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RESEARCH AND EDUCATION

Removal resistance of locking taper implant-abutment connections at different strike numbers: An in vitro study

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ABSTRACT

Statement of problem. The consistency of the placement force provided by striking friction-locked abutments remains uncertain.

Purpose. The purpose of this in vitro study was to evaluate different strikes and repeated reconnection-disconnection cycles on the connection strength of a locking taper implant system with varying wells.

Material and methods. Sixty-three implant-abutment replica specimens (IARSs) with 2-, 2.5-, and 3-mm wells with a 1.5-degree locking taper design were fabricated from grade 5 titanium (Ti). Specimens were divided into 3 striking groups (3T, 6T, 12T, n=21/group) and 3 well subgroups (3T2, 3T2.5, 3T3; 6T2, 6T2.5, 6T3, and 12T2, 12T2.5, 12T3, n=7/subgroup). For subgroups 3T2, 3T2.5, and 3T3, the abutments were seated in the implants with finger load and then struck 3 times with a calibrated striker. The abutments were then disconnected using rotational movements and the removal torque values were recorded with a digital torque meter in newton centimeter (Ncm). This seating, striking, and removal cycle was repeated 7 times by the same operator. The same procedure was applied to the second and third subgroups with 6 and 12 strikes respectively. ANOVA and post hoc Bonferroni tests were used to analyze the data. Repeated measures ANOVA with the Greenhouse-Geisser correction was applied when the sphericity assumption was violated (α=.05).

Results. Removal torque values varied significantly with well diameter and the number of strikes ($P<.001$), peaking at 3 mm with 12 strikes. Torque values were statistically similar after repeated disconnection-reconnection cycles ($P>.05$).

Conclusions. Although wider wells and higher striking cycles tended to increase the removal torque and repeated disconnection–reconnection tended to decrease it, the values remained statistically similar. (J Prosthet Dent xxxx;xxx:xxx-xxx)

The design and retention characteristics of the implant-abutment connection (IAC) determine its strength, stability, and positional and rotational integrity.[1,2] IACs have been divided into 2 main groups (external and internal) based on their relationship with the coronal surface of the implant body. The most significant disadvantage of external connections is that the screw represents the weakest link, leading to frequent loosening and related complications.[3–6] Internal connection systems were developed to overcome the limitations of external connections, with the interface positioned within the implant body.[1,7] Internal connections have been primarily divided into slip-fit and friction fit. Friction-fit connections are formed by cone screw connection (8-, 11-, and 5.7-degrees Morse taper) and locking taper connection (LTC) or true Morse taper connections.[7] The 1.5-degree LTC, a type of connection that carries the characteristics of a true Morse taper, does not include screws in its system and is formed purely by friction.[1]

LTC system was first introduced in 1985 by Bicon Inc as an alternative to screw-retained IAC systems to overcome the disadvantages of the screws, which con-

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Clinical Implications

To ensure a more stable and reliable connection, the largest possible connection diameter should be used. Abutments should be seated with at least 6 strikes to facilitate clinical application, and the number of reconnection–disconnection cycles should be reduced during both clinical and laboratory procedures.

stituted the weak point of such systems.[8] I-System (I-System; Novodent SA) and NTA Shorter (NTA; Pilatus Swiss Dental Group) are implants with a 1.5-degree taper and other locked conical connections.

The abutment is placed by striking it into the corresponding hole in the implant body. The connection is formed by the friction of 2 matching surfaces which appear bright and smooth but are not at the micrometer level. A high compressive force is generated between the support and the inner wall of the implant due to the relative sliding between 2 friction surfaces at high contact pressure. This results in the breakdown of surface oxide layers and the fusion of roughness (cold welding). Thus, the gaps between the 2 surfaces are not visible, and, by behaving as a whole, it can withstand forces without loosening. Because of the high friction force, a higher removal torque than the embedding force is generated, and they resist tensile forces.[9–12] Clinical and meta-analysis studies have revealed minimal biological and mechanical complications with high survival and peri-implant tissue stability.[13,14]

Controlling the placement force exerted through striking can be challenging, leading to variations among clinicians. In LTC systems, 3 to 6 strikes have been reported to be sufficient to ensure a secure activation of the connection.[15,16] Physiological mastication forces may also generate additional pressure that stabilizes frictionally locked connections.[17,18] During prosthetic treatment processes, it may be necessary to remove the abutment after treatment for clinical and esthetic concerns. When the abutments are removed, a decrease in removal torque can be observed because of deformation between the materials on the connection surface.[19] Bozkaya and Müftü[20,21] determined that in frictionally locked tapered connections, an increase in placement forces—within the limits of elastic deformation—led to higher removal forces. Additionally, they reported that these forces increased linearly with the cube of contact area and insertion depth. Additionally, in LTC systems, where the application of removal force or torque when the abutment needs to be separated from the implant is required, the removal torque value (RTV) at which the abutments loosen is an indicator of the stability of the connection.[20–23] Certain implant systems require indexing, which restricts the abutment's rotational degree of freedom. It is not possible to remove the abutment from such systems or from complete arch restorations using torque alone. Both techniques (force and torque) can be used if there is no indexing limitation on the IAC.[24]

Given the limited number of studies investigating this topic,[25–30] this study aimed to compare the connection strength of friction-locked, screwless conical implant-abutment systems with different wells under in vitro conditions by evaluating the RTVs after different numbers of strikes and repeated reconnection-disconnection cycles (RDCs).

