DESIGN, DEVELOPMENT AND NUMERICAL VALIDATION OF NEURAL-CONTROLLED HAND-PROSTHESIS

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This article presents the mechatronic structure of a neural prosthesis developed within the NerveRepack project, emphasizing both its functional design and structural integrity. The prosthesis was designed to enable bidirectional communication with a neural interface equipped with motor and sensory electrodes. The mechanical components were modeled in SolidWorks 2024 and fabricated through 3D printing using PLA material. To evaluate functionality, simulated signals replicating motor neural activity from the median and ulnar nerves were applied. Alongside functional validation, structural analysis using numerical methods was conducted to assess the durability and load-bearing capacity of the prosthetic components. The results confirm that the NerveRepack prosthesis can be effectively interfaced with the peripheral nervous system of upper-limb amputees, ensuring both mechanical reliability and personalized motor function.

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Keywords: Control System; Exo-prosthesis; Design, validation and fabrication.

1. Introduction

According to the World Health Organization, over 60 million people worldwide live with limb amputations, more than 10% of whom are affected by upper limb loss [1], [2]. Existing commercial prosthetic solutions are predominantly based on myoelectric control and provide only a limited set of predefined functions, insufficient to support the full range of daily activities such as hygiene, feeding, and self-care. Consequently, individuals with upper limb amputations often face significant social and professional exclusion, being unable to reintegrate into the labor market. Many become dependent on family support and minimal disability benefits provided by governmental programs.

The fundamental limitation of myoelectric prostheses lies in the restricted number of usable myoelectric signals generated by the residual musculature in the amputation stump [3], [4]. In response to these limitations, recent years have witnessed an increasing interest in neurally controlled prosthetic systems, supported by systematic efforts from multidisciplinary research teams. These efforts encompass advances in neural interface design [5], biocompatible materials for long-term implantation [6], signal decoding algorithms [7], and robotic actuation mechanisms [8]. Notable contributors include institutions such as MIT's Biomechatronics Group, the Max Planck Institute for Intelligent Systems, and the Imperial College London Neural Prosthetics Lab. These developments reflect a broader, coordinated push toward more natural and responsive control systems for upper-limb prosthetics.

Neurally controlled prostheses overcome the limitations of myoelectric systems by utilizing motor neural signals directly acquired from peripheral nerves, such as the median and ulnar nerves within the amputation stump. The number and specificity of motor commands achievable are significantly enhanced using implanted motor electrodes. The neural prosthesis developed within the NerveRepack project, for instance, employs plug-type electrodes, as described in [9], which enable the recognition of 31 distinct movement patterns. This functionality is made possible by the implantation of three motor needles in the median nerve and two in the ulnar nerve, allowing fine-grained control tailored to the user's intent. Each neural signal pattern corresponds to a unique movement command, thus greatly expanding the range of motion and autonomy offered to the patient.

Beyond the increase in the number of controllable motions, this approach enhances the user experience by providing more natural, fatigue-resistant operation, reducing the cognitive load typically associated with learning to use conventional prosthetic devices. Moreover, the integration of such neural interfaces opens the

path for future bidirectional systems, which may restore not only motor control but also sensory feedback, further improving embodiment, motor learning, and the overall quality of life for individuals with upper limb amputations.

2. Design of the neural prosthesis mechatronic structure

The hand prosthesis presented in this article was developed as part of the NerveRepack project, in which all the authors of this article participate. This prosthesis was designed for a patient with right hand amputation.

The patient mentioned in this study was selected based on a protocol approved by the ethics committee of the partner institution, according to clear eligibility criteria, which included the level of amputation, overall health condition, and willingness to participate in the research. Participation was voluntary, and the patient signed an informed consent form in accordance with national and international ethical standards for biomedical research. All activities involving the human patient from this project was conducted in accordance with the Helsinki Declaration of 1975, as revised in 2008, and Regulation (EU) 2017/745 of the European Parliament and of the Council of 5 April 2017 and the patient signed an informed consent. In all activities that was involving the voluntary participation of the patient, the three essential ethical principles were respected: 1) the principle of respect for the person; 2) the principle of benefit and 3) the principle of equity. The enrolment of the patient was based on inclusion criteria in the protocol and was consist of patient understanding and signing of the Informed Consent. Inclusion of the human patient in research was absolutely voluntary.

The development of the mechatronic structure was initiated based on detailed anthropometric measurements of both the amputation stump and the healthy limb of the patient enrolled in the project. The finger dimensions of the prosthesis were based on the dimensions of the fingers on the patient's healthy hand. The anatomical parameters of the residual limb, along with the corresponding dimensions of the patient healthy hand, were accurately recorded for the design process (Figure 1, Figure 2 and Table 1).

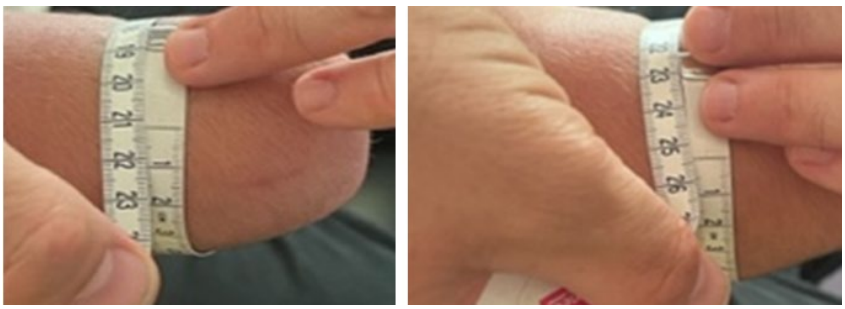


Fig. 1. Measuring the anatomical dimensions of the patient's amputation stump

The mechatronic structure was designed in SolidWorks 2024, based on the anatomical dimensions of both the patient's amputation stump and the contralateral healthy hand, as part of the NerveRepack project. The geometry of the palm and each finger was developed according to the required movement profiles of the prosthesis, which were defined during the preliminary analysis phase (Figure 3). Each finger's motion is actuated by a cam-and-connecting-shaft mechanism, driven by a dedicated motor and linear actuator, selected based on the initial mechanical and kinematic specifications. The initial kinematic specifications included the range of motion of the fingers, the speed and accuracy of actuation, the force required for gripping, as well as the number of degrees of freedom needed to reproduce functional and natural movements. The chosen kinematic solution for actuating each fingertip is illustrated in Figure 4 and Table 2.

