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Ayushi Pandey[1], Rajeev Kumar[1], Anuj Kumar Sharma[2]

[1]Department of Mechanical Engineering, Institute of Engineering & Technology, Lucknow, India

[2]Department of Mechatronics, Centre for Advanced Studies, Lucknow, India

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COMPARATIVE FINITE ELEMENT ANALYSIS OF POLYETHERETHERKETONE (PEEK) AND TITANIUM ALLOY DENTAL IMPLANTS: BIOMECHANICAL AND MATERIAL PERFORMANCE EVALUATION

Ayushi Pandey[1*], Rajeev Kumar[1], Anuj Kumar Sharma[2]

[1]*Department of Mechanical Engineering, Institute of Engineering & Technology, Lucknow, India*

[2]*Department of Mechatronics, Centre for Advanced Studies, Lucknow, India*

ayushipandey409@gmail.com, rajeev.kumar@ietlucknow.ac.in, anujksharma@cas.res.in

Corresponding author

Abstract: *Dental implants are widely utilized for oral rehabilitation, with their clinical success determined largely by the degree of osseointegration and biomechanical compatibility of the implant material. Titanium and its alloys, particularly Ti-6Al-4V, have been extensively used due to their high strength and corrosion resistance; however, their high elastic modulus (110 GPa) compared to that of cortical bone (13-15 GPa) leads to stress shielding and potential bone resorption. Polyetheretherketone (PEEK), a high-performance thermoplastic polymer, offers a lower elastic modulus (3-4 GPa) closer to bone, alongside excellent wear resistance, radiolucency, and biocompatibility. In this study, finite element analysis (FEA) was performed using ANSYS 19.2 to compare the mechanical performance of PEEK and titanium-based dental implants under vertical loading from 100 N to 760 N with an additional 40 N lateral force. The analysis revealed that PEEK implants exhibit lower localized stress concentrations and more uniform stress distribution across the bone implant interface, with deformation values around 0.34 mm at maximum load, compared to 0.015 mm for titanium implants. These findings demonstrate PEEK's capacity to replicate the biomechanical behavior of bone, thereby reducing stress shielding and enhancing load transfer. Hence, PEEK represents a promising alternative material for next-generation dental implants offering improved mechanical and biological performance.*

Keywords: Dental implants, Bioactive materials, Polyetheretherketone, ANSYS, Finite element analysis.

1. Introduction

The quality of life of those facing tooth loss, regardless of the cause, can be addressed with dental implants. A restorative solution was developed based on Branemark's principle of Osseo integration and the practicality of artificial dental prosthetics [1]. The use of dental implants for oral and orthodontic rehabilitation has grown substantially predominantly due to patient demand since 1969. Occlusal pressures typically induce stresses and strains inside a dental implant, impacting the adjacent mandible [2]. If the bone attrition is extremely severe for the human body, the effort becomes exceedingly difficult. Materials exhibiting osteoinductive, osteogenic , and osteoconductive properties are desirable [3]–[5]. The direct The chemical link between the implantation device and tissue surrounding it at their interface originates from a biological mechanism that activates bioactive materials. Consequently, bioactive materials are crucial for the progression of implants. Ti-6Al-4V are often major employed materials in the medical field. However, they exhibit distinctive shortcomings, such as hypersensitivity, inferior visual appeal, and notable leaching of aluminum and vanadium ions, which restricts their use [6]–[9]. Several patients have voiced allergies to titanium alloy [9]. The principal reason

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for the insufficient variation of mechanical stress The disparity in modulus of elasticity between titanium alloy (Ti) and bone resides between the dental prosthesis and the jawbone as a whole.Titanium's elastic modulus of 110 GPa, while bone has nearly between 13.8 and 14 GPa as elastic modulus [6] . The difference in modulus of elasticity leads to a condition that is termed stress shielding, in which the alveolar bone protects the implant from occlusal forces, triggering bone resorption and an erosion in bone volume[6]. As a result, alternatives are carefully analyzed and assessed to alleviate these limitations. The preferred material for implants in dentistry as ceramic-based, particularly zirconia, recognized as possessing elevated mechanical strength, reduced plaque affinity, and exceptional resistance to corrosion in biological fluids. Moreover, it fulfills the aesthetic demands of consumers. However, its elastic modulus, quantified at 210 GPa, limits prospective supplementary applications. In light of these constraints, synthetic polymers were examined more thoroughly due to their broad application possibilities (10-14).

Polymers including PLA, UHMWPE, PTFE, PMMA, and PGA were under review. Some were feeble, while others were excessively pliant and unable to endure the requisite power as stated. In composite polymerization, the monomer chains commence cross-linking, resulting in the generation of tension at the interfaces between the tooth and the restoration. The tension is influenced by the elastic modulus and the volumetric shrinkage during the polymerization process. Optimizing the curing rate enhances the composite's mobility during polymerization and reduces synthesis stress. The only documented approach to reduce shrinkage is to decrease the quantity of methacrylates or use polymerization chemical processes. The peak polymerization shrinkage occurs during the transition from a viscous state to a viscoelastic state, and subsequently during the elastic phase. The soft-start photo activation method yields a modest increase in light intensity, hence limiting polymerization shrinkage [15].

1.1 Polyetheretherketone as a Substitute Material for Implant Dentistry

Polyetheretherketone, a synthetic polymer, synthesized in 1978 from the monomer unit of ether-ether ketone (Fig. 1)[16]. The polymerization process involves sequential dialkylation of bis-phenolates. Polyetheretherketone's molecular structure consists of interlinked aryl rings featuring ether and ketone functional groups, placing it in the polyaryletherketone polymer category. Either of these methods can manufacture Polyetheretherketone[17], [18]. The preliminary method entails employing different amounts of 4,4'-difluorobenzophenone in conjunction with an electrophilic agent, such as 3,5-difluorobenzophenone or 2,4-difluorobenzophenoneThe components engage in nucleophilic aromatic substitution conjugation reactions with hydroquinone in the polar solvent diphenyl sulfone at a temperature of 300°C. Polyetheretherketone is a semi-crystalline material having a melting point of around 335°C. The amorphous Polyetheretherketone has a glass transition temperature of 145°C and a melting temperature of 345°C for the unaltered polymer.[19]–[22]. Polyetheretherketone is a thermoplastic organic polymer distinguished by a polycyclic aromatic structure that closely approximates the color of teeth. It features a combination of mechanical and chemical qualities that make it suitable for biomedical applications. PEEK, a biocompatible polymer, has been employed in fixation of tissue and rehab treatments [16], [20]. This approach was first used in

