

Article

Finite Element Analysis of Prosthetic Finger Connectors: Comparison of Ti-6Al-4V and PEEK-Based Polymers Using COMSOL

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Abstract

Introduction/Objectives: The mechanical reliability of prosthetic finger connectors is critical for daily prosthesis use, as these small load-bearing elements transmit forces between finger segments and are prone to localized stress concentration at hinge transitions. The objective of this study was to quantitatively evaluate the mechanical performance of prosthetic finger connectors and determine how material selection influences stress distribution and deformation behavior when connector geometry is maintained constant. **Methods:** A controlled comparative finite element analysis was performed using COMSOL Multiphysics 6.2 on a single prosthetic finger connector geometry manufactured from four candidate materials: titanium alloy Ti-6Al-4V ($E = 110$ GPa), polyetheretherketone (PEEK, $E = 4.0$ GPa), AvaSpire AV-621 NT ($E = 3.0$ GPa), and KetaSpire KT-820 FP ($E = 4.1$ GPa). Three representative loading scenarios were simulated using an axial grip force of 50 N and a bending moment of 0.5 N·m: (1) axial loading, (2) combined axial and forward bending, and (3) combined axial and reverse bending. **Results:** Across all loading conditions, peak von Mises stresses consistently localized at the hinge-transition region of the connector. Ti-6Al-4V demonstrated the lowest peak stress (approximately 2.2 MPa under combined forward bending) and the smallest maximum displacement (approximately 0.0005 mm). In contrast, the PEEK-family polymers showed greater deformation (0.0125–0.0167 mm) due to their lower elastic modulus, while maintaining comparable peak stresses of approximately 2.5–2.6 MPa. The stress profiles along the connector arc length were nearly identical among all materials, indicating that connector geometry primarily governs stress localization, whereas material stiffness mainly determines deformation magnitude. **Conclusions:** This study demonstrates that material selection strongly affects prosthetic finger connector compliance, while stress concentration behavior is mainly controlled by geometry. The obtained quantitative comparison provides design-oriented guidance for selecting between Ti-6Al-4V and PEEK-based polymers for the development of lightweight and mechanically reliable prosthetic finger connectors.

Keywords: prosthetics; KetaSpire; AvaSpire; Ti-6Al-4V; PEEK; COMSOL Multiphysics; finite element analysis



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1. Introduction

Prosthetic finger mechanisms experience complex loading during daily use, including axial compression, bending moments, and torsional forces generated during grasping and manipulation. These loading modes apply substantial stresses on connector elements linking distal and proximal finger segments. Previous studies on artificial hands and upper-limb prosthetics have demonstrated that geometric discontinuities such as hinge transitions, sharp fillets, and thin connector cross-sections act as critical stress risers and common initiation sites for fatigue-related failures [1–4]. Similar behavior has been reported in orthopedic structures and splinting systems, where local topology significantly alters stress and structural durability [5,6]. Ensuring the mechanical reliability of these connector regions is therefore essential for functional stability, user safety, and long-term prosthesis performance.

Within the overall prosthesis architecture, the finger connector is the small load-bearing element that joins two adjacent phalangeal segments and transmits forces between them. Functionally, it serves three coupled mechanical roles: (i) it transfers axial grip forces from the distal to the proximal segment during pinching and grasping; (ii) it carries bending and shear loads generated when the finger contacts an object off-axis or at an oblique angle; and (iii) it acts as the rotational hinge that defines the kinematic articulation of the joint. Because these load paths are funneled through a small cross-section with abrupt geometric transitions, the connector is also the principal site of stress concentration and fatigue crack initiation in the assembly. Its design therefore directly governs the durability, stiffness, and load-distribution behavior of the prosthetic finger, and a poorly designed or under-specified connector can compromise the integration of the distal tip, intermediate phalanx, and actuation tendon, even if every other component is mechanically sound.

Material selection also plays a crucial role in determining the structural behavior of prosthetic components. Titanium alloy Ti-6Al-4V remains the gold standard for load-bearing biomedical applications due to its high stiffness, corrosion resistance, and excellent fatigue performance, as has been shown in several orthopedic and biomechanical studies [7–9]. Ti-6Al-4V is a Grade 5 $\alpha + \beta$ titanium alloy whose nominal composition is approximately 90 wt% Ti, 6 wt% Al, and 4 wt% V, with controlled impurity limits on Fe, O, C, N, and H. Aluminum acts as an α -stabilizer that strengthens the hexagonal close-packed (HCP) α phase, while vanadium is a β -stabilizer that retains a body-centered cubic (BCC) β phase at room temperature. The resulting two-phase microstructure (lamellar, equiaxed, or bimodal depending on heat treatment) gives the alloy its characteristic combination of high specific strength (yield strength, typically 800–900 MPa; ultimate tensile strength, 900–1100 MPa), low density (4.43 g/cm³), an elastic modulus of approximately 110–114 GPa, excellent corrosion resistance through a stable TiO₂ passive layer, and well-established biocompatibility. These properties make Ti-6Al-4V the standard against which polymeric prosthetic materials are benchmarked and motivate its inclusion as the reference material in the present comparative study [5,7–9]. However, titanium's high elastic modulus frequently induces stress shielding, shifting physiological loads away from surrounding biological structures.

To address this limitation, alternative materials and composite systems have been studied. Finite element analysis (FEA) showed that a femoral stem with a CFRPC (Carbon Fiber-Reinforced Polymer Composite) distal component produced a more uniform stress distribution in the femur compared to a fully Ti-6Al-4V stem. The CFRC configuration reduced stress shielding and increasing proximal bone loading (throughout this manuscript, the term “stem” refers to the elongated load-bearing component of an orthopedic implant commonly inserted into the medullary canal of a long bone, as described in previously cited FEA studies; however, the present work does not investigate stem geometries but instead

focuses on a distinct prosthetic finger connector joining two phalangeal segments). The maximum von Mises stress decreased from about 725 MPa (Ti-6Al-4V) to 236 MPa (CFRPC configuration), indicating improved load transfer and lower stress concentration. However, limitations include potential issues with implant fixation, modular interface complications, and simplifications in the numerical model such as isotropic bone properties and static loading assumptions [5].

Alternatively, depending on the design configuration and loading conditions, titanium-based frameworks might still exhibit superior mechanical performance. Finite element analysis showed that titanium frameworks provided the most favorable biomechanical behavior, transferring the least stress to bone, cement, and screws while maintaining structural stability. In contrast, modified polyetheretherketone (PEEK) and PEEK produced higher stress levels, especially under incisor loading, which may increase the risk of fatigue or debonding. Overall, titanium demonstrated the best stress distribution and mechanical reliability among the tested materials [10].

Finite element analysis is widely employed to study stress distribution, deformation, and potential failure in prosthetic and orthopedic systems. FEA comparing titanium alloys and PEEK-based materials for mandibular defect reconstruction showed that PEEK systems generated higher stresses in the mandible and fibula, sometimes exceeding bone yield strength, while titanium alloys maintained stresses within safe limits. However, carbon fiber-reinforced PEEK (CFR-PEEK) demonstrated improved stress distribution and lower bone stress compared with other PEEK variants, making it the most promising polymer-based option. Overall, titanium provides high mechanical stability but may cause stress shielding, whereas reinforced PEEK offers better modulus matching with bone but may have mechanical limitations [11]. These findings further highlight the complexity between mechanical stability and physiological load transfer when selecting biomaterials.

Finite element analysis showed that PEEK and Polylactic Acid (PLA) composite femoral implants produced a more physiological load transfer and reduced stress shielding compared with traditional titanium stems. The composites distributed stress more uniformly across the bone–implant interface, with a higher percentage of bone nodes experiencing physiologically relevant stress levels (~8.9–9.2% for PEEK/PLA vs. ~6% for titanium), suggesting improved bone stimulation. However, while titanium provides higher stiffness and structural reliability, polymer composites may introduce higher interface stresses and require further experimental validation [12]. Similarly, these results emphasize that improved load transfer does not necessarily guarantee superior mechanical reliability.