The null hypotheses were that the use of the same number of strikes would have no effect on RTVs in different wells, that the use of a different number of strikes would have no effect on RTVs in same wells, that, in different wells, the same number of strikes would not affect the RTVs after repeated RDCs, and that, in the same wells, a different number of strikes would not affect the RTVs after repeated RDCs.

MATERIAL AND METHODS

Sample size was determined using a software program (G*Power 3.1.9.6; Heinrich-Heine-Universität Düsseldorf) based on the study by Da Silva et al.[25] For a medium effect size (f=0.25), a significance level of .05% and 80% power, a minimum of 63 specimens was required to detect a significant difference.

Since RTVs cannot be measured using mass-produced implants and abutments, a test specimen that mimics the original IAC and allows for testing was designed. The elasticity and rigidity of the embedding material play a critical role in torque measurements, as deformation of low-modulus materials may absorb part of the applied torque and compromise accuracy. Kernen et al[31] reported that embedding implants in a stiffer material such as brass influenced the mechanical behavior of implants. To minimize mechanical interference, a 1-piece titanium replica of the original connection, without embedding, was used to ensure that the measured values reflected the actual torque at the interface. Since the rounded geometry of implant components may have caused rotation during torque measurement, such tests would be more difficult with original parts, leading to erroneous results. The design of the implant-abutment replica specimens (IARSs) was based on the connection of I-System implants (I-system; Novodent SA) with 3 different wells and abutment post diameters of 2, 2.5, and 3 mm, featuring a 2-part, 1.5-degree conical locking taper mechanism. Bozkaya and Müftü[20] stated that a small conicity angle and larger contact area provided a safe range of placement forces, resulting in