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medicine during the 1980s and gained prominence in the 1990s period as an alternative to metal implants for many orthopedic treatments, including spine surgery, cardio-vascular treatments, fracture fixation, maxillofacial surgery, and prosthetic limbs[23], [24]. It served as an alternate for stainless steel bone plates. Its remarkable mechanical qualities and biocompatibility frequently make it a preferred choice for temporary abutment. The lasting efficacy of an implant is primarily contingent upon the care of soft tissues, a process that is complicated and challenging to execute [25]. Thus, a temporary prosthesis is typically utilized before the definitive prosthesis to facilitate the development and healing of the soft tissues. Polyetheretherketone is a radiolucent material with exceptional mechanical qualities. It is non-toxic, thermally stable above 300°C, and devoid of mutagenesis effects. The material has exceptional resistance to abrasion and wear, together with a low coefficient of friction.The primary objective in medicine is to improve the power-to-weight ratio via the use of sophisticated materials identified through research [19], [21], [26], [27]. Polyetheretherketone can be altered before polymerization using reinforcing materials like glass and carbon fibers, or after polymerization using chemical processes such as sulphonation, and nitration [28]–[32]. Polyetheretherketone implants enhance the stability of soft tissues. Furthermore, these implants exhibit X-ray transparency, which is advantageous for evaluating therapeutic progress using computerized tomography or nuclear magnetic resonance amid the surgical convalescence [23], [33].

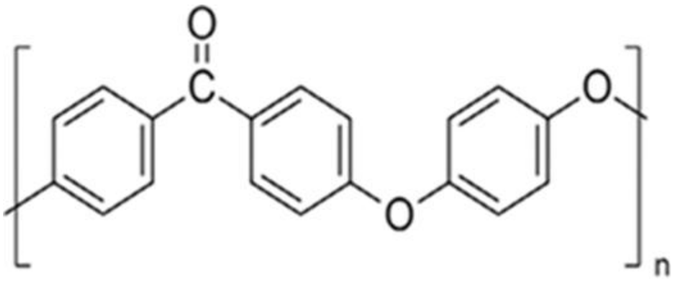

Figure 1. Illustrates the chemical structure of Polyetheretherketone [23].

Mutation of Polyetheretherketone via Pre-polymerization. An important benefit of employing Polyetheretherketone for implants is its lower modulus of elasticity (3-4 GPa), which roughly matches that exhibited by human bone (14 GPa) and dentin (15 GPa). Enhancing Polyetheretherketone with glass or carbon fibers, as demonstrated in Table 1, may elevate its modulus to a range of 12-18 GPa[19]. A variety of reinforcement Polyetheretherketone composites have been manufactured to enhance the characteristics of pure Polyetheretherketone material. This involves nano-TiO_2/Polyetheretherketone, which combines Polyetheretherketone with titanium alloy oxide nanoparticles [34], [35], as well as short carbon fiber-reinforced Polyetheretherketone (CFR-Polyetheretherketone) and glass fiber-reinforced Polyetheretherketone (GFR-Polyetheretherketone). CFR Polyetheretherketone comes with an elastic modulus of 18-20 GPa, while GFR-Polyetheretherketone possesses an elastic modulus of 12 GPa[29] CFR-Polyetheretherketone is frequently utilized in the palatal area of maxillary obturator dentures for individuals with extensive oral-nasal deformities [30]. The life expectancy and exceptional fatigue resistance of CFR-Polyetheretherketone may potentially result in a decrease in implant fractures. n-TiO_2/Polyetheretherketone has shown the capacity to enhance the bioactivity of

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Polyetheretherketone and facilitate osteoblast adhesion. Employing finite-element analysis (FEA) is recommended for evaluating the efficacy of dental implants made from short/continuous carbon-fiber reinforced Polyetheretherketone (CFR-Polyetheretherketone) [35]. This material is proposed as a successor in comparison to traditional titanium alloy alloys, with the objective for lowering stress shielding. However, clinical research on Polyetheretherketone dental implants is rarely conducted, leaving the potential disparities in bone resorption between Polyetheretherketone and titanium alloy implants unconfirmed [29]. Additional clinical study is necessary to comprehensively ascertain this particular fact. CFR-Polyetheretherketone is favored for how well it complies and adapt with contemporary imaging techniques. The energy dissipation theory asserts that negligible deformation of an implant signifies that the applied loading conforms to the energy conservation principle inherent to stiff implants conducted a study leveraging the finite element technique (FEM) to analyze the stress variation of 30% CFR-Polyetheretherketone and titanium [31]. The study focused on a potential dental implant made from infinite carbon fiber, considered an advanced alternative to CFR-Polyetheretherketone. The results demonstrated that this type of implant reduces stress peaks at the bone-implant contact by alleviating elastic deformation performed a comparative examination of powder-filled Polyetheretherketone and 60% parallel-oriented continuous carbon fiber CFR-Polyetheretherketone. The conclusion suggested that the continuous carbon fiber CFR-Polyetheretherketone implant could not be endorsed due to heightened stress concentrations. CFR-Polyetheretherketone is cosmetically unpleasant due to its dark hue that arise from carbon fibers. Consequently, contradicting evidence exists that does not yield a definitive knowledge about Polyetheretherketone composites [36].

Table 1 The tensile strength and elastic moduli of different materials

MATERIAL	Tensile Strength (MPa)	Young's Modulus (GPa)	References
Polyetheretherketone	80-85	3-5	Sandler, 2002 [37]
Cortical Bone	103-120	15	Rho, 1993; Martin, 1989; [38], [39]
CFR-POLYETHERETHERKETONE	120	18	Sandler, 2002 [37]
Polymethylmethacrylate	48-76	3-5	Zafar,2014;Vallittu, 1998;[40], [41]
Titanium	954-976	102-110	Lee, 2012[22]
Dentin	104	15-30	Chun, 2014; Rees, 1993[42]–[44]
Enamel	47.5	40-83	Staines,1981; Cavalli,2004;Rees, 1993;[27], [43], [45]
GFR-POLYETHERETHERKETONE	147-154	12	Lee, 2012[46]
Zirconia	77-85	210	Lee, 2012[46]