High-performance polymers such as PEEK and its engineered derivatives, including KetaSpire and AvaSpire, have expanded material options for lightweight prosthetic devices. PEEK is a semi-crystalline aromatic thermoplastic of the polyaryletherketone (PAEK) family, composed of repeating ether and ketone linkages between phenyl rings. Its rigid aromatic backbone gives it a high glass transition temperature ($T_g \approx 143^\circ\text{C}$); a melting temperature near 343°C ; and exceptional chemical, hydrolytic, and radiation resistance, while the partially crystalline morphology provides toughness and dimensional stability. Mechanically, neat PEEK has a Young's modulus of approximately 3.6–4.1 GPa, a tensile strength of 90–100 MPa, a density of about 1.30 g/cm^3 , and well-established biocompatibility, which together make it the leading metal-replacement polymer in load-bearing biomedical and aerospace components. KetaSpire (KT-820 FP, Solvay, Bruxelles, Belgium) is a high-performance industrial PEEK grade formulated for friction- and wear-bearing applications; it retains the chemical and thermal stability of PEEK but is engineered with a slightly higher stiffness ($E \approx 4.1\text{ GPa}$) and improved tribological behavior, which is directly relevant to the cyclic contact loading at a finger hinge. AvaSpire (AV-621 NT, Solvay) is a PAEK copolymer derived from the same chemistry as PEEK but with a modified backbone

that lowers the elastic modulus to approximately 3.0 GPa, increases toughness, and reduces the melt-processing temperature, making it easier to mold or 3D-print into thin-walled prosthetic components. Including these three polymers alongside Ti-6Al-4V therefore spans a representative stiffness range (3.0–4.1 GPa) of clinically relevant PEEK-family materials and allows the present study to evaluate, within a single identical connector geometry, how subtle differences in polymer modulus translate into differences in stress and deformation. These polymers combine favorable strength-to-weight ratios, transparency, manufacturability, and biocompatibility. Therefore, they have been successfully applied in cranial reconstruction, orthopedic fixation, and upper-limb splinting [10–12]. Finite element simulations further show that polymeric implants, particularly PEEK and its composites, provide a more physiological load transfer, maintaining bone stresses within the optimal 2–3 MPa range and reducing stress shielding compared with Ti-6Al-4V [5,10,13]. Nonetheless, the considerably lower elastic modulus of PEEK-family materials (3–18 GPa vs. 110 GPa for Ti-6Al-4V) results in increased deformation under load, thus raising concerns about their suitability for small and high stress connector regions. Taken together, these studies suggest that polymers improve biomechanical compatibility, and their structural limitations must be carefully considered in load-carrying applications.

Investigations of prosthetic hands, compliant mechanisms, and Shape Memory Alloy (SMA) actuated finger models demonstrate that bending moments dominate stress patterns and that even small geometric deviations strongly influence mechanical behavior [1,3,4,14]. Other research on cranial implants and splint geometries emphasizes the dependence of local stress gradients on material properties, boundary conditions, and load directions [6,11]. However, direct material-level comparisons within identical connector geometries remain a methodological challenge. This difficulty arises because polymers and metals do not deform or accumulate damage in comparable ways: polymers exhibit strain-rate sensitivity [15,16], viscoelasticity [17], localized shear banding [18], and time-dependent deformation [17]. Meanwhile titanium alloys demonstrate linear elasticity followed by predefined yield and hardening behavior [19,20]. These fundamental differences create a non-uniform stress transfer; distinct fatigue mechanisms; and different sensitivities to combined bending, torsion, and axial compression. Moreover, identical meshing strategies do not always converge equivalently for polymers and metals due to their separate deformation magnitudes, and common stress criteria such as von Mises may not fully capture polymer failure modes. As a result, the mechanical response of high-performance polymers versus titanium alloys under multiaxial bending has not been systematically evaluated within a controlled, geometry-invariant framework.

To address these gaps, the present study uses COMSOL Multiphysics 6.2 to compare the structural behavior of four candidate materials, KetaSpire, AvaSpire, PEEK, and Ti-6Al-4V, under three physiologically relevant loading cases involving axial loading and forward/reverse bending. The relevance of this comparison is anchored in the fact that prosthetic finger connectors are the smallest and most highly loaded structural element in an upper-limb prosthesis: they concentrate axial grip forces and bending moments into a millimeter-scale hinge cross-section, and their durability directly determines the service life of the device for the user. The specific research objective is therefore to quantify, within a single identical connector geometry, how the choice of material (Ti-6Al-4V vs. PEEK, KetaSpire, and AvaSpire) modifies the magnitude and spatial distribution of von Mises stress and the maximum displacement under realistic axial and bending load cases and to determine whether geometry or material is the dominant driver of stress localization. By holding geometry, mesh, and boundary conditions constant across all four materials, the design and material-selection contribution of the connector to durability and functionality under practical loading conditions can be isolated and discussed on a like-for-like basis.

The primary contribution of this work is the implementation of a controlled comparative framework in which identical connector geometry and boundary conditions are maintained, enabling the isolation of material-specific mechanical responses without geometric bias. A second novelty lies in the systematic evaluation of stress distribution, deformation behavior, and sensitivity to bending direction, allowing for the identification of how different materials respond to combined loading conditions representative of finger manipulation. Finally, the study provides design-oriented insights into evidence-based material selection in prosthetic finger connectors, supporting the development of next-generation prosthetic components. Therefore, this research will contribute to emerging strategies in polymeric implants, compliant mechanisms, and material-assisted biomedical design [21].

2. Materials and Methods

2.1. Geometry and Model Assembly

The three-dimensional geometry of the prosthetic finger connector with proper meshing (a physics-controlled mesh with a normal element size) was imported and finalized in COMSOL Multiphysics 6.2 using the Form Union operation to ensure mechanical continuity between adjacent domains (Figure 1A). COMSOL Multiphysics 6.2. is a general-purpose finite element analysis (FEA) and multiphysics simulation software (COMSOL AB, Stockholm, Sweden) that solves systems of partial differential equations describing coupled physical phenomena (in this work, the solid mechanics and fatigue modules) on a discretized geometry. The connector CAD geometry itself was prepared upstream in SolidWorks 2023 (Dassault Systèmes, Vélizy-Villacoublay, France) and exported as a STEP (.stp) file before being imported into COMSOL. The nominal overall dimensions of the modelled connector are approximately 28 mm in total length, with a hinge section measuring 6.0 mm in width and 4.0 mm in thickness, a hinge pin bore of $\varnothing$ 2.5 mm, and proximal and distal coupling stems of approximately 8 mm in length and 5 mm in diameter; these dimensions were reproduced from the fabricated prototype shown in Figure 1B,C and were kept identical for every material so that geometry was not a confounding variable. The assembled model consisted of two solid objects forming three domains. Automatic repair tolerance was applied to eliminate geometric inconsistencies without altering dimensional accuracy. This procedure ensured continuous stress transfer across interfaces prior to meshing and structural analysis. Photographs of the fabricated prosthetic finger connector and its size relative to a human finger are presented in Figure 1B,C.

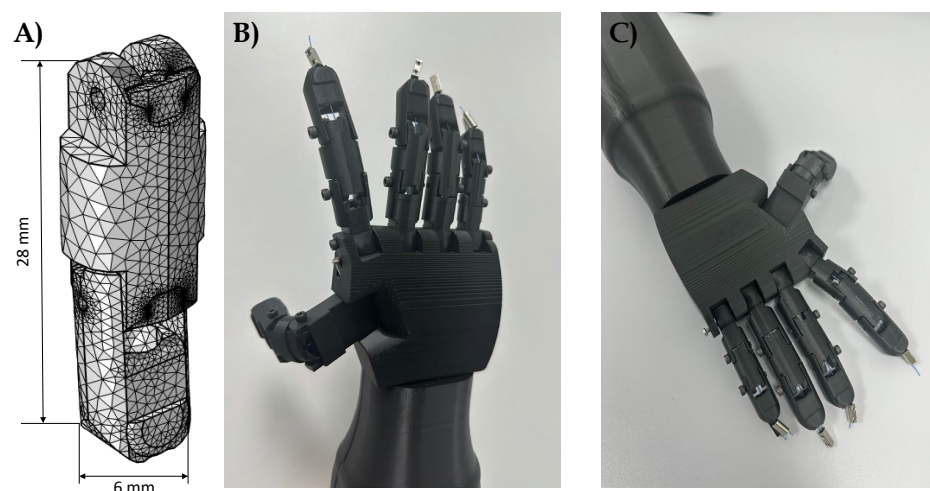


Figure 1. (A) Finite element mesh of the prosthetic finger connector generated in COMSOL Multiphysics, showing the tetrahedral discretization used for structural and fatigue analysis. (B) Photograph

of the hand prototype showing the integrated finger connectors in a functional configuration. (C) Top view of the fabricated prosthetic hand, highlighting the connector placement, joint structure, and overall geometry of the system.