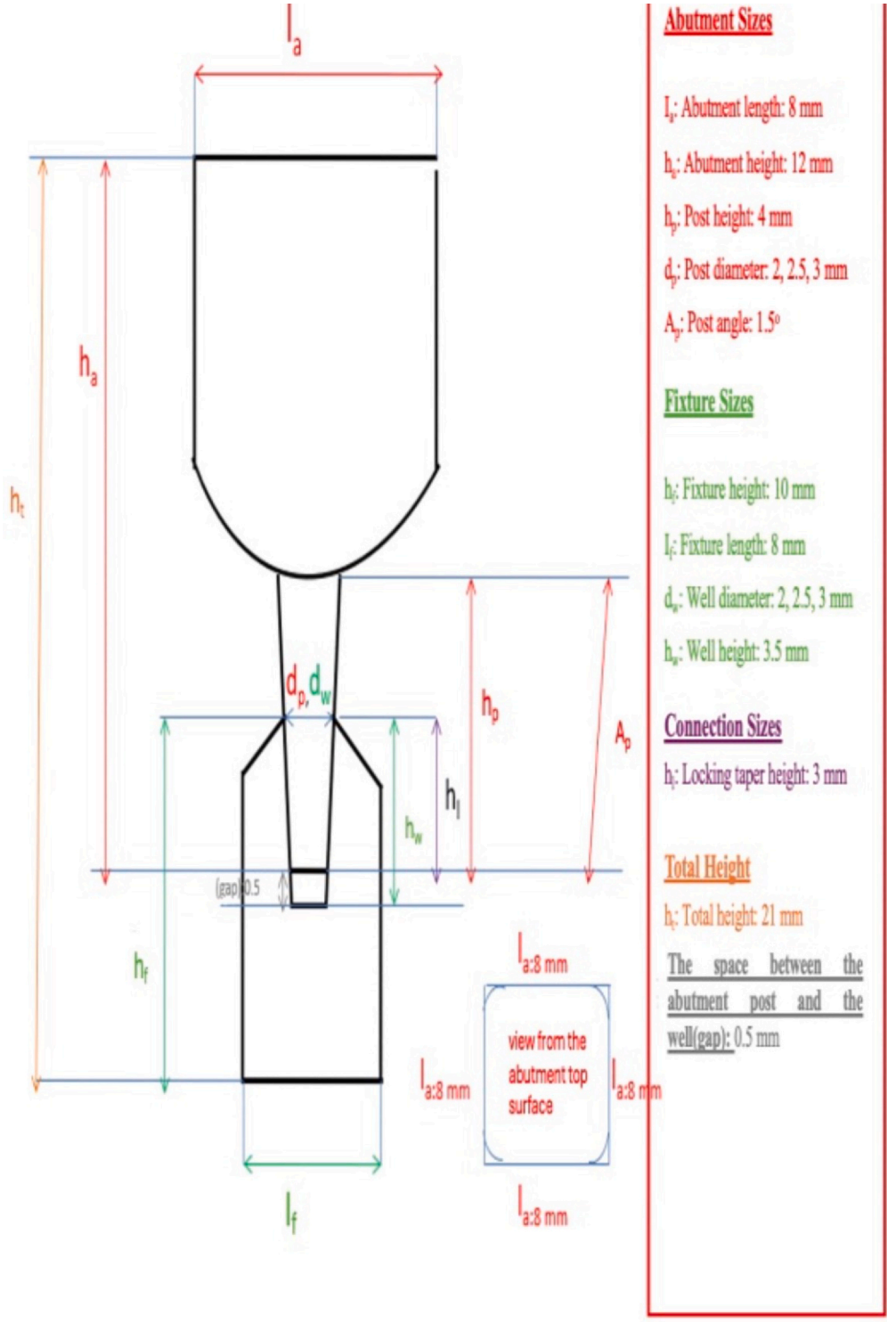

Figure 1. Linear design of specimens.

higher efficiency. The IARSs with a 1.5-degree taper angle used in the present study complied with the International Organization for Standardization (ISO) 296 standard.[32] Determined draft was then modeled in 3 dimensions using a software program (SolidWorks; SolidWorks Corp) software (Figs. 1, 2).

Friction at the implant-abutment interface has been reported to be influenced by factors such as material hardness, surface treatments, and saliva, making it difficult to determine the friction coefficient.[33,34] Depending on tribological conditions, friction coefficients between 0.2 and 0.5 have been reported for titanium and titanium alloy

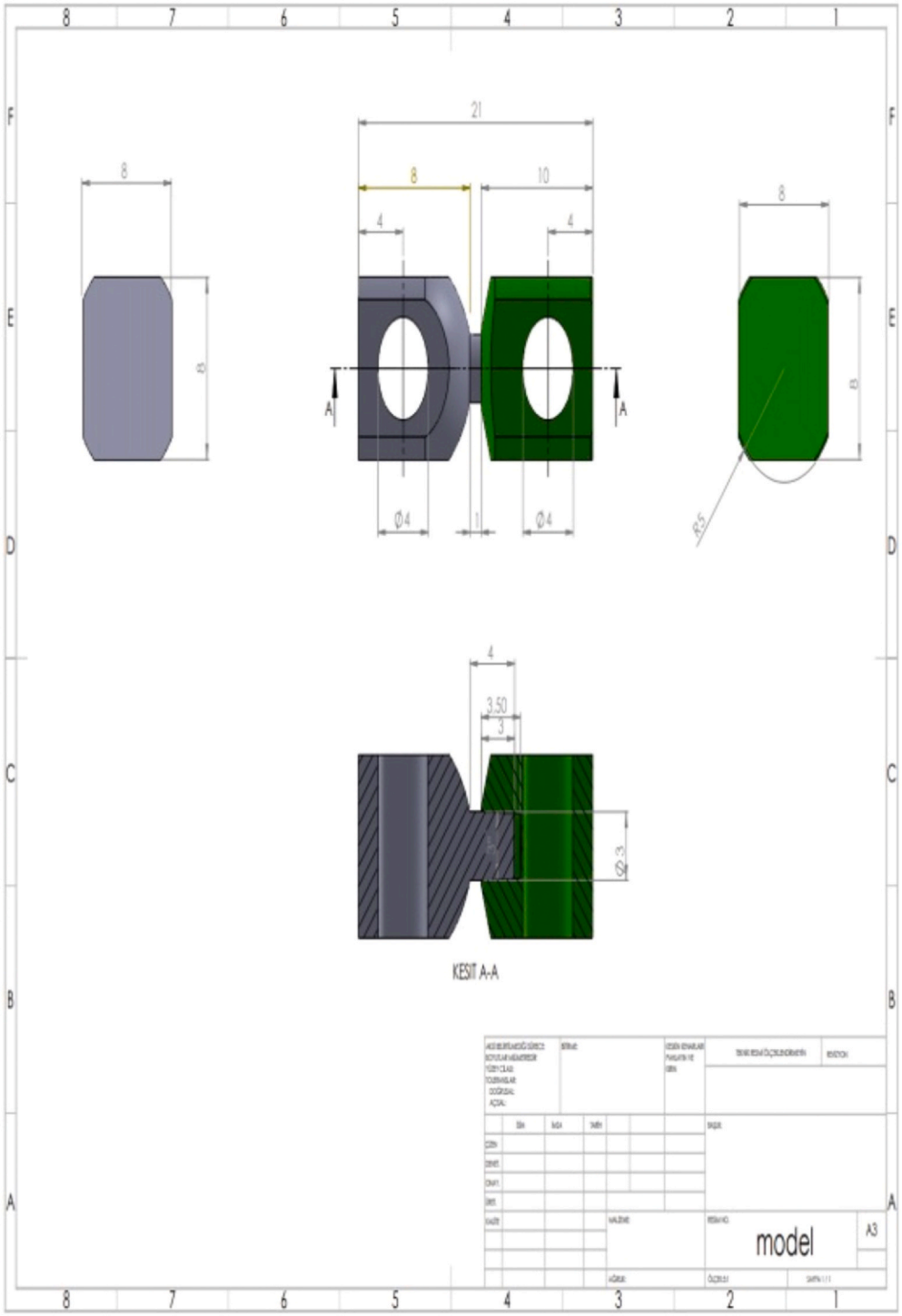

Figure 2. Three-dimensional design of specimens.