1.2 Post-Polymerization Surface Modification of Polyetheretherketone.

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Pure Polyetheretherketone exhibits intrinsic hydrophobicity, shown by a water contact angle of 78-89°, and demonstrates a significant degree of hydrophobicity. It exhibits superior bioinert characteristics, although demonstrates inadequate tissue in growth in the adjacent region. Furthermore, Polyetheretherketone can be disinfected via ethylene oxide, gamma radiation, or steam under pressure. Polyetheretherketone has had nanoscale modifications to improve its bioactivity. Conventionally, Polyetheretherketone is coated by plasma spraying to integrate bioactive materials such as calcium hydroxyapatite (HAp) or titanium alloy particles. The plasmapresent will eventually elevate the temperature of the bioactive materials, leading to their liquefaction and altering the implant's surface, so creating an irregular layer. Although the application of the bioactive substance via spraying may be appropriate for bigger implants, the resultant coating is ineffective for smaller dental implants. The principal reason for dental implant failure is the formation of an irregular surface and a substantial layer of hydroxyapatite on the implant, which eventually becomes delaminated .A further disadvantage is the elevated temperature, which may jeopardize the integrity of the Polyetheretherketone structure owing to its low melting point of 340°C. Polyetheretherketone may be modified using many plasma treatments, including methane and oxygen plasma, hydrogen/argon plasma, ammonia/argon plasma, and nitrogen and oxygen plasma. Surface coatings may be applied using techniques such as physical vapor deposition and aerosol deposition. The used methods include plasma immersion ion implantation, cold spray method, electron beam deposition, ionic plasma deposition, radio-frequency deposition, vacuum plasma spraying, and spin coating.

Titanium alloy dioxide (TiO_2), hydroxyapatite (HAP), and hydroxyapatite-filled Polyetheretherketone (HAF-Polyetheretherketone) nanoparticles are combined utilizing melt-blending to fabricate bioactive nanocomposites and enhance surface roughness.The chemical modifications entail the amalgamation of bioactive compounds, which are preliminarily evaluated using experimental methods such as sulphonation, amination, and nitration (refer to Table 2). Moreover, the modified Polyetheretherketone has superior tensile properties relative to the unblended Polyetheretherketone. It is proposed that hydroxyfluorapatite nanoparticles (n-FHA) may inhibit bacterial adhesion and reduce the likelihood of peri-implantitis. Consequently, n-FHA demonstrates favorable potential for dental implant applications.

Heat treatment techniques may enhance crystallinity and reduce residual stress and shrinkage distortion, hence improving the mechanical performance of Polyetheretherketone components. Thus, the Polyetheretherketone composite may be tailored using carbon fibers to closely emulate the characteristics of human cortical bone [30] .

Table 2 Surface Modification of POLYETHERETHERKETONE [3]

Surface Modification	Processes	Material	References
Coating Techniques	Plasma deposition	Titanium alloy (Ti6Al4V); Hydroxyapatite (HA-p)	Rust-Dawicki 1995; Ha, 1994 Suska, 2014;
Surface conversion	Sandblasting	Aluminium oxide (Al_2O_3), TiO_2	Xu, 2015; Suska, 2014;

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Integration of bioactive properties	Bioactive inorganic substances	nano-fluorohydroxyapatite (n- FHA), Nano-TiO_2(n-TiO_2);	Wang, 2014 ;
Chemical Alteration	Sulphonation	Sulfonate groups (–SO_3–)	Yee, 2013

2. Methodology

2.1. Implant material

The decision to use specific materials for dental implants is dictated by their mechanical characteristics and biological compatibility. Dental implants withstand significant mechanical forces during mastication and fixation. Reports indicate that the apex biting force in the molar region is around 750-765 N, whereas the average biting force ranges from 75 to 155 N, and also varies on the tooth's position upon the mandible. It was observed that in most instances, these forces do not exceeds 100 N [47], [48]. Titanium alloy and Polyetheretherketone have been chosen for evaluating optimal structures that have been granted clinical approval owing to their superior bio compatibility, high tensile strength.

Table 3 The tensile strength and elastic moduli of different materials [49], [50]

MATERIAL	Tensile Strength (MPa)	Young's Modulus (GPa)
Polyetheretherketone	80-85	3-5
Cortical Bone	103-120	15
Dentin	110	18-20
Titanium alloy (Ti6Al4V)	954-976	105-110
Polymethylmethacrylate (PMMA)	48-76	3-4
Enamel	47.5	40-85
Zirconia (ZrO_2)	77-85	210

2.2 Metal alloys

They possess biomechanical attributes that make them exceptionally suitable for implantation. Furthermore, the metallic material possesses a sleek feel and displays an extraordinary level of luster. Metal implants may be disinfected using conventional sterilization techniques, hence facilitating their application. Advancements in technology and decreasing metal's pricing such as stainless steel, gold, and cobalt-chromium alloy have rendered these materials obsolete, resulting in their substitution with newer alternatives. Titanium alloy (Ti) and its alloys,

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particularly Ti6Al4V [51], are the preferred option for dental implants. Nonetheless, gold alloys, nickel-chromium alloys, stainless steel, cobalt-chromium, remain employed in the fabrication of implant prosthetic components.

Table 4 Experimental values of mechanical characteristics for titanium alloy

Titanium Alloy Properties	Value
Compressive Yield Strength	920 MPa
Density	4.6e-0066 kg/mm^3
Bulk Modulus	1.1429e+005 MPa
Coefficient of thermal expansion	9.4e-006/degC
Specific Heat (C_p)	5.22e+0055 mJ/kg-degC
Ultimate tensile strength	1077 MPa
Thermal Conductivity	2.198e-002 W/mm-degC
Resistivity	1.7e-0029 ohm-mm
Shear Modulus	35.3 GPa
Poisson's Ratio	0.359
Melting Temperature	22 deg C

Table 5 Experimental values of the mechanical characteristics of Polyetheretherketone material

Polyetheretherketone's Properties	Value
Density	1.23e-29 kg/m^3
Compressive Strength	135 MPa
Tensile Yield Strength	65 MPa
Shear Modulus	1.35 GPa
Poisson's Ratio	0.378
Melting Temperature	324 deg C
Ultimate Tensile Strength	115 MPa
Bulk Modulus	5.80 GPa
Thermal Conductivity	0.239W/m-K
Resistivity	3.3e+0.129 ohm.m

2.3 Modeling and Analysis

The objective of creating novel tooth abutment implant models was to enhance the existing implant design. The modeling is performed using ANSYS Software version 19.2.

-Workflow for finite element analysis
-Preprocessing involves defining the geometric domain of the problem.
-Elucidate the material element's properties.
-Choosing the type(s) of element to be employed.
-Detail the geometric attributes of the components (dimensions).

