



POLITECNICO
MILANO 1863

SCUOLA DI INGEGNERIA INDUSTRIALE
E DELL'INFORMAZIONE

A multi-DoF wrist soft exosuit: tendon driven prototype and elec- trohydraulic actuation feasibility

TESI DI LAUREA MAGISTRALE IN
MECHANICAL ENGINEERING - INGEGNERIA MECCANICA

Author: **Francesca Benatti**

Student ID: 259588

Advisor: Prof.ssa Marta Gandolla

Co-advisor: Dott. Emanuele Bianchi

Academic Year: 2025-26

Abstract

Wrist function is critical for the execution of Activities of Daily Living, enabling the orientation, manipulation, and stabilization of the hand required for interaction with the environment. Despite this clinical importance, the wrist remains understudied in wearable robotics compared to larger upper limb segments. Furthermore, existing soft exosuits are often restricted to single degree of freedom systems, which fail to replicate the complex coordination required for natural movement. This thesis presents the design, development, and validation of a novel tendon driven soft exosuit capable of providing simultaneous assistance for both wrist extension and supination. The mechanical architecture is based on a lightweight textile design actuated through custom tendon transmissions. Two servo motors provide actuation, two IMUs are used for sensing, and an ESP32 based controller manages the system. The control architecture integrates active assistive control, passive trajectory tracking, and teach and play modalities, enabling flexible rehabilitation and assistance strategies. To address the limitations of traditional servo motors, a theoretical feasibility study was conducted on advanced electrohydraulic actuation, specifically exploring HASEL and HALVE technologies, and demonstrating analytical feasibility at wrist-assistance force scales. The system was validated through experimental tests with twelve healthy subjects and a pilot study with two neurologically impaired patients. Results confirmed kinematic transparency in the unpowered mode and accurate trajectory tracking (mean RMSE = 2.01°) in the active assistive mode. Pilot tests demonstrated the feasibility of supporting functional wrist movements in patients, while the passive mode provided smooth and safe rehabilitation trajectories. Overall, this prototype offers a clinically pilot-tested solution for upper limb mobility restrictions, while the proposed electrohydraulic actuation holds significant potential, promising a transition toward fully compliant, silent, and biomimetic wearable robotics.

Keywords: Soft Exosuit, Wrist Rehabilitation, Tendon Driven Actuation, Electrohydraulic Actuation, Wearable Robotics, Multi Degree of Freedom Assistance

Abstract in lingua italiana

La funzionalità del polso è fondamentale per l'esecuzione delle Attività della Vita Quotidiana, poiché consente l'orientamento, la manipolazione e la stabilizzazione della mano nell'interazione con l'ambiente. Nonostante questa importanza clinica, il polso rimane poco studiato nella robotica indossabile rispetto ai segmenti dell'arto superiore di maggiori dimensioni. Inoltre, i soft exosuit esistenti sono spesso limitati a sistemi a singolo grado di libertà, che non riescono a replicare la complessa coordinazione richiesta dal movimento naturale. Questa tesi presenta la progettazione, lo sviluppo e la validazione di un innovativo soft exosuit azionato a cavi per l'assistenza simultanea di estensione e supinazione del polso. L'architettura meccanica si basa su un design tessile leggero con trasmissioni a cavi personalizzate. Due servomotori forniscono l'attuazione, due IMU il sensing e un controllore ESP32 gestisce il sistema. L'architettura di controllo integra assistenza attiva, traiettoria passiva e modalità teach and play, consentendo strategie flessibili di riabilitazione e assistenza. Per ovviare ai limiti dei servo motori tradizionali, è stato condotto uno studio teorico di fattibilità sull'attuazione elettroidraulica, esplorando le tecnologie HASEL e HALVE e dimostrando la fattibilità analitica alle scale di forza richieste. Il sistema è stato validato con dodici soggetti sani e uno studio pilota con due pazienti affetti da disturbi neurologici. I risultati hanno confermato la trasparenza cinematica in modalità non attuata e l'accurato tracciamento delle traiettorie (RMSE medio = 2.01°) in modalità assistiva. I test pilota hanno dimostrato la fattibilità del sostegno ai movimenti funzionali del polso, mentre la modalità passiva ha fornito traiettorie di riabilitazione fluide e sicure. Nel complesso, il prototipo offre una soluzione clinicamente testata in uno studio pilota per le limitazioni di mobilità dell'arto superiore, mentre l'attuazione elettroidraulica proposta prospetta una transizione verso sistemi indossabili pienamente conformi, silenziosi e biomimetici.

Parole chiave: Soft Exosuit, Riabilitazione del polso, Attuazione a cavi, Attuazione elettroidraulica, Robotica indossabile, Assistenza multi grado di libertà

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Introduction

The wrist is one of the most anatomically complex joints in the body due to its very small and compact size and its great mobility [52]. It is able to move in a 3D space with very complex kinematics organized in three degrees of freedom (DoF): flexion/extension, which is the bending of the hand towards the palmar or dorsal side of the forearm; radial/ulnar deviation, that is the sideways movement of the wrist toward the radius (thumb) or ulnar (little finger); pronation/supination, that is the rotation of the whole forearm around its longitudinal direction. Being a combination of movements acting within the forearm and the elbow, it is sometimes not considered among the wrist degrees of freedom [52]; however, it is fundamental in the execution of many ADLs, such as eating with a fork, opening a bottle, pouring water inside a glass, or turning a doorknob [21], and is therefore a strong focus of this thesis.

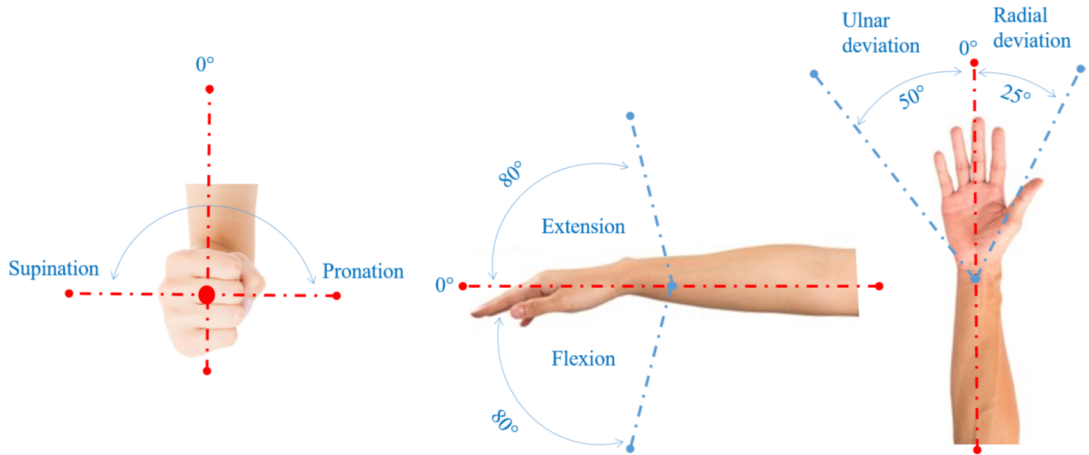


Figure 1: Degrees of freedom of the wrist with corresponding maximum range of motion (RoM) from [52].

In general, the wrist is a crucial joint in ADLs as it is critical to orient the hand with respect to the forearm and to transmit loads between the forearm and the hand. Moreover, wrist stability is directly related to the capacity of the fingers to flex and extend, meaning that wrist impairments can also reduce the ability to grasp and manipulate objects [38, 63].

Wrist impairments can result from neurological disorders such as stroke, cerebral palsy, and spinal cord injuries (SCI), or from musculoskeletal damage like carpal tunnel syndrome, tendon disorders, and sprains. In addition to limiting patients' independence, restricted wrist mobility often forces individuals to use unnatural compensatory movements with their shoulder and elbow, which can lead to further musculoskeletal strain [30].

Robotic assistance has been widely researched in past decades to aid patients with motor disabilities [42] with two main approaches: assistance and rehabilitation. Firstly, providing active robotic assistance during ADLs empowers patients to immediately compensate for their disabilities in real world scenarios and in occupational settings, it mitigates fatigue and prevents the onset of work related musculoskeletal pain. It must be noticed, however, that the effect of the device's action is not therapeutic but orthotic: this means that the beneficial effect to the user stops once the device has been removed. Secondly, rehabilitation is essential for regaining movement. To be effective, it must consist of repetitive, controlled exercises that retrain both muscles and motor pathways. Robotic support allows these movements to be performed with extreme precision, perfect repeatability and without the constant supervision of a therapist [2, 54].

Moreover, studies suggest that the recovery of distal functions is closely linked to the state of proximal functions, meaning that training the wrist can lead to positive therapeutic effects that extend up to the shoulder and elbow [38]. Therefore, successful upper arm therapy requires a strong focus on rehabilitating the wrist and hand, not just the broader arm [18].

Despite its clinical importance and its status as the fourth most common site of upper-limb musculoskeletal pain [53], few devices focus specifically on this joint. One possible explanation is the difficulty of designing a device that anchors securely to the forearm and actuates the wrist without impeding the natural grasping capabilities of the hand [21]. Furthermore, the wrist is positioned between the hand and the arm and generally an impaired wrist comes with an impaired hand or arm or both: therefore, in order to be used by patients, a device for the wrist might need to be accompanied by some kind of appliance for either the hand (for instance an orthosis to hold cutlery in order to eat) or the arm, increasing the difficulty of the study.

Several factors must be considered when designing an assistive device and one of them is portability. In fact, for rehabilitation, portability enables home based training [28]. For assistance, it allows users to perform functional everyday tasks independently, including outside clinical settings [38].

Wearability cannot be separated from concepts like comfort, low weight and low encumbrance, safety in the interaction with human joints and other properties that will be discussed more in detail in the state of the art. As a consequence of the popularity gained by soft robotics in the last few years, a recent rising trend in wearable robotics to meet these requirement has been that of soft exosuits instead of rigid exoskeletons [43]:

“The term ‘soft exosuits’ is used as a collective term for wearable robotic devices made from compliant materials. They wrap around the user’s body and rely on its structural integrity instead of an additional rigid frame to transfer reaction forces between body segments.” [60]

Given the critical role of the wrist in daily function and upper limb rehabilitation, the limited number of dedicated wearable devices, and the growing importance of portability and user comfort, this thesis proposes the design, development, and validation of a soft robotic exosuit for wrist assistance and rehabilitation. The device specifically targets flexion/extension and pronation/supination (which remains underexplored in the current literature compared to the other two degrees of freedom). The work includes mechanical design, control strategy development, prototype fabrication, and preliminary human subject testing.

1 | State of the Art

This chapter will initially provide a general overview of the categories of wearable robotic devices and will then present a complete classification of the soft actuators used to actuate soft exosuits, reporting for each category examples of devices for upper limb, mostly wrist. This is to analyze and frame the existing technologies.

1.1. Types of devices

The field of interest of this project is that of collaborative robotics, a subfield of robotics focused on the collaboration between robot and human. Collaborative robots, also called cobots, can be characterized by different degrees of interaction between the human and the robot. As already mentioned, this work focuses on a wearable and portable device; therefore, this chapter will not include examples of end effector robots, which are considered wearable in the literature but are definitely not portable. End effector systems are not worn by the user but operate externally, with only the end effector physically connected to the human body. They also represent one of the earliest forms of collaborative robotic systems. [56].

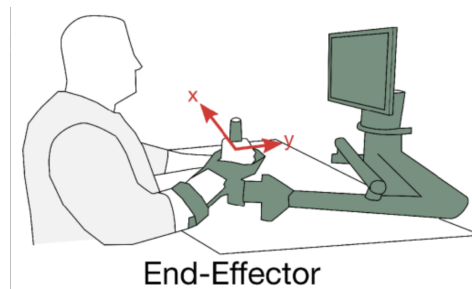


Figure 1.1: End-effector robotic device from [56].

Having clarified what is going to be excluded from this overview, the focus can now shift to the systems included in this section.

Wearable cobots can be classified according to characteristics other than the degree of interaction: an extremely relevant classification criterion for introducing the already men-

tioned soft exosuits is stiffness. Under this perspective, three main categories of devices can be identified, ordered from the stiffest to the softest: rigid exoskeletons, compliant devices, and soft exosuits. [53].

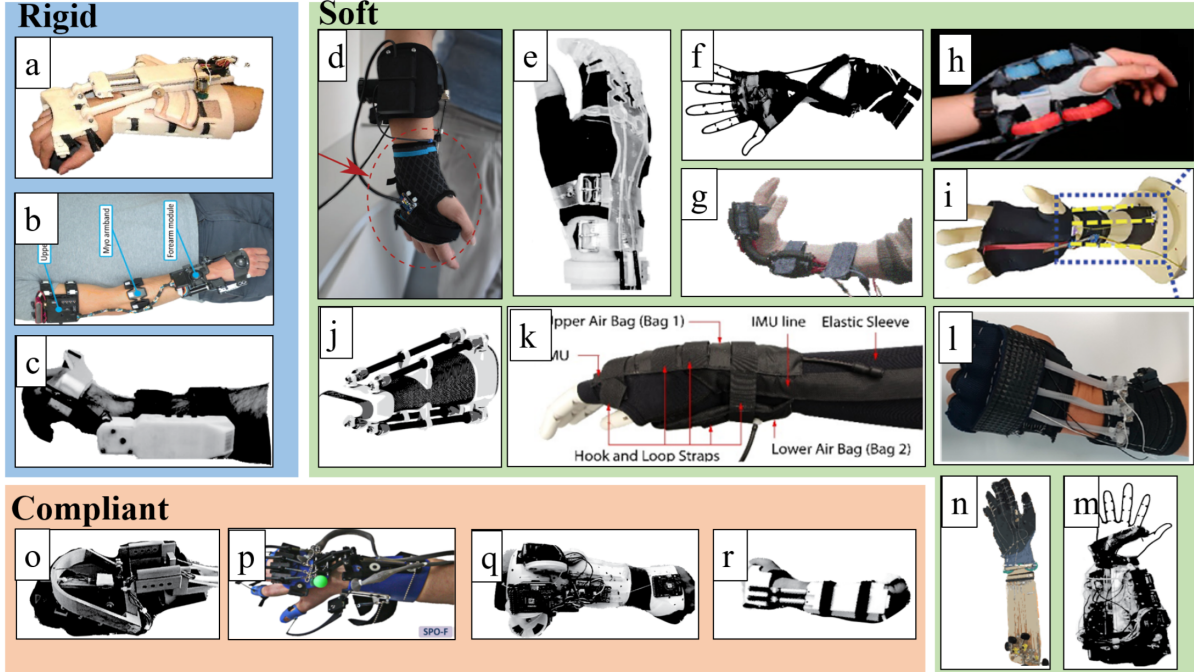


Figure 1.2: Classification of wearable robotic devices according to stiffness from [53].

Rigid exoskeletons Rigid exoskeletons are devices made of hard and stiff materials whose axes, unlike end-effector robots, are aligned with the anatomical axes of the subject.

They are generally, but not always, actuated by servo motors capable of directly controlling individual joints and transmitting high loads while, thanks to the rigidity of their materials, they can also bear significant loads, relieving the user from them [11, 41]. For this reason, although it is not subject of this project, it is worth mentioning that rigid exoskeletons are also widely used in industrial environments, helping operators in handling heavy loads safely and with reduced physical effort. Furthermore, rigid devices are generally much easier to model and control than soft ones, providing high transmission efficiency and highly precise and reliable trajectory tracking [52].

However, rigid exoskeletons also have some limitations. First of all, they require extreme precision in the alignment between robotic and human joints, otherwise they could put the wearer at risk [38]. Additionally, they tend to be quite bulky and heavy, limiting portability, restricting the wearer's agility and comfort and increasing the metabolic cost required to move [2]. Lastly, they can be expensive due to their mechanical complexity.

Compliant devices Compliant devices pose themselves in the middle between totally rigid and totally soft devices, being them constituted by elements like springs or flexible 3D printed structures.

Unlike rigid exoskeletons that require precise alignment with the human joint's center of rotation, compliant mechanisms naturally deform and adapt to the user's anatomy during motion; they are also generally lighter and less cumbersome than rigid devices [27]. Moreover, compared to completely soft systems, the presence of semi rigid components introduces a "stiff as needed" approach, helping in bearing loads [12].

On the other hand, the deformability of the material can reduce control precision and introduce unpredictable deformations or even unwanted shear forces, especially when the actuator is not manually backdrivable [53].

In short, finding the optimal balance of stiffness is challenging. The compliant system must be stiff enough to handle functional loads and transmit forces effectively but still soft enough to prevent localized pressure spots and maintain high comfort levels. Therefore, despite theoretically combining the advantages of both rigid and soft devices, compliant systems can be difficult to manage in practice.

Soft exosuits Soft exosuits are wearable robotic devices made from highly compliant, flexible materials such as fabrics, webbing, rubber, elastomers, thermoplastics, and artificial tendons. Unlike traditional rigid exoskeletons, they do not use an external rigid kinematic frame: instead, they rely on the wearer's own biological skeletal structure to support compressive forces and transfer reaction forces between body segments [35, 60]. Soft exosuits represent the forefront of research in collaborative wearable robotics. In fact, while traditional rigid exoskeletons have been studied for industrial and medical applications since the 1960s and 1970s, soft robotics has only emerged in recent years as a novel solution to the bulkiness, discomfort, and joint misalignment problems of rigid devices. Suffice it to say that the first soft wearable device capable of supporting wrist pronosupination dates back only to 2018 [21].

The advantages compared to rigid exoskeletons are several and significant.

First, soft exosuits are undoubtedly lighter thanks to the reduced weight of soft materials [35]. As it will become clearer later on, the weight largely depends also on the actuator used in the system but the choices of lightweight actuators, especially when considering collocated ¹ weight, are numerous [52].

Then, these systems are inherently compliant and offer much better ergonomics and com-

¹The term collocated is used to indicate the parts of the system that are directly mounted on the body segment or joint of interest, whereas non collocated parts are located remotely, for instance a motor mounted as a backpack or a pump placed on a separate support device such as a deambulatory device.

fort, granting an overall better wearability [43]. This aspect is particularly important in assistive devices intended to be worn for extended periods of time, potentially even underneath everyday clothing.

Next, one of the main advantages of soft exosuits is that, due to the absence of rigidly imposed joints and the resulting virtually infinite number of degrees of freedom, they naturally accommodate the wearer's joint movements [74]. This way, they are transparent and much safer than rigid exoskeletons [35].

After that, it is worth mentioning that the use of flexible materials such as fabric (for example in the form of gloves and sleeves in wrist devices) makes it much easier to adapt the system to different users with different anatomies and sizes [43].

Finally, soft exosuits are generally characterized by a lower cost due to the inexpensive nature of materials such as fabric and elastomers [43].

On the other hand, soft structures exhibit more complex and often nonlinear kinematics compared to rigid exoskeletons, making them more difficult to model and control precisely [74].

Moreover, purely soft, textile based devices often suffer from low force transfer efficiency because the fabrics stretch, slip, or deform under high tension, which dissipates the intended assistive torque at the human-device interface [12].

Furthermore, soft actuators are generally highly susceptible to failure under large payloads. Pneumatic artificial muscles, for example, are prone to punctures and can suffer from a "ballooning effect" leading to rupture if internal pressures become too high [52].

Lastly, even though the collocated weight is generally low, depending on the actuator type the non collocated weight and encumbrance might be non negligible [38].

However, with a proper choice of actuator, a good mechanical design and a clever control strategy, the disadvantages just presented can be mitigated: this is why soft exosuits are the most promising technology among the ones just listed and, therefore, the perfect choice for an innovative assistive and rehabilitative wearable device for the wrist.

1.2. Actuation technologies

In order to better understand the existing soft exosuits and their functioning, it is necessary to introduce their key element: the actuator. Of course, for a device to be considered truly soft, not only the moving elements must be soft but also the actuating technology and this opens the discussion to the very broad category of soft actuators.

Several review articles in the literature propose classifications of the different types of soft actuators: among the most comprehensive are [45, 46, 74]. After consulting many

of them, the classification shown in Figure 1.3 was developed with the aim of being as exhaustive as possible.

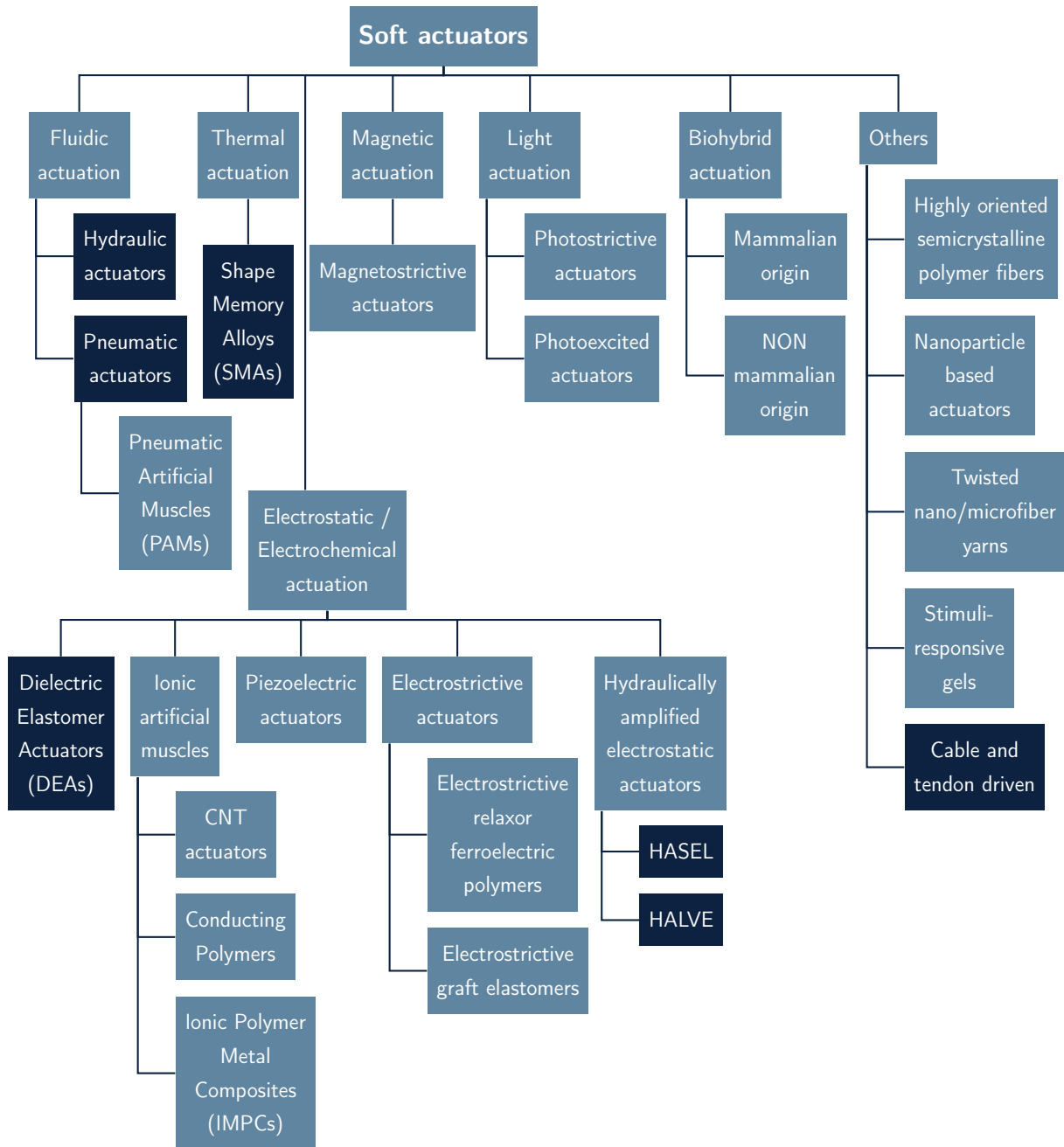


Figure 1.3: Classification of soft actuators. In dark blue, the categories discussed in detail in this chapter.

It's important to notice that this overview aims to cover as many categories as possible, but not all of them are suitable for actuating soft exosuit. In fact, many actuator types belonging to the “Electrostatic / Electrochemical actuation” category, as well as

all magnetic, light-driven, biohybrid, and “other” actuators, are currently too low on the Technology Readiness Level scale [48], shown in Figure 1.4, to actuate systems as large and heavy as wearable exosuits, since most of them are only capable of producing very small displacements and under negligible external loads.

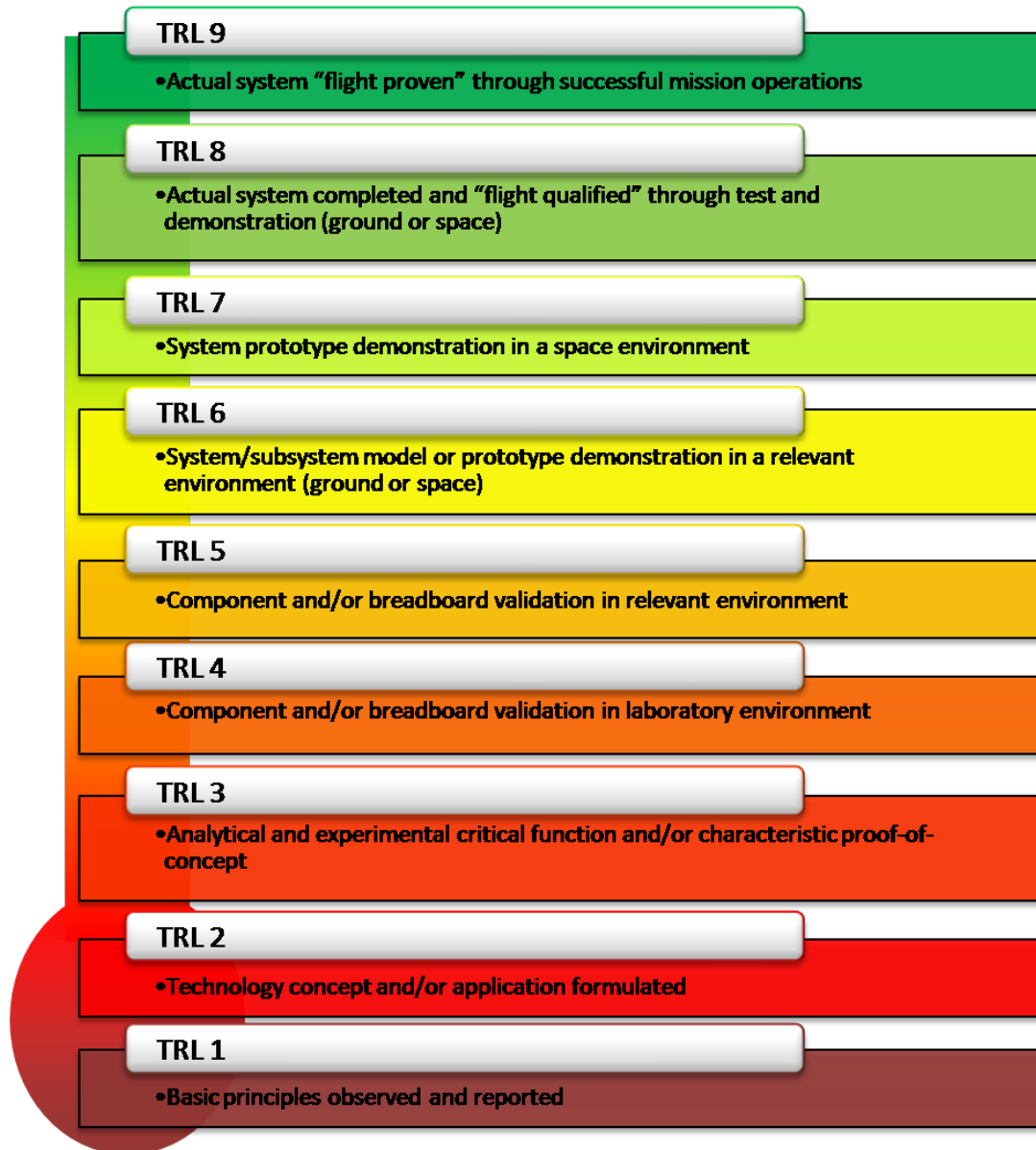


Figure 1.4: Technology Readiness Level scale from NASA [48]

Among these actuators, those suitable for actuating soft exosuits or similar biomedical and bioinspired systems are often referred to as “artificial muscles”. The most commonly used are soft fluidic actuators, mainly soft pneumatic, shape memory alloys (SMAs) and dielectric elastomer actuators (DEAs). Another category has recently started to emerge

and it is that of hydraulically amplified electrostatic actuators, primarily in the form of HASEL or HALVE actuators. Finally, one of the most diffused technologies in this field is cable or tendon driven, meaning that the actuator is an electric or servo motor and the transmission happens through some cables or tendons, frequently Bowden cables: this category is generally included when discussing soft actuation systems [12]; however, due to the rigidity of the servo motor, it lies at the boundary of soft robotics and cannot strictly be considered an artificial muscle, since it does not mimic the behavior of mammalian muscles [46]; it does mimic the action of tendons though [40].

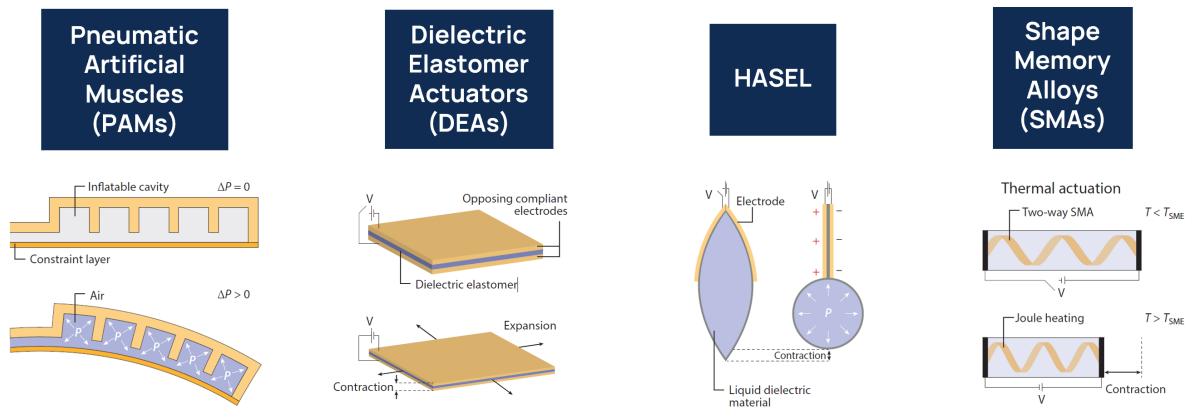


Figure 1.5: Main types of artificial muscles, pictures from [74].

Let us discuss each of these main categories and their existing applications in wrist soft exosuits.

1.2.1. Soft fluidic actuators

Soft fluidic actuators achieve motion through the control in space and in time of fluidic pressure inside internal inflatable cavities, such as channels and chambers. They are primarily composed of elastomers or other flexible materials generally wrapped by inextensible reinforcement layers (such as fibers, fabrics or stiffer materials) to constrain the direction of deformation: in this way, depending on the geometry of these strain limiting layers, these actuators can generate a wide range of motions, including bending, twisting, and elongation. They are pressurized by rotary or reciprocating pumps, in the case of hydraulic actuators, or compressors, in the case of pneumatic ones. A key feature is their ability to maintain an actuated state without drawing continuous power simply by closing their valves [74].

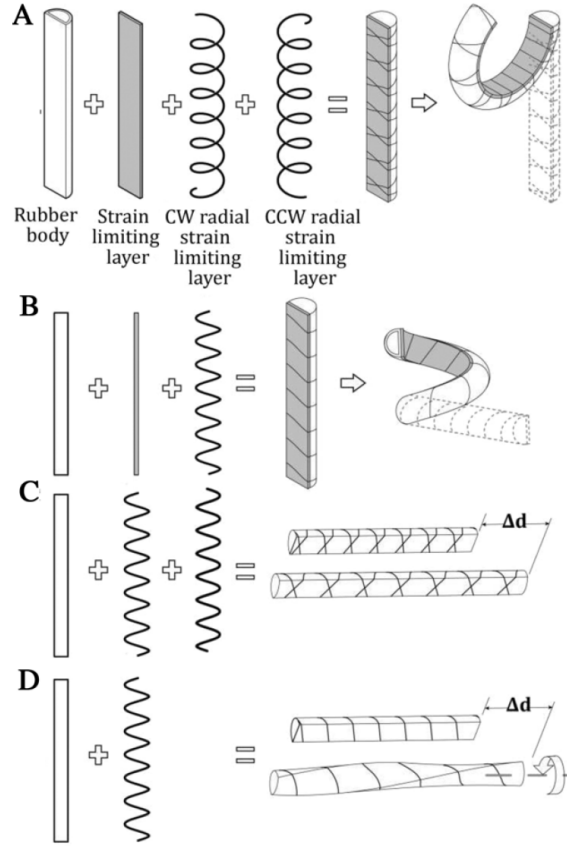


Figure 1.6: Examples of different uses of strain limiting layers in soft fluidic actuators from [54].

However, soft fluidic actuators suffer from several disadvantages. They are generally not energy efficient, are prone to puncture, and are difficult to accurately control due to the nonlinear response of soft materials.

Since they rely on pumping fluid through a network of tubes and channels, they suffer from fluid transport losses that limit their actuation speed and efficiency.

But the most critical drawback for wearable devices is that they require cumbersome peripheral components such as large reservoirs, valves, and compressors which severely increase non collocated weight and bulkiness, reducing portability [34].

Soft hydraulic actuators

Hydraulic actuators are fluidic actuators that use an incompressible fluid as the working medium. In industrial applications it is generally oil, not water because oil provides essential internal lubrication, prevents metal corrosion, and offers a much broader operating temperature range without freezing or boiling. However, since liquid can leak from

actuators, oil is not suitable for applications in contact with the human body, therefore water is generally used.

The use of an incompressible fluid ensures higher force density compared to compressible fluids. However, the weight of the liquid is also higher and this is conflicting with the necessity of low weight, especially collocated, required by portable devices.

There are no wrist soft exosuits actuated by soft hydraulic actuators, probably because the disadvantages outweigh the advantages in this kind of application, especially with such a wide range of possible actuators to choose. However, Polygerinos et al. [54] developed a soft hydraulic actuators driven soft exosuit for the fingers, shown in Figure 1.7.

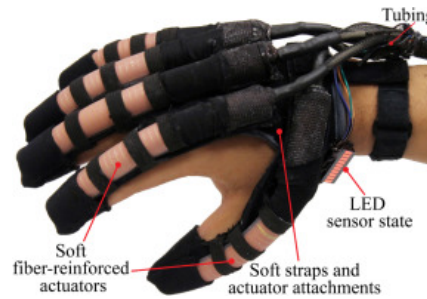


Figure 1.7: Soft exosuit for fingers flexion/extension with soft hydraulic actuators from [54].

The only other existing upper limb soft exosuits with soft hydraulic actuators are from Kosaki et al. [36], shown in Figure 1.8, and from Sy et al. [69], shown in Figure 1.9, both for the elbow. However, the former is only a theoretical study, no real prototype has been realized; the latter, instead, is not purely a soft hydraulic actuator because the hydraulic expansive action is combined with the passive action of a spring, responsible for the contraction.

Other disadvantages include the need for a return line, that increases once again volume and weight, the risk of water leakage and risk of significant pressure oscillations and overshoots during operation caused by the delayed valve response times due to water inertia.

Overall, these actuators are probably the least suitable, among the ones discussed in this section, for the wrist soft exosuit of this thesis [24].

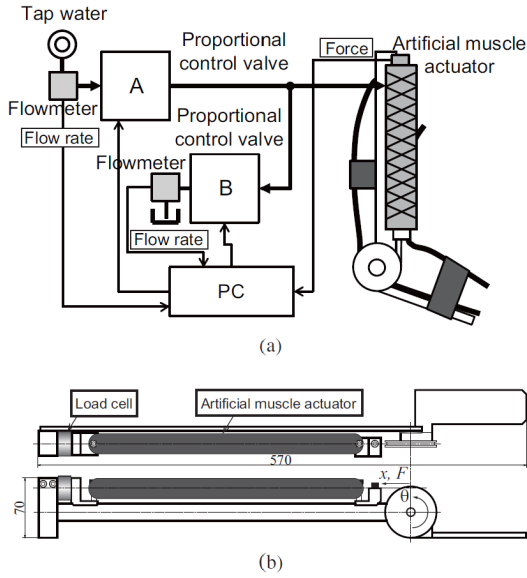


Figure 1.8: Theoretical soft exosuit for elbow flexion/extension with soft hydraulic actuator from [36].

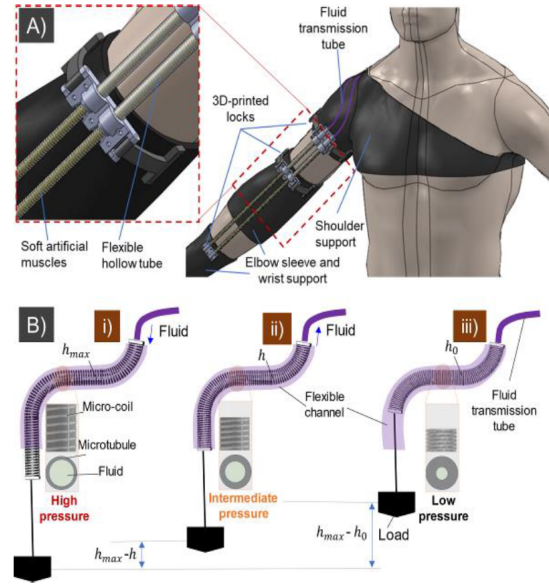


Figure 1.9: Soft exosuit for elbow flexion/extension with soft hydraulic actuators combined with passive springs from [69].

Soft pneumatic actuators

Soft pneumatic actuators, on the other hand, operate using a compressible fluid: compressed air. They are made of channels and chambers that are inflated and can either expand or contract in the active direction, generating mechanical work; the most renowned example of contracting soft pneumatic actuators is that of McKibben actuators, sometimes referred to as Pneumatic Artificial Muscles (PAMs). These actuators consist of a highly compliant inner rubber bladder enclosed within an inextensible braided tubular sheath. When pressurized, the air inflates the inner bladder and tries to expand outward radially. Because the outer braided fibers cannot stretch, the radial expansion forces the geometric angle of the braided mesh to change, which pulls the ends of the actuator together, resulting in a linear contraction [46].

Examples of soft pneumatic actuators that expand are, instead, pleated, bellows, and origami actuators which, instead of relying on the elastic stretching of a bladder, exploit predefined folds, such as basic origami structures [46].

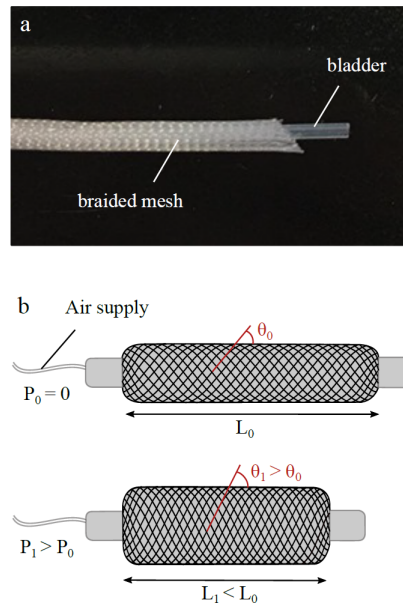


Figure 1.10: Layout and actuation principle of a traditional McKibben actuator from [39].

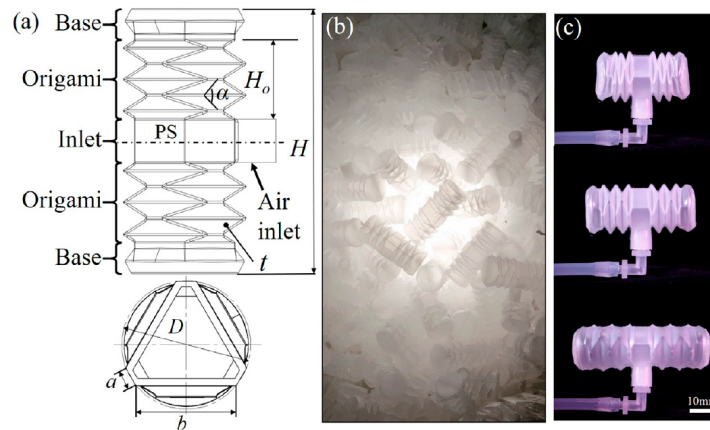


Figure 1.11: Design of a soft origami actuator from [43].

Among their numerous advantages, soft pneumatic actuators are particularly valued for their high power to weight ratio, inherent compliance, and safe human-robot interaction. In addition, the use of air as the working medium, rather than liquids, helps maintain a clean operating environment and allows the actuator to withstand heavy loads without sustaining damage, especially when pressure relief valves or similar safety mechanisms are employed. Furthermore, soft pneumatic actuators are extremely lightweight, resulting in low collocated mass.

There are numerous wrist soft exosuits actuated by soft pneumatic actuators, which represent one of the most widespread categories of soft actuators for this type of device. In

this section, only a few representative examples will be presented, while a more comprehensive overview of existing devices will be provided in the Table A.1 and discussed in the Section 1.3.

Starting from contractile soft pneumatic actuators, Bartlett et al. [5] present a very interesting device, shown in Figure 1.12, driven by two couples of McKibben actuators arranged in two crosses on the top and bottom of the forearm, capable of obtaining all three degrees of freedom of the wrist depending on which pair of actuators is activated. Not only does this device present a remarkably smart and unique design strategy, but it is also one of the few existing soft exosuits that manage to achieve wrist pronosupination.

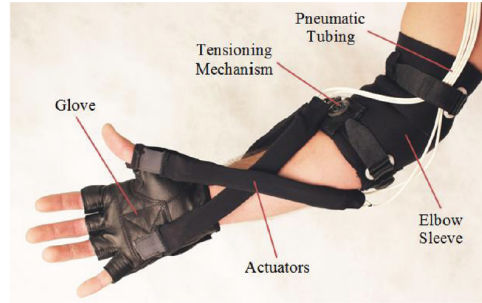


Figure 1.12: Soft exosuit for wrist flexion/extension, radial/ulnar deviation and pronation/supination with McKibben actuators from [5].

Another pneumatic wrist pronation/supination soft exosuit reported in the literature is that from Ferroni et al. [21], shown in Figure 1.13: it is still contractile but with a custom geometry and it uses a different strategy from the previous one, that consists of a pair of counter rotating actuators wrapped around the forearm to achieve pronation and supination based on the activated actuator.

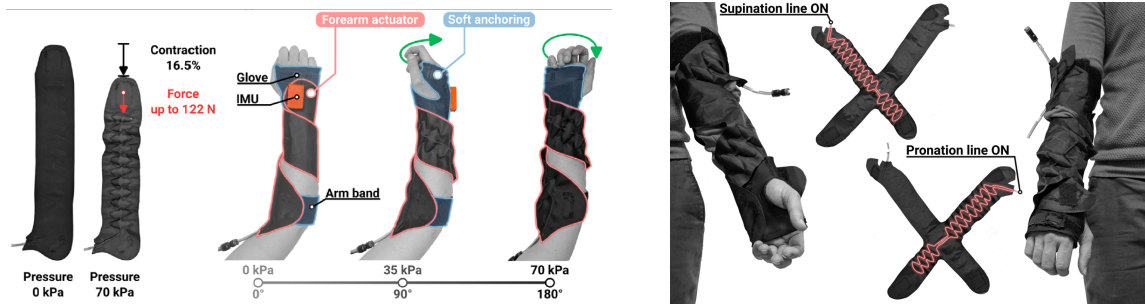


Figure 1.13: Soft exosuit for wrist pronosupination with one or two ²contractile pneumatic fabric-based actuators that twist around the forearm from [21].

²Two prototypes were created: a reconfigurable one with a single twisting actuator that could be

Moving on to expansive actuators, Liu et al. [43] developed a clever solution to obtain both flexion/extension and radial/ulnar deviation using origami actuators, shown in Figure 1.14.

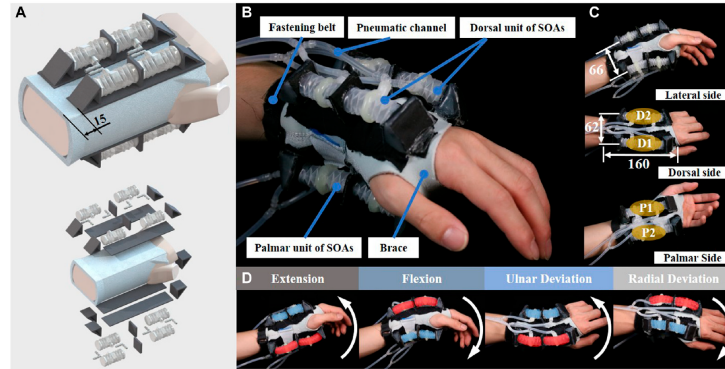


Figure 1.14: Soft exosuit for wrist flexion/extension and radial/ulnar deviation with origami actuators from [43].

To conclude this synthetic overview, it is worth noting that strain limiting layers, previously introduced for hydraulic actuators, can also be employed in pneumatic actuators. They can result in actuators that, instead of expanding or compressing, mainly twist, as in the case of the device by Realmuto et al. [57], shown in Figure 1.15, or they can be combined with creative geometries, as demonstrated by Feng et al. [20] in their device for finger assistance shown in Figure 1.16.

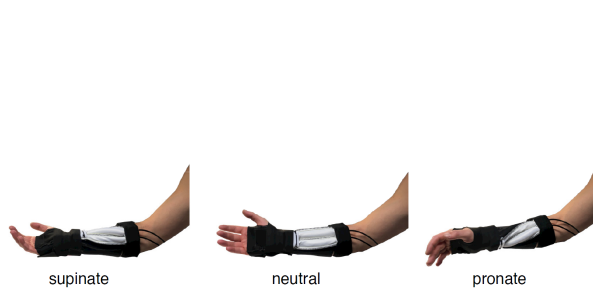


Figure 1.15: Soft exosuit for wrist pronation/supination with soft pneumatic actuators with strain limiting elements from [57].



Figure 1.16: Soft exosuit for fingers flexion/extension with pneumatic actuators equipped with strain limiting layers from [20].

positioned for pronation or supination according to the need, which was also validated on spinal cord injury (SCI) patients, and one with two counter-twisting actuators enabling both pronation and supination at the same time, which was not tested on SCI patients.

Apart from the already mentioned ballooning effect, pneumatic actuators suffer from other defects.

