Biomaterials Journal
http://www.biomatj.com
Online ISSN: 2812-5045

Type of the Paper (Mini-Review)

Bioengineered Models in Oral Health: From Development to Clinical Applications

Citation: Tamer M. Hamdy. Bioengineered Models in Oral Health: From Development to Clinical Applications. *Biomat. J., 4 (5),1 – 2 (2025).*

https://doi.org/10.5281/zenodo.18055475

Received: 20 December 2025
Accepted: 22 December 2025
Published: 29 December 2025

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Abstract: Oral bioengineered models have emerged as a leading platform for oral health research because they can closely resemble the structure, biology, and function of native oral tissue. Through the combination of living cells with sophisticated biofabrication technologies and smart biomaterials, these models outperform other model systems, such as conventional 2D in vitro cultures and animal models. This article summarizes recent advances in the generation and characterization of bioengineered oral models, as well as their applications in modeling human oral diseases, regenerative dentistry, and biomaterial testing. The actual limitations and forthcoming opportunities towards translational and personalized oral care is also emphasized.

Keywords: Bioengineered models; Oral health; Tissue engineering; Biomaterials; Regenerative dentistry.

Until now, studies on oral health have primarily made use of animal models and cell cultures in two dimensions. The above techniques have greatly furthered scientific understanding but are limited in their ability to recreate the well-organized three-dimensional structure, cellular diversity and mechanical environment of oral tissues. Therefore, bioengineered models have received attention as more physiologically relevant systems for studying oral biology, pathology and therapeutic responses [1].

Successful bioengineered oral model development depends on the unification of appropriate cell sources, biomaterials and fabrication technologies. The most commonly used cells are human oral keratinocytes, gingival fibroblasts, periodontal ligament cells and dental pulp stem cells due to their regenerating capacity and clinical relevance. Scaffolds to mimic the extracellular matrix need to be composed of natural biomaterials (collagen, gelatin, chitosan) or synthetic biodegradable polymers. With the recent progress in 3D bioprinting and microfluidic techniques, it is now possible to achieve better spatial control over cell placement, enhanced cell–cell interaction and fluid flow-based dynamic culture [2].

Stringent characterization is critical for the biological relevance and the reproducibility of bioengineered models. Histological staining and state-of-the-art imaging are used for morphologic identification. Gene and protein expression for tissue-specific markers are examined by molecular analyses. Functional testing includes tests for cell proliferation, differentiation, inflammation and mineralization. It is even more important for the oral tissues because they are exposed to forces produced by mastication [3].

Oral bioengineered models are broadly used for investigating periodontal diseases, dental caries, oral mucosa related pathologies and oral cancer. In regenerative dentistry, such models enable the study of regeneration of dental pulp, periodontal and alveolar

bone. They additionally represent established bases for biomaterials testing, testing of dental implants, drug screening and the development of individualized therapies.

Even with these benefits, bioengineered oral models suffer from production expenses, technical complexity, limited vasculature and are an incomplete immune system representation. The upcoming research will be likely directed toward organ-on-a-chip rights, multi-tissue integration, and the use of artificial intelligence-supported analysis to enhance the translational significance [4].

Bioengineered models are an innovative aspect of oral health research. Rapid progress in biomaterials science, stem cell biology and biofabrication technologies is likely to improve further their predictive potential and medical relevance.

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Biomaterials Journal
http://www.biomatj.com
Online ISSN: 2812-5045

Type of the Paper (Mini-Review)

The Advent of 4D Dentistry: Integrating Time and Technology in Contemporary Oral Care

