RESEARCH PAPER

Numerical and Experimental Investigation of Wearable Body-Powered Finger Prosthetics: A Comparative Study

Madhu Mohan R[1], Subashchandra Kattimani[1]*, Poornesh Kumar Koorata[1]* and Prashanth M Vishwanath[2]

[1]Department of Mechanical Engineering, National Institute of Technology Karnataka, Surathkal, Karnataka, India.
[2]Department of Biochemistry, JSS Medical College, JSS Academy of Higher Education and Research, Mysuru 570015, Karnataka, India
*Email of corresponding author: subhaskatti@nitk.edu.in

Received: 28th Feb, 2026; Revised: 6th March 2026; Accepted: 7th April, 2026; Available Online: 20th April, 2026

ABSTRACT
The individual person with partial finger amputation which affect the various day to day activities, difficult to performing grasping, pinching action and presided movement controls previous research works are based improving results in simulation reading but there is a large gap in real time operation inaccuracy in validation, assumption, type of material used in simulation and lack of user-centric test procedure for the prosthetic. This paper identified drawback with simulation and experimental test result the prosthetic finger is evaluating by comparing both the result and to study the discrepancies between simulation reading and real-time experimental test result and proposes the number of finding to improve the simulation test and experimental test result the prosthetic is tested with customized real-time operation test setup with curved slot having angular position to mount the prosthetic which mimic the natural movement of the finger number of test are performed with varying anglar positions .This study finds average discrepancies of 15.85% comparing with simulation and experimental test result this difference occurred due to calibration error in load cell , assumption arrived for hinge point in simulation and post processing in 3d printing of prosthetic finger .This study helps to bridge the gap between simulation and experimental test to improve the test procedure and advance material to design the prosthetic finger.
Keywords: *Passive body powered prosthetic finger, CAD simulation, mechanical testing, DIP distal interphalangeal, PIP proximal interphalangeal, MIP metacarpophalangeal.*
How to cite this article: R MM, Kattimani S, Koorata PK and Vishwanath PM, Numerical and Experimental Investigation of Wearable Body-Powered Finger Prosthetics: A Comparative Study. Int J Drug Deliv Technol. 2026;16(5): 155-169. DOI: 10.25258/ijddt.16.5.18
Source of support: Nil.
Conflict of interest: None

1. INTRODUCTION

A limb is a very essential part of human body without which the person loses either the proficiency or the control over the several daily tasks and in turn loses his normalcy [1]. When compared to other amputation the effects for amputee not only limits to functionalities handling and physical limitations but also effects person's social success, Self-realization and Quality of life [2]. The study and survey have been carried out on the 28 individual who had lost their limbs particularly the upper limb in order to address, understand the need, their preferences and the extent of effects those individual feels and go through which are discussed above [3].

Prosthetics has very long history of development and usage. During old times, they used to reduce the effect of disabilities and hence re-introduce normalcy of a person. Continuous adaptation and evolution of the prosthetics over the year has led to tremendous growth which was the effect of both technological advancement and deep understanding of the human anatomy, its effects and behavior to the external bodies [4]. FLUIDHAND III is one of those development which falls in this category. It was invented by Gaiser in 2009 [5]. In this the inventor has incorporated fluid actuation system coupled with versatile control mechanism. This was included with extra features like vibrotactile force feedback which produced some hopes for the persons who had gone upper limb amputation.

*Author for Correspondence: subhaskatti@nitk.edu.in

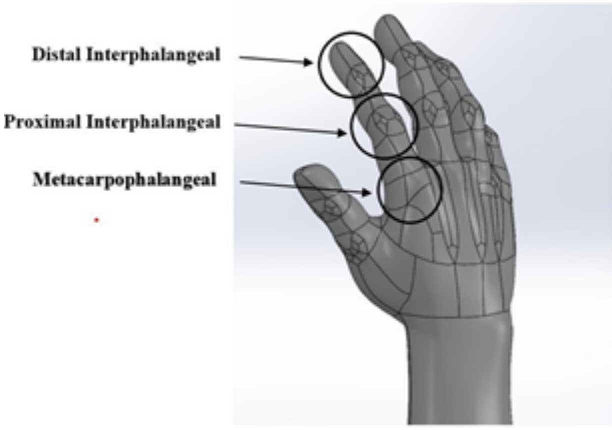

Figure 1: Distal interphalangeal, Proximal interphalangeal, and Metacarpophalangeal joints [6].

The prosthetics should be design and should function in such a way that it should cater each individual need without affecting their comfort which keeps on varying from person to person. This is the area where the FLUIDHAND III had not concentrated with some extra areas like aesthetics and user specific conditions which are very important while developing. Achieving each individuals' satisfactions and comforts is very important while development of these which in-turn enhances user approval, satisfaction and their usage.

There has been a major development concerting the upper limb amputation but the gap in the development of solution for Distal Interphalangeal (DIP) remained as without completely served. In General, the DIP joint plays prominent role in handling dexterity, fine works impacting accuracy, overall affecting hand functionality [8]. Hence loss of DIP joints affects those areas with additionally affecting Individual's Confidence and quality of life reducing the social success. Hence it is essential to address the problem and providing a solution for those who lost their DIP joint which should re-initiate the mechanism or movement of normal finger anatomy.

This Research paper explains the mechanical properties of the finger prothesis which is developed to address the issues of the person who lost particularly DIP joint which will Replicate and Re-initiate the normal mechanical movement of the human finger without affecting the Anatomy variations and serves the individual need towards functional requirement [9]. Primarily An 3d model of finger prosthesis was designed using CAD software and then it is subjected to Virtual simulation using Ansys Workbench for analyzing the various load points and failure loads that the finger can withstand at different positions. On the other hand, 3D model is bought into reality by 3d Printing and physical load testing has been carried out in order to compare the values of virtual simulation and Realtime mechanical testing in order to evaluate the Functional capability and mechanical properties of the finger prosthesis. Proximal phalanx acts as the base for the finger prothesis which is used for a person who lost his DIP joint as shown in the Figure 1. Initially 3D model design virtual simulation helps to analyze basic mechanical design requirement, corrections in the errors and degree of functional requirement it satisfies, reducing the physical requirement for iteration. The CAD design helps to design the finger prosthesis depending on the individual requirement and in customizable size which increases the comfort and the functional satisfaction. The 3D model is then subjected to Virtual simulation by applying suitable mechanical properties for predicting the behavior of the model under different conditions and positions and then evaluated using various Engineering principles. Necessary design adjustments, corrections for achieving functional requirements are carried out by analyzing the results of this virtual simulations.

This comparison made between virtual simulation and practical real-world testing helps to evaluate the degree of which the finger satisfies the desired specifications and perform as required in the real-world applications. Recently there has been more studies carried out on the abilities of 3d printed parts in the prosthetics development of customizable shapes, sizes, functional satisfaction with reduction in both time and cost [10]. Although there are lot of studies, there is a lack of research and empirical evidences on 3d printed prosthetics of upper limb on long term usability, functionality and Acceptance by user particularly for the used suffered from partial limb amputations.