On the one hand, air is much lighter than water, which explains why soft pneumatic actuators are much more used than hydraulic ones in portable and wearable devices but, on the other hand, the compressibility of air compared to water causes pneumatic actuators to suffer from low accuracy and struggle with precise velocity control [24].

Then, flat fabric pneumatic actuators can suffer from "air binding" at low pressures, where the fabric layers stick together and prevent air from entering, leading to highly nonlinear force outputs at the start of actuation [20].

Furthermore, pneumatic compressors also generate loud, continuous noise during operation [51].

But, as mentioned in the introduction about fluidic actuators, the most relevant and problematic drawback of soft pneumatic actuators is the reliance on a compressed air supply: the required compressors, storage tanks, regulators, and valves make the overall system bulky, heavy, and difficult to make fully portable [52]. This problem constitutes one of the most important points that this thesis aims at overcoming.

1.2.2. Shape Memory Alloys (SMAs)

Shape memory alloys, most commonly Nickel-Titanium (NiTi or Nitinol), are materials that exhibit unique properties due to reversible phase transformations between martensite and austenite, induced either by mechanical stress or temperature variations. They are mainly characterized by two phenomena: pseudoelasticity and the shape memory effect. Pseudoelasticity is a stress-induced transformation that occurs at high temperatures, when the material is in the austenitic phase, and results in a superelastic behavior, allowing the alloy to withstand very large elastic deformations. At lower temperatures, when the material is in the martensitic phase, martensite variants are initially arranged in a twinned configuration and can reorient under applied stress into a detwinned configuration, producing a rubber-like mechanical response. The shape memory effect, instead, is a temperature-induced phenomenon that enables the alloy to recover its original shape after undergoing large deformations when heated above a specific transformation temperature, causing the transition from martensite back to austenite [46]. This second effect is exploited by SMA actuators to generate mechanical work.

In robotics and wearable devices, the necessary heat is usually generated via the Joule effect by passing an electrical current directly through the SMA material [63]. When the current is removed, the material cools via convection or conduction and can be stretched back to its deformed state. An alternative to natural convection to reduce the cooling

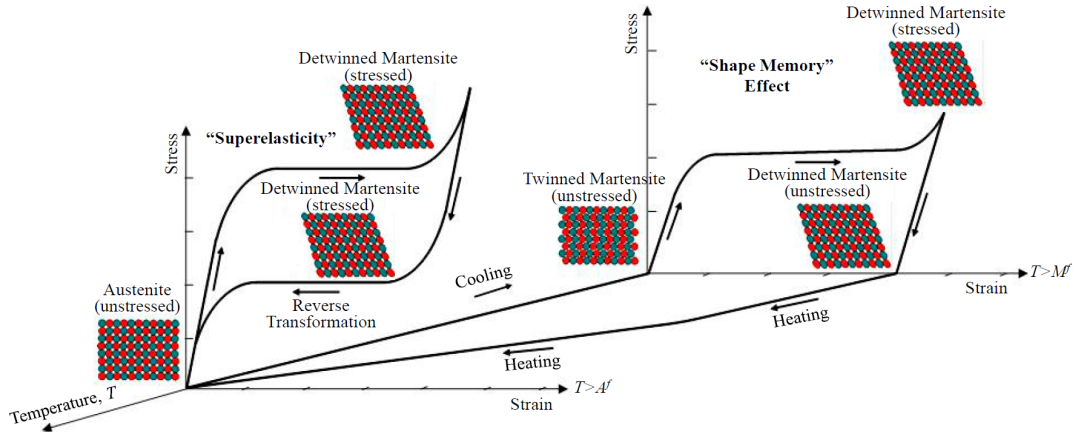


Figure 1.17: Behaviour of SMAs from [62].

time is to use a compressed air flow directed by some plastic tubing [28].

While a straight SMA wire can only contract by about 3% to 8% of its length, shaping the SMA into a coil spring or zig-zag configuration can amplify the contraction strain to over 200%, giving it capabilities comparable to human biological muscles and even much more performing [30]. In fact, the work density of human muscles is around 0.07 J/cm^3 whereas that of Nitinol shape memory alloys is quoted as between 1 and 100 J/cm^3 depending on its deformation mode (for reference, small DC motors are around $4 \div 10 \text{ J/cm}^3$, and PAM around 2 J/cm^3) [28].

This introduces two of the main advantages of the shape memory alloys: they possess an exceptional work density, unmatched by most other materials, and they can be manufactured under the form of wires that can be configured or assembled together in a huge variety of geometries in order to obtain even better performances than one wire on its own.

Moreover, the wire shape is extremely lightweight and small, constituting an overall light and compact system compared to the previous categories of actuators discussed. Still on this aspect, SMA actuators completely eliminates the need for cumbersome external compressors, fluid reservoirs, or heavy DC motors, making them ideal for highly portable and low profile wearable devices [63].

Lastly, since they rely entirely on thermal material deformation rather than gears or pumps, SMA actuators operate completely noiselessly, which greatly improves user acceptance [28].

Since the most peculiar advantage of SMA that has been described is the possibility of creating a variety of geometries to improve the system performance, it's appropriate to

start from the exosuit designed by Park et al. [51] shown in Figure 1.18 even though it's for elbow and not for wrist.



Figure 1.18: Soft exosuit for elbow flexion/extension actuated by fabric SMA muscle from [51].

In fact, the highly innovative strategy of this device is to combine the actuator part and the soft part of the exo in one component by weaving SMA fibers in order to create a soft actuating fabric mimicking the action of mammalian muscles. Since, as mentioned, the potential of SMA is not completely exploited by wires but instead it's achieved with much better performance by coils, the authors of the paper wrapped SMA wires around some core wires that they then dissolved in an acid solution, leaving behind a fabric of SMA coils, as shown in Figure 1.19.

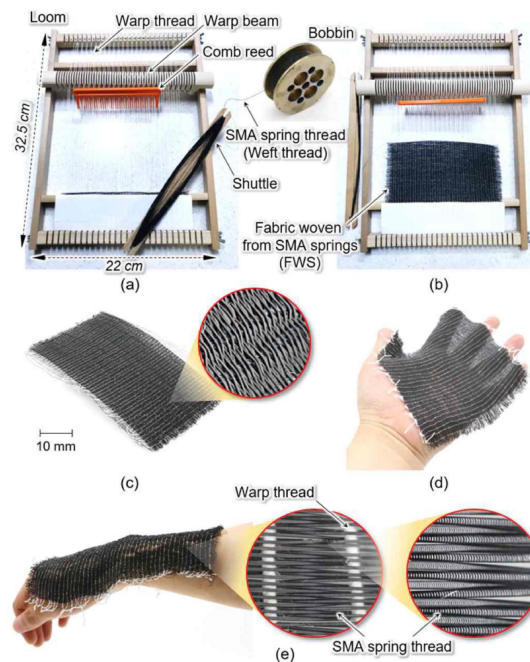


Figure 1.19: Fabric SMA muscle preparation and weaving from [51].

Passing on to a more classic design, Jeong et al. [30] present the design of a device, shown in Figure 1.20, for wrist flexion/extension and radial/ulnar deviation made of coils that contract upon heating and therefore pull the hand in the corresponding direction.

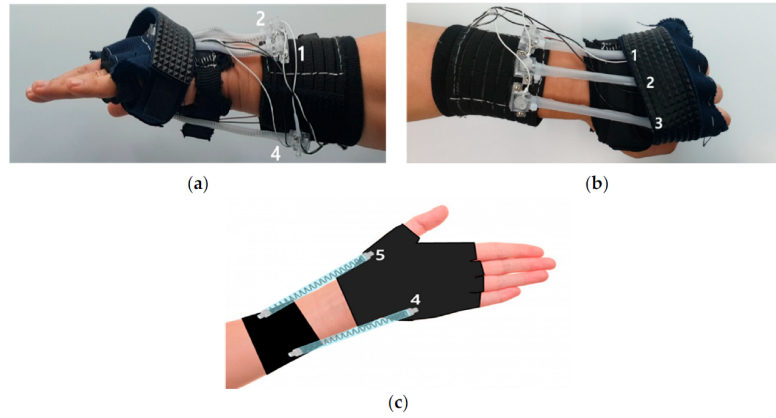


Figure 1.20: Soft exosuit for wrist flexion/extension and radial/ulnar deviation with SMA coils from [30].

It is interesting to observe that, unlike pronation/supination, flexion/extension is a conceptually simple movement to reproduce. In most soft exosuit designs, it is achieved by placing actuators or transmission elements, such as cables, along the dorsal and palmar sides of the hand, where they generate flexion or extension through pulling forces (or pushing forces in the case of expansive pneumatic actuators). This design principle is consistently adopted across the literature and is particularly relevant to the present work, as flexion/extension is one of the two wrist degrees of freedom addressed in this thesis.

This approach is used also for fingers flexion/extension as presented by Hadi et al. [26] and shown in Figure 1.21.

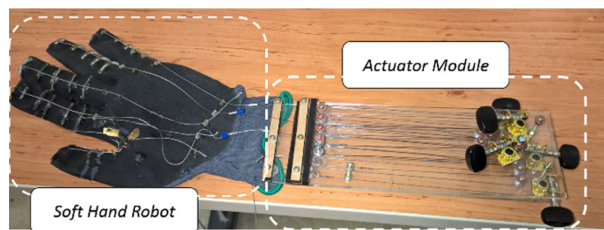


Figure 1.21: Soft exosuit for fingers flexion/extension with SMA wires from [26].

This device is notable because it addresses the need for increased strain by utilizing pulley systems rather than coils. These pulleys trade force, which is inherently high in SMA wires

and more than sufficient for lifting lightweight fingers, for increased strain. This design principle will also be a central theme in this project.

Pronation/supination, on the other hand, is a more complex movement to achieve than flexion/extension. As a result, the relatively few soft exosuits designed for this degree of freedom adopt different geometries and actuation strategies. This topic will be analyzed more in detail in the Materials and Methods chapter, in the section dedicated to the mechanical design of the pronation/supination movement. Nevertheless, among the soft exosuits actuated by Shape Memory Alloys (SMAs), the only example specifically designed for forearm pronation/supination is the device proposed by Jeong et al. [31]. Interestingly, it is based on the same principle adopted by the pneumatic wrist exosuit developed by Ferroni et al. [21], where the actuators are arranged around the forearm in a twisting configuration to generate rotational motion.

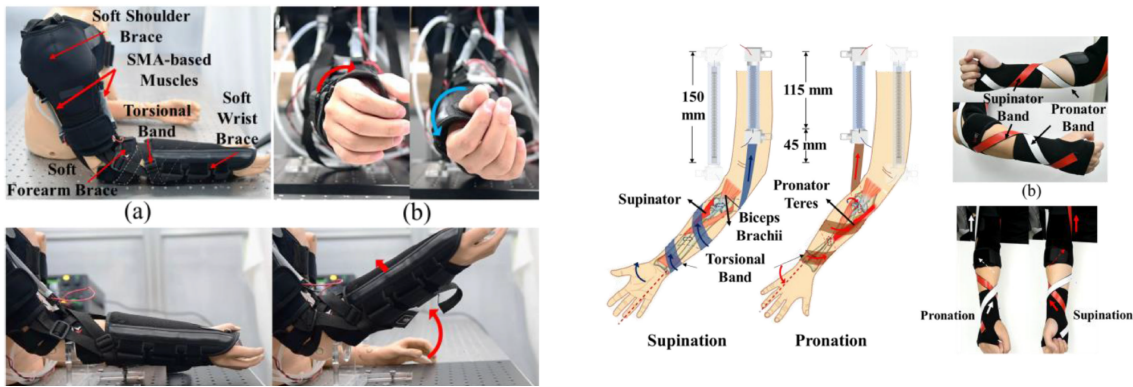


Figure 1.22: Soft exosuit for elbow flexion/extension and wrist pronation/supination with SMA twisting actuators from [31]; on the right, a representation of the SMA paths inside the glove.

Lastly, it can be observed that SMAs possess mechanical properties so advantageous that they are employed even in rigid exoskeletons. Since rigid exoskeletons, by definition, do not require compliance, they generally rely on servo motors because of their ease of control, precision, and ability to generate high forces. Therefore, the fact that SMAs can represent a valid alternative even in systems where softness is not a requirement further highlights the remarkable mechanical properties of these materials. Two examples are the devices by Hope et al. [28] and Serrano et al. [63] shown, respectively, in Figure 1.23 and Figure 1.24.

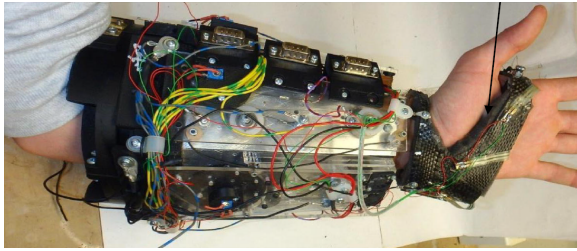


Figure 1.23: Rigid exoskeleton for wrist flexion/extension, radial/ulnar deviation and pronation/supination actuated by SMAs from [28].

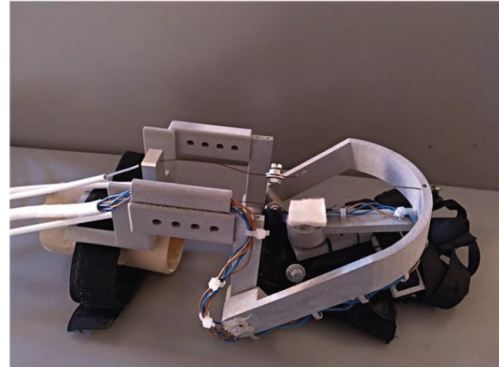


Figure 1.24: Rigid exoskeleton for wrist flexion/extension and radial/ulnar deviation actuated by SMAs from [63].

Despite SMAs being among the best possible choices of soft actuators for wearable devices, they present some limitations that this thesis aims at overcoming.

First of all, the speed of SMA actuators is heavily bottlenecked by their cooling rate. While heating them with electricity is fast, waiting for the heat to dissipate to return to the martensite phase takes much longer even with forced convection through an airflow, limiting their use in high frequency applications [30].

Secondly, SMAs exhibit a wide thermal hysteresis loop (where the heating strain path does not match the cooling strain path) that makes precise position and force control incredibly difficult [46].

Thirdly, generating sufficient heat to trigger actuation means that the SMA wires can reach high temperatures: when used in wearable devices, this poses a significant risk of burning the user's skin unless careful thermal insulation or active cooling is applied [30]. Still connected to this heating aspect, because SMAs are generally heated through Joule effect, they require high electrical currents and suffer from very low overall energy efficiency [74] (often less than 16% [46]), limiting battery life in portable settings.

Lastly, shape memory alloys are generally expensive to fabricate and suffer from thermal fatigue, which significantly reduces their life compared to other actuators [46].

1.2.3. DEAs

Dielectric Elastomer Actuators (DEAs) are a type of electrically driven soft actuator that essentially function as compliant, stretchable capacitors. They are considered part of the category of electroactive polymer (EAP) actuators [6] and consist of an elastic dielectric layer (typically made of silicone or acrylic elastomers) sandwiched between two mechanically compliant, conductive electrodes [46]. When a high direct current (DC)

voltage is applied across the electrodes, the opposite charges attract each other, creating an electrostatic force known as Maxwell stress [6]. This stress squeezes the dielectric layer in the thickness dimension. Because the elastomer is incompressible, this squeezing forces the material to expand outwards in area, generating actuation and movement. When the voltage is lowered, the elastic forces of the material return it to its original shape [74].

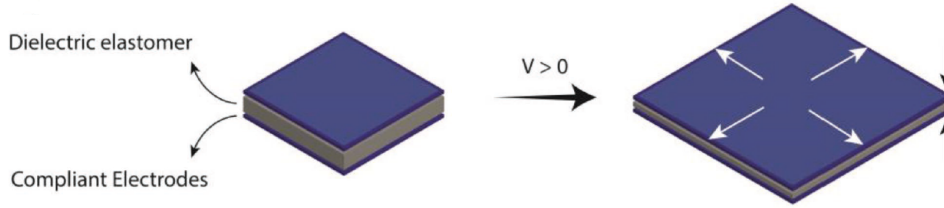


Figure 1.25: Functioning principle of dielectric elastomer actuators (DEAs) from [46].

When evaluated for wearable robotic applications, DEAs show several characteristics that make them attractive as actuators.

First, they can generate large electrically induced area strains, ranging from approximately 120% to over 300%, while producing stresses of up to 3 MPa. These values are comparable to, and in some cases higher than, those of mammalian skeletal muscle [6, 45, 46].

In addition, DEAs have very fast response times, as low as 1 ms, and can operate at frequencies up to the kilohertz range, making them suitable for dynamic human movements [46].

Another important advantage is their self-sensing capability. Since the electrical capacitance of a DEA changes when the actuator deforms, it can also function as a proprioceptive sensor. This allows closed-loop control without the need for additional external sensors [6].

Furthermore, DEAs are intrinsically compliant and operate almost silently, both of which are highly desirable features for wearable assistive devices.

Finally, at low operating frequencies, DEAs can theoretically achieve electromechanical efficiencies of up to 80-90% [7].

Despite the many advantages, DEAs are not yet feasible to be successfully used in soft exosuits and not even rigid exoskeletons. The two devices that come closest to simulate wearable ones are those by Behboodi et al. [6] and Duduta et al. [19], shown in Figures 1.26 and 1.27 .

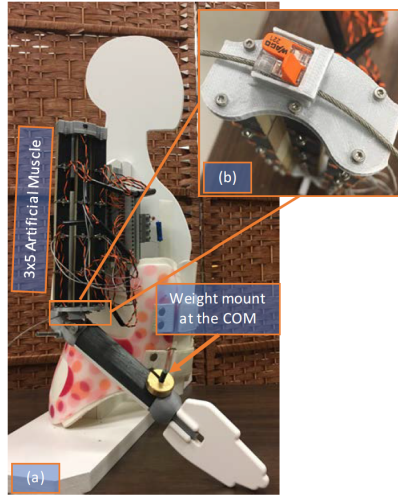


Figure 1.26: Rigid device for elbow flexion/extension on a infant phantom arm with dielectric elastomer actuators from [6].

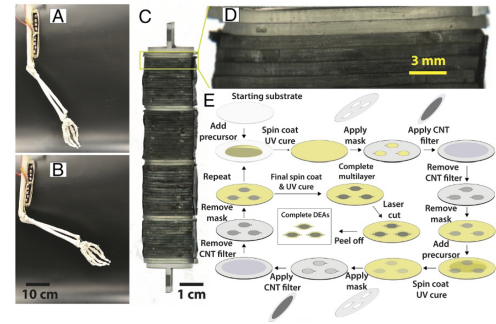


Figure 1.27: Device for elbow flexion/extension on a polymer mock up skeleton consisting of a stack of dielectric elastomer actuators directly mounted on the artificial arm from [19].

The first device consists of a mechanical upper limb phantom equipped with a rigid assembly of stacked DEA artificial muscles arranged in a 3×5 array. The system was developed to actuate elbow flexion in a phantom arm designed to reproduce the dimensions, mass, and motion of the upper limb of a two year old child [6]. The second device uses a linear contracting DEA made of UV curable strain stiffening elastomers and carbon nanotube electrodes to actuate a polymer replica of a human arm bone [19]. Although these systems demonstrate the potential of DEA based actuation for upper limb motion, neither can be considered a true wearable device. In both cases, the actuators are attached directly to artificial limb structures rather than interacting with a human user through compliant attachment mechanisms such as textile sleeves, cuffs, or straps. As a result, these devices do not address important wearable robotics challenges such as force transmission, comfort, and user safety.

Despite their promising characteristics, DEAs are not yet considered a mature technology for wearable robotic systems. Their limited use is mainly due to poor long term durability, susceptibility to dielectric breakdown and material tearing, high voltage requirements, and limited force output when scaled to human relevant dimensions and loads [7].

One of the main limitations of DEAs is their requirement for very high operating voltages, typically ranging from 1 to 15 kV [61]. Besides raising safety concerns when used close to the human body, these voltages require bulky electronic components such as high voltage power supplies and DC-DC converters, reducing system portability [46]. In addition, to

achieve large actuation strains while avoiding dielectric failure, DEA membranes are often pre-stretched and mounted on rigid frames. This requirement partially removes the benefit of the actuator's intrinsic softness and makes integration into fully compliant wearable systems more difficult.

From a control perspective, the viscoelastic behaviour of elastomeric materials introduces nonlinear effects such as hysteresis, creep, and stress relaxation, which can reduce positioning accuracy and make control more challenging [7]. Furthermore, DEAs naturally expand in plane when activated, whereas biological muscles produce contraction. Therefore, obtaining muscle-like linear contraction often requires complex actuator designs or large stacks of individual DEA elements. Although stacking can increase force and displacement output, it also increases system complexity and the probability of failure [19].

For these reasons, despite their excellent compliance, silent operation, and high strain capability, DEAs are currently more suitable as a research technology than as a practical actuation solution for wearable exoskeletons and exosuits.

1.2.4. Cable and tendon driven

The last three categories of soft actuators discussed in this chapter are also the most relevant to this thesis. The first of these is cable and tendon driven actuation.

Cable driven and tendon driven actuation uses electrical motors combined with pulleys and a cable transmission (such as tendons, wires, or Bowden cables) to actuate robotic joints [52]. Even if it does not mimic the contractile action of muscles, the mechanism is heavily bioinspired, functioning like the tendons in the human body which mechanically transfer power from the muscles to the skeleton. By pulling these artificial tendons, the wearable device can control the posture and movement of the user's limbs [26].

The main distinction among transmission types is between Bowden cable transmissions and tendon driven transmissions. A Bowden cable consists of an inner cable sliding inside a flexible outer sheath. This configuration allows the actuator to be placed remotely while the sheath constrains the transmission path and reacts to the cable tension, making it possible to route forces along complex trajectories independently of the relative positions of the motor and the assisted joint. In contrast, tendon driven systems use unconstrained tensile elements that are routed directly across the body or through lightweight guides and anchors integrated into the wearable structure. Since tendons are not enclosed within a sheath, they transmit forces directly to the body along their entire routing path and can more closely reproduce the action of biological tendons. While Bowden cables provide greater flexibility in motor placement and cable routing, tendon driven architectures

generally offer lower transmission losses, improved mechanical transparency, and a more direct interaction with the wearer's anatomy [52].

Among the actuating strategies discussed in this chapter, cable and tendon driven are arguably the most technologically mature mostly because electric motors position themselves on the highest level of the Technology Readiness Level scale already discussed and shown in Figure 1.4.

In fact, the electric motors typically employed in cable driven systems feature high power-to-mass and power-to-volume ratios, allowing for highly compact and lightweight mechanical designs [18]. Moreover, compared to pneumatic artificial muscles (which have nonlinearities and slower response times) or shape memory alloys (which are thermally limited), cable driven systems can produce high torques at a high bandwidth with excellent position, speed, and torque controllability [38]. Finally, the use of cables to transmit forces allows the heavy motors and power units to be placed remotely from the actuated joint (for example in a backpack in case of elbow or shoulder exosuits or proximally on the arm in case of hand or wrist exosuits) reducing the overall collocated weight, encumbrance, and inertia on the user's limb, which significantly lowers the metabolic impact on the wearer [30].

This last aspect is represented in an emblematic way by the device by Lessard et al. [40] shown in Figure 1.28. It is a multi-joint upper extremity exosuit that uses a backpack motor module to actuate seven cables routed specifically along the human body's natural "lines of non-extension" (preferential routes on the skin which neither stretch nor compress as the human moves) to assist the shoulder, elbow, and wrist.

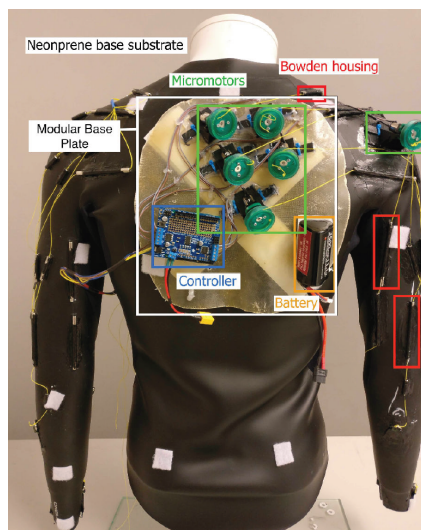


Figure 1.28: Soft exosuit for wrist pronation/supination, elbow flexion/extension, shoulder flexion/extension and abduction/adduction with cable driven actuation from [40].

Among the many joints and degrees of freedom covered by this device, the most interesting one for this thesis is wrist pronation/supination, that relies on the already seen strategy of twisting cables around the forearm, as is clearly shown by the cables' paths in Figure 1.29.

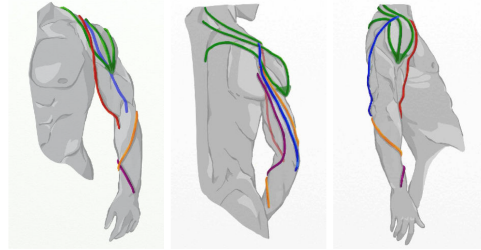


Figure 1.29: Cables' path from [40].

Unlike the previously discussed pneumatic and SMA based devices, where the contractile action is distributed along the entire actuator length, in Bowden cable systems the force is typically applied only at the cable termination point. In the device of Lessard et al. [40], the torque responsible for the forearm rotation is therefore generated primarily at the wrist interface. Although this approach successfully produces pronation/supination, it does so by acting mainly at the distal end of the forearm. Anatomically, however, pronation/supination is not a wrist movement but a distributed rotation of the radius around the ulna along the entire forearm. Consequently, a transmission strategy that applies forces over a larger forearm region may provide a biomechanically more representative interaction and better exploit the characteristics of this degree of freedom.

Another cable driven device, which lies at the boundary between soft and compliant, responsible for wrist pronation/supination with the same disadvantage of pulling only from the wrist is that by Su et al. [68] shown in Figure 1.30, that exploits a new configuration in which the cables pull a rotating wrist band that then transfers motion to the wrist itself.

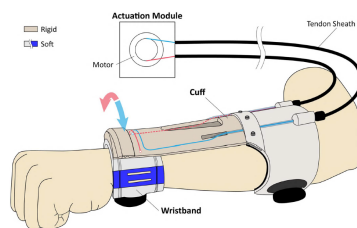


Figure 1.30: Soft exosuit for wrist pronation/supination with cable driven actuation from [68].

The two examples just reported are both driven by Bowden cables. Examples of soft exosuits for upper limb that are, instead, tendon driven are, for instance, those by Little et al. [42] and Alapati et al. [2] shown, respectively, in Figures 1.31 and 1.32. The latter obtains wrist pronosupination with the same strategy of the rotating wrist ring as [68].



Figure 1.31: Soft exosuit for elbow flexion/extension with tendon driven actuation from [42].

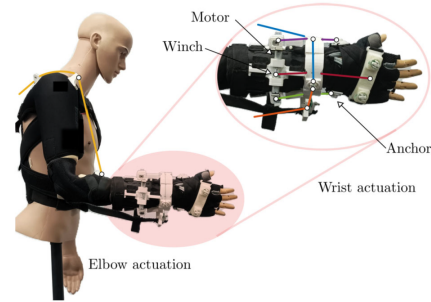


Figure 1.32: Soft exosuit with compliant elements for elbow flexion/extension, wrist flexion/extension, radial/ulnar deviation and pronation/supination with tendon driven actuation from [2].

The maturity and robustness of this actuation technology are further evidenced by the presence of commercial devices alongside research prototypes, as is the case for Carbonhand and Ironhand by Bioservo [8], two devices to assist fingers flexion (shown respectively in Figure 1.33 and Figure 1.34), the first for medical assistance purposes and the second to increase grip strength and reduce wrist strain in industrial environments.



Figure 1.33: Carbonhand - commercial soft exosuit for fingers flexion for health-care assistance and rehabilitation with cable driven actuation from [8].



Figure 1.34: Ironhand - commercial soft exosuit for fingers flexion for industrial applications with cable driven actuation from [8].

Most of the discussed applications, and cable driven exosuits in general, employ Bowden cable transmissions rather than free tendons because they allow the routing of forces

along arbitrary paths while keeping the motors remotely located. However, Bowden cables introduce a limitation that is absent in free tendon systems: friction between the cable and its sheath. This friction depends on factors such as cable tension and routing curvature, leading to transmission losses, cable slack, reduced mechanical transparency, and increased control complexity [12]. Additional limitations common to both cable and tendon driven systems include their ability to generate forces only in tension and the possibility of energy losses caused by deformation or slippage of the soft wearable structure. For example, part of the actuator work may be absorbed by deformation of the textile anchors rather than being fully transferred to the user's limb [52].

Nevertheless, these limitations are often manageable in practice. The unidirectional nature of tendon transmissions can be addressed through antagonist actuator pairs or, in some applications, by exploiting gravity or passive elastic elements to provide the return motion. Likewise, transmission losses associated with the soft structure can be significantly reduced through careful mechanical design, optimized attachment strategies, and appropriate load distribution across the wearable interface.

Despite the advantages of cable and tendon driven transmissions, the electric motors that actuate them introduce several limitations that must be considered in wearable applications.

First of all, they are often noisy during operation: this constant mechanical noise reduces user acceptance and comfort, particularly when the device is intended for unobtrusive daily use, whereas other soft actuators (like DEAs or SMAs) operate completely silently [24].

Then, even if small motors can be used (especially for upper limb applications since the weights involved are not huge), they are still bulky, heavy and rigid, reducing the overall volume, weight, aesthetic and compliance especially compared to light and totally soft actuators such as pneumatic, SMAs, DEAs and electrohydraulic [6].

Furthermore, servo motors do not inherently possess a low power catch state, they must consume continuous electrical power to hold a joint in a fixed position against a load, impacting on energy consumption.

Finally, the presence of electric motors creates electromagnetic fields that can have negative impact on sensors such as IMUs (that cannot orientate appropriately because their magnetometer cannot properly align with the north pole due to electromagnetic interference).

In summary, cable and tendon driven systems represent one of the most mature and widely adopted actuation strategies for soft wearable robots. Their ability to combine the controllability and efficiency of electric motors with lightweight and compliant force trans-

mission makes them particularly attractive for upper limb assistance. While Bowden cable transmissions enable motor relocation at the cost of friction and reduced transparency, tendon driven architectures offer a simpler transmission path and more direct force transfer. For these reasons, tendon driven actuation was selected for the exosuit developed in this thesis, as it provides a favorable compromise between controllability, mechanical simplicity, wearability, and compatibility with the targeted wrist and forearm motions.

1.2.5. HASEL

Electrohydraulic actuators are a class of soft transducers that combine the high speed, self-sensing abilities and muscle-like electrical operation of dielectric elastomer actuators with the mechanical versatility of fluidic actuators. They are most prominently known in soft robotics as HASEL (Hydraulically Amplified Self-healing ELectrostatic) actuators.

They typically consist of a flexible, often inextensible, polymeric shell (such as a plastic pouch) that is filled with a liquid dielectric, like vegetable-based transformer oil or silicone oil. A pair of flexible electrodes is placed on the outside of the shell, partially covering it.

When a high voltage is applied to the electrodes, it creates an electric field that generates an electrostatic force known as Maxwell stress. This stress causes the opposing electrodes to rapidly attract each other and pull together (this process is referred to as "electrostatic zipping"). As the electrodes zip shut, they locally pressurize and physically push the liquid dielectric into the regions of the shell that are not covered by the electrodes. Because the fluid is incompressible and the shell is flexible, this internal hydraulic pressure forces the structure to deform (for example, passing from a flat pouch to a circular tube), which generates mechanical work and movement [1, 32, 47].

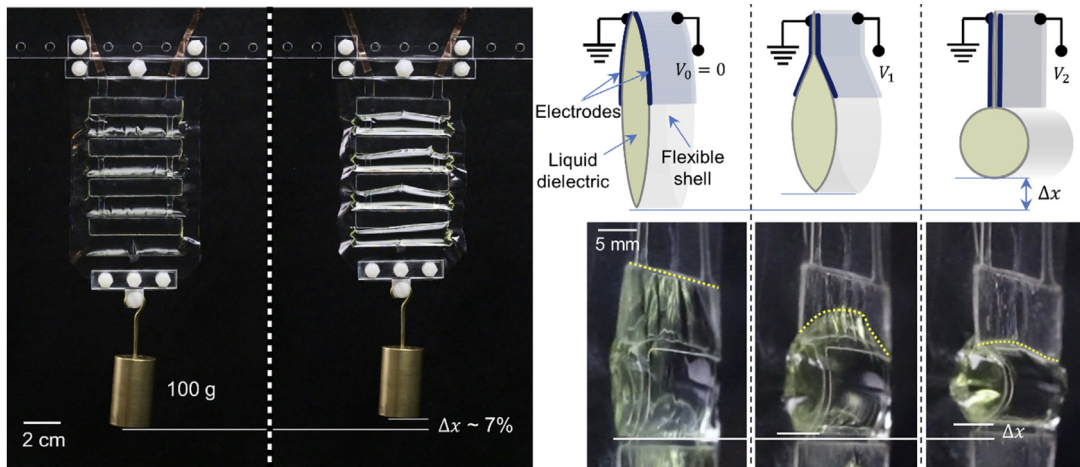


Figure 1.35: Zipping process in HASEL actuators from [33].

The kind of HASEL actuators shown in Figure 1.35 is referred to as Peano-HASEL: they are shaped as a series of rectangular pouches connected in series and are characterized by an overall contraction [32]. However, this is not the only possible design: the actuating principle is very versatile and several different geometries can be created in order to obtain a custom actuation. For example, the pouches can be arranged in a circle in order to emulate the contraction of circular muscles (such as gastrointestinal muscles) [71]. Another interesting design to increase the overall linear contraction relies on the exploitation of custom made pulleys around which the actuators can wrap [71]. Also, despite the actuators' functioning being contractile, an expansive overall action can be obtained with doughnut shaped actuators, mostly if stacked one on top of the other [1]. Finally, strain limiting layers can be used to create curling or angular deformations [34, 47]. All these geometries are shown in Figure 1.36.

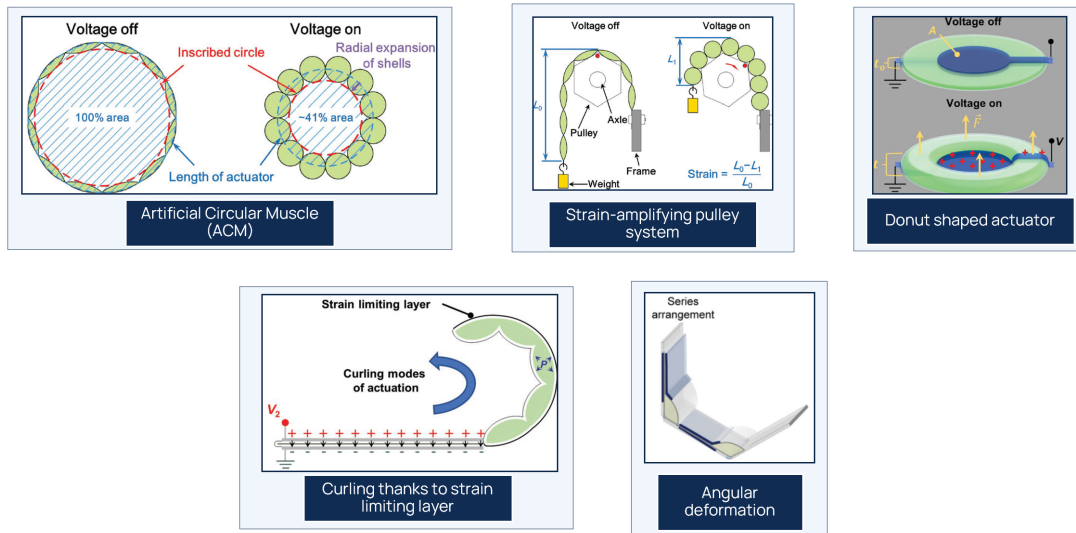


Figure 1.36: Some possible geometries of HASEL actuators from [1, 34, 47, 71].

This extreme customizability makes HASEL actuators perfect in several different fields, ranging from robot end effectors to bioinspired robots, as shown in Figure 1.37.

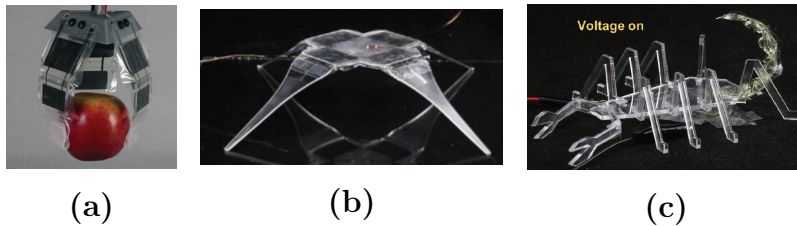


Figure 1.37: Different applications using HASEL actuators. (a) Robot manipulator end effector from [34] (b). Spider inspired robot from [34]. (c) Scorpion robot from [47].

Other than being extremely versatile, HASEL actuators not only share both the advantages of DEAs and hydraulic actuators but also compensate some of the individual defects of DEAs with the hydraulic part and vice versa.

For example, unlike solid DEAs that are permanently destroyed if a high voltage spark burns through them, the liquid dielectric in HASELs immediately returns to an insulating state after a dielectric breakdown event, allowing the actuator to seamlessly recover and continue functioning (hence the appellation "self healing") [47]. Moreover, because HASELs generate hydraulic pressure locally, they avoid the viscous delays and energy losses associated with pumping fluid through networks of tubes in standard pneumatic actuators: therefore they can react in as little as 12 milliseconds, operate smoothly at high frequencies (20 Hz to over 100 Hz [47]), and deliver specific power outputs that double or greatly exceed that of mammalian skeletal muscle (up to 614 W/kg) [1].

Furthermore, similarly to DEAs, they are self sensing thanks to the electrodes functioning like a capacitor: by superimposing a low voltage AC signal over the high voltage driving signal, the changing capacitance can be measured in real time, allowing the actuator to self sense its exact deformation and position without needing external bulky sensors [1]. Lastly, since they do not require stretchable electrodes, they can be fabricated rapidly using cheap, commercially available materials (like BOPP films used in food packaging) and simple heat sealing techniques; depending on the chosen materials, they can even be made highly optically transparent [32].

Despite all these incredible advantages that picture HASEL actuators as the perfect solution for soft exosuits, this technology is so recent (it has been first introduced to the scientific community in January 2018 [32, 47]) that it just started entering the field of collaborative robotics. Up to now, the only pseudo-wearable device using HASEL actuators is an active wrist tremor suppression device designed by Shagan Shomron et al. [65] and shown in Figure 1.38.

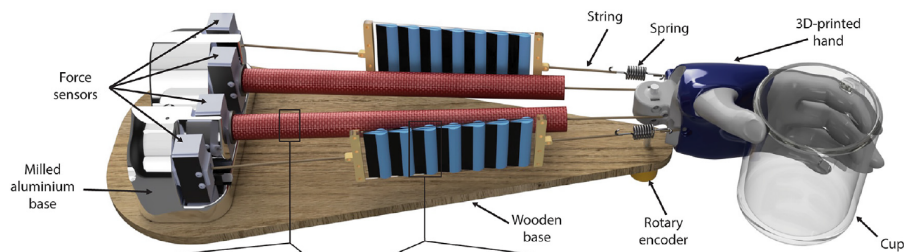


Figure 1.38: Device for wrist tremor suppression in flexion/extension direction with HASEL actuators from [65].

Even though this paper has the merit of being the first collaborative robot actuated by

HASEL actuators, it is not an actual wearable device since it is not designed to be worn by a human: instead, it is mounted on a 3D printed biomimetic robotic testbed of a forearm. Another important aspect to highlight is that this device is only responsible for suppressing an artificial sinusoidal tremor generated by a couple of pneumatic actuators: it is not responsible for moving a real hand in order to do real life activities, meaning the dynamic profile and load requirements differ significantly from those of a generalized assistive exoskeleton.

As a consequence, to this day a fully wearable human-worn exosuit using HASELs has not yet been realized nor designed. Also, not only developing such a soft exosuit would be at the forefront of research in the field of soft exosuits, it would also be a cutting edge application in the field of soft actuators since applications of HASELs with forces as big as those involved in soft exosuits have not been created yet not even in other fields.

Nevertheless, HASEL actuators exhibit certain limitations.

First of all, because the Maxwell stress relies on intense electric fields, HASEL actuators require extremely high operating voltages, typically ranging from 2 kV to over 10 kV [1]: generating and routing these high voltages in an untethered system in close proximity to the human body raises major electrical safety and shock concerns [55]; however, with the proper electrical insulation this problem can be overcome [64].

Secondly, HASEL actuators are not extremely robust: while the liquid dielectric self heals after an electrical breakdown, the repeated high voltage arcing can eventually melt or burn physical holes through the thin thermoplastic shell, causing the liquid dielectric to leak out and leading to irreversible actuator failure; moreover, some dielectric films used to build HASELs tend to accumulate and retain electrical charge over time without dissipating it and causing the actuator not to fully relax to its initial position. This latter aspect can be solved by actively alternating the polarity of the applied voltage during every actuation cycle, even though this increases the bulk and complexity of the power supply [32].

Lastly, especially at lower voltages, the forces and strains of HASEL actuators are not very high and this is the primary reason of why they have been only used for small and light applications up to now. However, a smart design of the geometry can reduce this problem and make HASELs appropriate also for wearable robotics. The wrist is also the perfect joint because, compared to others (such as lower limbs), it involves forces that are not too big.

In conclusion, with the proper precautions and clever design solutions, HASEL actuators represent an extremely promising actuating mode for a wrist soft exosuits and will be largely discussed in this thesis.

1.2.6. HALVE

HALVE (Hydraulically Amplified Low-Voltage Electrostatic) actuators are an advanced class of soft electrohydraulic artificial muscles developed to overcome the high voltage limitations of traditional HASEL actuators.

Like HASEL actuators, HALVE actuators consist of a liquid filled pouch (using dielectric oil as a hydraulic amplification medium) with electrodes covering a portion of the pouch. When a voltage is applied, opposite electric charges build up on the electrodes, creating a force that causes them to zip together, causing a linear contraction of the pouch.



Figure 1.39: HALVE actuator being filled with oil using a syringe with a filed tip not to break the Mylar[®] film.

In standard HASEL actuators, a single thermoplastic film (like BOPP or BoPET, commercially known as Mylar) is used for both the structural shell and the dielectric layer. This functional duality forces a compromise between mechanical strength and dielectric properties, leading to the requirement of extremely high actuation voltages, generally greater than 6 kV to generate enough force and strain.

HALVE actuators solve this by decoupling the structural and dielectric functions into a three layer composite structure shown in Figure 1.40:

- an outer structural shell made from a *mechanically* strong polymer (like BoPET or BoPVDF), acting as both the load bearing element and an outer electrical insulator;
- thin aluminum or chrome/gold *electrodes* placed inside the shell;
- a very thin (5 μm) layer of a relaxor-ferroelectric terpolymer called P(VDF-TrFE-CTFE) cast directly over the electrodes and functioning as a high-permittivity *dielectric* layer.

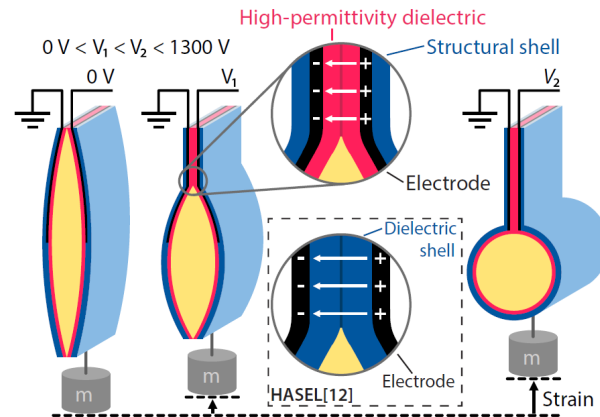


Figure 1.40: Three layer structure of HALVE actuators and comparison with two layer structure of HASELs from [25].

By using two distinct layers, beside the electrodes, instead of just one, both mechanical and dielectric properties are increased. In particular, by using a dedicated high energy density dielectric layer, HALVE actuators significantly boost electrostatic performance, achieving the same force-strain performance as HASEL actuators but at 4.9 to 6.6 times lower driving voltages (e.g., 1100 V instead of 5400 V or more). Moreover, with this new configuration, the electrodes are not exposed anymore: therefore, the actuator is water resistant and safe to touch [25].

The former property introduces this category of actuators to the huge world of marine and underwater applications, like the robotic fish developed by Gravert et al. [25] shown in Figure 1.41.

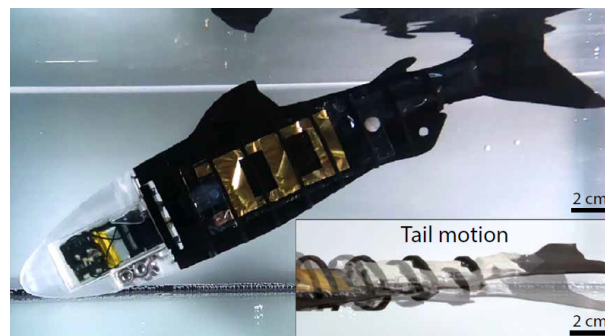


Figure 1.41: Robotic fish with HALVE actuators from [25].

The latter property, instead, is perfect for applications involving direct interaction with humans, such as wearable robots.

In general, HASEL actuators are very appropriate for soft exosuits but HALVE actuators

might be even more because they present advantages with respect to their predecessors HASELs other than the ones already discussed. For example, since they operate at around 1000 V rather than 6000 V or more, they can be driven by commercially available, cheap, and tiny MOSFET components rather than bulky, expensive, and power consuming high voltage optocouplers: this allows the power supply to be drastically reduced in weight (for example to just 13–15 grams including the battery), making them ideal for truly untethered wearable exosuits. Moreover, while in standard BOPP HASELs an electrical spark (dielectric breakdown) melts a hole through the shell, leaking the oil and permanently destroying the actuator, in HALVE actuators, if a breakdown occurs, the thin inner electrode is locally destroyed (which clears the short circuit), but the robust outer structural shell survives intact. This prevents oil leaks and allows the HALVE actuator to endure dozens of dielectric breakdowns while remaining fully operational [25].

Another positive aspect of HALVE actuators to consider in comparison with all the other analyzed categories is that they match and even surpass the performance of mammalian skeletal muscle, achieving an average specific power density of 50.5 W/kg and peak strain rates of 971% per second [25]. This is extremely valuable when studying wearable systems actuated by artificial muscles because it allows the design logic to be more similar to the real human body.

Conversely, HALVE actuators are subject to some inherent drawbacks absent in HASEL technology.

