Designing an Accessible Neural Interfacing Hand Exoskeleton

By GROUP 15

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Abstract

The hand is undoubtedly one of the most critical parts of the human body. When investigating the uses of the hand, it’s been shown that its role is ever-present in the world we live in. Daily tasks become a tremendous challenge without a properly functioning hand, often leading to a decrease in quality of life. By investigating the current market of available active hand exoskeletons, the paper has deduced possible improvements to make a more user-friendly and widely available product. The hand exoskeleton was divided into five fundamental sections to investigate: the neural interface, mechanics, manufacturing, materials and design. These areas were researched, and the possible solutions were evaluated to determine their viability. The study found that existing neural interface and mechanics technologies were optimal, coupled with the use of 3D printing and composite materials, allowing for a slimmer and more accessible design.

Personal Statements

Ciara Ball

Within the project, I was elected to be the group leader and therefore was responsible for the general organisation and communication of the group. I booked a study room for every meeting that we had throughout the project as well as organising meeting times and discussions. I also recorded 'meeting notes' and posted them into teams as well as posting in pre-discussed timelines and current tasks. My primary roles once the project started were the research of the neural interfacing technology as well as the development of the pitch. I recorded the audio for the pitch and worked on the submission with the group.

After the pitch was submitted, I took over the research and development of the mechanics of the device as well as the neural interface. This involved spending large amounts of time reviewing academic papers, comparatively examining different options for the mechanics and neural interface as well as calculations into the strength and load bearing requirements of the mechanical system. I then decided on specific parts. The next task that I carried out was developing my slides & script for the presentation. I also helped to finalise the slides and during the presentation, due to the nature/content of the questions, I had to answer every question that the group was asked.

For the final dissertation, I wrote the methodology sections for both of my researched topics (Mechanics & Neural Interface). Having written these, I was then able to write the results and discussion sections for these topics as well. I was also able to make use of an appendix (Appendix A) for further information. Following this, I also wrote the section entitled 'Future Work' as well as writing the 'Conclusion'. Furthermore, I also then wrote the methodology section for 'Manufacturing & 3D Printing'. Finally, I went through the entire dissertation, checking for errors and formatting. Overall, I played a key role as the group leader, and I think that the group communicated effectively and achieved all our goals within the timeline of the project to a very high quality.

Signed: Ciara Ball
Date: 16th November 2022

John Chen

In the beginning of the project, I was assigned the task of researching the mechanics and designs for our hand exoskeleton. I took a few pages of notes on possible skin sensors, finger biomechanics, and designs that other groups pushed out. Then, I added a quick portion on the design and testing for the video pitch script; I also helped with creating the slides on Canva. I spent most of the extra time thinking of different concepts for our exoskeleton: the style of actuation and how it interacted with the finger.

Once we had decided on an idea, I worked with another person on the CAD using Onshape. I created a few simple models that incorporated elements that we wanted mechanically, such as how the string was wired through the exoskeleton and the general aesthetics. After our middle iteration and development stage, I switched over to designing the wrist/lower forearm frame for housing the electronics and linear actuators. For the presentation, I made a circuit diagram on how we would power each component and how they were connected and controlled. I also did a brief analysis on the expected power consumption over a day to derive ideal battery specifications. I presented these slides in addition to the future works and FEAs I did with the help of the group.

Additionally, I went to uCreate labs to borrow their Arduino kit and servos for the purposes of creating a circuit that directly linked two potentiometers and their positions to their respective servos. This was meant to actuate a printed exoskeleton for the presentation, but we could not do it in time. For the report, I wrote both the design methodology and design results. This included the FEAs I did on SimScale. I wrote up the electronics work I did for the presentation, which can be found in the report's appendices. It was an overall great learning experience, working as a group with strangers yet producing such awesome work. I definitely learned a lot.

Signed: John Chen
Date: 16th November 2022

Oscar Hellier

During the course of this group project, I believe I have learnt a great deal and contributed significantly to the concept, research and design of our projects aim in designing an Accessible Neural Interfacing Hand Exoskeleton.

In this group project, I have had a number of tasks and roles. During the video pitch, I was researching, compiling and writing the script for the "why" of the project, what the project actually wants to achieve, why there is a gap in the technology available on the market and how this product can help so many people with impaired hands.
After this initial video pitch was concluded, my role was to compile and expand my research on the "why" of this project and to research the FEA of the biomechanics of a human hand. Within this role, using CAD I designed and mated a mechanical Hand on Onshape to perform an FEA on it, which we hope might give us useful information regarding the forces on the hand and each finger when someone grips, grasps, pulls or pinches, however, this was later not used.
Towards the mid to later stages of the project, through the course of numerous meetings, my role became to focus on writing and structuring Aims and Objectives, the Manufacturing and 3D printing research and their respective methodologies and finally to extensively research and come to a conclusion and decision with regard to which exact material and manufacturing process to use when producing the exoskeleton.
I conducted thorough research, charted graphs on Granta, excel and material comparison websites and sought advice from professors at the university to come to a conclusion as to the two materials used for the exoskeleton (Rigid – Short Carbon Fiber Thermoplastic composite and Flexible – Polyurethane Elastomer). I also analysed and compared each type of 3D printing technology with my colleagues to decide upon the chosen method of manufacturing (Fused Deposit Modelling (FDM)).
Having accumulated this research and decided upon the materials I then compiled, analysed and evaluated it in writing the materials results section.

I believe that during this project we have achieved the formulation, methodology and detailed design of a neural interfacing hand exoskeleton product which has the potential to dramatically improve the quality of life of many people without the use of their hand having suffered an injury or condition. I have learnt a vast amount from this project, particularly with regard to material selection, the manufacturing process of Plastics, the application and function of all the different types of 3D printing, and the inner workings of FDM 3D printing technology. Not to mention, learning from the experience of creating a time-constrained video pitch to sell our idea, extensively researching and writing a comprehensive dissertation as a group and compiling slides and a script to present in front of a room of engineers and Professors.
I believe we have worked well in our group and in so doing we have achieved an intensively researched and well-informed video pitch, presentation and dissertation detailing the design of a product that we can be proud of.

Signed: Oscar Hellier
Date: 16th November 2022

Brandon Henwood

In the beginning of the project, I ventured into researching the background of hand exoskeletons. This enabled me to have better understanding of how our project will be able to make a difference. For the pitch, I volunteered to look at what improvements can be made to current hand exoskeleton's with respect to material and manufacturing options. Through researching options while keeping the themes of accessibility and slimness in mind I was able to conclude that the use of 3D printing and composite materials were going to be key part of our investigation. I then created the slides for these sections as well as writing the script for these aspects in the pitch. I also played a role in producing the overall Prezi presentation.

The next stage was to decide on the design of exoskeleton, this process was an iterative one. The first main decision was made where we finalised to used wires as our actuation system, then after toying with the design we settled on the current one. I played a large part in solving the use of support's where the idea came to me from the use of sports strapping, which I used when I injured my finger years ago. For the final presentation I was tasked with dealing with the overview, where I created slides and my script as well as prepared for any questions which might be asked about my section.

Finally, for the report, I wrote the abstract, introduction, literature review and the materials methodology, I also contributed to the results of the materials. In addition, I also played a role in formatting of the final report with touch ups before submitting.

Signed: Brandon Henwood
Date: 16th November 2022

Cormac O'Brien

My initial responsibility within this project was a member of the design team, myself and John were tasked with designing the exoskeleton structure and the actuator housing. Using literature, ergonomics and requirements set by manufacturing techniques and mechanical systems we developed several initial concept designs. Following this I modelled the exoskeleton's structure using OnShape CAD software, incorporating variables to ensure the design was adjustable for different users. After many iterations, manufacturing prototypes using the Prusa i3 MKS5 in addition to input from other group members I reached a final design.

For the final dissertation, I wrote the Manufacturing results section. Employing a combination of literature based evidence and first-hand experience printing prototypes of our design, to justify the chosen manufacturing method of FDM printing. Further to this I contributed to the materials section of the results, evaluating the use of composite materials within the exoskeleton.

Other areas I have contributed to in this project include: Editing the video pitch, FEA analysis, modelling a hand for FEA analysis, slides for both the video pitch and final presentation and assisting compiling the report. On reflection, I feel I have been a very proactive member of the group, regularly contributing to discussion and teams posts. I have gained a lot of experience during this process, particularly improving my CAD skills and knowledge of both materials and additive manufacturing techniques.

Signed: Cormac O'Brien
Date: 16th November 2022

Acknowledgements

We would like to express our gratitude and appreciation to our supervisor, Dr Yang, and the University of Edinburgh Engineering lab staff. Dr Yang's suggestions and critiques were vital in the progression of our designs and played a crucial role in guiding our writing of the report. Secondly, we would like the technicians of the engineering lab for their help in 3D printing our model.

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1. Introduction

The human hand plays a strategic role connecting our lives to the world we live in today. As we develop and grow, the hand plays a vital role for both 'human play' and everyday tasks. The hand's multiple grips and motions have multifaceted uses, ranging from opening a door to coordinating fingers when typing. The use of the hand is ubiquitous with all human functions, so natural that we fail to stop and contemplate the complexities of its movement.

The hand is clearly a significant part of our daily lives; unfortunately, many people across the world suffer from various conditions which have adverse effects on the hand, such as reduced mobility or even total loss. Today, these people are overlooked in society, limited access to rehabilitation regimes means that many are unable to restore the hand's complete function. This study aims to investigate the viability of designing an exoskeleton in order to cater to those around the world who suffer from these impairments.

Exoskeletons are common in the field of medicine, for example the first powered exoskeleton was patented in 1890 to augment the physical capabilities of humans. Exoskeletons can be divided into two groups, passive and powered. Passive exoskeletons provide additional functional support and works in tandem with the body to reduce strain. [1] By contrast, powered exoskeletons where the structure works in conjunction with the electronics. The device picks up the body's signals and transfers this to actionable data to then execute. For the purpose of this paper, this investigation will be solely focusing on powered exoskeletons.

Research into these technologies has demonstrated significant issues that prohibit the advancement of people who suffer from these impairments. For example, current exoskeletons are bulky and inaccessible to the average person worldwide. Moreover, hands are different, requiring the production of individual items, which is both expensive and time-consuming.

The study aims to determine possible areas of advancing hand exoskeleton models, focusing on the construction of a device assisting in day-to-day life and rehabilitation. Through investigating key factors in the creation of exoskeletons, the study has identified areas which can be improved. We seek to interrogate these suggestions to assess their viability.

2. Literature review

Motivation

Around the world disabilities affect over 1 billion people and ageing populations are contributing to its rise. [17] Disabilities range from mental to physical and their adverse effects on peoples' lives are undeniable. The world is complex and filled with hardship which no one is immune to, however, disabilities create everyday 'barriers' impacting things most people take for granted. [5] Furthermore, disabilities prevent social mobility as the world we live in does not cater for the needs of these people. [1]

Hand disabilities are caused by diseases, accidents and can even develop with age. These can leave the patient with an intact cortex, however, limit their motor control. One fifth of all emergencies in European emergency rooms are hand-related injuries, and one third of these patients experience chronic hand disabilities as a result. [1] The consequences of these hand disabilities are of three-fold - psychological, physical and social.

As we know, hands are a vital part of many daily tasks such as making butter and toast in the morning and using your hand while communicating. Therefore, when people are able to perform these motions without assistance, fully functioning hands create independence. Disabilities are connected to increased stress and anxiety levels, lowering a person's quality of life. [14] For example, people in occupations with a high-risk of hand related injuries, such as a construction worker who uses their hand for multiple functions, may be more susceptible to increased mental stress due to their dependency on a working hand. This stress can be transferred to their social life, possibly causing a deterioration of important relationships.[15]

2.1 Reviewing current technologies

2.1.1 Overview

Exoskeletons have typically been investigated for use in two specific areas, in the medical field and for use in military operations. These two fields often have two different purposes; the consumer product in the medical field is created as an assistive and rehabilitation device. In contrast, the military product is censored towards augmenting human's ability by amplifying one's strength. This is also the case when looking at hand exoskeletons; this paper will emphasise using a hand exoskeleton in the medical field.

As emphasised earlier, a hand disability adversely affects one's life. Thus, the engineering world needs to utilise its skillset and fulfil its responsibility to make the world a better place for these patients. Looking widely at the field, there is expected to be tremendous growth within the biomedical field of exoskeletons with the emergence of newer technologies such as composite materials and 3D printing. [4] This evidently shown below with the rise of 3D printing set to exponentially expand, especially regarding devices such as exoskeletons.

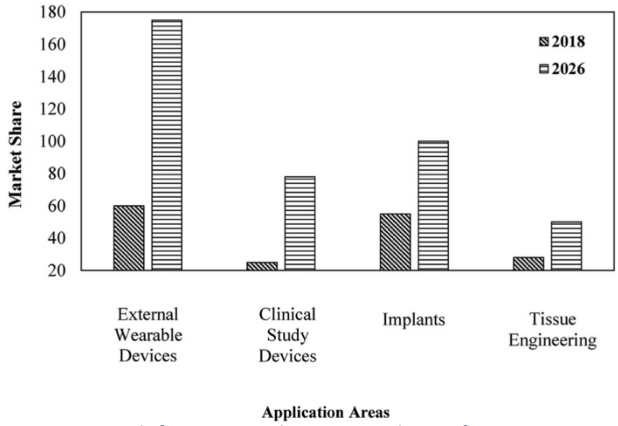


Figure 1 Graph forecasting the increased use of 3D Printing technology in the biomedical field [18]

2.1.2 Types of Hand Exoskeleton

Figure 1.2 showing examples of Rigid exoskeletons [8]

Figure 1.3 showing examples of Soft exoskeletons [8]

Hand Exoskeleton's can fundamentally group between soft or rigid hand exoskeleton. These groups will be the base of the review of the current technologies. These groups are derived from the how the force is applied to the hand, where rigid exoskeleton's force is applied through a rigid mechanical structure while soft exoskeleton refers to a glove like structure made from a flexible material and act without electricity. Illustrated above we can visually see the difference between the two.

2.1.3 Driving Mechanisms for Hand Exoskeletons

There are two primary components of the mechanism of an active hand exoskeleton, the actuation unit and the transmission unit. Where the force is delivered by the actuation unit, and these forces are transferred to the device by the transmission unit.

Actuation

Figure 1.4 showing an electromechanical actuation technique [12]

Figure 1.5 showing a pneumatic actuation through an inflatable soft hand exoskeleton [9]

In both types of exoskeletons, actuation is responsible for the movement of the exoskeleton, with the main types being electromechanical, pneumatic and hydraulic methods. Electromechanical methods are comprised of a servomotor, where a DC motor is connected to a set of gears and a control circuit. A sensor is also incorporated into the servo, which enables it to have position feedback. [12] Usually, a standard DC motor is used in the servo; however, for more advanced designs, a servo with a brushless DC motor is used. These motors are far better than standard DC motors, where maintenance and efficiency are of greater quality. [13] Depending on the design, these servos could be acting rotationally or linearly. Electromechanical options are the most popular mechanism used, with roughly 70 % of designs using them; most are rigid exoskeleton types. [8]
The second most common type of mechanism used is Pneumatic actuation, with 16% of current designs in the literature using this method. [8] These mechanisms use compressed air to actuate the fingers when needed. The mechanism comprises the cylinders, compressor, storage tank, and air tubes. Due to the necessary equipment, these designs are very bulky. They often incur a considerable weight, leading to most items needing a storage area where some designs incorporate a belt or an attachment to a wheelchair. These actuators have less control than the electromechanical type and are more likely to be used for soft hand exoskeletons.

The last most common type of mechanism is the use of hydraulic actuation. The system is very similar to the pneumatic system; however, a fluid is used instead of a gas. This type of mechanism would need far more power but, in return, produce a more significant force and thus are primarily the focus for augmentation devices. [8]

Transmission

Moving to the transmission methods, this is where the given actuator conveys the force to the hand. The transmission is responsible for closing the hand by moving the fingers. The type of transmission used is generally connected to the primary differentiation factor of the hand exoskeleton, referred to earlier.

The rigid hand exoskeleton is associated with a mechanical transmission type. These are usually situated on the top of the hand, where the options are between a direct drive and a gear structure. These two mechanisms are similar and have the same underlying mechanistic philosophy. These systems are often underactuated due to the restrictions of the hand. Thus, to increase the number of degrees of freedom (DOF), mechanical links were designed in newer models, allowing an underactuated system to actuate more degrees of freedom. [7]

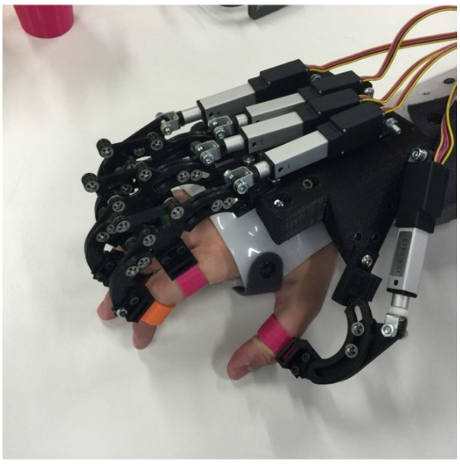


Figure 1.6 showing the use of mechanical links on the top of the hand [7]

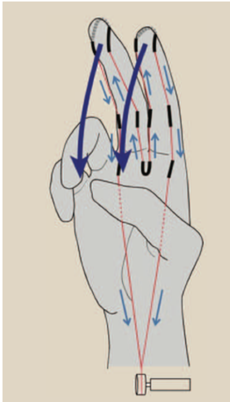


Figure 1.7 showing an underactuated cable transmission type [10]

The soft exoskeleton type is associated with a cable transmission type, where the cable is connected to the actuation system. These systems are usually modelled after the hand itself. The systems, commonly called 'tendon-driven devices', use a wire which is connected to the end of the fingers. Depending on the action required of the hand, the cables either flex the fingers or extend the fingers through the actuation system. These systems are also often under actuated where one actuator controls the action of multiple fingers through looping wires strategically around the shown above [10].

