8	171.3
109.4	51.0	–	59.2	98.1	150.1	213.7	–	209.1	136.4	128.4
83.6	48.9	39.4	46.7	85.4	122.9	181.0	186.2	163.0	118.9	107.6
159.0	–	154.5	–	113.9	93.8	–	122.5	–	117.8	126.9
133.2	105.9	–	115.9	137.4	88.9	123.1	–	149.0	98.7	119.0
122.5	136.6	70.7	100.5	111.3	80.3	85.5	122.9	90.5	84.0	100.5
156.6	–	153.8	–	141.6	106.9	–	116.7	–	88.6	127.4
155.4	128.2	–	123.9	137.2	96.2	141.6	–	149.9	101.5	129.2
120.6	126.1	70.2	103.2	131.9	94.3	91.9	121.3	113.0	96.1	106.9
165.7	–	154.9	–	140.4	112.9	–	127.3	–	100.7	133.6
160.2	104.4	–	109.8	142.1	123.3	123.5	–	147.1	121.5	129.0
122.2	135.2	69.5	99.3	139.1	95.8	91.8	125.7	110.8	102.3	109.2
162.5	–	153.4	–	148.7	122.0	–	121.1	–	92.3	133.3
110.5	106.6	–	124.1	137.3	81.4	163.0	–	122.3	91.3	117.1
150.0	142.1	70.8	128.0	127.8	105.0	121.0	143.0	104.7	88.9	118.1
108.3	–	114.0	–	141.5	156.1	–	148.1	–	106.1	129.0
107.8	106.8	–	124.9	146.3	107.0	159.2	–	154.2	91.8	124.7
100.9	121.4	68.6	116.6	141.0	116.1	94.8	125.3	125.1	89.6	109.9
108.3	–	112.6	–	143.5	165.0	–	171.0	–	90.8	131.9
105.3	133.1	–	127.5	139.4	135.0	176.1	–	154.5	104.0	134.4
108.5	122.4	110.9	120.7	117.5	201.0	120.1	130.0	98.0	91.0	122.0
158.1	–	132.7	–	139.0	191.8	–	256.2	–	198.6	179.4
163.3	103.9	–	111.6	151.4	79.9	209.8	–	219.2	108.3	143.4
156.2	107.8	130.1	102.9	133.5	164.2	171.9	190.5	133.6	106.3	139.7

Post = posterior with diverse loading magnitudes; Unilat = unilaterally applied.

Table 4 Analysis of Variance for the Mean Peak Equivalent Stress in the Implant System

Parameter	DF	SS	MS	*P* value	Contribution (%)
Connection type	1	11,978.7	11,978.7	< .0001*	43.0
Loading condition	6	9,813.1	1,635.5	< .0001*	35.2
Implant number	2	3,182.2	1,591.1	< .0001*	11.4
Connection type x Loading condition	6	770.5	128.4	.0203*	2.8
Connection type x Implant number	2	6.5	3.2	.9059	0.02
Loading condition x Implant number	12	1,722.9	143.6	.0077*	6.2

*Statistically significant; $P < .05$. DF = degrees of freedom; SS = sum of squares; MS = mean.

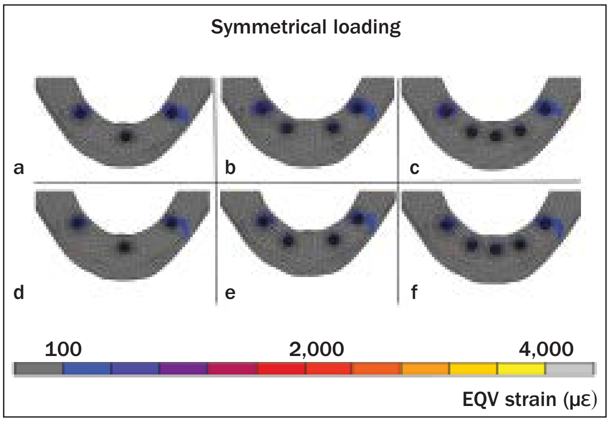

Fig 3 Occlusal view of EQV strain (με) distribution in bone for the external hex (EH) (*[a]* three implants, *[b]* four implants, *[c]* five implants) and Morse-taper (MT) (*[d]* three implants, *[e]* four implants, *[f]* five implants) models, for a symmetrical 480 N loading distribution. Note the higher strain concentration for EH connection, principally in distal peri-implant bone.