First of all, while their average specific power is excellent, currently their maximum force density is generally less than that of high-voltage BoPET HASELs, that was still described as limited. This is because the P(VDF-TrFE-CTFE) films are manufactured in standard lab environments (through a technique called blade casting) rather than clean-rooms, resulting in the possible introduction of dust and impurities. These defects cause the actuators to prematurely break down at field strengths of around 120 V/ μm , rather than their theoretical maximum of 350 V/ μm : therefore, at very high voltages (more than 6 kV), traditional industrial HASELs still output more force.

Then, during durability testing, failure often occurs at the heat seals: the sealing process typically only fuses the inner P(VDF-TrFE-CTFE) layers and not the robust outer structural shell, creating a weak point that can degrade over thousands of cycles. To improve the pouch's heat seal, researchers had to use BoPVDF as the outer shell instead of BoPET substantially increasing material costs, equipment requirements, and manufacturing time and labor [25].

The current limitations of HALVE actuators are primarily developmental, highlighting the broader challenge with this technology: while theoretically excellent and potentially the

most suitable for this thesis, HALVEs are in their infancy (introduced in 2024 with only one primary publication [25]). Consequently, they sit very low on the TRL scale (Figure 1.4) and require further refinement to become robust enough for practical applications beyond small scale laboratory prototypes.

1.3. Existing wrist devices

A comprehensive review of the current literature reveals a growing interest in wearable soft robotics for upper limb assistance and rehabilitation; however, devices specifically targeting the wrist remain limited. The comparative analysis detailed in Table A.1 highlights several distinct trends and technological limitations within the current state of the art.

Primarily, the vast majority of existing soft exosuits focus almost exclusively on the flexion/extension (FE) degree of freedom. While this movement is critical for grasping and basic manipulation, it is conceptually simpler to actuate by routing actuators along the dorsal and palmar sides of the hand. Conversely, pronation/supination (PS) remains largely underexplored. The few devices that do attempt PS often rely on twisting pneumatic actuators or complex cable routing strategies that pull primarily at the distal end of the wrist. Because PS is a distributed rotation of the radius around the ulna along the entire forearm, these localized actuation strategies often fail to provide a biomechanically optimal force transmission, leading to inefficiencies and reduced user comfort.

Furthermore, multi degree of freedom soft exosuits are fairly rare. Very few devices successfully combine both FE and PS in a single, cohesive, and lightweight platform. Devices that do manage multiple DoFs (and not all of them manage to control all the DoFs combined and, instead, control them one at a time instead) often suffer from the inherent drawbacks of their chosen actuators: pneumatic systems generally present significant non collocated bulk due to compressors and valves, severely limiting true portability; Shape Memory Alloy (SMA) devices, while lightweight, struggle with thermal hysteresis and low actuation speeds, making them difficult to utilize for dynamic, high frequency daily activities.

Lastly, while cable and tendon driven systems offer superior control, high bandwidth, and reliability, the current tendon driven wrist devices lack an optimized mechanical architecture to simultaneously support FE and a biomechanically accurate PS movement without adding excessive weight or friction. This demonstrates a clear need for a highly portable, multi-DoF soft exosuit capable of delivering precise, biomechanically sound assistance for both wrist flexion/extension and pronation/supination.

1.4. Technological gap

Despite the rapid advancements in soft wearable robotics, a significant technological gap persists in the development of wrist exosuits. Specifically, there is a lack of fully portable, comfortable, and highly controllable soft devices capable of simultaneously assisting both wrist flexion/extension and the biomechanically complex pronation/supination movements [2, 68]. Existing solutions are either confined by the bulkiness of pneumatic systems [21], the thermal limitations of SMAs [52], or an inability to accurately distribute rotational forces along the forearm [68]. Furthermore, while novel electrohydraulic actuators (such as HASEL and the recently developed HALVE) exhibit extraordinary, muscle-like performance metrics [1, 25], they have yet to be systematically applied or validated in wearable, portable robots capable of supporting human scale forces [65].

To bridge this gap, this thesis proposes the design, development, and validation of a novel, fully wearable soft exosuit tailored specifically for wrist flexion/extension and pronation/supination. The initial development utilizes a tendon driven transmission to guarantee robust, high bandwidth control and reliable force routing, resulting in a functional prototype that is subjected to a comprehensive validation protocol on both healthy subjects and target patients.

Building upon the mechanical and control architecture of this validated tendon driven system, the second primary contribution of this thesis is a complete feasibility study exploring the integration of cutting edge electrohydraulic actuators (HASEL [32] and HALVE [25]). By adapting the exosuit's design to accommodate these next generation artificial muscles, this work investigates their potential to entirely replace servo motors, paving the way for truly untethered, silent, and biomimetic soft wearable robotics for upper limb rehabilitation.

2 | Materials and Methods

The goal of this thesis is to develop and validate a tendon driven soft exosuit for wrist flexion/extension and pronation/supination, intended for both assistance and rehabilitation applications, and to carry out a complete feasibility study of the same soft exosuit actuated by HASEL and HALVE actuators. This chapter presents the different phases of the development and validation process.

First, the requirements of the system are defined, as they are essential for establishing the design choices from both a mechanical and a control perspective.

The mechanical design is then studied in detail, starting with an in depth analysis of the forces and moments involved and a search for the optimal geometry to achieve the required degrees of freedom. This is followed by a description of the construction of the tendon driven prototype and concludes with a feasibility study of the same prototype actuated by HASEL or HALVE actuators, together with a comparison between tendon driven and electrohydraulic actuation.

The chapter then moves on to the control design, first introducing the general strategy adopted to control the prototype and subsequently examining in detail the different high level control strategies implemented, as well as the graphical user interface developed for their use.

Lastly, a thorough description of the validation protocol is provided. The results of the tests described in this section are presented in Chapter 3 and discussed in Chapter 4.

2.1. System requirements

The first step towards the development of a robotic system is to determine its requirements, meaning how the hardware and software of the system should behave and the constraints associated with its operation. The process leading to the definition of these requirements is called Requirements Engineering (RE) and generally consists of different phases: *elicitation* (the discovery of stakeholders' needs), *analysis and negotiation* (selection of the most useful requirements from the set of elicited ones), *specification* (writing

the list of selected requirements), *validation and verification* (decision of the tests to verify if the requirements are met) and *management* (process of documentation and controlling change in the requirements) [37].

In this thesis, the elicitation, analysis and negotiation phases were conducted by accurately studying the literature and extracting the desired properties from similar articles. The validation and verification phase, instead, is thoroughly discussed in Section 2.4.

This section focuses on the specification and reports the list of the chosen requirements divided into *functional* and *non functional*.

2.1.1. Functional requirements

Functional requirements define the basic system behaviour, meaning the features that allow the system to function as intended: if the functional requirements are not met, the system does not work.

The functional requirements for the soft exosuit of interest are listed in Table 2.1.

Table 2.1: Functional requirements

REQUIREMENT	DESCRIPTION
F1 - Assisted movements	The system must support the movement of the wrist, in particular its flexion/extension and pronation/supination.
F2 - Operating modes	The system must operate in an active assistive configuration and in a passive configuration ^a , covering both assistance and rehabilitation purposes.
F3 - User interaction	The system must allow the user to start/stop the device and trigger an emergency stop.

^a A detailed explanation of the different modalities of high level control is present in Section 2.3.

2.1.2. Non functional requirements

Non functional requirements define the system features and general characteristics that affect the user experience. Therefore, they represent product properties that focus on user expectations without altering the basic functionality of the system.

The non functional requirements for the soft exosuit of interest are listed in Table 2.2.

Table 2.2: Non functional requirements

REQUIREMENT	DESCRIPTION
N1 - Range of motion	The exosuit must be able to provide the range of motion for ADL in an adult wrist, consisting in 50° of flexion/extension and 50° of pronation/supination ^a . It should not overcome the maximum RoM, that is 70–80° of flexion, 70–80° of extension, 70–85° of pronation and 70–85° of supination ^b [52].
N2 - Assistance capability	The system shall provide assistance corresponding to at least 50% of the moment acting on the forearm and hand during active assistive operation ^c and it shall be capable of providing up to 100% compensation of the moment acting on the forearm and hand during passive operation.
N3 - User controllability	The system shall allow the user to control wrist motion with sufficient precision ($RMSE < 6^{\circ d}$) and to maintain intermediate positions within the range of motion.
N4 - Speed	The system should be actuated at speeds comparable to those of mammalian muscles (peak strain rate of 500%/s [25]).
N5 - Transparency	When unpowered, the suit shall allow for comfortable execution of ADLs with minimum perceived resistance.
N6 - Comfort of wearing	The system should not provoke discomfort for the user during operation.
N7 - Ease of operating	The device should be intuitive to wear. Also, donning should take less than two minutes and doffing less than one minute.
N8 - Portability and weight	The system must be portable enough to be usable without restricting user movements in a typical seated/desktop environment and it should have a collocated weight < 300 g.
N9 - Adaptability to different users	The device should be suitable for people with different hand sizes.
N10 - Emergency stop	The system shall stop safely when triggered.
N11 - Electrical safety	The system shall ensure electrical insulation of the user.

^a The RoM of ADLs is defined as 15° of flexion and 35° of extension from the horizontal position. The requirement specifies only the total RoM because the system aims to be adjustable according to user needs. The same principle applies to pronation/supination. Furthermore, this limitation is necessary to avoid pushing the joint to its anatomical limits, which could be painful and potentially hazardous.

^b Flexion/extension angles are conventionally measured with the hand in the horizontal position, while pronation/supination angles are measured with the hand in the vertical position, as shown in Figure 1. However, in the remainder of this thesis, the horizontal position is considered the default reference for all movements.

^c A 50% compensation level in *assistance as needed* approaches is considered sufficient for upper limb applications [4].

^d The quantities reported in this list were defined through a comparative analysis of the wrist devices presented in Chapter 1 and Table A.1.

2.2. Mechanical design

This section focuses on the design and construction of the mechanical components of the wrist soft exosuit. Since the wrist is located between the forearm and the hand, the actuators responsible for wrist motion are anchored to two wearable interfaces: a forearm sleeve and a fingerless cycling glove worn on the hand. A detailed description of the physical components used in the final prototype is provided in Section 2.2.3. The purpose of this initial part is to analyze the geometry and arrangement of the actuators connecting the glove to the sleeve to enable wrist motion.

2.2.1. Flexion/extension

Introduction and general strategy

As discussed in Chapter 1, the principle behind flexion/extension assistance involves placing one or more actuators on the dorsal side of the hand and one or more on the palmar side. Depending on the chosen actuator technology, their contraction or expansion generates wrist motion.

In the case of tendon driven actuation, force transmission is possible only in tension, as tendons cannot transmit compressive forces. Consequently, tendons routed on the dorsal side of the hand produce wrist extension, whereas tendons routed on the palmar side produce wrist flexion. In this design, however, active actuation is limited to wrist extension. This is primarily because typical functional hand postures are with the palm facing down, meaning gravity naturally assists flexion but opposes extension. Therefore, an extension only approach eliminates the need for a palmar actuation unit¹, allowing the system to use just one servomotor and significantly reducing overall mass and volume.

To determine the required motor characteristics and to define the geometry of both the transmission system and the exosuit, a preliminary analysis of the forces and displacements involved in wrist motion is necessary.

Forces and moments

In this study, the main load considered is the gravitational moment generated by the weight of the hand acting at its center of mass. Additional contributions could also be

¹While a single servo motor could theoretically drive both flexion and extension through antagonistic cable routing (e.g., wrapping two wires in opposite directions on the same pulley), a single tendon per actuator architecture was deliberately selected. This restriction ensures that the mechanical transmission remains logically identical to the alternative HASEL actuator configuration introduced in Section 2.2.4, allowing for a direct performance comparison between the two actuation methods.

taken into account, such as the dynamic moment associated with hand acceleration and inertia, as well as the resistive action of passive tissues and antagonist muscles. However, for the relatively slow wrist motions considered in this work, the dynamic moment is expected to be significantly smaller than the gravitational moment, while the contribution of passive tissues and antagonist muscles is considered negligible in comparison. For this reason, the analysis is limited to static forces and moments under quasi-static conditions.

It is important to note that this assumption relies on modeling a soft, non-spastic limb. In many clinical scenarios, particularly those involving patients recovering from stroke, spasticity may present a significantly higher active resistance that can lock the hand in a flexed posture. Accounting for pathological spasticity would require vastly different dynamic models and higher actuation forces. Therefore, for the scope of this thesis, only the gravitational moment is considered. This provides a baseline and a reasonable order of magnitude for the target torque that the exosuit must provide to overcome standard resistance, establishing a model that could eventually be scaled to address spastic profiles.

The mass of the hand and the position of its center of mass with respect to the wrist joint can be computed as described by Winter et al. [72], as shown in Figures 2.1 and 2.2.

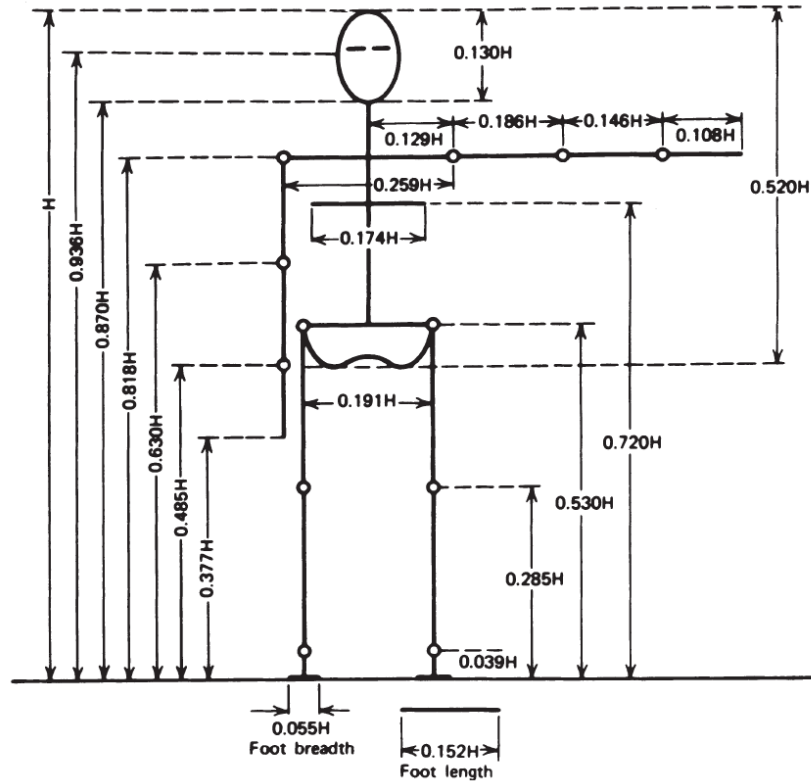


Figure 2.1: Body segment lengths expressed as a fraction of body height H from [72].

Segment	Definition	Segment Weight/Total Body Weight	Center of Mass/ Segment Length		Radius of Gyration/ Segment Length			Density
			Proximal	Distal	C of G	Proximal	Distal	
Hand	Wrist axis/knuckle II middle finger	0.006 M	0.506	0.494 P	0.297	0.587	0.577 M	1.16
Forearm	Elbow axis/ulnar styloid	0.016 M	0.430	0.570 P	0.303	0.526	0.647 M	1.13
Upper arm	Glenohumeral axis/elbow axis	0.028 M	0.436	0.564 P	0.322	0.542	0.645 M	1.07
Forearm and hand	Elbow axis/ulnar styloid	0.022 M	0.682	0.318 P	0.468	0.827	0.565 P	1.14
Total arm	Glenohumeral joint/ulnar styloid	0.050 M	0.530	0.470 P	0.368	0.645	0.596 P	1.11
Foot	Lateral malleolus/head metatarsal II	0.0145 M	0.50	0.50 P	0.475	0.690	0.690 P	1.10
Leg	Femoral condyles/medial malleolus	0.0465 M	0.433	0.567 P	0.302	0.528	0.643 M	1.09
Thigh	Greater trochanter/femoral condyles	0.100 M	0.433	0.567 P	0.323	0.540	0.653 M	1.05
Foot and leg	Femoral condyles/medial malleolus	0.061 M	0.606	0.394 P	0.416	0.735	0.572 P	1.09
Total leg	Greater trochanter/medial malleolus	0.161 M	0.447	0.553 P	0.326	0.560	0.650 P	1.06
Head and neck	C7–T1 and 1st rib/ear canal	0.081 M	1.000	— PC	0.495	0.116	— PC	1.11
Shoulder mass	Sternoclavicular joint/glenohumeral axis	—	0.712	0.288	—	—	—	1.04
Thorax	C7–T1/T12–L1 and diaphragm*	0.216 PC	0.82	0.18	—	—	—	0.92
Abdomen	T12–L1/L4–L5*	0.139 LC	0.44	0.56	—	—	—	—
Pelvis	L4–L5/greater trochanter*	0.142 LC	0.105	0.895	—	—	—	—
Thorax and abdomen	C7–T1/L4–L5*	0.355 LC	0.63	0.37	—	—	—	—
Abdomen and pelvis	T12–L1/greater trochanter*	0.281 PC	0.27	0.73	—	—	—	1.01
Trunk	Greater trochanter/glenohumeral joint*	0.497 M	0.50	0.50	—	—	—	1.03
Trunk head neck	Greater trochanter/glenohumeral joint*	0.578 MC	0.66	0.34 P	0.503	0.830	0.607 M	—
Head, arms, and trunk (HAT)	Greater trochanter/glenohumeral joint*	0.678 MC	0.626	0.374 PC	0.496	0.798	0.621 PC	—
HAT	Greater trochanter/mid rib	0.678	1.142	—	0.903	1.456	—	—

Figure 2.2: Anthropometric data from [72].

Given a reference subject of height H and mass M , the mass of the hand, the total hand length, and the distance of the center of mass from the wrist joint are computed as shown in Equations 2.1, 2.2, and 2.3, respectively.

$$m_{\text{hand}} = 0.006 M \quad (2.1)$$

$$P = 0.108 H \quad (2.2)$$

$$r = 0.506 P \quad (2.3)$$

The weight of the hand is calculated as:

$$\mathbf{F}_g = m_{\text{hand}} \mathbf{g} \quad (2.4)$$

(Note: The numerical evaluation of these variables, based on the anthropometric data of a representative reference user, is detailed subsequently in Section 2.2.3.)

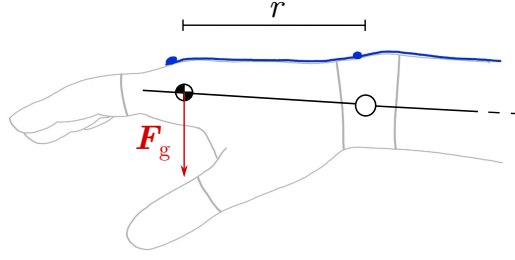


Figure 2.3: Force of gravity acting on the hand.

The moment required to counteract the force of gravity depends on the inclination angle θ of the hand in the flexion/extension direction. While the force vector remains vertical, its moment arm with respect to the wrist joint varies as $r \cos \theta$. The least advantageous configuration, corresponding to the highest possible moment, occurs at $\theta = 0^\circ$ (defined as the horizontal position). In this position, the maximum moment is:

$$\mathbf{M}_{\max \text{ g, FE}} = \mathbf{F}_g r =: \mathbf{M}_{\text{comp, FE}} \quad (2.5)$$

The notation $\mathbf{M}_{\text{comp, FE}}$ denotes the moment that must be compensated to overcome the entirety of the gravitational effect. As outlined in the *Assistance capability* non functional requirement, the system must provide at least 50% moment compensation for the active assistive mode and 100% for the passive mode. Therefore, setting $\mathbf{M}_{\text{comp, FE}}$ equal to the maximum gravitational moment $\mathbf{M}_{\max \text{ g, FE}}$ provides a general baseline for subsequent calculations.

Once the required moment to compensate is defined, it is equated to the moment provided by the exosuit, which is expressed as a function of the unknown tendon pulling force. Solving this equilibrium equation yields the required tendon tension, a fundamental parameter for motor sizing.

However, to formulate this equation, the geometry of the exosuit must first be defined.

Possible configurations

To determine the optimal tendon routing, several configurations are evaluated.

Stitched configuration The most intuitive routing strategy is the *stitched configuration*, illustrated in Figure 2.4.

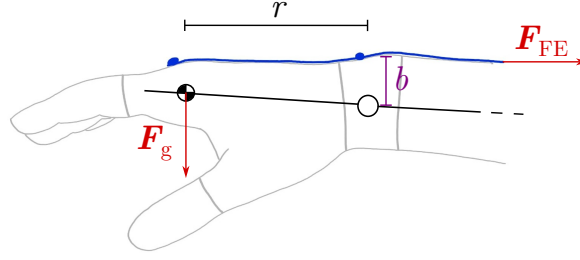


Figure 2.4: Stitched configuration.

The tendon (indicated in blue) is anchored to the dorsal side of the hand at two positions: the base of the fingers and the proximal region near the wrist. Mechanically, the proximal stitch serves as the primary load bearing anchor. The tendon force $\mathbf{F}_{FE}$ is entirely transmitted to this first encountered anchor point, which theoretically acts as a clamp preventing further force transmission along the tendon. The distal anchor near the fingers primarily ensures that the hand is pulled uniformly.

Because the force is transmitted exclusively to the proximal anchor, the resulting moment arm corresponds strictly to the distance b , which approximates half the thickness of the wrist. By establishing b and imposing the required exosuit moment $\mathbf{M}_{\text{exo, FE}} = \mathbf{M}_{\text{comp, FE}}$, the necessary tendon force $\mathbf{F}_{FE}$ is computed as:

$$\mathbf{F}_{FE} = \frac{\mathbf{M}_{\text{exo, FE}}}{b} = \frac{\mathbf{M}_{\text{comp, FE}}}{b} \quad (2.6)$$

Sliding configuration The primary limitation of the stitched configuration is the inherently small moment arm b , which remains constant even though a larger vertical distance becomes available when the flexion/extension angle deviates from 0° .

To address this, an alternative strategy involves replacing the proximal rigid anchor with a sliding band. This allows the tendon tension to bypass the wrist and transmit directly to the distal anchor near the fingers. The sliding band confines the tendon close to the anatomy without restricting its axial movement or adding significant volume.

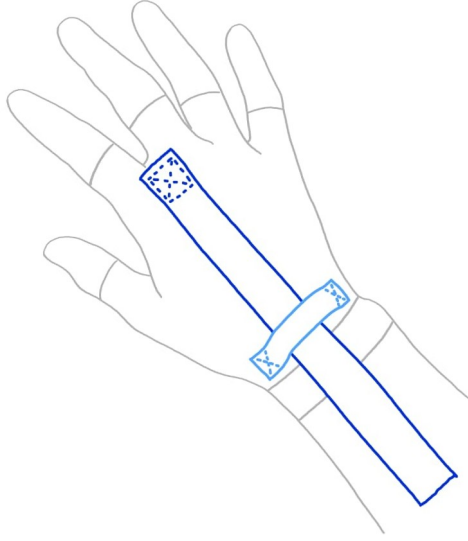


Figure 2.5: Sliding configuration.

However, analyzing the free body diagram in Figure 2.6 reveals no mechanical advantage over the stitched configuration. The increased moment arm of the horizontal force component is perfectly counterbalanced by the opposing moment generated by the vertical force component.

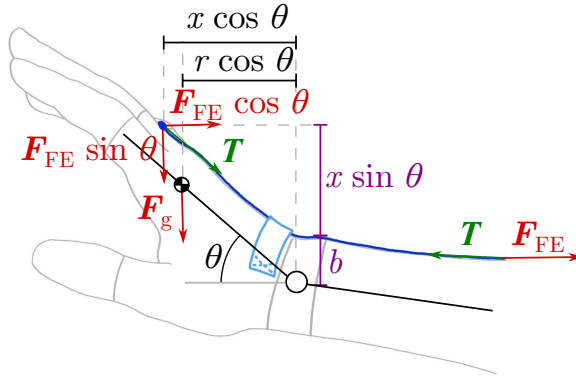


Figure 2.6: Forces in sliding configuration.

The actuator force $\mathbf{F}_{\text{FE}}$ is transmitted to the distal anchor via the tendon tension $\mathbf{T}$. Decomposing this force into its horizontal and vertical components, and calculating the resultant moment around the wrist joint $\mathbf{M}_{\text{exo, FE}}$ (using a positive clockwise convention), yields:

$$\begin{aligned}
 \mathbf{M}_{\text{exo, FE}} &= \mathbf{F}_{\text{FE}} \cos \theta (x \sin \theta + b) - \mathbf{F}_{\text{FE}} \sin \theta x \cos \theta \\
 &= \mathbf{F}_{\text{FE}} (x \cos \theta \sin \theta + b \cos \theta - x \sin \theta \cos \theta) \\
 &= \mathbf{F}_{\text{FE}} b \cos \theta \rightarrow \mathbf{F}_{\text{FE}} = \frac{\mathbf{M}_{\text{exo, FE}}}{b \cos \theta} = \frac{\mathbf{M}_{\text{comp, FE}}}{b \cos \theta}
 \end{aligned} \tag{2.7}$$

Evaluating this expression at $\theta = 0^\circ$ (the worst case scenario analyzed previously) reduces the required force to $\mathbf{F}_{FE} = \frac{M_{\text{comp, FE}}}{b}$, which is identical to the stitched configuration result.

This exact equivalence holds strictly when the sliding band aligns perfectly with the wrist joint. If an anatomical offset exists, such that the horizontal moment arm becomes $x \cos \theta + c$, where c is the horizontal distance between the band and the joint, the terms do not perfectly cancel. Nevertheless, the net mechanical difference between the sliding and stitched configurations remains negligible, necessitating a different structural approach.

Stitched bump configuration To effectively increase the moment arm and thereby reduce the actuation force required to overcome gravity, inspiration is drawn from human anatomy. Bony prominences known as *sesamoid bones* are positioned near specific joints to act as anatomical pulleys. These structures improve mechanical advantage by displacing tendons away from the center of rotation. A primary example is the patella in the human knee, illustrated in Figure 2.7.

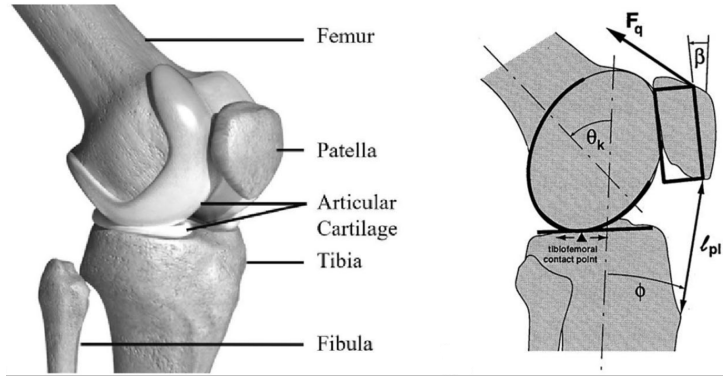


Figure 2.7: Biomechanics of the knee and mechanical advantage of the patella from [75].

Applying this biomimetic principle, a third configuration introduces a protrusion on the dorsal side of the glove. The tendon is anchored to the apex of this structure rather than resting flat against the hand, as shown in Figure 2.8.

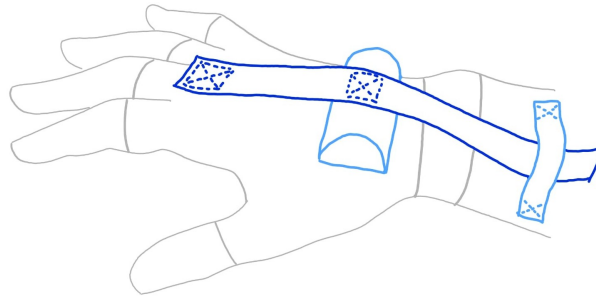


Figure 2.8: Stitched bump configuration.

The geometric shape and optimal positioning of this protrusion require careful evaluation. A more detailed analysis regarding the material and manufacturing of the structure is provided subsequently in Section 2.2.3.

Taking inspiration from the smooth contours of biological joints, the most intuitive profile is a semi elliptical shape lacking sharp edges, depicted in Figure 2.9.

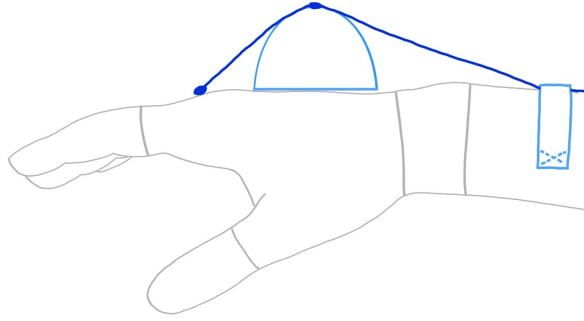


Figure 2.9: Semi elliptical protrusion in stitched bump configuration.

The corresponding free body diagram is represented in Figure 2.10.

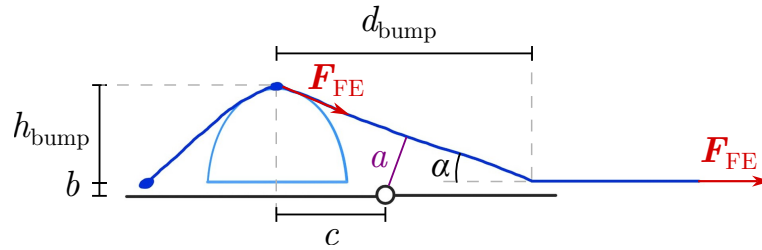


Figure 2.10: Forces in stitched bump configuration with semi elliptical protrusion: the hand (on the left) and forearm (on the right) are schematized as beams connected by a hinge, that represents the wrist joint.

In this schematic, b represents half the wrist thickness, h_{bump} is the height of the protrusion, and c is the horizontal distance from the tendon anchor point to the wrist joint axis. Additionally, d_{bump} represents the horizontal span between the anchor point and the proximal sliding band, $\alpha = \arctan\left(\frac{h_{\text{bump}}}{d_{\text{bump}}}\right)$ defines the angle of the tendon relative to the forearm, and a denotes the resultant moment arm of the actuation force $\mathbf{F}_{\text{FE}}$.

The exosuit moment is calculated by decomposing $\mathbf{F}_{\text{FE}}$ into its orthogonal components:

$$\begin{aligned}
 M_{\text{exo, FE}} &= \mathbf{F}_{\text{FE}} \cos \alpha (b + h_{\text{bump}}) - \mathbf{F}_{\text{FE}} \sin \alpha c \\
 &= \mathbf{F}_{\text{FE}} ((b + h_{\text{bump}}) \cos \alpha - c \sin \alpha) \\
 &= \mathbf{F}_{\text{FE}} a \rightarrow \mathbf{F}_{\text{FE}} = \frac{M_{\text{exo, FE}}}{a} = \frac{M_{\text{comp, FE}}}{a}
 \end{aligned} \tag{2.8}$$

where

$$a = ((b + h_{\text{bump}}) \cos \alpha - c \sin \alpha) \tag{2.9}$$

To determine the optimal position of the protrusion, a triangular profile is analyzed. This shape more clearly highlights the mechanical differences between vastly different anchor locations compared to a smooth curve. Specifically, the two opposing configurations depicted in Figure 2.11 are evaluated.

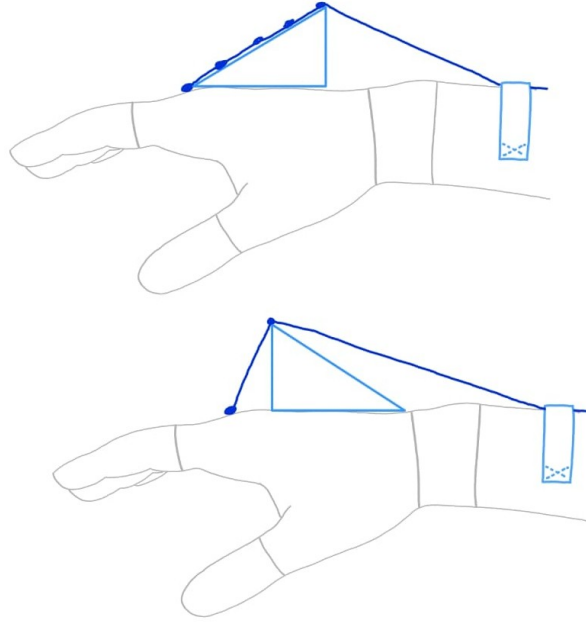


Figure 2.11: Two possible configurations with triangular bumps to study the optimal positioning of the protrusion in the stitched bump configuration.

Two distinct geometric properties emerge: the first configuration minimizes the horizontal offset c , while the second configuration minimizes the tendon angle α . Furthermore, the geometry of the first design provides a continuous sloped surface along its hypotenuse, potentially allowing the tendon to be secured along the entire face for a more uniform distribution of pulling forces.

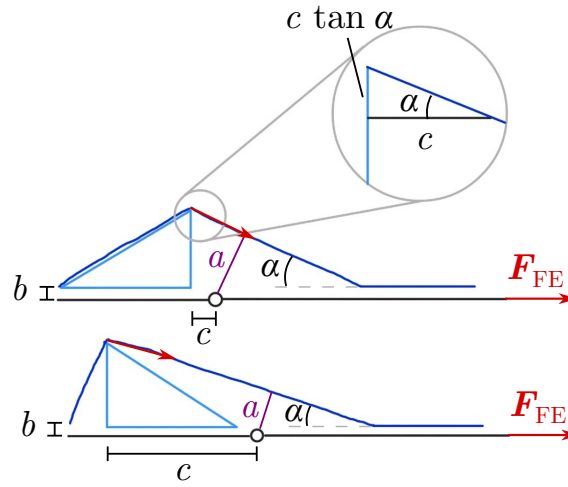


Figure 2.12: Forces in stitched bump configuration with triangular bumps.

Referring back to Equation 2.9, minimizing c yields a significantly larger increase in the moment arm a compared to minimizing α . Therefore, the first configuration is mechanically superior, establishing a critical design principle: the protrusion must be positioned as close to the wrist joint as possible.

This placement strategy is directly applied in the final prototype to maximize mechanical advantage. However, rather than utilizing a sharp triangular wedge, the physical device employs a semi elliptical profile. A rounded structure successfully maintains the necessary proximity to the joint to minimize c , while offering significantly improved ergonomics, user comfort, and a lower overall spatial profile.

Overall geometry

The final geometry for flexion/extension comprises a glove featuring a semi elliptical protrusion secured to its dorsal side, coupled with a forearm sleeve housing the actuator. A tendon anchored to both the protrusion and the servo motor establishes the mechanical transmission between the hand and the forearm. This tendon is exclusively responsible for driving wrist extension, while wrist flexion is achieved passively through gravitational forces.

A comprehensive description of the physical components comprising the final prototype is provided in Section 2.2.3.

2.2.2. Pronation/supination

Introduction and possible configurations

The second degree of freedom evaluated is forearm pronation and supination. A review of the state of the art reveals that soft exosuits addressing wrist pronation/supination are scarce. Existing devices employ various actuation strategies to achieve this movement, which are categorized into four main approaches. Several designs utilize actuators wrapping around the forearm [21, 31, 40]. Alternatively, Bartlett et al. [5] implemented two pairs of crossed actuators on the dorsal and palmar sides of the forearm, capable of achieving all three wrist degrees of freedom depending on the activated pair. Other approaches include pneumatic actuators that force rotation through twisting strain limiting layers [57], and cable driven systems utilizing a rigid wrist ring that is pulled to induce rotation [2, 68].

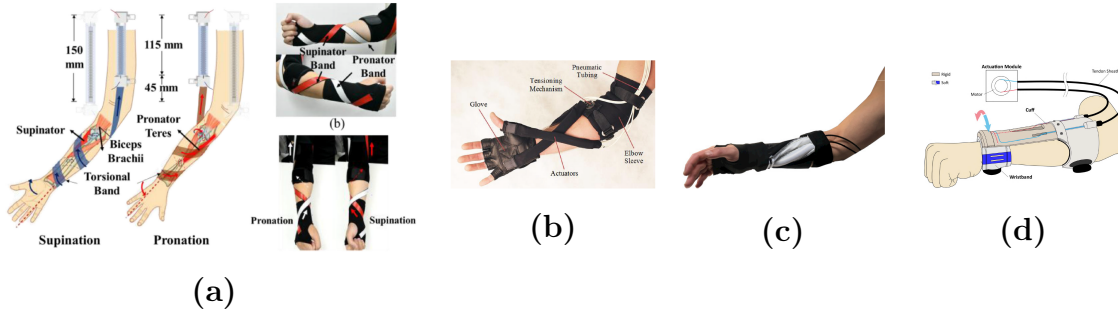


Figure 2.13: Different strategies to obtain pronation/supination. **(a)** Wrapping actuators from [31] **(b)**. Cross actuators configuration from [5]. **(c)** Twisting actuators with strain limiting layers from [57]. **(d)** Rotating wrist ring from [68].

Given the volume and routing of the tendons utilized for flexion/extension described previously, avoiding a crossed architecture similar to [5] is preferable, as it would place additional tendons almost parallel to the existing ones. This configuration would prove even more problematic for the electrohydraulic prototype discussed subsequently as it is already bulkier. Furthermore, incorporating strain limiting layers is not technically feasible with tendon driven actuation, rendering the approach of [57] unsuitable. Finally, the rotating wrist ring geometry introduces a rigid element. While rigid components are not strictly prohibited in soft wearable systems, they are suboptimal when fully compliant alternatives exist. Moreover, as discussed in Chapter 1, this solution generates rotation via a localized pulling action at the wrist, which contradicts the natural biomechanics of pronation/supination that involve a distributed rotation along the entire forearm. Therefore, the wrapping actuators architecture represents the most biomimetic and structurally appropriate option.

This geometry is achieved by anchoring one tendon to the palmar side and another to the dorsal side of the glove, wrapping them around the forearm in counter rotating directions to independently drive pronation and supination. However, this configuration presents specific limitations. Following the rationale established for flexion/extension, an active bidirectional system requires two separate motors. This addition increases the collocated weight and volume on the forearm sleeve, directly reducing user comfort. Additionally, physical contact between the two counter rotating tendons, as well as potential interference with the extension tendon, increases overall mechanical friction, thereby reducing force transmission efficiency and control precision. Finally, a primary objective of this thesis is to maintain a unified mechanical architecture across both actuation technologies to allow for a direct comparison. As anticipated in Chapter 1, HASEL actuators currently exhibit limitations regarding output force and strain. Consequently, as detailed in Section 2.2.4, the electrohydraulic design requires multiple actuators arranged in parallel. Accommodating multiple parallel actuators for both pronation and supination in a counter twisted configuration, alongside the extension actuators, creates an excessively bulky and practically unfeasible system.

To resolve these constraints, the proposed design actively drives only supination, leaving pronation to the natural elastic recoil of passive musculoskeletal tissues. The rationale behind prioritizing supination stems from the definition of the resting posture. Biomechanically, the neutral position of the forearm is vertical, with the thumb pointing upwards (as shown in Figure 1). However, as established in the system requirements, this device assumes a horizontal, palm down posture as the default resting state. From this fully pronated baseline, the primary active motion required to initiate and perform most typical daily tasks, such as turning the palm upward to hold an object or bringing a hand to the mouth, is supination. Consequently, pronation primarily serves as the return phase to bring the forearm back to its relaxed, palm down resting state. This return movement can be effectively achieved through passive tissue elasticity rather than requiring active motor assistance.

It is worth noting that the choice of actuating either supination or pronation could ultimately be tailored to the specific target user in future clinical applications. For instance, higher cervical lesions (at the C3-C4 level) impact the musculocutaneous nerve controlling the biceps and supinator muscles, resulting in an inability to perform active supination. Conversely, lower cervical lesions (at the C6-C7 level) primarily affect the median nerve, impairing the pronator muscles [21]. Therefore, while the current prototype exclusively assists supination, the underlying wrapping actuator architecture could easily be mirrored in future devices to provide a custom solution suited to the specific neurological deficit of

the end user.

To minimize bulk while ensuring the tendons maintain direct contact with the forearm (a requirement for distributing the pulling action along the entire forearm rather than localizing it at the wrist) the supination tendon is routed underneath the extension tendon.

Anatomy of the movement

To define the optimal routing of the tendons and their attachment points on the glove, an understanding of the biomechanics of forearm pronation and supination is required.

As described by Soubeyrand et al. [67], the rotation of the forearm can occur in two distinct configurations: immobile ulna and mobile ulna. In the immobile ulna configuration, the ulna remains stationary and serves as the central axis around which the radius rotates. In the mobile ulna configuration, both the ulna and the radius actively contribute to the overall rotation. Natural physiological rotations of the forearm typically involve a combination of both behaviors.

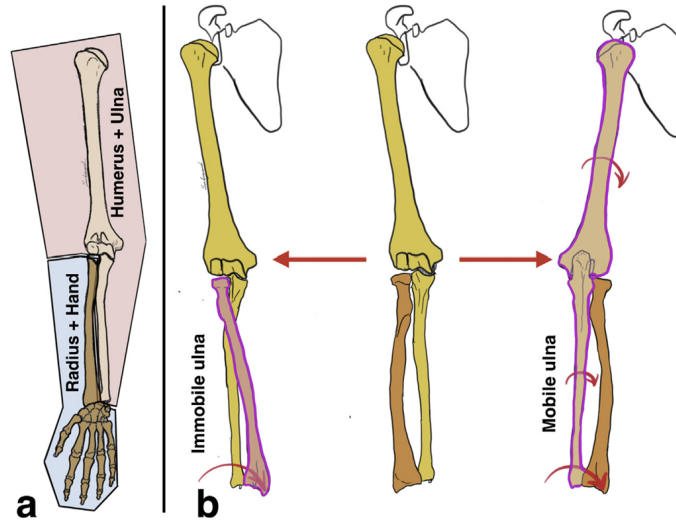


Figure 2.14: Rotation of the forearm with immobile and mobile ulna from [67].

Although the ulnar side might intuitively feel stronger during rotation, as evidenced by common gym exercises where cables or resistance bands are pulled from the ulnar (little finger) side of the palm to train supination, the actual physiological movement is driven by the displacement of the radius. Consequently, when applying an external mechanical torque via an exosuit, anchoring the actuation to the radial (thumb) side exploits a substantially larger moment arm and greater spatial displacement compared to pulling from the ulnar (little finger) side.

Furthermore, the true anatomical axis of rotation is slightly inclined relative to the longitudinal axis of the forearm, as illustrated in Figure 2.15.

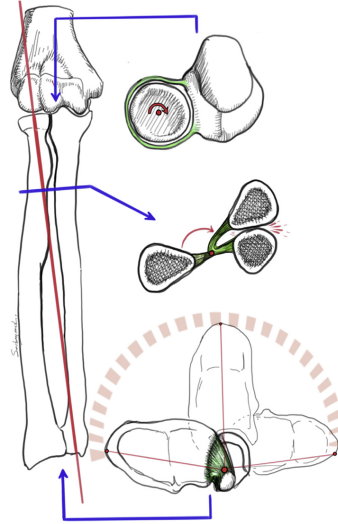


Figure 2.15: Forearm's rotational axis from [67].

These anatomical considerations are critical for identifying the regions of the forearm that undergo the largest displacement, as well as those that remain stationary (fixed points) during pronation and supination. High displacement regions provide optimal anchor points for the force transmitting tendons, whereas the fixed points are ideal for mounting stationary components such as motors and inertial sensors. The practical application of these anatomical principles to the physical prototype is detailed in Section 2.2.3.

However, for the sake of simplicity in the subsequent mechanical analysis, the rotation axis is assumed to be perfectly longitudinal and aligned with the geometric center of the forearm. This assumption represents a reasonable simplification for developing a tractable analytical model, as the discrepancy between this idealized axis and the true anatomical axis is expected to have a negligible effect on the resulting torque and force computations.

Design strategy

Although it has been established that pulling from the radial (thumb) side offers a greater mechanical advantage than pulling from the ulnar (little finger) side, a dual point actuation strategy pulling from both sides simultaneously is highly viable. In the case of the tendon driven prototype, to utilize a single motor, these two tendons must converge into a single main tendon routing up the forearm. Conversely, in the theoretical HASEL

driven exosuit, the two actuators remain separated to operate in parallel without physical interference, a configuration detailed further in Section 2.2.4.

The underlying kinematics of the wrapping tendon approach can be analyzed by initially considering only the tendon anchored to the dorsal side near the thumb.

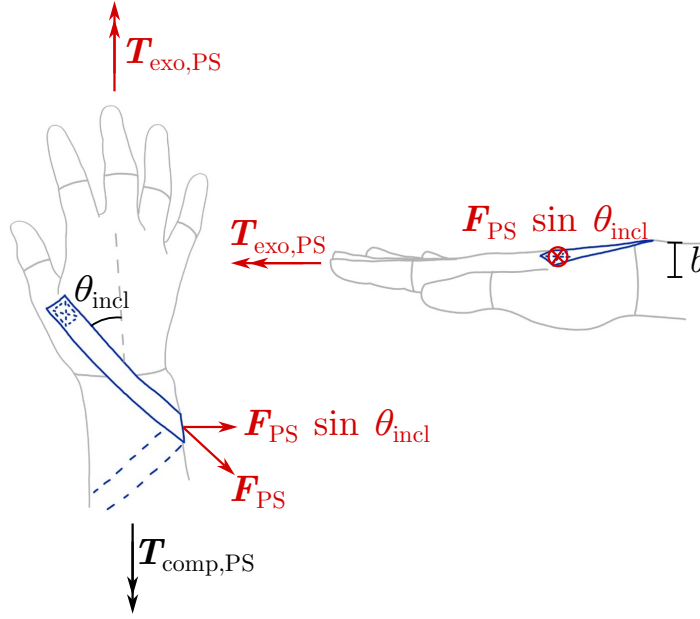


Figure 2.16: Forces and torques in simple wrapping configuration.

Without additional geometric elements, the moment arm of the torque exerted by the exosuit is limited to b , representing approximately half the thickness of the hand (approximated as the wrist thickness for simplicity), which provides an extremely low mechanical advantage. Similar to the strategy employed for flexion/extension, a surface protrusion must be introduced to increase the lever arm. Based on the previous kinematic analysis, the protrusion should ideally be positioned as close to the tendon anchor point (near the thumb) as possible. However, because the base of the thumb is a highly mobile anatomical region, a more practical solution utilizes the exact same dorsal protrusion already implemented for wrist extension. This shared component approach reduces overall system bulk and restricts rigid elements to areas of lower hand mobility.

