Design of a Multi-Functional End Effector Prosthetic

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Abstract* – *This study researches the effectiveness of a prosthetic designed for using hand tools for bicycle repair. This is an underdeveloped category of prosthetics and would improve the lives of amputees and enable their independence. While there is currently a large array of activity specific prosthetics, there is little information on prosthetics for using hand tools or doing repair work. This study explores this concept by designing and creating a prototype prosthetic specifically for the use of hand tools with the use of Solidworks, 3D printing, and a simulated test procedure. This will test the prosthetic's ability to use common tools in bicycle repair for common repairs and installations to their required torques. The final prototype could hold over 2 kg and had a 4.6 kg grip strength. A SolidWorks simulation found that the prosthetic could apply more than 20 Nm torque upon a 40 mm screwdriver and 20 Nm to an M10 spanner, making it more than sufficient for high strength tasks, but was insufficient in dexterity.

Keywords: Amputee, prosthetics, osseointegration, 3D printing.

I. INTRODUCTION

Prosthetics have existed for centuries to replace human limbs, and most of these functional prosthetics work purely mechanically [1]. Despite the abundance of highly capable electrical and hybrid prosthetics now available, the most widely used upper-limb prosthetic is still the body-powered split hook [2]. This is due to various factors, including but not limited to low weight, high durability, and practicality [2]. The most common causes of amputation are diabetes, trauma, infections, and tumours [3–6] and the most common type of upper-limb amputation is the transhumeral, closely followed by the transradial [3,5,7,8]. Considering the large range of causes and outcomes for amputation, designing this prosthetic for transradial amputees opens opportunities for a large number of users. Additionally, transradial amputees are the most likely upper limb amputee to return to work [9]. However, the human hand is an incredibly complicated body part, and constructing the 'perfect' human hand prosthetic is practically a 'holy grail' [5]. This is because of the complexity of the natural human hand, and task-specific prosthetics have been needed where a traditional or electronic prosthetic would be insufficient.

To outline the goals for this dissertation, the shortcomings of existing prosthetics will define what this project will achieve. Prosthetics have been shown to cause discomfort [10] and most current commercial hand prosthetics provide the user with insufficient dexterity and functionality due to the highly restricted number of prehensile patterns that may be achieved [11]. As a result of this, close to half of all upper limb amputees forego prosthetic use, additionally due to dissatisfaction, discomfort, cost, or poor performance of the prosthetic [12]. While activity specific prosthetics are by no means common, there is still a lack of manufacturing or repairing focussed prosthetics, which could affect the independence and confidence of people living with upper limb loss.

This project intends to create a new activity-specific prosthetic for a bike mechanic. The key goals include using a screwdriver and spanner for various common types of bicycle repairs. Comparable prosthetics are those made primarily for task applications, either specific jobs [13] or non-specific [14], and for physical activities [15,16]. Where some activity specific prosthetics can grab onto objects such as a bike handle or a fishing pole, this can grasp and control commonplace tools. The goals for this dissertation are as follows:

- Design a unique prosthetic prototype for the use of hand tools, with the main use case for bike repair.
- Test the prototypes on their ability to apply appropriate torques to fasteners outlined by manufacturers. This includes the most common tools for a bike mechanic: screwdrivers and spanners.
- The prosthetic's primary source of energy is the user, and no external assistance is required to manipulate the prosthetic or change the tool in use during testing.
- The prosthetic is constructed from reused or reclaimed materials. Some materials and components are common household or bike parts, such as nuts, bolts, and pulley parts.

II. DESIGN PROCESS

Initially, the causes and types of upper limb amputation and prosthetics were heavily studied. The design for this study focuses on the transradial amputation, one of the most common types of upper limb amputation [3,5,7,8]. A transradial amputation means the arm is amputated below the elbow and above the wrist with the elbow joint still intact. Each amputation will leave a "stump" on the person's residual limb that varies in shape and size after amputation. This makes designing a traditional socket-based prosthetic very difficult because each person's socket needs to be custom-designed for each user to ensure compatibility and comfort.

This is where osseointegrated prosthetics can help. Osseointegration is a technique to implant structures into the bones of amputees upon their residual limb to extend their bone profile. This osseointegrated extension connects to prosthetics and comes in many different forms, such as the 'Intraosseous Transcutaneous Amputation Prosthesis' (ITAP), 'Osseointegrated Prostheses for the Rehabilitation of Amputees' (OPRA) or 'The Compress Device'. There are many benefits of osseointegration prosthetics, and they are not without their shortcomings, but most users say the benefits outweigh the drawbacks [15]. Osseointegrated prosthetics do not cause discomfort or sweat at the connection area, allow for increased mobility and strength as the prosthetic is directly attached to the bone, and enhance osseoperception compared to socket prosthetics [17].

The selected audience for this prosthetic is people living with limb loss who want to regain independence. It is known that amputees benefit from exercising their body symmetrically to maintain a healthy balance of the muscles [18,19], and while prosthetics exist for general exercise, weightlifting, or riding bikes, there aren't many prosthetic products made specifically for bicycle repair [or any such repair, for that matter]. Because of this, the prosthetic's goals do not include performing activities of daily living (ADL), as is commonly expected of prosthetics.