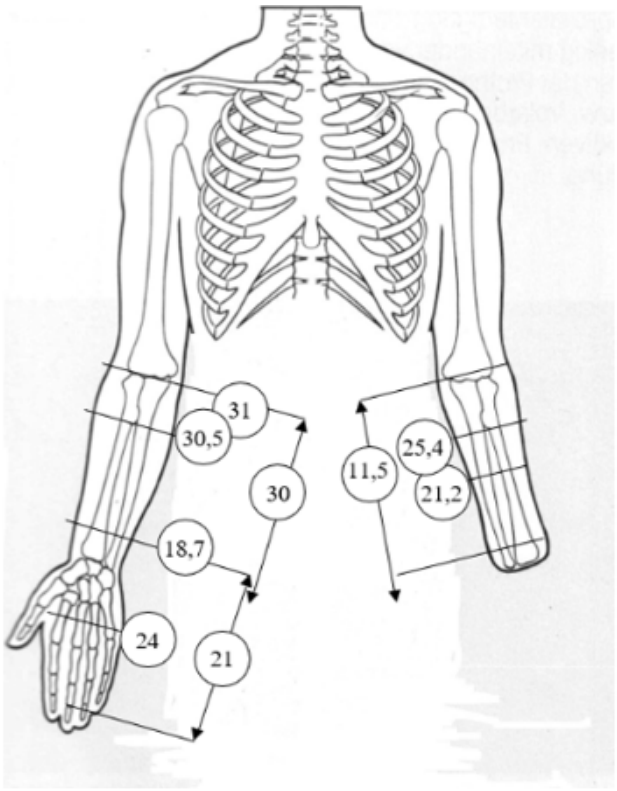


Fig. 2. Anatomical dimensions for the amputation stump and the patient's healthy hand

Table 1

Dimensions of the healthy arm and the prosthesis

	Length	Width	Thickness
Healthy Palm	21	24	3
Healthy Forearm	30	-----	18,7 / 30,5 / 31
Stump	11,5	----	21,2 / 25,4

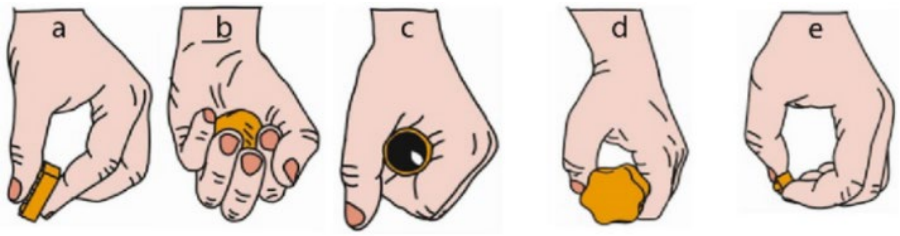


Fig. 3. Different types of hand grasp, prehension, and hook [6].

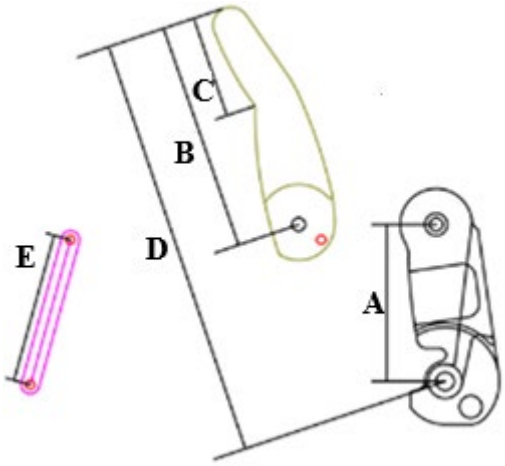


Fig. 4. The cam design - shaft connecting mechanism for each finger, actuated from a specific motor through a linear actuator.

Table 2

Dimensions of the connecting mechanism for each finger

	A	B	C	D	E
Finger 2	35	50,9	22,7	85	33,55
Finger 3	43,3	50,9	22,7	87	41,5
Finger 4	38	45,9	20,9	80,6	36,4
Finger 5	35	38,9	17,35	70,75	33,55

The 3D designs of the tip, middle part, and base of the index finger mounted on the palm and driven by a motor and a linear actuator are shown in Figure 5.

To facilitate the flexion and extension movements, a mechanism was designed in which the proximal phalanx, the distal element, and the metacarpal are interconnected. The movement of the distal phalanges is achieved through a pin connected to the distal element. For these reasons, related to simplifying the finger kinematics, the distal and middle phalanges were designed as a single piece. From a mechanical perspective, the movement of the kinematic chain associated with each finger can be performed in a single plane.

Therefore, abduction and adduction movements are not achievable with a mechanical system of this design. However, fundamental movements such as gripping, releasing, and object manipulation can be performed with high precision when an appropriate actuation system is used.

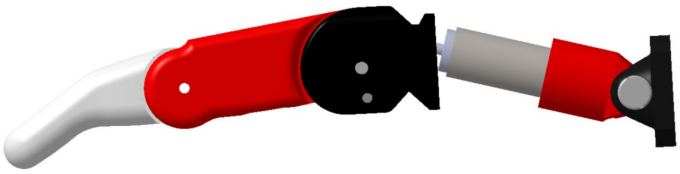
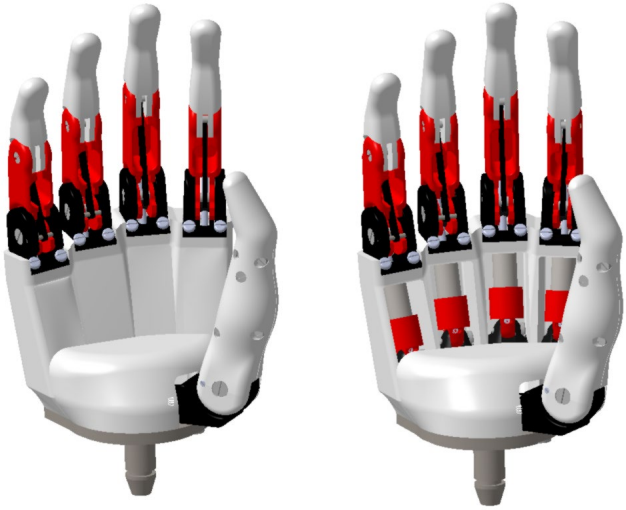


Fig. 5. The design of the index (pointer) finger of the prosthesis, including the base support system in the palm and the actuator device, as designed and assembled through SW 2024

The mechatronic structure of the prosthesis, which includes the mounting position of the motors and linear actuators, is presented in Figure 6.



a) b)

Fig. 6. a) A global view of the overall design of the exoprosthetic hand, b) DC motors positioning details

After designing the mechatronic structure of the prosthesis, a stress and strain analysis was carried out using the finite element method (FEM) to identify the critical points of the stress state and the deformation field within the structure. Preparing the computational model was essential, as the mechatronic structure of the prosthesis includes components with complex geometries that are not relevant

to the analysis. In the initial stage, all elements that did not influence the stress and strain distribution such as screw heads or chamfers without a functional role in stress transfer were removed. This resulted in a simplified model that was easier to discretize (Figure 7) and allowed for a more efficient solution. In the areas of interest, where stress concentrations occur, additional Boolean division operations were applied to achieve a finer mesh and more accurate results.

Fig. 7. Geometry simplification and revolute joint configuration

Stress concentrators, also known as stress raisers or stress concentration points, are features within the material or the structure that cause a localized increase in stress. These areas can significantly impact the mechanical integrity and performance of the materials, especially when subjected to loading. Stress concentrators are critical in the design and analysis of structures because they can lead to failure if not properly accounted for. To achieve a finer mesh around the screw through holes, each area was divided by Boolean operations. Thus, the revolution kinematic torques of the prosthesis model were processed in such a way that they can be controlled and refined around the stress concentrators. In Figure 8, the waist elements of the simplification of the phalanges and the kinematic joints are presented.

The definition of kinematic joints was realized according to the elements through which movement is achieved. To have a movement of the superior phalanx, all the degrees of freedom of each individual finger were defined by means of kinematic couples. Figure 9 shows the translational movements of the nut that sets the entire finger assembly in motion. Since the mechanism by which the motion of the electric motor is transmitted to each middle phalanx is a lead screw-ball nut type, the primary motion is the translation motion of the nut. This movement will move the entire finger assembly. Considering the movement of the real prototype, all the revolute joints of the entire kinematic chain were also defined.