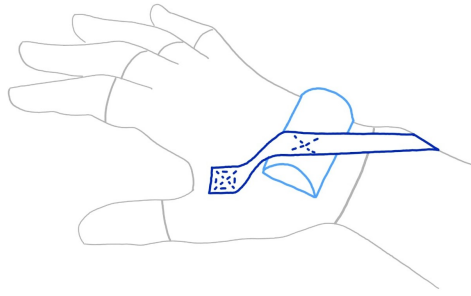


Figure 2.17: Tendon string from the dorsal side of the hand with relative protrusion.

This identical biomechanical reasoning applies to the palmar side, necessitating a second protrusion to increase the lever arm for the tendon originating near the little finger. Because the palm of the hand is a highly mobile functional surface essential for grasping, this palmar protrusion must feature a significantly smaller profile, as illustrated in Figure 2.18.

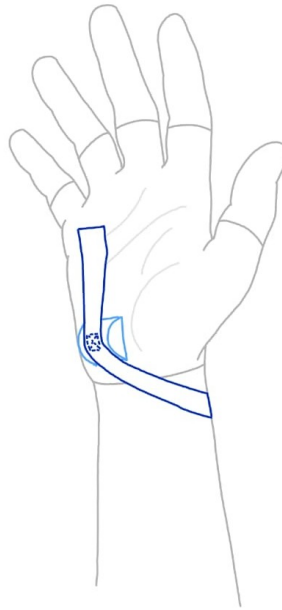


Figure 2.18: Tendon string from the palmar side of the hand with relative protrusion.

String from the thumb on the back of the hand

The analysis begins by evaluating the tendon originating from the radial (thumb) side. The corresponding actuation force is denoted as $\mathbf{F}_t$, where the subscript t refers to the thumb.

This force is transmitted as tension through the tendon strictly up to the anchor point on the dorsal protrusion. The stitch acts as a rigid mechanical constraint, preventing

further force transmission distally along the hand. The spatial configuration of this force is illustrated in Figure 2.19 and can be decomposed into three orthogonal components.

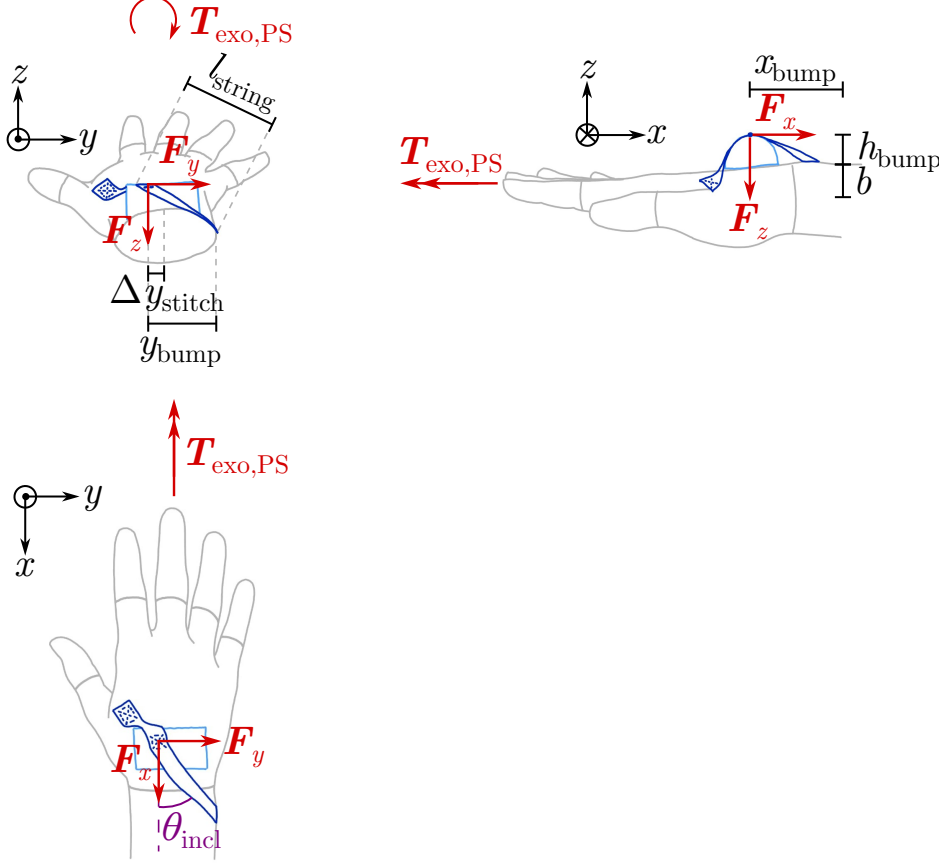


Figure 2.19: Forces in the tendon from the dorsal side of the hand.

Based on the defined geometry, the projected tendon length on the xy plane ($l_{\text{string, proj}}$), the true spatial tendon length (l_{string}), the elevation angle between the tendon and the forearm (α_{PS}), and the planar angle within the xy plane (θ_{incl}) are formulated as follows:

$$l_{\text{string, proj}} = \sqrt{x_{\text{bump}}^2 + y_{\text{bump}}^2} \quad (2.10)$$

$$l_{\text{string}} = \sqrt{x_{\text{bump}}^2 + y_{\text{bump}}^2 + h_{\text{bump}}^2} \quad (2.11)$$

$$\alpha_{\text{PS}} = \arccos\left(\frac{l_{\text{string, proj}}}{l_{\text{string}}}\right) \quad (2.12)$$

$$\theta_{\text{incl}} = \arctan\left(\frac{y_{\text{bump}}}{x_{\text{bump}}}\right) \quad (2.13)$$

It can be observed that the longitudinal component F_x acts parallel to the axis of rota-

tion and therefore does not contribute to the total exosuit torque $\mathbf{T}_{\text{exo}}$. The transverse component $\mathbf{F}_y$ provides the primary rotational contribution. The vertical component $\mathbf{F}_z$ contributes to the torque only if the anchor point is offset from the central longitudinal axis of the hand by a distance Δy_{stitch} (measured as positive in the negative y direction). Defining $\mathbf{T}_y$ and $\mathbf{T}_z$ as the torques generated by $\mathbf{F}_y$ and $\mathbf{F}_z$ respectively, the formulations are:

$$\mathbf{T}_y = \mathbf{F}_y(h_{\text{bump}} + b) = \mathbf{F}_t \cos \alpha_{\text{PS}} \sin \theta_{\text{incl}}(h_{\text{bump}} + b) \quad (2.14)$$

$$\mathbf{T}_z = -\mathbf{F}_z \Delta y_{\text{stitch}} = -\mathbf{F}_t \sin \alpha_{\text{PS}} \Delta y_{\text{stitch}} \quad (2.15)$$

The total torque contribution from the thumb side, $\mathbf{T}_{\text{exo, thumb}}$, is derived by summing these individual components:

$$\mathbf{T}_{\text{exo, thumb}} = \mathbf{T}_y + \mathbf{T}_z = \mathbf{F}_t(\cos \alpha_{\text{PS}} \sin \theta_{\text{incl}}(h_{\text{bump}} + b) - \sin \alpha_{\text{PS}} \Delta y_{\text{stitch}}) = \mathbf{F}_t a_t \quad (2.16)$$

where the effective moment arm a_t is defined as:

$$a_t = \cos \alpha_{\text{PS}} \sin \theta_{\text{incl}}(h_{\text{bump}} + b) - \sin \alpha_{\text{PS}} \Delta y_{\text{stitch}} \quad (2.17)$$

String from the little finger on the palm of the hand

The actuation geometry on the palmar side mirrors the architecture of the dorsal side. Variables associated with this region are denoted by the subscript *palm*², and the actuation force originating near the little finger is defined as $\mathbf{F}_l$.

Following the established methodology, the projected tendon length on the xy plane ($l_{\text{string, proj, palm}}$), the true spatial tendon length ($l_{\text{string, palm}}$), the elevation angle between the tendon and the forearm ($\alpha_{\text{PS, palm}}$), and the planar angle within the xy plane ($\theta_{\text{incl, palm}}$) are defined as:

$$l_{\text{string, proj, palm}} = \sqrt{x_{\text{palm}}^2 + y_{\text{palm}}^2} \quad (2.18)$$

$$l_{\text{string, palm}} = \sqrt{x_{\text{palm}}^2 + y_{\text{palm}}^2 + z_{\text{palm}}^2} \quad (2.19)$$

$$\alpha_{\text{PS, palm}} = \arccos \left(\frac{l_{\text{string, proj, palm}}}{l_{\text{string, palm}}} \right) \quad (2.20)$$

²While using the paired subscripts *dorsum* and *palm* might offer greater symmetry, the subscript *bump* is intentionally retained for the dorsal components to maintain strict mathematical consistency with the flexion/extension derivations presented earlier. Because both actuation mechanisms share the exact same physical protrusion, introducing a new subscript for the dorsal side in this section would create unnecessary confusion.

$$\theta_{\text{incl, palm}} = \arctan\left(\frac{y_{\text{palm}}}{x_{\text{palm}}}\right) \quad (2.21)$$

Applying the identical moment arm decomposition used for the dorsal protrusion, the transverse torque $\mathbf{T}_{y,\text{palm}}$ and the vertical torque $\mathbf{T}_{z,\text{palm}}$ are calculated as follows:

$$\mathbf{T}_{y,\text{palm}} = \mathbf{F}_{y,\text{palm}}(z_{\text{palm}} + b) = \mathbf{F}_l \cos \alpha_{\text{PS, palm}} \sin \theta_{\text{incl, palm}}(z_{\text{palm}} + b) \quad (2.22)$$

$$\mathbf{T}_{z,\text{palm}} = -\mathbf{F}_{z,\text{palm}} \Delta y_{\text{palm}} = -\mathbf{F}_l \sin \alpha_{\text{PS, palm}} \Delta y_{\text{palm}} \quad (2.23)$$

The total torque contribution from the ulnar side, $\mathbf{T}_{\text{exo, little finger}}$, is derived by summing these components:

$$\begin{aligned} \mathbf{T}_{\text{exo, little finger}} &= \mathbf{T}_{y,\text{palm}} + \mathbf{T}_{z,\text{palm}} \\ &= \mathbf{F}_l (\cos \alpha_{\text{PS,palm}} \sin \theta_{\text{incl,palm}}(z_{\text{palm}} + b) - \sin \alpha_{\text{PS,palm}} \Delta y_{\text{palm}}) \\ &= \mathbf{F}_l a_l \end{aligned} \quad (2.24)$$

where the effective moment arm a_l is defined as:

$$a_l = \cos \alpha_{\text{PS,palm}} \sin \theta_{\text{incl,palm}}(z_{\text{palm}} + b) - \sin \alpha_{\text{PS,palm}} \Delta y_{\text{palm}} \quad (2.25)$$

Forces and torques to overcome

Having determined the individual torque contributions from the radial and ulnar anchor points, the necessary tendon forces $\mathbf{F}_t$ and $\mathbf{F}_l$ are calculated. Similar to the methodology applied for flexion/extension, this requires equating the torque generated by the exosuit to the physiological resisting torque that must be compensated to achieve motion.

While gravitational forces dominate the resistance in wrist extension, gravity has a negligible impact on pronation and supination because the center of mass of the hand lies approximately on the longitudinal axis of forearm rotation. Consequently, the primary resistance to overcome is the internal passive stiffness of the antagonist muscles, tendons, and ligaments³.

Formica et al. [22] experimentally quantified the internal passive stiffness of the wrist

³As with the flexion/extension analysis, the target torque is calculated to provide a reasonable order of magnitude rather than an exact dynamic computation. Because gravitational forces dominate extension and passive joint stiffness dominates pronation/supination, focusing on these primary resistive factors establishes a solid mechanical baseline for the exosuit torque requirements.

joint per unit of rotation angle (in radians), as presented in Figure 2.20.

Movement	Mean Stiffness, Nm/rad		
	Males	Females	Overall
Toward flexion	0.605 ± 0.131	0.429 ± 0.192	0.554 ± 0.170
Toward extension	1.146 ± 0.327	0.717 ± 0.323	1.021 ± 0.379
Toward radial deviation	1.927 ± 0.521	1.205 ± 0.314	1.710 ± 0.573
Toward ulnar deviation	1.328 ± 0.468	1.035 ± 0.315	1.245 ± 0.448
Toward pronation	0.285 ± 0.120	0.135 ± 0.107	0.240 ± 0.135
Toward supination	0.217 ± 0.093	0.114 ± 0.083	0.186 ± 0.101

Values are mean ($\pm$ SD) stiffness estimates, obtained using the 1-dimensional (1-D) protocol, from all subjects (overall). Mean values for male and female subgroups are also shown.

Figure 2.20: Internal passive stiffness of the wrist per unit of angle from [22].

Consistent with the flexion/extension design criteria, the objective is to compensate for 100% of this passive resistance to fully support the passive operating mode.

The target passive torque depends on the maximum required rotation angle of the wrist $\beta_{\max}$. This angle corresponds to the required maximum linear displacement (excursion) of the supination tendon, denoted as $c_{\text{need},1}$. The ulnar (little finger) side is utilized for this geometric calculation because it features the smallest planar angle θ_{incl} , thereby maximizing the force component directed along the arc of rotation.

This analysis utilizes the *required* linear displacement rather than the *actual* obtained displacement to prevent a complex circular dependency. Specifically, the rotation angle dictates the required torque, which in turn influences the actual displacement, thereby feeding back into the achieved angle. Utilizing the theoretical required displacement bypasses the need for an iterative numerical solver while remaining a valid simplification, as the actual mechanical excursion is designed to match as closely as possible the required target.

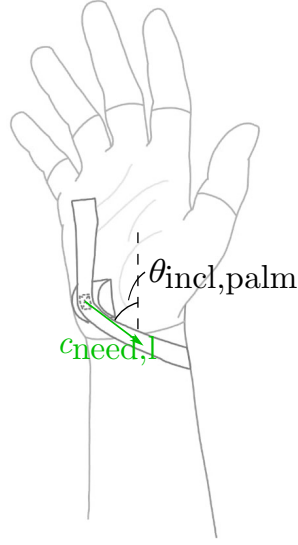


Figure 2.21: Needed linear displacement of the tendon from the little finger.

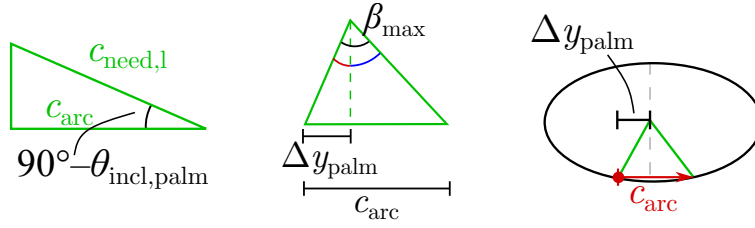


Figure 2.22: Geometric quantities involved in supination.

First, the horizontal component of the linear displacement $c_{\text{need},l}$ is calculated:

$$c_{\text{arc}} = c_{\text{need},l} \cos(90^\circ - \theta_{\text{incl,palm}}) \quad (2.26)$$

The corresponding maximum angle of wrist rotation is formulated as:

$$\beta_{\text{max}} = \beta_{\text{left}} + \beta_{\text{right}} = \arctan\left(\frac{\Delta y_{\text{palm}}}{b}\right) + \arctan\left(\frac{c_{\text{arc}} - \Delta y_{\text{palm}}}{b}\right) \quad (2.27)$$

Utilizing the passive supination stiffness per radian $k_{\text{per rad}}$ derived from Figure 2.20, along

with the maximum rotation angle $\beta_{\max}$, the target compensation torque is :

$$T_{\text{comp, PS}} = k_{\text{per rad}} \beta_{\max} \quad (2.28)$$

Imposing the equilibrium condition $\mathbf{T}_{\text{comp, PS}} = \mathbf{T}_{\text{exo}}$, the total torque is expressed as the sum of the radial and ulnar contributions:

$$\mathbf{T}_{\text{comp, PS}} = \mathbf{F}_t a_t + \mathbf{F}_l a_l \quad (2.29)$$

Solving for the two unknown forces $\mathbf{F}_t$ and $\mathbf{F}_l$ requires a secondary relationship. Defining an arbitrary force ratio coefficient k such that $\mathbf{F}_l = k \mathbf{F}_t$, the torque equation becomes $\mathbf{T}_{\text{comp, PS}} = \mathbf{F}_t(a_t + k a_l)$. The individual forces are therefore:

$$\mathbf{F}_t = \frac{\mathbf{T}_{\text{comp, PS}}}{a_t + k a_l} \quad (2.30)$$

$$\mathbf{F}_l = \frac{k \mathbf{T}_{\text{comp, PS}}}{a_t + k a_l} = k \mathbf{F}_t \quad (2.31)$$

A value of $k = 1$ is an appropriate assumption, given that the two tendon branches are mechanically coupled through the compliant structure of the glove, which acts as an elastic element that naturally distributes and balances the tensile loads.

Assuming symmetric force distribution, the tension required in each individual tendon branch is equivalent:

$$\mathbf{F}_{\text{PS}} = \frac{\mathbf{T}_{\text{comp, PS}}}{a_t + a_l} \quad (2.32)$$

Because the tendon driven prototype merges these two distal branches into a single proximal tendon connected to the servo motor, the total tensile force transmitted by the primary string is:

$$\mathbf{F}_{\text{PS, TOT}} = 2 \mathbf{F}_{\text{PS}} \quad (2.33)$$

³While spatial forces and torques ($\mathbf{F}$, $\mathbf{T}$) are generally denoted as vectors to represent their directional lines of action in 3D space, structural stiffness (k) and angular displacement (β) are inherently scalar properties. Consequently, their product defines the scalar magnitude of the required compensation torque, which is therefore not bolded in this formulation. This scalar notation is applied here and throughout the subsequent elastic band analysis in Section 2.2.2, where the mechanics are evaluated strictly along a single defined axis of motion. It is also applied in the tables including the numerical results of this analysis in Section 2.2.3.

Elastic band responsible for pronation

A theoretical kinematic model for an elastic band intended to assist pronation was developed and is presented below. During the preliminary physical design phase, empirical observations revealed that the natural elastic recoil of the user's passive musculoskeletal tissues was generally sufficient to return the wrist to a neutral posture from a supinated state. Consequently, the physical elastic band was omitted from the final prototype. Discarding this component prevents unnecessary friction between counter rotating elements, significantly reduces the mechanical complexity of the device, and facilitates a much more intuitive donning and doffing process for the user. However, this mathematical framework is retained within this study as it validates the mechanical feasibility of a bidirectional soft actuation strategy. Should future clinical applications or specific patient profiles require active pronation assistance, this model provides the necessary analytical foundation for immediate integration into the existing architecture.

Determining the optimal routing for the elastic band requires analyzing the relative displacement between anchor points. A circumferential routing situated entirely on the wrist would theoretically maximize the elongation Δl , as it aligns directly with the rotational path. However, this configuration lacks a stationary proximal anchor; the entire assembly would rotate simultaneously, resulting in a net elongation of zero ($\Delta l = 0$). Consequently, the proximal anchor must be located on the stationary forearm sleeve, while the distal extremity is secured to the rotating glove. Furthermore, the optimal proximal anchor point is located near the elbow, as it lies on the longitudinal axis of rotation and remains stationary during forearm pronation and supination.

The initial kinematic model assumes a longitudinal resting position for the elastic band, routed along the ulna from the elbow to the ulnar side of the hand (near the little finger).

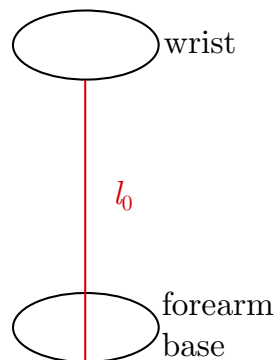


Figure 2.23: Resting position of the elastic band responsible for pronation.

To compute the elongation of the band during rotation, the kinematics can be modeled by analyzing a flexible band wrapped around a cylinder of radius r through a rotation angle β .

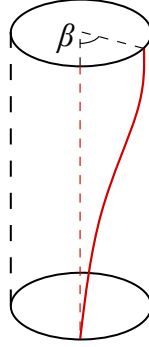


Figure 2.24: Band wrapped around a cylinder.

By virtually unrolling the cylindrical surface into a 2D plane, the instantaneous length of the band can be derived using the Pythagorean theorem.

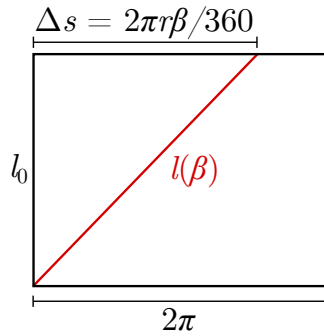


Figure 2.25: Unwrapped cylinder with band around it.

$$l(\beta) = \sqrt{l_0^2 + \Delta s^2} = \sqrt{l_0^2 + \left(\frac{2\pi r \beta}{360}\right)^2} \quad (2.34)$$

Although the anatomical cross section of the forearm is not perfectly cylindrical, approximating it as such is not required. The distal anchor point on the glove traces a perfect circular trajectory around the fixed longitudinal rotation axis of the forearm, regardless of the cross sectional profile of the wrist.

To maximize mechanical advantage, the distal anchor must be positioned laterally on the side of the hand. This placement ensures that the radius of rotation r corresponds to the

major semi axis of the wrist cross section, rather than the minor semi axis (which would occur if anchored centrally on the dorsal or palmar surfaces).

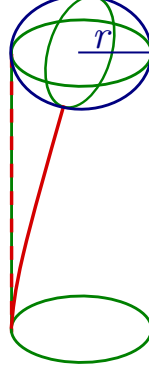


Figure 2.26: Radius of rotation of the elastic band extremity on the wrist.

The resting length l_0 of the elastic band is approximated as the standard forearm length defined previously in Figure 2.2:

$$l_0 = 0.146 \cdot H \quad (2.35)$$

Knowing the initial resting length l_0 and the extended length $l(\beta_{\max})$ at the maximum rotation angle, the total elongation of the elastic band is derived as:

$$\Delta l = l(\beta_{\max}) - l_0 \quad (2.36)$$

The final parameter required to define the elastic band is its stiffness, which depends on the necessary target force. The target pronation torque is calculated using the same methodology applied to supination. Using the passive pronation stiffness per radian ($k_{\text{pronation}}$) from Figure 2.20, the required torque magnitude is:

$$T_{\text{pronation}} = k_{\text{pronation}} \beta_{\max} \quad (2.37)$$

It is not strictly necessary for the elastic band to compensate for the entirety of this target torque. Because the forearm naturally tends to return to the neutral horizontal posture, the elastic band primarily serves to initiate the movement, after which passive tissue elasticity assists the remainder of the trajectory. Consequently, only a specified percentage ($\text{perc}_{\text{elastic}}$) of the total torque $T_{\text{pronation}}$ must be actively supplied.

From this reduced torque, the required tangential force $F_{\text{pronation}}$ and the corresponding tension required in the elastic band F_{elastic} can be determined (as illustrated in Figure

2.27).

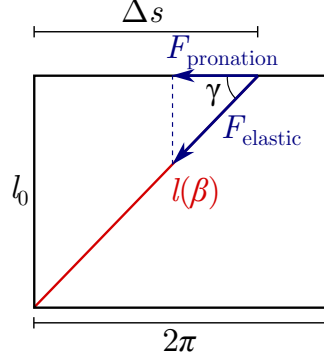


Figure 2.27: Force to obtain pronation and force given by the elastic band.

$$F_{\text{elastic}} = \frac{F_{\text{pronation}}}{\cos \gamma} = \frac{\text{perc}_{\text{elastic}} T_{\text{pronation}}}{r \cos \gamma} \quad (2.38)$$

Finally, the required stiffness coefficient of the elastic band is calculated as:

$$k_{\text{elastic}} = \frac{F_{\text{elastic}}}{\Delta l} \quad (2.39)$$

Overall geometry

In summary, the final architecture for pronation and supination uses two distal tendon branches. The first branch is anchored to the shared dorsal protrusion near the thumb, which is also used for wrist extension. The second branch is secured to a smaller dedicated protrusion on the palmar side near the base of the little finger. These two distal components converge into a single primary tendon that wraps around the forearm and connects to the proximally mounted servo motor. This specific tendon routing is exclusively responsible for actively driving supination. For the return phase of pronation, while the theoretical framework for an elastic band was established, the natural elastic recoil of the passive musculoskeletal tissues provides sufficient rotational force. Consequently, an active pronation mechanism or supplementary elastic band is omitted from the final prototype to prioritize mechanical simplicity and user comfort.

2.2.3. Prototype implementation

Reference user definition

To appropriately size the motor, numerical values must first be introduced into the previously derived symbolic computations.

To achieve this, it is first necessary to define the anthropometric characteristics of a reference user. Based on population level studies such as the NCD-RisC collaboration [49], average adult heights in Italy are approximately 175-178 cm for men and 162-165 cm for women. Accordingly, a representative height of $H = 1.70$ m is selected for this work. The reference body mass is then calculated using a Body Mass Index (BMI) of 24.2 kg/m^2 , which lies within the normal weight range defined by the World Health Organization [50]. This combination results in a reference body mass of $M = 70$ kg.

In wearable robotics, relying on an average reference subject is standard practice. While extreme values (such as the 90th or 95th percentile) are strictly required when designing for geometric clearance, wearable devices with adjustable interfaces are typically modeled around a representative subject [14]. Because this device is a prototype intended to be worn and evaluated by a general population, optimizing its dynamic design for the average user rather than the extremes is highly appropriate.

From a dynamic perspective, basing the model on a taller and heavier user would indeed inherently result in higher gravitational loads, making the initial design more conservative. However, the hand and forearm represent only a fraction of the total body mass. Consequently, scaling the overall body size upwards produces only moderate changes in the required wrist moments, rather than order of magnitude differences.

To ensure robustness against inter-subject variability and modeling uncertainties, conservative safety factors are applied during motor selection. Therefore, the design does not rely on worst case anthropometric values, but rather combines a representative reference subject with appropriate safety margins (as detailed in Section 2.2.3). Together, these factors ensure coverage for a sufficiently wide range of users while maintaining a consistent and physically based model.

Finally, this methodology was actively verified during the experimental phase. As reported in Section 2.4, the test subjects included in the study exhibit anthropometric characteristics directly consistent with this reference model.

Numerical design parameters

Having quantified the height H and mass M of the reference user, the mechanical and kinematic equations derived previously can be evaluated. The resulting numerical design

parameters are summarized in Tables 2.3, 2.4, and 2.5.

Table 2.3: Anthropometric parameters of the reference user.

Quantity	Symbol	Unit	Value
Body mass	M	kg	70
Body height	H	m	1.70
Hand mass	m_{hand}	kg	0.42
Hand length	P	m	0.184
COM proximal length	r	m	0.093
Wrist thickness	t	m	0.04
Half wrist thickness	b	m	0.02
Distance between stitch and wrist	x	m	0.08
Wrist width	a_{wrist}	m	0.05

Table 2.4: Flexion/extension design parameters.

Quantity	Symbol	Unit	Value
Forces and moments			
Gravity force	F_g	N	4.12
Maximum moment due to gravity	$M_{\text{max g, FE}}$	Nm	0.38
Percentage of gravity compensated	$comp$	%	100
Maximum moment to compensate	$M_{\text{comp, FE}}$	Nm	0.38
Stitched bump configuration			
Bump height	h_{bump}	m	0.02
Distance between bump and band	d_{bump}	m	0.1
Angle between string and arm	α	°	11.31
Distance between bump and wrist joint	c	m	0.01
Moment arm	a	m	0.037
Required force	F_{FE}	N	10.27

Sizing of the servo motor

A tendon driven transmission does not strictly mandate a specific type of electric actuator; alternative options such as stepper motors or open loop DC motors could theoretically be employed. However, the application of a wearable exosuit requires accurate and stable control of tendon tension and displacement under highly variable external disturbances, including human motion and changes in tendon routing geometry.

For this reason, a closed loop servo actuation architecture is adopted. Compared to open loop alternatives, servo motors provide integrated feedback control through encoders,

Table 2.5: Pronation/supination design parameters.

Quantity	Symbol	Unit	Value
Forces and torques to overcome			
Passive stiffness in supination	$k_{\text{per rad}}$	N/m	0.186
Needed linear displacement of the tendon ^a	$c_{\text{need, l}}$	m	0.02
Maximum arc of supination	c_{arc}	m	0.019
Maximum angle of supination	β_{max}	°	48.50
Maximum torque to compensate	$T_{\text{comp, PS}}$	Nm	0.16
Thumb/Dorsal side			
Distance stitch–bump along x	x_{bump}	m	0.020
Distance stitch–bump along y	y_{bump}	m	0.040
Distance stitch–hand centre along y	Δy_{stitch}	m	0.005
Projected string length	$l_{\text{string, proj}}$	m	0.045
String length	l_{string}	m	0.049
String inclination in plane xy	θ_{incl}	°	63.4
Angle between string and arm	α_{PS}	°	24.1
Moment arm	a_t	m	0.031
Force ratio coefficient	k	–	1
Force from thumb side	F_t	N	2.82
Little finger/Palmar side			
Distance stitch–bump along x	x_{palm}	m	0.015
Distance stitch–bump along y	y_{palm}	m	0.040
Bump height	z_{palm}	m	0.010
Distance stitch–hand centre along y	Δy_{palm}	m	0.005
Projected string length	$l_{\text{string, proj, palm}}$	m	0.043
String length	$l_{\text{string, palm}}$	m	0.044
String inclination in plane xy	$\theta_{\text{incl, palm}}$	°	69.4
Angle between string and arm	$\alpha_{\text{PS, palm}}$	°	13.2
Moment arm	a_l	m	0.026
Little finger force	F_l	N	2.82
Overall result			
Total required force	$F_{\text{PS, TOT}}$	N	5.63

^a The required linear displacement of the tendon was established experimentally to provide the Range of Motion (RoM) necessary for Activities of Daily Living (ADLs). This displacement value applies to both flexion/extension and pronation/supination, and serves as a critical baseline for the actuation feasibility analysis in Section 2.2.4.

enabling precise regulation of position, velocity, and, indirectly, tendon force. This is particularly relevant in tendon driven systems, where small position errors at the motor can translate into significant variations in tendon tension due to the transmission ratio and intrinsic structural compliance. Furthermore, servo based architectures enable continuous error correction, which mitigates the effects of nonlinearities such as friction, backlash, and tendon elasticity by dynamically adjusting the motor command based on measured output. These characteristics make servo motors particularly suitable for safe human-robot interaction, where predictable and repeatable force delivery is a critical safety requirement.

Having defined the required tendon tension forces for both flexion/extension and pronation/supination, the appropriate servo motors can be sized. Given the required linear displacement $\Delta L = c_{\text{need}}$ and the required servo rotation θ_{servo} (expressed in radians), the minimum pulley radius required is:

$$r_{\text{servo}} = \frac{\Delta L}{\theta_{\text{servo}}} \quad (2.40)$$

To optimize manufacturing and maintenance, an identical servo motor model is selected for both the flexion/extension and pronation/supination actuation units. This modular approach allows for bulk purchase of the motors and straightforward component interchangeability. Consequently, the chosen motor must be sized to accommodate the maximum expected load. Comparing $\mathbf{F}_{\text{FE}}$ and $\mathbf{F}_{\text{PS, TOT}}$ from Tables 2.4 and 2.5, the required force for flexion/extension ($\mathbf{F}_{\text{FE}}$) represents the governing maximum load case.

The theoretical maximum torque required from the motor is therefore:

$$\tau_{\text{max}} = \mathbf{F}_{\text{FE}} r_{\text{servo}} \quad (2.41)$$

However, to prevent operating the servo at its absolute mechanical limit, account for unmodeled dynamic losses, and accommodate users larger or heavier than the reference profile, a safety factor (SF) is introduced. The required target design torque thus becomes:

$$\tau_{\text{req}} = SF \tau_{\text{max}} \quad (2.42)$$

The numerical results of the sizing calculations are reported in Table 2.6.

Table 2.6: Sizing of the servo motor.

Quantity	Symbol	Unit	Value
Required linear displacement	ΔL	m	0.02
Servo rotation	θ_{servo}	°	180
Minimum pulley radius	r_{servo}	m	0.0064
Safety factor	SF	–	3
Required torque ^a	τ_{req}	kg·cm	2.00

^a The unit of measurement for the torque required by the servo motor is listed as kg·cm instead of N·m, as this is the standard commercial unit used by manufacturers for servo motor specifications.

Hardware components

The hardware architecture of the exosuit is composed of four main subsystems: actuation, sensing, control electronics, and the wearable mechanical structure. Additional materials and fabrication tools utilized for assembly are also detailed for completeness.

Actuation A practical and widely adopted method for classifying servo motors is by their rated output torque. In hobby and educational contexts, this is frequently expressed in kg·cm, whereas professional and industrial systems utilize N·m. Based on this criterion, commercially available servo motors generally fall into the following categories: micro and mini ($\sim 1 \div 5$ kg·cm), standard ($\sim 5 \div 15$ kg·cm), high torque ($\sim 15 \div 60+$ kg·cm), and industrial (a fundamentally different class of actuators with torques strictly expressed in N·m).

For the prototype developed in this thesis, standard servo motors were selected. While certain micro or mini servos with nominal torques around 2 kg·cm theoretically satisfy the calculated requirements, the continuous operating torque is generally assumed to be half of the stall limit; therefore, motors with a higher torque rating are required for reliable operation. However, excessively large motors were avoided for two primary reasons. First, a larger servo inherently introduces increased mass and volume, which degrades the ergonomics of a wearable device. Second, selecting a servo with a continuous torque rating moderately close to the calculated requirement provides an excellent physical validation of the theoretical design model. This secondary point is particularly relevant for the subsequent feasibility study regarding electrohydraulic actuators, as both systems share identical kinematic routing and torque requirements. The specific actuator selected was the Carson Servo CS-6 Waterproof MG (500502041), shown in Figure 2.28.



Figure 2.28: Servo CS-6 Waterproof MG (500502041) from Carson.

Sensing The orientation of the hand relative to the forearm was measured using two BNO055 Inertial Measurement Units (IMUs). The BNO055 is a compact nine degrees of freedom sensor integrating a three axis accelerometer, a three axis gyroscope, and a three axis magnetometer within a single package. A key feature of this specific sensor is an onboard microcontroller that performs internal sensor fusion, outputting processed orientation estimates directly in the form of Euler angles or quaternions. The sensor supports multiple communication interfaces, including I2C and UART, and offers configurable operating modes ranging from raw data acquisition to fully fused orientation outputs. One IMU used was manufactured by Adafruit (Figure 2.29), and the other IMU was manufactured by Bosch (Figure 2.30).

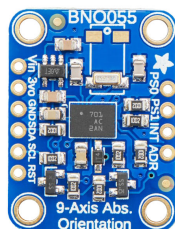


Figure 2.29: IMU BNO055 from Adafruit.



Figure 2.30: IMU BNO055 on custom breakout board.

The image in Figure 2.30 is provided strictly for reference; the actual sensor utilized on

the forearm was housed within a sealed, custom 3D printed enclosure salvaged from a previous laboratory project [44], as shown in Figure 2.31.



Figure 2.31: 3D printed casing containing the IMU BNO055 from Bosch.

Control electronics The electronic architecture of the prototype is governed by an ESP32 microcontroller (specifically, the ESP32-S3-DEV-KIT-N8R8, shown in Figure 2.32), which is programmed using the Arduino framework and acts as the central processing unit. The ESP32 is responsible for acquiring data from the IMUs, processing the sensor measurements, and generating the Pulse Width Modulation (PWM) control signals required to actuate the servo motors. This specific development board was selected due to its compact dimensions, low cost, sufficient computational capabilities, and integrated wireless communication features, which heavily facilitate debugging and data monitoring during development.

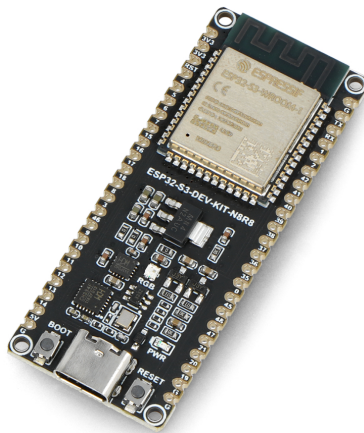


Figure 2.32: ESP32-S3-DEV-KIT-N8R8.

To supply power to the actuators, a USB-C breakout board was integrated into the circuit, as shown in Figure 2.33. This board provides a robust interface to connect the servo motors directly to a dedicated external power source (such as a power bank or wall adapter). This power separation is a critical hardware safety measure. Servo motors draw substantial current, particularly under mechanical load or during stall conditions. Attempting to power multiple servo motors directly through the ESP32 while it is connected to a computer's USB port for programming or serial monitoring risks heavily exceeding the port's maximum current output limit. This overcurrent condition can trigger forced shutdowns or cause permanent hardware damage to the computer's motherboard. Additionally, utilizing this dedicated power distribution board isolates the high current motor delivery from the logic circuitry of the ESP32, significantly reducing the risk of sudden voltage drops and electrical noise that could otherwise cause the microcontroller to reset or behave erratically.

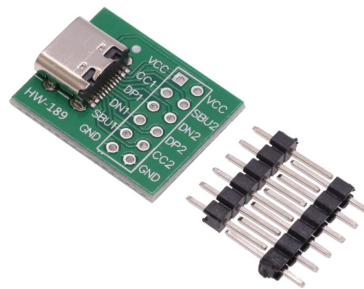


Figure 2.33: Breakout board USB C.

Wearable mechanical structure The wearable garment consists of a fingerless glove (shown in Figure 2.34) and a forearm sleeve (shown in Figure 2.35), which provide the foundation for integrating the sensing, actuation, and tendon routing hardware.



Figure 2.34: Fingerless glove.



Figure 2.35: Forearm sleeve.

The two moment arm protrusions discussed in the model are anchored to the dorsal (Figures 2.36 and 2.37) and palmar (Figures 2.38 and 2.39) sides of the glove. During the design phase, two candidate materials were evaluated for these components: molded silicone and 3D printed Thermoplastic Polyurethane (TPU). Silicone offers excellent compliance and user comfort; however, it possesses minimal shear tear resistance, meaning it cannot be reliably sewn (the thread would slice through the material) or glued to the textile. Conversely, TPU exhibits comparable density but greater structural rigidity. This rigidity allows the protrusions to be designed as hollow shell structures, significantly reducing overall component mass. Furthermore, TPU is rapid to manufacture, can be easily glued, and allows for direct integration with textiles. The primary disadvantage of a rigid TPU component is that a single large block could restrict the natural transverse arching of the hand.

To mitigate this, the dorsal protrusion was printed as two independent, adjacent blocks⁴. To securely attach these blocks to the glove, a layer of porous gauze fabric was manually laid over the print bed after the first three layers of TPU were deposited, and the print was subsequently resumed over it. This technique successfully embedded a sewable fabric flange directly into the base of the TPU blocks.

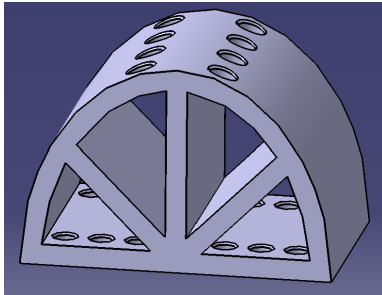


Figure 2.36: CAD model of the single block of the protrusion on the dorsal side of the hand. The hollow design optimizes the weight-to-volume ratio, and the internal supports provide necessary structural reinforcement against actuation forces.

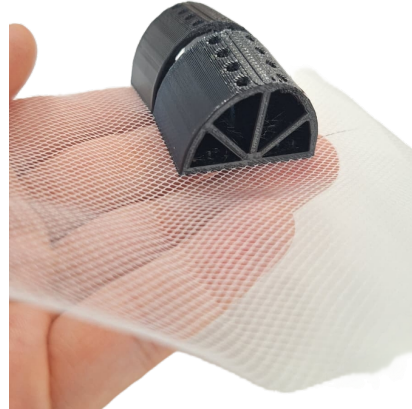


Figure 2.37: Dorsal protrusion composed of two discrete blocks, 3D printed directly onto a porous gauze fabric base to allow for textile integration.

⁴As detailed in Section 2.2.3, the flexion/extension mechanism utilizes two parallel strings to ensure uniform force distribution. Additionally, the subsequent feasibility study (Section 2.2.4) demonstrates that the HASEL prototype requires two parallel actuators. Utilizing two discrete blocks perfectly accommodates these parallel actuation geometries while maintaining transverse hand flexibility.

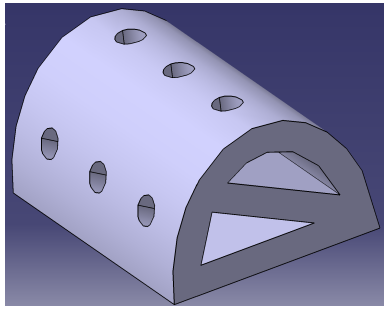


Figure 2.38: CAD model of the palmar protrusion. A single diagonal internal support is employed due to reduced space, specifically aligned to withstand the applied tendon tension.



Figure 2.39: Palmar protrusion 3D printed directly onto a porous gauze fabric base to allow for textile integration.

Mechanical force transmission is achieved through inelastic nylon tendons routed from the motors, along the sleeve, to the distal anchor points on the glove (Figure 2.40).



Figure 2.40: Nylon tendon of 2 cm width.

Tendon tensioning and length adjustment are facilitated by plastic buckles equipped with integrated string adjusters (Figure 2.41) and custom 3D printed linear adjusters (Figure 2.42). These mechanisms allow the baseline tendon length to be calibrated to the specific forearm dimensions of the user and the desired initial posture.



Figure 2.41: Plastic buckle with integrated string adjuster.

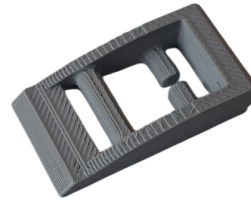


Figure 2.42: 3D printed string adjuster.

To ensure the servo motors and the proximal IMU remained firmly anchored to the sleeve under load, the standard sewn thread bindings were structurally reinforced with custom crocheted guides. Utilizing a rigid nylon thread and a slipstitch technique (the tightest and least elastic crochet stitch) resulted in a highly stable, form fitting harness that prevented actuator shifting during operation. Finally, all electrical connections between the sensors, actuators, and control electronics were established using flexible insulated jumper wires (Figure 2.43) routed along the dorsal surface of the sleeve.



Figure 2.43: Jumper wires.

Assembly materials and tools The prototype was assembled using conventional textile fabrication materials and tools. Standard orange sewing thread was employed for stitching components to the garments, while a heavy gauge black nylon thread was utilized for the crocheted structural elements securing the hardware. A 3 mm crochet hook was utilized to produce these compact fastening structures. Additional standard assembly tools included needles, temporary positioning pins, and fabric scissors.



Figure 2.44: Assembly materials and tools, from left to right: standard orange sewing thread, heavy gauge black nylon thread, 3 mm crochet hook, needle, pins and safety pins, fabric scissors.

Design and functioning

With the mechanical design formulated and the hardware components defined, the overall system architecture and its physical implementation can be analyzed.

Glove The dorsal and palmar protrusions are firmly stitched to the fingerless glove. To ensure a uniform distribution of the pulling force, the distal tendon extremities are anchored on top of these protrusions and additionally secured to the distal edges of the glove near the metacarpophalangeal joints. Four individual tendon branches depart from the glove. Two parallel branches are dedicated to flexion/extension: these originate on the dorsal side (even though they are fixed also to the palmar side for reinforcement) and converge into a single main tendon via a reinforced stitch. The remaining two branches are dedicated to pronation/supination: one originates on the dorsal side near the thumb, while the other originates on the palmar side near the base of the little finger. These two branches converge into a single tendon through a tunable string adjuster. This configuration allows the user to calibrate the adjuster position, ensuring both branches maintain optimal, symmetric tension.

To facilitate the donning process, the tendons are embroidered with the identifiers *FE* (Flexion/Extension) and *PS* (Pronation/Supination). The distal tendon segments termi-



Figure 2.45: Tendon driven prototype.

nate with plastic buckles, allowing the user to easily connect them to their corresponding proximal segments on the forearm sleeve.

The Adafruit IMU is sewn onto the right side of the dorsal protrusion. This lateral placement was selected to prevent spatial interference with the FE and PS tendons both routed along the left side. The ESP32 microcontroller is non permanently nested beneath the tendon routing between the fingers and the dorsal protrusion; the orientation of the GPIO pins and the stiffness of the wiring harness inherently lock the board in place without requiring rigid fasteners. A USB C cable routes from the ESP32 to an external laptop to provide serial communication and control. Additionally, the USB C power distribution breakout board is stitched to the thumb branch of the PS tendon, providing an independent power line routed directly to the servo motors.



Figure 2.46: Dorsal and palmar sides of the glove.

Sleeve The forearm sleeve houses the two servo motors and the proximal Bosch IMU. The FE servo motor is mounted dorsally on the proximal portion of the forearm, whereas the PS servo motor is positioned near the elbow. The proximal IMU is mounted directly above the PS servo. This specific placement is anatomically critical: the proximal IMU serves as the absolute reference frame for calculating the relative wrist joint angle. Consequently, it must be mounted on a skeletal landmark that remains stationary during both

FE and PS movements. While any dorsal point on the forearm is stationary during wrist extension, the elbow is the only anatomically fixed region during forearm rotation (in both mobile and immobile ulna kinematic configurations).

The servo motors are secured within a custom harness composed of crossed fabric bands, further stabilized by crocheted bindings anchored around the motor housings. The IMU is similarly secured atop the PS motor using crossed crocheted bindings. Stacking the IMU on top of the servo introduces a potential source of measurement noise, as mechanical vibrations or shifting of the servo on the soft textile base could displace the sensor. However, the inverse configuration (mounting the servo on top of the IMU) was rejected, as elevating the PS servo would disrupt the close contact tendon routing required for the wrapping actuation mechanism. To mitigate actuator shifting, a circumferential tensioning strap equipped with a string adjuster wraps around the forearm and over the servos. This additional structural constraint is particularly vital during passive operating modes, where precise trajectory tracking is needed. Finally, the proximal halves of the tendons, complete with mating buckles and embroidered labels, are anchored to the servo motor horns.



Figure 2.47: Forearm sleeve with one servo motor on the top and one servo motor and one IMU on the right, corresponding to the elbow side.

Tendon routing As previously detailed, the two distal branches originating from the dorsal side of the glove converge into a single primary FE tendon. This tendon is routed longitudinally along the dorsal surface of the forearm. Due to geometric constraints and limited mounting space on the sleeve, the FE servo motor is slightly offset laterally, resulting in a minor leftward deviation in the proximal tendon path.