interfaces.[20,33,35] In previous studies, it has been shown that titanium alloys (Ti-6Al-4V) had higher strength[36,37] and, therefore, were more stable under functional loads.[38] A total of 63 titanium grade 5 (Ti6Al4V) IARSs were fabricated on a computer numerical control (CNC) machine (Citizen Machinery Co, Ltd). Specimens were ultrasonically washed (CC HYDROSONICS Ltd) using a cleaning fluid (Vertrel SION; DuPont) for 180 seconds and dried with compressed air for 60 seconds after a 60-second steam bath (F810; CC HYDROSONIC Ltd).

Sixty-three IARSs were fabricated and divided into 3 striking groups (3T, 6T, and 12T, n=21/group) and 9 subgroups (3T2, 3T2.5, 3T3; 6T2, 6T2.5, 6T3, and 12T2, 12T2.5, 12T3, n=7/subgroup) (Fig. 3). To ensure the

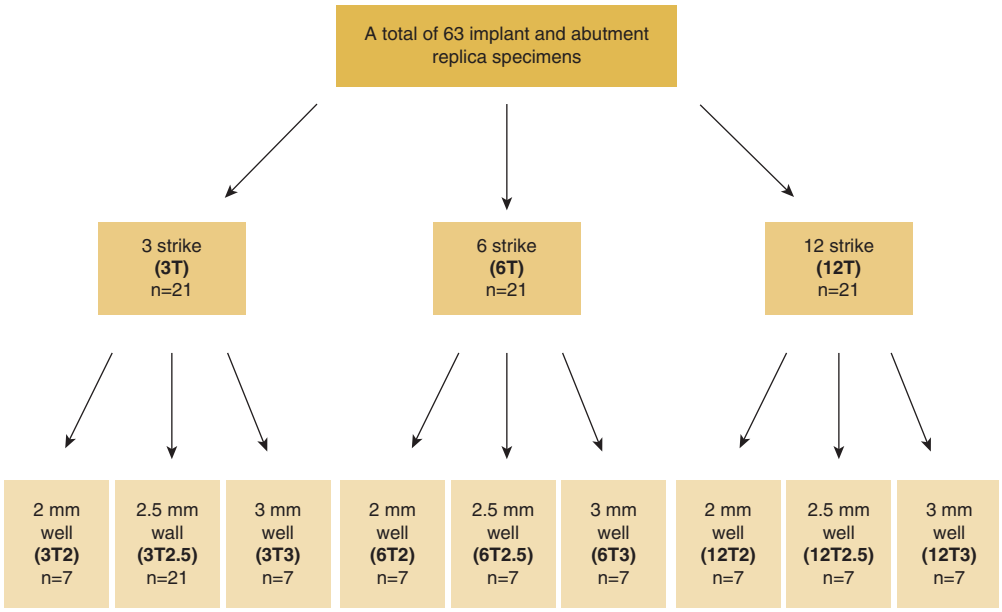

Figure 3. Grouping of implant and abutment specimens.

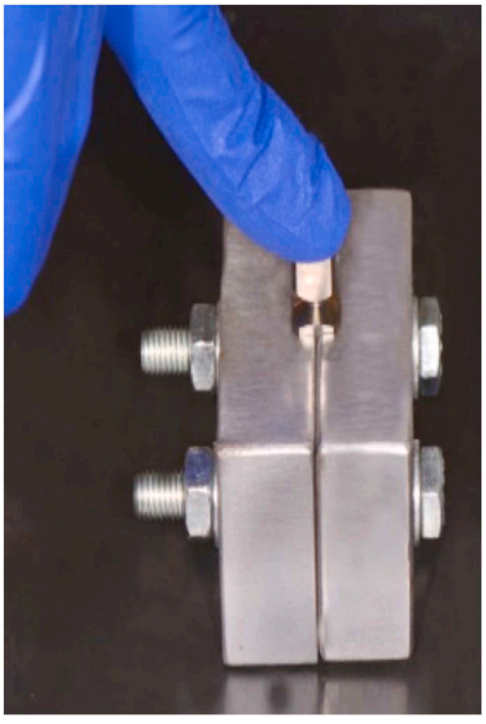

Figure 4. Finger load application.