While some of the biggest prosthetic manufacturers, such as Ottobock, Steeper, and London Prosthetics, provide a vast array of upper limb prosthetic options, many rely on expensive myoelectric control with motorised movement. While these work for some people, they are not suitable for many people or tasks, such as people with muscle atrophy [20] or performing quick and precise tasks. The advantages of body-powered prosthetics can be seen in the case of Austin Anderson's prosthetic, who works with a chainsaw prosthetic [21], or Manami Ito, a single-handed swimmer, who plays the violin [22], where both of which use body-powered prosthetics, and can perform critical and/or high precision movements. Currently, there are minimal options for activity-specific prosthetics outside of a bespoke product.

Before the designing phase, an extensive research phase was conducted to explore how to test the prosthetic authentically without an amputee volunteer. 'Prosthetic simulators' (Fig. 1) are a type of training device for people to practice using prosthetics before amputation, or to research them without an amputee [23].

Figure 1 - Prosthetic simulator [24].

The use of a prosthetic simulator also allows a deeper understanding of the customer's needs, as it allows the user to experience the benefits and shortcomings of the product and to correctly make trade-offs during design development. Therefore, for this project, a similar design was used, where a handle was chosen as a simple and effective way of creating a prosthetic simulator. The design simply uses the user's hand to hold the prosthetic so they can control it as a person with an amputation would. The handle is simply a part of the practical simulation experience and is not a part of the study because the final design would be attached via osseointegration.

A Product Design Specification (PDS) was developed to help focus the design process (Table 1) using references to anthropomorphic dimensions and safety guidelines.

Table 1 – PDS table.

Parameter	Range	Justification	Importance
Maximum length	150-230 mm	Matching the size of a human hand [25] is essential for comfort.	5
Weight	0.3-0.6 kg	One of the largest criticisms of prosthetics is their weight [26], and the average hand weighs only about 0.4 kg [27].	4
Actuation force	Weight of held object +/- 25%	A low actuation force will enable precise work and movements, without worrying about resistance.	3
Minimum lifting capacity	1.4 kg	1.4 kg is heavier than most single-hand manual tools, the maximum by CCOHS [28].	3
Max gripping force	47 kg	Average grip force for adult males [29,30].	3
Toolset	2 tools	Bike repairs need spanners and screwdrivers minimum.	5
Time to change tool	30s	Prosthetic can change tools within 30 seconds.	5
Maximum grasp size	43-53 mm	These dimensions are precision and power grab grasp sizes [31].	4
Range of motion	180-360°	180° each turn on fasteners for effective tool use.	5
Operating temperatures	-10-40 °C	Prosthetic must withstand year-round temperatures [32].	1
Materials	PLA, PETG	3D printing is essential for rapid prototyping.	3

Unlike standard body-powered prosthetics, this study did not use a full-body harness to control the motion of the prosthetic hand. Instead, a similar and simpler mechanism was used. A strap attaches to the user's upper arm, which has a pulley attached to it to pull the prosthetic's fingers down. For a final product, a full body harness must be used to enable full arm control, as the simplistic mechanism used here restricts the user's elbow movement range to that of the prosthetic.

The three primary inspirations (Fig. 2) are activity specific prosthetics with rather different methods of grabbing objects which could be repurposed for tools. Starting from the left: the first method (Fig. 2A) uses a strap, somewhat like a ratchet strap, to wrap around weightlifting bars and handles tightly. This prosthetic has the benefit of strength but at the cost of versatility and convenience for changing grasp. The next design (Fig. 2B) uses a helical curve to grab around hockey sticks or cricket bats. However, this novel design would likely be very difficult to manipulate the tools, as it lacks a feasible way of gripping the tools while applying torque because it holds objects by using their weight as a counterbalance, which is not practical in the use case of a tool. The final design is a more traditional one, which has the most flexible mechanism for grabbing various shapes, but with a limited minimum grasp. A dedicated design for holding typical tool shapes with the third prosthetic design (Fig. 2C) was deemed to be the most practical method for this study by a SWOT analysis, especially because it has room for developing a custom gripper design to optimise holding an array of tools. Other designs like a multitool and a torque wrench connected to a prosthetic wrist were considered but fell short due to not being significantly new or unique designs. Other novel concepts involved a mechanical iris diaphragm or compliant mechanisms to grab the tools, but these were rejected due to complexity or lack of strength.

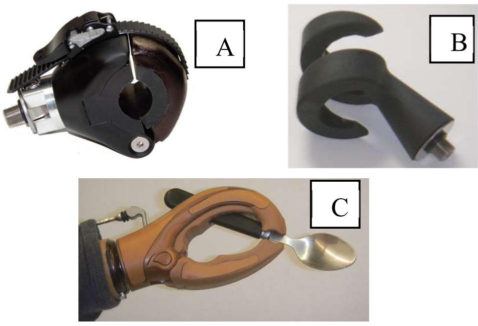

Figure 2 - Various types of activity specific prosthetics [33].