This research addresses the gap by experimentation carried on comparison of the results of CAD simulation with real-world physical test. This type of study Contributes for developing the more user-friendly prosthesis that restores the normal movement of the natural finger with consideration of anatomical variations and requirements for functional satisfaction. This comprehensive evaluation on the developed finger prosthesis particularly for partial finger amputees ensures that it satisfies their functional requirements and daily needs.

2. NEED FOR SIMULATION AND TESTING

A bionic prosthetic finger functions just like mimicking the articulation motion of real biological human finger for enhanced experience in absence of partial finger like PIP, DIP, MIP, etc. The mathematical modelling, finite element method of analysis helps us understand the mechanical motion and behavior of links and joints specially for disorders like arthritis according to study [13], it also emphasizes grip force distribution as one of the prime parameters for performance evaluation [14].

In case of adaptive strategies, a comparison between natural fingers and prosthetic hands force distribution plays a vital role. Previous researchers [15] and [16] have stressed upon grip and pinch force case studies involving load distribution across finger to palm.

To notice the maximum and minimum force load throughout, testing of prosthetic finger must be done during fingertip activities such as grasping action of rock climbing and manual labor. This monitoring is required because of high loads on the finger tips is applied during the actions such as crimp, half crimp, pinch grips and other high risking musculoskeletal injuries. [17], [18], [19], [20], [21]. Therefore, designing for extreme fingertip loads is essential for real-world applications.

3. SIMULATION

Ansys non-Linear Structural Analysis is used for simulation of the Finger prosthesis 3D model which was created using Solid works CAD Software. This non-Linear structural analysis is Suitable for non-linear stress-strain behavior under constant load.

Type of simulation: Nonlinear Static structural Analysis is, under constant applied load it follows non-linear stress strain curve.

Procedure:
3D model is prepared for analysis by understanding its geometry. Boolean operation is used for achieving error free analysis of the 3D prosthesis in the Ansys Workbench. Material selected as Polylactic Acid (PLA), structural attributes of the 3D model was given in the table 1 which was divided into 3 different parts as Distal Interphalangeal (TOP), Proximal Interphalangeal (Middle), Metacarpophalangeal (Bottom). Moment of Inertia, Volume, mass and Centroid Coordinates are the Attributes. As mentioned earlier, since the 3D model is manufactured by process of 3D printing and hence for matching the physical model's density, new material definitions were established which is necessary for accurate simulation.

Figure 2: Designed Prosthetic Finger.

Generating Mesh
The cleaned model was meshed using tetrahedral elements, with different target element sizes applied as per the requirement for the entire model, as showed in Figure 3.

Type of element used: 3D Tetrahedral elements.

Element size used: 1mm for the entire finger and 2mm for the mountings and links.

Applying boundary condition

After meshing the model perfectly, the forces acting on the finger were defined. Single forces were applied to the top frame, and fixed supports were provided at the required locations to ensure the simulation matched the experimental conditions.

Solving
After applying the required boundary conditions, this step solved the model using the solve option in the Ansys Workbench. The system ran a set of calculations to solve the given problem. It generated the solutions for the required parameters, such as deformation, stress, and strain depicted in Figure 4 & 5.

Table 1: Properties of Prosthetic Finger

Physical	
Density	952.43 kg/m^3
Structural	
Derive from	Young's modulus and poisson's Ratio
Young's modulus	3.45 GPa
Poisson's ratio	0.39
Bulk modulus	5.2273 GPa
Shear Modulus	1.241 GPa
Tensile ultimate Strength	59.2 MPa

Tensile yield strength	54.1 MPa
Properties of links	
Volume	$5.1245 \times 10^{-7} m^3$
Mass	0.00048807 kg
Centroid X	0.091502 m
Centroid Y	-0.046365 m
Centroid Z	0.018068 m
Moment of Inertia Ip1	7.273×10^{-7} kg·m^2
Moment of Inertia Ip2	7.9366×10^{-8} kg·m^2
Moment of Inertia Ip3	6.5589×10^{-7} kg·m^2
Properties of Bottom part	
Volume	$1.229 \times 10^{-6} m^3$
Mass	0.00097703 kg
Centroid X	0.064524 m
Centroid Y	-0.043847 m
Centroid Z	0.0059993 m
Moment of Inertia Ip1	7.7551×10^{-7} kg·m^2
Moment of Inertia Ip2	1.2521×10^{-6} kg·m^2
Moment of Inertia Ip3	1.0372×10^{-6} kg·m^2
Properties of Middle part	
Volume	$1.1451 \times 10^{-6} m^3$
Mass	0.0089891 kg
Centroid X	0.090419 m
Centroid Y	-0.049329 m
Centroid Z	0.0060664 m
Moment of Inertia Ip1	7.6195×10^{-6} kg·m^2
Moment of Inertia Ip2	9.9983×10^{-6} kg·m^2
Moment of Inertia Ip3	1.2145×10^{-5} kg·m^2
Properties of Top part	
Volume	$1.4794 \times 10^{-6} m^2$
Mass	0.0011995 kg
Centroid X	0.10989 m
Centroid Y	-0.058919 m
Centroid Z	0.0060009 m
Moment of Inertia Ip1	7.5249×10^{-7} kg·m^2
Moment of Inertia Ip2	5.6347×10^{-7} kg·m^2
Moment of Inertia Ip3	8.7124×10^{-7} kg·m^2