Conversely, the two PS tendon branches, originating from the thumb and little finger, wrap around the distal forearm before converging into a single primary PS tendon. This combined tendon completes one full helical wrap around the wrist and forearm before anchoring to the proximally mounted PS servo motor.



Figure 2.48: Tendon routing for flexion/extension.

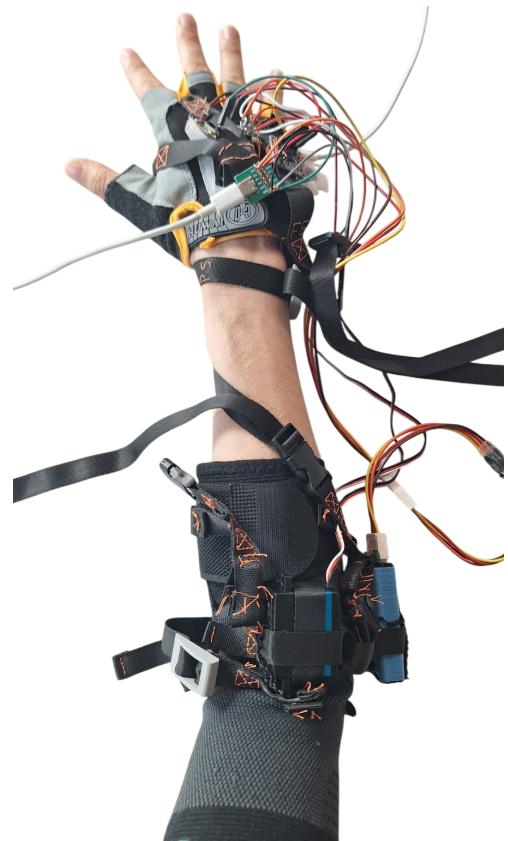


Figure 2.49: Tendon routing for pronation/supination.

Electronic connections The connections between all the electronic elements with jumper wires and USB-C cables are reported in Figure 2.50.

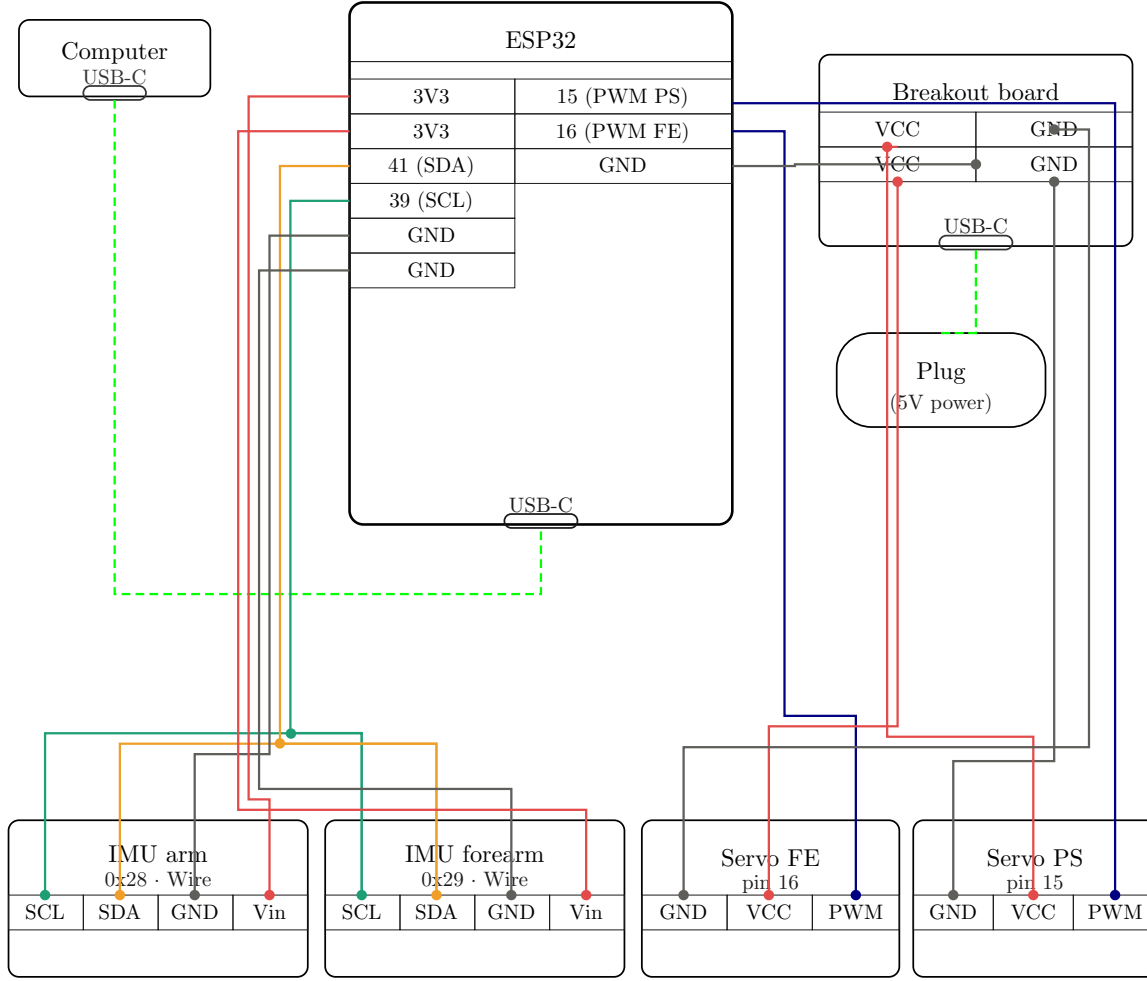


Figure 2.50: Connections between electronic parts.

Functioning The fundamental operating principle of the prototype, independent of the high level control strategy, integrates dedicated sensing and actuation loops. On the sensing side, the instantaneous wrist joint angle is computed continuously by calculating the relative orientation of the distal hand IMU against the fixed reference frame of the proximal forearm IMU. On the actuation side, the FE servo retracts the longitudinally routed FE tendon to actively drive wrist extension, while the PS servo retracts the helically routed tendon to actively drive forearm supination. As established in the biomechanical models, wrist flexion relies on gravitational return, and forearm pronation relies on the natural elastic recoil of the passive musculoskeletal tissues, successfully negating the necessity for active counter actuation or supplementary elastic bands.

2.2.4. Feasibility study with electrohydraulic actuators

Before detailing the electrohydraulic design, it is necessary to explicitly distinguish the scope of this section from the preceding ones. While the tendon driven exosuit discussed up to now was physically manufactured and experimentally validated on human subjects, the following analysis of HASEL and HALVE actuators constitutes a theoretical and mathematical feasibility study. By applying the identical mechanical geometry and routing architecture used in the physical prototype, this study aims to analytically determine whether these novel soft electroactive materials can meet the same operational requirements as traditional servo motors.

Requirements

Because overall system requirements are fundamentally dictated by the functional and clinical needs of the user rather than the underlying actuation technology, the core specifications remain identical to those established in Section 2.1.

However, soft electrohydraulic actuators currently exhibit lower force outputs at high strains compared to traditional rigid tendon driven systems. Consequently, this feasibility study strategically adjusts the mechanical target for wrist extension. Rather than attempting the 100% gravity compensation required for a fully passive control strategy, the objective is scaled to a realistic but still ambitious 50% compensation threshold. This level of force output is entirely sufficient for active assistive control modes, where the user retains partial motor capability. Conversely, the 100% compensation target for pronation and supination is maintained, as the resistive forces associated with internal joint stiffness are significantly lower than the full gravitational loads encountered during extension.

Despite these force limitations, substituting rigid servo motors with HASEL or HALVE actuators introduces distinct functional advantages. The resulting theoretical system is projected to be significantly lighter, possess faster actuation dynamics, and operate completely silently. Therefore, while the raw force expectations are slightly relaxed, the requirements regarding system mass, operational speed, and acoustic noise are anticipated to be not just met, but exceeded.

Assumptions

Consistent with the established system requirements, the target Range of Motion (RoM) for both the flexion/extension and pronation/supination degrees of freedom is strictly limited to the functional range required for Activities of Daily Living (ADLs). This delib-

erate restriction prevents the application of potentially harmful forces near the anatomical extremes of the user's joints.

As previously stated, the electrohydraulic actuation strategy for wrist extension aims to compensate for only a designated percentage of the total gravitational moment rather than achieving 100% compensation. This partial assistance approach is highly suitable for active assistive control schemes and provides the critical advantage of limiting the internal skeletal loads imposed on the user. Because soft exosuits inherently lack a rigid load bearing frame in parallel with the human body, all applied mechanical forces are transmitted directly through the musculoskeletal system of the wearer. For individuals with long term motor impairments, prolonged disuse often results in reduced bone mineral density and joint vulnerability. Minimizing the peak actuation forces thus serves as a vital safety measure to prevent excessive skeletal stress and potential injury.

A final kinematic assumption reflects the natural posture of the upper limb during ADLs. In typical functional tasks, the dorsal surface of the hand is rarely oriented perfectly parallel to the ground: instead, the forearm is frequently maintained in a partially supinated posture. Consequently, only a geometric projection of the total gravitational force acts directly against the flexion/extension plane. Therefore, the subsequent dynamic computations for wrist extension model the hand in this typical partially supinated orientation to provide a more accurate and realistic estimation of the required assistive torque.

Forces, moments and torques

Because the kinematic geometry of the electrohydraulic driven exosuit is assumed to be identical to the tendon driven prototype, the foundational biomechanical models and geometric force derivations remain valid. However, the revised analytical assumptions, specifically the partial compensation threshold ($comp_{FE} = 50\%$) and the natural supination angle ($\beta = 30^\circ$) during the resting posture, must be integrated into the flexion/extension load calculations.

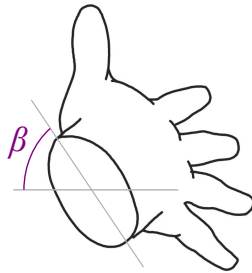


Figure 2.51: Supinated resting position in flexion/extension.

Taking this angular projection and partial compensation into account, the effective gravitational force to be compensated, $\mathbf{F}_{g, \text{comp}}$, replaces the absolute gravitational force $\mathbf{F}_g$:

$$\mathbf{F}_{g, \text{comp}} = \mathbf{F}_g \frac{\text{comp}_{\text{FE}}}{100} \cos \beta \quad (2.43)$$

The corresponding target moment required for wrist extension, $\mathbf{M}_{\text{comp, FE}}$, is then calculated as:

$$\mathbf{M}_{\text{comp, FE}} = \mathbf{F}_{g, \text{comp}} r \quad (2.44)$$

The remainder of the required actuation force derivation follows the identical methodology previously established for the stitched bump configuration.

For the pronation and supination degrees of freedom, the target compensation remains at 100% of the passive joint stiffness. Consequently, the required torque and force parameters are identical to those calculated for the tendon driven prototype.

The governing numerical results for this electrohydraulic feasibility study are summarized in Table 2.7.

Table 2.7: Design parameters for the electrohydraulic actuator driven exosuit.

Quantity	Symbol	Unit	Value
Flexion/extension			
Gravity force	F_g	N	4.12
Percentage of gravity to compensate	comp_{FE}	%	50
Supination angle in resting position	β	°	30
Gravity force to compensate	$F_{g, \text{comp}}$	N	1.78
Maximum moment due to gravity	$M_{\text{max g, FE}}$	Nm	0.38
Maximum moment to compensate	$M_{\text{comp, FE}}$	Nm	0.17
Required force in stitched bump configuration	F_{FE}	N	4.45
Pronation/supination			
Percentage of passive stiffness to compensate	comp_{PS}	%	100
Force from thumb side	F_t	N	2.82
Force from little finger side	F_l	N	2.82

Force, strain and voltage

Having determined the required actuation force, the subsequent step is to calculate the corresponding actuator strain. In electrohydraulic actuators, output force and linear strain exhibit an inversely proportional relationship, as illustrated in Figure 2.52.

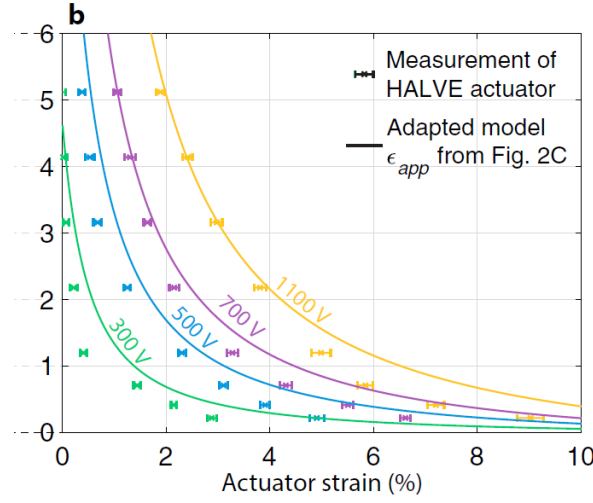


Figure 2.52: Force/strain curves at different voltages for HALVE actuators from [25].

Before detailing the specific actuator sizing for the exosuit, an analytical relationship between force and strain at various operating voltages must be established. This analysis begins by deriving the mechanical models for HALVE actuators, followed by an extension of this methodology to HASEL actuators.

In the supplementary materials provided by Gravert et al. [25], an empirical formula correlating force and strain at an operating voltage of 1300 V is defined as:

$$F(\varepsilon) = \frac{2.81}{\sin(0.17(\varepsilon - 0.23))} - 2.86 \quad (2.45)$$

Because the voltage converter utilized in this thesis is limited to a maximum output of approximately 780 V, a new empirical relationship must be derived to fit the experimental data points presented in Figure 2.52.

First, discrete data points were extracted from the published force strain curves using a web plot digitizer [58]. To ensure the highest fidelity, only the raw empirical data points were selected rather than interpolating from the continuous curves. These coordinates were subsequently structured into matrices categorized by their corresponding operating voltage.

To correct for initial digitizer calibration errors, the raw coordinates were rescaled using linear transformations:

$$x_{\text{real}} = ax \cdot x_{\text{digitized}} + bx \quad (2.46)$$

$$y_{\text{real}} = ay \cdot y_{\text{digitized}} + by \quad (2.47)$$

The transformation coefficients (ax, bx, ay, by) were determined by sampling known reference points on the axes (e.g., $(0, 0)$, $(10, 0)$, and $(0, 6)$) and mapping the digitized pixel coordinates to their true physical values.

With the correctly scaled data points, the characteristic curves can be reconstructed. To maintain mathematical consistency with the established 1300 V model, the four numerical coefficients from the original formula are treated as unknowns (A, B, C, D) and solved for each voltage level:

$$\mathbf{F}(\varepsilon) = \frac{A}{\sin(B - C\varepsilon)} - D \quad (2.48)$$

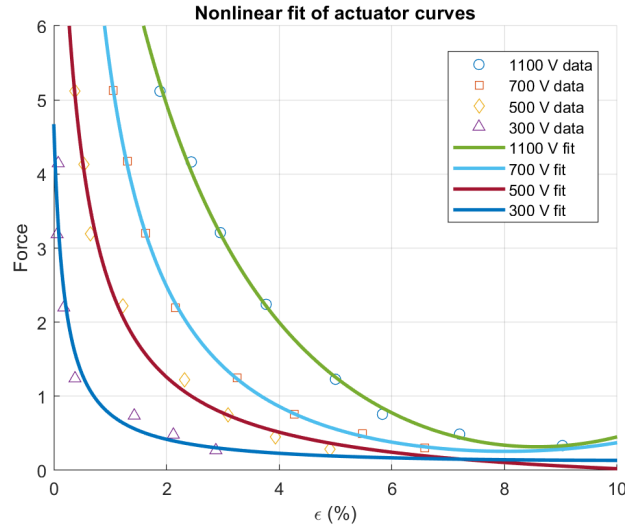


Figure 2.53: HALVE actuators curve reconstruction.

To determine the coefficients for the 1100 V curve, the parameters from the 1300 V equation were utilized as initial numerical guesses. For each subsequent lower voltage, the computed coefficients from the preceding curve served as the initial guesses. The sole exception occurred during the 300 V curve fitting, where numerical instability required the parameter B to be manually fixed to its 1100 V value. This adjustment was empirically derived to ensure convergence rather than being based on strict theoretical considerations.

While the equations for the published voltage curves were successfully identified, the specific 780 V target required for this prototype was not among them. To enable the extraction of a force strain curve for any arbitrary voltage, a continuous 3D force strain voltage surface must be reconstructed from the discrete 2D profiles.

To interpolate between the known curves, the numerical trends of the four calculated coefficients were plotted against the operating voltage.

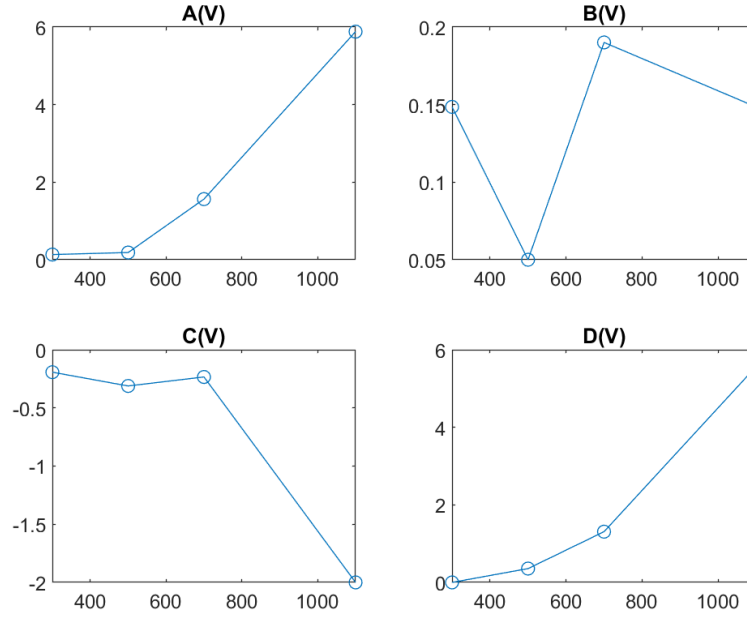


Figure 2.54: Behaviour of the coefficients of the force/strain relationship at different voltages.

With the exception of B (which exhibits discontinuity due to the manual constraint at 300 V), the remaining coefficients demonstrate almost parabolic trends. Consequently, their behavior was accurately approximated using second order polynomial regressions. By evaluating these polynomials across the continuous voltage domain, the complete 3D actuation surface can be generated.

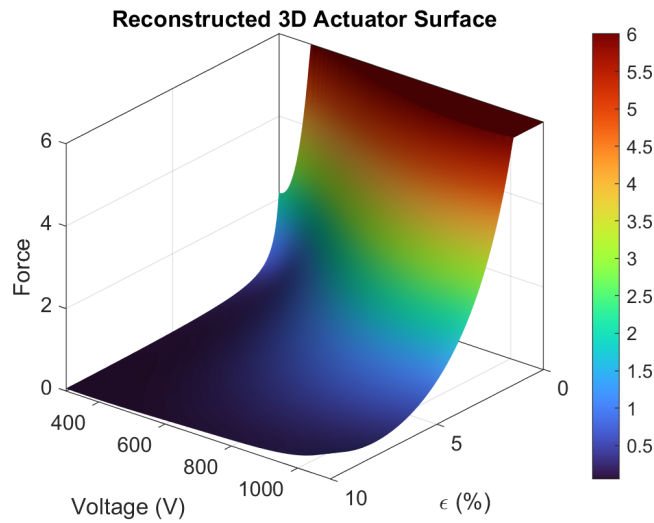


Figure 2.55: HALVE actuators reconstructed 3D surface force/strain/voltage.

Finally, the specific 2D force strain curve operating at 780 V can be extracted from this reconstructed surface.

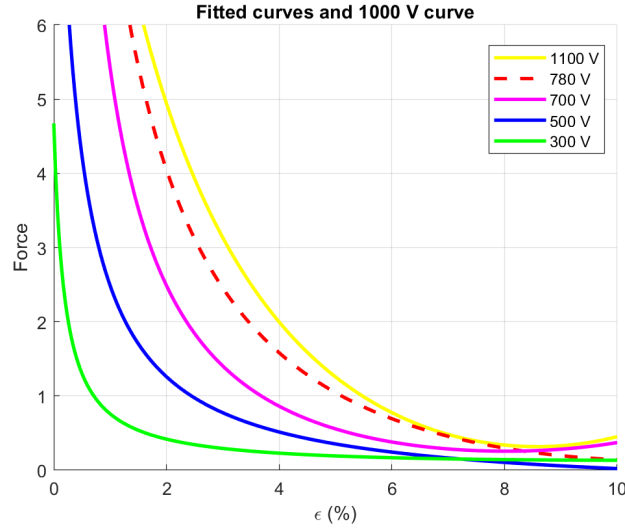


Figure 2.56: HALVE actuators reconstruction of curve at 780 V.

The resulting explicit force equation and its algebraic inverse are:

$$\mathbf{F}(\varepsilon) = \frac{2.0787}{\sin(0.1375(\varepsilon + 0.5723))} - 1.9518 \quad (2.49)$$

$$\varepsilon(\mathbf{F}) = \frac{1}{0.1375} \left(\arcsin\left(\frac{2.0787}{\mathbf{F} + 1.9518}\right) \right) - 0.5723 \quad (2.50)$$

The architectural dimensions of the pouches utilized in [25], alongside other relevant physical parameters, are detailed in Table 2.8. A geometric reference for these variables is provided in Figure 2.57.

Table 2.8: HALVE actuators' pouch dimensions and other parameters.

Quantity	Symbol	Unit	Value
Height of the whole pouch	h_{tot}	mm	17
Height of the region of the pouch occupied by electrodes	h_{el}	mm	9
Length of the pouch	l	mm	60
Thickness of the film	t	μm	12.5
Voltage	V	V	780
Permittivity of vacuum	ε_0	F/m	$8.854 \cdot 10^{-12}$
Permittivity of the film (BoPET)	ε_r	F/m	3.0

Before applying these models to the exosuit case study, the equivalent force strain voltage relationship must be derived for the HASEL actuators.

Kellaris et al. [33] present a comprehensive analytical model to compute the force and strain outputs of HASEL actuators. Unlike the HALVE literature [25], which primarily provided empirical plots, this analytical framework allows for direct equations derivation. However, the variables are parametrically decoupled: linear strain is expressed as a function of the pouch's central folding angle (originally denoted as α in literature, but reassigned here as α_{pouch} to prevent conflict with the flexion/extension routing angle), and force is expressed as a function of both voltage and this central angle. The parametric equations are defined as:

$$\varepsilon = 1 - \frac{\alpha_{\text{pouch}, 0}}{\sin \alpha_{\text{pouch}, 0}} \left(1 + \frac{\sqrt{2A}}{h_{\text{tot}}} \frac{\sin \alpha_{\text{pouch}} - \alpha_{\text{pouch}}}{\sqrt{\alpha_{\text{pouch}} - \sin \alpha_{\text{pouch}} \cos \alpha_{\text{pouch}}}} \right) \quad (2.51)$$

$$\mathbf{F} = \frac{l}{4t} \frac{\cos \alpha_{\text{pouch}}}{1 - \cos \alpha_{\text{pouch}}} \varepsilon_0 \varepsilon_r V^2 \quad (2.52)$$

where the geometric dimensions (illustrated in Figure 2.57) and physical parameters are defined as:

- h_{tot} is the height of the whole pouch;
- h_{el} is the height of the region of the pouch occupied by the electrodes;
- l is the length of the pouch;
- t is the thickness of the film;
- V is the voltage;
- $\alpha_{\text{pouch}, 0}$ is the initial central angle;
- $\alpha_{\text{pouch}, f}$ is the final central angle;
- ε_0 is the permittivity of vacuum;
- ε_r is the permittivity of the film;
- $A = \frac{(h_{\text{tot}} - h_{\text{el}})^2}{\pi}$ is the cross section of the pouch.

While it might seem intuitive to analytically invert the strain equation to solve for α_{pouch} and substitute it into the force equation, the severe mathematical nonlinearity of the strain formulation prevents a closed form algebraic inversion. This geometric complexity

necessitates the use of a parametric representation. Consequently, a numerical approach is required to link force directly to strain.

Unlike the HALVE actuators, which used fixed, pre established dimensions from the literature [25], HASEL actuators can be custom manufactured, necessitating a specific geometric sizing phase for the pouches.

Following an iterative optimization process (which benchmarked custom dimensions against viable manufacturing capabilities documented in external literature), the finalized HASEL pouch dimensions were established. These parameters are detailed in Table 2.9 and illustrated in Figure 2.57.

Table 2.9: HASEL actuators' pouch dimensions and other parameters.

Quantity	Symbol	Unit	Value
Height of the whole pouch	h_{tot}	mm	20
Height of the region of the pouch occupied by electrodes	h_{el}	mm	9
Length of the pouch	l	mm	60
Thickness of the film	t	μm	18
Voltage	V	kV	6
Permittivity of vacuum	ε_0	F/m	$8.854 \cdot 10^{-12}$
Permittivity of the film (BOPP) ^a	ε_r	F/m	2.2

^a Biaxially Oriented Polypropylene (BOPP) and Biaxially Oriented Polyethylene Terephthalate (BoPET) are standard dielectric materials for HASEL films, possessing relative permittivities of 2.2 F/m and 3.0 F/m, respectively. The lower permittivity of BOPP was deliberately selected to enforce a conservative design margin; validating the system with BOPP inherently guarantees functional feasibility if the higher performing BoPET is ultimately used.

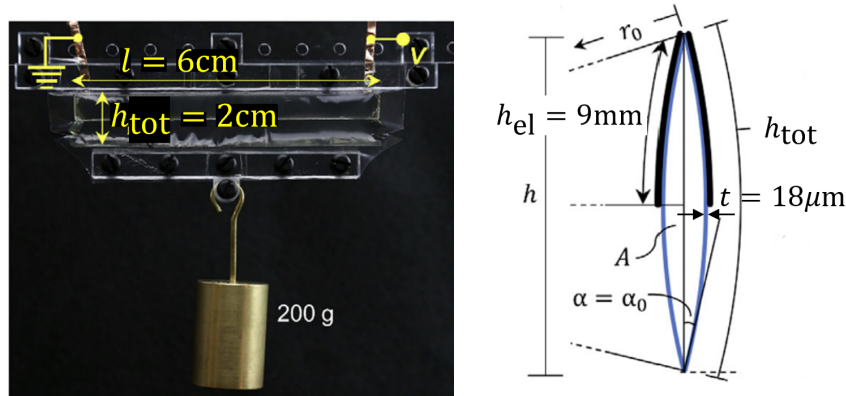


Figure 2.57: HASEL pouches dimensions from [33] and then edited.

Using the finalized geometry, the parametric response curve at 6 kV (the maximum voltage

threshold for the proposed control electronics) can be generated using Equations 2.51 and 2.52.

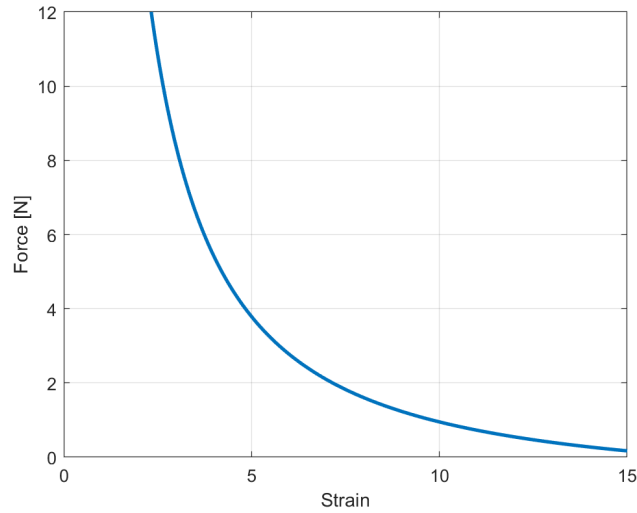


Figure 2.58: HASEL curve force/strain at 6 kV according to the analytic model presented in [33].

While visualizing the discrete parametric relationship is useful, a continuous, direct analytical function linking force to strain must be derived to mirror the HALVE analysis. Because HASEL and HALVE actuators share identical underlying physics, the fitting model previously established in Equation 2.48 was applied. The resulting fit is presented in Figure 2.59.

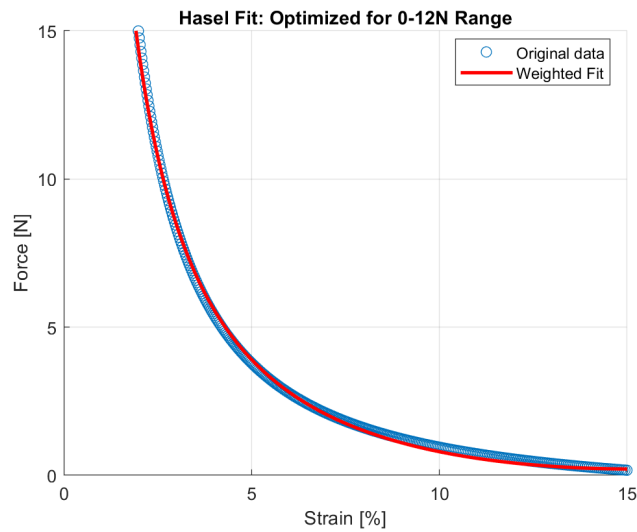


Figure 2.59: HASEL curve force/strain obtained by using the same model as HALVE.

While the fitting methodology is conceptually similar to the one applied to the HALVE actuators, the MATLAB optimization function `lsqnonlin` was utilized instead of `lsqcurvefit`. This substitution enabled the application of weighted priorities across specific regions of the response curve. The analytical model naturally generates exceptionally high forces (approaching 1000 N) at near zero strains, severely skewing an unweighted regression away from the functionally relevant lower force region. Applying strict weights ensures high fidelity fitting strictly within the operational range. This weighting was unnecessary for the HALVE actuators, as the manual point extraction naturally filtered out data outside the functional bounds.

The resulting optimized regression equation defining force as a direct function of strain for the HASEL actuators is:

$$\mathbf{F}(\varepsilon) = \frac{1.8694}{\sin(0.0976(\varepsilon - 0.0233))} - 1.7801 \quad (2.53)$$

$$\varepsilon(\mathbf{F}) = \frac{1}{0.0976} \left(\arcsin\left(\frac{1.8694}{\mathbf{F} + 1.7801}\right) \right) + 0.0233 \quad (2.54)$$

Actuation system design

As evidenced by the force-strain plots in the preceding section, the force magnitudes required for the exosuit correspond to exceedingly low actuator strains⁵. This inverse relationship presents a significant design challenge: the prototype requires both sufficient force to overcome gravity and passive muscle stiffness, and sufficient linear contraction to cover the target Range of Motion (RoM) for ADLs. Consequently, simply trading force for strain, or vice versa, is not a viable strategy. Traditional mechanical transmissions, such as pulley systems, amplify force directly at the expense of linear displacement (or the opposite), making them unsuitable for overcoming this dual constraint. Therefore, the optimal solution requires determining a specific hardware configuration of actuators that natively satisfies both the force and contraction requirements without relying on supplementary mechanical amplification.

To establish this configuration, two foundational kinematic and kinetic principles regarding modular actuator scaling must be applied:

- Actuators configured in parallel proportionally divide the required target force among themselves (subject to a minor efficiency reduction due to mechanical friction between adjacent units).

⁵Although the referenced plot specifically illustrates HALVE actuators, HASEL actuators generate only marginally higher forces; therefore, this fundamental reasoning applies equally to both.

- Actuator pouches configured in series cumulatively increase the absolute total contraction, as linear strain is defined as a percentage of the initial resting length of the actuator.

Because simultaneously satisfying both force and contraction limits is complex, an iterative analytical strategy is employed. The methodology involves constraining one primary parameter (either force or contraction) to its target requirement, computing the resulting output of the secondary parameter for a given number of parallel actuators (n_{par}) and series pouches (n_{pou}), and comparing this result against the system requirements. This comparison yields a theoretical amplification ratio, which dictates how much mechanical advantage would be required if a physical transmission were to be used.

The first analytical approach imposes the target required force ($\mathbf{F}_{\text{need}}$) as the governing constraint. This procedure is mathematically generalized and applies equally to both the flexion/extension and pronation/supination subsystems. For a given number of parallel actuators (n_{par}), the individual force each actuator must provide is:

$$\mathbf{F}_{\text{sing}} = \frac{\mathbf{F}_{\text{need}}}{n_{\text{par}}} \quad (2.55)$$

Using the inverse parametric equations defining strain ε as a function of force $\mathbf{F}$ (Equation 2.54 for HASEL or Equation 2.50 for HALVE), the corresponding operational strain is calculated. For an actuator comprising n_{pou} pouches in series, each with a total resting height h_{tot} , the absolute linear contraction of a single pouch is evaluated as:

$$c_{\text{pou}} = \frac{\varepsilon}{100} h_{\text{tot}} \quad (2.56)$$

The total linear contraction of the entire series actuator is then:

$$c_{\text{tot}} = n_{\text{pou}} c_{\text{pou}} \quad (2.57)$$

Finally, comparing this theoretical output to the target total contraction required for the ADL RoM (c_{need}), the required amplification ratio for contraction (amp_c) is derived:

$$\text{amp}_c = \frac{c_{\text{need}}}{c_{\text{tot}}} \quad (2.58)$$

A mechanically viable configuration is achieved if both amplification ratios (amp_c and amp_F) are ≤ 1 , meaning no external amplification is required. Alternatively, if a mechan-

ical transmission is introduced, one ratio may exceed 1 exclusively if the reciprocal ratio is proportionally lower, adhering to the fundamental conservation of work constraints inherent to passive mechanical systems.

Alternatively, the system can be evaluated from the opposite boundary condition by enforcing the target contraction (c_{need}) as the primary constraint. Given n_{pou} pouches in series within a single actuator, the specific linear displacement required from each individual pouch is:

$$c_{\text{need,pou}} = \frac{c_{\text{need}}}{n_{\text{pou}}} \quad (2.59)$$

Given the total resting height h_{tot} of a single pouch, the corresponding required actuator strain is defined as:

$$\varepsilon_{\text{need}} = \frac{c_{\text{need,pou}}}{h_{\text{tot}}} \cdot 100 \quad (2.60)$$

With the target strain established, the empirical models defining force as a function of strain (Equation 2.53 for HASEL or Equation 2.49 for HALVE) are evaluated. Inputting $\varepsilon_{\text{need}}$ yields the maximum theoretical force generated by a single actuator unit, designated as $\mathbf{F}_{\text{sing}, 2}$ ⁶.

When multiple actuators (n_{par}) are configured in parallel, their individual force contributions sum linearly, resulting in a total available system force of:

$$\mathbf{F}_{\text{tot}} = n_{\text{par}} \mathbf{F}_{\text{sing}, 2} \quad (2.61)$$

Finally, by comparing this total available force against the biomechanically required target force ($\mathbf{F}_{\text{need}}$) imposed by the mechanical geometry, the required amplification ratio for the output force (amp_F) is determined:

$$\text{amp}_F = \frac{\mathbf{F}_{\text{need}}}{\mathbf{F}_{\text{tot}}} \quad (2.62)$$

Exosuit with HASEL actuators

Following an iterative optimization process, the optimal actuator configurations were established. The selected hardware architectures are detailed in Table 2.10 for flex-

⁶The notation $\mathbf{F}_{\text{sing}, 2}$ is deliberately used to distinguish the actual force capable of being produced by a single actuator operating at the required strain (contraction requirement $\rightarrow$ available force), as opposed to $\mathbf{F}_{\text{sing}}$, which denotes the theoretical force strictly required from a single actuator to perfectly compensate the target load (force requirement $\rightarrow$ resulting contraction).

ion/extension and Table 2.11 for pronation/supination. These tables summarize the numerical results derived from the analytical framework presented in Section 2.2.4.

Table 2.10: Design for flexion/extension of the exosuit with HASEL actuators.

Quantity	Symbol	Unit	Value
Number of actuators in parallel	n_{par}	—	2
Number of pouches in series per actuators	n_{pou}	—	20
Needed force	F_{need}	N	4.45
Needed contraction	c_{need}	m	0.02
Force requirement $\rightarrow$ contraction			
Force to each actuator in parallel	F_{sing}	N	2.23
Strain	ε	%	5.0
Contraction of each pouch	c_{pou}	m	0.001
Total contraction	c_{tot}	m	0.02
Contraction amplification	amp_c	—	1
Contraction requirement $\rightarrow$ force			
Needed contraction per pouch	$c_{\text{need, pou}}$	m	0.001
Needed strain	$\varepsilon_{\text{need}}$	%	5
Force from each actuator	$F_{\text{sing, 2}}$	N	2.23
Total force	F_{tot}	N	4.45
Force amplification	amp_F	—	1

As demonstrated in Table 2.10, a configuration involving two parallel actuators, each comprising 20 pouches in series, is perfectly capable of providing the necessary force and contraction to achieve wrist extension (while compensating for 50% of gravity from a 30° supinated resting position) without any need for mechanical amplification.

This represents a highly significant milestone: it constitutes both the first documented feasibility study of a soft wearable exosuit powered by HASEL actuators, and the first theoretical application of HASEL technology operating under biomechanical load scales of this magnitude.

However, a practical spatial limitation arises from this optimal configuration. Twenty series pouches of 2 cm each equate to a total actuator length of 40 cm, which exceeds the anatomical limits of a standard human forearm. To resolve this clearance issue, alternative routing geometries must be evaluated.

The first potential configuration (illustrated in Figure 2.60) extends the routing proximally across the elbow joint, utilizing the upper arm to accommodate the extensive actuator length.

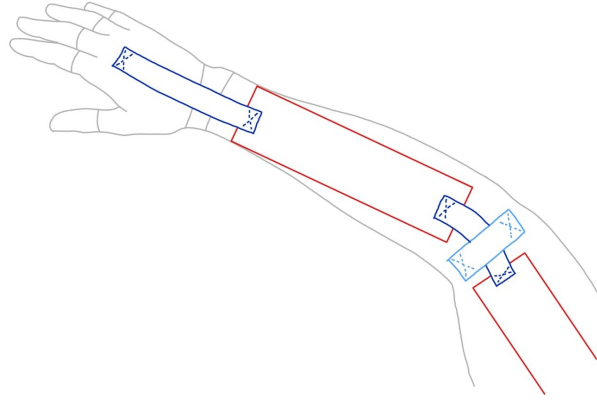


Figure 2.60: Possible proximal routing geometry for flexion/extension with electrohydraulic actuators. Note: This illustration is a simplified schematic; while the actual design requires two parallel actuators, only one is depicted to clearly demonstrate the underlying routing logic.

However, this approach introduces two major biomechanical and functional drawbacks. First, routing the active tension band across the elbow joint inherently applies parasitic compression and shear forces to the elbow during actuation, placing an unassociated joint under unwanted stress (a critical safety concern for motor-impaired users). Second, it compromises a fundamental advantage of soft wearable robotics: localized actuation. If the system necessitates anchoring components across distant body segments, conventional back-mounted rigid motors with Bowden cables would be mechanically simpler and more efficient.

To bypass these limitations while remaining confined to the forearm, a folded pulley configuration is proposed, as shown in Figure 2.61.

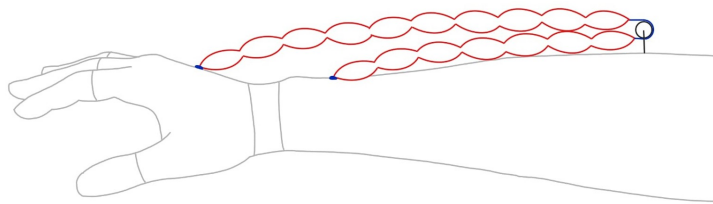


Figure 2.61: Folded pulley geometry for flexion/extension with electrohydraulic actuators. The image displays the top actuator layer; the second parallel actuator is located directly behind it. Note: This is a schematic representation, and the depicted pouch quantity is purely illustrative.

Inspired by the mechanical folding concepts described by Wang et al. [71], this geometry effectively packages an extensive series of pouches within a strictly limited linear distance.

Rather than utilizing two continuous 40 cm actuators, the system employs four 20 cm actuators (10 pouches each). By physically folding these units back on themselves via a distal inactive string routed around a passive pulley, the required contraction and force are maintained within half the longitudinal footprint. The primary trade-off of this solution is a minor increase in the radial profile (thickness) of the device over the forearm.

The analytical methodology was subsequently applied to the pronation/supination sub-systems, with the resulting design parameters summarized in Table 2.11.

Table 2.11: Design for pronation/supination of the exosuit with HASEL actuators.

Quantity	Symbol	Unit	Value
Number of actuators in parallel for each side (thumb, little finger) ^a	n_{par}	—	1
Number of pouches in series per actuator	n_{pou}	—	20
Needed force	F_{need}	N	2.82
Needed contraction	c_{need}	m	0.02
Force requirement $\rightarrow$ contraction			
Force to each actuator in parallel for each side	F_{sing}	N	2.82
Strain	ε	%	4.32
Contraction of each pouch	c_{pou}	m	0.0009
Total contraction	c_{tot}	m	0.0173
Contraction amplification	amp_c	—	1.16
Contraction requirement $\rightarrow$ force			
Needed contraction per pouch	$c_{\text{need, pou}}$	m	0.001
Needed strain	$\varepsilon_{\text{need}}$	%	5
Force from each actuator	$F_{\text{sing, 2}}$	N	2.23
Total force	F_{tot}	N	2.23
Force amplification	amp_F	—	1.27

^a The overall pronation/supination architecture uses two wrapping actuators in parallel. However, because the biomechanical load was symmetrically divided into discrete thumb and little finger force vectors during the kinematic derivations, the sizing analysis is conducted per individual branch. Consequently, $n_{\text{par}} = 1$ denotes a single dedicated actuator for the thumb routing and a single dedicated actuator for the little finger routing.

Although the calculated amplification ratios slightly exceed the ideal 1.0 threshold, this configuration remains highly feasible and structurally sound. Primarily, substituting the conservative BOPP dielectric film with higher-permittivity BoPET would substantially decrease these required amplifications. Furthermore, because the pronation/supination design originally targeted an aggressive 100% torque compensation (unlike the 50% target strictly set for wrist extension), a fractional reduction in output force simply yields a slightly lower overall compensation percentage. This remaining output is still overwhelm-

ingly abundant for active assistive control schemes.

Interestingly, this analysis reveals that pronation/supination, often the most mechanically complex and functionally neglected degree of freedom in traditional rigid wrist exoskeletons, proves to be exceptionally well suited to the inherent wrapping nature of this novel soft actuation technology.

The final theoretical implementation featuring the parallel wrapping actuators is conceptualized in Figure 2.62.

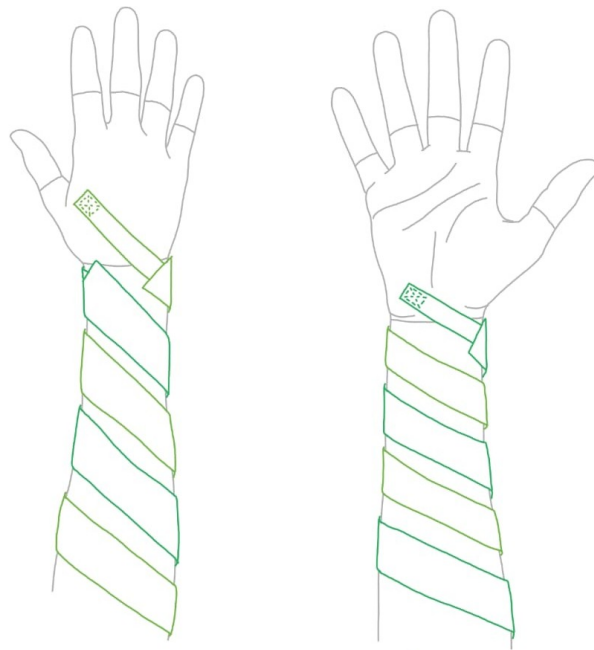


Figure 2.62: Wrapping geometry for pronation/supination with electrohydraulic actuators. Note: This drawing is a simplified schematic highlighting the helical routing of the parallel actuators; the underlying glove anchoring protrusions have been omitted for clarity.

While a continuous 20-pouch actuator per anatomical branch represents the theoretical optimum, practical manufacturing and modularity considerations favor utilizing the identical 10-pouch actuator modules (placing two of them in series for each branch) developed for the flexion/extension subsystem. This component standardization would allow a single, universally sized actuator type to be produced and interchangeably integrated across all degrees of freedom in the exosuit.

Exosuit with HALVE actuators

The mechanical architecture of the exosuit equipped with HALVE actuators directly parallels that of the HASEL system, involving parallel flexion actuators routed through a folded pulley configuration and parallel supination actuators wrapped helically around the forearm.

The primary advantages of HALVE actuators are their operation at significantly lower voltages, which translates to a drastically lighter and less cumbersome power electronics subsystem, and their inherent safety for human-robot interaction due to fully encapsulated, non exposed electrodes. Conversely, they currently exhibit lower structural robustness and generate comparatively lower forces at equivalent strains than HASEL actuators. Consequently, to maintain a viable design without exceeding realistic spatial dimensions, the compensation target was deliberately adjusted to 30% for both flexion/extension and pronation/supination (maintaining the 30° supinated resting posture). While lower than the original tendon driven requirements, this reduced threshold remains highly clinically relevant for active assistive modes; for instance, Bardi et al. [4] demonstrated functional continuous assistance using gravity compensation levels as low as 25%.

The numerical results dictating the optimal actuator configurations are reported in Table 2.12 for flexion/extension and Table 2.13 for pronation/supination.

The proposed design for flexion/extension necessitates three parallel actuators, each comprising 21 series pouches. This specific pouch quantity was deliberately selected as a multiple of three to accommodate current manufacturing constraints: HALVE actuators are currently most reliably fabricated in short, robust modules to prevent premature structural failure under tension. Therefore, each full 21-pouch actuator is functionally constructed by connecting seven independent 3-pouch modules in series.

While this analysis successfully demonstrates the theoretical feasibility of a HALVE driven soft wrist exosuit, it highlights a distinct spatial trade-off. Integrating three parallel actuators within the folded pulley configuration inherently increases the overall radial bulk of the device on the forearm compared to the dual actuator HASEL design.

For pronation/supination, the optimal routing architecture retains the exact same two actuator parallel wrapping geometry (one for the thumb branch, one for the little finger branch) as the HASEL driven exosuit, perfectly preserving its compact volumetric profile. However, this dimensional parity is achieved strictly at the cost of functional output capability, limiting the total active torque compensation to 30% compared to the full 100% achieved by the HASEL model.

Table 2.12: Design for flexion/extension of the exosuit with HALVE actuators.