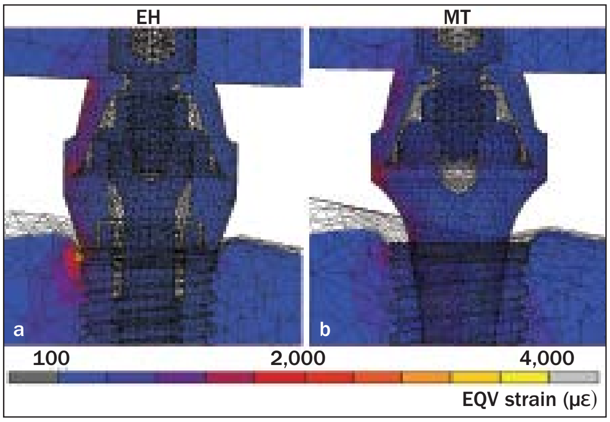

Fig 4 EQV strain (με) distribution in a median buccopalatal plane. *(a)* External hex implant system. *(b)* Morse-taper implant system. Note the higher strain concentration for external hex connection.

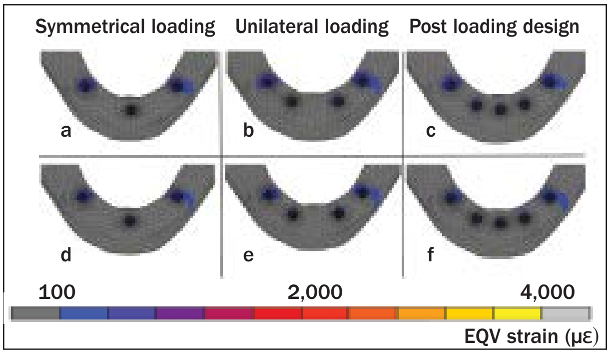

Fig 5 Occlusal view of EQV strain (με) distribution in bone for the external hex (EH) (*[a]* three implants, *[b]* four implants, *[c]* five implants) and Morse-taper (MT) (*[d]* three implants, *[e]* four implants, *[f]* five implants) models, 480 N force in different loading distributions. Note the better strain distribution and magnitudes for the Sym loading design.

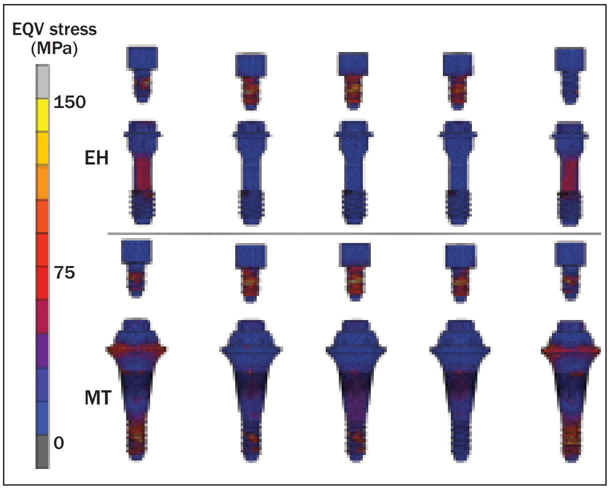

Fig 6 EQV stress (MPa) distribution on abutment screws of external hex and Morse-taper connections for five implants, Post 480 N models. Note the higher stress concentration on distal abutment screws and medial bar retaining screws.

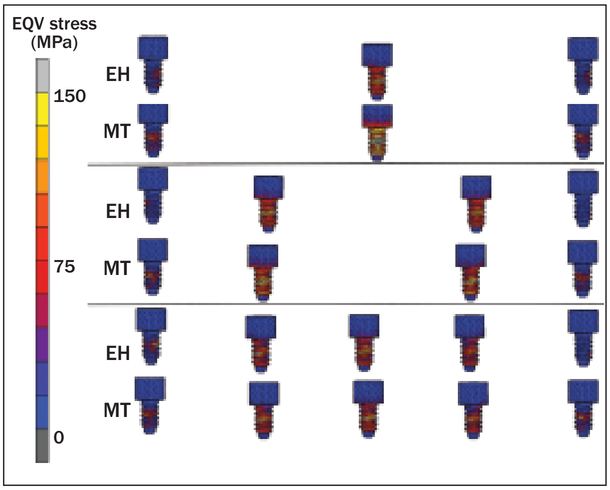

Fig 7 EQV stress (MPa) distribution on bar retaining screws of external hex and Morse-taper connections: three, four, and five implants, Post 480 N models. Note the higher stress concentration for three-implant bar retaining screws and for MT bar retaining screws.

configurations, regardless of the loading design and connection type (Fig 7).