2.2. Finite Element Discretization

The geometry was discretized using a physics-controlled tetrahedral mesh. Finite element discretization is the procedure by which a continuous solid domain is subdivided into a finite number of small, non-overlapping geometric sub-domains (elements) connected at nodes; the unknown displacement field is approximated within each element by a polynomial shape function, and the governing partial differential equations of solid mechanics are converted into a coupled algebraic system in the nodal displacements that can be solved on a computer. In this study, the connector volume was meshed with second-order (quadratic) tetrahedral elements, which interpolate displacements quadratically inside each element and allow for an accurate representation of curved boundaries such as the hinge fillet without an excessive element count. A Normal Element Size was selected, with both solid mechanics and fatigue physics activated as mesh contributors. This approach enables automatic refinement in regions of high stress gradients, particularly near geometric discontinuities and connector interfaces where fatigue crack initiation is expected. The resulting mesh provided adequate spatial resolution while maintaining computational efficiency (Figure 1A). To ensure that the reported stresses are not biased by element size, a mesh convergence study was carried out for all the models under Load Case 2 by re-running the simulation with the built-in COMSOL element-size presets: Coarser (≈ 6800 elements), Coarse ($\approx 12,400$ elements), Normal ($\approx 24,700$ elements), Fine ($\approx 48,100$ elements), and Finer ($\approx 92,300$ elements). The peak von Mises stress at the hinge transition was monitored as the convergence metric. The change in peak stress between successive refinement levels decreased monotonically and dropped below 5% when moving from Normal to Fine ($<3\%$ in our runs) and below 2% from Fine to Finer, indicating that the Normal Mesh already lies in the converged regime and that further refinement does not change the engineering conclusions of the study. The Normal Mesh was therefore retained for all four materials and all three load cases, providing a consistent compromise between accuracy and computational cost.

2.3. Boundary Conditions and Load Definition

A stationary structural analysis was performed under predefined loading scenarios. The proximal region of the connector was constrained using a Fixed Constraint boundary condition, imposing zero displacement ($u = 0$) to simulate attachment of the prosthetic connector to the adjacent structural segment and prevent rigid body motion. Specifically, the Fixed Constraint was applied to the entire cylindrical outer surface of the proximal coupling stem (length ≈ 8 mm; $\varnothing$, 5 mm; lateral-surface area ≈ 126 mm²), corresponding to the region that would be bonded into the adjacent phalangeal segment in the physical assembly. Nodes on this surface had all three translational degrees of freedom set to zero ($u_x = u_y = u_\phi = 0$); no rotational constraints were imposed because the underlying solid elements have only translational degrees of freedom. The remaining outer surfaces (hinge, fillets, and distal stem) were left unconstrained.

Load transfer was implemented through a rigid connector boundary condition applied at the load introduction region. The connector enforced rigid body kinematics according to

$$u = u_c + R \times (X - X_c) \quad (1)$$

where u_c is the displacement of the center of rotation, and R represents rotational components. X_c is the centroid of the selected region. The global coordinate system and automatic

center of rotation were used to ensure uniform displacement and realistic load transmission. In Equation (1), u is the displacement vector at an arbitrary material point, X , on the rigid-connector boundary; u_c is the translational displacement of the rigid-body reference point (the center of rotation, defined automatically by COMSOL as the centroid of the loaded surface); R is the infinitesimal rotation tensor of the rigid body about that center; X is the spatial position vector of the material point in the reference configuration; and X_c is the position vector of the center of rotation. Equation (1) therefore enforces that every node on the loaded boundary moves as a rigid body, so that the applied axial force and bending moment are transmitted uniformly into the connector without producing artificial local distortion.

The rigid connector boundary condition described above was applied specifically to the cylindrical outer surface of the distal coupling stem of the connector (length ≈ 8 mm; $\varnothing$, 5 mm; total loaded area ≈ 126 mm², calculated as πdL), with its automatic center of rotation located at the centroid of that surface. Two parameterized load groups were then defined within the solid mechanics interface and applied to that rigid connector (Figure 2). The load magnitudes were chosen to be physiologically representative of daily grasping activities of the human hand: an axial grip force of $F = 50$ N (within the typical range of 30–100 N reported for sub-maximal lateral and tip pinch in healthy adults and prosthesis users [3,4,12]) was applied through the rigid connector in Load Group 1, and a bending moment of $M = 0.5$ N·m (consistent with off-axis loading of the distal phalanx when contacting an object at a 10–15 mm lever arm from the hinge) was applied in Load Group 2. Both values are conservative upper-bound estimates for routine daily use and are kept identical for every material and load case. These quantities are applied force and moment boundary conditions; no pressure load was prescribed.

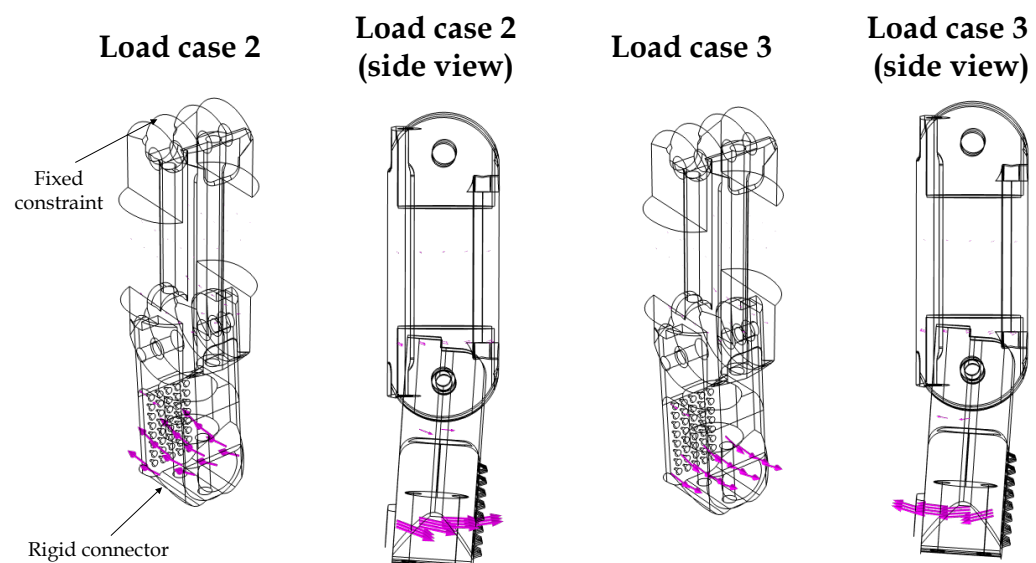


Figure 2. Schematic illustration of loading configuration applied to prosthetic finger connector. Loads were implemented using two load groups (LG): LG1 (applied axial force through a rigid connector) and LG2 (applied bending moment through the same connector). Three loading cases (LCs) were considered: LC1—axial force only; LC2—combined force and forward bending moment (LG1 + LG2); and LC3—combined force and reverse bending moment (LG1 + LG2 with the opposite sign). Arrows indicate the direction of the applied loads.

Load Group 1 (LG1): Axial force applied through the rigid connector, representing compression or tension during grasping.

Load Group 2 (LG2): Bending moment applied through the same rigid connector, simulating flexural loading during finger manipulation.

Three physiologically relevant load cases were considered:

- Axial loading (Load case 1): LG1 active (+1) and LG2 inactive;
- Forward bending (Load case 2): LG1 and LG2 active with the same sign (+1, +1);
- Reverse bending (Load case 3): LG1 active (+1) and LG2 applied with the opposite sign (+1, −1).

This parameterized configuration enabled systematic evaluation of stress responses under combined and reversed multiaxial loading while maintaining identical geometry and boundary conditions across all simulations.