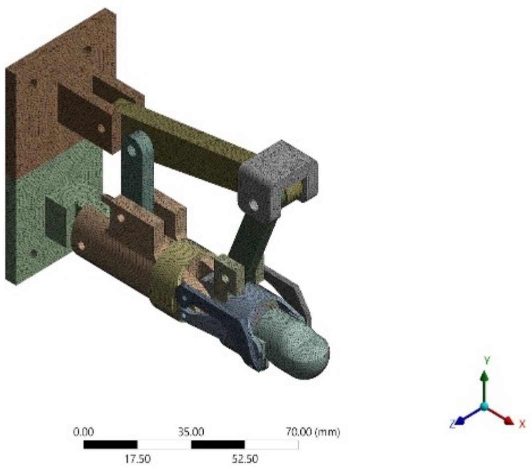

Figure 3: Meshed 3D Model

Stress Analysis and Functionality of the Jig

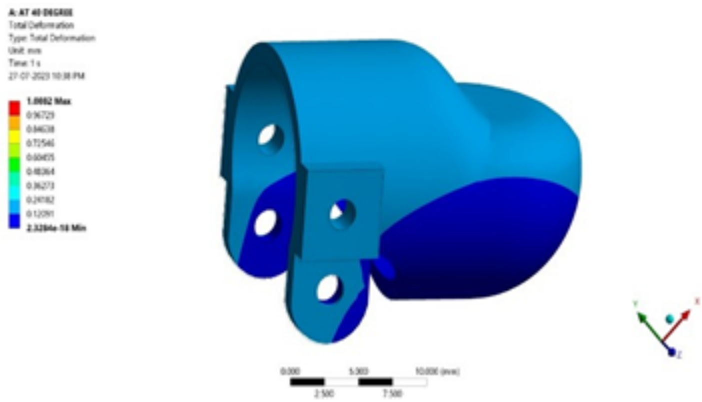

Figure 4: Simulated Distal Phalanx

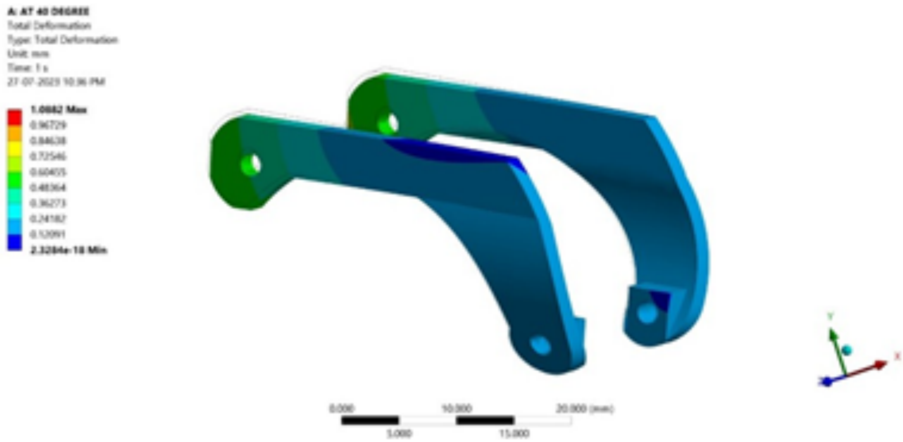

Figure 5: Simulated Linkages

The jig in the present experimentation is an important setup in order to articulate the natural action and loading conditions experienced by a prosthetic device specially designed for partial finger amputation. The jig facilitates the application of internal loading through a body powered prosthetic finger, which helps in comprehensive and a wholesome assessment of prosthetic finger including its performance and structural integrity.

The specially designed jig helps in resolving stresses through experimentation. It also ensures smooth and naturally articulated moment of the prosthetic finger through accurately engineered linkages. The jig enables experimentation of applying loads along the same direction as of the joints motion and testing various flexion. This plays a vital role for evaluating prosthetic response under various constraints. The design of jig is meticulously engineered with linkages, ensuring the smooth and natural articulation of prosthetic finger. The strategic placement of load applying points allows comprehensive evaluation of strength and stability of prosthesis. Additionally, a curved slot accommodates the finger, simulating its full range of motion, and assist in precise data acquisition including angular measurements as shown in figure 8.

Stress analysis during the simulation is obtained by prosthetic finger interfaced with load cell, this enables real time monitoring of forces applied. The motion of prosthetic responding to different loads and stresses enhances the understanding performance of prosthetic finger under real time conditions.

Analyzing Results

The 3D model of the jig with mounted prosthetic finger is depicted in figure 6. The set-up results in examining one newton load at 0 degrees, which induced a very minimal deformation. Key findings were notices in different aspects of prosthetic model.

The maximum equivalent stress of 0.88 MPa in the upper linkages were determined, with the majority of stresses remains within 0.12 MPa in prosthetics. The distal phalanx is observed with 1.65 MPa, while the yield stress was established with 54.1 MPa at the same point.

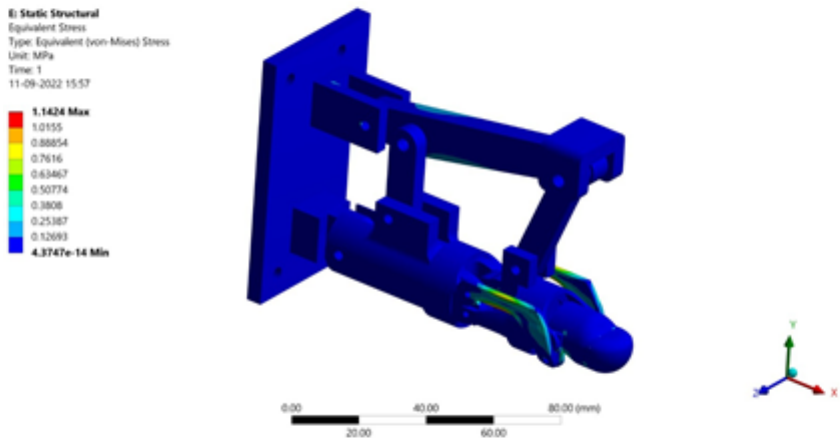

Figure 6: 3D model of the Jig with the Mounted Prosthetic Finger.

Table 2: Test Results of Computer Simulation of Prosthetic Finger under Loading.

Sl No	Elevation of Distal Tip (Degrees)	Failure Load (N)
1	0	31.96
2	10	63.74
3	20	64.20
4	40	96.10
5	50	156.90

Factor of Safety (FOS) Calculation

$$\text{FOS} = \frac{\text{Yield Stres}}{\text{Maximum Stress}} = \frac{54.1\ \text{MPa}}{1.65\ \text{MPa}} = 32.78 \text{ -------(1)}$$

At 10-degree angle, 1N load was applied on to prosthetic in the simulation.as a result the load carrying capacity will be 63.74 N. which depicts its ability to support such a load without failure. The minute bending or stretching of prosthetic was seen in the middle phalanx region indicating the maximum deformation of 0.01mm. it was noted that 0.36MPa was the maximum stress in the distal phalanx indicating the clear indicative stress experienced by the prosthetic. A consistent load carrying capacity of 64.20N demonstrating stability in the prosthetic's load bearing capacities at 20 degree. At 40 and 50-degree angles there was a considerable increase that is 96.10N and 156.90N, respectively. This demonstrates a progressive improvement in load support with the raise in angle of distal tip, hence promising the performing capacities of prosthetics.

4. MECHANICAL TEST PROCEDURE

The body powered prosthetic finger is tested with customized test rig which mimic the natural movement of human finger the applied load is transferred through internally connected links of the test rig. The designed prosthetic finger is interlinked with test rig, the load transfer from test rig to prosthetic is effectively occurred through the link, the tip of prosthetic finger is place on external curved slot having different angular position. The load transferred creates the movement such as bending the finger inward towards the palm with the fingertip the mechanical test used to determine the manual activities of the finger. A specialized test rig was developed to replicate these conditions on the partial body-powered finger prosthetic, as illustrated in Figure 7. The experimental apparatus interlinked with the prosthetic finger, adopting a load cell to capture real-time force developed during testing. A screw-driven mechanism was used to apply progressive load increments, while a guide rod positioned within a curved slot controlled the prosthetic's flexion angle.