DISCUSSION

The present FEA was carried out to evaluate the effect of different implant numbers (three, four, and five implants), connection type (external hex and Morse-taper), and loading condition on the biomechanical behavior of a mandibular full-arch implant-supported rehabilitation. It was demonstrated that implant number has a negligible contribution to the strains encountered in bone in delayed loaded implant simulations (ie, osseointegrated implants). However, although connection type and loading condition had a great influence on peri-implant strain levels, the values of bone strains were below the pathologic overload threshold,[16,35] regardless of the implant number, connection type, and loading design. In this parameter, Morse-taper presented lower strain magnitudes in comparison with the external hex connection type. On the other hand, the EQV stress in the implant system (ie, abutment screw and bar retaining screw) had an inverse relationship regarding the implant number. Notwithstanding, a balanced adjustment of the loading (ie, symmetrical loading distribution) improved the biomechanics of a mandibular full-arch rehabilitation.

The present FEA demonstrated a significant reduction in strain levels for the Morse-taper connection and for symmetrical loading distributions. Merz and coworkers[31] compared, by experimental and finite element methods, the stresses induced by off-axis loads on tapered and butt-joint connections. They concluded that the tapered interface distributed the stresses more evenly when compared with the butt-joint connection. In other finite element studies, Hansson[21] and also Pessoa et al[22] observed that a Morse-taper implant-abutment at the level of the marginal bone substantially decreased the bone stress peak. Moreover, it improved the stress distribution into the supporting bone. This can be explained by the difference in surface area between connections. The conical interface of the Morse-taper helped to dissipate the forces to the implant.[21,22]

Although precise determination of the loading level that separates mechanical loading into acceptable, osteogenic, or failure-inducing levels is difficult and until now unresolved, some authors focused on the bone strain amplitudes as the mechanical stimulus determinant to the bone adaptive process. In this regard, Duyck et al,[16] in an experiment in rabbit tibiae, proved that the stress/strain concentration, caused by an excessive dynamic loading, is capable of inducing marginal bone loss around osseointegrated implants. The authors estimated, by FEA based on CT images of tibiae samples, 4,200 με as the strain value associated with overload-induced resorption. A possible threshold for pathologic bone overload was also considered by Frost as 4,000 με.[35] In the present study, bone strain levels remained below 4,200 με, regardless of the connection type, implant number, and loading magnitude or distribution. These results are in agreement with long-term clinical data on implant and prosthesis survival, which demonstrated results that were similar regardless of the number of implants (ie, three, four, five, or six implants) used to support the fixed prostheses.[9,10,36–42] On the contrary, Fazi et al,[14] as well as Silva-Neto et al,[43] in recent FEAs, concluded that four or five parallel implants showed lower stress/strain concentration in bone in comparison with a configuration with three parallel implants. However, in these studies, the models were quite simplified and the implants were designed without body threads, abutment housing, and internal threads for prosthetic screws.[14,43] In addition, the loading was applied to a single point at the distal edge of the bar in some FEA studies.[14,43] Although in a comparative FEA study, having the same loading conditions would be enough to compare different implant aspects on the stress/strain values and distribution, accurate reproduction of the complex forces exerted during chewing function is desirable. Furthermore, bite forces ranging from 50 to 400 N in the molar regions and 25 to 170 N in the incisor areas have been reported.[33] These variations are influenced by patient sex, muscle mass, exercise, diet, bite location, parafunction, number of teeth and implants, type of implant-supported prosthesis, physical status, and age.[33]

In this way, some assumptions made during the process of developing numerical models, especially regarding the model macrogeometry, the assignment of material properties, and interface conditions, may limit the validity of FEA results in some studies. Even generic finite element models, which intend to focus only on the relative influence of some implant parameter rather than to the absolute in vivo results, may be evaluated in respect to their coherence with biologic available data.[44] Hence, it is possible to determine whether numerical models are consistent in their predictive capacity and whether the provided information could be extrapolated, or at least be useful, to the clinical context.