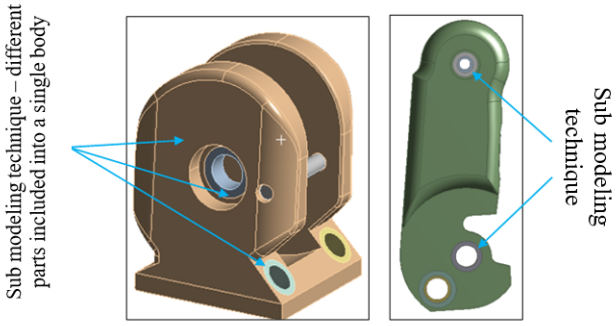


Fig. 8. Geometry simplification and revolute joint configuration

For generating the calculation model, a concentrated force of 50 N (equivalent to approximately 5 [kg]) was considered. This value was chosen based on data from the biomechanics literature, which indicates that the average gripping force exerted by a human finger during typical daily activities ranges between 30–60 [N], depending on the object and task complexity [9].

The selected force of 50 [N] represents a realistic upper-bound scenario for evaluating the mechanical performance of the prosthetic finger under functional loading conditions. Additionally, it allows for conservative stress estimation in critical components, ensuring structural integrity and reliability of the prosthesis during regular use. Tetrahedral and hexahedral elements were used in the areas where no stress concentrators appeared, while the maximum size of the elements was 1 [mm]. For the elements in the stress concentrator area, the mesh is controlled locally by tetrahedral elements, imposing a maximum size of 0.2 [mm]. All these discretization settings resulted in a computational model with 645770 nodes and 446053 meshing elements.

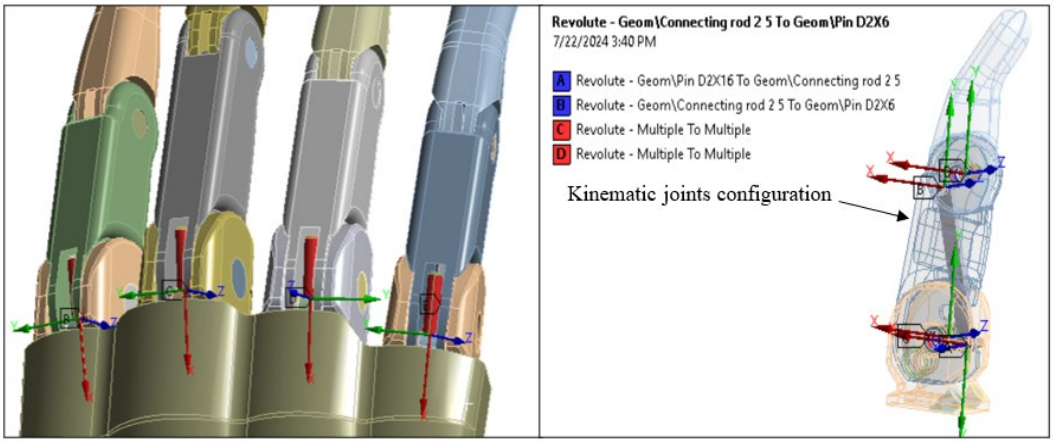


Fig. 9. Kinematic joints definition for each finger

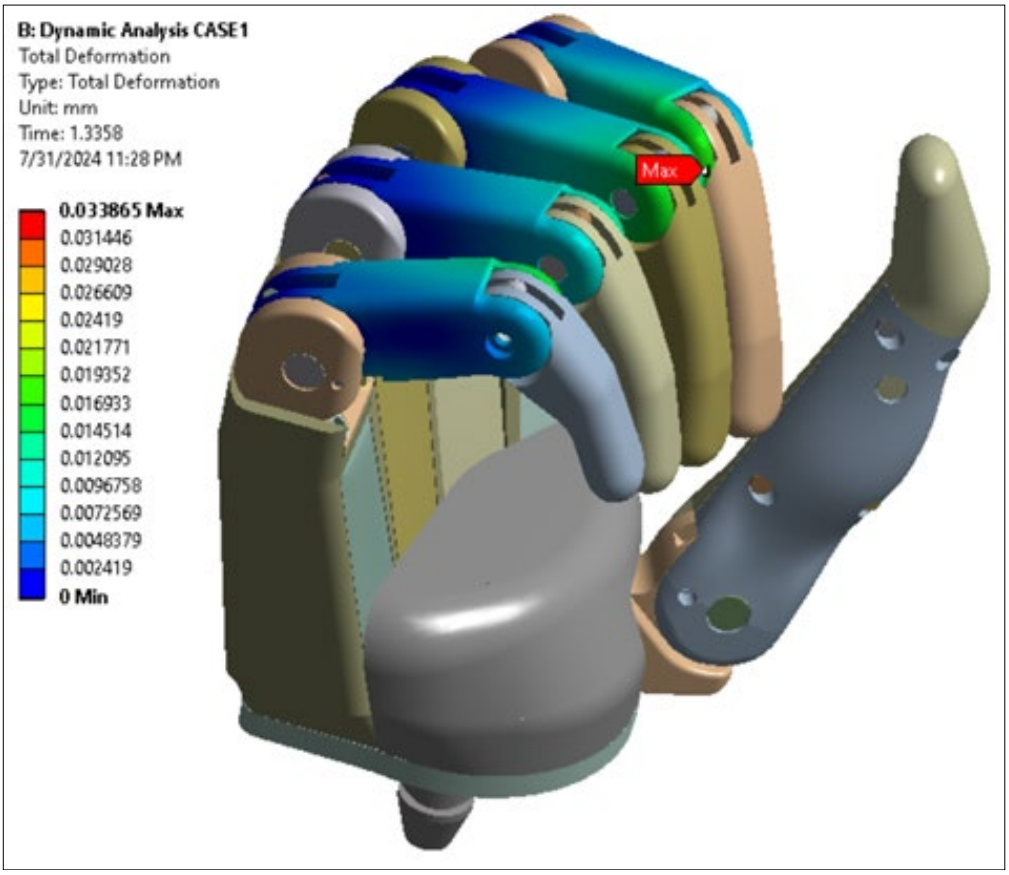


Fig. 10. Concentrated mass position

The stress and strain distributions to which the prosthetic mechanism is subjected were obtained through the conducted numerical analysis. Figure 10 illustrates the total deformation across the finger structure, along with the critical points where maximum strain occurs. The results indicate a maximum displacement of approximately 0.33 [mm], which is considered minimal in the context of dynamic loading. Such a small deformation suggests that the prosthetic structure maintains sufficient rigidity and mechanical stability during actuation, without compromising the accuracy or repeatability of finger movement.

Figure 11 presents the results obtained in the metacarpal elements, highlighting the distribution of von Mises equivalent stresses during mechanical loading. The maximum stress value reaches 34.805 [MPa] and is localized in the contact area between the shaft and the first metacarpal element, suggesting a possible stress concentration due to initial movement or the application of a dynamic load. The remaining components exhibit significantly lower stresses, mostly below 10 [MPa], indicating stable structural behavior in the support regions. These results underscore the importance of optimizing the geometry and material

of the connecting shaft, which represents the critical point of the analyzed metacarpal assembly.