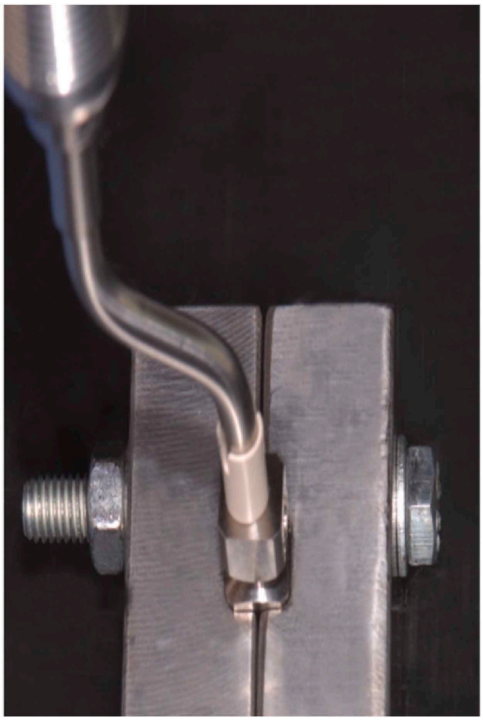

Figure 5. Striking process.

same direction of repeated forces along the long axis of the specimen without movement, each specimen was securely fixed into a 2-piece mold. The force was applied using finger load and a hammer and was quantified using a Basic Force Gauge (Mecmesin). Lee et al[39] conducted measurements for each type of force applied by the same individual to determine average values, reporting a mean ±standard deviation (SD) finger load of 58.0 ±5.7 N and a mean hammer strike force of 32.9 ±2.8 N. Specimens for subgroups 3T2, 3T2.5, and 3T3, abutments were seated in the implants with a finger load (Fig. 4) and then struck 3 times perpendicular to the abutment surface using a calibrated striker (Conometric fixation tool; Dentsply Sirona) according to the manufacturer's instructions (Fig. 5). The striker applied a force of 29 N in a single tap.

The ultrasonic method, removal torque, and rotational angle measurement have been commonly used

techniques for the indirect assessment of placement force.[40–42] A digital torque meter (Cap Torque Testers, Series TT03; Mark10) linked to a computer using a data acquisition software program (MESURLite; Mark-10) was used to measure the removal torque values. The body of the digital torque meter was fixed to a metal plate to standardize the torque value measurements. Specimens were firmly secured within the torque meter's 3-edged locking chuck to prevent any movement during rotational removal test. A metal bar was placed that passively passed through the hole within the abutment replica. For LTC, axial-torsional force to overcome frictional resistance and detach the abutment from the implant body has been recommended.[1,16] Abutment replicas were disconnected from the implant replicas using a rotational movement. RTVs were analyzed, and the highest RTV for each specimen was recorded in Ncm. This seating, striking, and removal cycle was repeated 7 times by the same operator (BG) experienced in implant dentistry for at least 10 years and cyclic RTVs were recorded.

For subgroups 6T2, 6T2.5, and 6T3, abutment replicas were seated in the implants using finger force and struck 6 times per trial, whereas for subgroups 12T2, 12T2.5, and 12T3, they were struck 12 times per trial. This procedure was repeated 7 times for each specimen.

Statistical analyses were performed using a statistical software package (IBM SPSS Statistics for Windows, v25.0; IBM Corp) with 95% confidence interval in all groups. RTVs, mean and SDs were calculated for each group. Data were analyzed using ANOVA, normal distribution was verified with the Kolmogorov-Smirnov and Shapiro-Wilk tests, and intragroup and intergroup comparisons were made using the post hoc Bonferroni test.

To compare the intercycle RTVs, the repeated measures ANOVA was used, and the sphericity assumption was examined. Sphericity assumption requires that the variances of differences between measurements at different time points are equal in analyses involving repeated measures. Sphericity was checked using the Mauchly Test of Sphericity. If sphericity were severely violated, the Greenhouse-Geisser correction was used to prevent the test from producing misleading results. The Greenhouse-Geisser correction is particularly appropriate for small to medium sample sizes, specifically when n<100.

RESULTS

The mean values and SD of the RTVs for different well sizes (2, 2.5, and 3 mm) with the same number of strikes, measured across 7 specimens and 7 cycles per subgroup in Ncm, are presented in Table 1. According to the

Table 1. Mean ±standard deviation values of removal torque values (Ncm) obtained during 7 cycles for different wells (2, 2.5, and 3 mm) and number of strikes (3, 6, 12)