The maximum admissible stress value, above which titanium fractures tend to occur, is approximately 900 MPa. The present result demonstrated that no implant system screw reached values even near that limit. Nevertheless, the abutment screw, as well as the abutment retaining screws, are susceptible to screw loosening, due to smaller magnitudes of fatigue stress, generated by oral forces.[45,46] Various factors may contribute to screw complications, such as connection design, inadequate preload on the screws, overtightening of the screws leading to stripping and/or screw deformation, occlusal overload from parafunction (applied loading magnitudes and frequency), occlusal interferences, and excessively long cantilevers.[45,46] The present FEA confirmed that loading magnitude and distribution greatly influenced the EQV stress values for implant components in both connection types. Additionally, a unilateral loading distribution induced significantly higher stress in the implant system, compared with a balanced (ie, symmetrical) loading distribution.

A reduction in the EQV stress in external hex abutment screws was demonstrated in the current FEA, with the increase in implant number. Also, Silva-Neto et al,[43] in a FEA study of external hex inferior full-arch rehabilitations, suggested that by reducing the number of implants from five to three, the stress significantly increased in abutment screws. However, for the Morse-taper connection, the number of implants had no influence on the EQV stress in abutment screws.

In the present FEA, the Morse-taper connection type presented significantly lower EQV stress values in the abutment screw, in comparison with the external hex connection. This observation corroborates data presented by Merz et al[31] and Pessoa et al,[22] who demonstrated that when loads are applied over the abutment in an external hex configuration, there is no positive or geometric locking. In this way, under lateral or oblique loading, the abutment separates from the implant and tends to tilt about a small area on the implant shoulder; thus, the rising stress is absorbed mainly by the abutment screw. This factor could also be used to explain the lower EQV stress levels found for external hex bar retaining screws. The resilient component in the external hex connection (ie, abutment screw) was shown to absorb some of the load, which could have affected the bar retaining screws for this connection type. On the contrary, in a taper connection, the loading is resisted mainly by the taper interface. It prevents the abutment from tilting off, allowing stable retention of position by frictional forces.[22] The lateral wall of Morse-taper abutments helps to dissipate the lateral forces and protects the abutment screw from excessive stress. However, the rigid aspect of the Morse-taper connection resulted in a higher EQV stress in bar retaining screws for this connection type.

The present study demonstrated a slightly higher stress concentration in bar retaining screws for the three-implant configuration, compared with the four- and five-implant configurations, regardless of the loading design and connection type. Thus, it could be speculated whether a higher incidence of screw loosening would actually happen for the three-implant rehabilitations. The smaller number of bar retaining screws could probably explain a possible greater need for more follow-up appointments for three-implant–supported prostheses, as only one screw loosening is capable of inducing uncomfortable prosthetic instability. However, as the number of bar retaining screws is higher for four- and five-implant-supported prostheses, only one screw loosening would possibly not be perceived by the patients. A higher need for prosthetic maintenance should possibly be expected for three-implant rehabilitation. Unfortunately, similar FEA studies evaluating the quantity and arrangement of implants did not consider the prosthetic connection and components in detail.[14,43] Additionally, clinical research has mainly focused either on implant survival or on biologic and technical complications in partial edentulism.[47–51] The incidence of prosthetic complications of fixed implant-supported complete dentures has been addressed to only a minor extent.[27,28]

The biomechanical effects of different implant/prosthetic designs for a total-arch implant-supported inferior rehabilitation were investigated by FEA in the present study. Although it is an incontestably useful tool for obtaining information that is difficult to acquire from laboratory experiments or clinical studies, the results obtained by FEA should be interpreted with some caution. The assumptions made during the process of developing a finite element model limit the validity of the absolute values of the stress/strain and displacement calculated in a model in which an experimental validation was not accomplished. Otherwise, the association of the FEA with statistical analysis has been demonstrated as capable of accurately interpreting the relative influence that each of the input parameters have on the encountered results of implant FEAs.[22,30,34,52,53] In this regard, the mean strain values in bone and stress values in the implant system screws have been used to statistically evaluate model sensitivities to variations of input parameters and their relative influence on the finite element results. However, the average values should not be used for a direct comparison between the evaluated parameters. Comparison should only be made between the values of peak strain and stress in each of the models components.