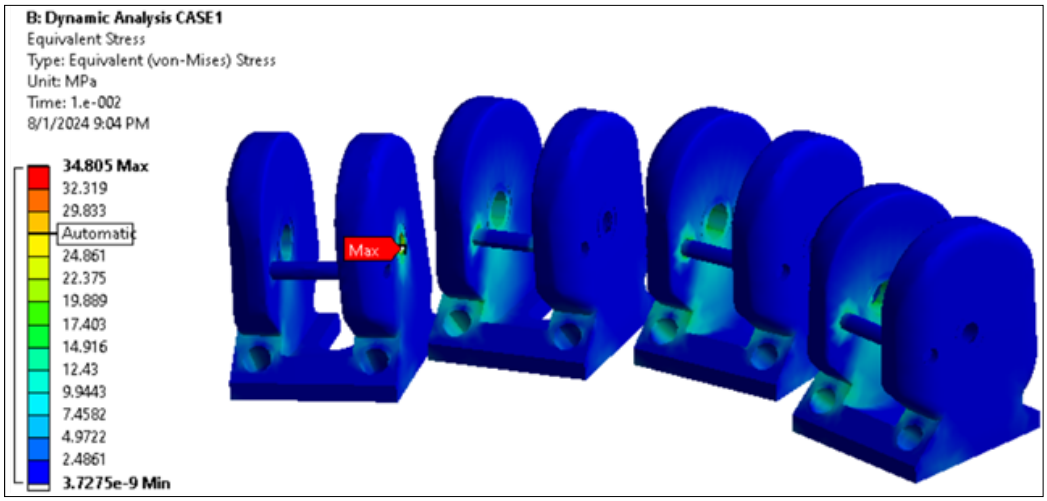


Fig. 11. Equivalent von Mises stress obtained using PLA

Figure 12 presents the result of the dynamic analysis performed on the proximal phalanx, highlighting the distribution of equivalent von Mises stresses under mechanical loading conditions. The maximum stress reaches a value of 35.8 [MPa] and is located near the mounting area, around the fastening holes, indicating a potential stress concentration caused by geometry and mechanical contact. The rest of the component shows low stress values, below 10 [MPa], suggesting a favorable distribution in the central and lateral regions of the phalanx.

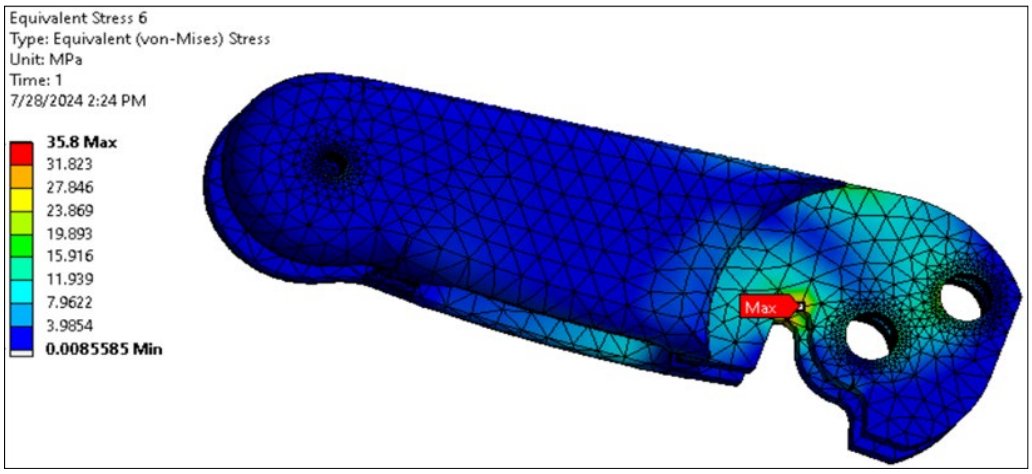


Fig. 12. Equivalent von Mises stress obtained in the joint between the connector and the fingertip using Duralumin alloy.

The mesh is properly refined in critical areas, ensuring good result accuracy. Considering the biomechanical nature of the part and the fact that it is part of a hand prosthesis, the area of maximum stress should be carefully analyzed to prevent structural failure, by optimizing the shape or adjusting the material used.

In Figure 13 is depicted the distribution of von Mises equivalent stresses for the mechanism's link plate. The maximum stress value, 16.335 [MPa], is localized in the contact area of the lower hole, indicating a stress concentration in this critical region. The rest of the component exhibits significantly lower stresses, suggesting stable structural behavior in most areas.

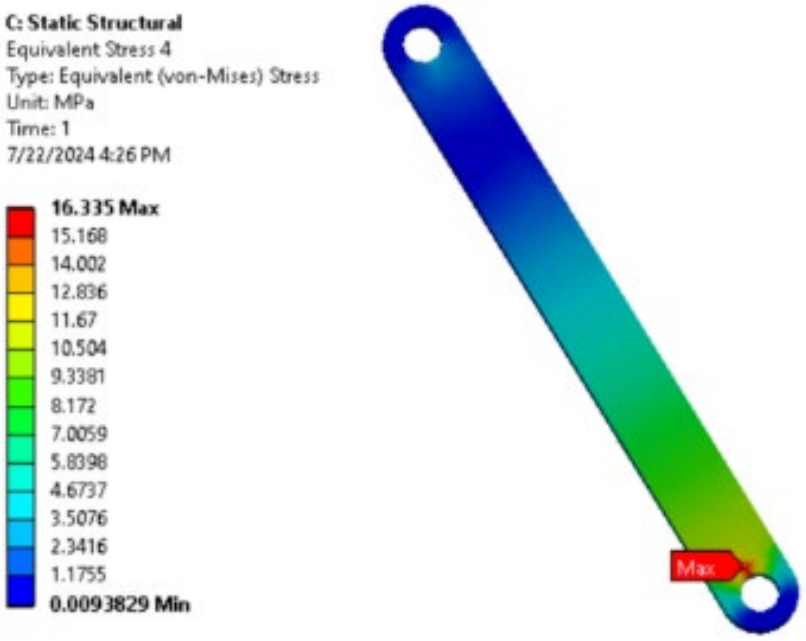


Fig. 13. Equivalent von Mises stresses obtained using Duralumin alloy for the connecting rod.

The metacarpal structures are subjected to significant mechanical stress, considering the use of PLA as the manufacturing material. Given that the material's yield strength is 43.83 [MPa] and applying a safety factor of 1.3 it can be inferred that these components present an elevated risk of yielding when exposed to external loading conditions. The safety factor of 1.3 was chosen because the structures are subjected to relatively controlled loads, and their failure does not pose a critical risk. This value optimizes material usage and component weight, maintaining a balance between safety and structural efficiency.

The use of non-reinforced PLA does not satisfy the necessary strength requirements. Accordingly, the following calculations demonstrate that the stress developed within the metacarpal structures may lead to the initiation of cracks or structural failures, which are undesirable for the reliable operation of such a system.

$$\sigma_a = \frac{\sigma_c}{c},\ where\ c\ is\ the\ safety\ factor,\ \sigma_c = Yield\ strength; \quad (1)$$

$$\sigma_a = \frac{43.8}{1.3} = 33.69\ [MPa] \quad (2)$$

From equation (2), the threshold value of the von Mises stresses expected in the metacarpal components can be identified. In the evaluated scenario, the equivalent von Mises stress reaches a maximum of 34 MPa ($\sigma_{\max\ \_equivalent\ stress}$), a value that exceeds the material's allowable strength when considering a safety factor of 1,3.

$$\sigma_{\max\ (equivalent\ stress)} \geq \sigma_{a\_PLA}, (using\ a\ 1.3\ safety\ factor) \quad (3)$$

Given that the maximum equivalent stress according to the von Mises criterion is below the yield strength of the material (43.83 [MPa]), the use of standard PLA is justified when considering a minimum safety factor. However, since the application involves the fabrication of a prosthesis where structural safety is a critical requirement it is recommended to use carbon fiber reinforced PLA due to its superior mechanical properties, such as increased stiffness and improved resistance to fatigue and long-term deformation.