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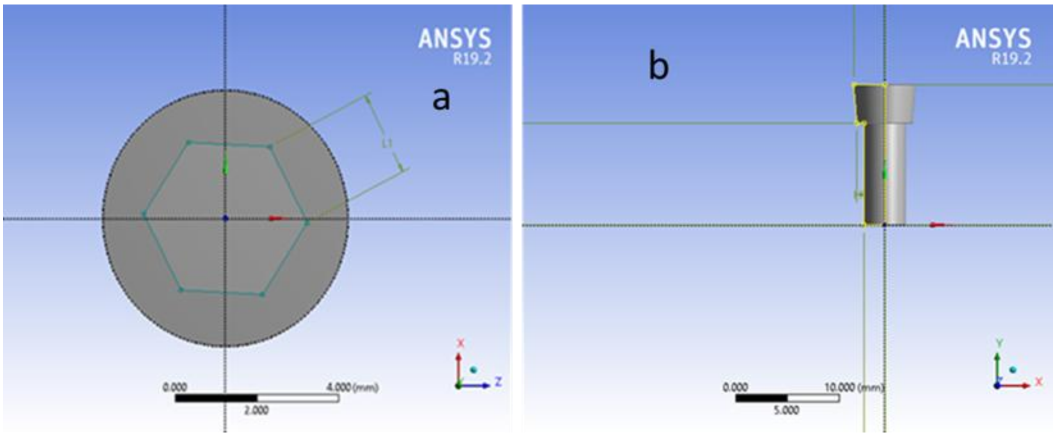
- Please provide details about the correlations among the components in the model (mesh).
-Assemble a compendium of physical restrictions (boundary conditions).
-Determine the weight or loading that must be supported
The solution identifies the unknown values (s) of the primary field variable. Thereafter, back substitution is employed to calculate supplementary variables, such as Von-Mises stress and primary stress.

Case 1: The implant is devoid of threads, enhancing the contact surface area and strengthening the bone-implant interface. The upper portion of the implant is tapered and features an internal hex system, which reduces issues such screw loosening and fractures, hence assuring a more solid fit. The decision to adopt the internal hex system was driven by the aim of reducing crestal bone loss. Previous studies demonstrated that after five years of employing an external hex system, there was a 95% survival rate with merely 1mm of crestal bone loss. After ten years, the survival rate was roughly 90-95%, accompanied by 1-2mm of marginal bone resorption[52][53].

The geometry of the implant is a tapered cylindrical shape devoid of threads. The implant measures 13.5 mm in length, while the abutment is 6 mm long and is designed to be secured to the implant by a screw mechanism. The implant has a minimum diameter of 3.75 mm.



Case 2 : The second variant includes a square thread implant that provides improved resistance to shear forces. This signifies that a heightened compressive force is transmitted during contact, leading to an enhanced bone-implant interaction. The upper segment is a conical bar featuring an interior hexagonal structure that reduces mechanical problems such screw loosening and fracture, hence improving friction fit.

The device comprises a cylindrical implant featuring a square thread [fig3.5(a), (b)]. The implant measures 13.5 mm in length and 3.75 mm in lower breadth. An abutment can be attached to this implant by a screw mechanism.

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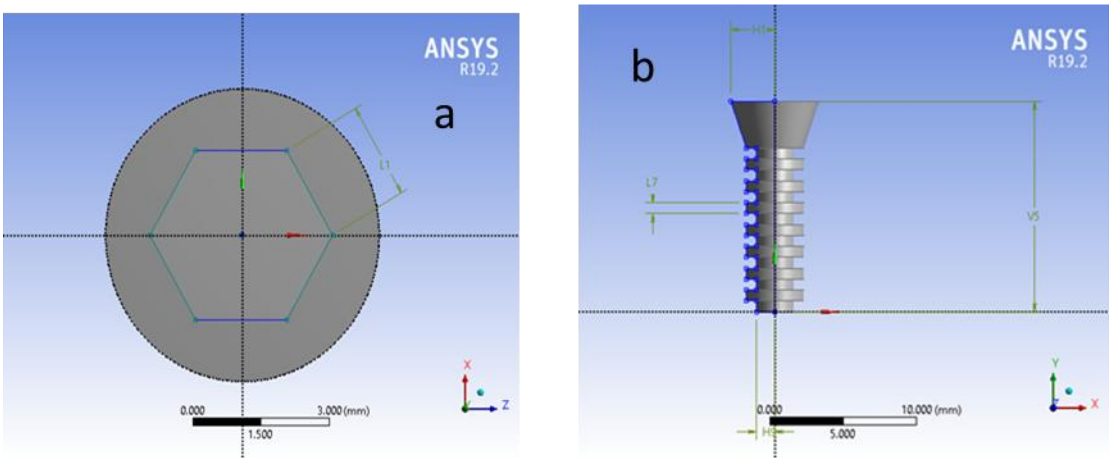

3. RESULTS AND DISCUSSIONS

3.1 The Constraints

In all instances, implants exhibit consistent mechanical properties, linear and isotropic elasticity, and are constructed from Titanium alloy and Polyetheretherketone in every design. Models are confined to all the three axes X, Y, and Z on the implant's surface.

3.2 Loading

The median masticatory force varies from 70 to 150 N, but the greatest force exerted during chewing and biting is between 760 and 800 N. The forces applied may be vertical, horizontal, or oblique, requiring the use of diverse loadings in various combinations to determine the stress variation on the implant system for each design. First, transmit a force of 100 N to assess stress under standard conditions, subsequently applying a lateral loading of 40 N. Subsequently, during the mastication process, analyze three unique loading scenarios: a minimum of 100 N vertical and 40 N lateral loadings acting simultaneously, the maximum biting force of 760 N vertical and 40 N lateral loadings, and an intermediate loading of 300 N with 40 N of lateral loading.

***3.3* Examination of Case1:** The dental implant is designed with no threads, thereby offers an augmented surface area of contact and improving bone-implant connection (BIC). The upper segment has a tapered bar with an internal hexagonal mechanism that reduces the mechanical dangers of prevention of screw slippage and fractures, hence ensuring a more secure friction fit.

In Case 1, the inferior surface area remains constant when subjected to various stresses, both normal and lateral, as well as in combination.