Quantity	Symbol	Unit	Value
Number of actuators in parallel	n_{par}	–	3
Number of pouches in series per actuator	n_{pou}	–	21
Percentage of gravity to compensate	$comp_{\text{FE}}$	%	30
Needed force	F_{need}	N	2.67
Needed contraction	c_{need}	m	0.02
Force requirement $\rightarrow$ contraction			
Force to each actuator in parallel	F_{sing}	N	0.89
Strain	ε	%	5.0
Contraction of each pouch	c_{pou}	m	0.0009
Total contraction	c_{tot}	m	0.0193
Contraction amplification	amp_c	–	1.04
Contraction requirement $\rightarrow$ force			
Needed contraction per pouch	$c_{\text{need, pou}}$	m	0.001
Needed strain	$\varepsilon_{\text{need}}$	%	5
Force from each actuator	$F_{\text{sing, 2}}$	N	0.82
Total force	F_{tot}	N	2.45
Force amplification	amp_F	–	1.09

Table 2.13: Design for pronation/supination of the exosuit with HALVE actuators.

Quantity	Symbol	Unit	Value
Number of actuators in parallel for each side	n_{par}	–	1
Number of pouches in series per actuator	n_{pou}	–	21
Percentage of torque to compensate	$comp_{\text{PS}}$	%	30
Needed force	F_{need}	N	0.84
Needed contraction	c_{need}	m	0.02
Force requirement $\rightarrow$ contraction			
Force to each actuator in parallel for each side	F_{sing}	N	0.84
Strain	ε	%	5.0
Contraction of each pouch	c_{pou}	m	0.0009
Total contraction	c_{tot}	m	0.0197
Contraction amplification	amp_c	–	1.01
Contraction requirement $\rightarrow$ force			
Needed contraction per pouch	$c_{\text{need, pou}}$	m	0.001
Needed strain	$\varepsilon_{\text{need}}$	%	5
Force from each actuator	$F_{\text{sing, 2}}$	N	0.82
Total force	F_{tot}	N	0.82
Force amplification	amp_F	–	1.03

Tendon driven vs electrohydraulic actuation: pros and cons

A comprehensive comparison between the physically developed tendon driven prototype and the theoretical electrohydraulic actuation concepts investigated in this work highlights the distinct engineering trade offs associated with each technology.

The tendon driven solution benefits from a high degree of technological maturity, mechanical robustness, and ease of maintenance. The actuation system is based on inexpensive commercial motors and simple tendon transmissions, making component replacement straightforward and significantly reducing the overall system cost. Furthermore, position control can be implemented using well established techniques, resulting in predictable and highly repeatable actuator behavior. From a safety perspective, the system operates at low voltages and relies on commercially available insulated components, substantially mitigating electrical hazards for the user. The current tendon driven implementation is also inherently capable of generating substantially higher assistive forces compared to the electrohydraulic alternatives evaluated.

Conversely, the reliance on traditional electric motors introduces several distinct drawbacks. Motors are relatively heavy and rigid components, which fundamentally contradicts the core compliance objectives of soft wearable robotics. Additionally, they generate audible acoustic noise during operation, and their mechanical pulling action is highly localized. Although the helical and linear tendon routing distributes part of the load along the forearm, the tendons themselves remain rigid structural elements; they strictly transmit force without intrinsically contracting.

Electrohydraulic actuators, such as HASEL and HALVE modules, offer several characteristics that are remarkably attractive for wearable devices. They are inherently soft, lightweight, capable of rapid actuation dynamics, and operate almost entirely silently. Crucially, their contractile action is distributed over the entire active length of the actuator pouches rather than being transmitted through a limited number of discrete distal attachment points. This distributed strain results in a highly uniform interaction with the human body and a significantly more comfortable user experience. Furthermore, despite requiring a larger active surface area than a single routing tendon, electrohydraulic actuators entirely eliminate the need to mount bulky, rigid electric motors and gearboxes directly onto the user's limb.

These biomechanical advantages become particularly evident in the design of the pronation/supination assistance mechanism. Human pronation and supination are naturally generated by muscles whose mechanical action extends over a significant spatial portion of the forearm. Electrohydraulic wrapping actuators effectively biomimic this behavior

by generating continuous contraction along their entire helical length, allowing the assistive torque to be developed through distributed, widespread contact with the forearm skin. In this specific context, structural friction between the actuator and the limb becomes a beneficial mechanical advantage rather than a parasitic loss, as it contributes to transmitting the actuator force over a vastly larger surface area. Tendon driven systems attempt to exploit a similar principle, which heavily motivated the choice of bare tendons over Bowden cables (where force transmission is strictly concentrated at the rigid cable endpoints), but the distributed action achieved by tendons remains functionally limited and can lead to uncomfortable, localized pressure concentrations on the skin.

Despite these groundbreaking theoretical advantages, electrohydraulic actuators currently present several practical limitations. Their absolute force output remains lower than that achievable with conventional motor driven systems, their long term material durability under cyclic loading remains a critical challenge, and their control is mathematically more complex due to the severe nonlinear behavior of soft electroactive dielectrics. In the specific case of HASEL actuators, the fundamental requirement for kilovolt range driving voltages introduces massive secondary challenges related to portable power electronics, dielectric insulation, and strict human electrical safety regulations.

Considerations on electrical safety

Electrical safety is one of the primary concerns associated with the use of electrohydraulic actuators in wearable robotic devices. HASEL actuators typically operate at voltages of several kilovolts, which raises legitimate safety concerns regarding accidental contact between the user and the actuator electrodes or power electronics. However, the physiological severity of an electrical shock is not determined by voltage alone. According to electrical safety studies on dielectric elastomer actuators, the principal hazards associated with high voltage exposure are tissue burns and cardiovascular injuries, particularly ventricular fibrillation. The actual physiological risk heavily depends on the current flowing through the body, the duration of the exposure, and the total energy available in the electrical system.

Pourazadi et al. [55] investigated the electrical safety of Dielectric Elastomer Actuators (DEAs) operating in close proximity to the human body and analyzed several accidental shock scenarios. These included direct contact with the high voltage power supply, contact with a charged actuator disconnected from the power source, and contact with a charged actuator while still connected to the power supply. Their analysis, based on IEC and UL safety standards, highlights the absolute importance of limiting both the available continuous current and the stored electrical energy within the actuator. In particular,

continuous currents below approximately 20 mA are generally considered incapable of inducing ventricular fibrillation, while capacitive discharges can be rendered entirely safe if the stored energy and discharge duration remain strictly below critical physiological thresholds.

These fundamental safety considerations are directly applicable to HASEL actuators, which share the exact same electrostatic operating principle and capacitive nature as DEAs. Recent work on a Peano HASEL based wrist tremor suppression device [65] explicitly evaluated electrical safety according to the IEC 60479-1 standard [29]. The authors reported a maximum stored energy of 98.2 mJ and a current hardware limited to 2 mA, both well below the corresponding safety limits established in their analysis. Such results definitively demonstrate that high voltage electrohydraulic actuators can be engineered to operate safely in wearable applications when appropriate electronic limitations are proactively adopted.

Nevertheless, accidental contact with energized electrodes must be physically prevented through adequate dielectric insulation. Several structural insulation strategies can be considered. A highly robust solution proposed by Shibuya et al. [64] consists of encapsulating the actuator completely within silicone layers, a technique successfully implemented for electrohydraulic actuators operating in underwater robotic applications.

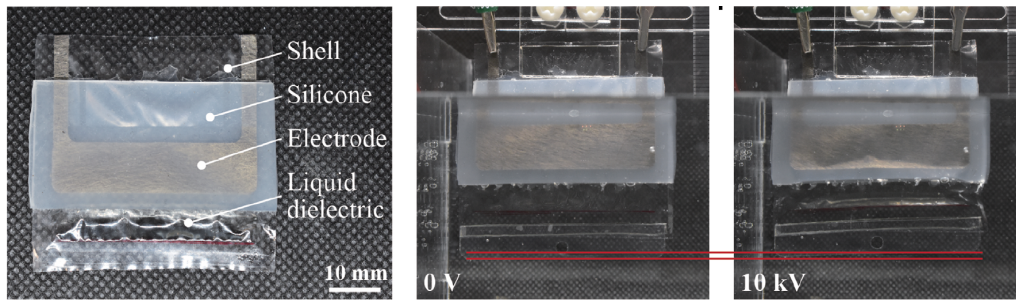


Figure 2.63: HASEL actuators encapsulated in an insulating silicone layer from [64].

Because underwater environments impose exceptionally stringent insulation requirements, this approach represents a highly promising and conservative solution for wearable devices subject to sweat or environmental moisture. Alternative methods include the application of spray on silicone coatings or the integration of electrically insulating films and tapes (such as Kapton tape), which offer the practical manufacturing advantage of simple self adhesive application. An even more conservative approach involves enclosing the entire actuation hardware within a sealed insulating textile sleeve or dielectric bag, thereby establishing a macroscopic physical barrier between all high voltage components and the

user.

Conversely, HALVE actuators operate at substantially lower voltages and are inherently safe to touch due to their fully encapsulated, non exposed electrodes. Although current HALVE technology is less mature than HASEL technology, and the resulting exosuit geometry is radially bulkier, these actuators organically eliminate many of the electrical safety challenges associated with high voltage systems, thereby drastically reducing the need for heavy external insulation. While the current developmental state of HALVE technology does not yet render it the most practical choice for the proposed exosuit prototype, future improvements in mechanical performance and robustness will position these actuators as an exceptionally attractive solution for wearable robotics, seamlessly combining structural compliance, low mass, ergonomic comfort, and intrinsically safe operation.

2.3. Control design

This section details the high level control modalities that the exosuit is capable of executing. In this context, a high level modality refers to the control strategy that dictates the desired human-robot interaction behavior during operation, remaining largely independent of the underlying hardware implementation.

Based on the physical interaction between the subject and the robotic device, Dalla Gasperina et al. [15] classify high level training modalities into four distinct control strategies:

- **Passive:** The robot performs the task autonomously without requiring voluntary effort from the subject, continuously correcting any trajectory errors;
- **Active assistive:** The robot and the subject perform the task cooperatively (e.g., the robot provides weight counterbalance or trajectory based corrective assistance);
- **Active:** The subject actively performs the task while the robot provides neither assistance nor resistance, operating with mechanical transparency;
- **Resistive:** The subject actively performs the task while the robot opposes the movement by providing resistive forces⁷.

⁷In Table A.1, devices that suppress tremor or deliver haptic feedback are classified as resistive.

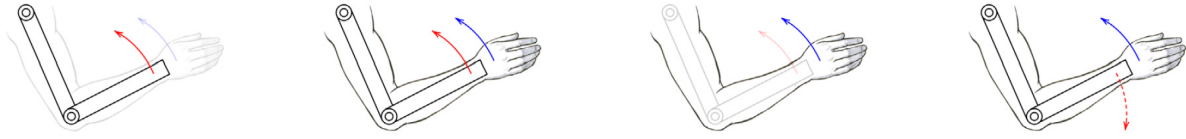


Figure 2.64: High level control strategies from [15], from left to right: passive, active assistive, active, resistive. Blue arrows indicate the subject’s voluntary effort; red arrows indicate robot assistance.

As established in the previous sections of this thesis, the proposed exosuit is specifically designed to implement both active assistive and passive control modes.

The active assistive mode aims to help patients with reduced wrist mobility successfully achieve the functional Range of Motion (RoM) required for Activities of Daily Living (ADLs). Because this mode is intended for continuous, all day assistance in everyday life, the 50% load compensation target discussed earlier is highly appropriate. This partial assistance ensures the user remains actively engaged in the movement while keeping the internal skeletal stress significantly lower than what would occur under full 100% compensation.

Conversely, the passive mode is intended primarily for structured clinical rehabilitation. In this modality, the exosuit drives the user through repeated, predefined movements for the duration of the therapy session. These movements can be defined either via a mathematically imposed trajectory or through a Teach and Play strategy (where a physical movement is first manually recorded by the device and subsequently replayed). Although passive mode strictly requires 100% load compensation to fully mobilize the joint, the associated skeletal risks are heavily mitigated because clinical rehabilitation sessions are conducted in highly controlled environments and last for a small fraction of the time compared to continuous all day wear.

2.3.1. General strategy

The fundamental control strategy of the exosuit relies on tracking the kinematics of the user’s wrist to provide appropriate motor assistance. To achieve this, the system utilizes Inertial Measurement Units. However, a single IMU placed on the hand is fundamentally insufficient for this application. A single sensor measures absolute orientation in global space, meaning it cannot distinguish between an isolated movement of the wrist and a movement of the entire arm. To accurately isolate the wrist joint angle, a second IMU is required. By placing this second sensor on the proximal forearm near the elbow,

a skeletal region that acts as a fixed reference frame during both flexion/extension and pronation/supination, the system can compute the relative orientation of the hand strictly with respect to the forearm.

Despite being conceptually straightforward, extracting a reliable relative angle from two independent IMUs introduces significant mathematical and hardware challenges, specifically regarding the choice of kinematic representation and the management of sensor drift.

Kinematic representation: Euler angles vs Quaternions

To represent 3D rotations, Euler angles are often the most intuitive choice, as they describe orientation through three consecutive rotations about predefined axes (roll, pitch, and yaw). However, they present severe mathematical limitations: the final orientation depends heavily on the chosen order of rotation, and they suffer from gimbal lock, a phenomenon where one degree of freedom is entirely lost when two axes align. Additionally, Euler angles are strictly bounded within a fixed mathematical range (e.g., $\pm 180^\circ$). When a rotation exceeds this limit, the value abruptly jumps to the opposite extreme, causing a discontinuity known as angle wrapping.

During the initial development of the flexion/extension control, a specific Euler angle was successfully isolated, and angle wrapping was mitigated through targeted software unwrapping techniques. However, applying this same Euler based approach to pronation/supination proved highly problematic. Human forearm rotation is not a pure single axis motion; it results from the complex kinematic rolling of the radius bone around the ulna, and its spatial trajectory is heavily influenced by the overall global orientation of the arm. Consequently, the effective axis of rotation is not fixed in the global reference frame. Because of this dynamic coupling, the exact same physical supination movement might be mathematically distributed across multiple different Euler angles depending on how the user is holding their arm in global space. While theoretically solvable through complex relative coordinate transformations, extracting a reliable pronation/supination angle from a predefined Euler sequence is computationally fragile and highly susceptible to sequence dependent errors.

To overcome these critical limitations, a unified quaternion based approach was adopted for both the flexion/extension and pronation/supination degrees of freedom. Quaternions provide a highly compact mathematical representation using four parameters that never suffer from gimbal lock or wrapping discontinuities. Although they are less intuitive to visualize, they are vastly superior for managing complex, coupled 3D motion. By

computing the relative quaternion rotation between the hand IMU and the forearm IMU with respect to a captured neutral calibration pose, the system obtains a stable, continuous representation of the wrist angle. This relative quaternion methodology is completely independent of the absolute global orientation of the sensors and inherently robust to coupled anatomical movements.

Sensor fusion and the drift problem

Translating this dual IMU mathematical model into a real world application revealed severe hardware challenges regarding sensor fusion. Standard IMUs (operating in NDOF mode) determine their orientation by fusing data from three internal sensors: the accelerometer finds the direction of gravity to establish pitch and roll, while the magnetometer identifies magnetic north to anchor the yaw heading.

However, in standard indoor environments, magnetometers perform exceptionally poorly. The presence of structural metals (such as steel rebar in concrete walls) and external electromagnetic fields heavily distorts the local magnetic field. This interference is severely amplified in the exosuit because the forearm IMU is mounted almost directly adjacent to the electromagnetic servo motors. Because the two IMUs experience different local magnetic distortions, their internal reference frames rapidly misalign. This frame misalignment manifests as a rapid relative drift; even if the physical arm remains perfectly still, the computed relative angle drifts uncontrollably, causing a perfectly functioning control loop to fail within thirty to sixty seconds.

To solve this critical heading estimation error, several iterative solutions were designed, tested, and evaluated:

1. **Periodic stillness recalibration:** The first attempted strategy was to automatically recalibrate the IMUs whenever the system detected that both sensors were completely still for a few seconds, assuming a stationary arm indicates a neutral resting posture. This approach possessed fatal clinical and functional flaws. Clinically, it completely fails for spastic patients (whose natural resting posture is highly contracted rather than neutral) and flaccid patients (whose hand passively drops when lifted). Functionally, if a user held a supinated posture for a few seconds to perform a static task (such as carrying a box), the system would incorrectly register that supinated angle as the new zero position, instantly ruining all subsequent angle computations.
2. **Accelerometer gravity tracking:** The second strategy attempted to bypass the magnetometer entirely by tracking the orientation of the hand relative to the fore-

arm strictly using the 3D gravity vectors measured by the accelerometers. While theoretically robust for static poses, this failed during dynamic motion. An accelerometer cannot naturally distinguish the static acceleration of gravity from the dynamic acceleration of human movement. Theoretically, this ambiguity could be partially mitigated by applying a low pass filter to the raw accelerometer data. Because gravitational pull presents as a constant low frequency signal, while dynamic human movements introduce high frequency transient spikes, a low pass filter can attempt to mathematically isolate the pure gravity vector. However, this signal separation is practically imperfect and highly complex to tune. Human biomechanical motion frequently occurs at low frequencies that overlap with the necessary filter cutoff threshold. Furthermore, aggressive signal filtering inherently introduces severe phase delays, creating a physical time lag that fundamentally compromises the instantaneous responsiveness required for a real time assistive control loop. Consequently, if the user simply lifted or lowered their arm, the dynamic acceleration still tricked the sensor algorithm into reading a false wrist angle, erroneously triggering the assistive servo motor.

3. **IMUPLUS mode:** The third and most successful approach was to completely switch the BNO055 sensors from their default NDOF mode to the IMUPLUS operating mode. This mode entirely disables the magnetometer, heavily relying on the accelerometer to correct pitch and roll while integrating data from the gyroscope to track yaw. This solution is 100% effective at eliminating magnetic drift but, because gyroscopes inherently possess microscopic mechanical biases, integrating their data causes a slow, gradual yaw drift over time. However, this gyroscope drift is exceedingly slow (becoming noticeable only after 10 to 15 minutes of continuous use), making it infinitely less problematic than rapid magnetic drift. For the short duration typical of a rehabilitation session, this trade off is highly acceptable. For longer continuous usage, a manual recalibration command can simply be sent to the system.
4. **Custom sensor fusion:** As a final theoretical consideration to eliminate the slow gyroscope drift, the possibility of implementing a custom sensor fusion architecture was evaluated. The conceptual solution involves bypassing the unreliable physical magnetic environment by imposing an artificially defined "fake north" via software. This artificial vector does not need to align with the actual magnetic north; it simply needs to provide a shared, mathematically identical reference direction for both IMUs. If both sensors anchored their yaw calculations to this exact same virtual vector, relative drift would be completely prevented. However, because the

BNO055 is a closed proprietary sensor, it is impossible to inject external reference vectors into its internal microcontroller. To implement this fake north, one would theoretically have to bypass the internal sensor fusion entirely, extract the raw accelerometer and gyroscope data, and run a custom sensor fusion algorithm (such as a Madgwick or Mahony filter) directly on the ESP32 microcontroller. This theoretical approach was ultimately discarded because developing a custom filter that matches the dynamic performance of the integrated BNO055 processor is practically unfeasible. The proprietary internal Kalman filter, engineered by Bosch, is heavily optimized specifically for human biomechanical motion. It possesses sophisticated dynamic tuning that intelligently weights gyroscope data to filter out the different transient linear accelerations experienced by the hand and the elbow during rapid arm lifting. Replicating this level of highly tuned proprietary filtering from scratch would inevitably result in poor dynamic stability, where normal arm movements would cause severe false angle readings and erroneous servo triggers. Consequently, accepting the light and highly predictable gyroscope drift of the integrated IMUPLUS mode remains the superior and safer engineering choice.

Mathematical framework: relative quaternions and orthogonalization

Because the mathematical tracking of the wrist relies on two independent sensors mounted onto flexible fabric, the system must digitally compensate for physical mounting misalignments, anatomical cross talk, and complex 3D rotations. To achieve this, a universal kinematic framework was developed. This framework processes IMU data through a strict sequence of quaternion operations to output pure, perfectly isolated single axis angles. This mathematical pipeline runs continuously and acts as the foundational engine for all high level control modalities.

Phase 1: The global reference frame and relative rotation When the IMUs boot in IMUPLUS mode, their internal sensor fusion algorithms lock their pitch and roll axes to the physical Earth gravity vector, while their yaw axes initialize based on their starting orientation. Because the sensors are initialized simultaneously while stationary, they share an identical 3D global coordinate system.

To determine the spatial relationship between the hand and the forearm, the system computes a relative quaternion. Let q_{forearm} and q_{hand} represent the absolute orientations of the two sensors in the global frame. To understand the mathematical logic behind extracting their relative difference, quaternions must be viewed as sequential spatial transformations.

The quaternion q_{hand} represents the transformation from the global origin to the hand, while q_{forearm} represents the transformation from the global origin to the forearm.

To compute the direct path from the forearm to the hand, the system must mathematically rotate backwards from the forearm into the shared global frame, and then forwards into the hand frame. This reverse transformation is achieved by taking the quaternion inverse (q^{-1}). Thus, the combined relative sequence is $q_{\text{forearm}}^{-1} \otimes q_{\text{hand}}$. Because the IMUs internally normalize their tracking data to output strictly unit quaternions, their mathematical inverse is computationally identical to their conjugate (q^*). Consequently, the relative rotation q_{rel} is efficiently evaluated as:

$$q_{\text{rel}} = q_{\text{forearm}}^* \otimes q_{\text{hand}} \quad (2.63)$$

This operation effectively makes the forearm the static origin of the universe, with q_{rel} describing exactly how the hand is oriented relative to it.

Phase 2: The neutral mounting offset When rigid electronic housings are sewn onto a flexible textile exosuit, perfect mechanical alignment is physically impossible; the hand sensor might be slightly twisted, and the forearm sensor might be slightly tilted. To correct this, the user is instructed to hold their hand in a flat, neutral resting posture while the system captures the current relative rotation and stores it as q_{neutral} .

This stored quaternion acts as a digital mounting offset. For all subsequent real time computations, the system mathematically subtracts this offset by multiplying the conjugate of the neutral pose by the live relative rotation to obtain the true, zeroed joint rotation (q_{joint}):

$$q_{\text{joint}} = q_{\text{neutral}}^* \otimes q_{\text{rel}} \quad (2.64)$$

This operation guarantees that when the user returns to their physical resting position, q_{joint} perfectly evaluates to the identity quaternion ($w = 1, x = 0, y = 0, z = 0$), artificially creating a flawless mechanical alignment.

Phase 3: Biological axis extraction The human wrist is not a perfect mechanical hinge aligned to standard Cartesian planes. When a human extends their wrist or supinates their forearm, the joint rotates around a highly specific, anatomically unique 3D vector.

To capture these biological hinges, the user is asked to perform a full wrist extension and hold the position. The system evaluates q_{joint} at this extended state, denoted as $q_{\text{calib, FE}}$.

To understand how the physical hinge is extracted from this measurement, it is necessary to recall that a unit quaternion is fundamentally defined by a rotation angle θ around a 3D unit vector $\mathbf{v}$:

$$q = \left(\cos \frac{\theta}{2}, \mathbf{v} \sin \frac{\theta}{2} \right) = (w, x, y, z) \quad (2.65)$$

This mathematical definition demonstrates that the x , y , and z components of the quaternion contain the true rotation axis $\mathbf{v}$, scaled by a factor of $\sin(\frac{\theta}{2})$. Therefore, by extracting just the x , y , and z components of $q_{\text{calib, FE}}$ and mathematically normalizing them (which cleanly cancels out the sine scale factor), the system directly isolates the pure 3D spatial vector $\mathbf{v}_{\text{FE}}$ representing the user's specific biological flexion/extension hinge. This exact process is repeated during a supination movement to extract the biological pronation/supination axis, $\mathbf{v}_{\text{PS}}$.

Phase 4: Gram-Schmidt orthogonalization Extracting the biological axes reveals a critical biomechanical reality: human wrist movements are naturally coupled. When a person fully extends their wrist, the forearm naturally supinates a tiny amount. Consequently, the captured $\mathbf{v}_{\text{FE}}$ and $\mathbf{v}_{\text{PS}}$ vectors are rarely perfectly orthogonal. If left uncorrected, moving in purely one direction would generate a false mathematical error in the other direction, causing the active controllers to aggressively fight each other.

To mathematically eliminate this anatomical cross talk, a Gram-Schmidt orthogonalization process is applied. The system designates the pronation/supination axis ($\mathbf{v}_{\text{PS}}$) as the strict reference vector. This specific hierarchy was deliberately selected based on functional priority and user comfort. Unwanted wrist flexion/extension occurring during a pure forearm rotation task is functionally far more disruptive to Activities of Daily Living than a slight unwanted forearm twist during wrist flexion. By utilizing the pronation/supination axis as the mathematical anchor, the system guarantees that a pure twisting movement does not erroneously trigger a disruptive flexion/extension motor response. The algorithm then projects the flexion/extension axis ($\mathbf{v}_{\text{FE}}$) onto $\mathbf{v}_{\text{PS}}$ using the dot product to find their exact mathematical overlap. By subtracting this overlapping "shadow" from the original flexion/extension axis, the system generates a perfectly orthogonalized vector, $\mathbf{v}_{\text{FE, ortho}}$:

$$\mathbf{v}_{\text{FE, ortho}} = \mathbf{v}_{\text{FE}} - (\mathbf{v}_{\text{FE}} \cdot \mathbf{v}_{\text{PS}})\mathbf{v}_{\text{PS}} \quad (2.66)$$

This spatial correction strictly decouples the axes, mathematically forcing the flexion/extension software channel to be blind to any pronation/supination movements.

Phase 5: Swing-twist angle decomposition With the biological axes perfectly isolated and orthogonalized, the final step is to extract a human readable angle in degrees from the live 4D quaternion data (q_{joint}) every control loop.

Instead of executing complex and unstable conversions to Euler rotation matrices, the system utilizes a swing twist decomposition. To isolate only the rotation (the twist) that aligns perfectly with a specific biological axis $\mathbf{v} = (v_x, v_y, v_z)$, the system projects the vector portion of the live quaternion onto the biological axis using a standard dot product:

$$t = q_{\text{joint},x}v_x + q_{\text{joint},y}v_y + q_{\text{joint},z}v_z \quad (2.67)$$

This t value represents the magnitude of rotation strictly along the target hinge, mathematically discarding all off axis rotation (the swing). Finally, this isolated magnitude is converted back into a standard angle $\tilde{\theta}$ in degrees using the quaternion twist formula ⁸:

$$\tilde{\theta} = 2 \arctan 2(t, q_{\text{joint},w}) \cdot \left(\frac{180}{\pi} \right) \quad (2.68)$$

This robust operation yields a continuous, perfectly decoupled angle for both flexion/extension and pronation/supination. These pure angular values are subsequently fed into the various control loops to drive the servo motors according to the specific active or passive modality.

2.3.2. Active assistive mode

The primary clinical objective of the active assistive mode is to support patients who possess residual, yet insufficient, voluntary motor function. In this modality, the robotic exosuit does not dictate the movement; rather, the user actively initiates the physical task. The system continuously monitors the user's kinematic intention and provides proportionally amplified motor assistance. This cooperative human robot interaction is meant to help the user overcome structural joint stiffness and muscular weakness to successfully achieve the functional RoM required for everyday ADLs.

Control logic

The active assistive controller relies directly on the continuous stream of decoupled angular data generated by the kinematic framework detailed in Section 2.3.1. Because human

⁸Throughout this control architecture, the tilde notation ($\tilde{\cdot}$) is used to denote variables that have been mathematically decoupled or mapped into a user specific functional space, distinguishing them from raw hardware or anatomical limits.

intention is highly dynamic, the control law must be exceptionally responsive while remaining completely stable against anatomical cross talk.

This stability is guaranteed by the prior Gram-Schmidt orthogonalization. By strictly feeding the active assistive loop with angles derived from the isolated $\mathbf{v}_{\text{FE, ortho}}$ and $\mathbf{v}_{\text{PS}}$ biological axes, the system ensures that when the user exerts a pure voluntary effort in one anatomical direction, the assistance provided is completely isolated to the flexion/extension servo. The pronation/supination servo remains entirely undisturbed, physically preventing the actuators from fighting the user or each other. Conversely, if the user deliberately initiates a combined movement spanning both degrees of freedom, the control architecture seamlessly drives both servo motors in parallel. This simultaneous actuation enables the natural execution of complex multi joint tasks without mechanical restriction.

For each specific degree of freedom, the live decoupled angle $\tilde{\theta}$ is processed through a precise sequence of mathematical operations every 20 milliseconds to compute the final servo command.

Step 1: Anatomical normalization First, the live angle $\tilde{\theta}$ is bounded by predefined, hardware specific anatomical limits ($\theta_{\min}$ and $\theta_{\max}$) set to the extremes of the functional RoM of ADLs. It is then mathematically mapped to a dimensionless ratio $\tilde{N}$, bounded strictly between 0 and 1:

$$\tilde{N} = \frac{\tilde{\theta} - \theta_{\min}}{\theta_{\max} - \theta_{\min}} \in [0, 1] \quad (2.69)$$

This operation converts the absolute biomechanical angle into a standardized fraction of the available functional RoM.

Step 2: Power curve During early development, a simple proportional linear mapping was utilized to translate the functional range of motion of the hand directly to the 180° range of the servo motor. However, empirical testing revealed that a purely linear proportional gain only provided perceptible assistance toward the end of the movement. This insufficiency during the critical early phase of motion is driven by three primary biomechanical and mechanical factors. First, initiating a physical movement is inherently the most demanding phase due to high passive resistance from biological muscles, tendons, and ligaments. Second, when the wrist is in its neutral, horizontal resting position, the gravitational moment generated by the mass of the hand is at its absolute maximum, since the center of mass sits at the largest perpendicular distance from the joint rotation axis. Third, the mechanical advantage of the tendon driven prototype is not constant

throughout the motion, as shown in Figure 2.65; the effective lever arm is minimized at the onset of motion and strictly increases toward the middle of the extension range, meaning the actuator must exert significantly more initial force to generate the same effective joint torque.

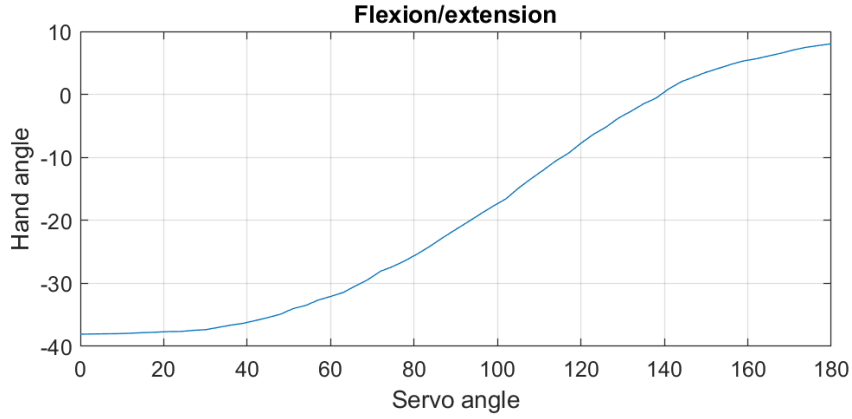


Figure 2.65: Relationship between hand angle and servo angle during wrist flexion/extension. The circular motion of the servo arm dictates that only one component of its rotation contributes to the linear tendon displacement, resulting in a harmonic hand elevation profile that minimizes the effective lever arm at the onset of motion.

To mathematically compensate for these combined initial resistances, the linear mapping was discarded in favor of a non linear power curve. The normalized ratio $\tilde{N}$ is raised to an empirically tuned exponent of 0.7 before being mapped to the physical range of the servo:

$$P_{\text{pos}} = \tilde{N}^{0.7} \cdot 180^\circ \quad (2.70)$$

Because the normalized input $\tilde{N}$ lies strictly within the interval $[0, 1]$, applying an exponent strictly less than 1 generates a concave mathematical function. This curve aggressively amplifies very small initial input angles while smoothly attenuating the amplification near the end of the range. Consequently, when the user actively initiates a movement, a negligible initial bending of the wrist instantly requests a disproportionately massive positional displacement from the servo. This creates a highly responsive assistance profile that perfectly neutralizes the high initial biomechanical and mechanical resistances of the system.

Step 3: Velocity based intention detection To further enhance the fluidity of the human robot interaction, a dynamic derivative term is incorporated. The system

calculates the instantaneous angular velocity ω ⁹ by subtracting the previously recorded angle from the current live angle. This velocity is multiplied by an empirically tuned velocity factor K_v to generate a dynamic positional boost P_{vel} :

$$P_{\text{vel}} = \omega \cdot K_v \quad (2.71)$$

This term functions simultaneously as a digital shock absorber and an intention detector. This dynamic response is particularly crucial during the initiation of movement, a phase previously identified as biomechanically demanding. Because the velocity instantaneously transitions from zero to a positive value the exact moment the user begins to move, this derivative term injects an immediate assistive surge that directly helps the user overcome initial static resistance. Furthermore, if the user decides to move their hand very rapidly, the velocity value spikes, adding a massive instantaneous boost to the target servo position; essentially, the motor accelerates to get out of the user's way and prevent the sensation of dragging heavy hardware. Conversely, if the user stops moving to hold a static posture, the velocity drops precisely to zero, and the motor seamlessly settles into a smooth, steady holding pattern based purely on the positional curve.

Step 4: Signal smoothing and actuation The positional curve and the dynamic velocity boost are summed to create a raw target angle. However, feeding this highly reactive raw signal directly into the servo motor would result in erratic, high frequency jitter that is physically uncomfortable for the user. To guarantee a smooth mechanical output, the raw target is passed through an Exponential Moving Average (EMA) low pass filter:

$$S_{\text{filtered},t} = \alpha (P_{\text{pos}} + P_{\text{vel}}) + (1 - \alpha)S_{\text{filtered},t-1} \quad (2.72)$$

where α is the smoothing coefficient (empirically tuned to 0.2) and $S_{\text{filtered},t-1}$ is the servo command from the previous control cycle. This filtered angle represents the final, smoothed positional target that is dispatched directly to the servo motor hardware.

The complete control architecture for the active assistive mode is visually summarized in Figure 2.66.

⁹Actually, the computed term ω represents an angular difference between consecutive control loops rather than a true angular velocity, as it is not explicitly divided by the time interval. However, because the microcontroller executes the control loop at a highly consistent frequency (every 20 milliseconds), this angular delta is directly proportional to the true velocity and functions identically within the mathematical logic.

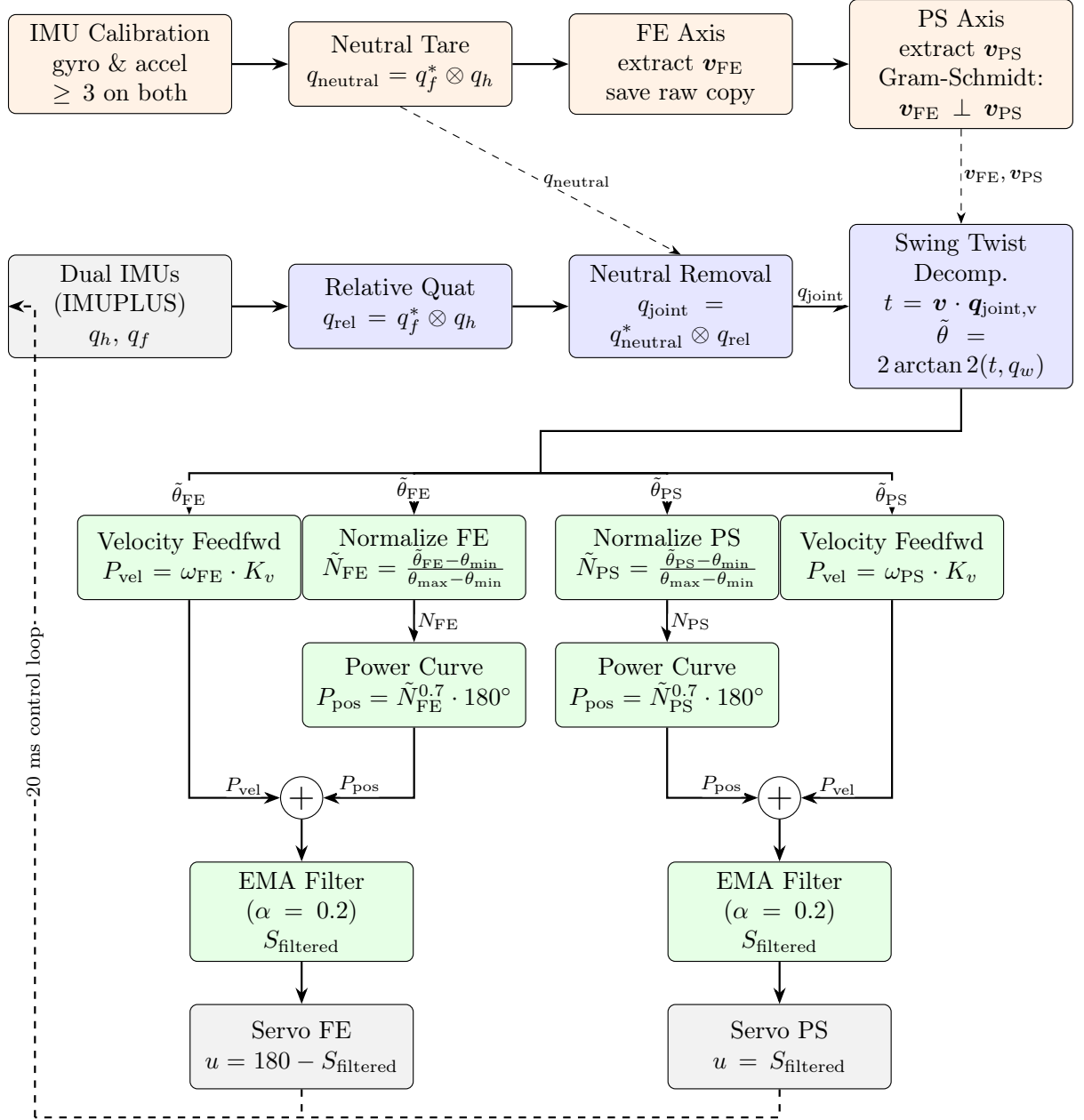


Figure 2.66: Control architecture for the active assistive mode. *Orange:* offline calibration phase capturing the neutral posture and biological axes with Gram-Schmidt orthogonalization ($v_{\text{FE}} \perp v_{\text{PS}}$). This calibration sequence is managed via a graphical user interface, detailed in Section 2.3.4. *Blue:* continuous 50 Hz kinematics pipeline executing neutral removal and swing twist angle decomposition for both degrees of freedom. *Green:* parallel control columns that normalize the isolated angles, apply a non linear power curve and velocity feedforward term, and smooth the signal using an EMA low pass filter before actuation. *Note:* the flexion/extension servo command is mathematically inverted ($180^\circ - S_{\text{filtered}}$) to account for the physical mounting orientation of the actuator.

2.3.3. Passive mode

In passive mode, the user remains completely relaxed while the robotic exosuit assumes full control of the joint kinematics, driving the wrist through continuous movements to prevent tissue contracture, stimulate neuroplasticity, and maintain healthy joint mobility.

Mapping phase

A critical prerequisite for any automated passive control in a tendon driven system is the execution of an accurate spatial mapping phase. If the exosuit featured rigid mechanical linkages, physically moving the user's hand in a transparent state would mechanically back drive the motors in both directions, making the registration of specific joint angles easy. However, tendons operate strictly under tension; they can pull, but they cannot push. If a therapist manually moves the hand in the direction opposite to the established tendon tension, they simply fold, leaving the motor completely stationary.

Consequently, to establish a reliable and safe correspondence between the physical angle of the hand and the absolute position of the servo motor, an automated mapping phase must be executed. The system incrementally steps the motors through a controlled sweep while continuously reading the decoupled IMU data. This routine mathematically links the hardware specific 0 to 180 degree range of the actuators directly to the physical range of the exosuit.

Automated passive trajectory

Once the mapping phase is complete, the system can execute automated passive trajectories. The exosuit continuously dictates the position of the motors to force the hand to follow a mathematically generated angular profile. While the current prototype implementation utilizes a synthetic harmonic curve to validate the responsiveness and safety of the control architecture, this software framework is designed to be highly modular. In a clinical application, medical professionals could easily substitute this synthetic curve with specific, evidence based rehabilitation trajectories tailored exactly to the therapeutic needs of the patient.

To enforce this trajectory accurately despite the inherent friction and elasticity of the tendons and the possible movements of the user, the control software employs a combination of empirical feedforward prediction and closed loop feedback correction.

Feedforward and temporal lead During every 20 millisecond control cycle, the system calculates the desired target angle. To proactively compensate for inherent mechan-

ical slack and hardware communication delays, the algorithm samples the mathematical trajectory with a 250 millisecond lead time. This anticipatory target angle is then passed directly through the empirical lookup table generated during the mapping phase, immediately yielding a baseline open loop positional command for the servo motors.

Dual read orthogonalization and PI feedback Because tendon driven systems suffer from unpredictable hysteresis, the open loop command alone cannot guarantee perfect trajectory tracking. Therefore, a Proportional Integral (PI) controller continuously monitors the live IMU data to apply dynamic micro corrections. To manage anatomical cross talk without distorting the absolute tracking accuracy, this feedback loop employs a unique dual read kinematic architecture.

During each 20 millisecond control cycle, the software evaluates the live relative orientation twice. First, it extracts the joint angle using the raw, non orthogonalized biological axes. This raw reading, denoted as $\tilde{\theta}_{\text{raw}}$, is used exclusively to calculate the instantaneous equilibrium error:

$$e_{\text{raw}} = \tilde{\theta}_{\text{target}} - \tilde{\theta}_{\text{raw}} \quad (2.73)$$

Bypassing the Gram-Schmidt spatial correction for this calculation guarantees that the system evaluates the true absolute position of the hand relative to its anatomical plane, completely preventing the introduction of systematic mathematical bias.

Second, the pipeline extracts the angle utilizing the strictly orthogonalized axes ($\mathbf{v}_{\text{FE, ortho}}$ and $\mathbf{v}_{\text{PS}}$). This decoupled angle, $\tilde{\theta}_{\text{ortho}}$, is used to calculate the orthogonalized error:

$$e_{\text{ortho}} = \tilde{\theta}_{\text{target}} - \tilde{\theta}_{\text{ortho}} \quad (2.74)$$

This value is fed exclusively into the integral accumulator. By strictly utilizing orthogonalized data for the integral term, the software ensures that error accumulation in one degree of freedom remains blind to perturbations in the other. The final commanded angle, $\tilde{\theta}_{\text{adj}}$, is computed as the sum of the feedforward target and the dual error terms:

$$\tilde{\theta}_{\text{adj}} = \tilde{\theta}_{\text{target}} + (e_{\text{raw}} \cdot K_p) + \left(\int e_{\text{ortho}} dt \cdot K_i \right) \quad (2.75)$$

This effectively decouples the active feedback response, preventing the flexion/extension and pronation/supination tracking algorithms from mathematically fighting each other during combined trajectory execution.

The complete control architecture for the passive mode is summarized in Figure 2.67.

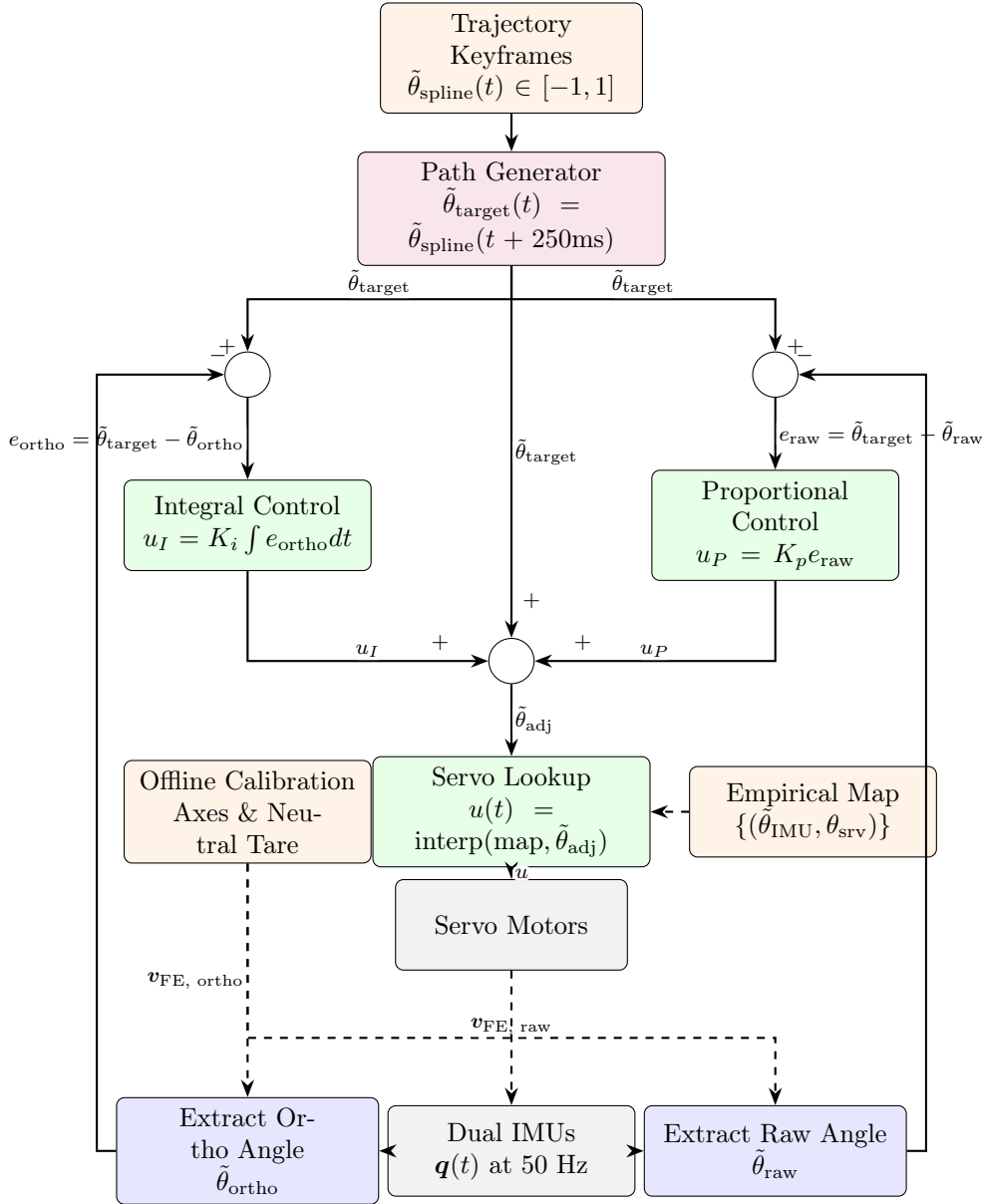


Figure 2.67: Control architecture for the automated passive trajectory mode. *Orange and red*: offline setup phase including the anatomical calibration, empirical mapping, and generation of the 250 millisecond feedforward target trajectory. *Blue*: dual read kinematics pipeline operating at 50 Hz, extracting both the raw anatomical angle and the Gram-Schmidt orthogonalized angle. *Green*: Proportional Integral feedback loop with a baseline feedforward term. The raw error calculates the absolute equilibrium correction, while the orthogonalized error drives the stable integral accumulation. All three components are summed together, and the adjusted target angle ($\tilde{\theta}_{\text{adj}}$) is resolved through the empirical lookup table to generate the final servo command.