Osseointegration prosthetics attach to the user via a threaded titanium implant, the 'Abutment screw', in Fig. 3. The bone grows around the fixture, making for true osseointegration. This gives the user a direct sense of osseoperception and provides a secure connection between the prosthetic and the user. This study's designs assume the threaded titanium abutment screw will be where this study's final prosthetic design will attach to within either the ulna or radial bone of the forearm [34].

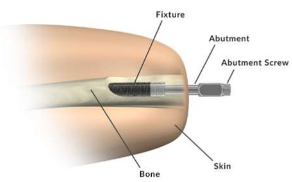

Figure 3 - OPRA osseointegration components [35].

III. CAD PROCESS.

The initial CAD phase began with a very primitive design to realise the prosthetic simulator concept. The first design, V1 (Fig. 4), was printed and was immediately noticed to lack a suitable connection for the springs, causing them to be unreliable and fail to hold the prosthetic fingers. V2 improved the spring connections by fixing them to the lower and upper fingers directly via small hooks. V2 also included holes to connect the pulley system to the fingers for direct control, and another hole on the holder to thread the pulley through to feed the movement through easily. The flat bottom finger was a temporary experimental model while checking how the mechanisms would connect. The location of the pulley on the top fingers meant the prosthetic was a voluntary opening one, where the user pulls the pulley to open the jaw and relies on the spring to pull the fingers together. This was suboptimal because, without designing an adjustable spring tension mechanism, the fingers wouldn't be reliable enough to hold various sized tools and likely not turn them with sufficient torque. This was fixed by moving the pulley connection beyond the pivot point of the fingers, so that when the user pulls the extension upwards, towards their body, the fingers close together, like a scissor mechanism. This allows the user to vary the force they apply to the prosthetic and allows for a direct connection with the tool in the hand.

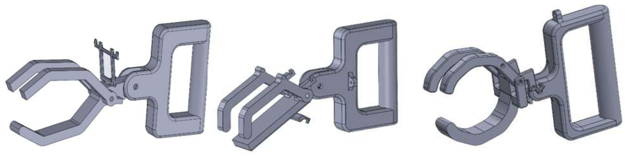

Figure 4 - Solidworks models of CAD versions V1, V2 and V3.

IV. PROTOTYPING

All prototypes were designed in Solidworks 2023 and printed on an FDM (Fused Deposition Modelling) based Artillery Genius 3D printer (filament extrusion machine, outlined by ISO/ASTM 52900) using a 0.4 mm diameter nozzle with Elegoo 1.75 mm PLA.

Prototype V1

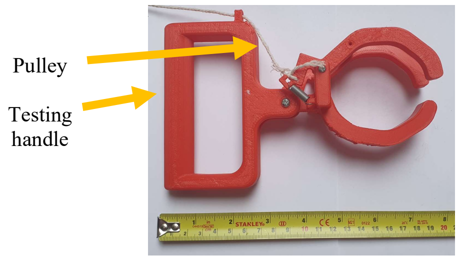

Figure 5 - Prototype V1.

Print time and settings: Sliced in UltiMaker Cura, settings: 200 °C printing temperature, 60 °C bed temperature, top/bottom layers at 1.12 mm thickness, 0.28 mm layer height, 100 m/s print speed, support enabled at 45° overhang, 10% cubic infill with 0.28 mm layer height. The total print time was 5 hours 11 minutes, and used 69 g of PLA, including supports. The constructed prototype weighed 69 g with all components. It had a grip range of 10-57 mm and can carry up to 1.4 kg.

Table 2 – Advantages and disadvantages of prototype V1.

Advantages	Disadvantages
Can withstand high loads – can hold 276 g.	Fingers have low friction.
Easy to use.	No holes dimensioned to ISO standards.
Has a large grasp range – 57 mm maximum.	The connection piece didn't fit - fingers splay.
Lightweight – only 68 g.	No bushing around the pulley.
1.7 kg grip strength.	Bolts and nuts aren't flush with prosthetic.
	Only one bolt on the handle.

This prototype proved successful, with a summary in Table 2, it was able to hold various objects much heavier than itself, despite the pulley only being connected to one finger. Above a certain velocity, objects tended to fall out, so an increase in surface roughness was required.

The prosthetic works by attaching the pulley to the user's bicep by a simple strap. By extending the elbow, the pulley is pulled taut from the prosthetic, pulling the upper fingers of the prosthetic downwards using a scissor-like mechanism, and closing the hand. By contracting your elbow, the springs pull the hand open again to its default state. This makes the prosthetic voluntary closing.