2.4. Fatigue-Critical Region Assessment

Fatigue life assessment was performed using the Stress-Based Fatigue module in COMSOL. A multiaxial critical-plane approach based on the Findley criterion was employed to account for the combined influence of shear stress amplitude and tensile normal stress on potential crack initiation planes. The Findley criterion was selected because it is a well-established multiaxial fatigue model that evaluates crack initiation on critical planes by combining shear stress amplitude with tensile normal stress, making it suitable for components subjected to combined bending, torsion, and axial loading conditions [22].

The normal stress sensitivity coefficient was set to $k = 0.05$, indicating moderate sensitivity to tensile normal stresses, and the fatigue limit factor was defined as $f = 5 \times 10^7$ Pa. The critical-plane search resolution was set to $Q = 11$, providing sufficient angular discretization of candidate failure planes. The shear stress range was evaluated using the circumscribing circle method.

Fatigue usage factors were calculated for each of the three load cases and for four candidate materials: Ti-6Al-4V alloy, KetaSpire, PEEK, and AvaSpire. This approach enables a systematic comparison of material durability and fatigue susceptibility under physiologically representative multiaxial loading scenarios.

A complete fatigue life prediction is outside the scope of the present study.

3. Results

The mechanical properties of the candidate materials used in the finite element simulations are summarized in Table 1, including Young's modulus, Poisson's ratio, and density for Ti-6Al-4V, PEEK, AvaSpire AV-621 NT, and KetaSpire KT-820 FP. The Ti-6Al-4V properties are taken from established orthopedic-titanium reviews [7,8], while the PEEK, AvaSpire AV-621 NT, and KetaSpire KT-820 FP values were obtained from the manufacturer technical data sheets (Solvay Specialty Polymers) and cross-checked against the FEA literature on PEEK-family biomedical components [10–13,23]. The Young's modulus values reported for the PEEK-family polymers correspond to the nominal room-temperature tensile properties provided in the manufacturers' technical data sheets and were used as linear-elastic input parameters in the present finite element simulations. Although material properties may vary with testing methodology, loading mode, and temperature (for example, physiological conditions), identical material-property assumptions were applied consistently across all simulations to enable a controlled comparison of material-dependent mechanical behavior. These parameters were implemented assuming linear elastic behavior in the structural and fatigue analyses.

A significant difference in stiffness is observed between the metallic and polymer-based materials. Ti-6Al-4V exhibits a Young's modulus of 110 GPa, which is substantially higher than those of the polymers (3.0–4.1 GPa). In contrast, the lower stiffness of PEEK, AvaSpire, and KetaSpire is closer to that of natural bone, which may promote a more physiological load transfer and reduce stress shielding in prosthetic applications.

Table 1. Mechanical properties of the four candidate materials (Ti-6Al-4V, PEEK, AvaSpire AV-621 NT, and KetaSpire KT-820 FP) used as inputs for the finite element simulations: Young’s modulus (E), Poisson’s ratio (ν), and density (ρ). Ti-6Al-4V data were taken from the orthopedic-titanium reviews of Long & Rack [7] and Marin & Lanzutti [8]; PEEK, AvaSpire AV-621 NT, and KetaSpire KT-820 FP data were taken from the Solvay Specialty Polymers technical data sheets and cross-checked against [10–13,23].

Material	Young’s Modulus, GPa	Poisson Ratio	Density, kg/m ³
Ti-6Al-4V	110	0.33	4430
PEEK	4.0	0.37	1300
AvaSpire AV-621 NT	3.0	0.39	1290
KetaSpire KT-820FP	4.1	0.36	1320

The Poisson’s ratios of the materials are relatively similar (0.33–0.39), indicating comparable deformation behavior under loading. However, a major difference is observed in density, where the polymers (~1290–1320 kg/m³) are significantly lighter than Ti-6Al-4V (4430 kg/m³). This reduced density can contribute to lighter prosthetic components and improved user comfort.

Polymer-based materials provide lower stiffness and weight compared with titanium alloys, which may improve biomechanical compatibility, although their lower modulus requires careful structural design to ensure sufficient mechanical stability.

Following the material characterization, finite element simulations were conducted to evaluate stress distribution and deformation behavior under three loading scenarios.

Figure 3 shows the von Mises stress distributions for KetaSpire, AvaSpire, PEEK, and Ti-6Al-4V under three loading scenarios. Rows correspond to Load Case 1 (axial force), Load Case 2 (axial force + forward bending), and Load Case 3 (axial force + reverse bending), while columns represent the materials, respectively. Identical view settings and a unified color scale were applied to enable direct comparisons across all configurations.

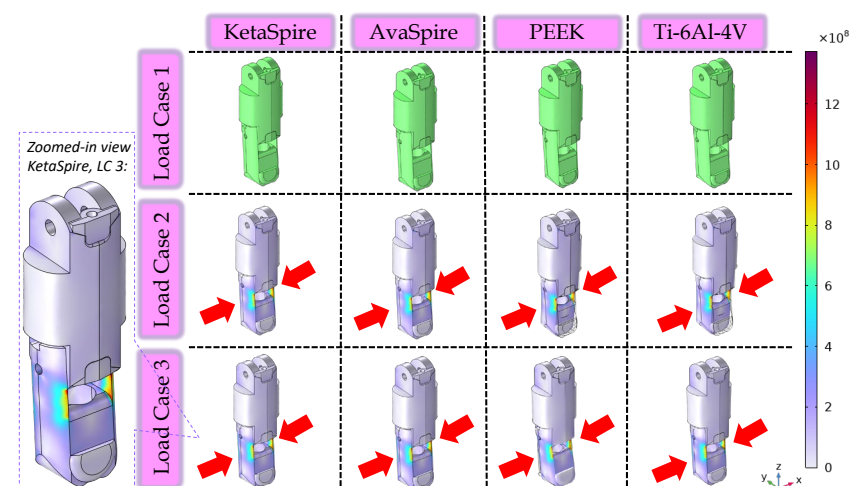


Figure 3. Finite element simulation of stress distribution in the prosthetic finger connector for four candidate materials, KetaSpire, AvaSpire, PEEK, and Ti-6Al-4V, under three loading scenarios. Load Case 1 represents axial loading, Load Case 2 corresponds to combined axial and forward bending, and Load Case 3 represents axial loading with reverse bending. The color scale indicates the von Mises stress magnitude ($\times 10^8$), where blue denotes low stress, and red indicates maximum stress concentration. Red arrows highlight the primary stress concentration regions located near the connector hinge and load-transfer interface, which represent potential fatigue-critical zones under repeated mechanical loading.

In all loading conditions, stress constantly concentrated at the hinge connector transition region, indicating that geometric discontinuities mostly dictate the location of peak stresses. Ti-6Al-4V exhibited the smallest and lowest intensity stress zones due to its high stiffness. Meanwhile, polymeric materials have shown broader distributions and higher local deformation, consistent with studies of titanium components in orthopedic implants [7–9]. On the other hand, polymeric materials displayed broader stress distributions and larger deformation regions, mostly under bending. The bending direction influenced the stress pattern: forward bending (Load Case 2) shifted peak stresses in a forward direction, whereas reverse bending (Load Case 3) shifted them in a backward direction. Axial loading alone (Load Case 1) produced relatively low gradients.

Figure 4 summarizes the maximum displacement observed for all materials under the three load cases. The minimum displacement remained zero due to fixed support boundary conditions. Load Case 2 produced the greatest deformation, followed by Load Case 3, with Load Case 1 generating minimal displacement.