In calculation relation 4, the data refers to the case of using a carbon fiber reinforced PLA, with a yield strength of 53.4 [MPa].

$$\sigma_{a_{PLA+CF}} = \frac{\sigma_{c\_PLA+CF}}{c} = \frac{53.4}{1.3} = 41[MPa] \quad (4)$$

$$\sigma_{\max\ (equivalent\ stress)} \leq \sigma_{a_{PLA+CF}}[MPa] \quad (5)$$

3. Fabrication and testing of the prosthesis mechatronic structure

The selection of raw materials for the fabrication of core body components in upper limb Exo-prosthesis is based on the structural analysis presented in the previous section. Specifically, the choice of materials is guided by the distribution of equivalent stresses and the resulting nodal displacements obtained from the simulations. In this context, polylactic acid (PLA), potentially reinforced with carbon fibers, has been identified as an optimal material for the manufacturing of rigid structural elements and internal inserts of the forearm, palm, and fingers. PLA is a thermoplastic, plasticizer-free polymer available in various commercial forms including pellets, films, and filament. It exhibits a low density (1.24 g/cm³), excellent mechanical performance, water resistance, low electrical conductivity, and thermal stability under moderate room temperature conditions. Furthermore, PLA is biodegradable and recyclable, and it supports flexographic and offset

printing without the need for surface pretreatment. These attributes make PLA a suitable and sustainable option for use in the additive manufacturing of prosthetic components subject to mechanical loading.

Prototype parts of the exoprosthesis structure were manufactured to apply Design for Manufacturing (DfM) principles in the final structure and verify all the analytical and numerical findings that were applied in the study. Each part was 3D printed in BAMBU LAB X1 CARBON COMBO–3D Printer with high-quality printing with 7 [μm] Lidar Resolution and a High-speed Core XY with 20000 [mm/s²] acceleration. Each part was printed separately, and the transmission and motion mechanisms were assembled to be tested for their functionality. The procedure was continued with 3D printed production of parts and fitting of parts to achieve a smooth assembly with appropriate tolerances and accuracy.

The motor block of the prosthesis includes five Faulhaber Series 1016M006SR DC motors (one for each finger). The choice was mainly made due to the space constraints imposed by the mechatronic structure and because of the balance in size and power that they provide. They fit in the mechatronic structure and provide the required mechanical power for adequate finger movement and grip force. The motors work at a nominal voltage of 6V, with a torque constant of 4.39mNm/A, more than enough to emulate the strenght of a healthy human hand. One of the main advantages of this motor is their form factor, allowing them to fit in the mechatronical structure and providing high spatial efficiency. The current consumption of the motors is also low, drawing from 9.1[mA] (no load) up to 2A (stall).

The motors were controlled using five DC drivers (Polulu 8838), an external signal generator (Artix-7 XC7A35T-1CPG236C FPGA), and a 6V power source. They were generated five PWM signals (one for each finger) with varying duty cycles and five digital signals, which change the motor's direction of rotation. By changing the duty cycles of the PWM signals, it was tested the mechatronic structure at different movement speeds. The mechatronic structure of the prosthesis is presented in figure 15.

The prosthesis was equipped with a motor interface for implementing personalized movement algorithms (figures 16).

The motor interface is composed of a controller board, a glove with finger flexion electrodes, and a graphical interface that displays glove and prosthesis movements. The interface is implemented in C++ and offers visual cues about the state of all devices on the prosthesis. This includes the state of the motors (if they are idle or functioning and their consumption) and the state of the pressure sensors mounted on the exo-prosthesis.

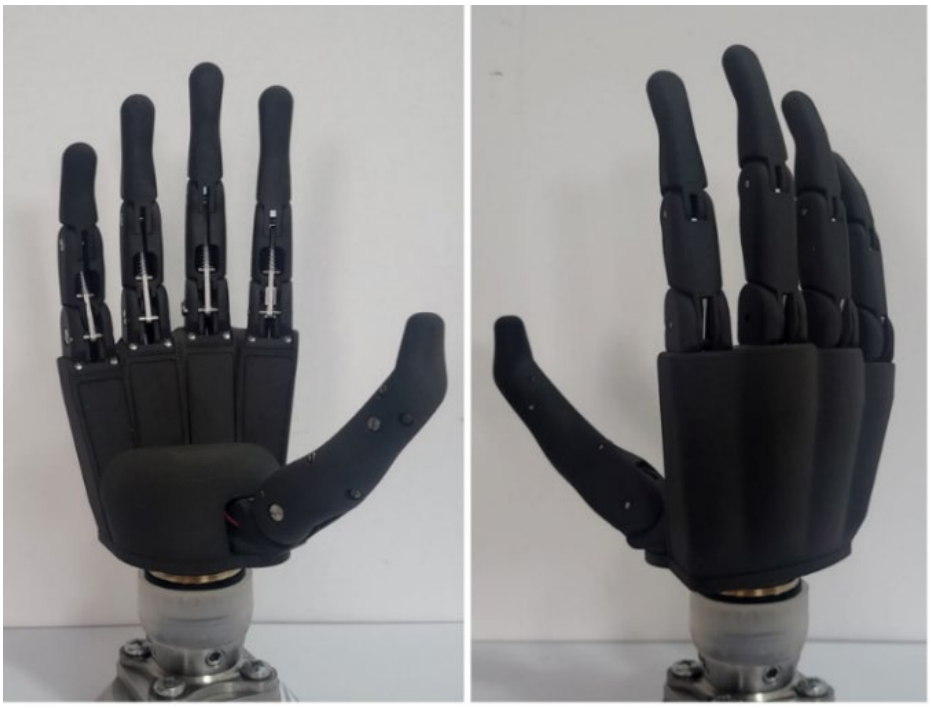


Fig. 15. The mechatronic structure of the prosthesis

The movement is handled inside the controller and is triggered by the movement of the 5DT Data Glove. The controller samples the positional values sent by the glove and decides, for each finger, if they are either flexed or relaxed, based on a threshold. If the sensor value exceeds the threshold, then the finger is flexed, otherwise it is relaxed. All positions of the fingers are then mirrored by the prosthesis by generating drive commands, which are sent to the mounted motors. The movement of an exo-prosthesis finger stops when the amperage consumption of a motor exceeds a certain threshold, which indicates mechanical blockage. A certain movement is considered to be completed when all fingers have stopped because of exceeding the amperage threshold.

The motor interface offers the patient the added advantage over other prostheses on the market of being able to upload personalized movement algorithms into the prosthesis control system. To define personalized movement algorithms, the patient will put the glove on their healthy hand and perform the movements they want to perform with the prosthesis with their healthy hand, and these movements will then be loaded with the help of the motor interface into the prosthesis control system. Figure 17 and Figure 18 show the graphical interface and the tests performed to define personalized movement algorithms for the prosthesis.