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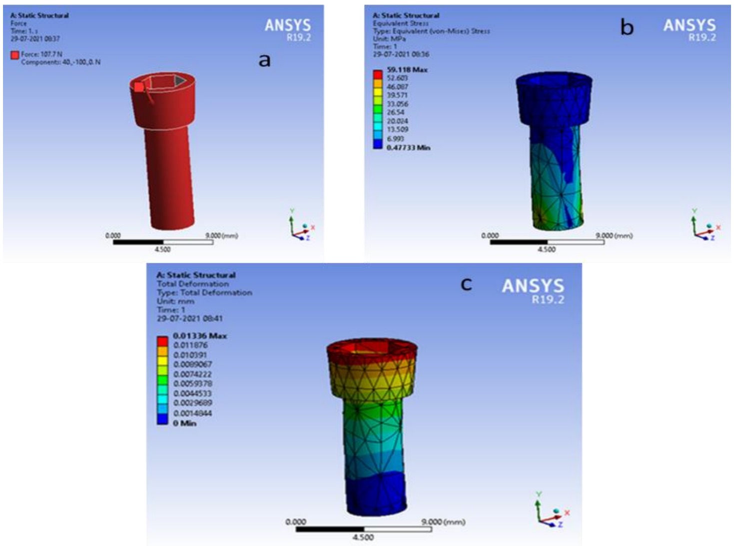

Figure 4.1(a) Force distribution , (b) Von Mises variation, (c) deformation

The maximum Von Mises stress variation for the non-thread titanium-based implant in case 1, subjected to a vertical loading of 100 N and a lateral loading of 40 N on the upper surface, is 59.12 MPa. The Net Deformation reaches a maximum of 0.0134mm at the upper surface, with 806 elements and 1625 nodes involved.

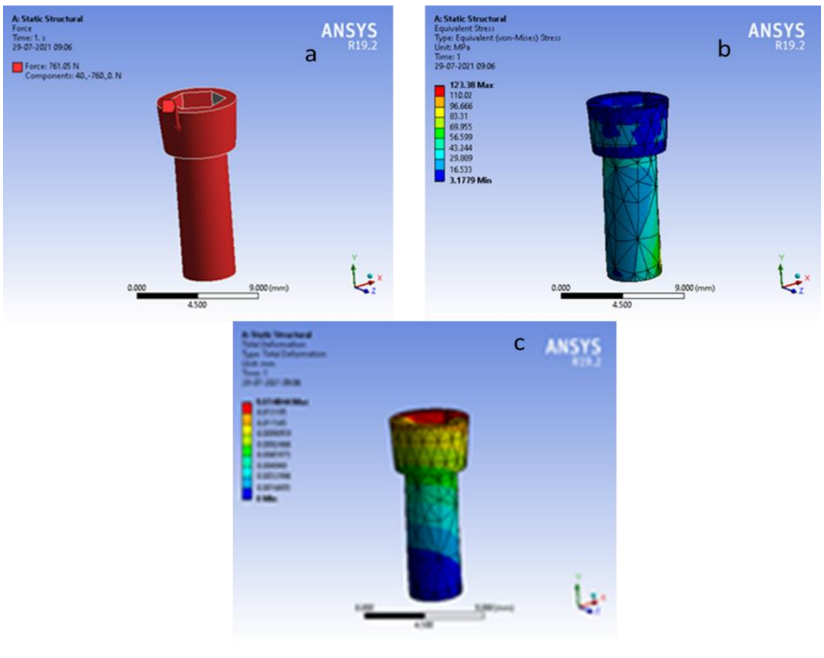

Figure 4.2(a) Force distribution (b) Von Mises stress variation (c) Net Deformation

The maximum stress calculated using the Von Mises variationfor the no thread titanium-based implant in case 1, subjected to a vertical force of 760 N and a lateral loading of 40 N on the upper surface, is 123.40MPa. The maximum Net Deformation at the upper surface is 0.01485mm, with 806 elements and 1625 nodes involved.

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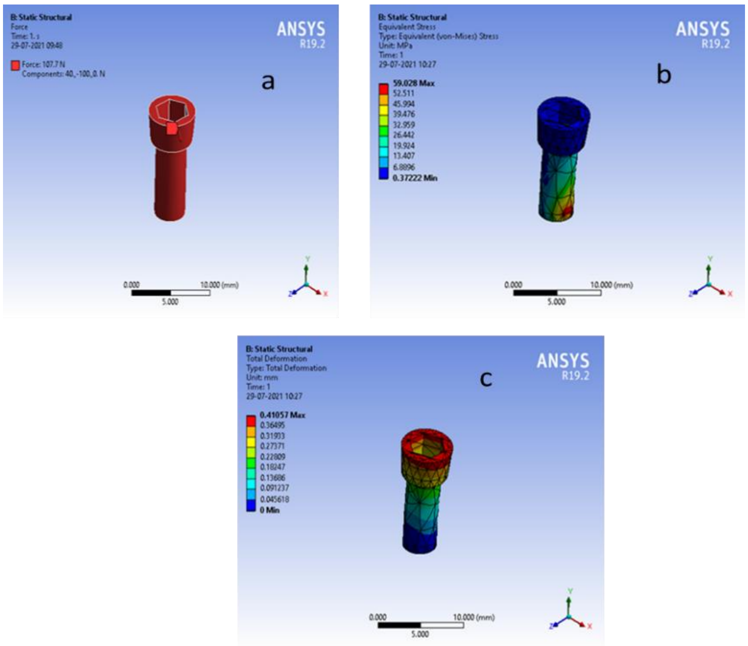

Figure 4.3(a) Force's orientation, (b) Von Mises stress variation (c) Deformation

The maximum Von Mises stress variation for the Polyetheretherketone -based no thread implant in case 1, subjected to a vertical loading of 100 N and a lateral loading of 40 N on the upper surface, is 59.03 MPa. The overall deformation reaches a maximum of 0.4106 mm at the upper surface, with a total of 1625 nodes and 806 components involved in the analysis.