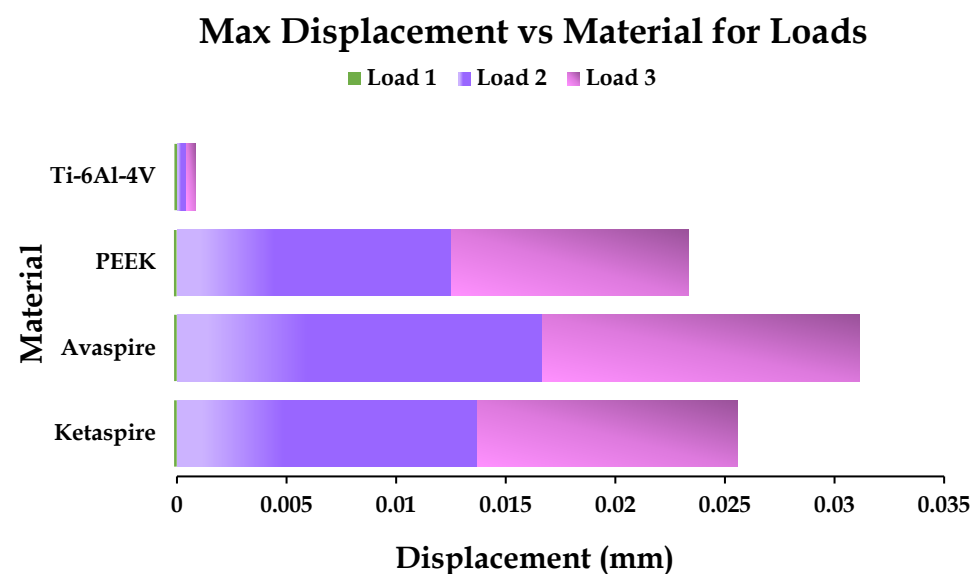


Figure 4. Maximum displacement ($u_{\max}$) of the prosthetic finger connector for the four candidate materials (Ti-6Al-4V, PEEK, AvaSpire AV-621 NT, and KetaSpire KT-820 FP) under the three load cases (LC1: axial only, $F = 50$ N; LC2: F + forward bending $M = 0.5$ N·m; LC3: F + reverse bending $M = -0.5$ N·m). Bars are grouped by load case.

Among the polymeric materials, AvaSpire showed the highest displacement, followed by KetaSpire and PEEK, reflecting their relative stiffness values. Ti-6Al-4V demonstrated an order-of-magnitude lower displacement.

Figures 5 and 6 provide longitudinal and side-view COMSOL contour screenshots for Load Cases 2 and 3. These panels illustrate the location and overall pattern of stress concentration, whereas the quantitative stress values were taken from the separately post-processed results in Table 2 and Figures 7 and 8. Stress consistently localized near the hinge-transition region, and forward bending produced slightly higher global peak stresses than reverse bending. Stress steadily localized at the hinge connector transition. Ti-6Al-4V again demonstrated the lowest stress levels and smallest concentrated zones. Meanwhile polymeric materials showed broader stress regions due to their reduced stiffness. Forward bending (Load Case 2) generated slightly higher stresses than reverse bending (Load Case 3), indicating a modest geometric asymmetry. This directional sensitivity mirrors asymmetric loading responses observed in mandibular fixation and femoral-stem implant studies [10,13,23].

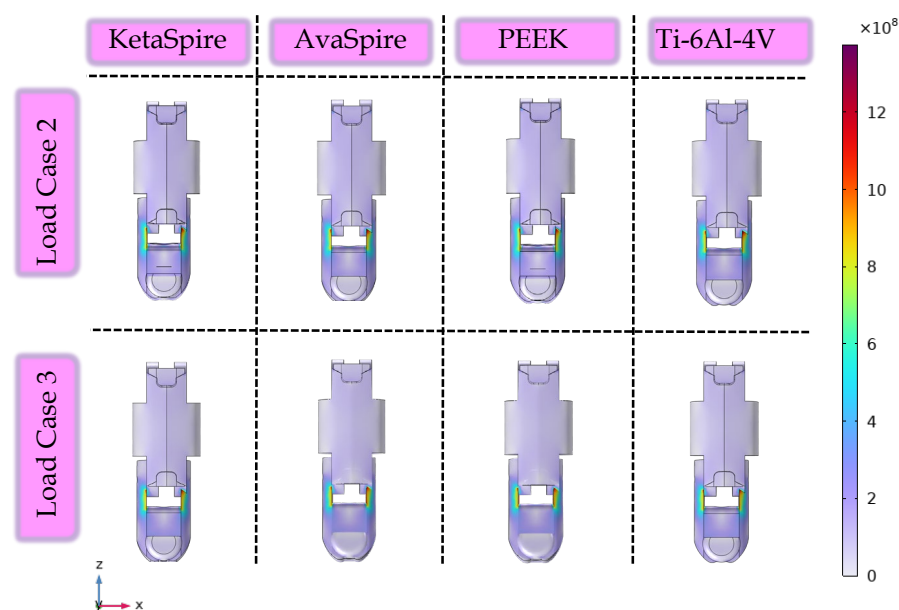


Figure 5. Von Mises stress distributions by 1D view of the prosthetic finger connector for Load Case 2 ((**top row**), axial force of $F = 50$ N combined with forward bending moment of $M = 0.5$ N·m) and Load Case 3 ((**bottom row**), axial force combined with reverse bending moment), shown from a 1D (longitudinal) viewing angle. Columns correspond to KetaSpire KT-820 FP, AvaSpire AV-621 NT, PEEK, and Ti-6Al-4V, respectively. The color scale is identical across all panels to enable a direct visual comparison.

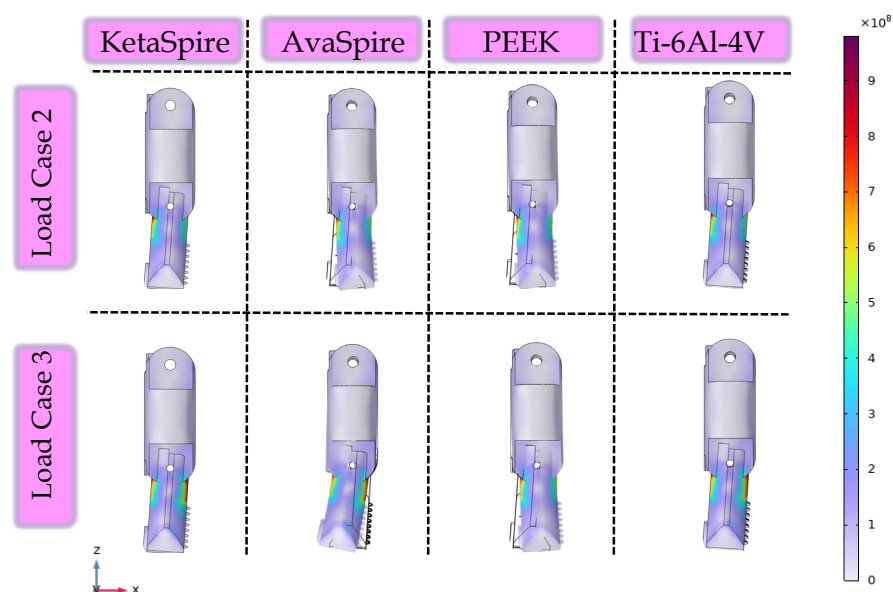


Figure 6. Von Mises stress distributions by side view of the prosthetic finger connector for Load Case 2 ((**top row**), axial force of $F = 50$ N combined with forward bending moment of $M = 0.5$ N·m) and Load Case 3 ((**bottom row**), axial force combined with reverse bending moment), shown from a side (lateral) viewing angle so that the hinge-to-stem stress transition region is fully visible. Columns correspond to KetaSpire KT-820 FP, AvaSpire AV-621 NT, PEEK, and Ti-6Al-4V, respectively. The color scale is identical across all panels.

Table 2. Quantitative summary of the finite element simulations: peak von Mises stress ($\sigma_{\max}$), maximum displacement ($u_{\max}$), and safety factor ($SF = R_e / \sigma_{\max}$) for each of the four materials under the three load cases (LC1: axial only, $F = 50$ N; LC2: F + forward bending $M = 0.5$ N·m; LC3: F + reverse bending $M = -0.5$ N·m).