Additionally, the modeling of bone-adaptive processes was not one of the aims of the current FEA. Although some authors have considered 4,200 $\mu\varepsilon$ as a possible threshold for pathologic bone overload, rather than only strain amplitude, loading frequency and number of loading cycles are parameters capable of greatly influencing the cortical bone adaptive response.[54] The loading applied in the presented simulation was static, and bone responds to dynamic rather than to static loads.[16] Furthermore, besides the implant-abutment connection design, preload applied to the abutment screw was also considered to be a factor influencing the abutment stability.[31] The tightening process causes interference in the abutment screw, which in turn causes the threads of the abutment screw and the implant to engage with a positive force. The implant-abutment joint efficiency, and therefore the strain state in the implant connection region, is considered a function of the design characteristics of the implant-abutment connection as well as, to some extent, of the preload stress achieved in the abutment screw when the suggested tightening torque is applied. Considering that the clamping force has a considerable effect on the maintenance of abutment complex stability, a decreased amount of strain and separation is likely to be observed when preload is incorporated in FEA.[55] In this way, preload in the abutment screw should be included for a realistic FEA evaluation of the implant-abutment connection.

CONCLUSIONS

Within the limitations of the present FEA, the following conclusions can be drawn. The Morse-taper connection type significantly decreases the strain levels in peri-implant bone and the EQV stress in the abutment screw.

No important effect from the number of implants could be noted in peri-implant bone strain, regardless of the connection type. A smaller number of implants slightly increases the stress in abutment and bar retaining screws. The Morse-taper connection type significantly decreases stress in the abutment screw, while increasing the stress in the bar retaining screw. A balanced adjustment of the loading (ie, symmetrical loading) improves the biomechanics of a mandibular full-arch rehabilitation.

ACKNOWLEDGMENTS

The authors reported no conflicts of interest related to this study.