The testing protocol proceeded as follows:

1. The prosthetic fingertip was connected to the pressure rod and ensure the rigidity, the rod's position varied for very trials for specific flexion angles.
2. Load was systematically increased via the screw mechanism until the prosthetic reached the point of static failure.
3. The comprehensive failure data and performance metrics from these trials are detailed across table 3 to table 7.

Table 3: Experimental Trial 1 Results of testing the prosthetic finger in the Test Rig.

Sl No.	Elevation of Distal Tip (Degrees)	Failure Load (N)
1	10	70.63
2	20	76.49
3	40	113.75
4	50	184.36

The experimental trial one results of testing the failure load of a partial finger prosthetic device at different elevations of the distal tip. The testing protocol involved locking the prosthetic's distal end at specific angles and applying progressive pressure until the unit collapsed. We observed a direct correlation between geometry and strength: as the distal tip angle increased, so did the failure load. At a shallow 10-degree inclination, the device failed at 70.63 N, inching up to 76.49 N at 20 degrees—proving it can sustain moderate loads even at low angles.

However, the capacity spiked significantly at higher elevations, reaching 113.75 N. This represents a substantial improvement over the 40-degree benchmark, reinforcing the conclusion that the prosthetic's load-bearing efficiency improves drastically as the distal tip angle steepens. The load carrying capacity at a higher elevation of 50 degrees went up to 184.36 N, which means that the load supporting capacity decreases with lower elevation

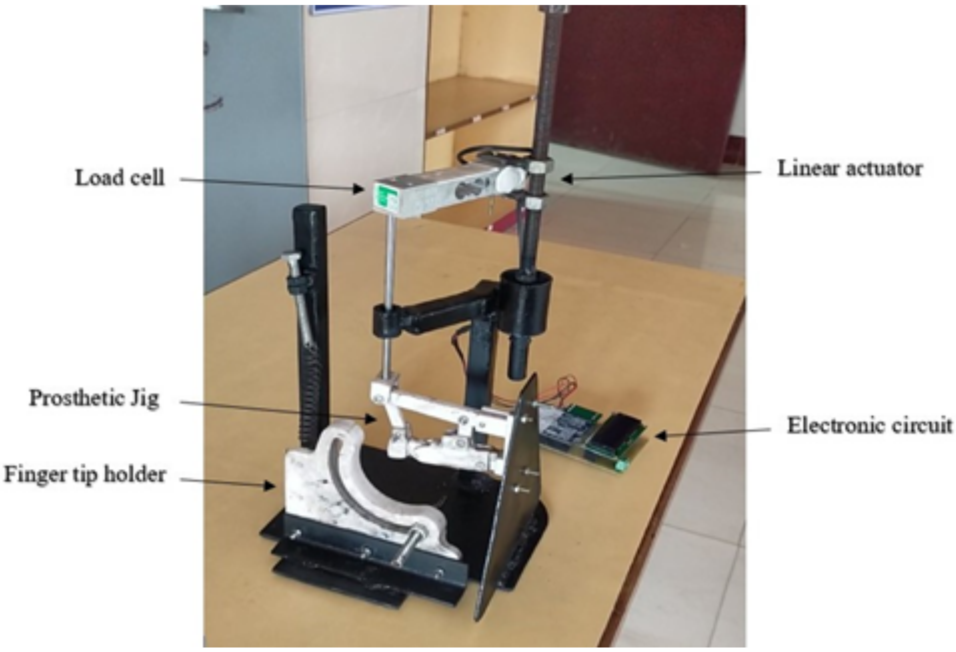

Figure 7: Specialized Test Rig [24].

Similarly, the experiment repeats for trial 2, the observation is noted at a distal tip elevation of 10 degrees, the prosthetic failed at a load of 72.59 N, and at 20 degrees, the prosthetic failed at 76.49

N. The prosthetic can sustain a moderate load when the distal tip is at a low angle. As the angle of the distal tip increased, the failure load of the prosthetic also increased. At 40 degrees of distal tip elevation, the prosthetic failed at 115.75 N. significant increase from the previous failure load.

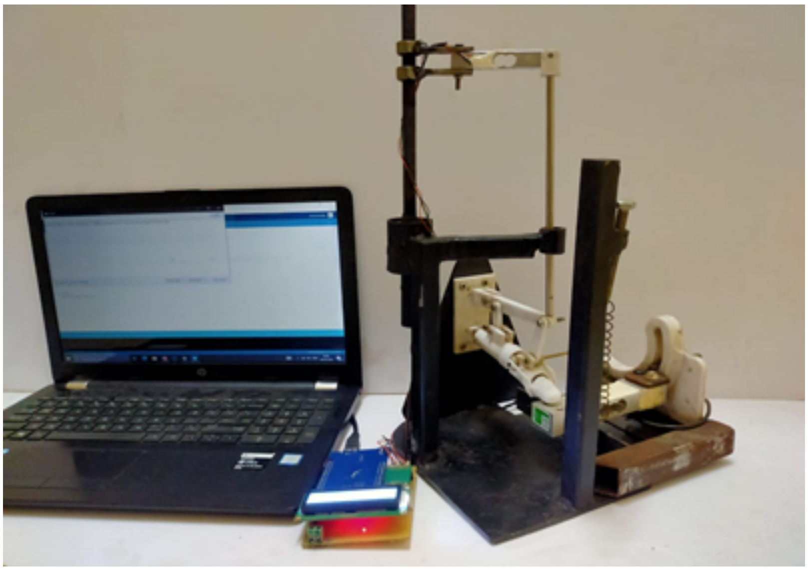

Figure 8: Finger Loaded in Gripper Setup on the Test Rig in an Experimental Setting.

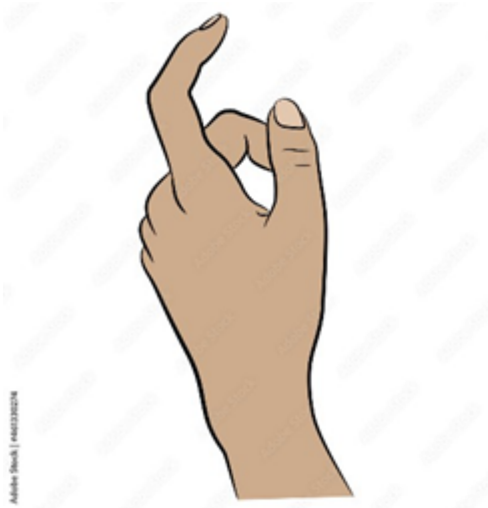

Figure 9: Bent Finger Position used during Clawing, Scratching, Strumming (Source: Adobe Stock).

Table 4: Experimental Trial 2 Results of testing the prosthetic finger in the Test Rig.