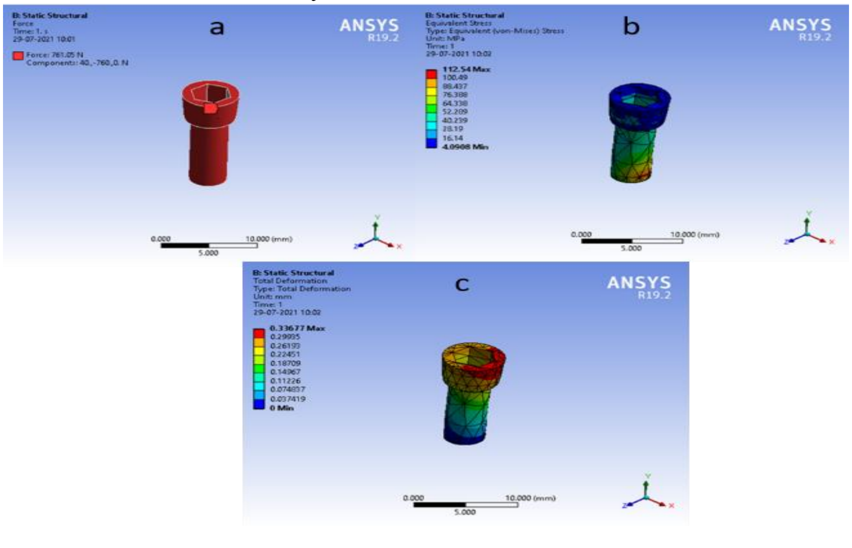

Figure 4.4(a) Force's orientation (b) Von Mises stress variation (c) Net Deformation

The maximum stress calculated using the Von Mises variation for the Polyetheretherketone-based non-thread implant in case 1, subjected to a vertical loading of 760 N and a lateral loading of 40 N on the upper surface, is 112.55 MPa. The Net Deformation reaches a maximum of 0.3368 mm at the upper surface, with a total of 1625 nodes and 806 elements utilized in the analysis.

Case 2: The implant incorporates a square-thread configuration, facilitating superior shear force and augmented compressive force transmission at the contact interface,

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hence enhancing bone-implant connection (BIC). The friction fit is enhanced by the presence of a tapered bar with an internal hexagonal system in the upper section, which mitigates the mechanical hazards of screw loosening and fracturing.
Geometry: It is composed of a cylindrical square-thread implant. The implant is 13.5 mm in length, has a lower diameter of 3.75 mm, and is equipped with a 6 mm abutment fastener.
This design incorporates a constant lower surface area, with changing loadings applied both perpendicularly and laterally, as well as in conjunction.

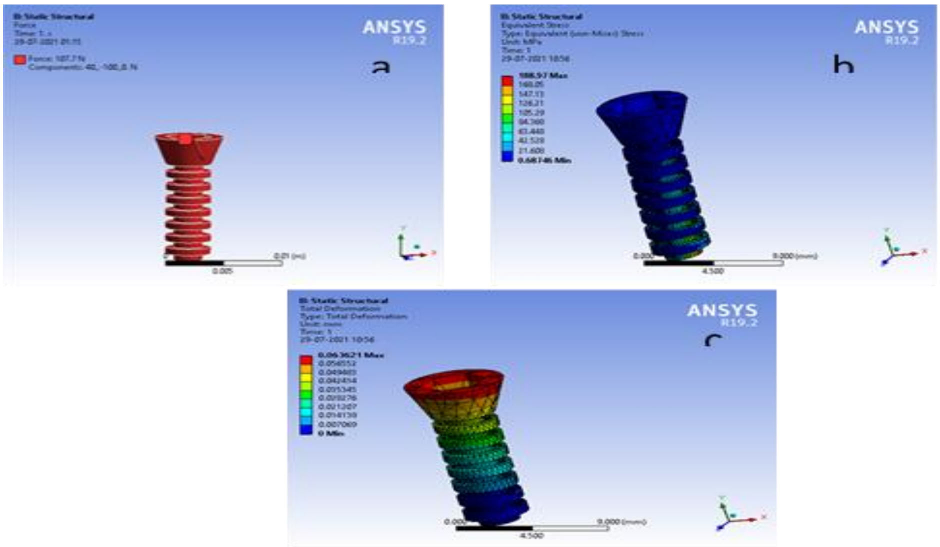

Figure 4.5(a) Force's Orientation, (b) Von Mises stress variation (c) Net Deformation

The result for the titanium-based implant in case 2, characterized by a square-thread configuration, exhibits a maximum Von Mises stress variation of 188.97 MPa under a vertical loading of 100 N and a lateral loading of 40 N on the upper surface. The Net Deformation reaches a maximum of 0.06362 mm at the upper surface, utilizing 4502 elements and 8205 nodes.

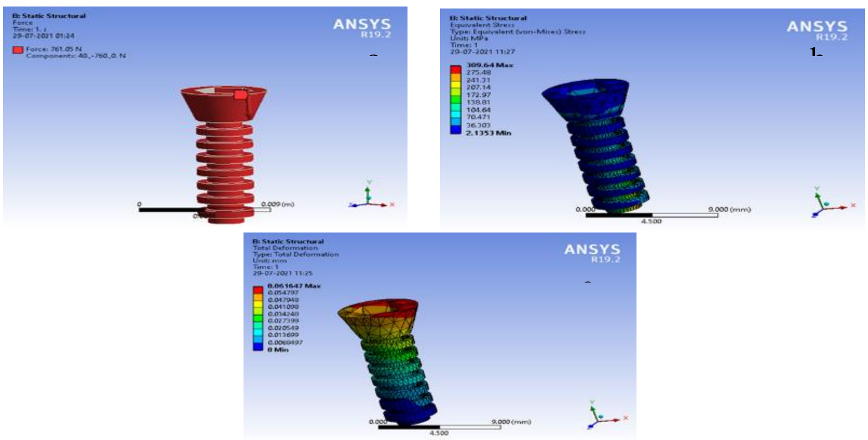

Figure 4.6 (a) Force's Orientation, (b) Von Mises stress variation (c) Net Deformation

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The maximum Von Mises stress variation for the titanium-based square-thread implant in case 2, subjected to a vertical loading of 760 N and a lateral loading of 40 N on the upper surface, is 309.65 MPa. The Net Deformation reaches a maximum of 0.0617 mm at the upper surface, with a total of 4502 elements and 8205 nodes utilized in the analysis.