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Citation: *Rasha M. Abdelraouf. The Advent of 4D Dentistry: Integrating Time and Technology in Contemporary Oral Care. Biomat. J., 4 (5), 3 – 5 (2025).*

https://doi.org/10.5281/zenodo.18056110

Received: 10 December 2025

Accepted: 22 December 2025

Published: 29 December 2025

Abstract: The development of digital dentistry is now moving from a 3D static to a 4D dynamic model. Although 3D modalities (e.g., CAD/CAM, CBCT) have overcome the issue of spatial accuracy, they are constrained by an inability to interpolate temporal physiological dynamics occurring in the oral environment. 4D dentistry adds the dimension of time, expressed in two fundamental ways: smart materials that change shape according to programmed sequences (4D printing) and real-time functional motion integrated into digital workflows. Here, we discuss how responsive polymers and hydrogels are currently transforming the fields of myofunctional orthodontics and tissue engineering, whereas dynamic jaw tracking is taking prosthodontic accuracy to the next level. From "inert" to bio-responsive systems. Developing from the old to a new paradigm: The transition from inert restorations to bio-responsive materials in 4D dentistry, the New Normal for these smart systems, would be associated with self-directed readiness, anagenic biocompatibility, and superior clinical longevity.

Keywords: digital dentistry, 4D printing, dental restorations.

Introduction

For years the gold standard for construction systems in dentistry was based in 3D replicate anatomic structures. Intraoral scanning and 3D printing Discovery of independent practice to be highly accurate prosthetics possible, but these are still "passive" structures that fail to adapt dynamically pressure biological?? from the mouth. The oral cavity is known to be a dynamic environment, which is subjected to changes in pH and temperature and to variable masticatory loads over time. Therefore, 3D-printed appliances frequently encounter "static failure" or material fatigue in which the appliance is unable to fit when the biological base has changed. This limitation has led to the need of 4D dentistry, which aims at transitioning from synthetic to live tissue by reshaping morphology [1].

At the heart of the 4D shift is using shape-memory alloys (SMAs) and shape-memory polymers (SMPs). That's the concept behind 4D printing: a 3D-printed object is designed to behave differently—from, say, solid to ultra-squishy—when it's hit with some particular trigger (like the heat energy in human saliva or the acidity of bacterial biofilms). In orthodontics, for example, an aligner that is 4D-printed could be designed to apply certain forces at temperatures

in the mouth as it progresses around a sequence of specific positions and thereby remove the requirement for multiple dozens of different plastic trays. Beyond materials, the 4D "New Normal" includes functional diagnostics. Contemporary workflows also include 4D motion capture of the patient's mandibular movements during speech and swallowing, so as to not only spatially restore a final restoration but for its neuromuscular pattern – functional [2].

Applications of 4D Dentistry

Current applications of 4D technology in dentistry The current application of 4D in dentistry is wide-ranging [3]:

Orthodontic: With 4D imaging it is possible for dentists to predict tooth movement in time, simulate outcomes, and change orthodontic apparatuses while the patient views what has been simulated.

Implantology: Dynamic imaging allows accurate guided implant placement by taking into account bone density, healing and soft tissue adaptation over time.

Restorative and Prosthetic Dentistry: Time-related digital models are useful to observe wear patterns, occlusal adjustments, and prosthesis adaptation at follow-up observations over time; increasing the duration of restorations.

Periodontics and Oral Surgery: The real-time imaging helps in tracking tissue healing/ regeneration after surgery, allowing for preemptive interventions if complications develop.

Advantages

The 4D concept provides a range of advantages: improved diagnostics, increased treatment predictability, patient-specific care planning and higher levels of patient compliance thanks to visual representation of the progress during treatment. Patients are allowed to visualize the expected results in a timeline fashion, leading to higher compliance and satisfaction. Additionally, 4D systems can work together with AI algorithms for predictive analytics and to better optimize treatment [4].

Challenges and Limitations

Although the future is bright, 4D dentistry also confronts a series of challenges. Challenges: High expense of equipment and software, as well as data storage demands, along with necessary training can be considered barriers to wider adoption. Also, standardization of protocols for dynamic imaging and transmission to electronic dental records is ongoing. Ethical concerns about patient data privacy and surveillance also need to be carefully addressed [5].