Sl No.	Elevation of Distal Tip (Degrees)	Failure Load (N)
1	10	72.59
2	20	76.49
3	40	115.75
4	50	183.44

Table 5: Experimental Trial 3 Results of testing the prosthetic finger in the Test Rig.

Sl No.	Elevation of Distal Tip (Degrees)	Failure Load (N)
1	10	72.59
2	20	73.81
3	40	112.81
4	50	183.44

This means that the higher the distal tip angle is the greater is the load bearing capacity of the prosthetic. The failure load was higher at a height of 50 degrees at 183.44 N meaning that the load bearing ability was lower at lower angles. The implications of these results are the possibility of optimization of designs in partial finger prosthetics.

The prosthetics to bear desired loads at different angles due to the relation between elevation of distal tips and the failure load. In cases when excessive loading is needed, the loads that the human hand has to handle in everyday and certain activities in a bent finger position have to be analyzed in details. There are also the data of loading tests with a spring connected. A vital and an important component of the assessment procedure was the dynamic loading test, which was to evaluate the prosthetic failure at realistic movements and loads at real-life. The prosthetic did not pass at 44.13 N which is much less than the failure loads of the tests in standing pose. This means that the prosthetic might not sustain the same loads in dynamic conditions as they are in the case of static conditions.

The reduced failure load of the dynamic test highlights the fact that additional testing and investigation of partial finger prosthetics in dynamic applications is important

Table 6: Experimental Trial 4 Results of testing the prosthetic finger in the Test Rig.

Sl No.	Elevation of Distal Tip (Degrees)	Failure Load (N)
1	10	72.59
2	20	74.55
3	40	118.7
4	50	185.4

Table 7: Experimental Trial 5 Results of testing the prosthetic finger in the Test Rig.

Sl No.	Elevation of Distal Tip (Degrees)	Failure Load (N)
1	10	72.59
2	20	75.53
3	40	117.72
4	50	185.4

Modifying the design or materials may be necessary to improve dynamic load resistance. This could involve changes to the structural integrity of the prosthetic or the materials used in its construction. Further testing and analysis are required to fully understand the prosthetic's behavior under dynamic conditions and determine the best improvements. A detailed study of maximum loading on a human finger during dynamic loading in a bent position is essential. [Figure 9], These findings can be used to perform appropriate design modifications to make the prosthetic stronger to bear dynamic load conditions

5. COMPARISON

The determination of the efficiency and dependability of the prosthetic design has a crucial dependency on the results comparison between the real-time mechanical testing of the partial finger prosthetic and CAD simulation. The ANSYS software was utilized for the CAD simulation, which precisely depicts the functioning of the prosthetic under various loads. Regardless, the real-time mechanical testing of the prosthetic necessitates the equipment and physical configuration, and it might produce variation between the simulation and real testing results. One of the predominant causes of the inconsistency in results is the calibration of the load cell employed for gauging the applied load on the prosthetic. The value of the imposed load amid each test might be different, hence causing the variation in the results betwixt real-time assessment and simulation. The prosthetic finger hinge point is also an additional aspect that induces the variances in the outcomes. The hinge point is composed of PLA material; additionally, no heat is applied to it in an experimental setting, while the revolved motions are represented as a hinge in the simulation. This variation in the representation of the hinge would result in load-carrying capacity variations at the time of load testing. The joints consist of rivets and do not have the same material as the rest of the structure. Rivet to structure clearance is also there. The fact that the two materials and clearances are not matched as they are assumed to be perfect in simulation might cause a difference between the simulated and the 3D model.

The mating of parts is correctly simulated in the CAD simulation and this minimizes the clearance as well as loss of load transfer. Conversely, 3D printing PLA hinge point is used in experimental testing, which causes a clearance diameter at the mating part, causing the greatest losses in the transfer of loads. Therefore, the prosthetic will be able to take additional load compared to the simulation. used lead screw for loading and this caused highest peaks and abrupt curve of the load on the prosthetic. This can cause the load cell to register a higher value during prosthetic failure, figure 10 shows the images of prosthetic fingers' hinges coming while experimentally testing on the Test Rig [24].

There could also be unnoticed slips in the tip of the prosthetic and the contact surface, slip between the link upon which load is applied, and the lead screw in the test rig which is not accounted, the prosthetic is made using a 3D printing process which involves a layer-by-layer building process. As a

Figure 10: PLA Based Prosthetic with Broken Hinges caused due to Load Bearing Capacity Failure during Experimental Testing Trials

result of this process, the corners and hinge points in the prosthetic are imperfect, which leads to variations in the load-carrying capacity compared to the simulation results. In CAD modeling, the material is PLA throughout. Whereas, as indicated by the research work of [22], in reality, the material of the structure consists of several layers of PLA (as a consequence of the 3D printing manufacturing process) which might have improper bonding between different layers, leading to errors observed. Even though simulation test result results give accurate results there is a need for experimental rest setup to validate the prosthetic finger by consider real time behavior during application of load on prosthetic and subjected to environmental conduction.

The difference in results between simulation and experimental test occurs due to various factors are:

1. The clearance in mating part and hinge point was differing due to the post processing operations in the load cell calibration and 3D printing process.
2. The set of assumptions and mathematical models on which the test results of the simulation are dependent might not be in line with behavior in real-world leading to the differences in the results.
3. Discretization errors, a finite number of elements and finite meshing may not be a continuous structure implying higher errors in the results of the simulation.
4. References obtained from external sources for the simulation materials property might not entirely reflect the genuine properties of the experimental material.
5. In mechanical experimental testing loading conduction may vary due to load applied by lead screw and position of load cell may vary while applying load. In simulation test all the parameter is abstract this contribute to difference in test results.

Table 8: Comparison Table for Mechanical Testing Results and Computer Simulation of the Finger Prosthetic for Trial 1.

Sl No	**Elevation of Distal Tip (Degrees)**	**Mechanical Testing Rig Failure Load (N) Trial 1**	**Computer Simulation (ANSYS) Failure Load (N)**	**Percent Difference (%)**
1	10	70.63	63.74	10.2
2	20	76.49	64.20	17.4
3	40	113.75	96.105	16.8
4	50	184.36	156.90	16

The comparison results of mechanical trial one testing with that of computer simulation of the finger prosthetic. At the 10-degrees angle of the distal fingertip, the simulation failure load of the prosthetic is approaching 63.74 N load, whereas the experimental testing failure load of the prosthetic is 70.63N. There is a significant difference of 10.2% in total. Equally, the simulation indicates the maximum load carrying capacity of the prosthetic at 20-degrees angle is up to 64.20 N load carrying. In contrast to this, the carrying capacity of the prosthetic on a real-time load is up to 76.49 N. The total result on this is an increase of 17.4 percent on a comparison of the simulation and the empirical findings. The difference between the load-carrying capacity in the simulation and the empirical result could be one of the possibilities in that there might be a difference in the precision and accuracy of the models in use during simulation and the actual use in the real environment. The simulation model might have failed to consider all the real-world conditions e.g. the variation of surface textures, frictional forces, joint stability which may have serious influence on the load carrying capability of the prosthetic. Also, variables or factors might have been present in experimental settings that are not factored in the simulation resulting in increased load-carrying capacity. The other explanation might be because at 20-degrees angle, the posture of the prosthetic arm is mainly straight and has less load than the curved beam as the load-applied position is always perpendicular to the fixing, the entire load is transferred to the prosthetic.