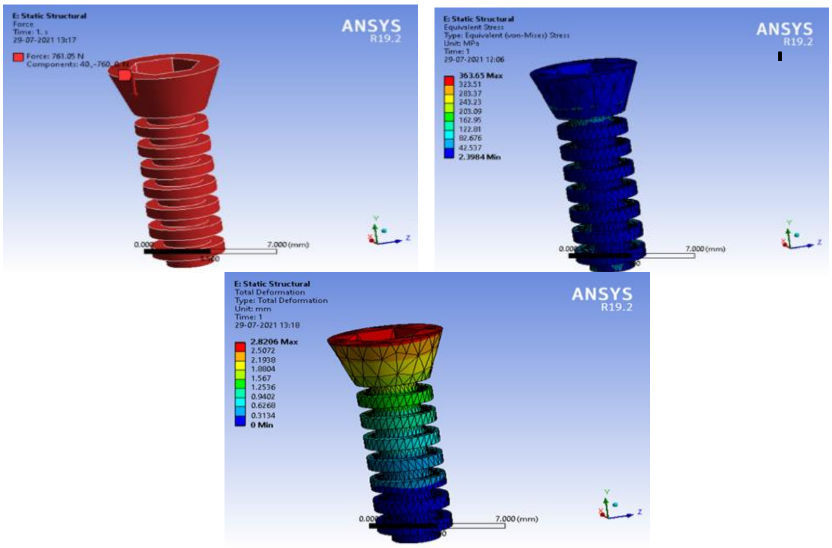

Figure 4.7 (a) Force's Orientation, (b) Von Mises stress variation (c) Net Deformation

The maximum stress from the Von Mises variation for the Polyetheretherketone-based square-thread implant in case 2, subjected to a vertical loading of 760 N on the upper surface, is 363.66 MPa. The overall deformation reaches a maximum of 2.821 mm at the upper surface, with a total of 4502 elements and 8205 nodes utilized in the analysis.

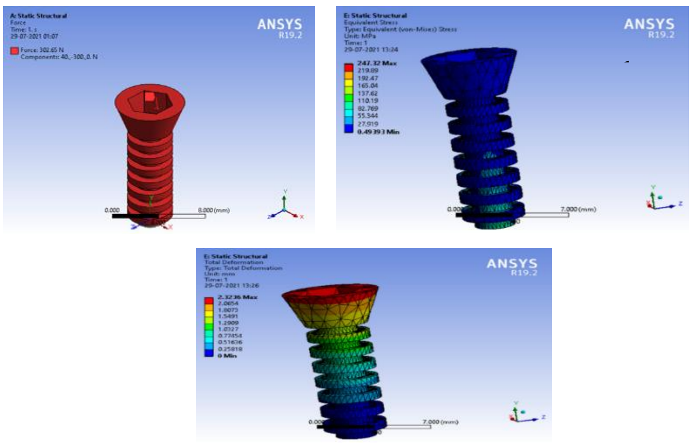

Figure 4.8(a) Force's Orientation, (b) Von Mises stress variation (c) Net Deformation

The maximum Von Mises stress variation for the Polyetheretherketone-based square-thread implant in case 2, subjected to a vertical loading of 300 N on the upper surface, is 247.40 MPa. The overall deformation reaches a maximum of 2.33 mm at the upper surface, utilizing 4502 elements and 8205 nodes.

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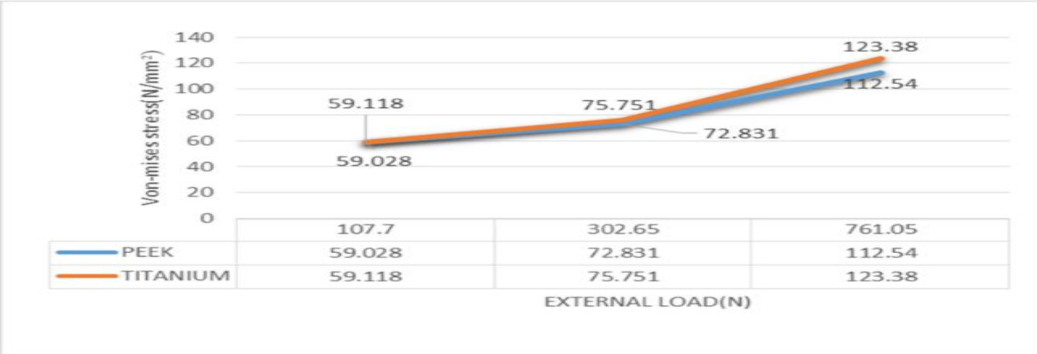

Figure 4.9. Graph depicting the correlation between Von-Mises Stress and externally applied loading (N) for Case 1.

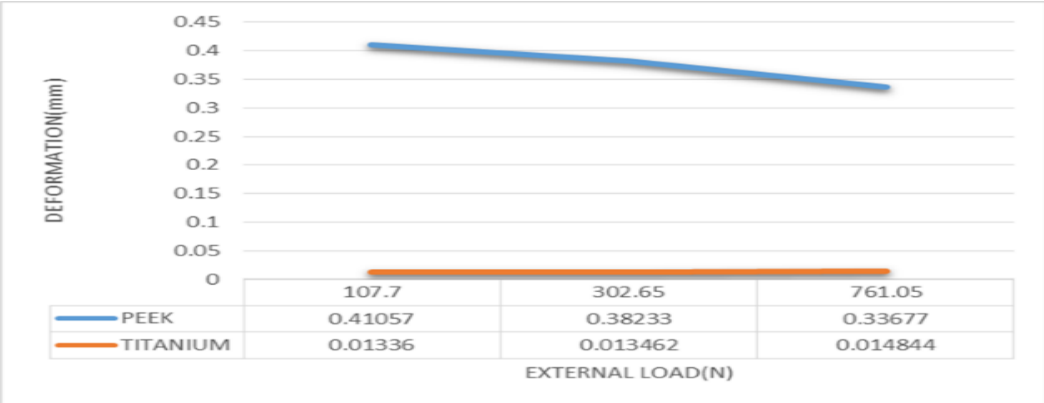

Figure4.10. Graph depicting the correlation between Deformation (mm) and externally applied loading (N) for Case 1.

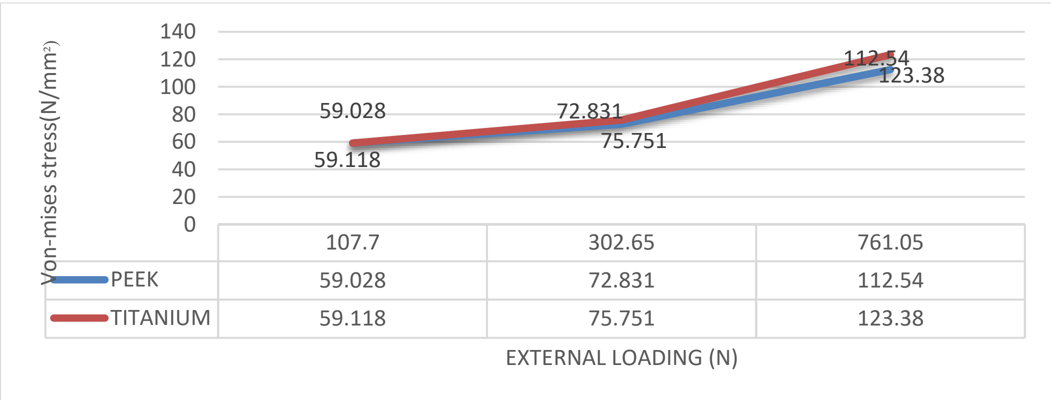

Fig 4.11. Graph depicting the correlation between Von-Mises Stress (N/mm²) and externally applied loading (N) for Case 2.