Future Perspectives

Integration of 4D dentistry with AI, machine learning and tele-dentistry has the potential to revolutionize patient care to allow for remote surveillance and predictive intervention. With technology becoming more affordable, 4D dentistry is poised to shift from being a high-tech fad to the standard of care that focuses on preventive, patient-centered oral health with evidence-based practice.

Conclusion

The advancement to 4D dentistry is a leap forward in dental health. Notwithstanding the potential for this technology, entrance into 4D dentistry requires a rethinking of established clinical procedures. Normalized fatigue life of smart material and high computational load for 4D motion processing are still big concerns. But as material science and digital processing moves closer to convergence, the 4D paradigm is ready to transition from an experimental frontier to being standard of care providing restorations that heal, adapt and move in concert with the organism. The possible advantages for both clinicians and patients indicate that 4D dentistry is not simply a vision of the future but rather an upcoming reality in dental practice.

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Biomaterials Journal
http://www.biomatj.com
Online ISSN: 2812-5045

Type of the Paper (Research Article)

Role of Thread Geometry in the Compressive Strength of Dental Implants

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Citation: *Adil Elmokhtar A Esslami, Asaad Nuri Elbalog, Sirageddin mohamed alhmadi. Role of Thread Geometry in the Compressive Strength of Dental Implants. Biomat. J., 4 (5), 6 – 15 (2025).*
https://doi.org/10.5281/zenodo.18077433
Received: 10 December 2025
Accepted: 22 December 2025
Published: 29 December 2025

Abstract:

The objective of this study was to evaluate the effect of different dental implant thread geometries on the compressive strength of dental implants. Based on thread form, sixty custom-made grade 4 titanium dental implant screws were prepared and grouped into; V-Shape (Group 1), Square Shape (Group 2), Buttress Shape (Group 3), and Reverse Buttress Form (Group 4); Standard lab analysis set up as recommended in ISO14801. The implants were embedded in an acrylic block and tested under a 30 °off-axis compression load. The compressive strength test was carried out using a Universal Testing Machine (UTM). The resulting data were analyzed by one-way ANOVA followed by Tukey's test. The results showed that Group 1 (V-Shape) was found to be statistically significantly higher than Group 2 (Square Shape Threads) and Group 3 (Buttress Thread Shapes). Group 4 (Reverse Buttress Shapes) was found to be statistically significantly higher than Group 2 and Group 3. There is no statistically significant difference between Group 2 and Group 3 ($p>0.05$). There is no statistically significant difference between Group 1 and Group 4 ($p>0.05$). The results of this study suggest that various thread designs may play a critical role in the fracture load of implants under static load, where the reverse buttress and V-shape thread designs show better resistance than the other shapes.

Keywords: *Dental Implants; Thread Shapes; Compressive strength.*

1. Introduction

Long-term clinical success of dental implants depends mainly on osseointegration and the effective transmission of occlusal forces to host alveolar bone. The thread design is aimed to provide not only better initial fixation but also improved load transfer, where axial forces are transformed into compressive stresses at the bone–implant junction. Because bone is more tolerant to compressive stress than to tensile or shear and its complexity increases dramatically the stresses under compressive load, the design of implant threads may influence considerably their strength in

compression and overall mechanical behavior when subjected to function loads [1]. How and where those forces are transferred is determined by the particular profile of the thread, be that V-shape, Square, Buttress or Reverse Buttress. V-shape threads are often used for their self-tapping properties, but square or buttress designs are usually also proposed in theory to transfer more of the occlusal load into advantageous compressive rather than disadvantageous shearing stress [2].

Although numerous forms of designs afford available on the commercial market, there is still a crucial clinical issue that has not been elucidated yet: what impact do these particular changes on geometry generate from an intrinsic compressive strength perspective for implant body. The majority of literature depicts the stress distribution in peri-implant bone, although the structural response of the titanium body to axial deformation or fracture is equally important, especially in posterior areas where chewing forces are a maximum [3]. A cutting thread profile that enhances bone-implant interface but excessively reduces the core diameter or generate a stress concentration points into the metal could cause early failure. It follows that the mechanical limits of such geometries need to be studied under standardized loading conditions, so as to prevent that aesthetic and biological improvements are being achieved at the expense of structural integrity [4].