2.1.4 Materials

The materials play a strategic role in a hand exoskeleton, mainly in comfortability. Most devices are not in mass production and are developed for prototype use. This leads to most structures being created from polylactic acid (PLA) and Acrylonitrile Butadiene Styrene (ABS) for the rigid type and silicon for the soft type. [3] The main reason for this is due to how readily available these materials are and the wide variety of available manufacturing techniques.

2.1.5 Neural Interface

The neural interface refers to the methods of sensing the bio-signal from the user. Two predominant methods for detecting these signals are Electroencephalogram (EEG) and electromyography (EMG).

EEG measures brain activity using surface electrodes placed on the user's head. These electrodes pick up the activity in the brain. A brain-machine interface (BMI) computer processes this to transfer this to actionable information allowing for better control of the hand exoskeleton. [2] An EMG fundamentally operates similarly to an EEG; however, the electrodes are placed on the forearm and measure the electrical signals going through the muscle. [11]

2.1.6 Current limitations

The limitations of current hand exoskeletons differ according to the type of hand exoskeleton designed. Most models are also used as prototypes or for educational purposes, further contributing to these limitations. As such, we will address three overarching themes that restrict hand exoskeletons: adaptability, accessibility and weight.

Firstly, hand exoskeletons need to be moulded for each individual raising adaptability issues. These one-off models, also increase production time. Alternatively, hand exoskeletons can be made for the average size adult hand. While, in principle, this sounds favourable, a 'one size fits all' approach causes discomfort for the user and increases the likelihood of forces being transmitted incorrectly.

Additionally, the current cohort of devices are exclusively available to developed countries, limiting their accessibility. Research on hand exoskeletons is continually developing, and any advances in the field require funding which is very scarce in developing countries. [6] Furthermore, this makes exoskeleton's cost a premium due to the associated technology leading to further inaccessibility.

Finally, current hand exoskeletons are incredibly bulky and heavy. The literature review reveals that the materials used to produce these devices have not been thoroughly investigated. Again, this arises from the absence of a universal design, which remains a strong focus of current research.

2.1.7 Summary

In conclusion, through investigating current hand exoskeletons we believe we have identified two possible solutions to the current limitations of hand exoskeletons. By analysing the current trends in manufacturing and biomedical engineering, we believe that taking advantage of the use of 3D printing as a manufacturing method will allow us to create cheaper and more readily available hand exoskeletons. In addition, 3D printing allows us to use composite materials for the construction of these materials. This will allow the use of materials with a high strength to weight ratio allowing us to cut down on the bulkiness while catalysing on the accessibility of 3D printing.

3. Methodology

3.1 Neural Interface

The main question that we wanted to answer was whether a neural interface would establish a more useful and functional hand and would it operate more closely to that of an unimpaired hand? A secondary question was would it work and would it interface seamlessly with our mechanical exoskeleton design?

The neural interface is the component which replaces the damaged/impaired nervous system for our proposed users. Our goal was to incorporate electromyographic technology in a non-invasive but still powerful way to then provide commands to a mechanical hand exoskeleton. Many considerations were taken into account to start the research into this field. We had to consider the ethical implications of invasive vs non-invasive technology, as neural interfacing requires electrodes either implanted into the body through a surgical procedure or an electrode resting on skin to detect electrical signals from skeletal muscles. Research also had to be done to discover the accuracy and signal stability of specific electrode types, including noise data and its minimisation, as well as a cost-benefit analysis.

Following on from this, we were aware that a significant limitation of current technology is bulkiness and seamlessness of equipment. We researched the possibility of a wearable neural interface that uses surface electrodes to record muscle signals and then convert them to a motor input signal [1]. We wanted to propose a wireless option for users to minimise the above limitations and so data had to be researched and discussed from current experimental papers to prove that this adaptable concept would work in tandem with our device.

A mix of qualitative and quantitative data was used to support our choices and proposals as well as a comparative analysis of assembled data from academic papers. This was because it was too difficult to achieve our own quantitative data from the proposed neural interface technology due to budget and time constraints within the project as well as the complexity of the technology and limitations within our own engineering knowledge that would take too long to overcome. However, there are plenty of secondary sources and academic papers to support the proposal which mimic our project to a degree which makes the data comparable and usable in our justifications. In addition to this, the expertise of the electrical engineers, neuroscientists & biomedical engineers that have studied and experimented with this technology greatly increases our breadth of understanding and the validity of the design. This is necessary due to the span of STEM fields that the project covers, as it is incredibly multi-disciplinary and requires collaborative work to achieve the desired outcome.

Particularly useful academic papers were those of comparative existing neural interface technology as they conducted experiments to determine the data needed for comparative analysis (including accuracy, performance, confusion matrices & gesture recognition). The data protocol used in this specific experiment was approved by the Comités d'Éthique de la Recherche avec des êtres humains de l'Université Laval [2] and thus is justifiably valid.
Upon evaluation, there wouldn't have been a more sensible way for the methodology of designing/utilising a neural interface for our device due to its complexity and expense. Another method would've been to create a wireless neural interface for our device completely from scratch rather than by incorporating existing technology in an intelligent and low-cost way. However, this would've been near impossible given our time constraints and budget and it also would've brought the cost of the overall project up significantly as all the parts would've had to be bought separately or designed and then printed which can be costly. In addition, there would be minimal data for us to utilise and we would have to make our own comparison tables against existing products to demonstrate improvements or justify its creation. It is much more practical then to incorporate the existing technology that we found and researched extensively into the project, hence our method decision.

3.2 Mechanics

The mechanical aspect of the device design was especially important as there are a multitude of factors to consider and many different mechanical options to examine. The desired function of the hand for our project would be to close in the 'Power Grip' around the handle of a 3kg shopping bag and lift it without the fingers buckling or losing grip. The aim of this task would be difficult to achieve without an understanding of the biomechanics of the human hand and so we chose to research this in order to gain an understanding of hand functionality and to attempt a 'biomimetic' approach to the mechanical engineering design. This was also an appropriate method with which to approach the design task as we wanted to create a sleek design that closely resembles an internal hand skeleton to limit the bulkiness of current designs for users.

For this, we began with secondary data analysis and design research. The methodology included reading biological papers describing the mechanics of the hand, tendon behaviours and hand movement evaluations. We discovered that the motion and range of hand grips, bending and rotation are incredibly large. Because of the length of time of our project and our page limitations, we decided it best to limit the mechanical requirements of our design to a proposed task/grip. This way, we could devote more of our limited research time to one motion and this would mean our understanding would be significantly more in depth and developed for that specific task.

We conducted research into wires that mimic tendon behaviours to continue our line of biomimetic technology which led us into analysis of wire materials, types and behaviours. We were also concerned with the compressive and tensile nature of tendons and muscles within the hand because our wire choice would need to withstand tensile and compressive forces when the exoskeleton hand is loaded or when pushing objects. It was important to obtain quantitative data surrounding load forces during loading on our exoskeleton hand design (and the hand itself) so as to understand what force would be needed in opposition. This was done by first calculating the tension needed for a straight cable hanging from the ceiling to simply hold (and then lift at a specified acceleration) a weight. This is a simplified free body diagram of a hand picking up a bag in order to obtain quantitative data for our understanding of the wires that would be necessary. With an estimated acceleration, we are able to obtain a value for the force/tension required in the cables to lift the weight. From this, it was important to calculate a value for the Grip Force needed to lift a 2kg which involved utilising existing graphical data for lesser weights and then extending the trendline on the graph to obtain a mean value for the Grip Force needed for a 2kg weight. This gave us a minimum force value that our cables and actuators needed to be able to cope with/provide.

We did this by going through the following steps:

1. Created three graphs in excel based on the given data for the three trials.
2. Extended the trendline to read from the graph the Grip Force needed for a 2kg weight.
3. Averaged the three graphs for the three trials to get a mean result.
4. Used this result to choose an Actuator with appropriate maximum force and a cable that can withstand said force.

The next step in our methodology for the mechanics of the device involved researching cables/wires that were capable of undergoing compressive and tensile forces as well as having enough flexibility to bend yet retain their load strength under bending (as a finger curls inwards). In addition to this, the cable/wire would also have to have a relatively small diameter so as not to be a hindrance to the user when wearing the exoskeleton and utilising it, as fingers themselves are of small diameters and we are also mimicking tendons which (for the distal phalanges) have an average diameter of 5.35mm for fingers and 6.9mm for the thumb [1] to give an idea of what 'small' corresponds to.

We then collected data from secondary sources such as academic papers discussing wire types and cable comparison and also E-Commerce websites to compile data from different types of wires and cables to do a comparative analysis and determine the most effective and cost efficient cable option for our design. Important data that was collected included operating loads, diameter, length, bending angle performance/efficiency and the maximum push & pull loads. E-Commerce websites gave us datasheets with all the necessary data to make an informed choice of the suitability of the cable. Thus, in addition to being a reliable data source, this was also a good method to choose a cable as it confirmed that the exact cable that we were looking for exists and is usable for our device.

Having now chosen a cable, we had to find a device such as an actuator or a servo to force the movement required to make the cable bend the fingers of the exoskeleton and the hand combined as well as a motor to provide enough power to allow this. We approached this in a very similar way to the cable method, by extensive academic research and the compilation of data from existing products as we have already proven this to be a reliable and accurate method of choosing existing mechanical parts.

Upon evaluation, if we had more time and also a larger budget for the project, it would have been sensible to test a variety of cables and actuators/servos in a laboratory to determine their usability with our device and to record primary experimental data. However, this was not feasible given our limitations and so using mostly secondary sources and online data was the practical solution that allowed for both range and accuracy of data/results within the budget and time frame of the project.

3.3 Manufacturing & 3D Printing

The manufacture of our device is one of the most important aspects to our project. The requirements of our project were that it is cost effective, adaptable for custom sized exoskeletons, utilises our chosen materials (the methodology for which will be discussed in the next section), is accessible to countries with less manufacturing capabilities, able to produce products with a strength to weight ratio as well as a high mechanical performance and durability.

Wood, metals, and ceramic are clearly too heavy, inflexible and impractical to manufacture the intricate and geometrically complex structure of the exoskeleton which needs to ergonomically fit and support the user. Therefore, the material used to manufacture the designed exoskeleton must be high-performing plastic. Up to this point in our methodology, the decisions and choices made were through discussion-based research within the group as well as knowledge gained from the literature review into current available technology and how it is manufactured.

Descriptive data from online research regarding current technology demonstrated the options available to us especially with our prerequisites. Although other options included Injection Moulding & Vacuum Casting, the clear dominant choice was 3D Printing. We were able to conduct comparative analysis with the qualitative data of all the available options including those of FDM, SLS and SLA Printing. This choice was heavily supported with our research from academic papers, current technology and the literature review.

Similarly, we were able to conduct another comparative analysis of FDM Printing (Short Fibre vs Continuous Fibre) in order to optimise our manufacturing process and produce the best outcome for our exoskeleton design. It is extremely important to delve into the specifics of the chosen manufacturing process in order to enhance the quality of the final product and so the research into FDM 3D Printing is justifiably in depth. Quantitative research from secondary sources was also absolutely needed to reach this depth of understanding in order to select the most effective settings. Otherwise, successful manufacture would be impossible.

The next step in our methodology was to conduct further research into the availability of composite 3D Printing technology at The University of Edinburgh. This involved collecting primary data from interviews and email correspondences conducted with academics, our supervisor and PhD students to determine if we could feasibility print our design. We wanted to print the design in order to test the capabilities of using a 3D printer for our exact purpose and to establish any difficulties or challenges that could then be overcome.

Having identified the possibility of utilising a composite 3D printer at the University, we were to conduct a manufacturing experiment. This experiment was to 3D Print a single finger exoskeleton using a dual filament FDM printer to print a flexible polymer. This prototype broke once the supporting plastic was attempted to be removed. This demonstrated an issue with using a dual filament FDM printer to print a flexible polymer for the 'spine' and rigid polymer for the support structure.
Having acquired this experimental data, we were able to determine that the manufacturing issues were a result of multiple reasons. The most important ones were because of a lack of available complex printing technology as well as a design flaw and so we were able to take steps in the design and the manufacturing process to ensure that the failure did not occur again for our second attempt.

Thus, the second experimental 3D print was conducted, and it proved to be a success thanks to the prior steps recorded in this methodology. The successful print of the single finger exoskeleton is a tangible justification for our chosen methodology because we were able to achieve this in only two 3D prints, as opposed to multiple prints. This method reduced waste of the filament and it also ensured a breadth of knowledge and understanding of the manufacturing capabilities available to us and how to use them to our advantage. The option of printing multiple exoskeletons, each with a set of selected settings for testing was viable but had many flaws. Our method minimised filament wastage which also minimised the cost of our experimental procedure and was more environmentally friendly than doing multiple test/experimental prints.

3.4 Materials

In order to choose a suitable material, we will begin by referring to the current limitations of hand exoskeletons, specifically accessibility and bulkiness. The current market options are made from basic materials mentioned earlier, where the material's role has not been interrogated. Following on from the 3D printing section, the materials need to utilise the viability and accessibility of 3D printing. In addition to this, whilst looking at the functionality of an exoskeleton, the following unconditional matrix criteria for products were created. The matrix is compromised of the ability to perform in all weather conditions, affordability and availability abundantly.
In order to gain a more accurate result, the properties mentioned above matrix were applied to specific sections of the design. The overall design is thus broken down into three different material sections:

1. Flexible spine
2. Structural backbone skeleton

This ensures we accurately evaluate the whole device, thus leaving less room for error.

Firstly, looking at the flexible spine has a twofold purpose in the design. The spine needs to be able to take some of the load transferred to it when the user is carrying or holding onto something. Furthermore, it needs to contribute to helping the hand to an open position when the actuator is pushing the wires down. The material is thus required to be durable and flexible while being able to withstand much energy, thus very durable.

Finally, looking at the frame which revolves around putting together three pieces separately of manufacturing parts together - the base, mid-section and tip. All three sections need to house the mechanical wire, with the tip being the part where the wire is passed through and threaded to the other side of each finger. Referring to the placement of these pieces, loads could often be passed through them or directly applied to them. In addition, these pieces will also be the materials most likely exposed to the interaction of daily tasks.

Consequently, some conclusions need to be made about the properties of these materials. Due to these three pieces acting as the backbone of the structure of the exoskeleton as well as the main load-bearing piece of the exoskeleton, it needs to be a material which has a high load-bearing capability, rigid in structure and a high tolerance for wear and tear of everyday use.

The first step in deciding the required materials was syphoning down many materials to fulfil our unconditional properties, which were concluded earlier. One exception to the unconditional was the wire needed for the mechanism of opening and closing the hand. This was not viable for 3D printing; however, the other unconditional requirement still applied to it.

Then a quantitative comparative study was done per section, where each section's specific requirements mentioned would need to be found and then compared. GrantaEdupack was utilised per section, where the limit function from the selection stage was used in order zone in on the needed properties. The library of materials used was narrowed down to polymers and composite, thus allowing a large number of options to be scanned. The properties required were all different; thus, the scales used differed. Once the materials were narrowed down sufficiently, the final materials were placed on graphs where the axis properties were required; from here, the final materials were chosen.

3.5 Design

Our end goal with the finger exoskeleton was to enable users to perform everyday activities they could not do previously as a result of some inability to properly control their hand. To this end, we brainstormed the various aspects of a design most important to the user's experience (comfort, style, ease, reliability); the design also had to be able to satisfy our other design requirements as mentioned in the materials and mechanics sections.

Our approach going in was to do an expansive review of available literature as well as analyse other finger exoskeleton designs in order to gain a general understanding of the niche. Our research was primarily qualitative as we were looking for design ideas for inspiration. The quantitative analysis conducted was not especially in-depth as the goal was to get a grasp of the complex biomechanics of the finger and how other groups took on the challenge of modelling the finger with relation to their design. In most cases, other groups were focused on the rehabilitation of the hand and were, as such, more interested in the fine position and force control of various segments of the finger [3.5.1]. Including such dynamical control of the finger would have been a huge addition to the exoskeleton, but our general review lent us the opinion that such an inclusion would simply be too time-intensive and in violation of our primary goal, a sleek and simple frame.

Academic papers were the main source of the dynamic models and explanation of their respective exoskeleton mechanics; these papers were sourced largely from Google Scholar and ResearchGate. On the other hand, our qualitative research was sourced from a mixture of Youtube, Google Images, and aforementioned research sites; such a large pool provided a greater number of resources for inspiration and ideas. The designs were chosen on the basis of them being constructed with the purpose of serving as either rehabilitation devices or, like us, as commercial/experimental finger movement devices. Unfortunately, no experiments were conducted in the formative stages of our design process due to a lack of time; however, a great deal of brainstorming and initial designs were put forth to be judged based on their feasibility and satisfaction of the core project goals.

The next steps involved bringing our designs to life in order to be able to clearly visualise and analyse them. To this end, we utilised Onshape as our program for CAD, shaping and constructing the frame around a pre-made human finger and hand such that it would be ready to be 3D printed at any time. After the initial design discussion, we decided to go with two members each doing their own version of the CAD despite the limited timeframe. The process of sharing multiple designs can improve exploration, outcome, and group rapport in addition to the exploration of divergent sets of ideas, all integrable amongst each other [3.5.2]. Generally, results are better since more perspectives are considered, including different problems and solutions.

Of course, during the process of designing and modelling, we always kept in mind the goal; materials, electronics, and modes of actuation that would be operating alongside the frame. These are all core elements which heavily influence the design, but we did not set hard limits so as to not restrain our imagination.

This step in our design methodology also included the iterative design process, during which we would reconvene every so often to discuss and analyse our designs. Finite element analysis (FEA) was also performed on the designs given expected boundary conditions to ensure structural integrity and efficacy. This process would repeat until we had a satisfactory model of our product which could then be tested for 3D-printability and user fit.