REFERENCES

1. Adell R, Eriksson B, Lekholm U, Brånemark PI, Jemt T. Long-term follow-up study of osseointegrated implants in the treatment of totally edentulous jaws. Int J Oral Maxillofac Implants 1990;5:347–359.
2. Naert I, Quirynen M, van Steenberghe D, Darius P. A study of 589 consecutive implants supporting complete fixed prostheses. Part II: Prosthetic aspects. J Prosthet Dent 1992;68:949–956.
3. Ekelund JA, Lindquist LW, Carlsson GE, Jemt T. Implant treatment in the edentulous mandible: A prospective study on Brånemark system implants over more than 20 years. Int J Prosthodont 2003;16:602–608.
4. Attard NJ, Zarb GA. Long-term treatment outcomes in edentulous patients with implant-fixed prostheses: The Toronto study. Int J Prosthodont 2004;17:417–424.
5. Brånemark PI, Svensson B, van Steenberghe D. Ten years survival rates of fixed prostheses on four or six implants ad modum Brånemark in full edentulism. Clin Oral Implants Res 1995;6:227–231.
6. Wolfinger GJ, Balshi TJ, Rangert B. Immediate functional loading of Brånemark system implants in edentulous mandibles: Clinical report of the results of developmental and simplified protocols. Int J Oral Maxillofac Implants 2003;18:250–257.
7. Chiapasco M, Gatti C. Implant-retained mandibular overdentures with immediate loading: A 3- to 8-year prospective study on 328 implants. Clin Implant Dent Relat Res 2003;5:29–38.
8. Maló P, Rangert B, Nobre M. "All-on-Four" immediate-function concept with Brånemark System implants for completely edentulous mandibles: A retrospective clinical study. Clin Implant Dent Relat Res 2003;5:2–9.
9. Brånemark PI, Engstrand P, Ohrnell LO, et al. Brånemark Novum: A new treatment concept for rehabilitation of the edentulous mandible. Preliminary results from a prospective clinical follow-up study. Clin Implant Dent Relat Res 1999;1:2–16.
10. Engstrand P, Gröndahl K, Ohrnell LO, et al. Prospective follow-up study of 95 patients with edentulous mandibles treated according to the Brånemark Novum concept. Clin Implant Dent Relat Res 2003;5:3–10.
11. Henry PJ, van Steenberghe D, Blombäck U, et al. Prospective multicenter study on immediate rehabilitation of edentulous lower jaws according to the Brånemark Novum Protocol. Clin Implant Dent Relat Res 2003;5:137–142.
12. De Bruyn H, Kisch J, Collaert B, et al. Fixed mandibular restorations on three early-loaded regular platform Brånemark implants. Clin Implant Dent Relat Res 2001;3:176–184.
13. Sahin S, Cehreli MC, Yalçin E. The influence of functional forces on the biomechanics of implant-supported prostheses—a review. J Dent 2002;30:271–282.
14. Fazi G, Tellini S, Vangi D, Branchi R. Three-dimensional finite element analysis of different implant configurations for a mandibular fixed prosthesis. Int J Oral Maxillofac Implants 2011;26:752–759.
15. Ogawa T, Dhaliwal S, Naert I, et al. Impact of implant number, distribution and prosthesis material on loading on implants supporting fixed prostheses. J Oral Rehabil 2010;37:525–531.
16. Duyck J, Rønold HJ, Van Oosterwyck H, et al. The influence of static and dynamic loading on marginal bone reactions around osseointegrated implants: An animal experimental study. Clin Oral Implants Res 2001;12:207–218.
17. Hoshaw SJ, Brunski JB, Cochran GV. Mechanical loading of Brånemark implants affects interfacial bone modeling and remodeling. Int J Oral Maxillofac Implants 1994;9:345–360.
18. Misch CE, Suzuki JB, Misch-Dietsh FM, Bidez MW. A positive correlation between occlusal trauma and peri-implant bone loss: Literature support. Implant Dent 2005;14:108–116.
19. Isidor F. Loss of osseointegration caused by occlusal load of oral implants. A clinical and radiographic study in monkeys. Clin Oral Implants Res 1996;7:143–152.
20. Isidor F. Histological evaluation of peri-implant bone at implants subjected to occlusal overload or plaque accumulation. Clin Oral Implants Res 1997;8:1–9.
21. Hansson S. A conical implant-abutment interface at the level of the marginal bone improves the distribution of stresses in the supporting bone. Clin Oral Implants Res 2003;14:286–293.
22. Pessoa RS, Muraru L, Marcantonio E Jr, 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;12:219–234.
23. Shin YK, Han CH, Heo SJ, Kim S, Chun HJ. Radiographic evaluation of marginal bone level around implants with different neck designs after 1 year. Int J Oral Maxillofac Implants 2006;20:789–794.
24. Van de Velde T, Collaert B, Sennerby L, De Bruyn H. Effect of implant design on preservation of marginal bone in the mandible. Clin Implant Dent Relat Res 2010;12:134–141.
25. Kang IH, Kim CW, Lim YJ, Kim MJ. A comparative study on the initial stability of different implants placed above the bone level using resonance frequency analysis. J Adv Prosthodont 2011;3:190–195.
26. Bateli M, Att W, Strub JR. Implant neck configurations for preservation of marginal bone level: A systematic review. Int J Oral Maxillofac Implants 2011;26:290–303.
27. Bozini T, Petridis H, Garefis K, Garefis P. A meta-analysis of prosthodontic complication rates of implant-supported fixed dental prostheses in edentulous patients after an observation period of at least 5 years. Int J Oral Maxillofac Implants 2011;26:304–318.
28. Noda K, Arakawa H, Maekawa K, et al. Identification of risk factors for fracture of veneering materials and screw loosening of implant-supported fixed partial dentures in partially edentulous cases. J Oral Rehabil 2013;40:214–220.
29. Jaecques SV, Van Oosterwyck H, Muraru L, et al. Individualized, micro CT-based finite element modeling as a tool for biomechanical analysis related to tissue engineering of bone. Biomaterials 2004;25:1683–1696.
30. Pessoa RS, Jaecques SV. Finite element analysis in dental implant biomechanics. In: Cehreli M, Akca K (eds). Biomechanics of Dental Implants: Handbook of Researchers. 1st ed. New York: Nova Science Publishers, 2011:157–181.