Material	Load Case	$\sigma_{\max}$ (MPa)	$u_{\max}$ (mm)	R_e (MPa)	SF
Ti-6Al-4V	LC1	0.31	0	880	2839
Ti-6Al-4V	LC2	2.20	0.0005	880	400
Ti-6Al-4V	LC3	2.05	0.0004	880	429
PEEK	LC1	0.34	0	97	285
PEEK	LC2	2.52	0.0125	97	38.5
PEEK	LC3	2.35	0.0108	97	41.3
AvaSpire AV-621 NT	LC1	0.34	0	90	265
AvaSpire AV-621 NT	LC2	2.59	0.0167	90	34.7
AvaSpire AV-621 NT	LC3	2.41	0.0144	90	37.3
KetaSpire KT-820 FP	LC1	0.33	0	100	303
KetaSpire KT-820 FP	LC2	2.50	0.0137	100	40.0
KetaSpire KT-820 FP	LC3	2.33	0.0119	100	42.9

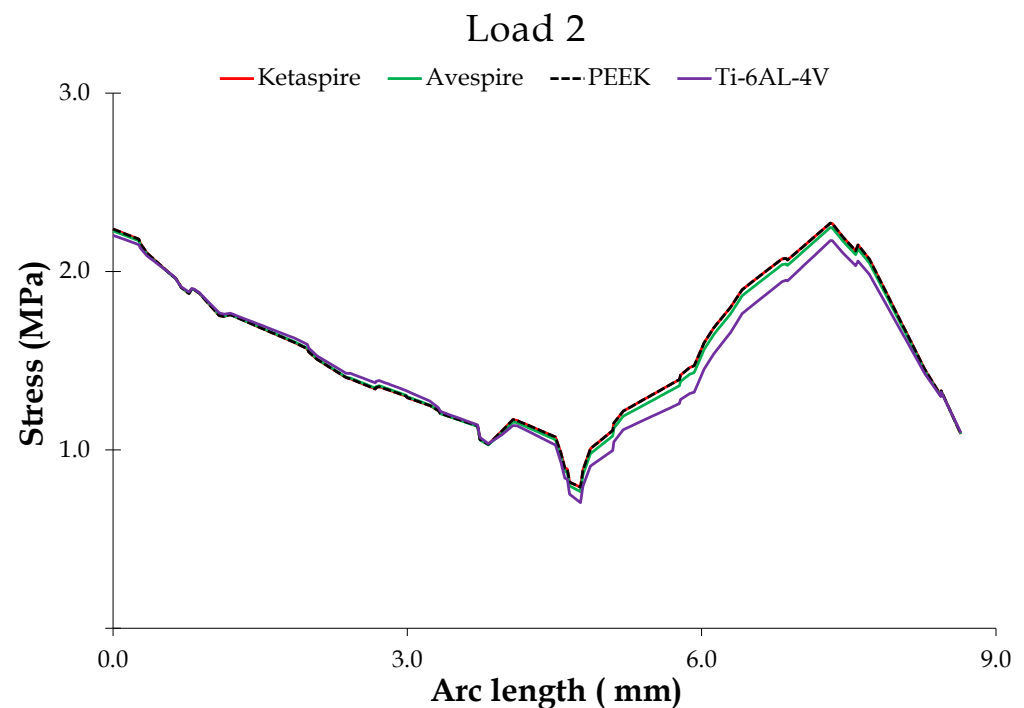


Figure 7. Variation of von Mises stress along the selected local arc-length path under Load Case 2. This figure is a separately post-processed COMSOL Line Graph, not a direct contour screenshot. The horizontal axis represents arc length (mm) along a user-defined local evaluation path on the outer surface of the hinge-transition region. The plotted path extends over approximately 0–9 mm and does not represent the full 28 mm connector length. The vertical axis reports the post-processed von Mises stress in MPa. The four curves correspond to KetaSpire KT-820 FP, AvaSpire AV-621 NT, PEEK, and Ti-6Al-4V.

Figure 7 illustrates the variation of von Mises stress along the arc length path for Load Case 2. All materials followed nearly identical stress trajectories. The stress curve displayed an initial high value ($\sim 2.2 \times 10^6$ Pa), followed by a mid-section minimum associated with a geometric transition, then a sharp rise under forward bending. Differences between the polymers and Ti-6Al-4V were minor and reflect only stiffness-dependent magnitude changes.

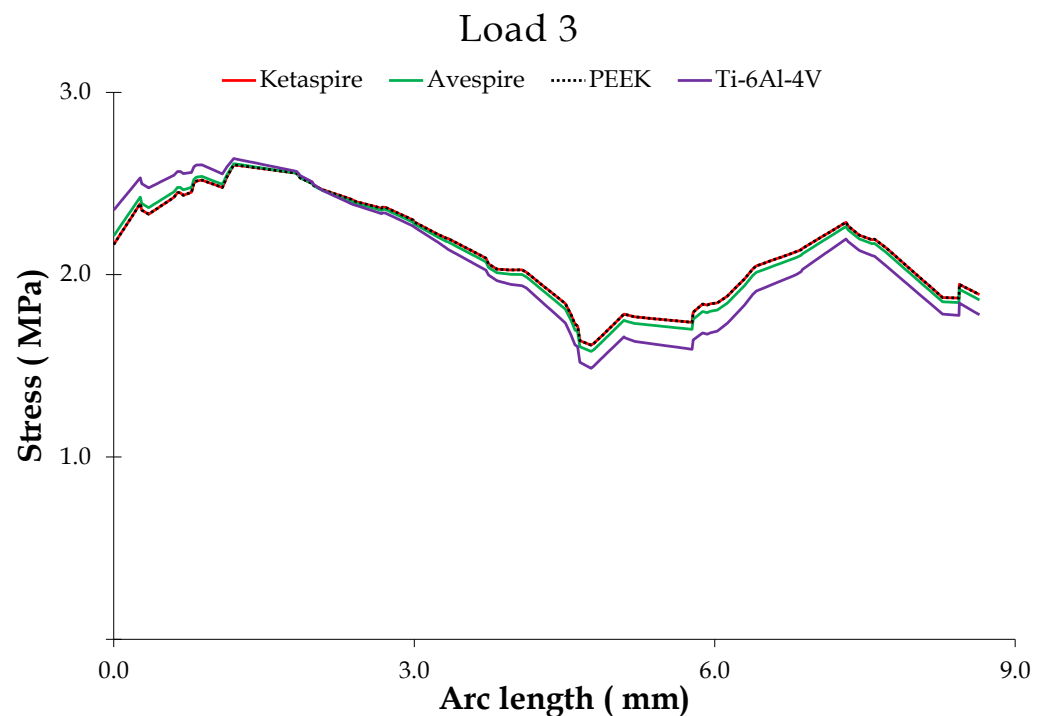


Figure 8. Variation of von Mises stress along the selected local arc-length path under Load Case 3. This figure is a separately post-processed COMSOL Line Graph, not a direct contour screenshot. The horizontal axis represents arc length (mm) along the same user-defined local hinge-transition path used in Figure 7. The plotted path extends over approximately 0–9 mm and does not represent the full 28 mm connector length. The vertical axis reports the post-processed von Mises stress in MPa. The four curves correspond to KetaSpire KT-820 FP, AvaSpire AV-621 NT, PEEK, and Ti-6Al-4V.

Figure 8 shows the separately post-processed stress distribution along the selected local path for Load Case 3. The materials produced similar stress profiles. The highest value along this local path was approximately 2.5 MPa. This local-path result should be distinguished from the global peak stress reported in Table 2.

To consolidate the quantitative outcomes of the twelve simulated combinations (four materials $\times$ three load cases) and to address how the simulated peak stresses compare against the material strength of each material, Table 2 summarizes the peak von Mises stress ($\sigma_{\max}$), maximum displacement ($u_{\max}$), and safety factor ($SF = R_e / \sigma_{\max}$) computed against the yield strength of each material. The $u_{\max}$ values presented in Table 2 were extracted from the post-processed numerical COMSOL results and provide the quantitative maximum displacement for each material and loading condition, complementing the graphical displacement comparison shown in Figure 4.

4. Discussion

The results demonstrate that stress distribution in prosthetic finger connectors is primarily governed by geometry rather than material selection, with consistent stress concentration observed at the hinge-transition region under all loading conditions. This observation aligns with previous findings in prosthetic hand mechanisms, artificial joints, and upper-limb splint structures where hinge regions and thin connectors act as stress risers [1,2,4,6,14]. While Ti-6AL-4V exhibited significantly lower stress magnitudes and minimal deformation due to its known high stiffness. PEEK-based polymers showed increased deformation and a wider stress distribution, which is a reflection of their lower elastic moduli. This behavior has also been noted in polymeric and composite implant analyses, where lower modulus materials deform more under bending loads [10,13,23]. The increased deformation observed for polymeric materials confirms the central role of

bending in driving connector displacement, which is consistent with compliant-mechanism and anthropomorphic hand simulations [2,3,14].