At 40-degree angle between the distal fingertip, the load carrying capacity of the prosthetic during real time use is up to 113.75 N. Contrarily, the simulation anticipates only a capacity of up to 96.10 N, which translates to 16.8 per cent prediction to actual performance. This difference can be explained by the fact that the position of the prosthetic (40-degrees) leads to a slight decrease in the load that is applied by the use of the prosthetic through the application of the fixture (the position of the prosthetic is 20 -degrees lower compared with the former). The fixture is one of the key factors in the load transfer process since it limits the load applied and affects the performance of the prosthetic.

The same trends are to be found in regard to the 50-degree placement of the distal tip. The theoretical load-bearing capacity is 156.90 N, and the real-life test has increased 16 percent to 184.36 N. This is caused by the fact that the prosthetic does not lie perpendicular to the 50-degrees of the fixtures and there are additional restrictive measures of how much weight that the fixtures transfer to the prosthetic

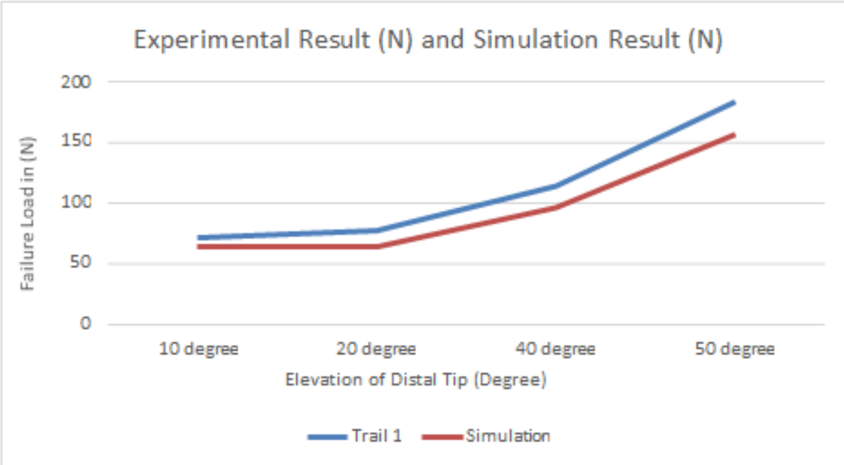

Figure 11: Graph comparing results of the experimental testing in Trail one and simulation testing. The X-axis depicts the failure load applied (N), and the Y-axis depicts the angle at the load was applied (degrees)

Similarly, Table 9, 10, 11, and 12 presents the comparison result of mechanical testing trial 2, and a similar trend is observed with the evident percentage difference when compared to simulation results of the load-bearing capacity of prosthetic. Figure 12, 13 presents this comparative study concerning different trials concerning the simulation results.

Table 9: Comparison Table for Mechanical Testing Results and Computer Simulation of the Finger Prosthetic for Trial 2.

Sl No	Elevation of Distal Tip (Degrees)	Mechanical Testing Rig Failure Load (N) Trial 2	Computer Simulation (ANSYS) Failure Load (N)	Percent Difference (%)
1	10	72.59	63.74	12.9
2	20	76.49	64.20	17.4
3	40	115.75	96.1	18.5
4	50	183.44	156.9	15.5

Table 10: Comparison Table for Mechanical Testing Results and Computer Simulation of the Finger Prosthetic for Trial 3.

Sl No	Elevation of Distal Tip (Degrees)	Mechanical Testing Rig Failure Load (N) Trial 3	Computer Simulation (ANSYS) Failure Load (N)	Percent Difference (%)
1	10	72.59	63.74	12.9
2	20	73.81	64.20	14.91
3	40	112.81	96.1	15.9
4	50	184.44	156.9	16.13

Table 11: Comparison Table for Mechanical Testing Results and Computer Simulation of the Finger Prosthetic for Trial 4.

Sl No	Elevation of Distal Tip (Degrees)	Mechanical Testing Rig Failure Load (N) Trial 4	Computer Simulation (ANSYS) Failure Load (N)	Percent Difference (%)
1	10	72.59	63.74	12.9
2	20	74.55	64.20	16.1
3	40	118.7	96.1	21
4	50	185.44	156.9	16.6

Table 12: Comparison Table for Mechanical Testing Results and Computer Simulation of the Finger Prosthetic for Trial 5.

Sl No	Elevation of Distal Tip (Degrees)	Mechanical Testing Rig Failure Load (N) Trial 5	Computer Simulation (ANSYS) Failure Load (N)	Percent Difference (%)
1	10	72.59	63.74	12.9
2	20	75.53	64.20	16.2
3	40	117.72	96.1	20.2
4	50	185.44	156.9	16.6

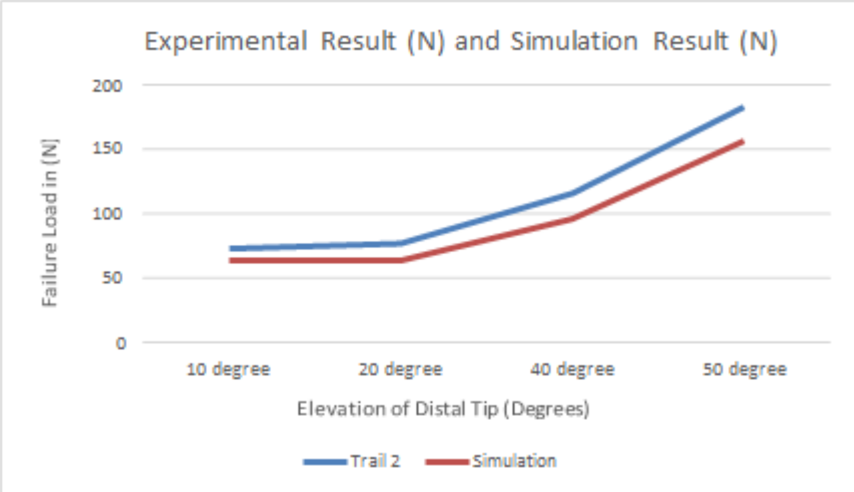

Figure 12: Graph comparing results of the experimental testing in trial two and simulation testing. The X-axis depicts the failure load applied (N), and the Y-axis depicts the angle at which the load was applied (degrees).