Another problem with the original prototype is that the connection piece between the fingers didn't fit right, causing the fingers to splay away from each other. Without the connection piece holding the fingers apart by a fixed distance, the fingers would splay away from each other because the spring was not lined up parallel to the finger. Another problem was the springs were not connected securely or safely. This is because the loops that the springs connect to weren't rounded or smooth, so both were fixed in the next iteration, allowing the springs to stay on securely. Bolt hole tolerances were not sized to standard, causing holes to be too large or too small to securely accommodate the screws. The nuts that hold the upper fingers screws into the main body weren't flush with the plastic around them. This meant that a spanner was needed at an awkwardly small area to tighten the nuts onto the fingers. Prototype V1 also had a single bolt holding the hand to the holder because originally there was going to be a wrist pivot point. This was considered and planned in some CAD designs but was ultimately decided to be beyond the scope of this study, so a fixed mounting point was used.

Prototype V2

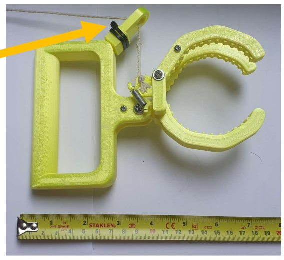

Figure 6 - Prototype V2.

Figure 7 - Weak layer orientation.

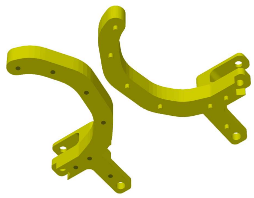

Figure 8 - The 2 halves of the lower finger, featuring pegs and slots generated in PrusaSlicer.

Print time and settings: Sliced in PrusaSlicer 2.6.1, settings: printing temperature, bed temperature, top/bottom layers at mm thickness, mm layer height, print speed, support enabled at 45° overhang, 20% adaptive cubic infill with 0.2 mm layer height. The total print time was 10 hours and 46 minutes and used 96 g of PLA, including supports. The constructed prototype weighed 89 g without the arm strap and weighed 39 g without the handle or arm strap. It had a grip range of 5-70 mm and can carry up to 1.6 kg.

Table 3 – Advantages and disadvantages of prototype V2.

Advantages	Disadvantages
Added a second bolt to the handle, preventing the hand from moving, but still wiggled a little.	Poor construction –critical layer separation.
Spring connectors are lined up with each other.	The bushing slot cracked open (temporary fix in Fig. 6)
Bumps are added to fingers to increase friction.	The hand was printed parallel to the print bed, making thin parts weaker.
All holes are sized to ISO standards.	1.2 kg grip strength.
Nuts slot into prosthetic so no spanner is needed for construction.	

Prototype V2 set out solely to improve on the problems from prototype V1. V2 added an M4 bolt and nut in addition to the M6 (Fig. 6) so the hand didn't turn towards the user when being pulled, but this connection still had some wiggle room. The spring connections were smooth, in line with each other, and more secure; the connector between the fingers fit perfectly and so the fingers no longer splayed; bumps were added to the contact surface of the fingers to help improve grasping ability; and all holes for the screws and nuts were sized appropriately for ISO 724. The distance between the finger pivot and the pulley was reduced because it was a potential stress point, but this prevented the hand from closing sufficiently. As a temporary repair for practical testing, a longer bushing holder was created to extend the actuation point on the handle so that the hand could close sufficiently. The original hole for the bushing was also enlarged, but this hole was too small and cracked open, causing the bushing to fall out. The slot for the bushing was too small for possibly two reasons: the PLA was printed in such a way that it expanded and became too tight for the bushing; or the orientation of the print, and perhaps supports within the hole, caused it to be misaligned and thus had too many imperfections for the smooth and round bushing.

The first attempt at printing prototype V2 failed to print one of the spring connections, like in prototype V1. This was likely due to the support structure connecting to the spring connection too firmly and pulling them off. The first print also had a weak layer orientation around the fingers (Fig. 7) due to printing the thin structure perpendicular to the filament layers, making them very weak. This was fixed by printing the lower finger parallel to the printer bed and splitting it into two halves and joining them together with pegs (Fig. 8) and glue. Glue is avoided at all costs in this project to improve repairability but is used to hold the fingers' halves together in this case as the piece is designed to be one and does not feature any components within. This was performed using PrusaSlicer 2.6.1, instead of UltiMaker's Cura, as it has this 'cutting' feature. V2's handle puts the nuts parallel to the plastic so that fewer metal pieces protrude, and the bolt is flush with the plastic. This also means a spanner isn't needed to secure the screw because the plastic holds the nut in place. A similar design is featured around the nuts for the finger's screws, fixing the problem with V1.

Prototype 3

Figure 9 – Prototype V3.

Figure 10 - 3D printed part with bushing implanted.

Print time and settings: Sliced in PrusaSlicer 2.6.1, settings: 200 ºC printing temperature, 60 ºC bed temperature, four top/bottom layers at 0.8 mm thickness, 0.2 mm layer height, 60.0 mm/s print speed, support enabled at 45º overhang, 20% adaptive cubic infill with 0.16 mm layer height. The total print time was 10 hours and 39 minutes, and used 97 g of PLA, including supports.

Table 4 - Cost breakdown of prototype V3.