Hence, there is a demand to understand carefully how geometry of the implant thread affects its compressive strength, an important mechanical property in relation with implant durability since it does not deform or fracture under occlusal forces. Although many studies have been done considering stress distribution through finite element analysis or primary stability in various bone qualities, relatively few looking directly at the axial load capacity of implants with different thread geometries under controlled mechanical test conditions. This knowledge gap restricts the ability of clinicians and industrial developers to make informed decisions on optimal implant design for improved mechanical behaviors and long-term clinical efficiency. Therefore, the research question that was evaluated in this study was whether several dental implant thread designs (V-shape, square shape, buttress shape and reverse buttress form) significantly influence the compressive strength of dental implants. As the null hypothesis, there is no significant difference between two groups in the compressive strength of implant body that has different thread design.

2. Material and method:

Sixty (N= 60) custom made bone level dental implants Ti-GR4 (ASTM F67) [5] ranging in diameter, length (4.1x10 mm), surface roughness, material and body design were tested with four different thread shapes namely V-shape,

Square shape, Buttress and Reverse buttress threads under laboratory conditions using test setup equivalent to that recommended by ISO 14801 protocols [6].

The materials utilized in this study, along with their specifications, and manufacture recommendations are listed in Table 3.

Table 1. Implant and Abutment Specifications.

Implant Specification				
Implant thread shape	V-shape	Square shape	Buttress shape	Reverse buttress shape
Diameter	4.1 mm	4.1 mm	4.1 mm	4.1 mm
Length	10 mm	10 mm	10 mm	10 mm
Implant Type	Bone Level	Bone Level	Bone Level	Bone Level
material	Ti-GR4(ASTM F67)	Ti-GR4(ASTM F67)	Ti-GR4(ASTM F67)	Ti-GR4(ASTM F67
Fixture design	Self-taping	Self-taping	Self-taping	Self-taping
Surface treatment	SLA	SLA	SLA	SLA
Surface roughness	1.8µm	1.8µm	1.8µm	1.8µm
Thread depth	0.44 mm	0.44 mm	0.44 mm	0.44 mm
Connection type	Conical	Conical	Conical	Conical
Abutment Specifications				
Material	Ti-GR5 (Ti–6Al–4V alloy)			
Gingival hight	4 mm			
Diameter	5 mm			
Torque	30 Ncm			
Lot no	PAD50GH4M18191004			

The 60 implant specimens were divided according to shapes of implant threads into four different groups as follow; Group #1: V- Shape Thread Implant Design (n=15). Group #2: Square Shape Thread Implant Designs (n=15). Group #3: Buttress Shape Thread Implant Designs (n=15). Group #4: Reverse Buttress Shape Thread Implant Designs **(**n=15).

The specimens were oriented vertically along the center of an acrylic resin block (Orthodentic Base Polymer Self-Cure Acrylic Resin) (Ref No: 0254), with a modulus of elasticity greater than 3 GPa According to the standard methods described in ISO 14801 and in order to keep the gap between the resin platform and top of implant fixture (apical from nominal bone level) at 3 ± 0.1 mm in order to simulate clinical bone loss.

A common titanium straight abutment by gingival height of 4mm and diameter of 5mm was connected to each implant specimen with a screw torque being at about 30 Ncm, according to the manufacture instructions, by means of using torque ratchet. In the surface of each specimen**.** A 5mm diameter hemispherical Cobalt-Chromium Alloy Crown was cast and seated onto an abutment of each specimen by using Zinc Phosphate Cement (Roorkee LOT#. ZP0802) and cap, this structure referred to as the "cap", Compliance with the specifications: The cap was suitably made up in accordance with prescribed requirements. An 11 mm separation between the simulated bone level and center of sphere of crown (The moment arm that is needed according to ISO 14801 that avoided lateral constraint.