Our design methodology was generally robust and effective in producing a final design that we were confident in. However, another methodology we could have followed would be one where a team closely worked together to produce one design, internally revising as a group. Given the amount of available free time, this process would have been too time consuming while lacking truly divergent perspectives since a group tends to think similarly unconsciously. Although there was no glaring weakness or obstacle in our methodology, the time allocated between design reviews and modelling could have been tighter to give more breathing room for the following analysis and product testing

4. Results & Discussion

4.1 Neural Interface Technology

4.1.1 Electrodes Results and Discussion

A study done by Farrel & Weir indicates that the classification accuracies of the opposing electrodes were not notably different and that choices made between the two should be made due to exterior factors such as cost and product design [1]. From this it is possible to justify the power of the surface electrodes in their signal detection capabilities as a design consideration as well as their other positives. Furthermore, the cost of surface electrodes is significantly less than inserted electrodes. You can buy adhesive surface electrodes for as little as £10 online with no medical certification. It is important to note that noise is introduced into the muscle signal the further away you record from the muscle fibres themselves [2] and so providing a clear signal becomes slightly more of a challenge than with inserted electrodes.

4.1.2 Integrating the Neural Interface

The following Box Diagram is an overview of how the Armband works in tandem with the proposed mechanical system of our device [3].

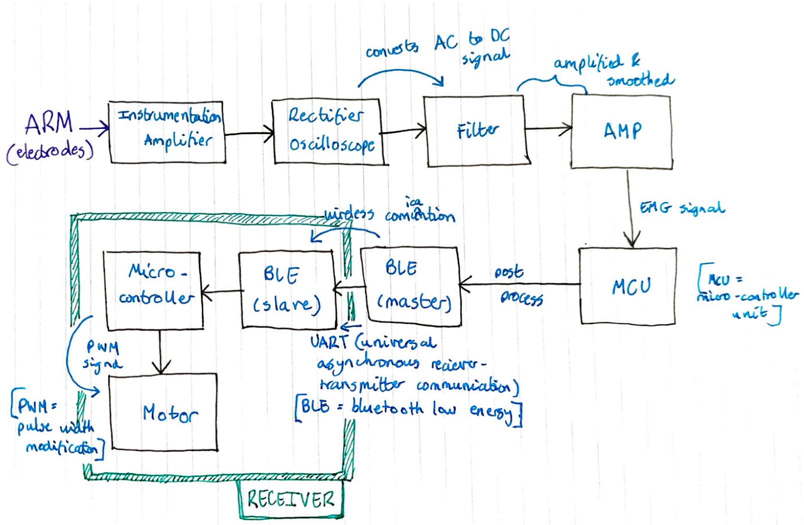


Figure 2: Box Diagram of Neural Interfacing System [3]

In more detail, on the surface of the skin are electromyographic electrodes which record signals using a 9-axes inertial measurement unit (3-axes gyroscope, 3-axes accelerometer and 3-axes magnetometer) which is then processed in a microcontroller unit (MCU). This processing usually involves a conversion of the signal from AC to DC, amplification of the signal, filtering to reduce noise and then smoothing. Following this, a transmission module sends the processed data related to the detected signals via a Bluetooth Low Energy transmitter and receiver to other electronic devices (such as an Arduino). These devices then act accordingly (depending on how they are programmed) in order to drive actuators or perform other specific functions [3].

4.1.3 Comparative Study of the Armbands

Having proven the capabilities of sEMG electrodes and understanding the function of the Armband, it was necessary to identify the most suitable design of such a neural interface and adopt it in our exoskeleton. Research showed the availability of the 'Myo Armband' as well as the '3DC Armband', both of which are wearable and wireless sEMG neural interfaces. A comparative research paper written by Côté-Allard *et al* describes the experimental process and consequent results of the performance of the two armbands [2].

The 3DC Armband is produced by Biomedical Microsystems Laboratory in Laval University and the Myo Armband is produced by Thalmic Labs (now rebranded as 'North').
Initially, it is helpful to compare all the existing data acquisition devices that are available in order to get an overview of the technological specifications. The paper describes this data within Figure 3 [2];

	Delsys Systems	Biometrics	Noraxon	Oymotion	Thalmic Lab	Hercules	3DC Armband
sEMG channels	up to 16	up to 16	up to 32 (at 2000 sps)	8	8	8	10
sEMG ADC *	16 bits	13 bits	16 bits	8 bits	8 bits	12 bits	10 bits (ENOB *)
sEMG Sampling rate	1960 sps	2000 sps	4000 sps	1000 sps	200 sps	1000 sps	1000 sps
Bandwidth or Built-in Filters	20–450 Hz or 10–850 Hz	10–490 Hz	5/10/20–500/1000/1500 Hz	20–500 Hz	~5–100 Hz	20–500 Hz	20–500 Hz
Contact Dimensions	5 mm2	78 mm2	N.A.	~66 mm2	100 mm2	78 mm2	50 mm2
Contact Material	Silver	Stainless Steel	N.A.	Stainless steel silver coated	Stainless Steel	Gold plated Copper	Electroless nickel immersion gold (ENIG)
Full Scale (Peak to Peak)	+/−11 sps or +/−22 sps	+/−6 sps	+/−24 sps	N.A.	~+/−1 sps (measured)	+/−6 sps	+/−3 sps
Input referred-noise (On system bandwith)	N.A.	<5 μV	<1 μV	N.A.	N.A.	N.A.	2.2 μV
IMU * sensors	9-axis Acc, Gyro, Mag	No	9-axis Acc, Gyro, Mag (if EMG set at 2000 sps or below)	9-axis Acc, Gyro, Mag	9-axis Acc, Gyro, Mag	No	9-axis Acc, Gyro, Mag
IMU Sampling rate	24–470 Hz (Acc), 24–360 Hz (Gyro), 50 Hz (Mag)	-	200 Hz	50 Hz	50 Hz	-	50 Hz
Transmitter	BLE 4.2	WiFi	2.4 GHz	BLE 4.1	BLE 4.0	Wi-Fi	Enhanced Shockburst **
Autonomy	4 to 8 h	8 h	8 h	N.A.	16 h	N.A.	6 h
Weight	14 g (per channel)	17 g (per channel)	14 g (per channel)	80 g	93 g	N.A.	62 g
Price	~$20,000 USD (for 16 channels)	~$17,000 USD (for 16 channels)	~$20,000 USD (for 16 channels and free battery replacement)	$1250 USD	$200 USD	N.A.	~$150 USD ***

Figure 3: Comparative Data Table for Neural Acquisition Systems [2]

As observed here, although the specifications of the two comparative armbands are similar, the 3DC Armband (column highlighted in purple) is cheaper, has smaller contact dimensions, has two more sEMG channels, had a significantly higher sEMG sampling rate and bandwidth and is also 31 grams lighter than the Myo Armband (column highlighted in grey). The Myo Armband has a 10 hour longer autonomy than the 3DC Armband.

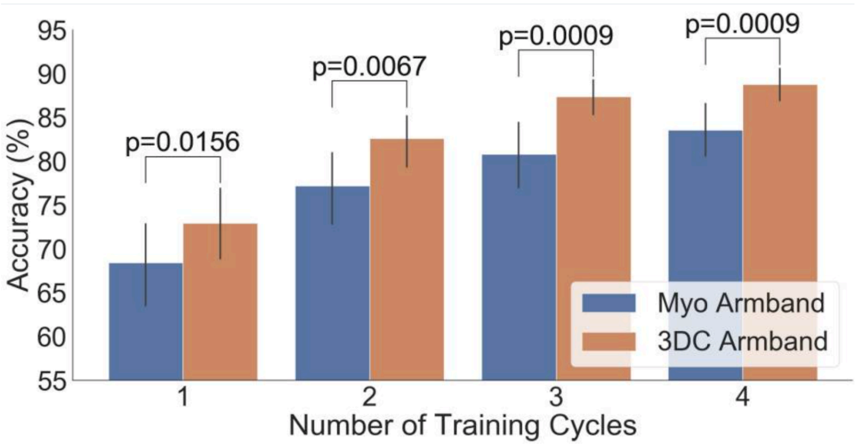


Figure 4: Wilcoxon Signed Rank Test for Accuracy determination [2]

In addition, it is important to note that the 3DC Armband has closer specifications to those of the extremely expensive sEMG acquisition systems that we are not considering to use due to their price which indicates better quality and performance.

It is important to apply the proposed devices to real-life scenarios to accurately compare their performances and usability. Figures 3 and 4 demonstrate that the 3DC Armband outperforms the Myo Armband. Figure 4 is a comparative graph employing the Spectrogram ConvNet for classification and then using The Wilcoxon Signed Rank test to determine the accuracy across 4 training cycles for each armband [2].

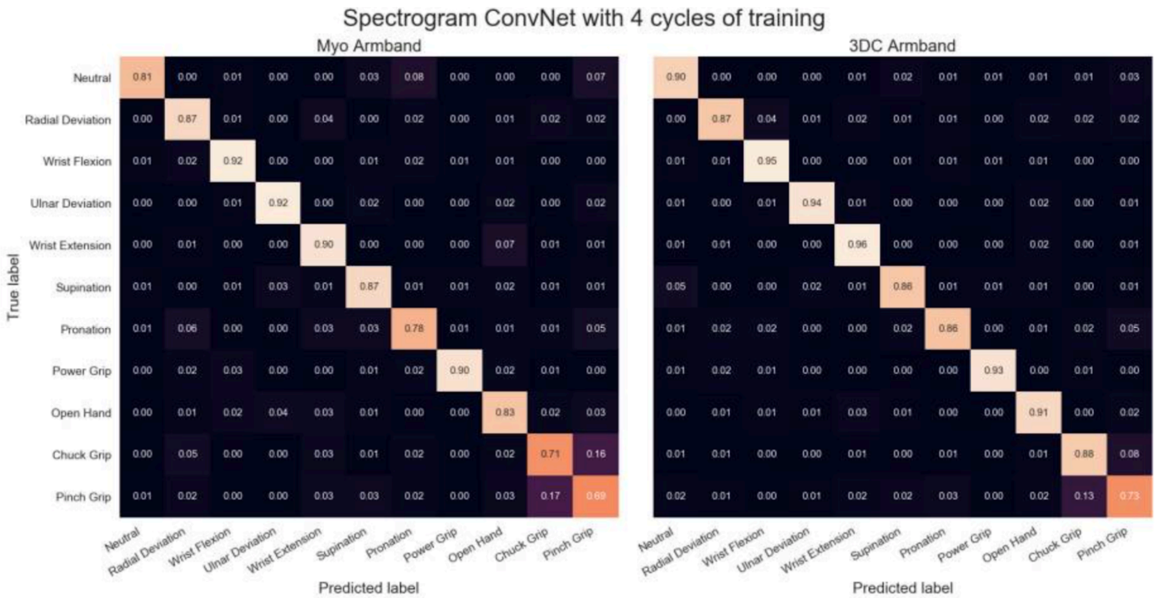


Figure 5: Confusion Matrix for Spectrogram ConvNet Classification [2]

Figure 5 is the corresponding Confusion Matrix for the Spectrogram ConvNet classification. As said in the paper, ‘a lighter colour is better’ [2] and it is clear that the 3DC Armband has a slightly lighter overall matrix. The figures found in *Appendix A* further demonstrate that across all measured cases, the 3DC outperforms the Myo Armband and details further description on the classifications. The other figures are for the Raw ConvNet classification & Linear Discriminant Analysis (LDA) classification.

4.1.4 Neural Interface Summary

From these technical and experimental comparisons, we can utilise the 3DC Armband as the best performing, low-cost, wireless & wearable neural interface in our project. We would need to incorporate a Bluetooth Low Energy receiver within the Arduino in order to ensure proportional myoelectric control which is possible to do with built-in hardware on the Arduino [4]. Proportional Myoelectric Control is the utilisation of a microcontroller to control a device/system with EMG signals which is exactly what our system is trying to achieve.

4.2 Mechanics

4.2.1. Overview

To justify the suitability of the chosen mechanical system, it is crucial to first calculate the requirements it must fulfil. Calculating the maximum bending angle of the finger enables the evaluation of the performance push pull cable under these maximum conditions. Additionally, it is fundamental to determine the force required to close a user’s hand, in order to establish both the power requirements for the linear actuator.

4.2.2 Performance of Push-Pull Cable in Bending

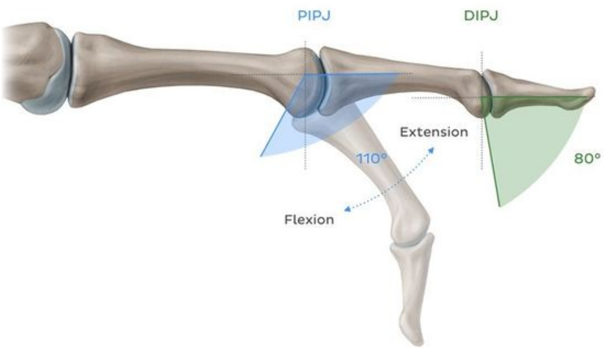


Figure 6: Maximum Bending Angles of the Proximal Interphalangeal Joint (PIPJ) and Distal Interphalangeal Joint (DIPJ) [1]

The maximum total bending angle of the proximal interphalangeal joint is 110° - 90° inwards (flexion) and 10° outwards (extension). For the distal interphalangeal joints the maximum angle is 80° inwards (flexion) [1]. This varies slightly by the finger, however by selecting the greatest angle we can ensure the device can replicate the maximum range of finger bending.

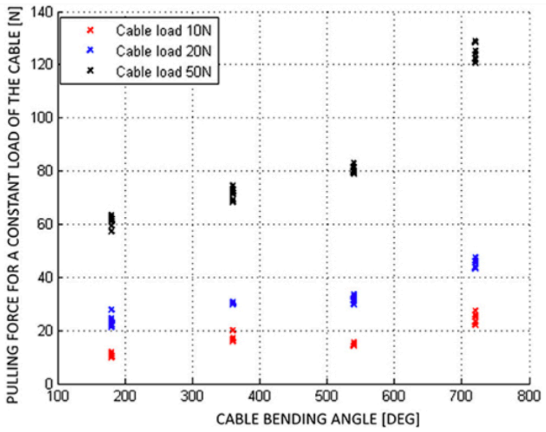


Figure 7: Pulling force of a push-pull cable under loads at varying bending angles

It is clear from Figure 7 [2] that the push pull cable functions extremely well at bending angles greater than the maximum flexion angle required of 90°. There is a slight increase in the pulling force required to support constant loads of 10N, 20N and 50N, under a cable bending angle; the average percentage increase to support a 50N load through a bending angle was measured to be 18.2%. This difference is not negligible, however, plotting a linear regression line and extrapolating this backwards for angles lesser than 100°, we can assume the required pulling force will decrease and the differential between the force required and the load will become negligible. It is therefore appropriate to assume a 100% pulling load efficiency for our cable at bending angles less than 100 degrees.

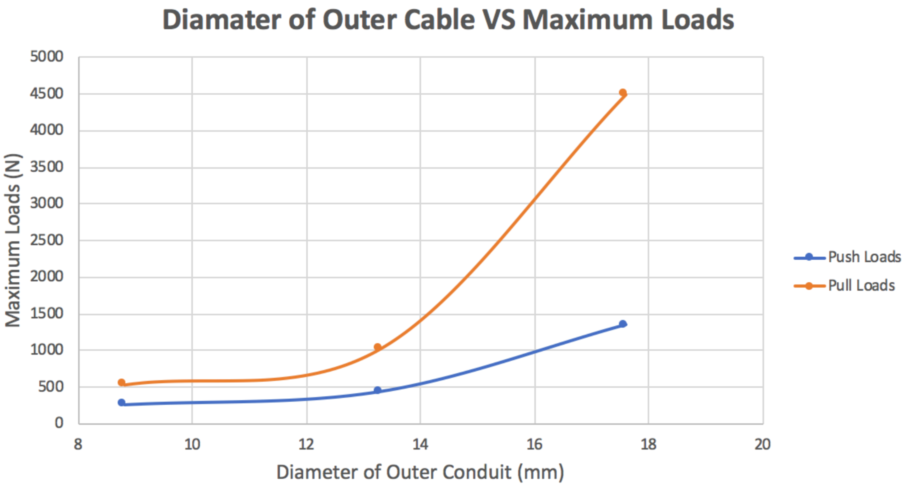


Figure 8: Maximum Load vs Diameter for Push Pull Cables under bending [3]

To further prove this, using data from source [3], Figure 8 was made. The data was for a push-pull cable of varying diameter and the corresponding maximum load it could lift. The manufacturers specification was for bending angles of 180 degrees through to 900 degrees. This graph demonstrates that that the bending angle is not a limiting factor as the maximum loads at small diameters are still relatively large. The bending angles of the finger are less than 100 degrees. The cables in the source prove that even at small diameters (i.e 9mm or less) and large bending angles, the maximum load is still much larger than our desired force which will now be discussed.

4.2.2 Required grip force

"During a manipulative action, grip force (normal to the surface) and load force (tangential to the surface) are highly coordinated, enabling grasp stability while guaranteeing adequate fingertip forces when load is changing." [4]. Hence, it is very important to consider grip force to determine the necessary specifications of our actuator and cable.

"The average healthy grip strength for men is a squeeze of about 72.6 pounds (32.93 kg) while women typically measure around 44 pounds (19.96 kg)" [5]. This is a force of 323.04 N for men & 195.81 N for women.

Considering the requirement of the exoskeleton, to be able to lift a 2kg weight; we must estimate the pulling force necessary to do this. Modelling the load as the tension in a vertical cable fixed at one end supporting 2kg weight, and applying Newtonian Laws:

$$F = mg = 2 * 9.81 = 19.62N.$$

What would be the maximum cable pulling force? If we were to accelerate the lifting at a = $3ms^2$, the force would need to be :

$$T = mg + ma = 2(9.81) + 2(3) = 25.62N.$$

Thus, from here our aims are to find a cable with enough strength to withstand 25.62 N of force. This is a minimum force calculation, in reality this is likely to be higher and we will attempt to find this value now.