Components	Quantity	Cost (GBP)
Adjustable arm strap.	1	1.80
Bronze bushing.	1	0.66
Aluminium crimps.	5	0.50
M4 screws, nuts, and washers.	15	0.17
Extension springs.	2	0.10
Stainless steel cable.	600 mm	0.04
M3 crosshead screw.	2	0.02

The final prototype cost approximately £1.84 in filament, £1.08 in electricity and £3.27 in components, making prototype V3 cost £6.17 in total, including the testing handle and arm strap. The constructed prototype weighed 98 g with all components, excluding the arm strap, and weighed 47 g without the handle and arm strap. It had a grip range of 5-40 mm and can carry over 2 kg.

Table 5 – Advantages and disadvantages of prototype V3.

Advantages	Disadvantages
The hand is connected securely by 3 bolts.	Difficult to use tools due to lack of wrist pivot.
Replaced bumps on fingers for high high-friction tape.	Fingers are fixed together, reducing flexibility.
Uses stainless steel cable and aluminium cable crimps.	Smaller grip range, only up to 40 mm.
The bushing is securely attached.	The bushing cannot be removed.
4.6 kg grip strength.	

Prototype V3 removed the M6 bolt in the handle and used only M4 screws for the handle in a triangle form, for maximum strength and minimising types of parts, and used M3 screws for the finger's pivots. The three screws on the handle support the hand much better than the two screws did. The fingers had the bumps removed for a high-friction tape. The tape on the fingers is an anti-slip tape which makes it perfect for applying torque on this prosthetic. The tape is highly adhesive and replaceable. V3 used a 1 mm diameter stainless steel cable for the pulley, fixed together at ends with aluminium cable crimps. To fix the bushing in place, V3 has the bushing implanted by pausing the print and placing the bushing inside mid-print (Fig. 10). This made the part very strong and was able to friction fit very tightly into the handle for the prosthetic.

V. TEST PROCEDURE

Table 6 – Bicycle components and their required torques.

Bike component	Info (bolt size, manufacturer, variant, etc)	Torque (Nm)	Reference
Seat post	Binder bolt, M4	1.9-3.9	[36–38]
	Dual-bolt clamps, M5	6.8	[39]
	Binder bolt, M6	9.0-15.6	[36–39]
	Binder bolt, M8	17.6-19.6	[36]
Brake levers		2.4-10	[36,38]
Brake calliper mount to frame	Side-pull, dual-pivot, centre-pull	7.7-10	[38]
Brake pad	Calliper brake M5, V-brake M6	5-9.8	[36,38]

Brake cable pinch bolt		4.5-9	[38]
Brake arm/handlebars		4-7.9	[37,39]
Stem handlebar binder 4 bolt faceplate	Various (excluding Control Tech)	6-8.8	[38]
Stem, stem clamp		4.5-10	[37,39]
Stem binder bolt	Threadless	5-10.1	[38]
Handlebar/stem		5-6	[37]
Chain ring bolt, aluminium		5-10	[38]
Chainring bolts, alloy		9.8	[39]
Shift lever/handlebars		2.9-10	[37–39]
Water bottle cage	M5	2.9-4.9	[36,39]

Originally, the prosthetic was going to be tested on its practical ability to manipulate fasteners and then the clamping forces of those fasteners would be calculated according to ASTM F606-09 to test the torsional strength. This was later evaluated to be beyond the scope of the study, so a static simulation study of the final prototype was used instead to find the torques applied in SolidWorks 2023. To achieve a pass, the prosthetic must be able to manipulate fasteners to a specified torque outlined in Table 5.

A 20 Nm torque was applied to a generic 40 mm diameter hand tool, closed around the prosthetic's fingers. A 40 mm diameter model was chosen as this is the average human grasp diameter [25], and 20 Nm because this is the upper limit of torques in the objectives for the prosthetic. PLA material properties were taken from online sources [40–43] and were used to simulate the prosthetic.

VI. Simulation Setup

The material properties used in the simulations (Table 7) are a close match to the prototype, using 25% grid infill density [42]. After establishing mesh independence with the setup in Fig. 10, the simulation was used to find the limits with a stronger structure because of the material property differences. The fasteners used in the simulations are from the SolidWorks toolbox and did not feature threads.

Table 7 – PLA material properties used in simulations.

Yield strength	Tensile strength	Young's modulus	Poisson ratio	Reference
16.59 Mpa	25.62 Mpa	0.835 Gpa	0.33	[42]

An initial test (Fig. 11) was run with the mesh parameters in Table 8. The results show that the prosthetic was weakest around the wrist connection, but while the maximum stress was not near the yield strength of PLA, it was an area of high priority, being so near the osseointegration's abutment. Thus, this area was refined with a stronger body (Fig. 12) before pursuing mesh independence.

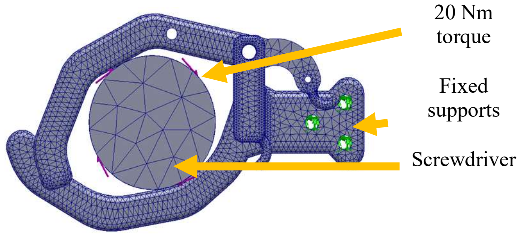

Figure 10 – Initial mesh on prototype V3.