Implant samples all were placed in the center of the resin; they were brought until static (compressing) load. In this study's experiment, a custom-made implant holder was made of stainless steel (Ufuk Kontrol Araçları, Gebze_istanbul) and 30 ± 2° away from the vertical load application according to ISO 14801 requirement regarding specimen holder. Afterwards the acrylic blocks with implant/abutment/crown assemblies were inserted in a testing device. The test fixture together with specimens was mounted 30° angle with the respect to the applied load according to ISO 14801 guidelines.

Compressive test was conducted in an INSTRON Universal Testing Machine (3345, 3345J7324; INSTRON, USA) maintained at the Hard Tissue Laboratory of Yeditepe University Maximum load of 5kN was evaluated using a small Instron tabletop tensile tester. It is suitable for tension and/or compression applications and able to be mainly operated by Bluehill Lite software equipped with this machine. Each test specimens were pre-loaded slightly (0.5N) to ensure that all parts and the acrylic blocks were fully seated prior to each testing session.

Off-axis loading was applied to every implant arrangement with the tensiometer's vertical piston that moved downward at a constant speed of 0.5 mm/min and 30° oblique force as prescribed by ISO 14801 standard, on the hemispherical cap of each implant, using flat indenter until either specimen breakage or indicating decrease in force (deformation) by the testing machine. The test didn't stop until the implants broke (some tests stopped with a bent implant), testing machine automatically stopped working when there was a sudden drop in specimen force.

Statistical analysis

Statistical analysis Data were analyzed with IBM SPSS 22.0 statistical package program and presented as the mean and standard deviation. Normality of quantitative data distributions were examined using the Shapiro-Wilk test. The parameters are compared between groups according to their normal distribution by the one-way ANOVA test, and difference is determined with Tukey HDS test. A p-value less than 0.05 was considered statistically significant.

3. Results:

The data were displayed as Minimum-Maximum loads (max) and as mean standard deviations (SD). One-way ANOVA test was utilized for the intergroup comparisons of parameters with normal distribution and Tukey HDS test was utilized for the determination of the group causing the difference. Statistical significance was accepted as $p<0.05$. In this study, V-Shape Threads applied samples are named as "Group 1", Square Shape Threads applied samples "Group 2", Buttress Thread Shapes applied samples "Group 3", Reverse Buttress Shapes applied samples are named as "Group 4". The mean and SD of Maximum Loads (N) was 482,38±45 in V-Shape Threads (Group 1), 422,08±41 in Square Shape Threads (Group 2), 407,95±46 in Buttress Thread Shapes (Group 3), and 484,53±57 in Reverse Buttress Shapes (Group 4). (Table 2). According to the results of the one-way ANOVA test, there are a statistically significant difference between the groups in terms of Maximum Loads (N) levels (p:0.000; $p<0.05$).

Table 2. Evaluation of the compressive strength of the groups in terms of maximum Load (N)

	Mean±SD
Group 1	482,38±45
Group 2	422,08±41
Group 3	407,95±46
Group 4	484,53±57

*p<0.05.

As a result of post hoc analysis, the fracture resistance of Group 1 was found to be statistically significantly higher than Group 2 and Group 3 (p_1:0.007; p_2:0.001; $p<0.05$). Group 4 was found to be statistically significantly higher than Group 2 and Group 3 (p_1:0.005; p_2:0.001; $p<0.05$). There is no statistically significant difference between Group 2 and Group 3 ($p>0.05$). There is no statistically significant difference between Group 1 and Group 4 ($p>0.05$). (Figure 1)