To find a more accurate value for the grip force needed in a hand to lift a 2kg weight, I researched the Grip Force to Load Force Ratio. I found a research paper which described

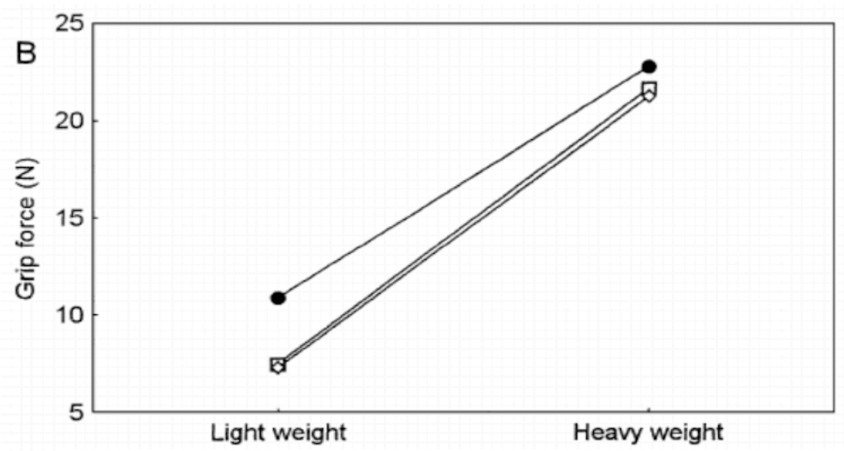


Figure 9: Grip-Load force ratio (A) and grip force (B) for heavy and light weight movements accompanied with a contralateral weight change as a function of trials

experiments of lifting weights through contralateral (opposite hands) and bilateral hand transitions (use of two hands together to grasp and manipulate objects) [4].

Figure 9 [4] shows the Grip Force in Newtons through three separate trials for a light weight (250g) and a heavy weight (750g). I was able to use this data to extend the existing graph for our 2kg weight. This is shown in Figure 10 and how I did this is detailed in the methodology.

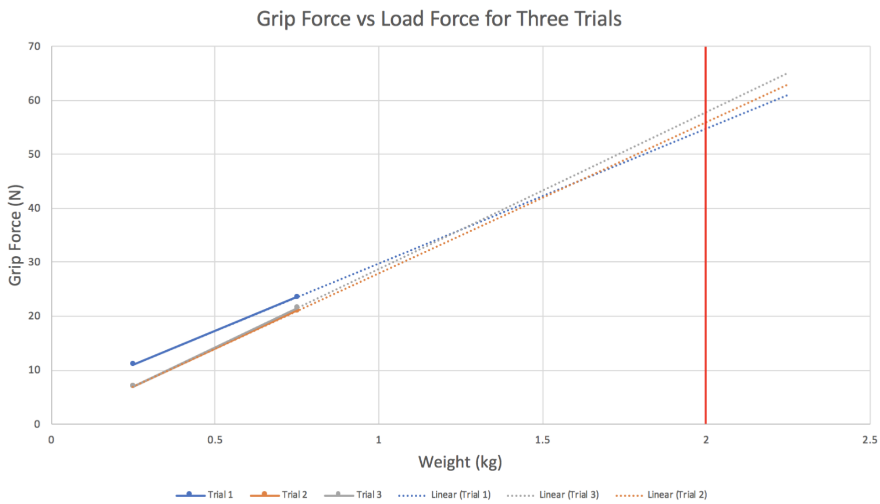


Figure 10: Grip Force vs Load Force with an extended trendline for three separate trials

Trial 1 gave a Grip Force of 55N for 2kg, Trial 2 gave a Grip force of 57N for 2kg, Trial 3 gave a Grip Force of 58N for 2kg.
The Average Grip Force needed to lift a 2 kg weight is thus $\frac{(55+57+58)}{3} = 56.67N$.

4.2.3 Summary

Therefore, our aims are to provide enough force to lift a 2kg weight (required 56.67N) and also to provide as close to average grip strength for a healthy person as possible (323N for men, 196N for women).

The actuator I chose can lift a Maximum Force of 100N. Its input voltage is 12V [6].

The cables I chose can be specially manufactured for any diameter/length and are widely used for truck, bus, boat or marine systems, thus are extremely capable of withstanding the necessary forces [7].

4.3 Manufacturing

4.3.1 Overview

Due to the geometrical complexity of our design, the available manufacturing methods are restricted to 3D printing techniques or injection moulding. Plastic extrusion, compression moulding, blow moulding and vacuum forming methods require a simple continuous part design, therefore these methods are unable to produce our required form. Due to the low cost and high accessibility we hypothesised FDM printing to be the most suitable manufacturing method for our exoskeleton.

To justify our chosen manufacturing technique, we must first evaluate and compare alternative methods. Assessing the strengths, weaknesses and the suitability of different procedures in relation to our design we can validate our chosen manufacturing technique. As outlined by our aims and objectives, three crucial requirements influenced by the chosen manufacturing technique of the exoskeleton are: good mechanical performance, high cost effectiveness and widely accessible. Each of these objectives will form the fundamental criteria for the justification of our chosen manufacturing technique.

4.3.2 Types of 3D Printing

Technique	State of Start Materials	Typical Materials	Working Principle	Advantages	Disadvantages
FDM	Filament	Thermoplastics, such as PC, ABS, PLA, and nylon	Extrusion and deposition	Low cost, good strength, multi-material capability	Anisotropy, nozzle clogging
SLA	Liquid photo-polymer	Photocurable resin (epoxy or acrylate based resin)	Laser scanning and UV induced curing	High printing resolution	Material limitations, cytotoxicity, high cost
SLS	Powder	PCL and polyamide powder	Laser scanning and heat induced sintering	Good strength, easy removal of support powder	High cost, powdery surface, material limitations

Table 1: Summary of Common 3D Printing Techniques: FDM, SLA and SLS. Adapted from [1]

Table 1 summarises the three most common 3D printing techniques, Fused Deposit Modelling, Stereolithography and Selective Laser Sintering. Both SLA and SLS methods have material limitations and are much higher cost than FDM printing. FDM printing is the most common 3D printing technique, suitable for a wide range of thermoplastics and is compatible with composite 3D printing. Consequently we will be evaluating FDM printing as our chosen 3D printing option.

4.3.3 Mechanical Performance

During the flexing of the exoskeleton the ‘spine’ elongates under tension to enable the finger to contract. Subsequently, it is fundamental to assess the effect of both FDM printing and injection moulding techniques on the stress-strain performance of a material.

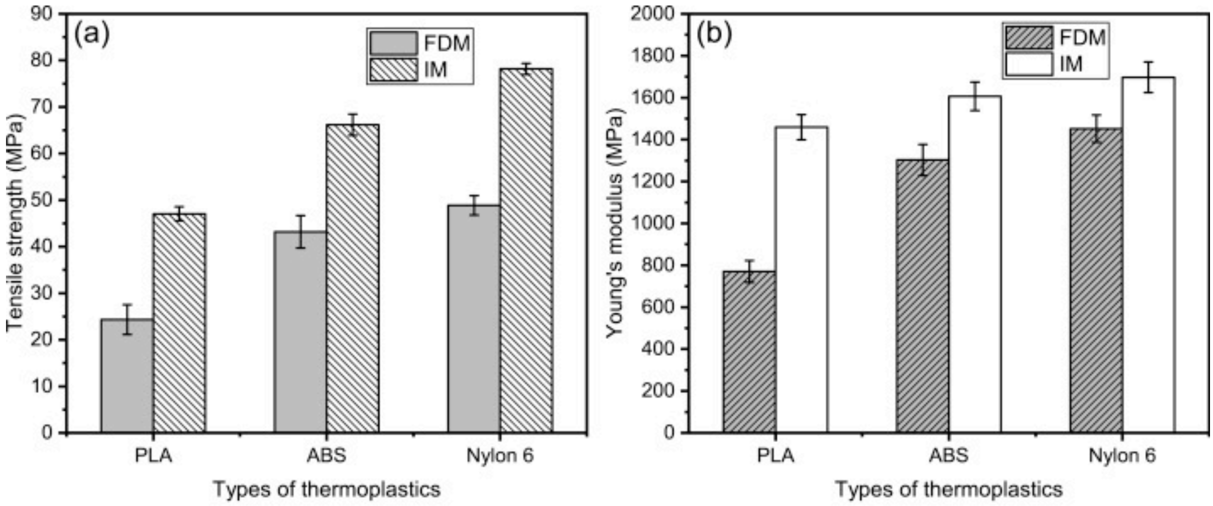


Figure 11: Analysing the Effect of Manufacturing Methods on the Mechanical Properties of Thermoplastics [2]

It is clear from *figure 11* that injection moulded components exhibit a much higher stress-strain response than FDM printed component, the tensile strength of FDM printed PLA is approximately 40% of its injection moulded counterpart. In order to understand the source of this disparity we must examine the microstructure of each component *(as shown in figure 12).*

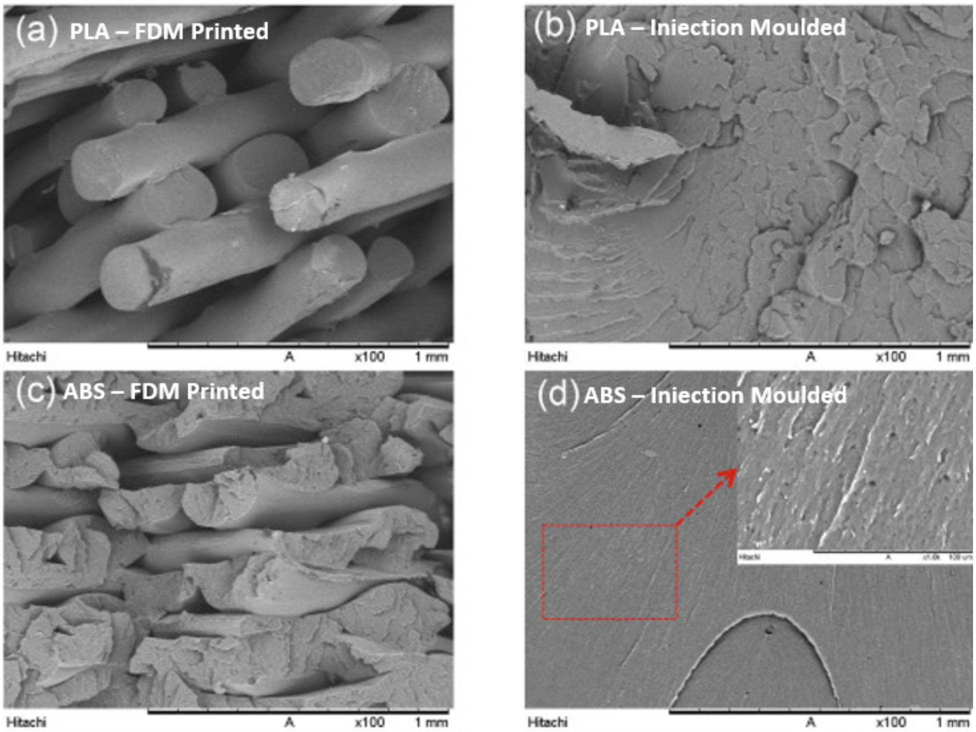


Figure 12: Examining the Microstructures of (a) PLA – FDM Printed, (b) PLA – Injection Moulded, (c) ABS – FDM Printed and (d) ABS – Injection Moulded [2]

During FDM printing a polymer filament is deposited one layer at a time *(see figure 13)*, this prevents uniform crystallisation within the component, creating large voids between layers. The interface between each of these layers is small, subsequently the force required to separate the layers is reduced.

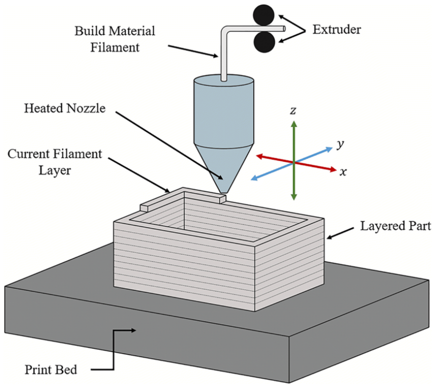


Figure 13: Schematic Diagram of the FDM Printing Process [3]

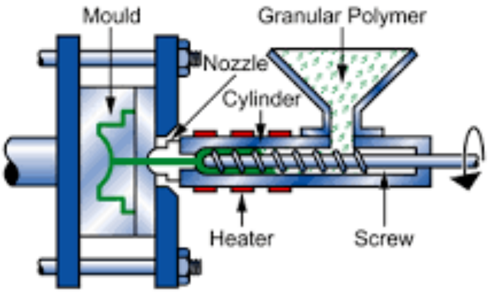


Figure 14: Schematic Diagram of the Injection Moulding Process Available from Granta Selector

Alternatively, in the injection moulding process, molten plastic is forced into a mould, packing the molecules closely together. Channels within the mould cool the part evenly, encouraging uniform crystallisation to occur within the part. Consequently, the microstructure of the injected moulded components is much more solid with very few voids. Packing the molecules together increases the density of crosslinking within the polymer, hence more force is required to deform and break these bonds.

4.3.4 Cost Analysis

Although injection moulding produces parts with increased mechanical performance, we must also consider the cost of injection moulding. Injection moulding moulds can be extremely complex and require precise machining making them very expensive.

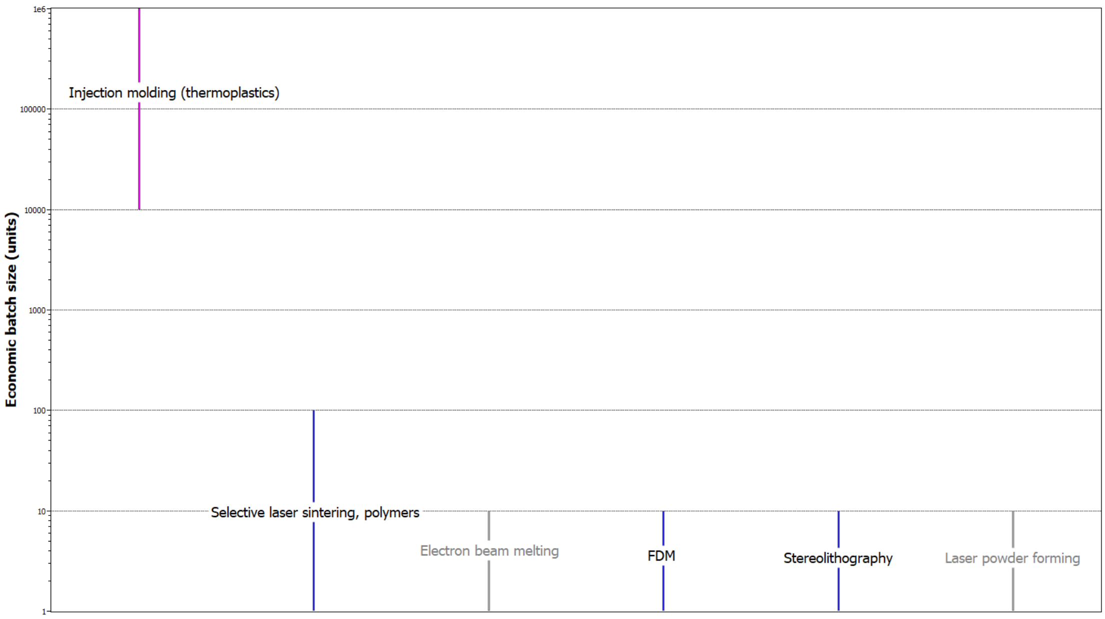


Figure 15: Granta Selector Plot of Economic Bach Sizes (number of units required for economic cost) for Available Manufacturing Methods

Considering each exoskeleton must be custom fitted to the user, the manufacturing method must be affordable for one-off production. The economic cost of manufacturing complex moulds for injection moulding makes the process only viable for batch sizes over 10,000 units *(see figure _)*. Moreover the time taken to produce each mould for each user, for each finger would make the process extremely inefficient.

4.3.5 Accessibility

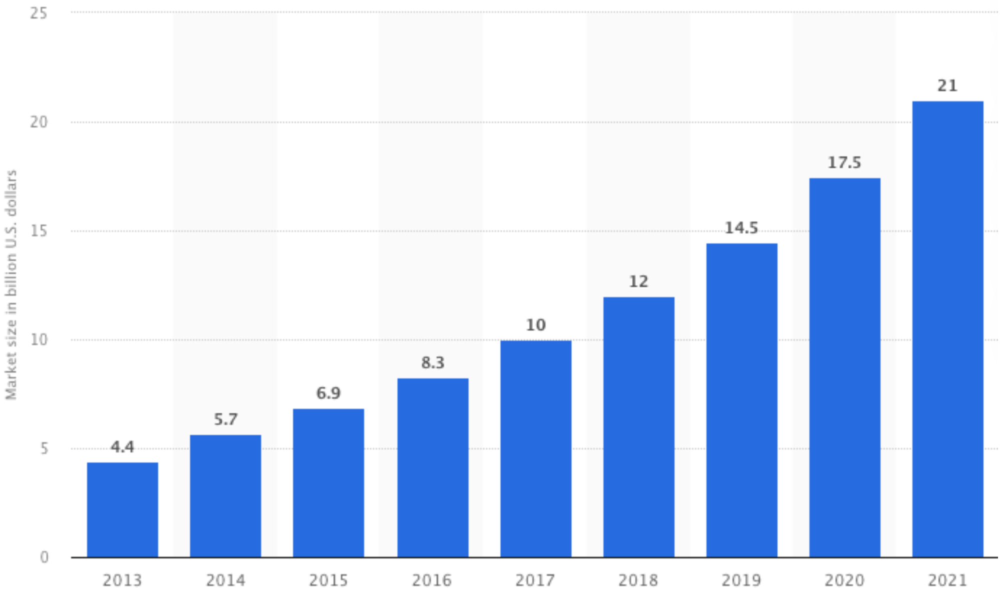


Figure 16: Global Market Growth of the 3D Printing Industry [4]

The 3D printing industry has been consistently expanding, the global market size stands at $15.1 billion in 2021 and is expected to reach USD 83.9 billion by 2029 [8]. A total of more than 140,000 industrial 3D printers and 2 million consumer 3D printers have been sold worldwide [9], by 2030 total shipments are expected to reach 21.5 million units [10]. The growth of the 3D printing industry will only increase the accessibility of this technology around the world, ensuring the maximum number of users can benefit from our rehabilitative exoskeleton.