Teach and Play mode

To offer a more personalized and versatile rehabilitation approach, a Teach and Play modality was developed. During the *Teach* phase, the therapist physically grasps the patient's hand and manually guides it through a specific, customized spatial movement. The dual IMU kinematic pipeline continuously records the pure decoupled angular states at a high frequency. During the *Play* phase, the control architecture translates these stored angular arrays into synchronized motor commands, allowing the exosuit to replicate the exact taught movement autonomously. This architecture heavily reduces the physical burden on the therapist while allowing the patient to receive the maximum prescribed dosage of repetitive therapy.

Teach phase To capture the exact nuances of the therapist's manual guidance, the recording phase operates at an accelerated 100 Hz execution loop. As the wrist is physically manipulated, the system continuously processes the live relative quaternion. During this recording state, it is absolutely critical that the extracted angles represent pure biological motion rather than corrupted multi joint data. Therefore, the kinematic pipeline extracts the live position using the strictly orthogonalized biological axes ($\mathbf{v}_{\text{FE, ortho}}$ and $\mathbf{v}_{\text{PS}}$).

As the system records the motion, it saves the decoupled, orthogonalized angular state $\tilde{\theta}_{\text{ortho}}$ and the relative timestamp Δt at each 100 Hz sampling interval i :

$$\text{Buffer}_i = \{\tilde{\theta}_{\text{ortho},i}, \Delta t_i\} \quad (2.76)$$

This array is capable of storing up to 20 seconds of highly complex, continuous multidimensional movement, ensuring that every microscopic adjustment made by the therapist is perfectly quantified and mathematically isolated.

Play phase Once the sequence is successfully captured, the system transitions to the Play phase, dropping to a stable 50 Hz execution loop to drive the servo motors. Because the recording frequency (100 Hz) and the playback control frequency (50 Hz) differ, stepping through the array index by index would artificially slow the movement down. To resolve this, the control architecture utilizes a dynamic time tracking algorithm. During every 20 millisecond playback cycle, the software checks the exact elapsed time and searches the memory buffer. If the exact millisecond timestamp does not align perfectly with a recorded frame, a continuous linear interpolation function mathematically calcu-

lates the exact intermediate angle between the two closest recorded frames.

$$\tilde{\theta}_{\text{target}}(t) = \tilde{\theta}_{\text{ortho},i} + \left(\frac{t - \Delta t_i}{\Delta t_{i+1} - \Delta t_i} \right) (\tilde{\theta}_{\text{ortho},i+1} - \tilde{\theta}_{\text{ortho},i}) \quad (2.77)$$

This guarantees a flawless temporal mapping, allowing the exosuit to replay the complex movement at the exact real time speed it was originally taught.

To execute these dynamic targets, the software utilizes the empirical mapping array generated during the initial calibration phase. The taught human angles are passed through this lookup table to generate the baseline open loop feedforward commands for both the flexion/extension and pronation/supination servo motors simultaneously.

Because the therapist frequently teaches complex combined movements, the physical mechanical cross talk between the artificial tendons is severely amplified during playback. To guarantee flawless trajectory tracking and compensate for dynamic tendon slack, the Play phase employs an independent Proportional Integral (PI) feedback architecture.

It is crucial to note that this phase diverges from the dual read architecture utilized in the automated passive trajectory. In the passive mode, the targets are strict numerical anatomical limits; using the raw, non orthogonalized axes for the proportional error guarantees the hand physically reaches these true absolute limits without mathematical distortion. Conversely, in the Teach and Play modality, the target is not a theoretical absolute angle, but rather a dynamic array of decoupled spatial states recorded directly in the orthogonalized domain. To track this dynamic target, the PI controller continuously calculates the orthogonalized error e_{ortho} and computes the final servo command u by summing the feedforward map value with the feedback corrections:

$$e_{\text{ortho}} = \tilde{\theta}_{\text{target}}(t) - \tilde{\theta}_{\text{ortho, live}} \quad (2.78)$$

$$u(t) = \text{interp} \left(\text{map, } \underbrace{\tilde{\theta}_{\text{target}}(t)}_{\text{Feedforward}} + \underbrace{(e_{\text{ortho}}(t) \cdot K_p)}_{\text{Proportional}} + \underbrace{\left(\int e_{\text{ortho}}(t) dt \cdot K_i \right)}_{\text{Integral}} \right) \quad (2.79)$$

Therefore, the playback cycle evaluates the live spatial orientation using exclusively the Gram-Schmidt orthogonalized axes ($\mathbf{v}_{\text{FE, ortho}}$ and $\mathbf{v}_{\text{PS}}$). By comparing the orthogonalized live reading directly against the orthogonalized recorded target, the system performs a mathematically pure comparison. This prevents the independent PI controllers from attempting to "correct" natural anatomical cross talk, a scenario that would otherwise

cause the dual servos to violently fight each other. By strictly utilizing isolated data for the entire closed loop calculation, the software allows the actuators to cooperatively replay the exact visual path taught by the therapist.

Geometric decoupling

Previous sections established that the natural biological axes of wrist flexion/extension and pronation/supination are inherently coupled. The offline calibration phase identifies these pure, user specific biological axes, which are subsequently orthogonalized to create a perfectly decoupled mathematical reference frame. However, the physical mounting of the exosuit actuators can never perfectly align with these idealized anatomical axes.

This physical misalignment is driven by three primary factors. First, due to strict spatial constraints on the forearm sleeve, the flexion/extension servo is mounted slightly off center, meaning tension on this artificial tendon inherently induces a slight pronation moment. Second, anatomical geometry varies significantly between individuals, making it impossible for fixed motor mounts to align perfectly with the biological axes of every user. Third, the inherent compliance of the soft textile exosuit dictates that the hardware anchor points shift slightly every time the device is worn.

Consequently, actuating a single servo never produces a pure, isolated movement along the mathematical orthogonalized axes nor along the natural pure axes recorded during calibration. This geometric reality formally justifies why advanced modalities like Teach and Play must continuously command both servos simultaneously. Replicating even a purely isolated anatomical extension requires the pronation/supination servo to actively adjust its tension to counteract the physical hardware misalignment and enforce the true biological path and vice versa.

To analytically quantify this geometric discrepancy between the physical actuator pulling vectors and the calibrated biological axes, a dedicated Python diagnostic script was developed. This testing algorithm commands the microcontroller to execute a controlled, automated sweep of one primary servo across its maximum functional range while strictly holding the secondary servo at its previously mapped neutral position. During this physical sweep, the script continuously reads the high frequency serial stream, logging both the primary (driven) joint angle and the secondary parasitic (undriven) joint angle.

By analyzing the captured arrays, the algorithm calculates the absolute Range of Motion (RoM) for both degrees of freedom. It then mathematically defines the mechanical cross talk as a strict percentage, calculated as the ratio of the parasitic RoM to the primary RoM. To aid in both clinical and engineering evaluation, the software automatically ex-

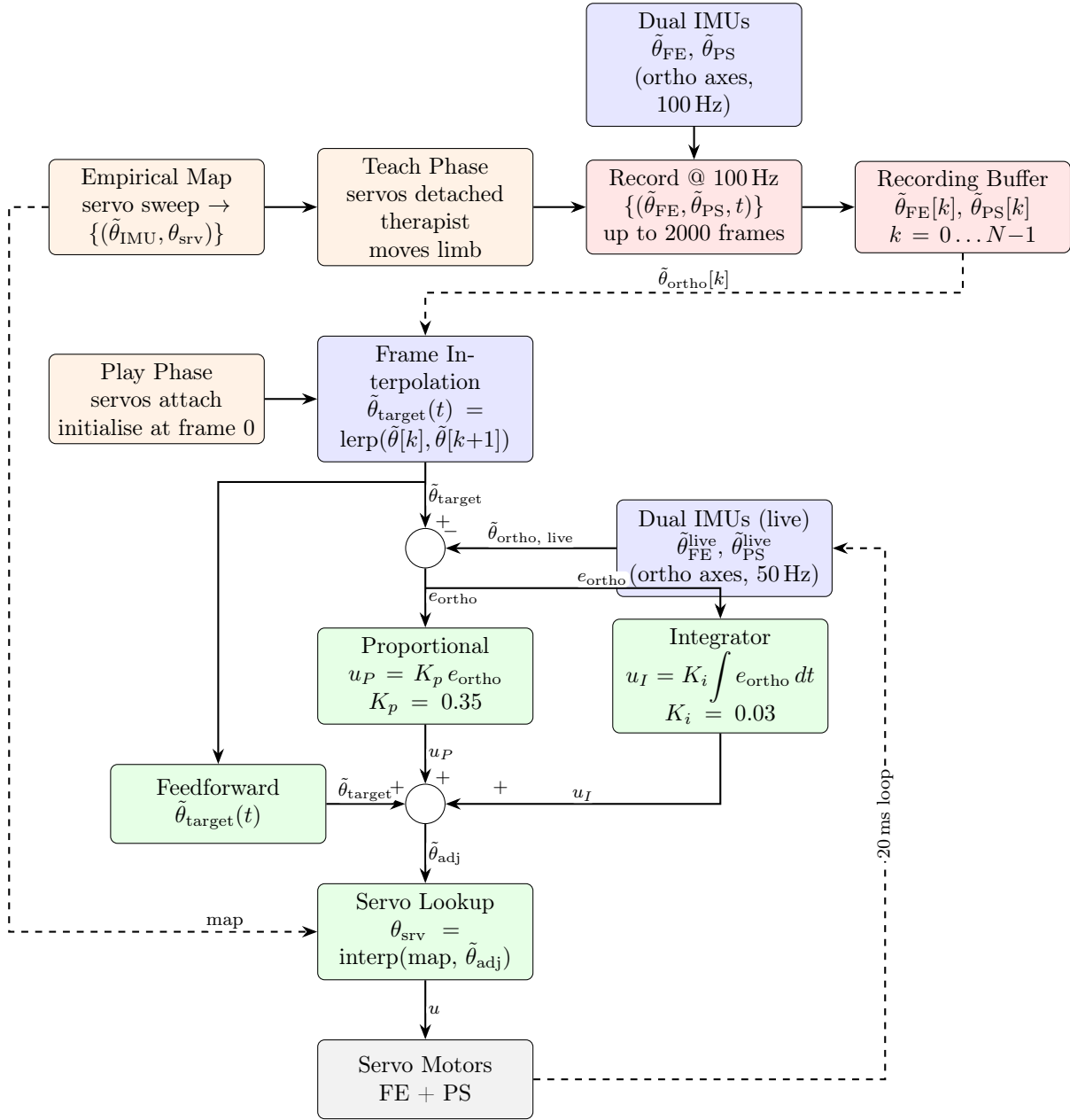


Figure 2.68: Teach & Play mode. *Top row (orange, red) – Teach phase:* the empirical servo angle↔wrist angle map is acquired once via a full ROM sweep. The servos are then detached so the therapist can move the patient’s limb freely; joint angles $\tilde{\theta}_{\text{FE}}$ and $\tilde{\theta}_{\text{PS}}$ are sampled at 100 Hz on orthogonalised axes and stored into a 2000-frame timestamped buffer. *Bottom rows (blue, green) – Play phase:* the servos reattach and are driven to frame 0. At each 20 ms tick the buffer is linearly interpolated to produce the reference trajectory $\tilde{\theta}_{\text{target}}(t)$. A standard PI controller closes the loop on live ortho-axis IMU feedback: the proportional branch ($K_p = 0.35$) and the integral branch ($K_i = 0.03$) produce corrections u_P and u_I which are summed with the feedforward reference $\tilde{\theta}_{\text{target}}$. The resulting corrected target $\tilde{\theta}_{\text{adj}}$ is resolved through the empirical lookup table to generate the final servo command.

ports the raw numerical data to a CSV file and generates two key visual diagnostics: a time series plot illustrating the simultaneous angular displacements, and a spatial scatter plot mapping the exact true trajectory of the physical tendon relative to the pure biological planes.

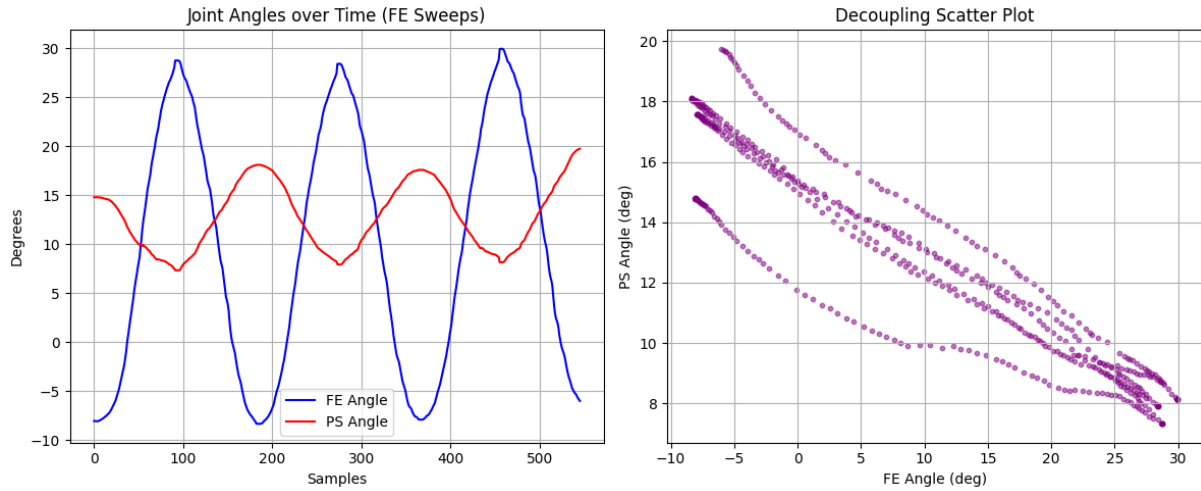


Figure 2.69: Parasitic pronation induced by the off center mounting of the flexion/extension servo. Conventionally, supination is recorded as positive and pronation as negative; therefore, the downward trajectory of the secondary curve during active extension clearly illustrates how the primary movement is physically coupled with a secondary pronation moment.

2.3.4. Graphic User Interface (GUI)

While the low level kinematic processing and motor control loops execute autonomously on the ESP32 microcontroller, the exosuit requires a centralized software hub to manage serial communication, guide the clinical operator through the mandatory physical setup, and trigger the various high level control modalities. To achieve this, a custom Graphical User Interface (GUI) was developed using Python.

The primary design objective of this interface is to abstract the underlying mathematical complexity of the system, providing the therapist with a clean, user friendly visual environment that ensures safety and procedural consistency during clinical rehabilitation sessions.

Architecture and serial communication

The foundation of the GUI is built upon a robust, bidirectional serial communication architecture. Upon launch, the software automatically scans the available hardware ports

and allows the operator to establish a high baud rate connection with the microcontroller. Once connected, the interface acts as a continuous real time monitor. It receives and parses the high frequency data stream from the dual IMUs, displaying live system status, sensor health, and calibration readiness. Conversely, it translates the therapist's button presses into specific, encoded command bytes that instantly trigger state changes within the microcontroller's execution loop.

The calibration protocol

As detailed in Section 2.3, the system cannot operate safely without first mapping the specific biological axes of the user and establishing a neutral resting baseline. To prevent operator error, the GUI forces the user through a strict, sequential calibration protocol before any motor control is unlocked.

This sequence is visually structured as a step by step graphical pipeline. First, the interface requests the name or identification of the patient. This data entry is essential, as all subsequent diagnostic routines and evaluation protocols automatically append the subject's identifier, alongside timestamp data, to exported data arrays and generated plots, ensuring rigorous data management.

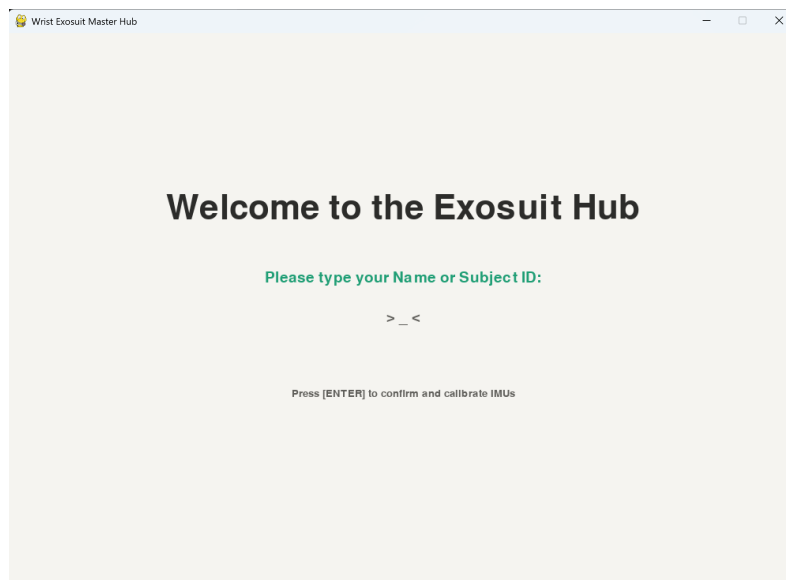


Figure 2.70: Initial user interface prompt for subject identification and data management.

Then, the interface displays the live gyroscope and accelerometer calibration statuses of both the hand and forearm IMUs, refusing to proceed until the hardware achieves maximum internal confidence (a value of 3/3 for both the gyroscope and accelerometer across both IMUs).

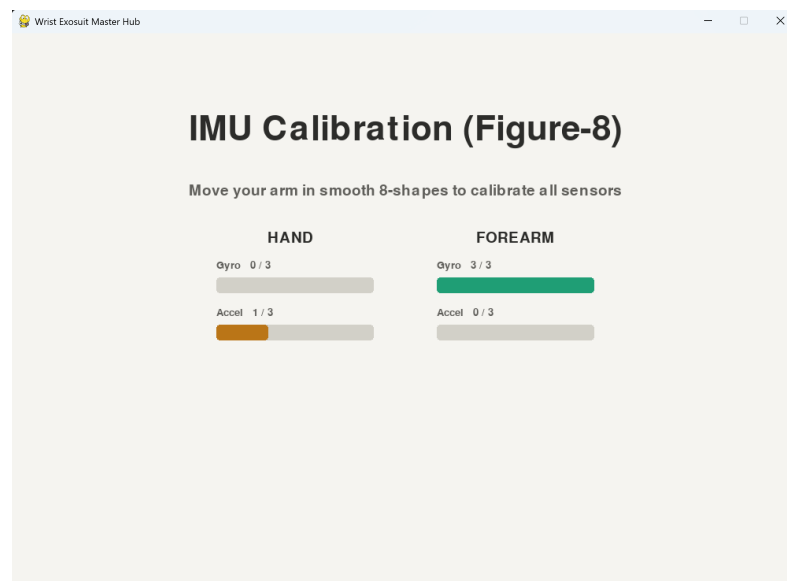


Figure 2.71: Hardware calibration interface. Dynamic visual indicators guide the operator by displaying the internal calibration confidence of the gyroscope and accelerometer for both IMUs.

Next, the interface prompts the therapist to hold the patient’s hand in a flat resting pose to capture the neutral quaternion offset.

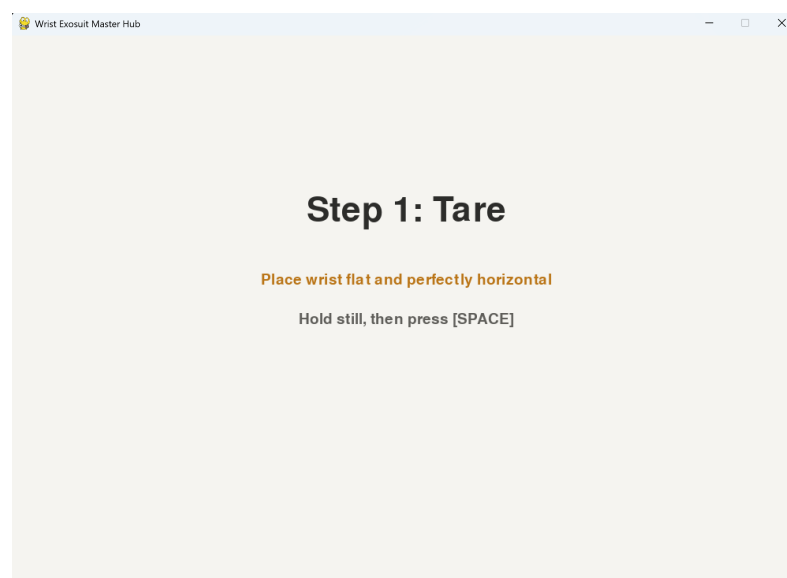


Figure 2.72: Neutral posture capture interface. This step establishes the mathematical baseline for the relative joint orientation.

Finally, it guides the user through isolated wrist extension and forearm supination movements.

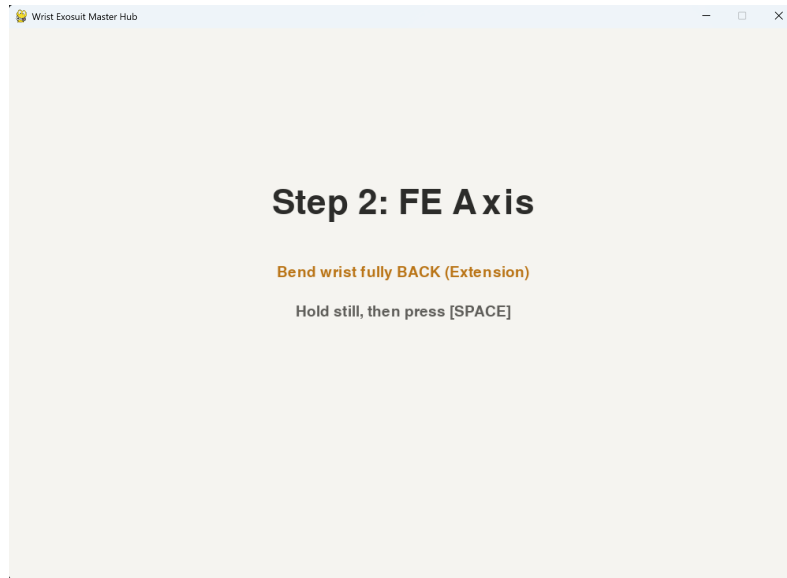


Figure 2.73: Biological axis extraction interface: capturing the peak wrist extension pose.

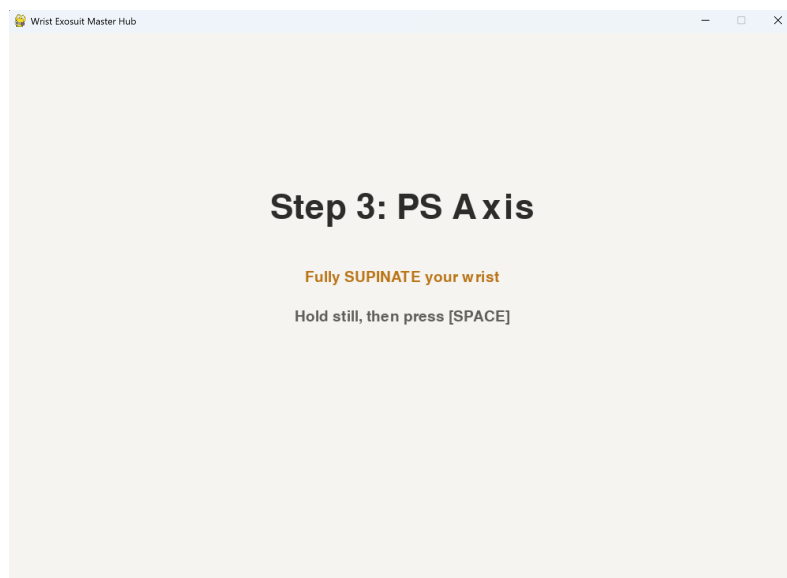


Figure 2.74: Biological axis extraction interface: capturing the peak forearm supination pose.

By clicking the corresponding capture buttons at the peak of these movements, the GUI commands the microcontroller to extract the pure biological vectors ($\mathbf{v}_{FE}$ and $\mathbf{v}_{PS}$) and execute the Gram-Schmidt orthogonalization.

By wrapping this highly complex mathematical calibration in a simple, visual step by step process, the GUI ensures the system is perfectly mathematically decoupled for every

unique user without requiring the therapist to understand the underlying quaternion mathematics.

Operational control hub

Once the exosuit is successfully calibrated and the empirical servo map is generated, the GUI transitions to the main operational control hub. This window serves as the primary dashboard for the duration of the therapy session, providing a clear visual hierarchy of the available rehabilitation strategies.

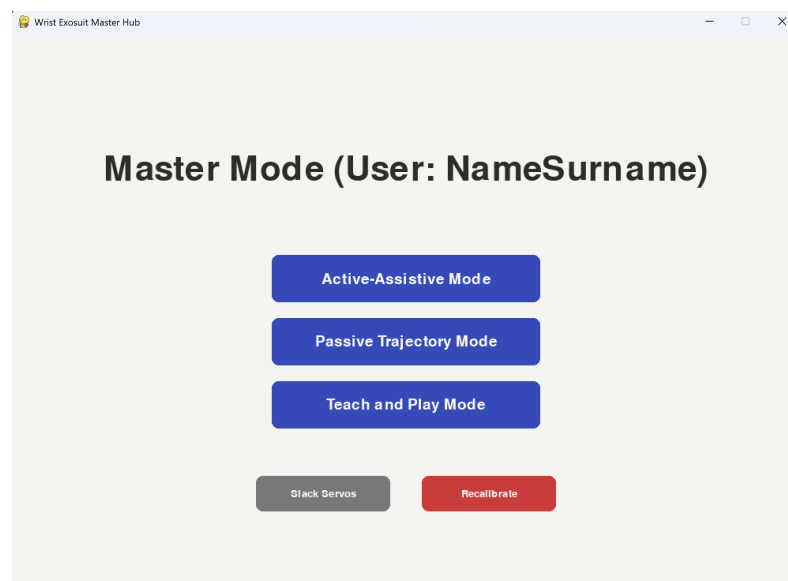


Figure 2.75: The main operational control hub. This centralized dashboard allows the therapist to select the primary high level modality, safely slack the servo motors, or restart the calibration protocol.

From this centralized menu, the therapist can seamlessly transition the exosuit between the primary high level modalities, such as active assistive control, automated passive trajectories, or the multidimensional Teach and Play framework. Clicking a primary mode button sends an immediate state command to the ESP32 microcontroller, instantly engaging the corresponding baseline control architecture.

Active assistive mode Engaging this mode instantly activates the cooperative power curves and unlocks access to interactive therapeutic exercises, such as a visual trajectory tracking game, as well as specific engineering diagnostic tools like a mechanical transparency test.

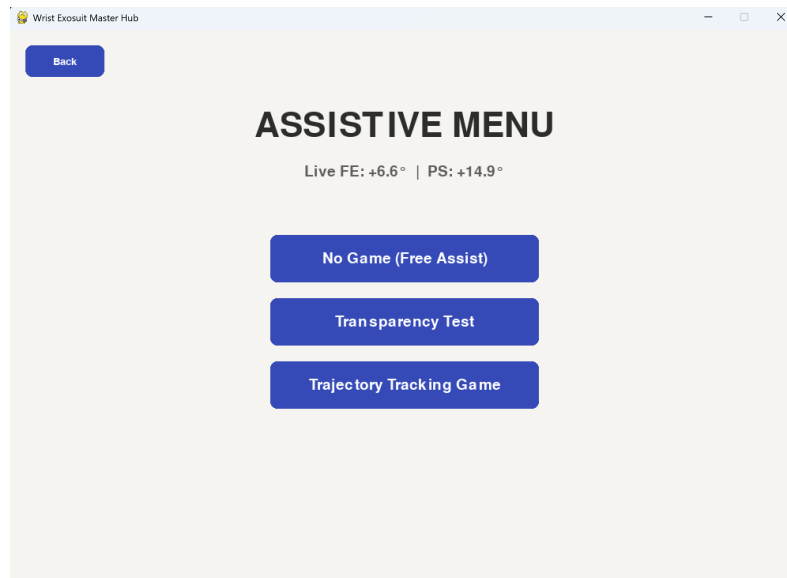


Figure 2.76: Active assistive sub menu. The interface displays real time kinematic feedback (top) and provides access to interactive tracking exercises and diagnostic tests (center).

Passive trajectory mode Selecting this modality provides the operator with the necessary parameters to launch mathematically generated angular profiles for automated stretching. Features include initiating the automated hardware mapping sequence, executing simple passive sweeps (which simultaneously generate the geometric decoupling diagnostics discussed previously), and launching complex, multi joint trajectories.

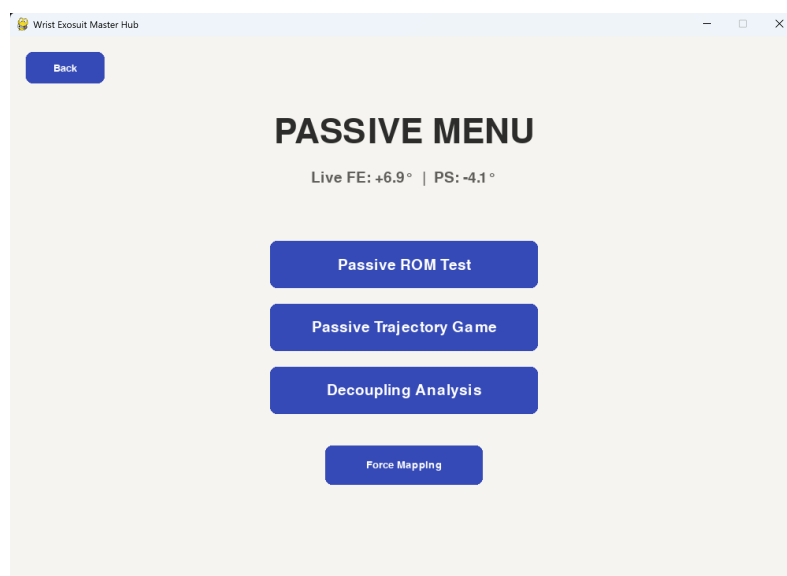


Figure 2.77: Passive trajectory sub menu, providing controls for hardware mapping and automated therapeutic movement execution.

Teach and Play mode This dedicated sub menu provides intuitive media controls (Record, Stop, Play) to safely manage the high frequency spatial recording buffer and trigger autonomous trajectory replication.

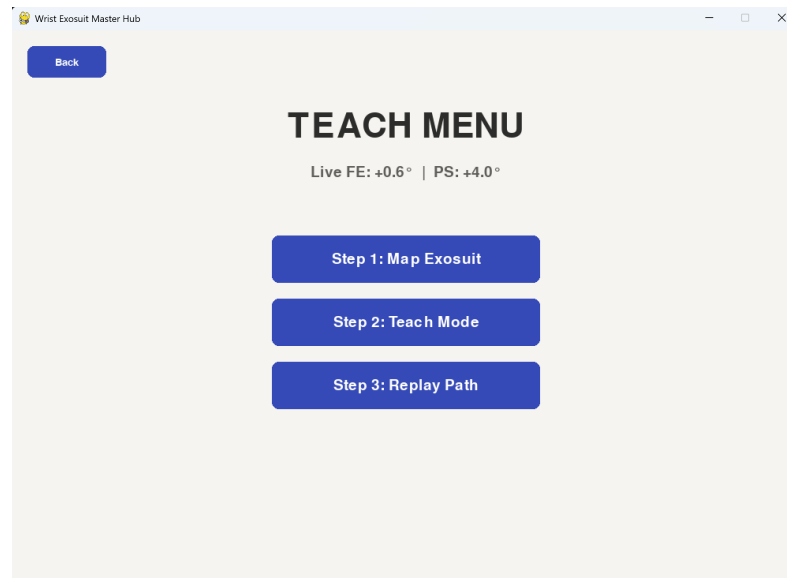


Figure 2.78: Teach and Play sub menu, featuring intuitive media controls to manage the high frequency spatial recording buffer and trigger autonomous trajectory replication.

Detailed descriptions and clinical evaluations of these specific sub functions and testing protocols are presented thoroughly in Section 2.4.

Across all modalities and sub menus, the interface maintains constant visibility of critical safety features. This design guarantees that the operator can safely halt any autonomous trajectory, interrupt a game, or instantly detach the servo motors at any moment via a master software emergency stop.

2.3.5. Control considerations for electrohydraulic actuators

While the current prototype utilizes traditional electromagnetic servo motors to validate the kinematic and software architecture, the ultimate goal of this research trajectory is the integration of electrohydraulic actuators such as HASEL or HALVE. Transitioning to these soft, electrohydraulic artificial muscles fundamentally alters the physical dynamics of the exosuit, necessitating specific considerations regarding the low level control architecture.

However, it is crucial to emphasize that the high level control strategies detailed in the previous sections remain entirely modular and valid for any hardware. The continuous dual IMU sensor fusion, the quaternion based spatial orthogonalization, the extraction

of the decoupled biological axes, and the velocity based intent detection algorithms are completely independent of the final actuation method.

The fundamental change occurs exclusively at the low level execution phase, where the control paradigm transitions from a position driven architecture to a force driven architecture. Electromagnetic servo motors are inherently position controlled devices; the microcontroller sends a specific pulse width signal, and the internal closed loop controller utilizes a rigid gear train to aggressively enforce that exact mechanical angle, regardless of the tension on the tendon. Conversely, HASEL and HALVE actuators operate through electrostatic zipping. Applying a high voltage signal across the integrated electrodes generates an electrostatic force that displaces the internal dielectric fluid, resulting in a volumetric contraction. Therefore, the control variable changes from commanding a rigid spatial position to commanding a contractile fluidic force. To accommodate this, the empirical mapping array utilized in the current software loop (which maps spatial angles to servo pulses) would simply be replaced by a force mapping function that translates the desired assistive torque into a specific high voltage output.

This shift to a force driven, electrohydraulic architecture provides massive clinical advantages regarding human robot interaction and patient safety. Because traditional servos use rigid mechanical gears, they are highly resistant to back driving. If a patient experiences a sudden, violent muscular spasm while the exosuit is executing a passive trajectory, the position driven servo will rigidly fight the biological resistance, potentially risking severe skeletal or tendon strain unless a secondary software safety threshold instantly detects the force spike and aborts the movement.

By contrast, HASEL actuators possess inherent mechanical compliance. Because they are constructed from flexible polymer films filled with liquid dielectric, their physical state is naturally soft. When driven by a specific voltage, they exert a proportional contractile force; however, if the patient resists with a biological force that exceeds this electrostatic threshold, the fluid safely redistributes and the actuator physically yields. This means that even if the software control loop fails to immediately detect a spasm, the physical hardware itself acts as a natural, fail safe shock absorber. This intrinsic compliance guarantees that the soft exosuit can never forcefully overpower the human skeleton, making electrohydraulic actuation the vastly superior choice for safe, comfortable, and continuous rehabilitation.

2.4. Validation protocol

The final phase of this research involves a comprehensive validation protocol designed to evaluate the physical and software performance of the prototype against the predefined functional and non functional requirements.

Traditionally, robotic development includes extensive isolated bench tests to quantify the exact torque, speed, and bandwidth of the mechanical actuators. However, because the current prototype utilizes commercially established electromagnetic servo motors (whose physical capabilities are already thoroughly documented by the manufacturer) and acts as a control testbed for future HASEL or HALVE (which are manufactured independently from this research and just purchased) integration, raw actuator bench testing was deemed unnecessary. Furthermore, literature analysis indicates that raw actuator bench testing is typically reserved for the introduction of novel, custom made actuators rather than for the integration of devices with externally validated specifications. Instead, the validation protocol is heavily focused on human robot interaction, assessing how the control architecture mathematically couples with human biomechanics.

The experimental validation is divided into two primary branches: tests on healthy subjects to establish the performance of the exosuit, especially baseline kinematics, control reliability, and system transparency; and tests on patients to evaluate the true clinical efficacy of the algorithms in the presence of motor impairment.

2.4.1. Tests on healthy subjects

Healthy subjects' inclusion / exclusion criteria

To ensure a standardized baseline for kinematic evaluation, healthy participants were recruited based on strict criteria.

The inclusion criteria required participants to be at least 20 years of age and no more than 65, possess no history of neurological or musculoskeletal disorders affecting the upper limb, experience no pain during wrist movements, and demonstrate the ability to perform a full active wrist range of motion. Furthermore, all subjects were required to understand the experimental protocol and sign an informed consent form prior to participation.

Conversely, individuals were excluded if they presented with any current or recent (within the last 6 months) upper limb injuries, experienced pain or discomfort during movement, or possessed skin conditions that would prevent the safe donning of the textile exosuit. Any underlying condition affecting motor control, such as essential tremor, also served as

a strict exclusion criterion.

Lastly, fluency in Italian was established as a mandatory inclusion criterion, as all experimental instructions and questionnaires were administered exclusively in Italian. This monolingual approach was deliberately selected over English only or bilingual alternatives. An English only protocol was dismissed because English was a secondary language for the entire participant pool, introducing the risk of misinterpreting the nuanced phrasing of the subjective evaluations. Conversely, a bilingual administration was avoided to eliminate the potential for translation discrepancies, ensuring that every participant received the exact same semantic tone and linguistic intention during the testing sequence.

Participants

Following the established criteria, a total of 12 healthy subjects voluntarily participated in the validation study. The cohort was evenly balanced by gender, consisting of 6 females and 6 males. The mean age of the participants was 28.3 years, ranging from 24 to 33 years. Notably, all 12 subjects were right handed. Although hand dominance was not an explicit inclusion criterion during the selection process, this demographic consistency proved highly advantageous, as the current exosuit prototype is specifically tailored to accommodate the right arm.

Experimental setup

To guarantee a repeatable and safe testing environment, a standardized experimental workstation was established. The subject was seated in a rigid chair equipped with a stable armrest, allowing the forearm to remain horizontally supported throughout the duration of the testing to isolate wrist kinematics from shoulder or elbow compensations. The primary component of the workstation was the exosuit itself, whose servo motors were powered via a standard electrical outlet.

The primary computational hub consisted of a laptop computer utilized to run the Python Graphical User Interface, manage the serial communication with the exosuit, log the high frequency kinematic data into CSV format for post processing analysis, and collect the demographic information of the users in an Excel file. A secondary tablet device was provided directly to the user to digitally administer the required questionnaires and Likert scale evaluations and the chronometer app on a phone was used.

Additionally, the experimental workstation included noise canceling headphones, a pre recorded audio track of the servo motors, and an opaque visual shield made from a piece of cardboard. These specific tools were used exclusively to isolate the user's sensory

perception during some phases of the protocol.



Figure 2.79: Experimental setup for tests on healthy subjects.

Prior to initiating the software tests, the operator collected standard anthropometric measurements of the user's forearm and hand using a flexible measuring tape to verify size compatibility. Including the initial onboarding, consent signing, and hardware calibration, the entire experimental protocol was designed to be completed in approximately 45 minutes per subject.

Tests

The testing sequence was structured chronologically into five distinct phases, carefully organized to minimize fatigue and progressively evaluate the different high level control modalities.

H1 - Transparency test The first test aimed to quantify how much the hardware, even unpowered, limited the Range of Motion. Specifically, it aimed to verify that the mechanical restriction of the RoM was minimal enough to permit ADLs, yet sufficient to provide stability for flaccid patients.

- **Conditions:** The test was evaluated under two comparative states: a pure baseline where the exosuit was completely unbuckled and disconnected from the hand (yet it was still worn because the IMUs were needed to get angle data of the hand), and a hardware condition where the exosuit was fully worn and buckled but electrically powered off.

- **Task:** With the forearm resting horizontally on the chair support, the subject performed three consecutive maximum sweeps of pure wrist flexion/extension. The procedure was then repeated for three sweeps of pronation/supination.
- **Measurements:** The dual IMU pipeline continuously recorded the joint angles. The maximum and minimum angular limits reached during the powered off state were directly compared to the unbuckled baseline to determine how severely the tension of the artificial tendons reduced the natural biological range of motion.

H2 - Noticing test To evaluate the true efficacy of the pulling action in the active assistive algorithm, a discrimination task with impaired sensory functions was executed.

- **Conditions:** To conduct this test without introducing cognitive bias, the user was completely blinded to the operational state of the exosuit. Visual feedback was blocked by placing the cardboard over the forearm, completely obscuring the hardware from the user's line of sight. Simultaneously, auditory feedback was eliminated by instructing the subject to wear the noise canceling headphones, which continuously played the pre recorded audio loop of the servo motors operating. This auditory masking prevented the subject from hearing the actual physical actuators trigger. Under these isolated sensory conditions, the software randomly alternated between two states: powered on (active assistive mode engaged) and powered off (transparent state). The sequence of on/off randomized actuation was different for each subject.
- **Task:** The subject was instructed to perform three consecutive sweeps of flexion/extension. This 3 sweep block was repeated 5 separate times, with the software potentially changing states between blocks. The entire 5 block sequence was subsequently repeated for the pronation/supination degree of freedom.
- **Measurements:** After every single block, the user used the tablet to record whether they physically felt the exosuit assisting their movement, rating their perception on a standard 5 point Likert scale (ranging from Strongly Disagree to Strongly Agree).

H3 - Trajectory tracking test This test assessed the dynamic responsiveness and control accuracy of the system during a continuous visual motor task.

- **Conditions:** The test compared three progressive states: an unbuckled anatomical baseline, the active assistive mode, and the automated passive trajectory mode.
- **Task:** The subject observed a custom video game interface on the computer screen displaying a dynamically moving target reference curve. The user was instructed

to continuously track this target curve by physically moving their wrist. This tracking task was performed independently for both flexion/extension and pronation/supination.

- **Measurements:** The software automatically logged the kinematic tracking performance, calculating the Root Mean Square Error (RMSE), the absolute maximum error spike, and the total percentage of time the user successfully maintained their angle within a predefined $\pm 5^\circ$ tolerance window.

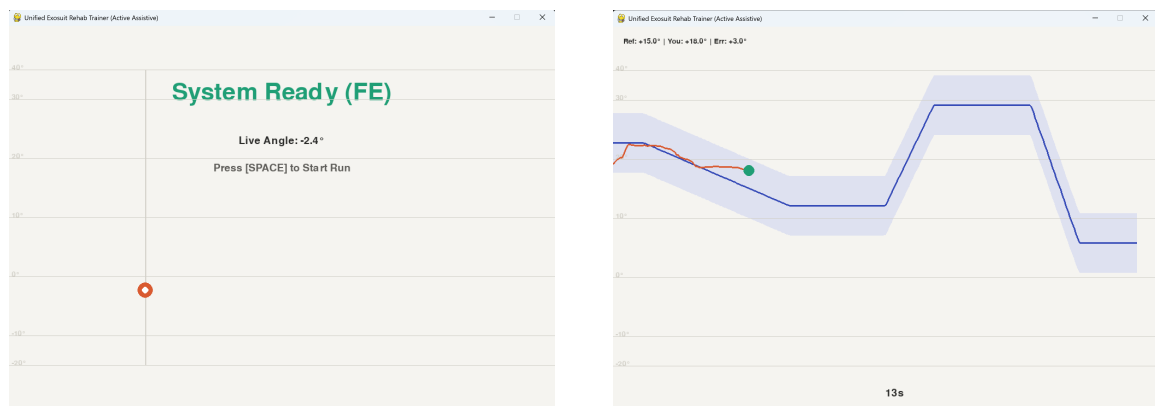


Figure 2.80: Trajectory tracking game in FE mode: on the left, the initialization page where the user can familiarize themselves with the control of the movement of the dot via exosuit's movement (the dot goes up and down according to the wrist extension and flexion); on the right the actual game (the trajectory arrives from the right side).

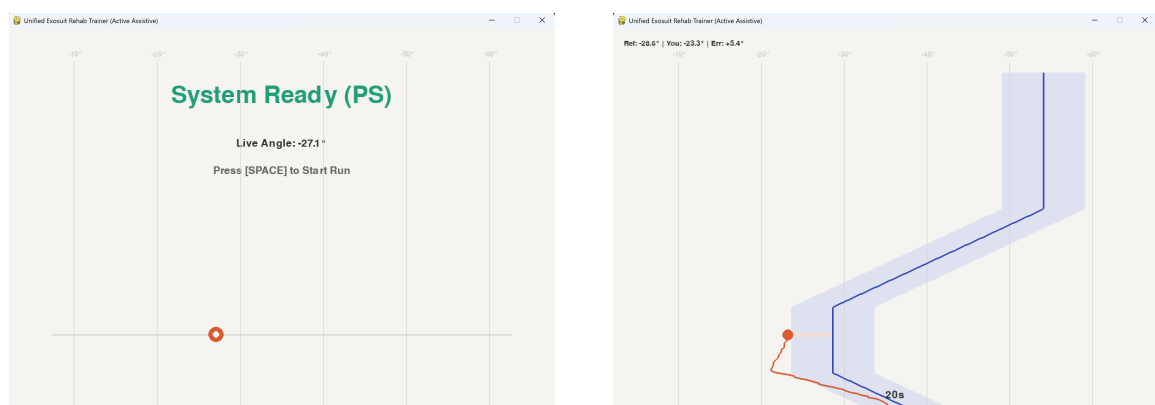


Figure 2.81: Trajectory tracking game in PS mode: on the left, the initialization page where the user can familiarize with the control of the movement of the dot via exosuit's movement (the dot goes left and right according to the wrist pronation and supination); on the right the actual game (the trajectory arrives from the upper part).

H4 - Donning and doffing To verify the ergonomic design and the usability of the exosuit, an evaluation of the time required to independently don and doff the device was conducted.