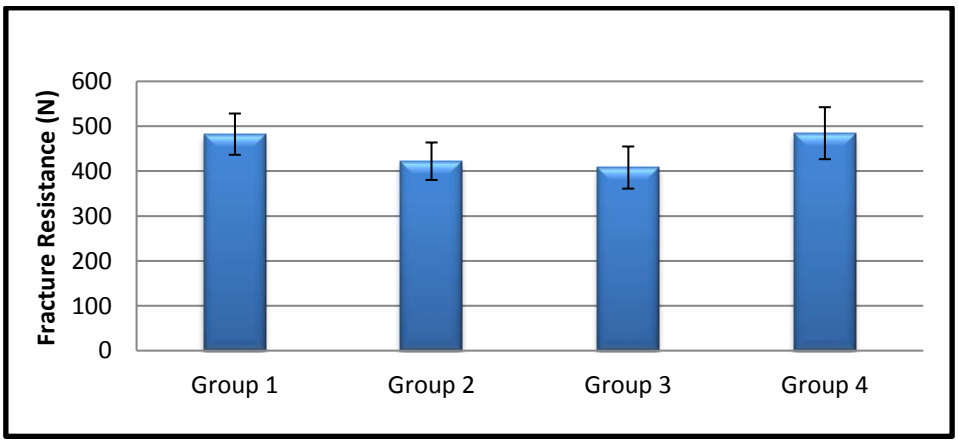

Figure 1. Bar Chart Illustrates the Mean of Maximum load (N).

4. Discussion:

The utilization of dental implants in the treatment of partial and full edentulous patients has made it feasible to restore function into the stomatognathic system. while preserving dental tissues and the longevity of the treatment. Because of the high efficacy, dental implant has frequently been used as a treatment modality in many patients [7].

Global implant survival rates in implants placed in partially edentulous portion after 3-7 years of loading are presented at 95.3% [8]. Nevertheless, and perhaps not surprisingly many problems exist with dental implants which appear to be encountered and dental surgeons need to be aware of in order to prevent additional biomechanical complications and implant failure [9]. These issues may be disengagement or fractures of the prosthesis and abutment screws and implant fractures [7]. Of all mechanical failures, implant fractures are considered the most annoying as they may occur after functioning over time. Implant Fracture The incidence of implant fracture varied significantly (0.0% to 3.45%) in the available literature [9].

Four front thread forms of screw-retained implants were used in this study: V-shape thread, square shape thread, buttress thread form, and reverse buttress thread form (defined according to the thickness of the threads and face angle) [10]. Misch *et al.* has described certain functions of threads; to enhance the resistance to initial contact, increase the surface area for functional use, add stability, and allow for dispersions and there have been other types distribution of stress in the zones of interfacial alongside suitably modified pitched-threads [11]. Steigenga *et al.* observed that both V-shape and reverse buttress threads exhibited similar stress figures, while square threads endured much lower stress in compression and, more importantly, shear forces [12]

By the other hand, posterior to initial healing the square thread has higher values of reverse torque when compared with V-shaped and reverse buttress screws [13]. The high level of stress is mainly transmitted through the implant surface of the thread valley, which decrease/stresses bone at this interface, that could favorably contribute to osseointegration and be advantage for threaded implants with improved bone-implant contact. By the way, there was more areas of low interfacial stress at the implant location in trabecular bone for the square thread compared to a triangular thread [13].

Triangle thread forms and straight–body forms, compared to non-threaded implants, square thread forms and tapered body form all showed higher maximum tensile and compressive stress by Rismanchian *et al.* [14]

A similar study was performed by Eraslan et al. [15], but with 4 different thread forms and a 100 N static axial load. The study found that the highest average stress was contained in cortical bone of cervix region around proximal thread Take square type Th least induced stress. At the same time, earlier FEA and animal experiments demonstrated that square thread shape results in effectively stress distribution and bone–to–implant contact area (BIC) [15–17].

At present, very little work has been done regarding the influence of implant thread design upon the overall strength of an implant and no direct comparisons between different available designs exist in literature. The purpose of this study was to compare and assess the impact of various dental implant thread shapes on fracture resistance.