4.3.6 Composite FDM Printing

It is possible to improve the mechanical performance of an FDM printed component by introducing composite fibres into the polymer. The two most common techniques to do this are continuous fibre

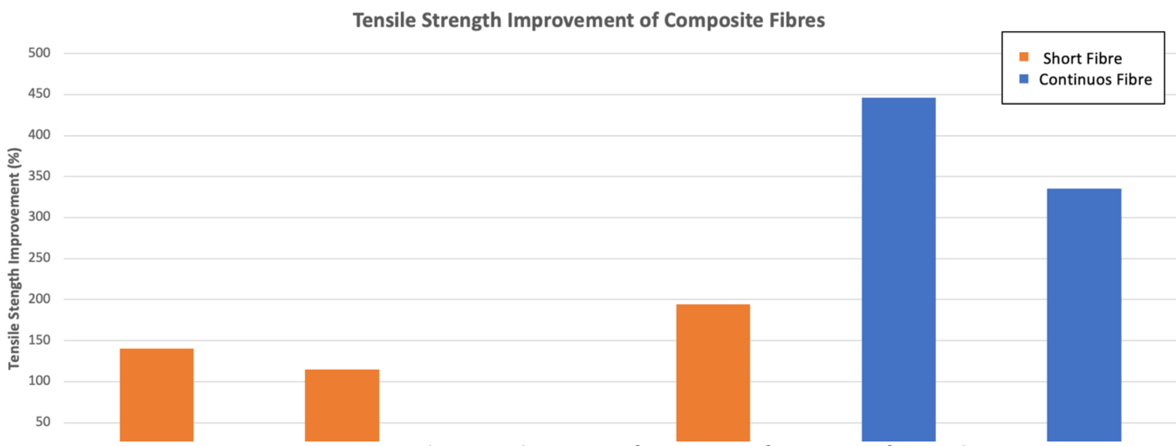


Figure 17: Percentage Improvements in Tensile Strength Compare for a Range of FDM Manufactured Composites (compared to a pure polymer). Data collected from [1]

extrusion and short fibre extrusion. By varying different compositions of the reinforcing material we can further adjust the mechanical performance of a composite.

Figure 17 illustrates the improvement in tensile strength for a range of composite compositions; short fibre composites exhibited a range of improvement in mechanical properties of 5% - 40%, this increased to over 300% when continuous fibre extruding. Evidently, the inclusion of reinforcing fibres within our polymer considerably improves its mechanical performance.

Reinforcing Fibre	Working Principle	Advantages	Disadvantages
Short Fibre FDM	Prepreg fibre reinforced filament used instead of pure polymer filament	No modification required, pre-processed, low cost	Uneven fibre distribution, omnidirectional
Continuous Fibre FDM	(a) modified nozzle impregnates polymer or (b) Dual headed printer coextrudes polymer and fibre	Significant improvement in mechanical performance	Specialist equipment required, expensive, anisotropic

Table 2: Comparison between Short Fibre and Continuous Fibre Reinforcing Techniques

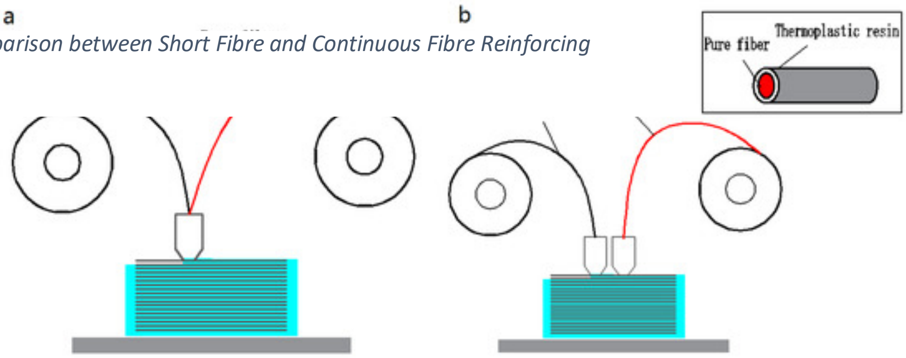
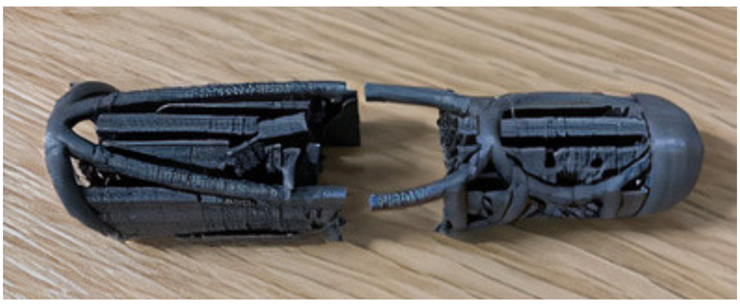


Figure 18: Schematic Diagram of Continuous Fibre Extrusion Methods; (a) In-Situ Fibre Impregnation, (b) Coextrusion [5]

Although continuous fibre extruded composites show a significantly increased mechanical performance, this only occurs along the direction of the fibres. Short fibres are randomly arranged in all directions, so they do not exhibit anisotropic properties, instead increasing the mechanical performance omni-directionally. Short fibre reinforced techniques are compatible with all FDM printers, subsequently this is a cheaper more accessible option.

4.3.5 Concept Prototyping



Initially the exoskeleton was designed as a single component, we intended to use a dual filament FDM printer to print a flexible polymer for the 'spine' and rigid polymer for the support structure. When printing a prototype of this design, the print failed. The flexible to rigid material interface was weak, and there was significant material wastage; 50.5% of the filament was used to print support material.

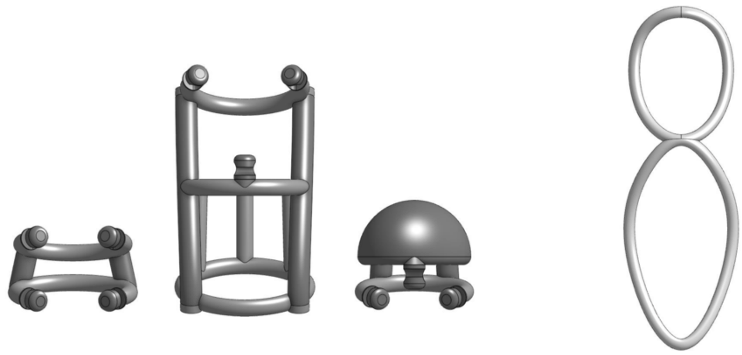



Figure 20: Manufacturing Optimised Finger Exoskeleton Design

Figure 21: Prototype of Improved Design, Printed on Prusa i3 MK3S

Following this initial prototype we modified the design, incorporating clips into the rigid structure so that we could manufacture the flexible spine separately. This removed the requirement to use the dual filament printer, replaced the rigid joint between the materials with a flexible joint and reduced the amount of support material used by over 40%. Consequently, we were able to successfully print a prototype.

4.3.6 Optimising FDM Printing

When FDM printing there are several controllable variables such as printing temperature, layer thickness, extrusion speed and infill density. Each of these factors directly influences both the quality

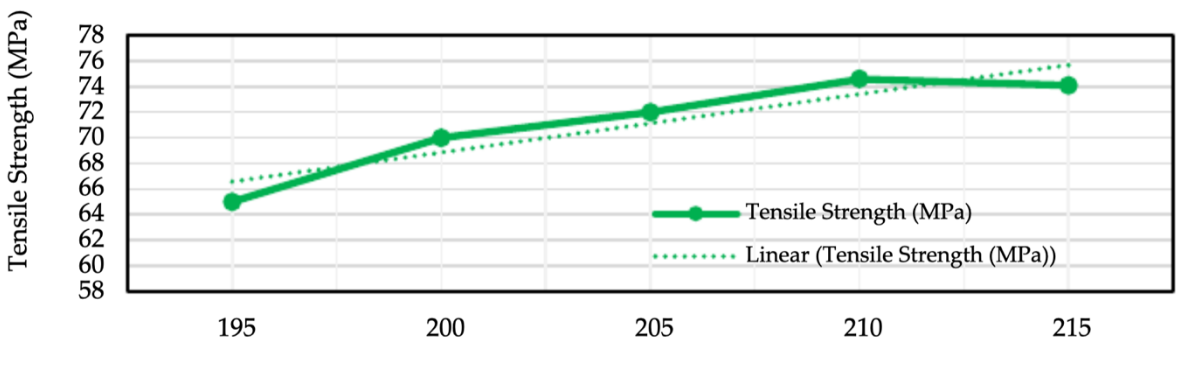


Figure 22: Effect of Extruder Temperature on Tensile Strength [13]

and properties of the final print. In order to ensure an optimal manufacturing technique it is crucial to understand the effect of each of these variables.

A higher extruder temperature increases the melt viscosity of the polymer, increasing the material interface between each of the layers. Although, if the extruder temperature is too high de-lamination of previous layers may occur. However, if the extruder is too cool poor lamination occurs between layers, reducing the maximum tensile strength of the printed part. Using data gathered by varying the extruder temperature for carbon fibre reinforced PLA, we can determine the most effective extruder temperature to be 210C. At 210C, when tested the printed component exhibited the highest tensile strength.

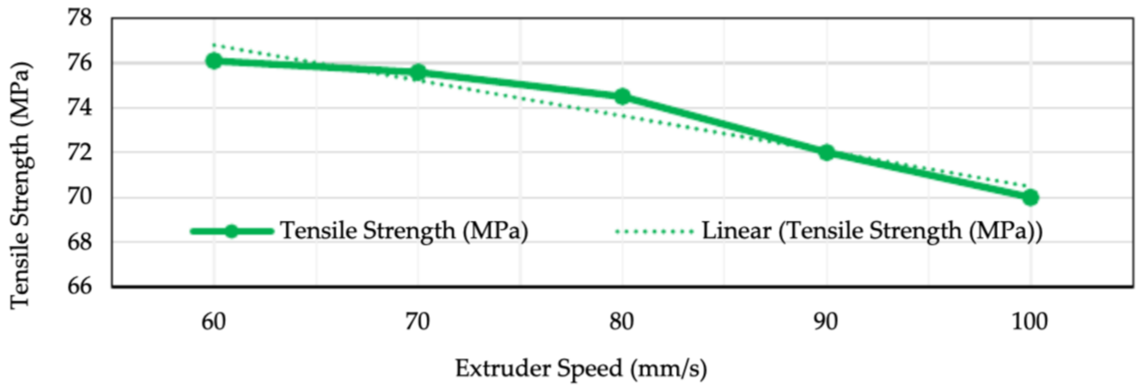


Figure 23: Effect of Extruder Speed on Tensile Strength [13]

Typically, reducing the speed of the extruder produces a better quality part, however if the speed is too slow filament can build up in the nozzle and cause defects. A slower speed will evidently increase the total print time, so the extruder speed must be evaluated against the quality improvements to select the most effective settings. During the experimentation the maximum tensile strength of 76 MPa was achieved using an extruder speed of 60 mm/s. Opting for a greater efficiency we can increase the print speed by 16.7% whilst only reducing the maximum tensile strength by 0.9%.

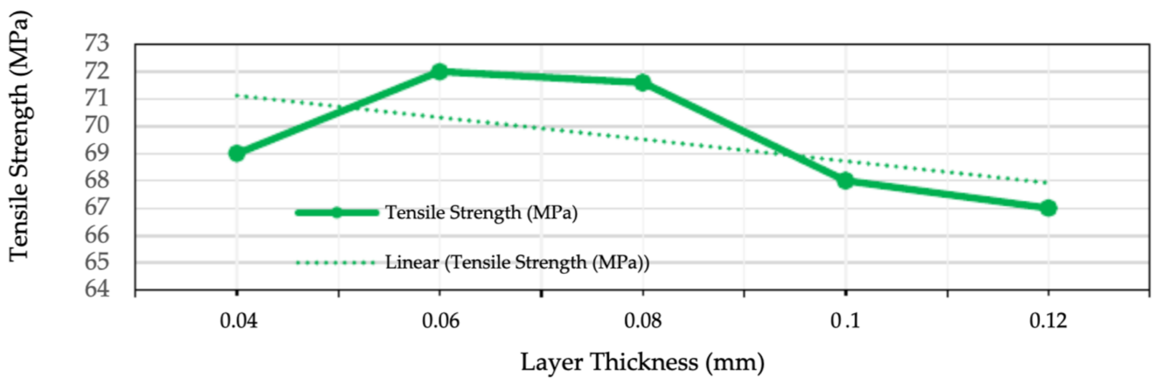


Figure 24: Effect of Layer Thickness on Tensile Strength [13]

By reducing the extrusion layer thickness we can improve the definition of the detailed sections within our printed component. However, this may reduce the maximum tensile strength; a direct result of thinner layers is an increasing total number of layers, consequently, the weaker layer to layer forces will dominate. If the layers are thick, delamination is likely occur as sufficient layer adhesion cannot occur. Considering the size of the smallest feature within the exoskeleton design to be 1 mm, a layer thickness of 0.06mm would provide sufficient detail, whilst optimising the tensile strength of the component.

4.3.7 Summary

FDM is the most widely used form of 3D printing technology in the world and therefore the most accessible and it is also the most cost-effective 3D printing technology process.
FDM has the distinct advantages of high accuracy, low cost, large material selection and particularly the capability of printing complex structures with fibre-reinforced and flexible material.
As detailed above FDM has the capability to accurately and effectively manufacture the exoskeleton Rigid frame structure from short carbon reinforced fiber PLA and also the flexible spine of the exoskeleton from Polyeurethane Elastomer.

It is evident that FDM 3D therefore meets the requirements of the exoskeleton specifications extremely well and proves to the optimal manufacturing process to produce it. Furthermore using FDM, it is possible to further improve the mechanical and structural performance of the exoskeleton by adjusting the variable settings of the FDM 3Dprinter; tuning the extruder temperature to 210°C, the extruder speed to 70 mm/s and layer thickness to 0.06 mm.

4.4 Materials

The exoskeleton design for each finger has been discussed in Section 4.3 and the results of this discussion are that the design must be specifically manufactured from two polymer 3D printed materials.

4.4.1 Rigid frame of the Exoskeleton material

The primary structure and supporting frame of the exoskeleton that fits around each finger of the patient must be rigid, strong and durable. It must be able to be printed using the very accessible FDM 3D printing technology, such that the customized part of the design can be tailored and manufactured anywhere in the world.
As previously discussed in Section 4.3, one of the reasons for choosing the widely used and accessible FDM 3D printing method is because of its ability to print fibre-reinforced composite polymers. Reinforcing the thermoplastic filament with fibres dramatically improves mechanical properties with some fibre-reinforced composite polymers having better mechanical properties than even metals.

4.4.2 Thermoplastic filament

FDM 3D printing involves extruding a hot thermoplastic filament to build the intended structure.
The most common thermoplastic 3D printing filament materials which can be reinforced with fibres are standard PLA, ABS filaments, but also sometimes Nylon.

The graphs below are sourced from Granta EduPack. They compare the following thermoplastic filaments against each other: Acrylonitrile butadiene styrene (ABS) in Red, Polyamides (Nylons, PA) in Green and Polylactide (PLA) in Blue.