Figure 11 – 20 Nm torque test on prototype V3.

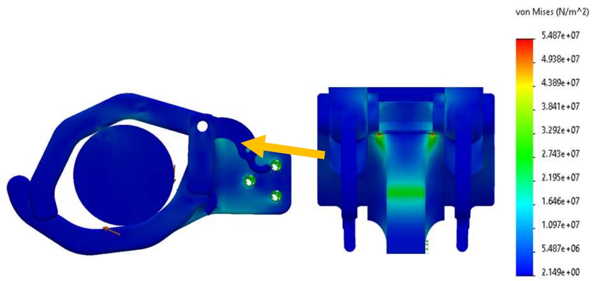

Figure 12 – 20 Nm torque test on updated prototype V3.

For the simulation, a mesh independence study was initially performed to determine appropriate mesh parameters for the results. All prosthetic components had their material properties defined from PLA material properties (Table 8) found online [40–43] and all fasteners used an alloy steel profile both found in SolidWorks. The screwdriver was assigned SolidWorks' AISI A2 tool steel, with the missing yield strength value supplemented by MatWeb [44]. All studies used a blended curvature-based mesh for more options for mesh parameters.

Table 8 –.Mesh parameters in the initial study.

Maximum mesh size	Minimum mesh size	Min number of elements in a circle	Mesh scale growth ratio
9 mm	0.45 mm	8	1.4

The handle was excluded from the simulations because it won't exist in a final osseointegrated prosthetic. Because of this, the fixed supports were assigned to the mounting holes, to remain true to the practical model.

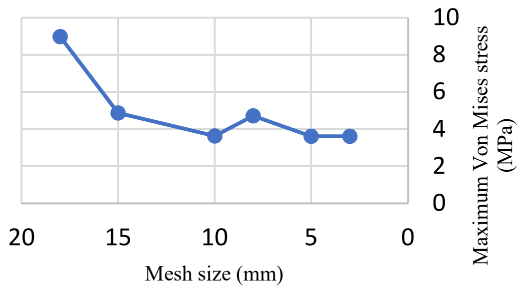

Figure 13 - Mesh convergence graph from stress simulations

The mesh sizes converged at a 10 mm maximum mesh size, with a maximum stress of 3.62 Mpa, which is marginally above the PLA's yield strength of 3.1 Gpa. This was again located solely at the wrist connection, so the fillets and the thickness were increased from 3 mm to 5 mm, and 8 mm to 10 mm respectively. This study found high-stress areas only in small corners, where cracks could occur, so smoother geometry was made to remove those stress concentrations.

After adjusting the model to remove the high-concentration areas, the prosthetic was able to withstand a rotational 20 Nm and had very little stress overall.

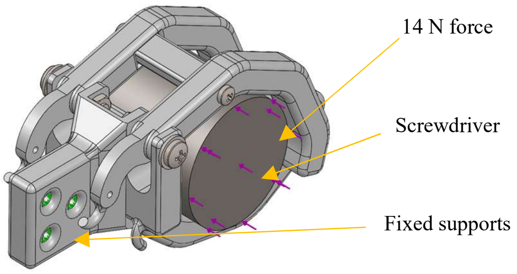

Figure 14 – Direct load simulation

Afterwards, a 14 N direct load was applied to the screwdriver model to replicate how well the prosthetic can hold a 1.4 kg weight. Initially, there were some high-stress areas around the finger's pivot points, but after adding some geometry to reduce this stress, the model was able to withstand the load entirely.

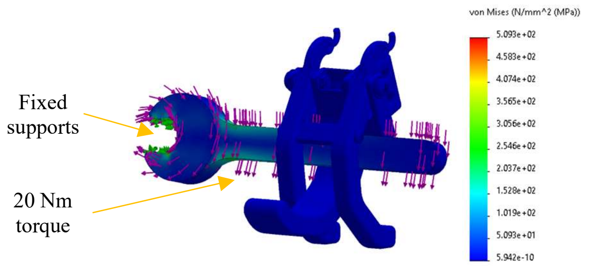

Figure 15 – Torque on spanner simulation

Lastly, a simulation was run to test how well the prosthetic could withstand the forces when applying a 20 Nm torque to a simple 150 mm M10 spanner (Fig. 15). The prosthetic was under very little stress, as shown by the maximum stress being only 2.71 Mpa, which is far lower than the PLA yield strength of 16.5 Mpa. To verify that SolidWorks was simulating this correctly, a simplified model was made, removing all holes and small surfaces, but this gave very similar results, with no significant stresses in the prosthetic. Another simulation was performed with only the force component of the 20 Nm torque being applied at the wrist part of the prosthetic. 250 N was used directly on the prosthetic from the wrist, to simulate the forces a user would have to apply to the prosthetic to manipulate the spanner. The only points on the prosthetic at stress over 3 Mpa were the finger pivot holes and the mounting holes directly next to the load.