The null hypothesis was rejected when we found significant differences in the compressive strength among various dental implant thread forms. Groups 4 (reverse buttress) and Group 1(V- shape threads) has statistically significant higher strength in the static load test than both square thread & buttress threads designs. V-shaped and reverse buttress normally have a face angle of 30 and 15 degree which has been shown to create more shear force at the interface than square threads that present no real face angles and thereby lowest shearing power from among the group. The axial loads transmitted by the v-threaded and reverse buttress threads are primarily a combination of compression, tension and shear [11]. This finding was consistent with Bumgardner *et al*. [13] that demonstrate altering the face angles may affect forces at the implant to bone interface. Small face angles type of facesolve.com tend to raise the tensile and compressive kind forces while an increase in the face angles has been shown to increase the shearing kind forces at implant to bone interface. This concept is observed to take place any of the thread shapes of their group. These shearing stresses have been observed to ultimately produce even more defects [18].

Additionally the results of this study show that thread macro-geometry has a clear impact on the maximum compressive load to failure of dental implants, with V-Shape (Group 1) and Reverse Buttress (Group 4) designs providing better mechanical resistance than Square (Group 2) and Buttress (group 3). There was no statistically significant difference between Group 1 and Group 4 indicating that elements of these two configurations retain specific structural features which except with less damage to the core during axial load application. On the other hand, as much lower compressive strengths were measured in G2 and G3; these also appear having a higher stress concentration factor or decrease effective area of implant core [18].

The improved behavior of V-shape thread could be due to its geometric advantage in load spread over the flank of screw threads. Wide under the head and to a point, in V-profile forming base of thread that continuously increases in

size until thread merges with implant body eliminating trenches that often lead to cut outs or mechanical failure. The observation is consistent with biomechanical theory that shows that L or rounded interfaces create a lower level of stress intensity factor than sharp, perpendicular interfaces. In the same manner, the structure of Reverse Buttress with subtle angled load-bearing face is apparently able to offer a sturdier structural configuration that withstands deformation from high-magnitude compressive forces and hence reminiscent the strength characterizes V-shape and traditional designs [11].

In comparison, the Square and Buttress thread types had decreased compressive strength values. In Square, the 90-degree angles of the thread-to-core junction result in high stress-concentration areas. When the corners are compressed, they serve as crack initiation regions. Also, to accommodate the broad flat abutment geometry of a Square, or rectangular Buttress thread and have some similarity of outside diameter the inner core on many implant designs has been reduced. Compressive strength is directly proportional to the cross-sectional area of the core, and this decrease in value clarifies the decreased load-bearing capacities by groups 2-3. The statistical equivalence of these two groups indicates that both geometries exerted equivalent mechanical sacrifices on the titanium base [13].

These findings draw attention to a fundamental compromise in implant design, although Square and Buttress threads are frequently promoted in clinical studies as facilitators for increasing the bone-to-implant interface and minimizing stress in the surrounding bone, they may also inherently compromise the integrity of the implant's central section. In these cases, a thread design such as V-shape and Reverse Buttress may be more appropriate to avoid fatigue-related fractures, particularly in patients presenting higher occlusal loads (e.g., bruxism patient or posterior mandible rehabilitation). Further investigation of internal stress vectors that form and results in the failure patterns we identified will also require FEA (Finite Element Analysis) for future studies.

In conclusion, the present study shows that dental implant thread macro-geometry has a significant effect on structural load-bearing capacity rejecting the null hypothesis. Statistically higher compressive resistance of the V-Shape (Group 1) and Reverse Buttress (Group 4) designs compared to the Square and Buttress type might be attributed to a more robust core diameter, leading to lower stress at thread-core junction. Although Square and Buttress designs are commonly employed to maximize load transfer between the surrounding bone, considering this evidence they may contribute as a limiting factor this effect in the implant body. Thus, in the presence of high masticatory loads, V-shape

or Reverse Buttress profiles could be preferred in clinic to increase the implant mechanical resistance and prevent fatigue.

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