31. 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;15:519–526.
32. Steinemann SG, Mäusli PA, Szmukler-Moncler S, et al. Betatitanium alloy for surgical implants. In: Froes FH, Caplan I (eds). Titanium '92. Science and technology. Warrendale, PA: The Minerals, Metals & Materials Society, 1993:2689–2696
33. Duyck J, Van Oosterwyck H, Vander Sloten J, et al. Magnitude and distribution of occlusal forces on oral implants supporting fixed prostheses: An in vivo study. Clin Oral Implants Res 2000;11:465–475.
34. Dar FH, Meakina JR, Aspden RM. Statistical methods in finite element analysis. Biomech 2002;35:1155–1161.
35. Frost HM. Perspectives: Bone's mechanical usage windows. Bone Miner 1992;19:257–271.
36. Östman PO. Immediate/early loading of dental implants. Clinical documentation and presentation of a treatment concept. Periodontol 2000 2008;47:90–112.
37. Esposito M, Grusovin MG, Achille H, Coulthard P, Worthington HV. Interventions for replacing missing teeth: Different times for loading dental implants. Cochrane Database Syst Rev 2009:CD003878.
38. Hatano N, Yamaguchi M, Suwa T, Watanabe K. A modified method of immediate loading using Brånemark implants in edentulous mandibles. Odontology 2003;91:37–42.
39. Hatano N, Yamaguchi M, Yaita T, Ishibashi T, Sennerby L. New approach for immediate prosthetic rehabilitation of the edentulous mandible with three implants: A retrospective study. Clin Oral Implants Res 2011;22:1265–1269.
40. Krekmanov L, Kahn M, Rangert B, Lindström H. Tilting of posterior mandibular and maxillary implants for improved prosthesis support. Int J Oral Maxillofac Implants 2000;15:405–414.
41. Rangert B, Jemt T, Jörneus L. Forces and moments on Brånemark implants. Int J Oral Maxillofac Implants 1989;4:241–247.
42. Rivaldo EG, Montagner A, Nary H, da Fontoura Frasca LC, Brånemark PI. Assessment of rehabilitation in edentulous patients treated with an immediately loaded complete fixed mandibular prosthesis supported by three implants. Int J Oral Maxillofac Implants 2012;27:695–702.
43. Silva-Neto JP, Pimentel MJ, Neves FD, Consani RL, Dos Santos MB. Stress analysis of different configurations of 3 implants to support a fixed prosthesis in an edentulous jaw. Braz Oral Res 2014;28:67–73.
44. Mellal A, Wiskott HW, Botsis J, Scherrer SS, Belser UC. Stimulating effect of implant loading on surrounding bone. Comparison of three numerical models and validation by in vivo data. Clin Oral Implants Res 2004;15:239–248.
45. Weinberg LA. The biomechanics of force distribution in implant-supported prostheses. Int J Oral Maxillofac Implants 1993;8:19–31.
46. McGlumphy EA. Keeping implant screws tight: The solution. J Dent Symp 1993;1:20–23.
47. Berglundh T, Persson L, Klinge B. A systematic review of the incidence of biological and technical complications in implant dentistry reported in prospective longitudinal studies of at least 5 years. J Clin Periodontol 2002;29:S197–S212.
48. Brägger U, Aeschlimann S, Bürgin W, Hämmerle CH, Lang NP. Biological and technical complications and failures with fixed partial dentures (FPD) on implants and teeth after four to five years of function. Clin Oral Implants Res 2001;12:26–34.
49. Goodacre CJ, Bernal G, Rungcharassaeng K, Kan JY. Clinical complications with implants and implant prostheses. J Prosthet Dent 2003;90:121–132.
50. Lang NP, Wilson TG, Corbet EF. Biological complications with dental implants: Their prevention, diagnosis and treatment. Clin Oral Implants Res 2000;11:S146–S155.
51. Pjetursson BE, Tan K, Lang NP, et al. A systematic review of the survival and complication rates of fixed partial dentures (FPDs) after an observation period of at least 5 years. Clin Oral Implants Res 2004;15:667–676.
52. Pessoa RS, Coelho PG, Muraru L, et al. Influence of implant design on the biomechanical environment of immediately placed implants: Computed tomography-based nonlinear three-dimensional finite element analysis. Int J Oral Maxillofac Implants 2011;26:1279–1287.
53. Pessoa RS, Vaz LG, Marcantonio E Jr, et al. Biomechanical evaluation of platform switching in different implant protocols: Computed tomography-based three-dimensional finite element analysis. Int J Oral Maxillofac Implants 2010;25:911–919.
54. De Smet E, Jaecques SV, Jansen JJ, et al. Effect of constant strain rate, composed by varying amplitude and frequency, of early loading on periimplant bone (re)modelling. J Clin Periodontol 2007;34:618–624.
55. Iplikçioğlu H, Akça K, Çehreli CM, Sahin S. Comparison of non-linear finite element stress analysis with in vitro strain gauge measurements on a Morse taper implant. Int J Oral Maxillofac Implants 2003;18:258–265.