Bending loads majorly affect mechanical response, with load direction impacting the spatial location of peak stress without affecting the overall stress pathway. This supports the finding that bending moments dominate the mechanical response of lean prosthetic connectors, as previously reported in SMA-actuated and tendon-driven finger simulations [2–4]. The observed directional sensitivity mirrors asymmetric loading responses reported in mandibular fixation and femoral-stem implant studies, where changes in bending direction altered local stress distribution without changing the overall mechanical pathway [10,13,23]. These findings align with bending-dominant stress trends investigated in prosthetic, craniofacial, and orthopedic biomechanics research works [1–4,10–13,23]. All the findings clearly indicate that although material choice impacts the magnitude of mechanical response, the connector geometry dictates stress localization. Thus, a proper geometry-driven design must be considered for any prosthetic system. Beyond confirming this geometry-versus-material trend, the simulations enable a deeper engineering interpretation. The arc-length stress profiles in Figures 7 and 8 show that all four materials reproduce the same characteristic three-zone signature: a high-stress region at the proximal load-introduction boundary, a low-stress mid-section corresponding to the slender hinge transition, and a secondary peak near the distal coupling. The identical arc-length stress trajectories further demonstrate that connector geometry, rather than material selection, primarily governs stress flow. Similar geometry-controlled stress pathways have been reported in finite element analyses of femoral stems and craniofacial implants, where stress localization was mainly dictated by structural geometry regardless of material choice [5,10,13,23]. The fact that this signature is invariant across materials, while the absolute stress magnitude scales weakly with modulus and the displacement scales strongly with modulus, is the mechanistic explanation for the geometry-dominated behavior: the elastic stress field in a stationary, geometrically identical part is set by the equilibrium equations and boundary conditions, and material modulus only acts as a scalar that re-scales the deformation field needed to satisfy them. This explains why the polymers, despite being approximately 27–37 times more compliant than Ti-6Al-4V, do not exhibit qualitatively different stress patterns but produce approximately 27–37 times greater maximum displacements under LC2 and LC3 (Table 2). Consequently, functional deformation may become the governing design limitation before static yielding occurs. This trend agrees with previous titanium biomechanical studies showing that high elastic modulus materials minimize deformation and maximize rigidity, whereas polymeric alternatives provide enhanced compliance, manufacturability, and reduced weight [7–12]. Comparing the simulated peak stresses against the material yield strengths (Table 2) shows that Ti-6Al-4V operates at less than 0.3% of its yield strength ($SF \approx 400$ – 2839) in every load case, whereas the PEEK-family polymers operate at approximately 2.3–2.9% of their yield strength under the bending load cases ($SF \approx 34.7$ – 42.9). Across all twelve material–load case combinations, the safety factor remains substantially above unity ($SF \geq 34.7$ for all polymer cases and $SF \geq 400$ for all Ti-6Al-4V load cases), indicating that under the assumed physiological loading conditions ($F = 50$ N, $M = 0.5$ N·m), none of the four materials is at risk of static yielding in the hinge region. These results suggest that static strength is not the limiting design criterion for the connector and that fatigue performance is likely to govern its long-term structural integrity. However, the relative safety margin of titanium is an order of magnitude larger, and under cyclic loading the higher fractional utilization of polymer strength translates directly into a shorter projected fatigue life. From a clinical perspective the trade-off is not driven by static strength alone: titanium offers superior mechanical reliability but at the cost of a fourfold-higher density (4430 vs. ~ 1300 kg/m³), greater stress shielding of the surrounding biological/prosthetic structures, more expensive

subtractive manufacturing, and the perception of a “cold” or heavy prosthesis by the user. PEEK, AvaSpire, and KetaSpire offer a much lighter, more comfortable, easier-to-mold and 3D-printable hinge that more closely matches the modulus of natural bone and surrounding soft tissue, at the cost of larger functional displacements and reduced fatigue safety margins. For a high-load occupational prosthesis Ti-6Al-4V is the safer choice; for everyday and pediatric prostheses, where weight, comfort, and rapid customization dominate, KetaSpire (the stiffest polymer of the three) is the most attractive compromise. The use of finite element analysis (FEA) is central to this comparison: FEA allows for the same geometry, same boundary conditions, and same load path to be applied to four materials whose physical samples would otherwise have to be fabricated and tested individually at a considerable cost. It provides full-field maps of stress and displacement that strain gauges cannot resolve, isolates the influence of one variable (material) at a time, and supports parametric sensitivity studies (for example, mesh, bending direction, and load magnitude) in a single virtual environment. These benefits are precisely what is exploited in the present geometry-invariant comparative framework.

5. Limitations

The present study has several limitations that should be considered when interpreting the results and that motivate the directions for future work. First, only static (stationary) load cases were simulated; cyclic loading, impact, and dynamic acceleration during rapid grasping were not modelled, and the reported safety factors therefore describe static yielding only, not fatigue life. Second, all four materials were modelled with linear-elastic constitutive laws. While this is a standard first approximation, it does not capture the viscoelasticity, strain-rate sensitivity, creep, and localized plastic flow that the PEEK-family polymers are known to exhibit; under repeated loading these mechanisms would increase the effective deformation and reduce the effective fatigue limit of the polymer connectors, so the polymer displacements reported here should be regarded as lower bounds and the polymer safety factors as upper bounds. Third, contact between the connector and the hinge pin/adjacent phalangeal segments was idealized as fully bonded (no contact mechanics, no friction, and no sliding), which neglects local pressure peaks and wear at the hinge bore that are particularly important for KetaSpire and AvaSpire in tribological service. Fourth, the boundary conditions are idealized: the proximal stem is treated as a perfectly rigid Fixed Constraint and the distal load surface as a perfectly rigid connector, whereas in practice the surrounding prosthesis introduces additional compliance. Fifth, no experimental validation against strain-gauge or digital image correlation (DIC) measurements on the fabricated prototype has been performed in the present work; although finite element analysis is widely accepted as a predictive tool for evaluating the mechanical behavior of biomedical structures, experimental verification remains essential for confirming numerical predictions and strengthening confidence in engineering-oriented studies. The simulation results are therefore consistent with the published FEA literature on PEEK and Ti-6Al-4V components [5,10–13,23] but have not been independently calibrated against measurements on this specific connector. Therefore, the present findings should be interpreted as predictive rather than experimentally validated. Future work will focus on validating the numerical results through mechanical testing of fabricated connector prototypes, enabling a direct comparison between simulated and experimentally measured stress and deformation under physiologically relevant loading conditions. Sixth, the fatigue analysis described in the Materials and Methods (Findley critical-plane criterion, $k = 0.05$, $f = 5 \times 10^7$ Pa, and $Q = 11$) was implemented but has not yet been reported in dedicated figures; full per-material, per-load-case fatigue usage maps are an explicit objective of a follow-up study.

Each of these limitations is addressed in the Conclusions and represents a specific item for future experimental and numerical work.

6. Conclusions

This study presented a comparative finite element analysis of the mechanical behavior of prosthetic finger connectors manufactured from Ti-6Al-4V and high-performance PEEK-based polymers (PEEK, KetaSpire, and AvaSpire) under three physiologically relevant loading scenarios. The most important quantitative outcomes of the study are as follows: (i) peak von Mises stresses at the hinge transition reached approximately 2.2 MPa for Ti-6Al-4V and 2.5–2.6 MPa for the three PEEK-family polymers under the combined axial + forward bending load case; (ii) maximum displacements were ~0.0005 mm for Ti-6Al-4V versus 0.0125–0.0167 mm for the polymers, corresponding to approximately 27–37 times greater displacements; (iii) the corresponding safety factors against static yielding (Table 2) were $SF = 400\text{--}2839$ for Ti-6Al-4V and $SF \approx 34.7\text{--}303$ for the polymers, indicating that none of the materials is at risk of static failure under daily grasping loads but that the polymers operate with a substantially smaller margin and that all materials provide substantial margins of safety against yielding; and (iv) all four materials produced nearly identical normalized stress trajectories along the connector arc length, demonstrating that geometry—not material—primarily governs stress localization. The simulations demonstrated that stress concentration consistently occurs at the hinge-transition region, confirming that connector geometry is the dominant factor governing stress localization in prosthetic finger mechanisms.

The results further show that bending loads play a significantly greater role than axial forces in determining deformation and stress patterns. Forward and reverse bending shifted the position of peak stresses but did not alter the overall stress pathway along the connector. Across all configurations, Ti-6Al-4V exhibited the lowest deformation and stress magnitude due to its high elastic modulus, indicating superior structural rigidity and mechanical stability.