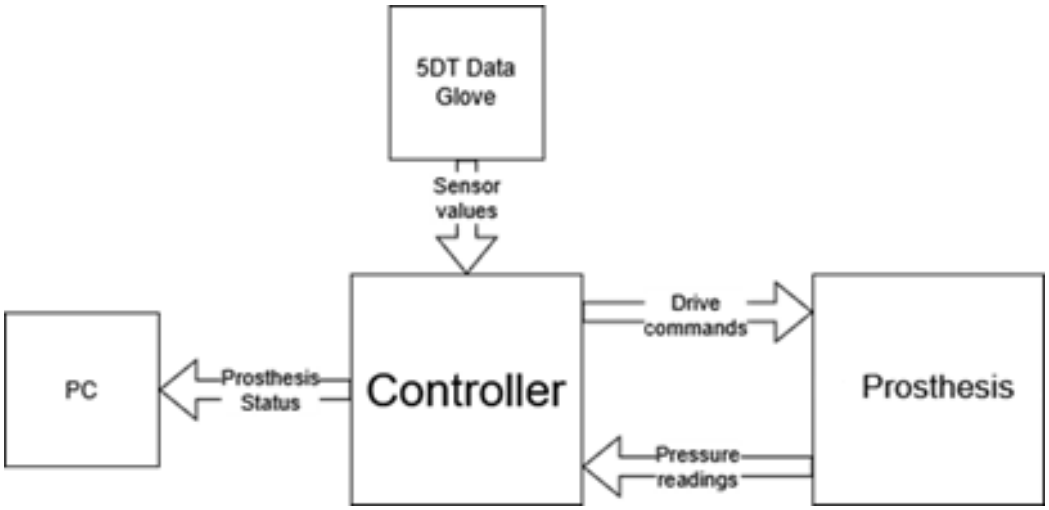


Fig. 16. Motor interface diagram

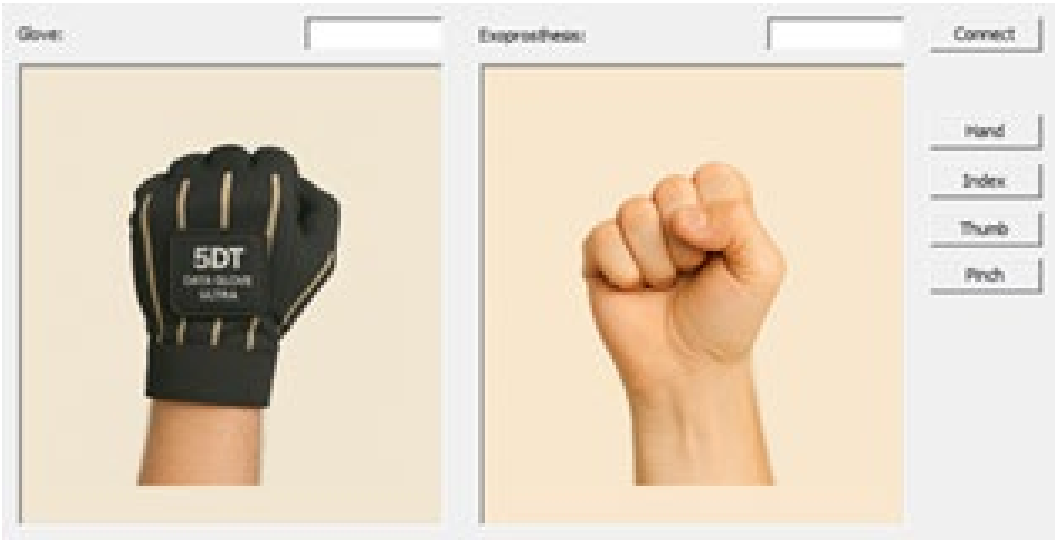


Fig. 17. Grafical interface for testing the connection between the glove and the prosthesis

The prosthesis in the NerveRepack project will be controlled with neural signals collected from the patient's stump. For this, a neural interface will be surgically implanted in the patient's stump, which will collect neural signals from the motor fascicles of the nerves from the patient stump, and will transmit them via wifi to the prosthesis control system. The neural interface has been tested so far in vivo and is presented in [10].

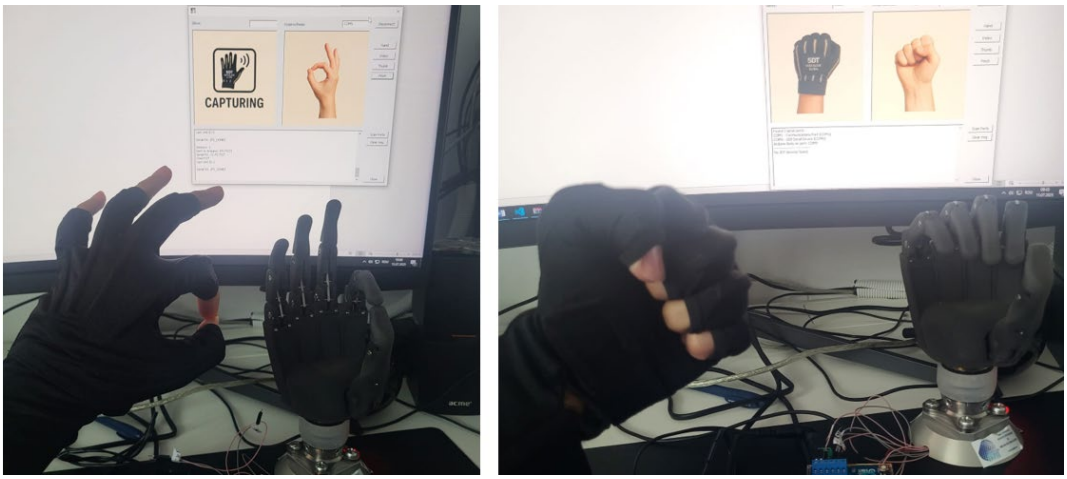


Fig. 18. Testing the functionality of the motor interface

The hand prosthesis was designed to be equipped with resistive pressure sensors, (RFP602, 5Kg, compatible FSR 402, 10MM [11]) mounted at the base of the motors that actuate the prosthesis fingers. The conception, design and implementation of pressure sensors is the subject of patent application no. A/00170/13.05.2025 [12]. When the fingers of the prosthesis grasp an object, the mechanical pressure exerted by the fingers on the object is transmitted through the structure of the drive system of each finger to the pressure sensor at the base of the sole of the motor, and then the sensor transforms the mechanical pressure into an electrical signal which is then transmitted to the sensory interface of the prosthesis and is displayed on the graphic interface.

The prosthesis' sensory interface is responsible for receiving signals from pressure sensors in the prosthesis' fingers and transmitting them to the prosthesis' control system and the graphic interface for real-time monitoring. When the prosthesis fingers grip an object, the controller also samples the readings from the pressure sensors. The sensors mounted on the prosthesis are FSR402 resistive tactile sensors (Figure 19), whose resistance decreases at a linear rate, depending on applied force (Figure 20).