Well	Strike	Cycle 1 (n=7)	Cycle 2 (n=7)	Cycle 3 (n=7)	Cycle 4 (n=7)	Cycle 5 (n=7)	Cycle 6 (n=7)	Cycle 7 (n=7)	Mean ±SD (n=49)
2 mm	3	34.43 ±2.46	28.77 ±2.00	28.26 ±1.85	27.86 ±1.69	27.29 ±1.90	27.27 ±1.59	26.91 ±2.50	28.68 ±1.44
2.5 mm	3	35.56 ±0.95	32.84 ±1.99	31.24 ±1.89	32.41 ±2.65	31.64 ±2.00	31.50 ±1.27	30.81 ±1.35	32.29 ±1.06
3 mm	3	44.56 ±2.39	38.27 ±3.43	35.26 ±2.36	36.79 ±3.42	38.19 ±2.44	37.04 ±3.32	34.46 ±2.14	37.80 ±1.15
2 mm	6	35.17 ±2.10	29.30 ±2.72	30.26 ±2.28	28.40 ±2.13	29.23 ±2.20	29.30 ±1.98	28.66 ±1.68	30.04 ±1.73
2.5 mm	6	44.01 ±1.72	40.66 ±2.82	37.49 ±2.12	36.49 ±2.78	38.70 ±2.55	36.49 ±0.99	37.87 ±1.92	38.81 ±1.14
3 mm	6	45.83 ±3.22	43.07 ±1.64	40.46 ±3.66	41.81 ±2.73	40.50 ±3.63	40.76 ±2.37	41.76 ±2.05	42.03 ±2.19
2 mm	12	38.86 ±2.63	34.67 ±2.73	32.10 ±2.23	34.07 ±1.55	32.57 ±2.32	31.74 ±1.99	32.69 ±1.64	33.81 ±1.63
2.5 mm	12	52.99 ±1.92	49.19 ±5.06	42.07 ±2.08	44.36 ±2.70	45.66 ±3.83	43.54 ±4.58	39.01 ±1.97	45.26 ±2.13
3 mm	12	54.86 ±4.37	51.23 ±4.38	48.87 ±1.93	48.97 ±4.42	49.49 ±3.12	46.71 ±3.59	48.83 ±4.37	49.85 ±1.56

SD, standard deviation

Table 2. Pairwise comparisons with Bonferroni test (post hoc)

Strike	(I) Well (mm)	(J) Well	Mean Difference (I-J)	P
3	2	2.5	-3.60408*	<.001
		3	-9.11429*	<.001
	2.5	3	-5.51020*	<.001
6	2	2.5	-8.76939*	<.001
		3	-11.98163*	<.001
	2.5	3	-3.21224*	.008
12	2	2.5	-11.44490*	<.001
		3	-16.03673*	<.001
	2.5	3	-4.56184*	<.001

* Statistically significant ($P<.05$).

ANOVA results, RTVs showed statistically significant differences depending on the well size for the same number of strikes (3 strikes: F(2.18)=97.797, $P<.001$, 6 strikes: F(2.18)=88.959, $P<.001$, and 12 strikes: F(2.18) =148.892, $P<.001$). Pairwise comparisons using the Bonferroni test revealed that the highest RTVs for all numbers of strikes were observed in abutments with a 3-mm well, while a decrease in well diameter resulted in a reduction in RTVs ($P<.001$) (Table 2).

Repeated measures ANOVA was used to compare the intercycle RTVs of specimens with 3, 6, and 12 strikes for different wells, and the sphericity assumption of this method was assessed. The Mauchly test of sphericity indicated that for specimens with 3 strikes, sphericity was achieved in the cyclic RTVs for 2-mm ($P=.068$) and 2.5-mm ($P=.621$) but not for 3-mm wells ($P<.001$). For specimens with 6 strikes, sphericity was achieved in the cyclic RTVs for the 2-mm well ($P=.760$), 2.5-mm well ($P=.726$), and 3-mm well ($P=.691$). For specimens with 12 strikes, sphericity was not achieved for the 2-mm well ($P=.004$) but was achieved for the 2.5-mm well ($P=.678$) and 3-mm well ($P=.081$).

For cyclic RTVs where sphericity was not achieved, the Greenhouse-Geisser correction values were considered, since the sample size was small. The univariate test (post hoc) was used to analyze the difference between cycles with the same number of strikes and different wells. A statistically significant difference was found in intercycle RTVs for 2-mm (3 strikes: F(6) =20.131, 6 strikes: F(6)=18.716, 12 strikes: F(2.95) =16.699), 2.5-mm (3 strikes: F(3.435)=7.113, 6 strikes: F (6)=12.215, 12 strikes: F(6)= 18.468), and 3-mm (3 strikes: F(3.233)=9.694, 6 strikes: F(6)=6.610, 12 strikes: F (6)= 3.217) wells for all numbers of strikes ($P<.001$). The Bonferroni test, applied to identify which cycles contributed to differences in intercycle RTVs, revealed that RTVs decreased after the first cycle but that the values remained statistically similar between cycles ($P>.05$) (Table 3).

The mean values and SD of the RTVs for different numbers of strikes (3, 6, and 12 strikes) with the same wells, measured across 7 specimens and 7 cycles per subgroup in Ncm, are presented in Table 1. According to the ANOVA results, RTVs showed significant differences depending on the number of strikes for the same well: 2-mm (F[2.18]=19.209; $P<.001$), 2.5-mm (F[2.18] =126.830; $P<.001$), and 3-mm (F[2.18]=91.993; $P<.001$). In pairwise comparisons made with the Bonferroni test, 12 strikes provided a higher removal torque than 3 and 6 strikes in all well groups, and 6 strikes provided a higher removal torque than 3 strikes except for the 2-mm well group (Table 4).