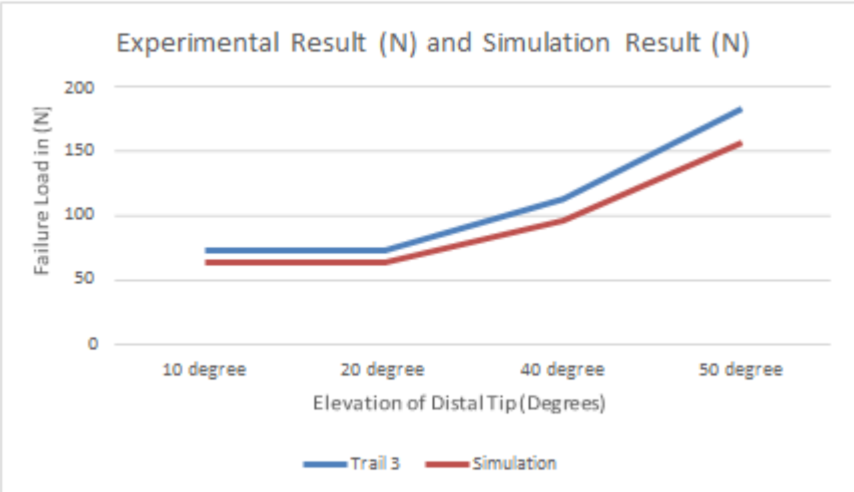

Figure 13: Graph comparing results of the experimental testing in trial three and simulation testing. The X-axis depicts the failure load applied (N), and the Y-axis depicts the angle at which the load was applied (degrees).

Research conducted by [23] highlights that holding, grasping, and seizing objects with the hand is prehension. It is noted that while the prehensile force exerted by the human hand can reach 500N, daily activities typically require only 70N. In their study, [23] conducted a structural analysis of a three-finger prosthetic hand prototype, focusing on the loading of the prosthetic finger. When positioning the finger forward, it was designated as having a 0-degree angle. Subsequently, as the finger was elevated from this initial orientation, positive degrees were attributed. Conversely, negative degrees were registered when the finger descended from this reference position, as exemplified from Figure 11 to Figure 15 for different experimented trials.

In the experimental setup, the forward-pointing orientation of the finger denotes the zero-degree position. Notably, in instances where the finger is directed downwards, as depicted in Figure 11, positive degrees are designated. This angular measurement is the angle between the distal phalanx and a plane parallel to the ground. The study conducted by [23] highlights that when the finger is directed downwards, its load-bearing capacity is high. Conversely, the load-bearing ability is notably reduced as the finger moves to a neutral zero-degree position. The results obtained in this study align with this trend, emphasizing the consistent correlation between finger orientation and load- bearing capacity as depicted in Figure 11. Upon elevating the finger from its neutral

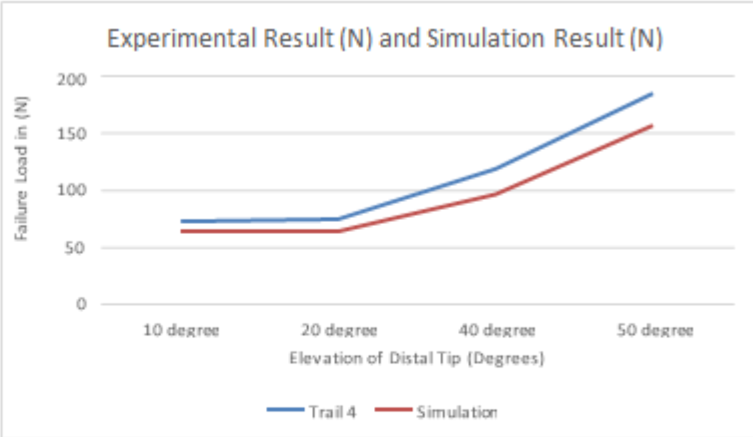

Figure 14: Graph comparing results of the experimental testing in trial four and simulation testing. The X-axis depicts the failure load applied (N), and the Y-axis depicts the angle at which the load was applied (degrees).

position, commensurate outcomes became discernible, thereby reflecting a congruity with previously observed results. This concordance substantiates the heightened credibility of the experimental apparatus. [19] conducted tests where they measured fingertip forces using the setup. The participants hung on a ledge using their fingertips, and the resultant forces were measured in anteroposterior and vertical directions. The study reports that the maximum force ranged from 350N to 576N, with anteroposterior forces ranging from 70N to 138N. It is noted that while the design under consideration in this research work can withstand anteroposterior forces, it may require more robust materials to withstand forces of 500 newtons. This change could be easily implemented by altering the construction material. However, it must be noted that the previous study recorded normal activities not exceeding 70N.

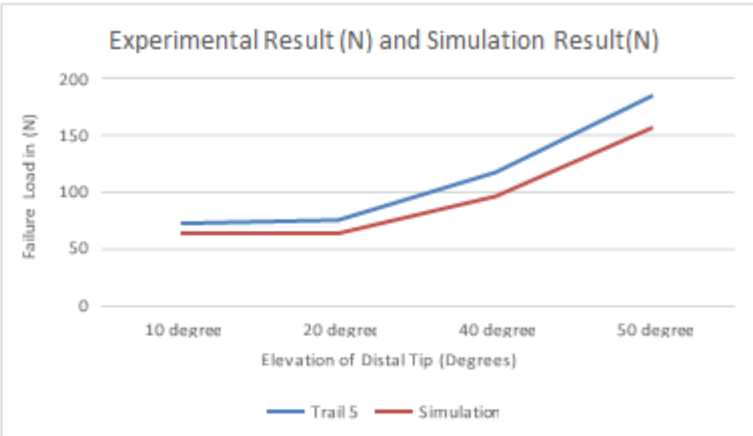

Figure 15: Graph comparing results of the experimental testing in trial five and simulation testing. The X-axis depicts the failure load applied (N), and the Y-axis depicts the angle at which the load was applied (degrees).

1. CONCLUSION

In comparison of the practical experimental results with the virtual simulation results as shown in the figure 12, 13, 14 and 15 carried out for various intermittent positions and at different trials, this study highlights the constant discrepancy in the results which is also confirmed by referring to the figures mentioned. Comparison of Virtual simulation test results, Real world test results with percentage difference of those to test respectively at each trial id depicted and shown in the form of Graph at respective Figures. These figures clearly shows that Value of the different degree of position of the prosthetic finger is directly proportional to its load carrying capacity at respective position, however the percentage difference between the virtual simulation and the real world test results value is due to the fact that in real world testing the Prosthesis will have Clearance in the hinge points, variation in the material thickness, some abnormal errors in 3D printing, variation in the working temperature, real world load slippage and friction difference at the finger gripping point in which all of these error causing parameters will have either a ideal values or user fed fixed values in the simulation test readings.

The quantification of the research outcome demonstrates a number of important findings.