Figure 19. FSR402 resistive sensor

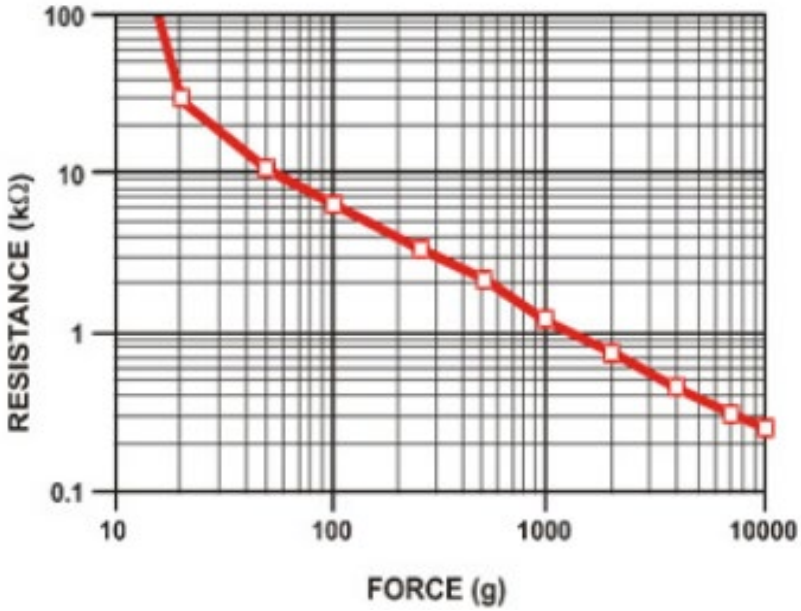


Fig. 20. Force – Resistance graph for the FSR402 resistive sensor

To better measure the force applied to the tactile sensors, it had implemented a voltage divider followed by a repeater (as shown in the tactile sensor datasheet). The result is a force (g) – voltage (V) dependency that grows approximately linearly in the 100g – 1000g force range. The voltage reading resulted from the tactile sensor circuit is injective, so each voltage level can be associated with a unique value of force applied to the sensor. The injective dependency between force and resistance is a very important factor, as each force value will produce a unique sensation, which grows in strength as the applied force grows.

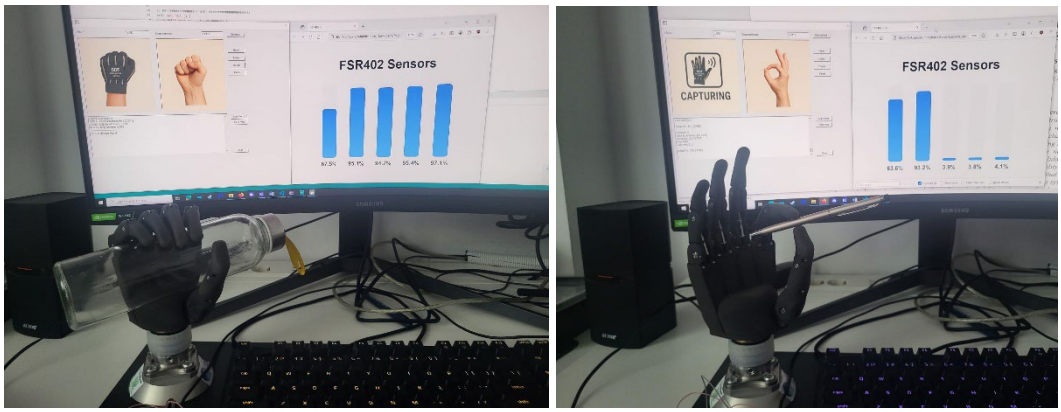


Fig. 21. Testing the functionality of the motor interface

The hand prostheses from the market use two types of systems to implement the tactile feedback function:

- small vibration motors placed on the skin of the stump;
- small motors that exert pressure on the skin of the stump;
- devices that generate skin stretching in different areas of the stump;

All these devices act indirectly because they generate sensations of pressure, vibration or skin stretching on the amputation stump to the patient, and the patient must learn to interpret these sensations and associate them with the movements of the prosthesis fingers when it manipulates different objects [13], [14], [15], [16], [17]. The prosthesis from the NerveRepack project uses a tactile feedback system that acts with electrical neural signals directly on the sensory fascicles of the median and ulnar nerves in the patient's stump. The tactile feedback system of the NerveRepack prosthesis will provide the patient touch sensations and pressure sensations similar to those they had before the amputation.

Tests carried out with pressure sensors mounted on the sole of the prosthesis motors showed that they can detect both the fingers touching an object and pressing forces equivalent to 5 kg. The tests carried out thus showed that the solution implemented in the NerveRepack prosthesis is similar to tactile feedback systems based on flexible wearable pressure sensors mounted on the fingers of the prosthesis [18], [19], [20]. In the case of flexible wearable pressure sensors, many problems must be solved related to the way they are mounted on the fingers, the way they are connected to the signal processing system, how to troubleshoot defects that may occur in these sensors during the operation of the prosthesis, how to replace these sensors when they fail during the operation of the prosthesis, etc.

Within the NerveRepack project, the prosthesis presented in this article will be equipped with a transmission/reception module to communicate bidirectionally with the neural interface that will be surgically implanted in the patient's amputation stump. This transmission/reception module will ensure the wifi reception of neural

signals acquired from the patient's stump and their use to control the prosthesis motors and the transmission of signals from the prosthesis' pressure sensors to the neural interface from the patient's stump to stimulate the sensory fascicles of the median and ulnar nerves and generate tactile sensations to the patient when he manipulates different objects with the prosthesis. The prosthesis will be equipped with a bidirectional transmission module with a neural interface that will be surgically implanted in the patient's amputation stump. The personalized prosthesis movement algorithms that the patient will upload into the prosthesis control system will be associated with the corresponding neural signal patterns that will be acquired from the patient's stump. After the neural interface implantation surgery, the patient will undergo training to learn to use the neural controlled prosthesis. During this training, the patterns of motor neural signals that will be acquired from the nerves in the patient's stump for each of the movements he wants to perform with the prosthesis will be identified.

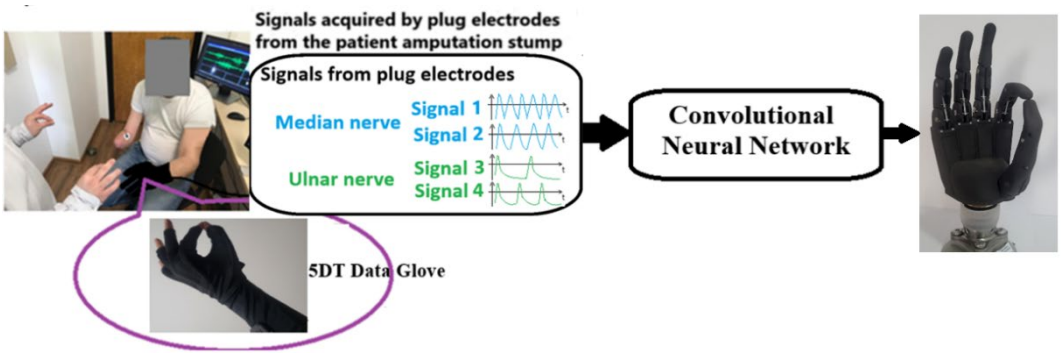


Fig. 22. The patient training for providing labeled data to the convolutional neural network for each movement

In order to do this, an artificial intelligence module based on a convolutional neural network will be implemented in the prosthesis control system that will identify the pattern of neural signals acquired from the patient's stump for each movement he wants to perform with the prosthesis. The patient will perform a specific training for each movement he wants to perform with the prosthesis. The patient will load the respective movement into the prosthesis using the glove and the algorithm described above. Then, for that selected movement, the patient will simultaneously trigger in his brain the same movement command for both hands: for the fingers of the healthy hand and for the amputated hand (figures 22). This command will trigger in the nerves of the patient amputated hand certain patterns of motor neural signals that will be identified by the neural network. The patient will repeat the same movement/brain command hundreds of times, and the motor neural signals from the patient's stump will be sent to the neural network. The 5DT glove will serve as an additional element for labeling the input data to the neural

network during this training. After the neural network has identified the motor neural signal pattern for that movement, that pattern will be associated with the corresponding movement for the prosthesis. The movement was loaded into the prosthesis at the beginning of the training. The patient will then move on to the next movement they want to perform with the prosthesis, and repeat the same training. At the end of the training sessions, the neural network will be able to identify the neural signal patterns for each movement the patient wants to perform with the prosthesis.