Repeated measures ANOVA was used to compare the intercycle RTVs of the specimens with 2-, 2.5-, and 3-mm wells at different numbers of strikes, and the sphericity assumption of this method was assessed. The Mauchly test of sphericity indicated that for specimens with a 2-mm well, sphericity was achieved in the cyclic RTVs for 3 strikes ($P=.068$) and 6 strikes ($P=.760$), but not for 12 strikes ($P=.004$). For specimens with a 2.5-mm well, sphericity was achieved in the cyclic RTVs for 3 strikes ($P=.621$), 6 strikes ($P=.726$), and 12 strikes ($P=.678$). For specimens with a 3-mm well, sphericity was not achieved in the cyclic RTVs for 3 strikes ($P<.001$) but was achieved for 6 strikes ($P=.691$) and 12 strikes ($P=.081$).

For cyclic RTVs where sphericity was not achieved, the Greenhouse-Geisser correction values were considered, since the sample size was small. The univariate test (post hoc) was used to analyze the differences between cycles with the same wells and different numbers of strikes. A significant difference in intercycle RTVs was found at 3 (2-mm: (6.38)=20.131; 2.5-mm F(6)=7.113; 3-mm: F(3.233)=9.694, 6 (2-mm: F(6.38)=18.716; 2.5-mm (F6)=12.215; 3-mm: F(6.233)=6.610, and 12 (2-mm: F (2.952)=16.699; 2.5-mm; F(6)=18.468; 3-mm: (F6.233) =3.217) strikes for all wells ($P<.001$). The Bonferroni test revealed that although RTVs decreased after the first cycle, the values remained statistically similar between cycles ($P>.05$) (Table 5).

DISCUSSION

The null hypotheses that the use of the same number of strikes would have no effect on RTVs in different wells and that the use of different numbers of strikes would have no effect on RTVs in the same wells were rejected, as increasing both the well diameter and the number of strikes resulted in higher RTVs. In the tests performed over 7 RDCs, the RTVs decreased after the first cycle; however, subsequent cycles showed statistically similar values (Tables 3, 5). Therefore, the hypotheses stating that repeated RDCs would not influence RTVs were partially rejected.

The findings of the present study demonstrated that increasing the placement forces applied over the abutments resulted in higher RTVs. This outcome indicated

Table 3. Post hoc (Bonferroni) analysis results of removal torque values between cycles of abutments with different wells for same number of strikes

Strike	Well	(I) Cycle	(J) Cycle	Mean Difference (I-J)	*P*
3 strike	2 mm	1	2	5.657*	.015
			3	6.171*	.009
			4	6.571*	.029
			5	7.143*	.006
			6	7.157*	.009
			7	7.514*	.019
	2.5 mm	1	3	4.314*	.043
			6	4.057*	.008
			7	4.743*	.017
	3 mm	1	3	9.300*	.003
			7	10.071*	.002
6 strike	2 mm	1	2	5.871*	.001
			3	4.914*	.038
			4	6.771*	<.001
			5	5.943*	.006
			6	5.871*	.001
			7	6.514*	<.001
	2.5 mm	1	3	6.529*	.007
			4	7.529*	.007
			5	5.314*	.024
			6	7.529*	.002
		2	3	3.171*	.040
	3 mm	1	4	4.014*	.046
12 strike	2 mm	1	2	4.186*	.011
			3	6.757*	.031
			5	6.286*	.027
			6	7.114*	.006
			7	6.171*	.008
	2.5 mm	1	3	10.914*	<.001
			4	8.629*	.003
			6	9.443*	.017
			7	13.971*	.001
		2	3	7.114*	.026
			7	10.171*	.050
		5	7	6.643*	.034
	3 mm	1	3	5.986*	.031
			6	8.143*	.016
			7	6.029*	.026

* Statistically significant ($P<.05$).

Table 4. Pairwise comparisons with Bonferroni test (post hoc)

Well (mm)	(I) Strike	(J) Strike	Mean Difference (I-J)	*P*
2	3	6	-1.36122	.390
		12	-5.13061*	<.001
	6	12	-3.76939*	.001
2.5	3	6	-6.52653*	<.001
		12	-12.97143*	<.001
	6	12	-6.44490*	<.001
3	3	6	-4.22857*	.001
		12	-12.05306*	<.001
	6	12	-7.82449*	<.001

* Statistically significant ($P<.05$).

that the magnitude of the applied load played a crucial role in determining the mechanical stability of LTCs. When higher loads were applied, the abutment was likely seated deeper into the implant well, improving the adaptation of the contacting surfaces and minimizing interfacial gaps. This enhanced fit promoted a stronger frictional engagement and mechanical interlocking—often referred to as the "cold welding" phenomenon—which contributed to greater resistance during removal.[9] These observations were consistent with those of previous reports showing that increased loading improved the implant–abutment interface and enhanced removal forces.[10–12,24] In LTC systems, abutments tend to embed vertically under applied forces until a stable limit is reached.[8,39,43] The present findings suggested that controlled increases in seating force, such as repeated striking, promote gradual embedment without damage and connection stability optimization.