1. Difference between experimental and simulated outcomes.
2. The error relativity assuring a praiseworthy conformity to accuracy.
3. Major differences were related to measurement differences of degrees and non-existence of rivets in ANSYS simulations.
4. Differences in the material properties and working temperatures between the ANSYS libraries and the materials.

Various differences identified on the qualitative and quantitative parameters are reported in this research paper with inclusion of all relative conclusions. The content in this paper highlights the optimization requirements in material, design and structural setup which can be utilized by various researchers who are working on development of Prosthetic limb. 15.85% is the overall difference between the values of real-world experimentation and virtual simulation. This value shows the accuracy level of the study carried out in this research paper on the Load carrying capacity of the finger prosthesis between Simulation test and Real-world Experimentation.

2. DISCLOSURE STATEMENT

The paper does not have any conflict of interest from any of the publication, institution and organization which affected this research results in presentation.

3. FUNDING

This research received no specific grant from any funding agency in the public, commercial, or not-for-profit sectors

REFERENCES

[1] R. Sinha, W. J. Van Den Heuvel, and P. Arokiasamy, “Factors affecting quality of life in lower limb amputees,” *Prosthetics and orthotics international*, vol. 35, no. 1, pp. 90–96, 2011.

[2] H. Belon and D. Vigoda, “Emotional adaptation to limb loss,” *Physical medicine and rehabilita- tion clinics of North America*, vol. 25, pp. 53–74, 02 2014.

[3] A. Chan, E. Kwok, and P. Bhuanantanondh, “Performance assessment of upper limb myoelectric prostheses using a programmable assessment platform,” *Journal of Medical and Biological Engineering*, vol. 32, 01 2012.

[4] A. J. Thurston, “Paré and prosthetics: the early history of artificial limbs,” *ANZ journal of surgery*, vol. 77, no. 12, pp. 1114–1119, 2007.

[5] I. N. Gaiser, C. Pylatiuk, S. Schulz, A. Kargov, R. Oberle, and T. Werner, “The fluidhand iii: A multifunctional prosthetic hand,” *JPO: Journal of Prosthetics and Orthotics*, vol. 21, no. 2, pp. 91–96, 2009.

[6] D. Sim, Y. Baek, M. Cho, S. Park, A. S. Sagar, and H. S. Kim, “Low-latency haptic open glove for immersive virtual reality interaction,” *Sensors*, vol. 21, no. 11, p. 3682, 2021.

[7] E. Difonzo, G. Zappatore, G. Mantriota, and G. Reina, “Advances in finger and partial hand prosthetic mechanisms,” *Robotics*, vol. 9, no. 4, 2020. [Online]. Available: https://www.mdpi.com/2218-6581/9/4/80

[8] S. A. Ovadia and M. Askari, “Upper extremity amputations and prosthetics,” in *Seminars in plastic surgery*, vol. 29, no. 01. Thieme Medical Publishers, 2015, pp. 055–061.

[9] P. C. Liacouras, D. Sahajwalla, M. D. Beachler, T. Sleeman, V. B. Ho, and J. P. Lichtenberger, “Using computed tomography and 3d printing to construct custom prosthetics attachments and devices,” *3D printing in medicine*, vol. 3, pp. 1–7, 2017.

[10] R. Mio, M. Sanchez, Q. Valverde, J. Lara, and F. Rumiche, “Mechanical testing methods for body-powered upper-limb prostheses: A case study,” *Advances in Science, Technology and Engineering Systems Journal*, vol. 4, no. 5, pp. 61–68, 2019.

[11] J. S. Cuellar, G. Smit, P. Breedveld, A. A. Zadpoor, and D. Plettenburg, “Functional evaluation of a non-assembly 3d-printed hand prosthesis,” *Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine*, vol. 233, no. 11, pp. 1122–1131, 2019.

[12] J. Ten Kate, G. Smit, and P. Breedveld, “3d-printed upper limb prostheses: a review,” *Disability and Rehabilitation: Assistive Technology*, vol. 12, no. 3, pp. 300–314, 2017.

[13] K. Butz, G. Merrell, and E. Nauman, “A biomechanical analysis of finger joint forces and stresses developed during common daily activities,” *Computer methods in biomechanics and biomedical engineering*, vol. 15, pp. 131–40, 06 2011.

[14] A. Kargov, C. Pylatiuk, J. Martin, S. Schulz, and L. Döderlein, “A comparison of the grip force distribution in natural hands and in prosthetic hands,” *Disability and Rehabilitation*, vol. 26, no. 12, pp. 705–711, 2004.

[15] X. Iruretagoiena-Urbieta, J. De la Fuente-Ortiz de Zarate, M. Blasi, F. Obradó-Carriedo, A. Ormazabal-Aristegi, and E. S. Rodríguez-López, “Grip force measurement as a complement to high-resolution ultrasound in the diagnosis and follow-up of a2 and a4 finger pulley injuries,” *Diagnostics*, vol. 10,

no. 4, p. 206, 2020.

[16] G. Smit and D. H. Plettenburg, "Efficiency of voluntary closing hand and hook prostheses," *Prosthetics and Orthotics international*, vol. 34, no. 4, pp. 411–427, 2010.

[17] D. Saul, G. Steinmetz, W. Lehmann, and A. F. Schilling, "Determinants for success in climbing: A systematic review," *Journal of Exercise Science & Fitness*, vol. 17, no. 3, pp. 91–100, 2019.

[18] B. Mugleston and C. McMullen, "Musculoskeletal injuries in climbers," *Current Physical Medicine and Rehabilitation Reports*, vol. 7, pp. 179–185, 2019.

[19] A. Amca, L. Vigouroux, S. Arıtan, and E. Berton, "Effect of hold depth and grip technique on maximal finger forces in rock climbing," *Journal of sports sciences*, vol. 30, pp. 669–77, 02 2012.

[20] E. N. Kubiak, J. A. Klugman, and J. A. Bosco, "Hand injuries in rock climbers," *Bulletin-Hospital for Joint Diseases New York*, vol. 64, no. 3/4, p. 172, 2006.

[21] F. Noé, F. Quaine, and L. Martin, "Influence of steep gradient supporting walls in rock climbing: biomechanical analysis," *Gait & posture*, vol. 13, no. 2, pp. 86–94, 2001.

[22] F. Johansson, "Optimizing fused filament fabrication 3d printing for durability: Tensile properties and layer bonding," 2016.

[23] J. A. Leal-Naranjo, C. R. Torres-San Miguel, and M. Faraón, "Structural numerical analysis of a three fingers prosthetic hand prototype," *International Journal of Physical Sciences*, vol. 8, no. 13, pp. 526–536, 2013.

[24] Madhu Mohan R, Subhaschandra Kattimani, Poornesh Kumar Koorata and Girisha C, "Design of novel test rig for prosthetic finger distal interphalangeal and phalanx strengths", *Prosthetics and orthotics International,* 2025 Apr 1;49(2):214--219. doi: 10.1097/PXR.0000000000000398.