VII. RESULTS

The practical prosthetic was successfully able to carry over 2 kg and hold various objects at high velocity. In the simulation and practical tests, the prosthetic is capable of lifting more than 1.4 kg, reaching the maximum recommended weight by the CCOHS [28] with a fully stretched arm, and is more than sufficient for holding single-handed manual tools. Weighing only 47 g (without the testing handle), this prototype competes well against most other currently available prosthetics on the market for weight (Table 9). The final prototype weighs less than even the small split hook, weighing only 55% of the medium-sized aluminium variant.

Table 9 – Weight and size comparison between prosthetics

Product	Weight (g)	Maximum grasp (mm)	Reference
Prototype V3	**47**	**40**	
Natural human hand	498	70-170	[45,46]
Split hook (medium, Al)	85	N/A	[47]
Split hook (small, Al)	68.5	N/A	[47]
MyoHand VariPlus	458	79-100	[48]
Ottobock Greifer	540	95	[49]
BeBionic	591	N/A	[50]
OpenBionics	322	N/A	[51]
Multi-D TD	272	45	[52]

Each prototype had its grip strength tested using an EH108 electronic dynamometer. Prototype 2 had a lower grip strength than prototype 1 due to the bushing housing being extended (Fig. 16), causing the pulling force parallel to the user's arm to the prosthetic to be more perpendicular. This is opposed to prototype 1, where it swivelled towards the force of the pulley, reducing the distance and angle between the prosthetic and the user, and thus reducing loss of pulley force.

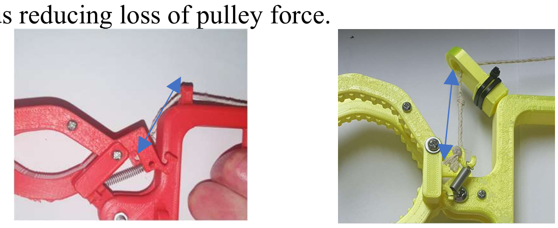

Figure 16 - Comparison of prototype V1 and V2 for the distance between the prosthetics and the pulley.

Before simulating, all models had their stress concentration areas filleted with at least a 0.5 mm radius. From the simulations, Fig. 17 shows the von Mises stress results for applying a 20 Nm torque to a spanner using the prosthetic. The prosthetic was capable of applying this torque, withstanding a maximum stress of 2.709 Mpa located near the spanner's head (Fig. 15). In this configuration, the force upon the user's osseointegration point can be found from basic calculations, using (1) [53] where force, F = 250 N, radius, r = 80 mm and the perpendicular angle between the force and torque is $\theta = 90^{o}$.

$$\tau = rFSin\theta \qquad [1]$$

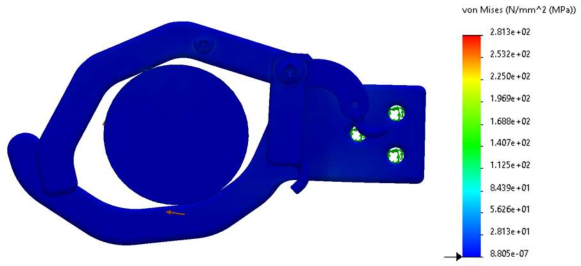

Figure 17 – Screwdriver torsion simulation stress results.

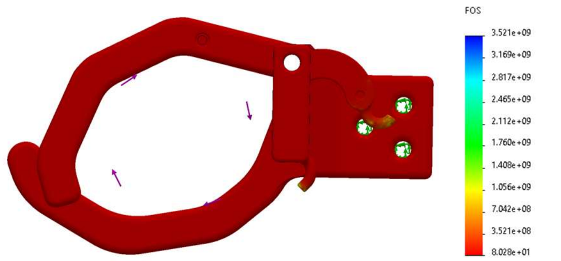

Figure 18 – Screwdriver torsion Factor of Safety results.

Examination of the results from a 20 Nm torque simulation shows the factor of safety goes up to 8.028 (except for minor stress concentration features) with a maximum stress of 5.626 Mpa located at the screws (Fig. 19). These simulation results show that the prosthetic is over-engineered because there is more material than necessary. Still, the simulations assume 100% density and perfectly isotropic material, which is unrealistic, and thus they can be taken as an estimate of a real scenario. Fig. 20 shows the prosthetic withstanding a 14 N force applied to the screwdriver, to represent holding such a weight. The locations of highest stress are again at the bolts holding the fingers together, and the bolt hole closest to the hand handle at 3.444 Mpa.

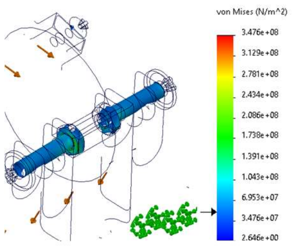

Figure 19 – During the torque simulation, the highest stresses are at the pivots and bolted points.