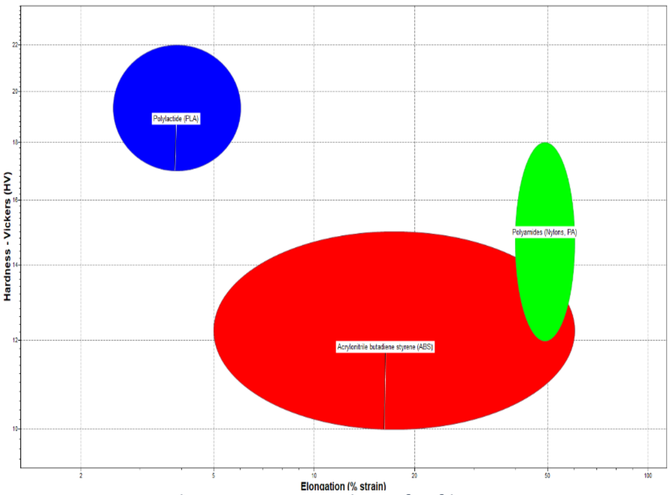


Figure 25: Elongation VS Hardness for filaments

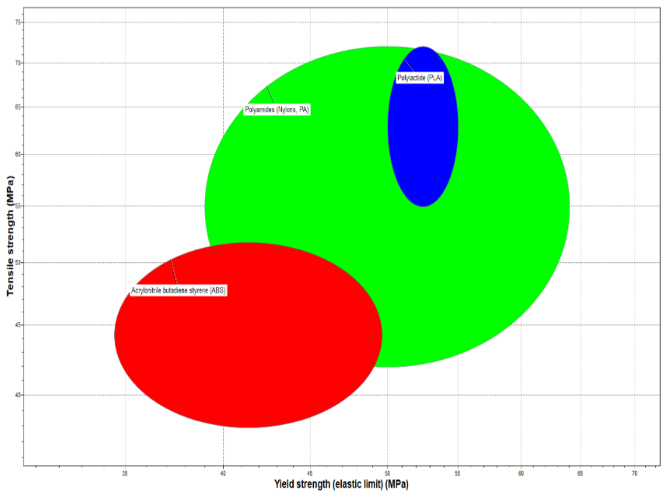


Figure 26: Yield Strength Vs Tensile Strength for filaments

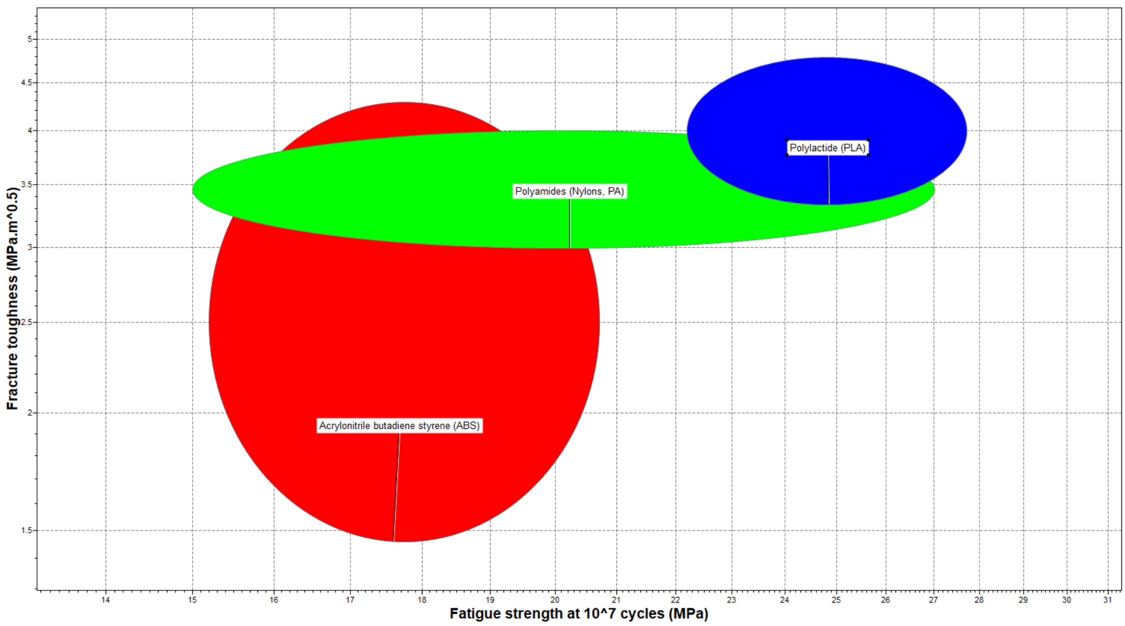


Figure 27: Fatigue Strength VS Fracture Toughness for filaments

Considering the thermoplastic material used for the rigid frame of the exoskeleton it is important to favour the mechanical properties of Strength, Durability, Toughness and Hardness. This is to maximise the strength of support that the exoskeleton gives to an impeded hand and serve as a strong long-lasting frame for the cables to operate within in order to close and open the fingers and hand. The material should be low cost and widely available in order that the design can be sent, customized and printed anywhere in the world with a FDM 3D printer. Flexibility & elongation can also be useful for this frame however, it is a less important property.

The price ranges of the materials (according to Granta) are PLA 1.94-2.47 £/kg, ABS 1.32-1.55£/kg and Nylon 3.37-4.97 £/kg. ABS is therefore the cheapest but not substantially cheaper than PLA. Nylon is the most expensive and the least widely available.
PLA and ABS are the thermoplastics that are most commonly reinforced with carbon fibre, and therefore the technology and resources to reinforce and print with these materials are cheaper and more widely available.

Properties	ABS	PLA
Tensile Strength	27 MPA	43 MPA
Elongation at break (XY axis)	3.5%	2.1%
Density	1.07 g/cm3	1.3 g/cm3
Biodegradable	No	Yes
Glass Transition Temperature	105C	60C
Price (per kg)	$USD 20	$USD 19

Figure 28: ABS vs PLA properties comparison table [1]

As the graphs in figures 24-26 show, Polyactide (PLA) in comparison, despite having a marginally worse elongation and therefore flexibility, has better strength, toughness, hardness, and durability properties.

For these reasons, PLA is the best option for the thermoplastic material to be reinforced with carbon fibre and manufactured into the rigid frame of the exoskeleton. The table below in figure 2816 shows the mechanical properties of PLA.

Mechanical properties

Young's modulus	ⓘ	3.3	-	3.6	GPa
Shear modulus	ⓘ	* 1.2	-	1.29	GPa
Bulk modulus	ⓘ	* 5.7	-	6.3	GPa
Poisson's ratio	ⓘ	* 0.38	-	0.4	
Yield strength (elastic limit)	ⓘ	50	-	55	MPa
Tensile strength	ⓘ	55	-	72	MPa
Compressive strength	ⓘ	66	-	86.4	MPa
Elongation	ⓘ	2.5	-	6	% strain
Hardness - Vickers	ⓘ	17	-	22	HV
Fatigue strength at 10^7 cycles	ⓘ	* 22.2	-	27.7	MPa
Fracture toughness	ⓘ	* 3.34	-	4.79	MPa.m^0.5
Mechanical loss coefficient (tan delta)	ⓘ	* 0.0747	-	0.0793	

Figure 29: Granta PLA properties table

PLA is biocompatible, meaning that it will not cause any chemical harm to the wearer and also biodegradable as it is primarily derived from annually renewable materials: maize, corn and milk. Therefore it is also the more sustainable and environmentally friendly option.

4.4.3 Composite reinforcement

Overview

It has been previously demonstrated that the mechanical performance of a FDM printed component can be significantly improved by integrating composite fibres. The two most common fibre reinforcing techniques compatible with FDM printing were; continuous and short fibre reinforcements. However due to the unidirectional nature of continuous fibre reinforced materials, they exhibit anisotropic properties. This would not be suitable for the purpose of an exoskeleton where multi-directional load bearing capacity is crucial. It was concluded that should it be required, short fibre reinforced composites would be the most appropriate, cost effective and accessible technique.

Reinforcing Material

The most evident variable affecting the performance of a composite is the reinforcing fibre material. There are several different types of fibre reinforcing with different materials and different lengths of fibres. The types of fibrous reinforcement available are carbon fibre, fiberglass, basalt, and Kevlar (Aramid fibrous reinforcement).

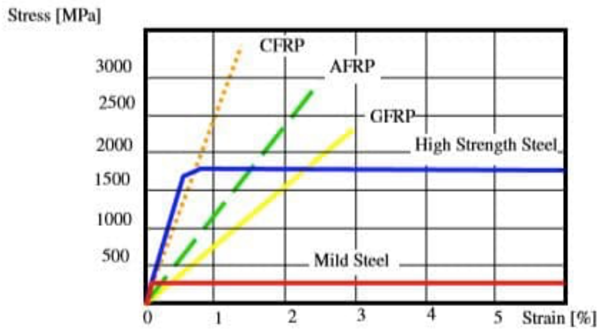


Figure 30: Stress-strain graph of different types of fibrous reinforcement compared with steel [2]

Carbon fibres are the highest performance in terms of strength, and stiffness and are the most commonly used and therefore very accessible. For these reasons, it is a better choice than the more competitively priced fiberglass next-best option.

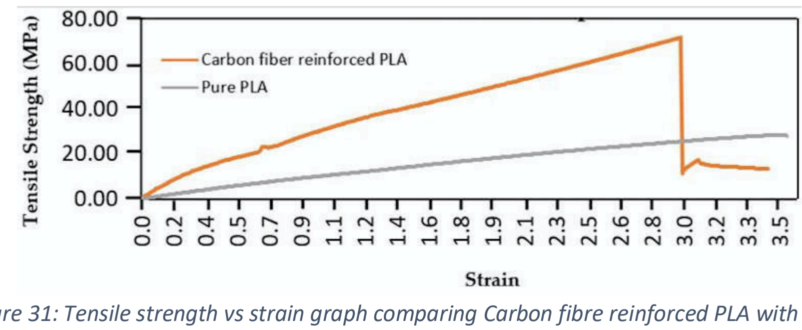


Figure 31: Tensile strength vs strain graph comparing Carbon fibre reinforced PLA with pure PLA. [3]

It is clear from *figure 2* there is considerable improvement in the tensile strength of a PLA when reinforced by carbon fibre. However, in order to optimise both the cost effectiveness and the mechanical performance of the chosen composite, it is necessary to understand how the addition of fibre reinforcements influences the mechanical properties of a composite.

Calculating the Mechanical Properties of a Composite

Short fibre composites materials are also referred to as chaos short fibre composite materials, due to their random distribution of fibres. Consequently, the mechanical properties of a short fibre composite material can be assumed as isotropic. Similar to homogeneous isotropic materials, it has the three elastic constants - Young's modulus E, shear modulus G, and Poisson's ratio μ, with relation of E = G/2(1 + μ).

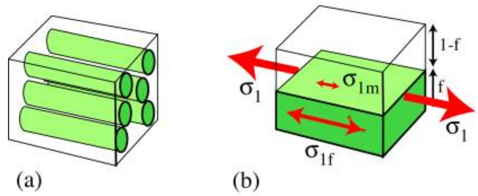


Figure 32: Parallel Slab Model [4]

By modelling a composite as a two parallel slabs, we cab estimate the Young's Modulus of the composite material. Assuming the strain in the fibre is equal to the strain in the matrix:

$$\varepsilon_1 = \varepsilon_{f1} = \varepsilon_{m1}$$

where,

$$\varepsilon_1 = \frac{\sigma_1}{E_1}, \quad \varepsilon_{f1} = \frac{\sigma_{f1}}{E_{f1}}, \quad \varepsilon_1 = \frac{\sigma_{m1}}{E_{m1}}$$

Hence,

$$E_1\varepsilon_1 = E_{f1}\varepsilon_{f1} = E_{m1}\varepsilon_{m1}$$

Assuming the composite fibres are much stiffer than the matrix ($E_f >> E_m$), the reinforcement fibre will also be subject to much higher stresses ($\sigma_{f1} >> \sigma_{m1}$). Consequently, the overall stress σ_1 can be expressed in terms of the two contributions:

$$\sigma_1 = (1 - f)\sigma_{m1} + \mathrm{f}\sigma_1$$

The Young's Modulus is therefore calculated as:

$$E_1 = \frac{\sigma_1}{E_1} = \frac{(1 - f)\sigma_{m1} + \mathrm{f}\sigma_1}{\frac{\sigma_{f1}}{E_{f1}}} = (1 - f)\mathrm{E}_m + \mathrm{fE}_f$$

This is commonly known as the rule of mixtures, and it demonstrates the stiffness of a composite is dependent on only the volume fraction of the fibres and the stiffness of the materials [4].

Volume Fraction of Fibres

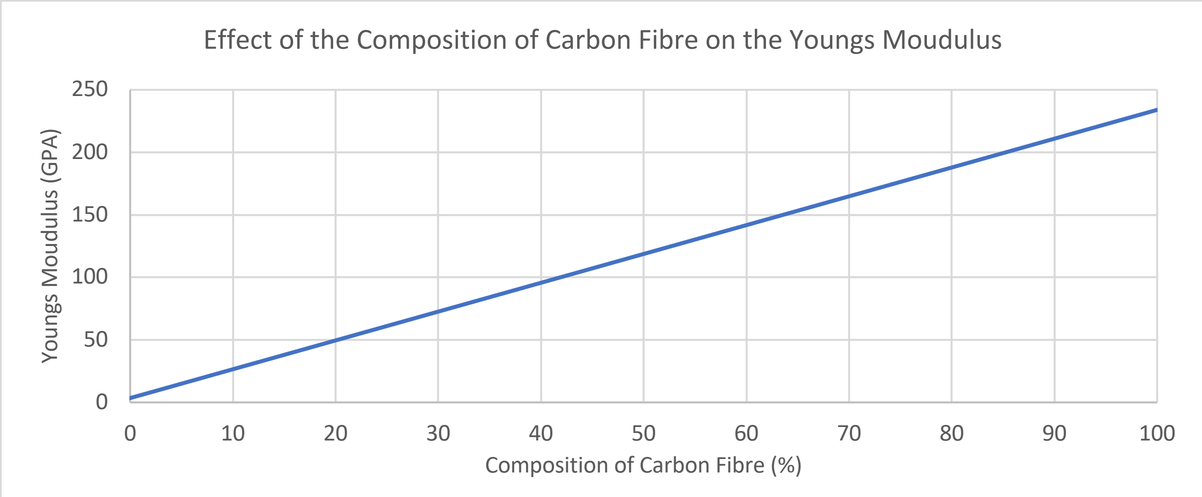


Figure 33: Effect of the Composition of Carbon Fibre on the Young's Modulus of a Composite, Applying the Rule of Mixtures

It can clearly be seen from *figure 4*, the composition of carbon fibre is directly proportional to the mechanical properties of the material. However, in reality this is not true. Polymer matrix composites (PMC) have two major phases – continuous phase and dispersed phase. "The continuous phase consists of different types of organic polymers that serve as the matrix to bind the dispersed phase, usually reinforcing fibers, together for an efficient transfer of load between them. The matrix is important as it serves as a platform to distribute the fibers evenly throughout the structure. Therefore, the properties of PMC depend on the matrix, reinforcement and the interphase." [5]

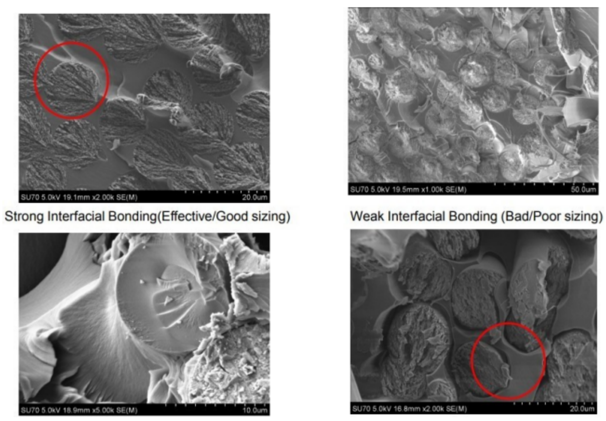


Figure 34: Micro Structure of Carbon Fibre Reinforced Polymer, Highlighting Strong Interfacial Bonding (left) and Weak (right)

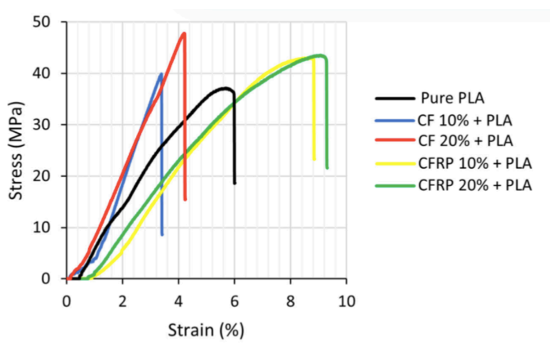


Figure 35: Stress-strain graph comparing different densities of carbon fibre reinforced with pure PLA [6]

Examining the microstructure of a carbon fibre reinforced polymer it is clear that poor interfacial bonding creates voids within the composite inhibiting effective transfer of stress.

Considering the application of the product, the exoskeleton is unlikely to experience extreme loads. It would not be necessary to use a high composition of carbon fibre, a composition of 20% carbon fibre reinforcement would provide a sufficient improvement in strength of approximately 30% *(shown in figure 5)*, whilst remaining cost efficient.

4.4.4 Flexible Spine Material

The secondary structure of the exoskeleton is the flexible spine, which will be attached to the rigid frame and running over the joints, providing the exoskeleton with flexibility and durability to support the joints of the fingers. The material that is used to manufacture this flexible spine must be highly elastic and flexible, tough, fairly hard, durable and 3D printable.

As established, using Fused Deposit Modelling (FDM) 3D printing technology, it is possible to print parts with flexible materials. These materials include thermoplastic elastomers, TPEs, thermoplastic polyurethane TPU or silicone.

Silicone, despite proving itself a very useful material in many other applications, is too soft and not elastic enough to be used as the material for the flexible spine. Therefore, it is necessary to compare

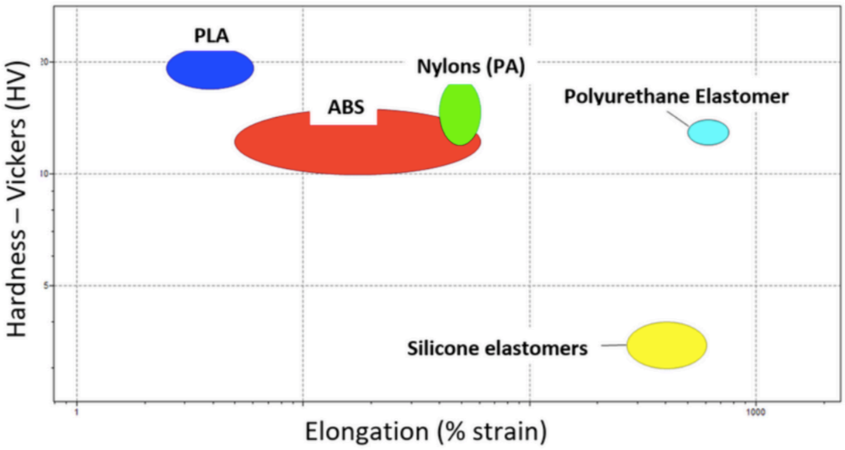


Figure 36: Hardness vs Elongation comparing the flexible 3D printable material options.