4. Conclusions

The hand prosthesis presented in this article was created within the NerveRepack project and it will be connected to the peripheral nervous system of a patient with hand amputation, after a bidirectional neural interface will be surgically installed in the patient stump. The hand prosthesis has a modular structure that allows it to be updated with new functions or interfaces for connection to the patient's nervous system. Compared to other prostheses on the market, the prosthesis from the NerveRepack project is equipped with pressure sensors to provide the patient with tactile feedback sensations when manipulating different objects with the prosthesis.

The prosthesis was equipped with a motor interface for loading movement algorithms and a sensory interface for acquiring signals from the prosthesis' pressure sensors that will be used to generate tactile feedback to the patient who will use the prosthesis. Pressure sensors were mounted on the sole of the motors that actuate the fingers of the prosthesis, to detect different levels of mechanical pressure, from touch to the pressing force corresponding to a weight of 5 kg (the maximum limit that the pressure sensor can detect). Mounting the pressure sensors on the soles of the motors offers the advantage of also detecting levels of mechanical pressure that come from mechanical forces that are not oriented perpendicular to the surface of the fingers of the prosthesis.

Another advantage of the NerveRepack prosthesis is that the patient can upload personalized movement algorithms into the prosthesis control system using the motor interface described in this article.

The mechanical structure of the prosthesis is similar to the mechanical structures of prostheses existing on the market but differs from them by the pressure sensors mounted on the sole of the motors that actuate the fingers of the prosthesis and by the motor and sensory interfaces with which it is equipped. After the neural interface implantation surgery in the patient's stump, which will take place in 2026, the prosthesis presented in this article will be connected to the peripheral nervous system of the hand amputee patient, and the patient will be trained to control it.

Acknowledgement

This study was funded by Project: NerveRepack (intelligent neural system for bidirectional connection with exoprostheses and exoskeletons), European Union's Horizon 2020 research and innovation programme under grant agreement No. 101112347/2023, and supported by the CHIPS Joint Undertaking and its members.

The authors thanks to Octavian Narcis Ionescu (IMT), David Dragomir (IMT) and Catalin Niculae (AREUS Technology) for the technical support in testing the sensorial interface of the hand prosthesis presented in this article.

R E F E R E N C E S

1. *Jibby E. Kurichi, Barbara E. Bates, Margaret G. Stineman,* Amputation, International Encyclopedia of Rehabilitation, 2010
2. *Yuan B, Hu D, Gu S, Xiao S, Song F.* The global burden of traumatic amputation in 204 countries and territories. Front Public Health. 2023; Vol. **11**, Iss:2, 1258853, Published 2023 Oct 20. doi:10.3389/fpubh.2023.1258853
3. *Osiceanu S, Dascalu M, Franti E., Barbilian A*, Intelligent interfaces for locomotory prosthesis, Proceedings of International Joint Conference on Neural Networks, 2009, Vols **1- 6**, pp. 1933-1938
4. *Franti E, Milea L, Dascalu M, Moga M, Florea C, Barbilian A, Teodorescu M, Schiopu P, Lazo F, Pogarasteanu M.E*, Personalized support system for the patients with forearm amputations, Romanian Journal of Information Science and Technology, Vol. **15**, Iss:4, pp.368-376, 2012
5. *Smith S. M., Green A. R., and Miller L. A.*, "Design and validation of implantable neural interfaces for bidirectional prosthetic control," Journal of Neural Engineering, Vol. **18**, no. 2, pp. 1–14, 2021. https://doi.org/10.1088/1741-2552/abd3d4
6. *Lee C. H., Park J. W., and Cho K. J.*, "Biocompatible materials for long-term neural implants: Current perspectives and future directions," Advanced Healthcare Materials, Vol. **9**, no. 17, 2020. https://doi.org/10.1002/adhm.202000487
7. *Garcia M. and Zhou P.*, "Machine learning-based decoding of peripheral nerve signals for upper-limb prosthesis control," IEEE Transactions on Neural Systems and Rehabilitation Engineering, Vol. **30**, pp. 785–795, 2022. https://doi.org/10.1109/TNSRE.2022.3159094.
8. *Tanaka K., Fukuda H., and Nakamura Y.*, "Soft robotics actuation technologies for bio-inspired prosthetic limbs," *Science Robotics*, Vol. **4**, no. 27, 2019. https://doi.org/10.1126/scirobotics.aav4494
9. *Cutkosky, M.R., & Howe, R.D., "Human Grasping and Dexterous Manipulation. In Springer Handbook of Robotics. Springer, 2005.*
10. *Ionescu O.N, Franti E, Carbunaru V, Moldovan C, Dinulescu S, Ion M., Dragomir D.C, Mihailescu C.M, Lascar I, Oproiu AM, Neagu T.P, Costea R, Dascalu M, Teleanu MD, Ionescu G, Teleanu R,* System of Implantable Electrodes for Neural Signal Acquisition and Stimulation for Wirelessly Connected Forearm Prosthesis, Biosensors, Vol.**14**, Iss:31, 2024
11. Resistive pressure sensor, RFP602, 5Kg, compatible FSR 402, 10MM, https://sigmanortec.ro/senzor-presiune-rezistiv-rfp602-5kg-compatibil-fsr402-10mm

12. *L. Székely, S. Olasz, I. Tüzes,* Actuation system equipped with threaded rod and pressure sensors for mobile segments of upper limb prostheses, Patent application nr. A/00170/13.05.2025.
13. *Güçlü H, Cora A*, "A survey on wearable hand robotics design for assistive, rehabilitative, and haptic applications", May 2023, International Journal of Intelligent Robotics and Applications Vol. **7**, Iss:2, pp.1-26,
14. *Barontini F, Obermeier A, Catalano M.G, Fani S, Grioli G, Bianchi M,* "Tactile Feedback in Upper Limb Prosthetics: A Pilot Study on Trans-Radial Amputees Comparing Different Haptic Modalities," in IEEE Transactions on Haptics, Vol. **16**, no. 4, pp. 760-769, Oct.-Dec. 2023.
15. *Dosen S, Prahm C, Amsüss S, Vujaklija I, and Farina D,* "Prosthetic feedback systems," in Bionic Limb Reconstruction. Berlin, Germany: Springer, 2021, pp. 147–167.
16. *Nemah M.N,* "A review of non-invasive haptic feedback stimulation techniques for upper extremity prostheses," Int. J. Integr. Eng., Vol. **11**, no. 1, 2019, Art. no. 3573.
17. *Schofield J.S, Evans K.R, Carey J.P, and Hebert J.S,* "Applications of sensory feedback in motorized upper extremity prosthesis: A review," Expert Rev. Med. Devices, Vol. **11**, no. 5, pp. 499–511, 2014.
18. *Li Y, Yang L, Deng S, et al.* "A machine learning-assisted multifunctional tactile sensor for smart prosthetics", InfoMat, 2023; Vol:5, Issue:9, e12463, doi:10.1002/inf2.12463
19. *Sankar S, et al.,* "A natural biomimetic prosthetic hand with neuromorphic tactile sensing for precise and compliant grasping", Sci. Adv.11, eadr9300, 2025
20. *Abd M.A, Paul R, Aravelli A, Bai O, Lagos L, Lin M, Engeberg E.D,* "Hierarchical Tactile Sensation Integration from Prosthetic Fingertips Enables Multi-Texture Surface Recognition", Sensors. 2021; Vol **21**, Issue 13, 4324. https://doi.org/10.3390/s21134324