In contrast, PEEK-based polymers produced broader stress distributions and higher displacement values, reflecting their lower stiffness. Nevertheless, these materials offer important advantages, including reduced weight, improved manufacturability, and enhanced mechanical compliance, which are beneficial for lightweight prosthetic devices and patient comfort.

Overall, the findings indicate that connector geometry primarily governs stress distribution, while material stiffness mainly influences deformation amplitude and stress magnitude. Titanium alloys provide maximum structural reliability, whereas PEEK-derived polymers represent promising alternatives for lightweight prosthetic components when appropriate geometric reinforcement is applied.

Future work should incorporate experimental validation and advanced fatigue evaluation using the Findley multiaxial fatigue model, enabling a more accurate prediction of long-term durability and guiding the development of next-generation prosthetic finger connectors.

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Abbreviations

The following abbreviations are used in this manuscript:

FEA	Finite Element Analysis
PEEK	Polyetheretherketone
PAEK	Polyaryletherketone
LG	Load Group
LC	Load Case
SMA	Shape Memory Alloy
CFRPC/CFR-PEEK	Carbon Fiber Reinforced Polymer/PEEK
PLA	Polylactic Acid
DIC	Digital Image Correlation

References

1. Abdelbaset, R.; Mohamed, B.A.; Afify, H.M. A 3-D biomechanical model for enhancing the performance of artificial hand. In *Proceedings of the 2018 IEEE 4th Middle East Conference on Biomedical Engineering, Tunis, Tunisia, 28–30 March 2018*; IEEE: New York, NY, USA, 2018.
2. Kaya, Z.E.; Yilmaz, A. Modeling and simulation of an anthropomorphic hand prosthesis with an object interaction. *Comput. Methods Programs Biomed.* **2020**, *183*, 105085. [[CrossRef](#)] [[PubMed](#)]
3. Simone, F.; Meli, D.; Rizzello, G.; Naso, D.; Seelecke, S. Finite element modeling and simulation of a robotic finger actuated by Ni-Ti shape memory alloy wires. In *Proceedings of the SPIE Smart Structures and Materials + Nondestructive Evaluation, Vancouver, BC, Canada, 16–19 March 2026*; p. 14. [[CrossRef](#)]
4. Young, K.J.; Pierce, J.E.; Zuniga, J.M. Assessment of body-powered 3D printed partial finger prostheses: A case study. *3D Print. Med.* **2019**, *5*, 7. [[CrossRef](#)] [[PubMed](#)]
5. Ceddia, M.; Solarino, G.; Giannini, G.; De Giosa, G.; Tucci, M.; Trentadue, B. A Finite Element Analysis Study of Influence of Femoral Stem Material in Stress Shielding in a Model of Uncemented Total Hip Arthroplasty: Ti-6Al-4V versus Carbon Fibre-Reinforced PEEK Composite. *J. Compos. Sci.* **2024**, *8*, 254. [[CrossRef](#)]
6. Mian, S.H.; Umer, U.; Moiduddin, K.; Alkhalefah, H. Finite Element Analysis of Upper Limb Splint Designs and Materials for 3D Printing. *Polymers* **2023**, *15*, 2993. [[CrossRef](#)] [[PubMed](#)]
7. Long, M.; Rack, H.J. Titanium alloys in total joint replacement—a materials science perspective. *Biomaterials* **1998**, *19*, 1621–1639. [[CrossRef](#)] [[PubMed](#)]
8. Marin, E.; Lanzutti, A. Biomedical Applications of Titanium Alloys: A Comprehensive Review. *Materials* **2024**, *17*, 114. [[CrossRef](#)] [[PubMed](#)]
9. Shen, X.; Shukla, P. A Review of Titanium Based Orthopaedic Implants (Part-I): Physical Characteristics, Problems and the need for Surface Modification. *Int. J. Peen. Sci. Technol.* **2020**, *1*, 301–332.

10. El-Sawy, M.A.; El-Khatib, B.; Borg, H.S.; Khater, M.T. Biomechanical effects of digitally constructed titanium, modified polyetheretherketone, and polyetherketoneketone subperiosteal implants on atrophied maxilla: A finite element analysis. *BMC Oral Health* **2025**, *25*, 1142. [[CrossRef](#)] [[PubMed](#)]
11. Smith, J.A.; Petersmann, S.; Arbeiter, F.; Schäfer, U. Optimization and manufacture of polyetheretherketone patient specific cranial implants by material extrusion—A clinical perspective. *J. Mech. Behav. Biomed. Mater.* **2023**, *144*, 105965. [[CrossRef](#)] [[PubMed](#)]
12. Lee, M.Y.; Lee, S.H.; Leigh, J.H.; Nam, H.S.; Hwang, E.Y.; Lee, J.Y.; Han, S.O.; Lee, G.M. Functional improvement by body-powered 3D-printed prosthesis in patients with finger amputation: Two case reports. *Medicine* **2022**, *101*, e29182. [[CrossRef](#)] [[PubMed](#)]
13. Chen, L. Finite element comparison of titanium and polyetheretherketone materials for mandibular defect reconstruction. *Am. J. Transl. Res.* **2024**, *16*, 6097–6105. [[CrossRef](#)] [[PubMed](#)]
14. Simone, F.; Rizzello, G.; Seelecke, S. A finite element framework for a shape memory alloy actuated finger. *J. Intell. Mater. Syst. Struct.* **2019**, *30*, 2052–2064. [[CrossRef](#)]
15. Vidakis, N.; Petousis, M.; Ntintakis, I.; David, C.; Sagrais, D.; Mountakis, N.; Moutsopoulou, A. Quantitative Insight into the Compressive Strain Rate Sensitivity of Polylactic Acid, Acrylonitrile Butadiene Styrene, Polyamide 12, and Polypropylene in Material Extrusion Additive Manufacturing. *J. Dyn. Behav. Mater.* **2024**, *10*, 251–269. [[CrossRef](#)]
16. Ma, L.; Liu, F.; Liu, D.; Liu, Y. Review of strain rate effects of fiber-reinforced polymer composites. *Polymers* **2021**, *13*, 2839. [[CrossRef](#)] [[PubMed](#)]
17. Aoyagi, Y.; Camboulives, L.N. Modeling of nonlinear viscoelastic-viscoplastic behavior of glassy polymers based on intramolecular rotation of molecular chains. *Int. J. Plast.* **2025**, *189*, 104328. [[CrossRef](#)]
18. Farotti, E.; Mancini, E.; Lattanzi, A.; Utzeri, M.; Sasso, M. Effect of temperature and strain rate on the formation of shear bands in polymers under quasi-static and dynamic compressive loadings: Proposed constitutive model and numerical validation. *Polymer* **2022**, *245*, 124690. [[CrossRef](#)]
19. Yang, X.; Liu, C.; Ding, Y.; Li, H.; Guo, Y.; Li, Y. Dynamic yielding and plastic flow behavior of Ti-6Al-4V under complex loading. *Int. J. Solids Struct.* **2023**, *283*, 112476. [[CrossRef](#)]
20. Cheng, J.; Li, J.; Yu, S.; Du, Z.; Zhang, X.; Gai, J.; Fan, J.; Zhang, W.; Song, H.; Yu, Z. Room Temperature Deformation and Superelastic Behavior of TLM Titanium Alloy Under Different Solution Conditions. *Front. Mater.* **2020**, *7*, 114. [[CrossRef](#)]
21. Suhas, P.; Quadros, J.D.; Mogul, Y.I.; Mohin, M.; Aabid, A.; Baig, M.; Ahmed, O.S. A review on mechanical metamaterials and additive manufacturing techniques for biomedical applications. *Mater. Adv.* **2024**, *6*, 887–908. [[CrossRef](#)]
22. Bibbo, N.D.; Baumgartner, J.; Arora, V. Comparative study of critical plane fatigue criteria on multiaxial variable amplitude loaded welded test specimens. *Int. J. Fatigue* **2022**, *158*, 106670. [[CrossRef](#)]
23. Milone, D.; Spataro, M. Biomechanical Evaluation of PEEK and PLA Composite Femoral Implants for Stress Shielding Reduction: A Finite Element Simulation Study. *Prosthesis* **2025**, *7*, 44. [[CrossRef](#)]

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