TPEs and TPU to determine which material the flexible spine should be 3D printed using.
The advantage of TPEs is that they tend to be more flexible, they have been on the market for a longer time and are therefore more commonplace, widely available and generally less expensive compared to TPU. However, TPU is still flexible but more rigid, tougher, harder and therefore a much easier material with which to print. TPU also perform well in heavier, tougher, and more durable prototypes, whereas TPE filament is more geared for lighter, softer, and more flexible models. [7]

For these reasons, we have chosen to manufacture the flexible spine of the exoskeleton with Thermoplastic Polyurethane Elastomer. This material gives us the critically useful and specification-fulfilling balance of properties; it has very high flexibility and elasticity, yet a high toughness and abrasion resistance, good tear strength, a low compression set, and high resilience and durability. TPU, possibly not to the extent of TPEs however, is also reasonably priced and widely available.

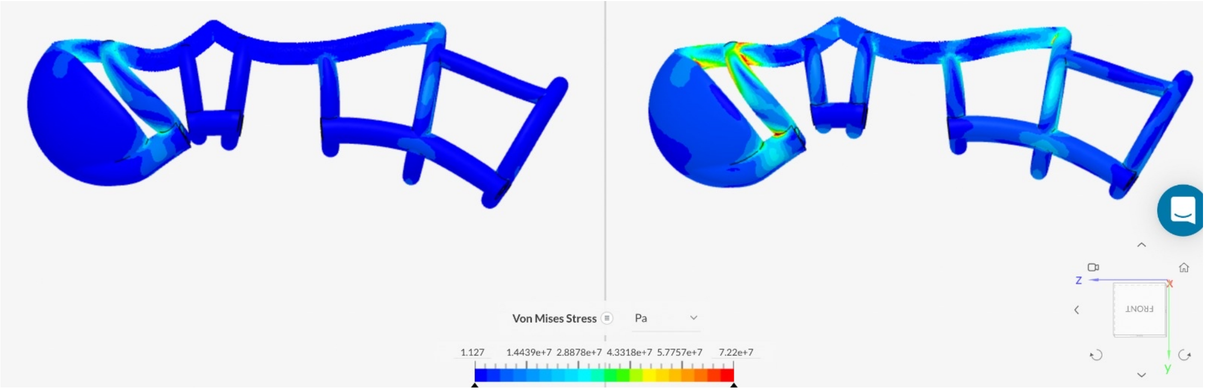


Figure 37: FEA Analysis Comparing Pure PLA (right) and Short Fibre Reinforced PLA (left)

It is clear from *figure 37* the improvement in mechanical performance when reinforcing the exoskeleton with short carbon fibres. The composite material is able to with stand much higher stresses than Pure PLA making it more suitable.

4.4.6 Summary

Part	Material
Finger Exoskeleton Rigid Frame	20% Carbon Fibre Reinforced PLA
Flexible Spine	TPU
Actuator Housing	20% Carbon Fibre Reinforced PLA

4.5 Design

4.5.1 Developmental Process and Reasoning

In our initial CAD models, we had extremely simplified designs since the main goal was to simply get a general idea of how we wanted the general mechanics to work in a visualised manner. As a result, the models were bulky and inaesthetic, containing a great deal of material for a device that had to fit ergonomically with a hand. In order to develop our design in a principled direction, we followed our standards of accessibility and mechanical performance at an effective cost. The key goal in mind was to put out an exoskeleton that could not only be manufactured in less capable locations but also provide a seamless experience for the user.

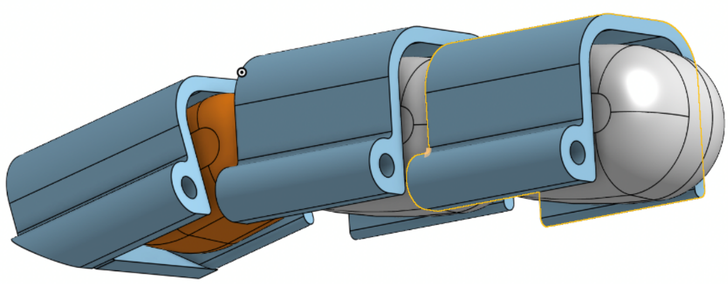


Figure 37: The blue components are the exoskeleton, and the grey and orange bits are the finger. The orange segment is the tip of the finger.

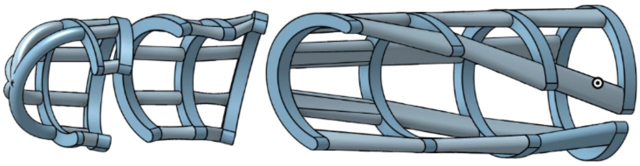
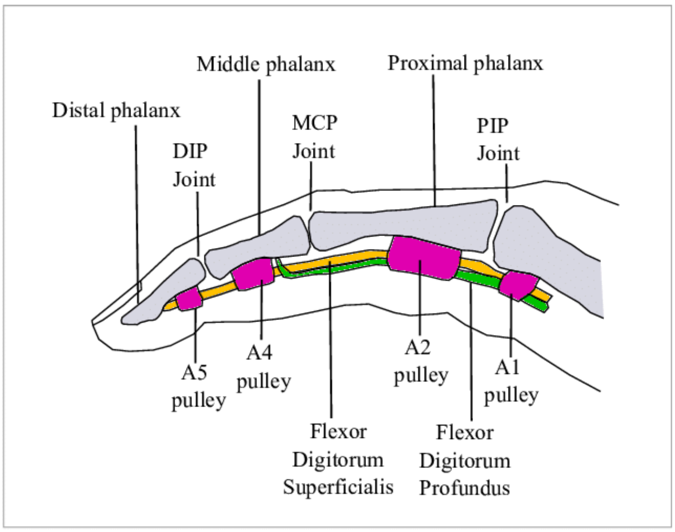


Figure 38: A much more complexly fitted early design pointing to the left.

Figure 35 hows one of the earliest and simplest models. It was composed of three rigid segments, one for each section of the finger, with the end cap having a slight lip at the tip to catch the finger if pulled up. There are channels for which a wire (connected to the tip) can be pulled through on one side to close the finger, mechanically similar to a real finger. The connecting elements between segments are not shown but are meant to be some elastic material that keeps the exoskeleton positioned correctly. On the other hand, Figure 36 shows a much more elaborate frame that wraps itself around the finger; the mechanics are generally the same in how they interact with the finger.

As mentioned before, our refining process follows a thematic based approach in order to create an exoskeleton that represents our principal vision. Additionally, a good amount of the design changes we made throughout the development cycle were highly intertwined. For example, the one-sided channels of the Figure 35 design is unstable due to the biased nature of its positioning. Under high loads, it may twist the exoskeleton off, so, to resolve this issue, we added a channel to the other side to even out any load distribution from the tensioned wire. This simple change not only improved the stability but also functions in the same way as a pulley, halving the load required from the actuators for the same amount of load. There is a huge cost benefit to having less powerful motors as well as the reduced stress on the wire channels.

Since our design methodology involves having multiple designs, we are always able to pick and switch based on objective performance. In this case, the design from Figure 36 offered more advantages, especially in the user experience. The model is less bulky, so it contains less material which is great for the user since it is lighter and less costly to produce. Culling the excess material slims down the design



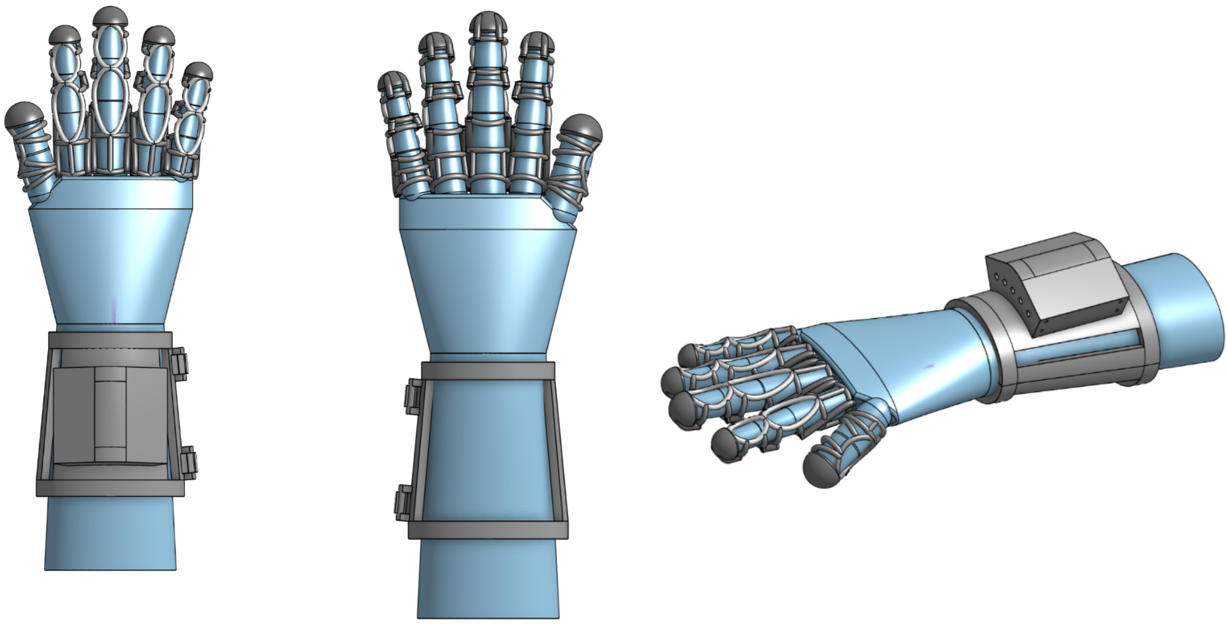
to a more aesthetically pleasing form factor in both movement and appearance. Mechanical performance is not sacrificed, yet cost has gone down while improving the actual user experience.

Figure 39: This is a diagram of the musculoskeletal system of the finger, including the tendons and ligaments [1]

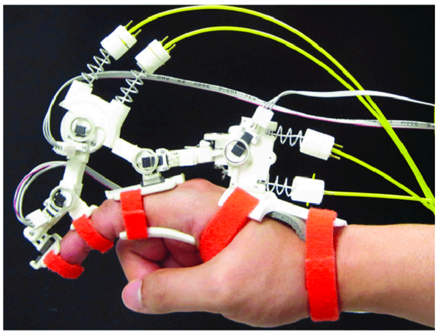
One common aspect of the design is the elastic connection between the rigid components of the exoskeleton and how it is actuated. Our goal was to replicate the musculoskeletal structure as seen in Figure 37, the main reason being its compactness and possibility of moving most of the actuators away from the fingers where there is limited space. Most importantly, Figure 37 shows the connection of the flexor digitorum profundus to the tip of the finger (distal phalanx) and the connection between the flexor digitorum superficialis to the middle finger segment (middle phalanx). In our case, the flexor tendons are combined into one wire and connected to the tip; when the wire is contracted, the end cap is pulled into the hand, pulling the rest of the finger with it.

This action is as close to a replicate of the real human model as possible. Emulating the human body's mechanics is highly ideal since we are utilising the existing musculoskeletal structure, excluding the activation of muscles. When contracted and carrying a load, the tensioned wire works in tandem with the compression of the finger's skeletal structure to distribute the load in a form of tensegrity. Therefore, the three exoskeleton pieces act as an external finger skeleton

4.5.3 Comparison of Proposed Design with Current Designs



With the final design, it is necessary to do a comparison against other designs to verify the key benefits our design provides in relation to others. The main objective is to be the best in the way that matters for an effective mechanical design that is accessible in production and manufacturing.



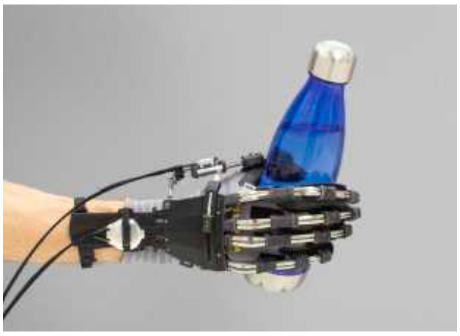


Figure 40: The top left image is of a finger exoskeleton with series elastic actuation for rehabilitation [2]. The top right image features the RELab tenoexo, a robot hand orthosis [3]. The bottom image is that of our group's design.

	Advantages	**Disadvantages**
Finger Exoskeleton	Fine position and torque control for each finger segment	Bulky, inaccessible due to complexity
RELab Tenoexo	Lightweight, ergonomic, slim	Limited strength (5N individual fingertip force [3]), complex assembly
Current Design	Slim, large force, lightweight, simple, 3D printable, strong	Limited control (positionally and force-wise)

Our current design serves both a functional purpose and is unobtrusive in the user's normal life. The design is an overall success in terms of the core objectives, providing an accessibility in production (3D-printing) and manufacturing (simplicity of construction). The design optimises force capacity, and the composite material offers great structural strength. Additionally, the open nature of the frame offers breathability and adaptability with minimal contact; the frame is lightweight and aesthetically pleasing, incredibly important in long-term usage. Given the complex dynamics of the thumb, we did not pursue the means of actuating it and instead decided to splint it in place. This is discussed further in the future works section below.

We also designed a wrist frame for housing the electronics and linear actuators on the lower half of the forearm. It is a simple two-piece frame with hinges that snap together on one side and a Velcro strap or buckle on the other. As with the fingers, material is limited to reduce weight. The electronics and circuit schematic in this housing can be found in Appendix B.

4.5.4 Finite Element Analysis

In this section we will use finite element analysis (FEA) to show the efficacy and soundness of our design. We will also compare it to our initial design to show the benefits in the changes. All printed parts will be simulated with the material properties of the carbon-infused PLA specifications as below.

Material	Young's Modulus (E)	Poisson's Ratio (v)	Density (ρ)	Yield Strength

	Pa	-	kg/m^3	Pa
Carbon PLA	1.3e+10	0.34	1130	88e+6

4.5.5 Initial to Final Comparison FEA

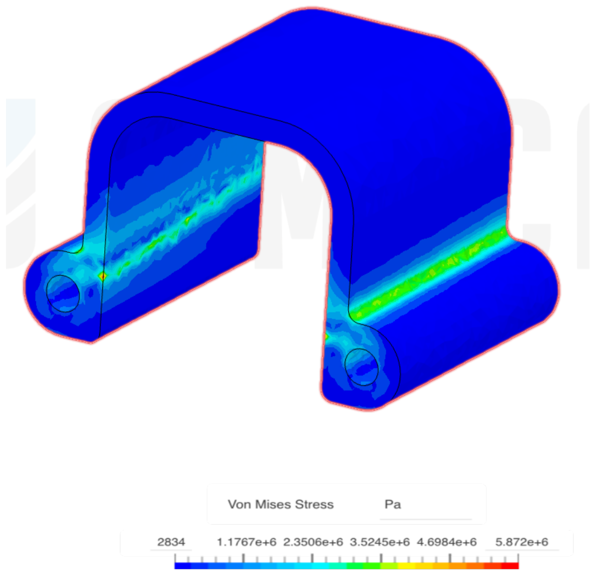


Figure 41: FEA of the initial frame design.

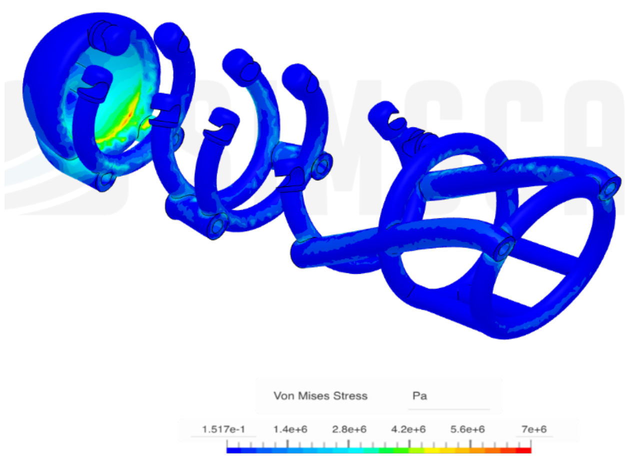


Figure 42: FEA of the final exoskeleton for the finger.

In Figure 5, the frame has a force boundary condition on the wire channels in the downwards direction. There is a finger that the frame rests upon but has been taken out for visual clarity of the stresses on the frame; the finger has a fixed support boundary condition and does not move, so the frame must press down on the finger. In Figure 6, there is also a fixed support finger that goes through the frame for which it can be pressed against. The wire channels have a force boundary condition that is directed inwards orthogonal to the general extended finger linkage. This condition is meant to represent the force of the wire pushing against the channel walls when contracted and under some load. Finally, there is a boundary condition at the tip of the finger where the wire wraps around; it is a force condition that is directed inwards into the hand, pushing the end piece into the tip of the finger. All three force boundary conditions are 120 N.

From the results, we can see that the initial bulky design fares well with the example load; however, the large amount of blue material suggests an abundance of unused material, which can be optimally removed. The final design also has a lot of blue regions under the yield criterion but contains far less material around the finger. The wire channels continue to experience the most load-bearing force in addition to the end cap, and the rest of the frame is there to just hold the channels in place.

4.5.2.2 Clip FEA

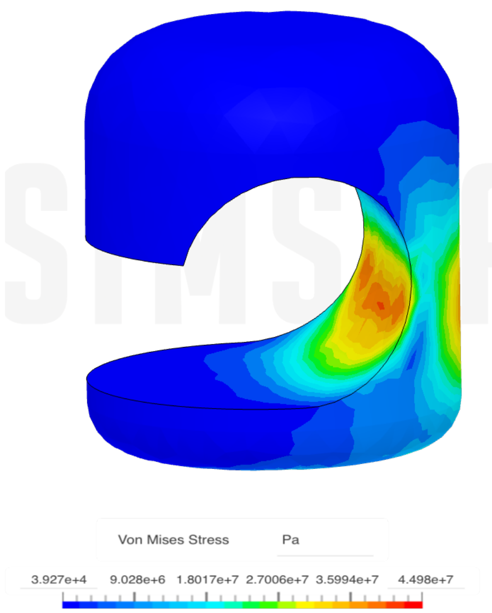


Figure 43: FEA of the clip (for holding the PU band) on the final design.