Specimens with a 3-mm well showed higher RTVs, indicating that a larger contact area improved the mechanical stability of LTCs. This supports the concept of gapless fitting through surface friction and cold welding under high contact pressure to resist loosening.[9] Previous studies[16,22,44] have shown that increasing the abutment diameter and the tapered joint surface enhanced frictional engagement and cold weld formation, thereby improving connection stability and reducing the risk of mechanical failure.

Santos et al[26] demonstrated that both taper angle and activation force influence removal resistance, with loading parallel to the implant axis (0 degree) producing higher retention. Similarly, Ren et al[27] and Zielak et al[28] reported that increasing the applied force, number of strikes, and abutment dimensions enhanced frictional engagement and removal torque. By ensuring parallel application of the seating forces to the implant axis, the present study demonstrated that higher striking numbers and larger well diameters led to a significant

Table 5. Post hoc (Bonferroni) analysis results of removal torque values between cycles according to different number of strikes of abutments with same well

Well	Strike	(I) Cycle	(J) Cycle	Mean Difference (I-J)	*P*
2 mm	3	1	2	5.657*	.015
			3	6.171*	.009
			4	6.571*	.029
			5	7.143*	.006
			6	7.157*	.009
			7	7.514*	.019
	6	1	2	5.871*	.001
			3	4.914*	.038
			4	6.771*	<.001
			5	5.943*	.006
			6	5.871*	.001
			7	6.514*	<.001
	12	1	2	4.186*	.011
			3	6.757*	.031
			5	6.286*	.027
			6	7.114*	.006
			7	6.171*	.008
2.5 mm	3	1	3	4.314*	.043
			6	4.057*	.008
			7	4.743*	.017
	6	1	3	6.529*	.007
			4	7.529*	.007
			5	5.314*	.024
			6	7.529*	.002
		2	3	3.171*	.040
	12	1	3	10.914*	<.001
			4	8.629*	.003
			6	9.443*	.017
			7	13.971*	.001
		2	3	7.114*	.026
			7	10.171*	.050
3	3	1	3	9.300*	.003
			7	10.071*	.002
	6	1	4	4.014*	.046
	12	1	3	5.986*	.031
			6	8.143*	.016
			7	6.029*	.026

* Statistically significant ($P<.05$).

increase in RTVs. However, differences in implant design, taper angle, applied force magnitude, and measurement techniques among studies may limit direct comparison of RTVs.

Bozkaya and Müftü[20,21] indicated that elastic deformation may enhance the removal force whereas plastic deformation could compromise this retention by reducing friction at the interface, a possible explanation for the progressive reduction in RTVs observed across successive abutment RDCs in this study. Previous investigations have evaluated the influence of mechanical loading and repeated RDCs, highlighting their effects on the integrity of the connection and associated RTVs. Alevizakos et al[19] reported that after cyclic loading, slight wear and polishing were noted on the abutment surface, which may have led to a slight reduction in removal forces. However, these changes were considered clinically negligible. Similarly, Ricciardi Coppedê et al[29] reported that while mechanical loading initially increased the removal torque, this value decreased with repeated insertion and removal cycles. Unlike the findings of the present study, Rabelo et al[30] demonstrated that the removal and reinsertion of abutments in LTC systems improved retention, although these changes may vary depending on the implant system used and the applied loading forces. Da Silva et al[25] reported insignificant changes in removal forces across 9 insertion and removal cycles, similar to the present study, although significant deformation or fractures of the implant abutments were reported. The findings of these studies suggest that the effects of mechanical loading and RDCs on IACs may vary depending on the implant system, applied force, and measurement methods used.

For a more stable and reliable connection, the largest feasible connection diameter implants and abutments should be used, the abutments should be seated with at least 6 strikes (considering the ease of clinical practice), and the number of RDCs should be limited during both clinical and laboratory procedures.

Limitations of the study included the in vitro design that cannot fully predict clinical outcomes. Future research simulating intraoral conditions or clinical studies might be necessary.

CONCLUSIONS

Based on the findings of this in vitro study, the following conclusions were drawn:

1. In locking taper IAC systems, variations in well diameter and the number of strikes influenced the strength of the connection.
2. The insertion and removal of abutments influenced the RTV.
3. To ensure a more stable and reliable connection, the largest possible connection diameter implants and abutments are preferred and seating the abutments with at least 6 strikes may be recommended.

DECLARATION OF GENERATIVE AI AND AI-ASSISTED TECHNOLOGIES IN THE WRITING PROCESS

During the preparation of this work the authors used ChatGPT 4.0 to improve language and readability. After using this tool or service, the authors reviewed and edited the content as needed and take full responsibility for the content of the publication.

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Leyla Kılıç: Conceptualization, Investigation, Writing - Original Draft, Writing - review and editing. **Tomurcuk Övül Kümbüloğlu:** Resources, Supervision, Writing - review and editing. **Mine Dündar Çömlekoğlu:** Supervision, Writing - review and editing. **Bülent Gökçe:** Conceptualization, Methodology, Resources, Writing - review and editing, Supervision.

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