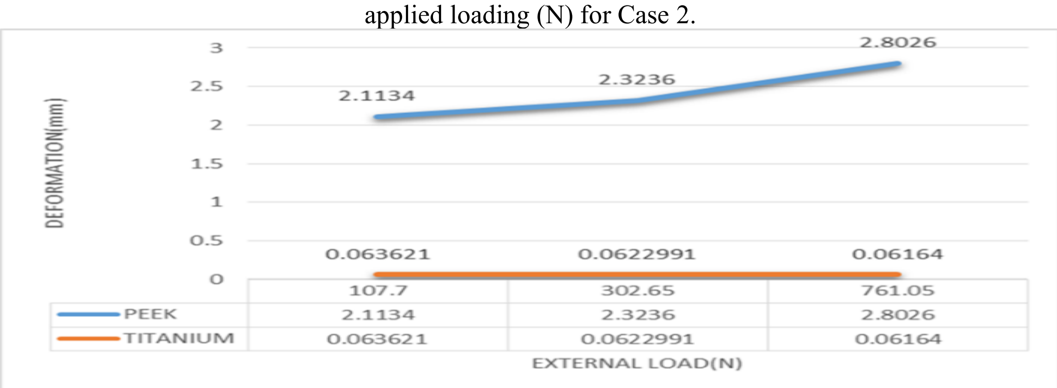

Figure 4.12. Graph depicting the correlation between Deformation (mm) and externally applied Loading (N) for Case 2.

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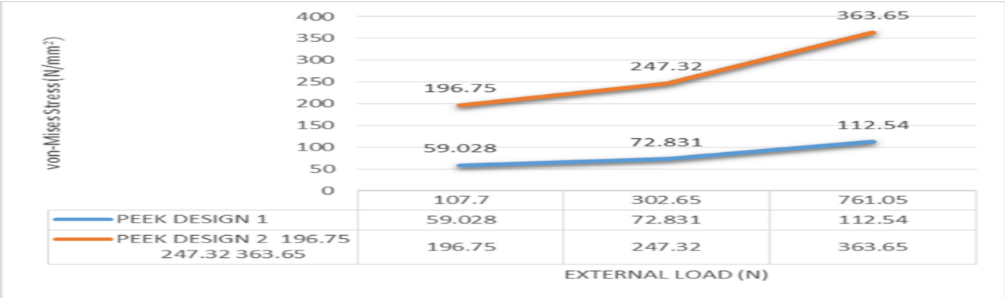

Figure 4.13 Graph illustrating the correlation between Von Mises stress (N/mm²) and externally applied loading (N) for Polyetheretherketone material, highlighting both cases.

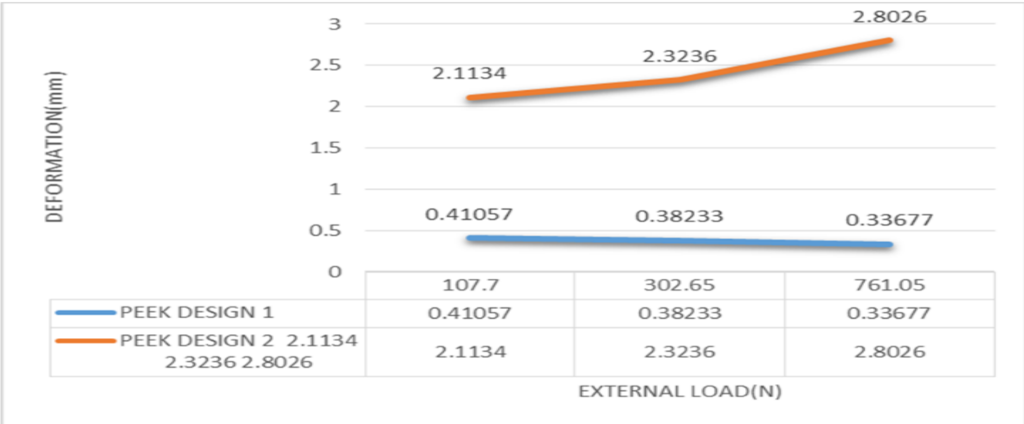

Figure 4.14 Graph illustrating the correlation between deformation (mm) and externally applied loading (N) for Polyetheretherketone material, encompassing both the cases parameters.

4. Conclusions and Future Scope

The comparative investigation of titanium alloy and Polyetheretherketone (PEEK) dental implants demonstrates that material selection plays a critical role in achieving long-term bio-mechanical stability and osseointegration. PEEK exhibits mechanical and biological properties that closely replicate those of natural bone, with an elastic modulus of 3-4 GPa, significantly reducing stress shielding effects observed in titanium-based implants (110 GPa). Finite element analysis (FEA) under various masticatory loads (100-760 N vertical and 40 N lateral) revealed that PEEK implants achieve a more uniform stress distribution across the implant–bone interface and higher deformation compatibility (≈0.34 mm), allowing better load sharing and energy absorption. Conversely, titanium implants, while mechanically stronger, exhibited localized stress concentrations that may contribute to bone resorption over prolonged use.

Among the evaluated geometries, square-threaded implants demonstrated superior stress transfer and stability under high occlusal forces compared to smooth designs. The internal hexagonal connection further minimized micromovements and mechanical failures. These results affirm PEEK's potential as a next-generation implant material, combining favorable mechanical behavior, radiolucency, and biocompatibility. Future work should focus on surface modification techniques such

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as plasma treatment and hydroxyapatite coating to enhance bioactivity and clinical integration. Long-term in vivo studies are also essential to validate the material's performance in complex oral environments.

The present study is limited to computational modeling and finite element analysis, which, although effective for predicting stress distribution and deformation, cannot fully replicate the complex biological responses occurring at the bone implant interface. Experimental validation under dynamic loading and long-term in vivo conditions is required to confirm the simulated outcomes. Furthermore, variations in bone density, patient-specific anatomy, and surface treatment effects were not incorporated, which may influence the overall clinical performance of PEEK implants.

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