One other important aspect of the current design that should be tested is the structural integrity of the clips holding the elastic band. When the finger contracts, there is some amount of strain applied to the band which, although elastic, will mean there is some inward force on the clip. This force will not be large, but given the small size of the clip, it is still important to test it out.

As seen in Figure 7, the FEA results prove that the clips do indeed work and will not break from the straining of the elastic band. The bottom of the clip has a fixed support boundary condition while the inside face has a 10N force directed away from the opening orthogonal to the base. This is to simulate the band pulling against the clip. The maximum von Mises stress is below the yield strength and so fulfils the yield criterion.

4.5.3 Summary

Our design has changed greatly with respect to the core objectives from the initial designs, and it now meets all the core requirements. Material has been shaved off in an optimised manner, making it more cost effective and aesthetic, and the structure is solid given the strength of composites. The FEAs verify as such.

5. Conclusion

With so many people in the world currently suffering from a life altering disability, the inhibited use of their hand, the drive to create a viable biomimetic exoskeleton to aid rehabilitation and promote independence is only getting stronger. With more funding and more research into this area than ever before, the links between a cost effective, strong exoskeleton and a neural interface is what our project was developing and researching.

We were able to conclude that wireless, neural interfacing, wearable technology exists in the form of an ‘armband’. We were also able to establish that this interface could be used to control miniature linear actuators and push-pull cables to mimic the behaviour of tendons and muscles within the arm and hand. These powerful mechanical components allow the user the ability to pick up objects in the ‘power grip’ and lift weights up to 2kg. In addition to this, the exoskeleton design itself is made from two separate materials, a flexible spine made of TPU and a rigid strong frame made from carbon reinforced PLA. This gave the exoskeleton finger the ability to bend over the joints yet support the rest of the finger, especially under compressive and tensile forces.
Finally, we were able to manufacture the exoskeleton using FDM 3D printing and we were successfully able to print one exoskeleton finger as proof of concept and viability for future work. We also did a cost estimation of the overall product which came to $469.50 and this can be found in Appendix C.

We hope that in the future, there is opportunity for more development in this area. Research into an opposable mechanism and dual extrusion FDM 3D printing is a promising and interesting option which will only further the functionality of our project beyond what we have already achieved.

6. Future Work Proposals

6.1 Overview

The main constraints of our project were that we had limited time to research, design and then manufacture our product and the budget was also restricted to £75. Unfortunately, the cost of even the neural armband itself would be $150 which was out of our price range already.

Because of this, there were a couple of proposals that we had that we did not consider during our research and design but that we were aware of. In addition to this, there were problems that we encountered during the 3D printing of our design that we were able to find a solution for but there were also other ways of fixing those issues. This section of the report will detail these issues.

6.2 Thumb Opposition Proposal

The first proposal that we initially considered was the design of the thumb. The thumb is proven to be the most freely moving digit. It has a 'saddle joint' which is when the joint has concave and convex portions that fit together [1]. The thumb can move in many directions, including flexion and extension (as the other fingers on your hand do). However, the thumb is able to move in opposition, abduction, adduction, and retropulsion also [2] which gives it a wider range of motion and makes the joint more complex to mimic mechanically.

Figure 44: Manual Thumb Opposition

Given more time, we would've liked to develop a system that would allow the thumb to move in opposition. A possible design solution for this would be similar to the RELab Tenoexo which is a robotic hand orthosis designed by ETH Zurich. Their design features a manually operated thumb opposition slider as pictured in *figure 42* [3].

A possible improvement to this design would be to replace the manual aspect of the design with an actuator (and cable) to push & pull the thumb in opposable movements instead of the manual slider. This actuator could be linked to the neural interface armband so that its movements are controlled using muscle signals, as the other fingers in the hand are. It would increase the bulkiness of the design we currently have somewhat but it is a promising idea as it greatly increases the function of the device if the thumb has closer movement capabilities to that of an uninhibited hand.

6.3 3D Printing Proposal

The other problem that we encountered during our 3D printing stage was the difficulty in printing our two different materials together and having them be connected at certain places. Our design solution during the project was to print the spine and the frame separately and to CAD small clips into the frame for the parts to clip together.

The two materials (TPU for the flexible spine & Carbon Fibre reinforced PLA for the rigid frame) are both capable of being 3D printed. Currently, you can print multiple materials using an FDM printer, which is the one we are using. This involves using a 'dual extruder' printer. Our materials require very different slicer settings to print well and we would have to change the nozzle and bed temperatures, the print speed as well as the retraction settings [4] to switch between them due to their differing properties. It is absolutely possible to print TPU and PLA in one design and have them bonded together, recommended guidelines include applying a slow print speed (30mm/s) and using a direct drive extruder for TPU specifically [5].

The University of Edinburgh 3D printing facilities unfortunately did not have the capacity to do this complex dual extrusion 3D printing however, the technology exists and it will only become more accessible in the future as more research and monetary support goes into 3D printing technology. This is why we believe it is a viable option for the future of our product. Furthermore, it will both improve the quality of our design and allow us to print the design in one go (rather than separately) and will allow the spine and frame to be bonded permanently rather than clipped together.

6.4 Summary

If we were able to spend more time researching and developing these proposals, they would elevate the design and functionality of our device for future users.

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7.3 Methodology

7.3.1 Neural Interface

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None

7.3.5 Design

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8.0 Conclusion

None

9.0 Future Work

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10.0 Appendix

10.1 Appendix A

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8.0 Appendices

8. 1 APPENDIX A

The figures below detail the Raw ConvNet classification & Linear Discriminant Analysis (LDA) classification.

Figure 44 details accuracy across 4 training cycles for each armband and Figure 45 details the confusion matrix. Both figures are for the Raw ConvNet classification. Figure 43 is the ConvNet Architecture and it is capable of using the raw sEMG signal for gesture recognition [1].

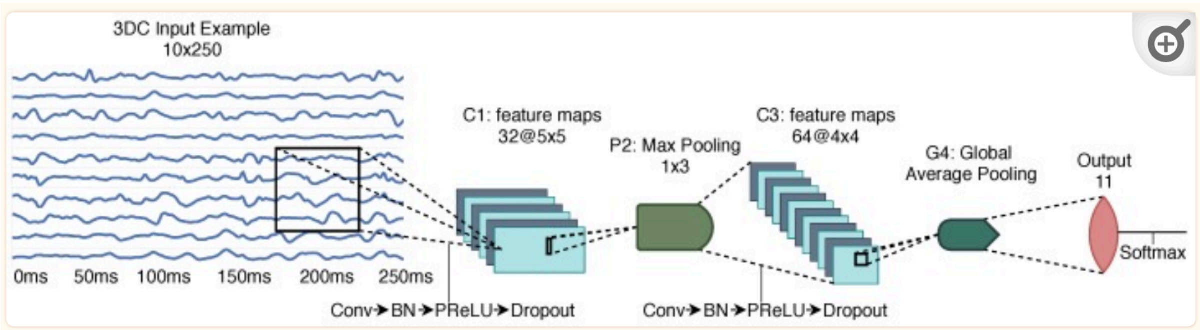

Figure A1: The raw ConvNet architecture employing 34,667 parameters [1]

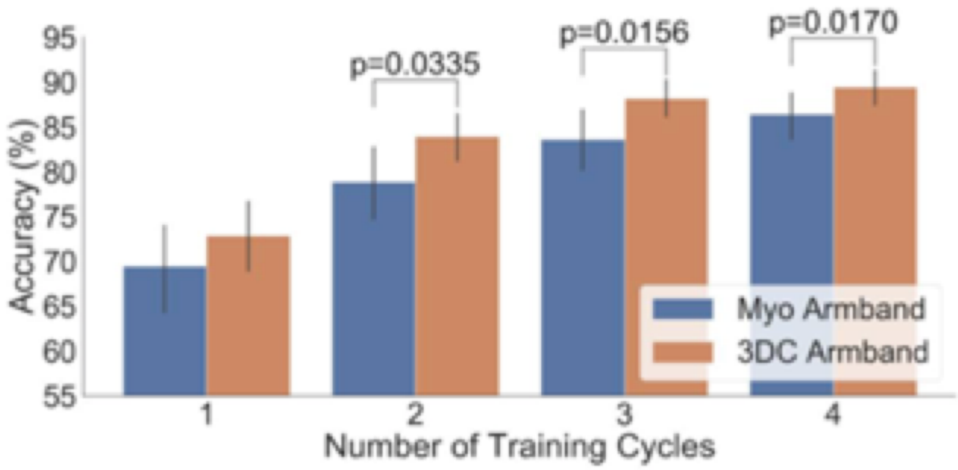

Figure A2: Accuracy of the Armbands for the Raw ConvNet Classification

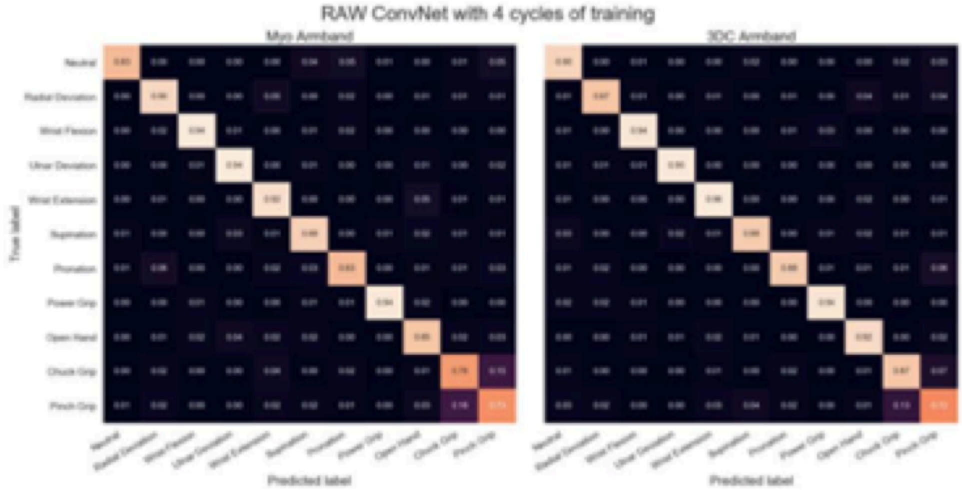

Figure A3: Confusion Matrices for the Raw ConvNet Classification

Figure 3 details accuracy across 4 training cycles for each armband and Figure 4 details the confusion matrix. Both of these figures are for the Linear Discriminant Analysis (LDA) classification. The LDA Classification is one of the most commonly used sEMG-based gesture recognition classifications due to its timely and computational efficiency classification while still being able to achieve high classification accuracies [1]. It also does not need any hyperparameter optimization unlike other classifications.

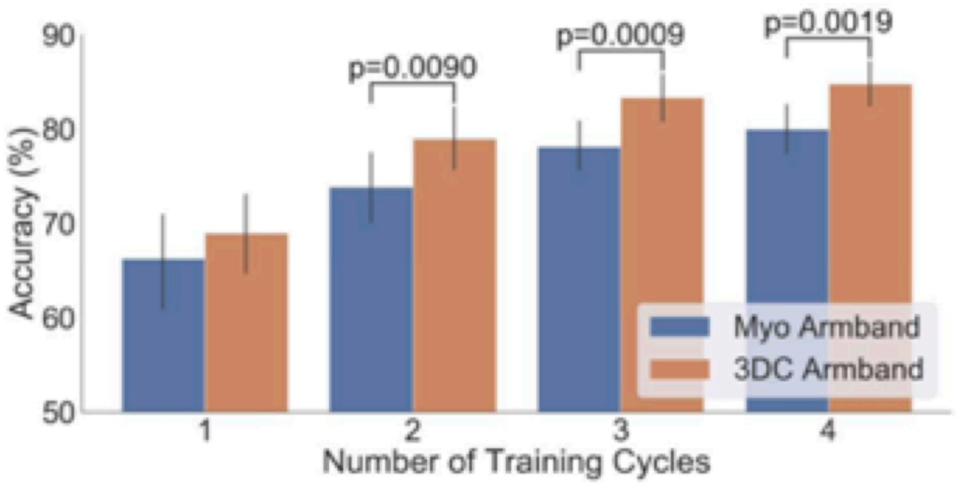

Figure A4: Accuracy of the Armband for the LDA Classification

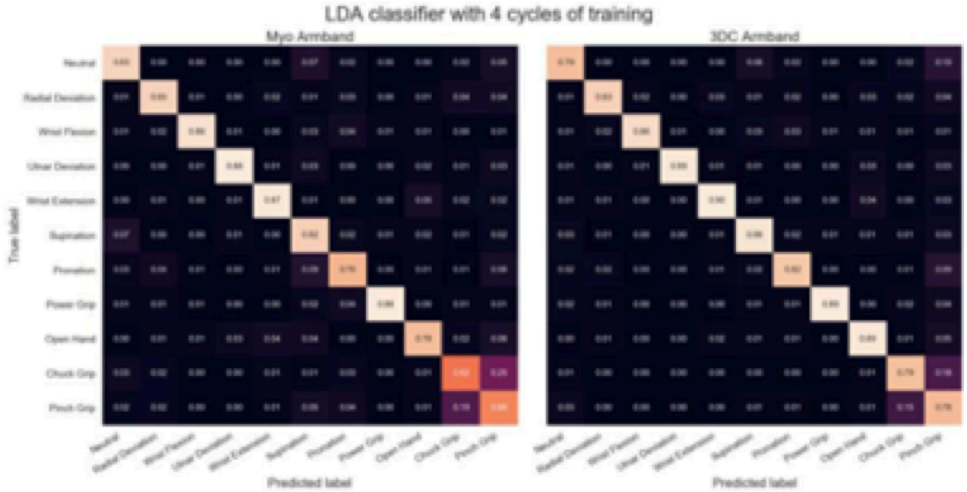

Figure A5: Confusion Matrices for the LDA Classification

They add further support to the performance of the 3DC Armband as opposed to the Myo Armband.

8.2 APPENDIX B

The electronics are highly essential for the actuation and control of the hand exoskeleton. However, it is also important to stay in line with our core objectives of accessibility and efficacy. We, therefore, want components that are cheap and simple to use while also providing all the necessary functionality we need in a consistent manner.

Arduinos are popular and proven microcontrollers, costing very little, while offering an immense array of functionality. The Nano line is one of their smallest microcontrollers which can come equipped with low energy Bluetooth. The L298N is a robust dual motor driver capable of controlling the motor direction through an H-bridge and its speed through pulse width modulation (PWM). One component offers all the necessary functionality. The L16-S linear actuators are the perfect size to limit bulkiness while being easily integrable.

8.2.1 Schematic

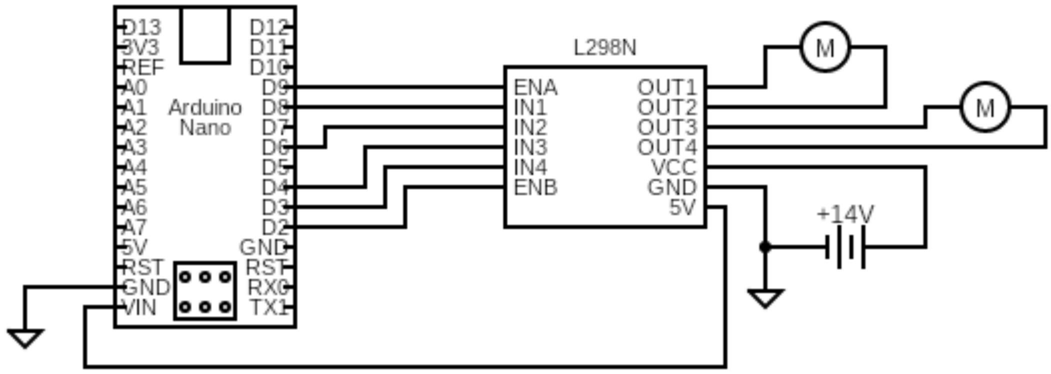

Figure B1: Schematic of circuit

The above schematic displays the simple electronic circuit. The 14V battery is connected to the L298N motor driver across the VCC and GND pins, while the 5V pin on the driver supplies the necessary voltage to the Arduino Nano. The two L16-S linear actuators are coupled to the motor driver, which takes the control commands through its input pins. All electrical components are connected to a common ground. The linear actuators are rated for 12V; however, the motor driver has a voltage drop of 2V, so the battery needs to be at a nominal 14V for the actuators to run at their rated performance.

8.2.2 Power Consumption

We can assume that, at a maximum, the user is running the linear actuators at full power for two hours a day while the Nano is running continuously throughout the day in the background.

	Voltage	Current	Time	Watt Hours
Nano	5 V	20 mA	24 hours	2400 mWh
2x Linear Actuators	12 V	650 mA	2 hours	31200 mWh
Total				33600 mWh

There is a voltage drop of 2V across the motor driver, so the battery will have to supply 14V in order for the linear actuators to run at their rated 12V.

33600 mWh / 14V =2400 mAh

Therefore, based on the predicted power consumption, we will need a 14V battery good for 2400 mAh.

8.3 APPENDIX C

Cost Estimation of the Device is shown in table C1 below.

Sections	**Specific Components**	**Price per component**	**Overall section cost in our design**	**Total**
Neural Interface	3DC Armband	$ 150.00	$ 150.00	**$ 150.00**
Mechanics	Push/pull wires	$ 30.00	$ 30.00	**$ 310.00**
	Actuators	$ 70.00	$ 280.00(4 actuators)	
Materials	Carbon 20% PLA Filament	$ 35.00 (500g)	$ 7.00 (for the 5 fingers)	**$ 9.50**
	TPU Filament	$ 40.00 (500g)	$ 2.50 (4 fingers)	
				$ 469.50

Table C1: showing expected cost of our design assuming the 3D printer is already in place