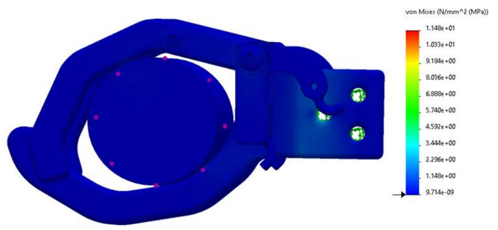

Figure 20 – Results of force simulation. 14 N applied upon screwdriver.

VIII. DISCUSSION

Through the creation process, there were occurrences which should be improved upon for future work. During the prototyping phase, numerous issues occurred, such as poor layer adhesion, poor print bed adhesion, and support weakening models. Some of these issues are shown in the first print for prototype V2, where there were significant structural weak points because of the orientation of the layers not supporting the small volume. This caused one piece to break off and could be a cause for future problems too.

After testing the prototypes, the lack of degrees of freedom was noticed to be a significant drawback. By lacking a wrist's rotation, it was difficult to turn a screwdriver about one point accurately. However, the prosthetic simulator demonstrated the prosthetic's ability to hold objects firmly at high velocities, even with a low friction interface with them and no separation of the finger digits. The simplistic mechanical design focussed purely on high strength but at the cost of dexterity and flexibility.

For future revisions, a wrist pivot is critical. This allows the user to control the tools and gives more freedom in the pulley attached to their arm more freely.

The infill patterns used for the prototypes are suboptimal in density and pattern. 20% adaptive cubic infill was used for the final prototype, which was sufficient for prototyping, but research shows that using higher densities leads to stronger material properties [40,42]. Should a final product be made via similar manufacturing methods, a stronger structure or infill should be used for stronger results.

For the simulation model, the SolidWorks simulations assume no friction and gravity, and that there is a perfect connection between the tool and prosthetic. While the simulation still demonstrates the ability of the prosthetic to withstand more than the desired torques, usage in practice may perform worse than simulated if the tool slips. The simulations also assume 100% density and a perfectly homogenous and isotropic material, which is not the case in practice.

The simulations revealed that the highest stresses were at the finger pivots and the connection points for the handle. While the handle won't exist in a final prosthetic, this shows that the stresses are generally where the most movement is which is essential information for reinforcing the prosthetic. To improve this, solidifying the prosthetic around these areas is essential to preventing failure in the long term due to fatigue. This can be done via stronger FDM materials, such as ABS or PETG, or by an AM metal, such as titanium, steel, or aluminium alloys [54].

The final prototype was very light, less than 100 g, had a grip strength of 4.6 kg, and was able to tolerate the 14 N load and 20 Nm torque. Because of this, a final version can be made from more dense material with similar or superior properties, as shown in the simulations. Refined and narrow geometry will take advantage of these superior properties for a more efficient prosthetic with high performance and robustness.

IX. CONCLUSIONS

This study demonstrates the creative process for designing a prosthetic prototype with a unique use case through CAD iterations, AM prototyping and simulations. The study found that the prosthetic is capable of:

- Holding tools with a 14 N force.
- Tolerating 20 Nm torque.
- Can hold over 2 kg in practice.
- The user is the sole source of energy for grabbing.
- Is made from common bolts, washers, and nuts.

The PLA material and geometry decisions were stronger than necessary, as seen in the stress simulations. While this increases the cost, the durability and long-term use of the prosthetic is improved, which is key for being superior to current prosthetics, as they are known for being weak and unreliable as outlined in the introduction. The design for this prosthetic prototype can be easily reproduced by other types of additive manufacturing (AM), as outlined in ISO/ASTM 53900. FDM was used for prototyping in this study, but a better alternative would be Powder Bed Fusion (PBF), as it is an AM method appropriate for metals that have been used for joint replacements in the past and is known for being highly effective [55,56]. Due to health risks associated with the open skin surrounding osseointegration abutments, a sterile prosthetic is essential. Therefore, for sterilisation by steam, manufacturing the prosthetic from stronger heat-resistant materials than PLA is required. By using AM, the prosthetic can also be customised to the user's body and needs by changing the weight and strength, modifying the infill densities, and customising the dimensions.

In practice, the prosthetic lacked dexterity and flexibility due to the lack of independent finger movement and wrist pivot. For future research, utilising a powered wrist to turn the hand to orientate the tools better would provide a unique contrast to the typical hybrid prosthetics, where the body-powered aspect controls the grasping function while the electronic aspect can manipulate the wrist. This would benefit the user in that they can feel the forces on the fingers more directly, while a motor can set the wrist to where it needs to be and stay there. Independent finger movement will also allow the user to grab tools and other non-uniform-shaped objects better. This could be enabled by a whippletree mechanism, which can control the fingers to move independently using a mechanical differential for each finger.

In conclusion, this prosthetic design is a capable prototype for proving that body-powered prosthetics are viable for bicycle repair. This study's prototype could hold heavy single-handed tools and withstand the forces that apply without the need for external influence, however, the lack of dexterity due to the lack of actuation points is acknowledged and further development here is necessary for a final product.

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