- **Conditions:** The exosuit was evaluated in its powered off state.
- **Task:** The subject was instructed to independently wear the exosuit, secure all anchoring straps, and subsequently remove the device.
- **Measurements:** The temporal duration required for donning and doffing was recorded via a digital chronometer.

HQ - Questionnaire Following the conclusion of all physical tasks, the subject completed a comprehensive digital survey to evaluate their subjective experience. To guarantee structured and scientifically validated results, the questionnaire was divided into specific sections. The hardware comfort and physical design were evaluated using metrics from the standard QUEST 2.0 (Quebec User Evaluation of Satisfaction with Assistive Technology) framework. To specifically assess the control design, custom 5 point Likert scale questions were developed to capture feedback regarding the comfort, intrusiveness, and perceived efficacy of both the active assistive and passive control modalities. Finally, an open text box was provided to allow subjects to record unstructured qualitative observations and suggestions for future iterations.

The standard QUEST 2.0 [17] was selected over alternative evaluation frameworks, such as the System Usability Scale (SUS) [10] and the Exoscore [66]. This decision was motivated by the fact that the QUEST 2.0 metric is explicitly tailored for the assessment of assistive technology, whereas the alternative methodologies offer a significantly more generic evaluation of system usability.

2.4.2. Tests on patients

These tests served primarily as a pilot study to evaluate the practical utility of the exosuit's assistance and to gather qualitative data and suggestions from real end users. For this reason, the experimental campaign was not conducted on a full cohort of twelve or more subjects, as was done for healthy individuals, but rather on a smaller sample of two patients.

The experimental trials were conducted at the “Villa Beretta” clinic, a rehabilitation center specializing in the recovery of human function following disabilities caused by Central Nervous System (CNS) lesions, including strokes, spinal cord injuries, traumatic

brain injuries, and degenerative diseases.

Patients' inclusion and exclusion criteria

To ensure the safety of the participants and the reliability of the collected data, specific clinical parameters were established for patient selection.

Eligible participants were required to be at least 18 years of age and exhibit reduced motor function in their right wrist stemming from neurological or musculoskeletal conditions. Furthermore, they needed a stable clinical profile outside of any acute phase, the cognitive capacity to understand and follow experimental instructions, and the ability to initiate voluntary wrist movements with enough residual active range of motion to perform partial trajectories.

Conversely, individuals were excluded from the study if they presented with severe spasticity (e.g., a score greater than 2 on the Modified Ashworth Scale [9] shown in Figure 2.82), fixed contractures, or severely limited passive range of motion that the exosuit could not mechanically overcome. Additional exclusion criteria encompassed a complete inability to initiate movement (as a kinematic input is necessary for active assistive mode) as well as pain caused by joint stiffness during motion. Finally, patients with severe tremors, uncontrolled movements, skin lesions preventing the safe donning of the exosuit, or any severe cognitive impairments that could compromise overall safety were also excluded from the testing cohort.

TABLE 1
Modified Ashworth Scale for Grading Spasticity^a

Grade	Description
0	no increase in muscle tone
1	slight increase in muscle tone, manifested by a catch and release or by minimal resistance at the end of the range of motion when the affected part(s) is moved in flexion or extension
1+	slight increase in muscle tone, manifested by a catch, followed by minimal resistance throughout the remainder (less than half) of the ROM
2	more marked increase in muscle tone through most of the ROM, but affected part(s) easily moved
3	considerable increase in muscle tone, passive movement difficult
4	affected part(s) rigid in flexion or extension

Figure 2.82: Modified Ashworth Scale from [9].

Participants

The pilot study involved two patients, specifically a 36 year old female and a 39 year old male. Both participants were characterized by hand spasticity and significantly reduced wrist mobility. Additionally, the male subject presented with slight joint rigidity in the extension direction.

Experimental setup

The experimental setup was similar to the one used for healthy subjects. Patients were seated in a chair with armrests (or in their own wheelchairs) and positioned in front of a table. Initial calibration and Range of Motion measurements were performed with the arm resting on the armrest, whereas the table served as a support surface for the functional activities. The equipment setup also included a computer, a power source, a smartphone, modeling clay (Figure 2.83), and a specialized fork orthosis named e-KIRO [23] (Figure 2.84).



Figure 2.83: Modeling clay used to represent food.



Figure 2.84: e-KIRO fork orthosis from [23].

Tests

The experimental protocol focused exclusively on the active assistive and passive modes. Although the Teach and Play mode was specifically designed for patient rehabilitation, it was excluded from this testing phase. It is worth noting that this modality was not evaluated during the trials with healthy subjects either; since they are not the intended target users, they would likely have corrupted the data by unconsciously performing the movements themselves.

For the patient cohort, the exclusion of the Teach and Play mode was driven by two primary factors. First, it was deemed unsuitable for the functional activity tests. From the very beginning of the trials, it became apparent that patients needed to heavily compensate with other body parts to successfully complete the tasks, which would have rendered the strictly isolated tracking of the Teach and Play mode meaningless in that context. Second, while this mode would have been highly appropriate for the isolated Range of Motion exercises, it was omitted to prevent excessive physical stress. The participants were already significantly fatigued from the previous test iterations and this fatigue was macroscopic and clearly visible in the kinematic results.

Similarly, to avoid overexerting the participants at the end of the protocol, which lasted approximately one hour per patient, a formal written questionnaire was not administered. Instead, qualitative oral feedback was collected directly from both the patients and the clinic operators.

P1 - RoM in different conditions To verify the effectiveness of the active assistive and passive mode in increasing the RoM of the wrist, full movements of flexion/extension and pronation/supination were performed not only in those two modes but also in baseline conditions.

- **Conditions:** The exosuit was always worn by the user, both because the IMUs were needed to compute the angles of the wrist and because donning and doffing with patients was complicated and very time consuming, and the test was conducted in several comparative states: unassisted and unbuckled active movement of the user, passive movement forced by an external clinical operator (necessary because the patient's structural rigidity could limit the overall possible range compared to healthy individuals), active movement assisted by the exosuit, and passive movement entirely driven by the exosuit. During the pronation/supination active assistive mode, the flexion/extension tendon was buckled tightly but unpowered, as it acted as a support to keep the hand horizontal.

- **Task:** Patients were asked to perform their maximum possible Range of Motion, first for flexion/extension, and subsequently for pronation/supination.
- **Measurements:** The kinematic data of the movements were continuously recorded using the IMU sensors and saved to a CSV file.

P2 - Functional activity flexion/extension

- **Conditions:** Unassisted active movement (wearing the unpowered exosuit) and active movement assisted by the exosuit.
- **Task:** A smartphone displaying an application with a large red button (Figure 2.85) was placed on the table. Participants were instructed to reach the phone, raise their hand to clear the device's thickness, and lower it to press the button, which incremented a digital counter. This sequence was repeated three times. Notably, patients were allowed to, and frequently did, use compensatory movements with other body parts to complete the task.

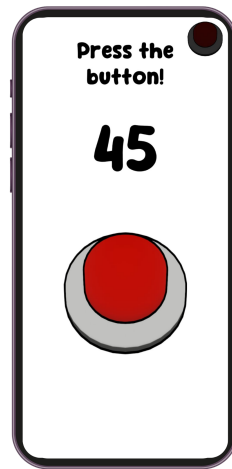


Figure 2.85: Phone application used for the flexion/extension functional activity.

- **Measurements:** The activity was marked as successful/unsuccessful for each of the 3 attempts and kinematic performance and compensatory tracking data were recorded using the IMU sensors.

P3 - Functional activity pronation/supination

- **Conditions:** Unassisted active movement (wearing the unpowered exosuit) and active movement assisted by the exosuit.

- **Task:** Patients wore the e-KIRO orthosis and were tasked with simulating the action of eating a piece of modeling clay using a fork. This functional sequence was repeated three times.
- **Measurements:** The activity was marked as successful/unsuccessful for each of the 3 attempts and kinematic performance and compensatory tracking data were recorded using the IMU sensors.

2.4.3. Requirements and tests

It is essential to remember that the experimental protocols are fundamentally designed to validate the fulfillment of the system requirements, ensuring that every defined functional and non functional parameter is rigorously evaluated by at least one specific testing procedure. The correspondence between requirements and tests that attempt to investigate them is reported in Table 2.14.

Requirement	Verif. by Constr.	Healthy Subjects					Patients		
		H1	H2	H3	H4	HQ	P1	P2	P3
F1 - Assisted movements			X	X		X	X	X	X
F2 - Operating modes			X	X		X	X	X	X
F3 - User interaction				X					
N1 - Range of motion				X			X		
N2 - Assistance capability	X			X				X	X
N3 - User controllability				X		X		X	X
N4 - Speed				X					
N5 - Transparency		X							
N6 - Comfort of wearing						X			
N7 - Ease of operating					X	X			
N8 - Portability and weight						X			
N9 - Adaptability to different users					X	X			
N10 - Emergency stop	X								
N11 - Electrical safety	X								

Table 2.14: Cross reference matrix mapping the functional and non functional requirements to their specific experimental validation methods. *Verif. by Constr.* means *Verified by Construction* and refers to those requirements that were used as design requirements and are, therefore, automatically satisfied.

3 | Results

3.1. Tests on healthy subjects

The experimental validation involved 12 healthy subjects, comprising 6 males and 6 females, whose demographic characteristics ¹ are depicted in Figures 3.1 and 3.2.

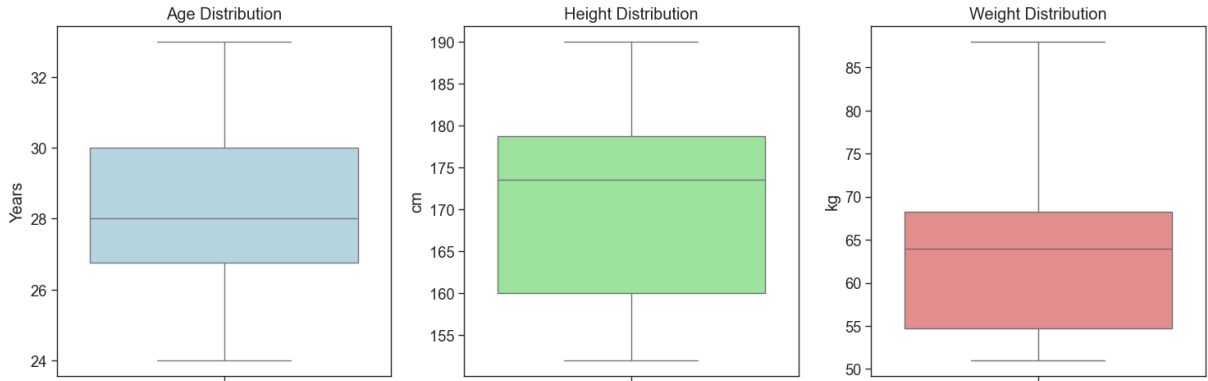


Figure 3.1: Distributions of age, height and weight of the healthy subjects.

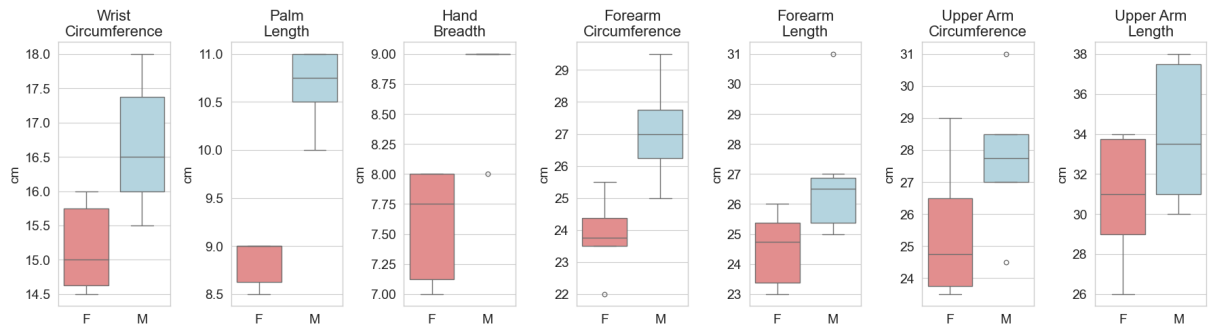


Figure 3.2: Distributions of hand, wrist and arm dimensions of the healthy subjects.

¹Anthropometric dimensions were recorded as follows: wrist circumference was taken just below the wrist bone; palm length from the base of the palm to the base of the middle finger; hand breadth across the widest part of the hand just below the knuckles (excluding the thumb); forearm circumference a few centimeters below the elbow, in the region covered by the sleeve; forearm length from the wrist bone to the elbow bone; upper arm circumference a few centimeters above the elbow, in the region covered by the sleeve; and upper arm length from the acromion process at the top of the shoulder to the olecranon process at the tip of the elbow.

Categorizing the demographic distribution by sex was particularly relevant since female subjects generally present smaller arm and wrist dimensions. During the execution of the experimental protocol, it was empirically observed that the prototype performed more effectively on female participants, a consequence of the fact that the soft exosuit was originally designed and patterned on a female reference user ².

3.1.1. H1 - Transparency test

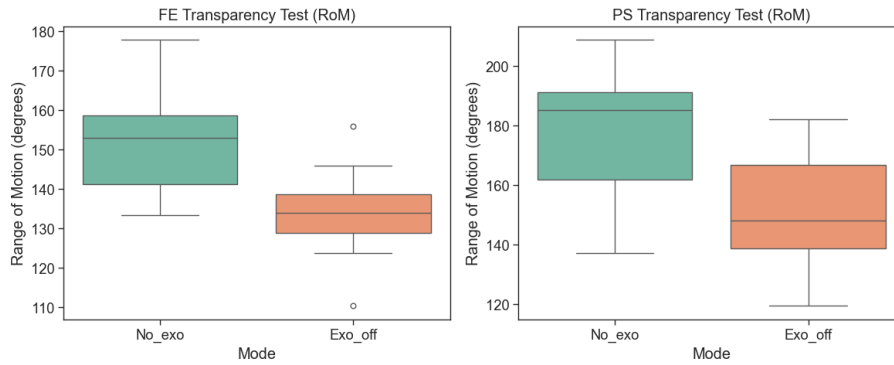


Figure 3.3: Reduction in RoM of the healthy subjects between unbuckled and buckled switched off conditions.

As expected, the experimental data highlighted a statistically significant reduction in the overall Range of Motion (RoM) when the users wore the unpowered device. A paired t-test confirmed this physical limitation for both flexion/extension (from $151.7^\circ \pm 13.5^\circ$ to $133.9^\circ \pm 11.7^\circ$, retaining $88.7\% \pm 9.2\%$ ³ of baseline; $p = 0.0023$) and pronation/supination (from $176.7^\circ \pm 23.0^\circ$ to $151.2^\circ \pm 18.2^\circ$, retaining $86.0\% \pm 6.9\%$; $p < 0.001$). This reduction is an inherent consequence of the artificial tendons being routed and tensioned along the arm. However, this phenomenon is not necessarily detrimental, as the slight tension provides stabilization to the wrist joint. Most importantly, the resulting limited RoM still vastly exceeds the physiological angles required for standard Activities of Daily Living (ADLs) specified in Table 2.2, meaning that the limitation poses limited functional impairment to the user.

Interestingly, the distribution of the RoM reduction among the participants shown in Figure 3.4 reveals that a subset of subjects experienced very minimal or even negative reductions. This anomaly occurred because some subjects, considering that they had to reach the same extreme angles recorded during the unbuckled baseline, voluntarily

²The author.

³Retention computed per subject, then averaged.

subjected their wrists to significant stress. Relying on their own initiative, they exploited the natural elasticity and softness of the textile sleeve and glove to push the joint to its extreme limits, artificially minimizing the recorded transparency loss.

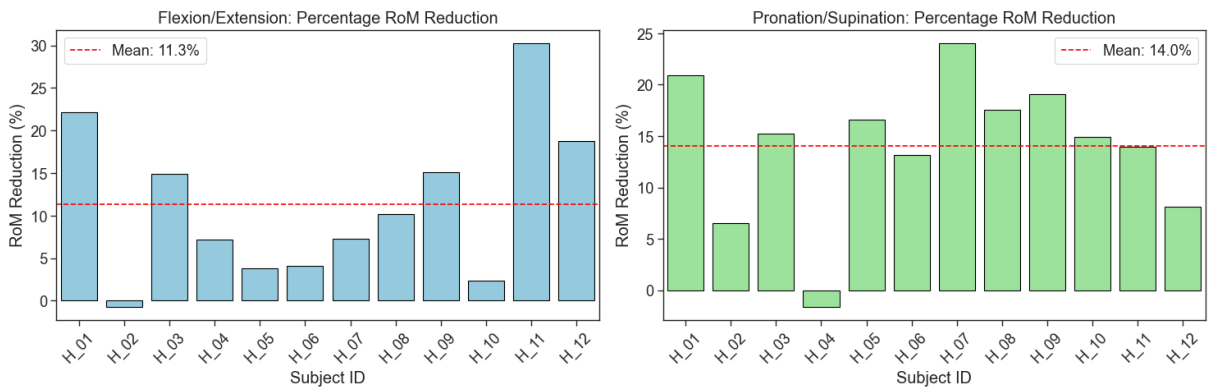


Figure 3.4: Percentage of the reduction in RoM of the healthy subjects between unbuckled and buckled switched off conditions subject by subject.

3.1.2. H2 - Noticing test

The outcomes of the noticing test were overwhelmingly clear, demonstrating that the subjects were consistently able to identify the correct actuation condition. By quantifying the perceptibility as the mean distance between the Likert scores assigned during the active and unpowered states, the analysis revealed a highly distinct separation: 3.92 ± 0.21 for flexion/extension and 3.97 ± 0.10 for pronation/supination (out of a maximum possible distance of 4). The extremely low standard deviations indicate that this distinct differentiation was uniform across the entire subject pool.

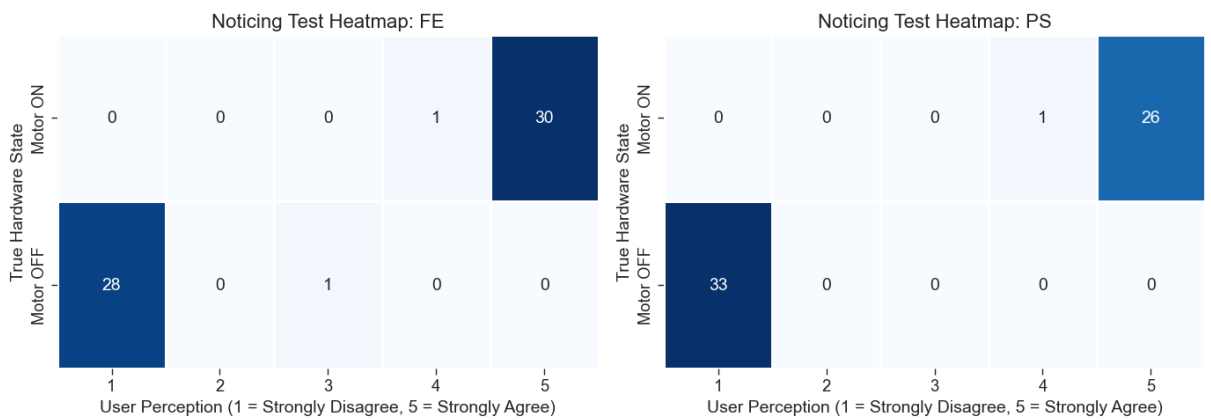


Figure 3.5: Heatmap of the noticing test results, illustrating subject responses to the randomized on and off states of the exosuit. Participants rated their agreement with the statement "I felt assistance" using a 1 to 5 Likert scale.

This specific test would represent an ideal validation benchmark for electrohydraulic artificial muscles, such as HASEL or HALVE actuators, where simply perceiving the assistive force would already be considered a significant achievement. Nevertheless, conducting this evaluation with servomotors remained highly valuable. Healthy subjects are accustomed to moving their wrists effortlessly and subconsciously, meaning that it is not automatic for them to consciously perceive external assistance. Therefore, the consistent ability to recognize when the exosuit was operating proves that the magnitude of the assistive action provided by the motors is sufficiently robust.

3.1.3. H3 - Trajectory tracking test

An example of the overlapped reference and actual trajectories is provided below to visually demonstrate the dynamic tracking behavior of the system.

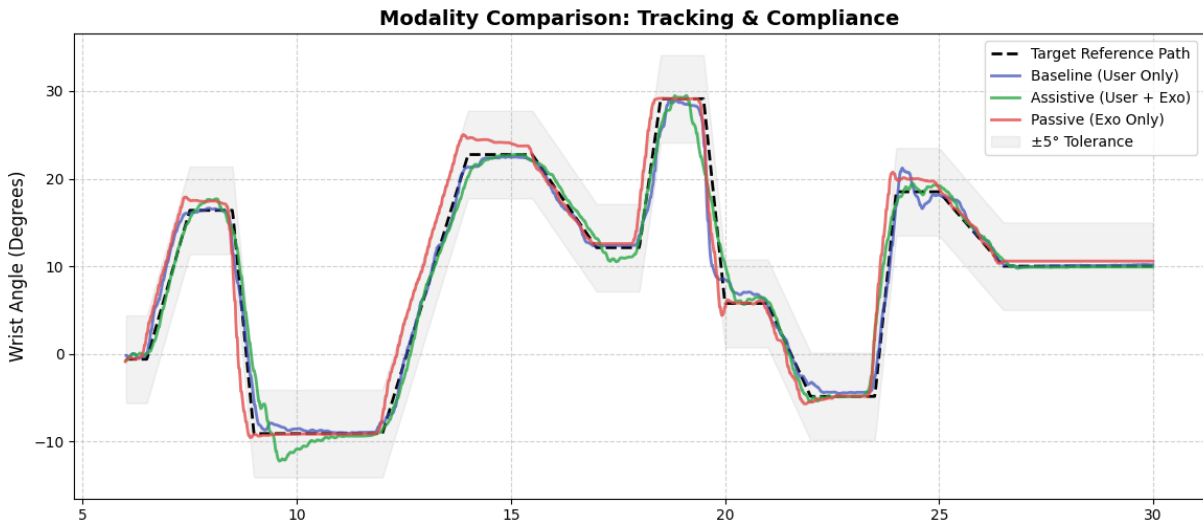


Figure 3.6: Overlapped reference and actual trajectories during a flexion and extension trial for a representative healthy subject.

The tracking performance was quantified using the Root Mean Square Error (RMSE) and the Time in Tolerance (TiT). RMSE provides a comprehensive measure of the overall deviation while heavily penalizing large errors, whereas TiT indicates the percentage of the task duration during which the user remained within a clinically acceptable 5° boundary from the target.

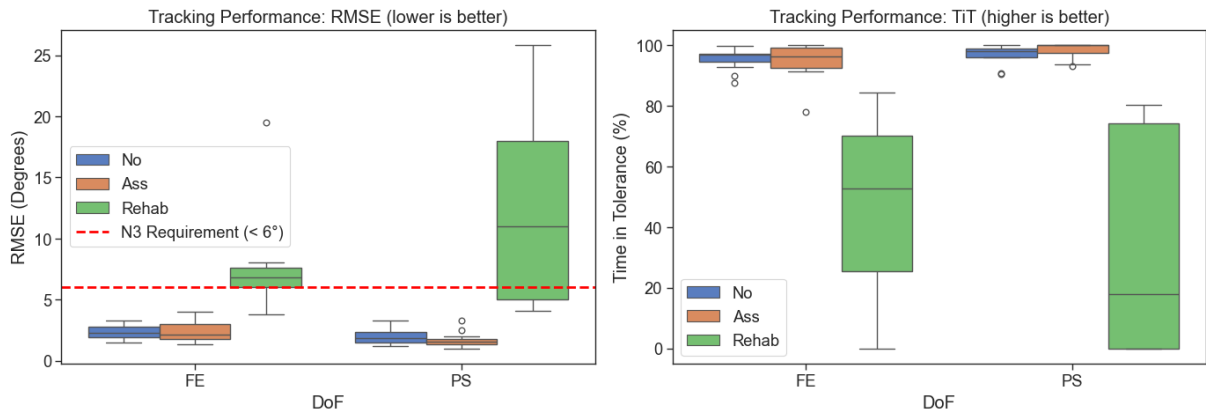


Figure 3.7: Overall trajectory tracking performance across all healthy subjects, quantified by Root Mean Square Error and Time in Tolerance.

For the active assistive modality, the overall RMSE ($2.01^\circ \pm 0.80^\circ$) successfully fell below the strict design threshold defined in the system requirements in Table 2.2. To compare the assisted and unassisted conditions, a conditional test selection was adopted according to the normality of the paired differences. For flexion/extension (FE), the differences were non-normal (Shapiro $p = 0.0058$), requiring the use of the Wilcoxon signed-rank test. Conversely, for pronation/supination (PS), the differences followed a normal distribution (Shapiro $p = 0.3532$), allowing the use of the more sensitive paired t-test. The analysis revealed a significant improvement in tracking accuracy for PS ($p = 0.0481$), whereas no statistically significant difference was observed for FE ($p = 0.4648$). While the overall RMSE for FE was not significantly reduced, the magnitude safely maintained the baseline error. This is a highly positive result, confirming that the exosuit is intuitive and easy to control without degrading natural human kinematics. Conversely, the passive modality exhibited worse and more varied performances, the mechanical causes of which are analyzed later in this section.

Regarding the active assistive mode, extension is the only actively supported part of the flexion/extension movement, while flexion must be slightly fought against to perform it rapidly (whereas executing it slowly allows gravity to govern the motion). One might hypothesize that the exosuit would facilitate the steep phases of extension while limiting the user during the steep phases of flexion. Furthermore, it was not known a priori whether the device helps maintain intermediate horizontal positions better than healthy subjects naturally can, or if it instead opposes them by continuously pulling toward the extremes of the RoM. For this reason, a specific analysis separating the going up, going down, and horizontal phases was conducted, alongside a parallel left and right analysis for pronation and supination.

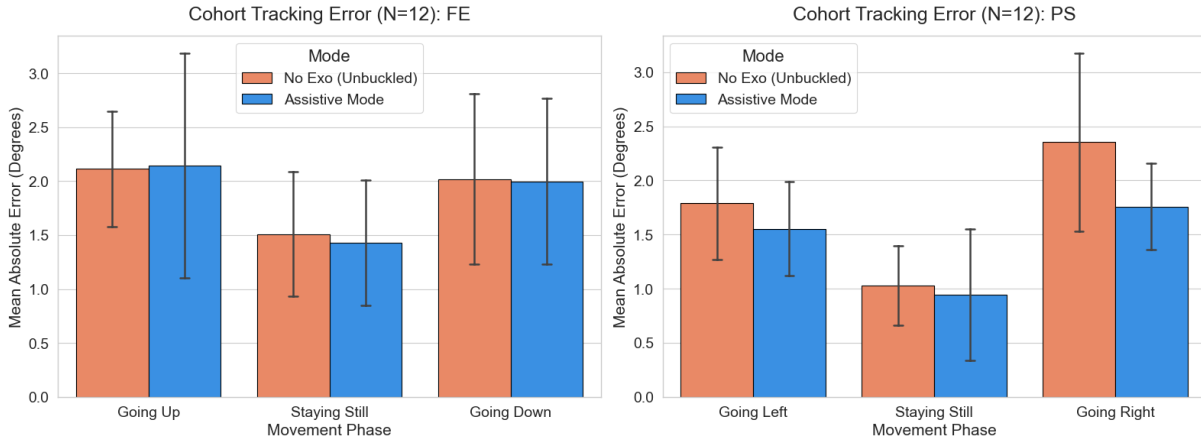


Figure 3.8: Directional analysis of the trajectory tracking test, detailing the Mean Absolute Error for unassisted and assisted conditions across distinct movement phases.

For this directional analysis, the Mean Absolute Error (MAE) was used instead of the RMSE. When breaking down a specific, 2 second movement like the downward phase, MAE translates directly into a more intuitive physical metric. If the MAE is 3.5, it literally means that the user's hand deviated from the target by an average of 3.5°. RMSE squares the errors before averaging them, which makes the final number harder to translate directly into physical degrees of deviation for such a short burst of time.

The directional analysis revealed that there is no massive difference in the overall error magnitudes, but it highlighted distinct interaction dynamics. Statistical testing demonstrated that the active assistance provided a significant improvement in pronation/supination, specifically during the "going right" phase ($p = 0.0068$) and the "staying still" phase ($p = 0.0425$). In contrast, none of the phases in flexion/extension yielded statistically significant differences (going up $p = 0.6848$, staying still $p = 0.3757$, going down $p = 0.9460$). Importantly, this lack of significance is not a problematic outcome; rather, it reflects a ceiling effect expected when testing on a healthy cohort whose baseline performance is already at the limit of human motor control. The fact that the active assistance did not significantly worsen this near perfect baseline proves the transparency and synchronization of the controller.

Looking at the raw MAE trends, an unexpected behavior emerged for flexion/extension. Contrary to initial predictions, the tracking error with the exosuit was marginally greater than the unbuckled baseline during the upward phase (extension) and smaller during the downward phase (flexion). In fact, the assisted modality yielded better mean tracking performance than the unassisted baseline across all testing conditions, including the horizontal holding phase and all phases of pronation and supination, with the single exception

of the upward phase. This specific increase in error contradicts initial expectations but is likely due to an overshooting effect. Because the exosuit provides substantial mechanical assistance during extension, users who instinctively applied the same muscular effort they used during the unassisted trials ended up overshooting the target angle upwards. Despite this isolated overshooting phenomenon, the lower error across all other phases confirms that the system effectively improves overall tracking performance and joint stabilization.

The target trajectory was intentionally kept identical across trials to facilitate comparison. Although the 30 second path is too long and varied to be perfectly memorized, users could learn how to correctly tune their hand angles to interact with the device. This implies that the assisted trajectory was slightly facilitated by the fact that it was always performed second. However, most subjects learned how to control the virtual game within the first 5 seconds of their initial attempt. To ensure a fair evaluation, the analysis was conducted by excluding the first 6 seconds of the trajectory, as can be seen in Figure 3.6. This crop also eliminated artificial errors caused by users starting from incorrect initial resting positions, which were not communicated to them beforehand.

In general, the overall ratio of improved and worsened tracking phases due to the exosuit intervention is shown in Figure 3.9, representing a very successful outcome.

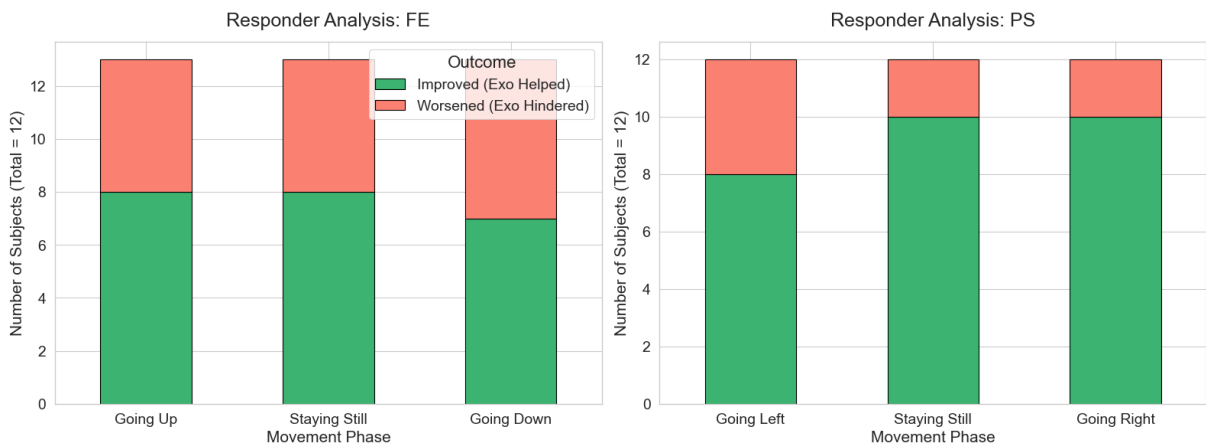


Figure 3.9: Proportion of improved VS worsened tracking performance in the assisted mode compared to unassisted, categorized by trajectory phase for healthy subjects.

The passive modality, however, highlighted two distinct mechanical challenges inherent to soft robotics:

1. **Lower Range of Motion:** The actual RoM was lower than the target due to the compliance of the fabric. Rather than pulling the hand to the extreme upper limit, the servomotors sometimes rotated or shifted onto the soft fabric sleeve, resulting

in incomplete movement. This parasitic displacement is a well known disadvantage of anchoring rigid components onto a soft exosuit. Furthermore, attempting to mathematically center a defective RoM around the middle position required the tendons to be tightened more than in the active assistive mode. If this tension was too high, reaching the downward limit of the RoM became physically impossible.

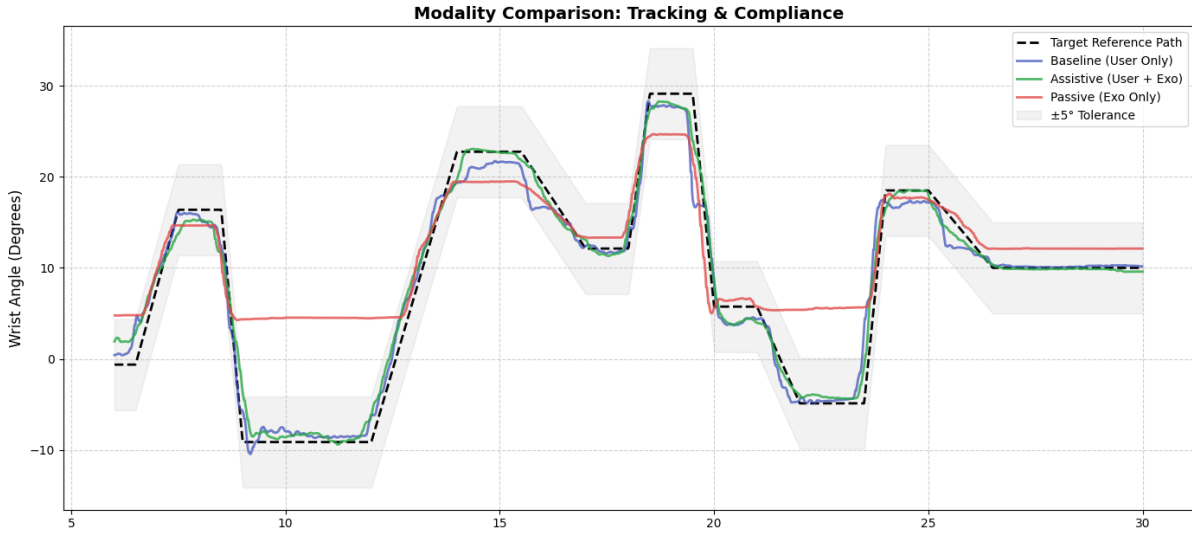


Figure 3.10: Visualization of the reduced Range of Motion in the passive trajectory (red) during a flexion/extension test for a representative healthy subject. The upper range is limited by the compliance of the sleeve fabric, while the lower range is restricted by excessive tendon tensioning.

2. **Translated trajectory:** The primary reason why the passive trajectory is frequently translated downwards is the mechanical slippage of the exosuit (either the sleeve shifting along the arm or the tendon slipping within the adjuster). This issue occurred much less frequently when the creator of the device wore it, due to a better understanding of how to properly secure the harnesses. The implemented PI controller, whose intervention is evidenced by the oscillatory behavior visible in the lower left section of Figure 3.12, is too weak to compensate for offsets of this magnitude, as it was designed strictly for small adjustments. The proportional gain is conservative, meaning it only attempts to correct 35 percent of the immediate error. More importantly, the integral windup is strictly capped. If the suit slips and creates a sudden 20 degree error, the integral loop quickly hits its 10 degree ceiling and gives up, as it is mathematically not allowed to reel in enough cable to pull the trajectory back to the reference line.

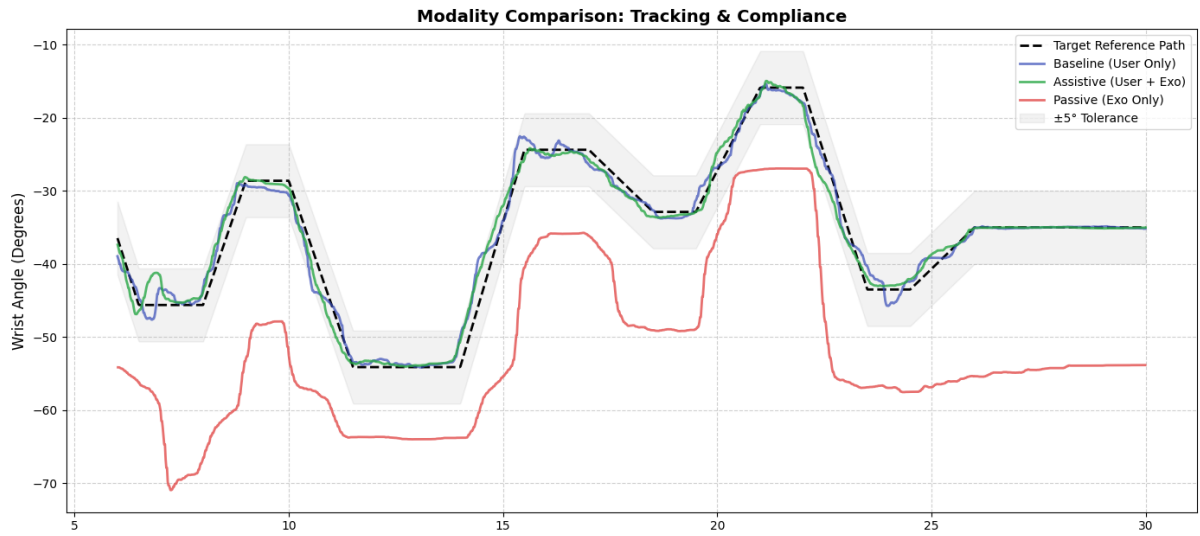


Figure 3.11: Visualization of the downward translation in the passive trajectory (red) during a pronation and supination test for a representative healthy subject. The overall downward shift is caused by the mechanical slippage of the exosuit.

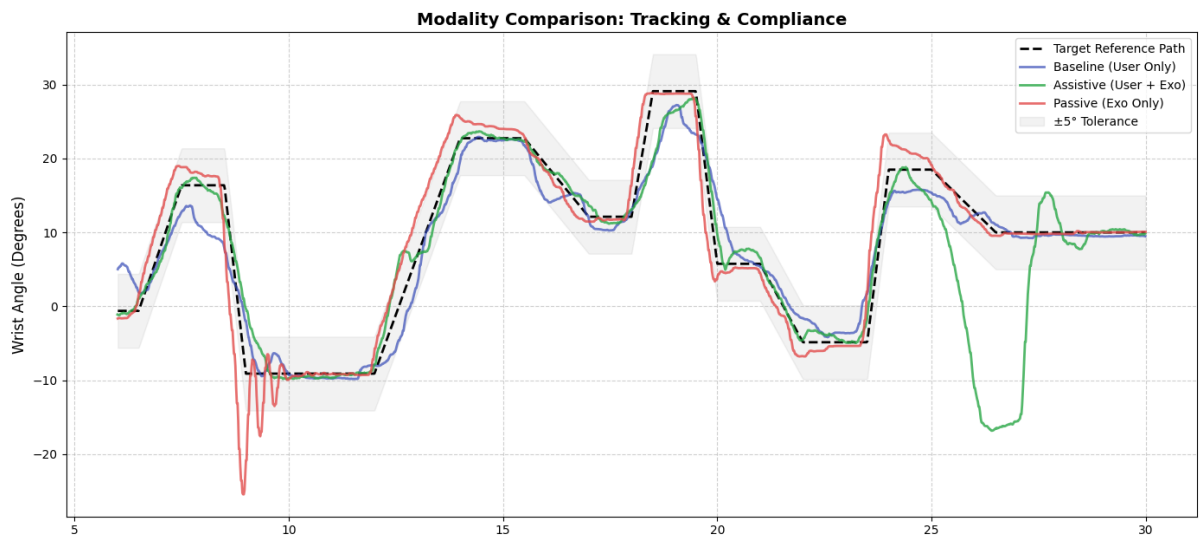


Figure 3.12: Visualization of a flexion and extension test for a representative healthy subject. The active intervention of the PI controller is highlighted by the oscillatory motion observed in the lower left portion of the graph, following the initial step descent.

The passive mode target completion plot (Figure 3.13) visually summarizes the system performance against the 100% requirement (mean completion was $85.80\% \pm 17.68\%$ for FE and $82.87\% \pm 19.89\%$ for PS). The observed variance, including instances where completion exceeded 100%, is attributed to three primary factors:

1. **Involuntary subject participation:** Healthy subjects often subconsciously fire agonist muscles to "help" the passive movement, compounding motor torque and leading to overshooting.
2. **Textile compliance and sensor displacement:** At the extremes of the joint RoM, fabric stretching and slippage of the IMU sensors over the skin can result in recorded angular displacements that exceed pure anatomical joint rotation.
3. **Dynamic overshoot:** The interplay between initial breakaway torque and the elasticity of the tendon routing induces a minor pendulum-like overshoot before the joint settles.

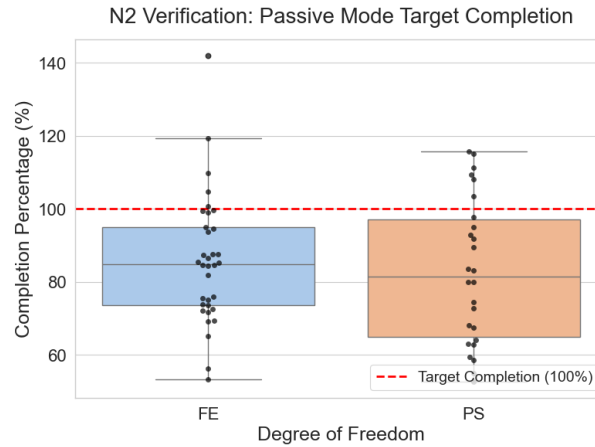


Figure 3.13: Target completion rate for the Range of Motion in the passive modality. The data represents the percentage of the actual recorded motion relative to the expected requirements outlined in Table 2.2.

3.1.4. H4 - Donning and doffing

Usability and independence were assessed by evaluating the time required for donning and doffing the device.

The temporal requirements were successfully satisfied across the entire cohort, with the exception of a single subject. This participant exceeded the two minute donning threshold because he voluntarily decided to remove the USB cable to find a more comfortable routing path, subsequently requiring extra time to reinsert the connector using his non dominant hand. Nevertheless, his total time was only marginally higher than the established requirement.

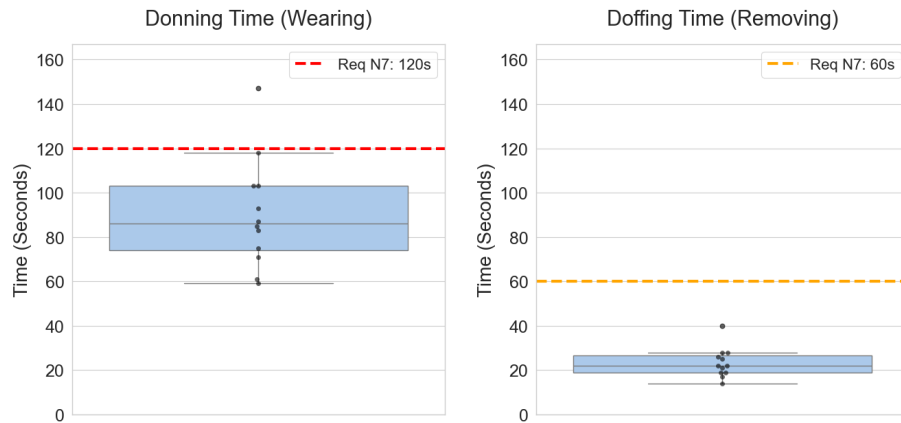


Figure 3.14: Donning and doffing time for healthy subjects with relative thresholds derived from Table 2.2.

During the experimental trials, it was noticed that individuals with smaller hands, especially female subjects, experienced significantly less trouble removing the exosuit. To mathematically investigate this observation, the potential correlation between hand dimensions and doffing time was initially analyzed using individual variables.

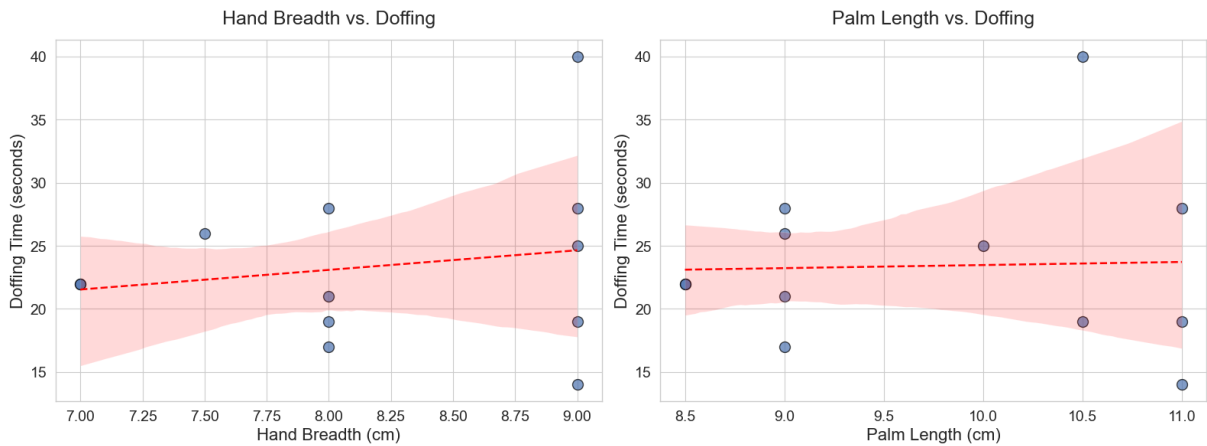


Figure 3.15: Relationship between individual hand dimensions and doffing time for healthy subjects. Note that only 11 data points are visible for the 12 healthy subjects, as two participants recorded identical measurements and times, resulting in completely overlapping markers. The shaded region represents the 95% confidence interval for the regression estimate.

When evaluated independently, although the individuals who took the longest time to doff the device possessed the largest hands, variables such as hand breadth ($r = 0.179, p = 0.577$) and palm length ($r = 0.036, p = 0.910$) did not exhibit a statistically significant

correlation with the recorded doffing times. A slight positive trend remains noticeable upon visual inspection when these measurements are analyzed concurrently to represent overall hand size (Figure 3.16); however, this trend does not correspond to a robust statistical correlation. The lack of statistical significance confirms that the current glove design provides sufficient clearance, effectively decoupling doffing duration from hand anthropometry, despite the slight trend observed.

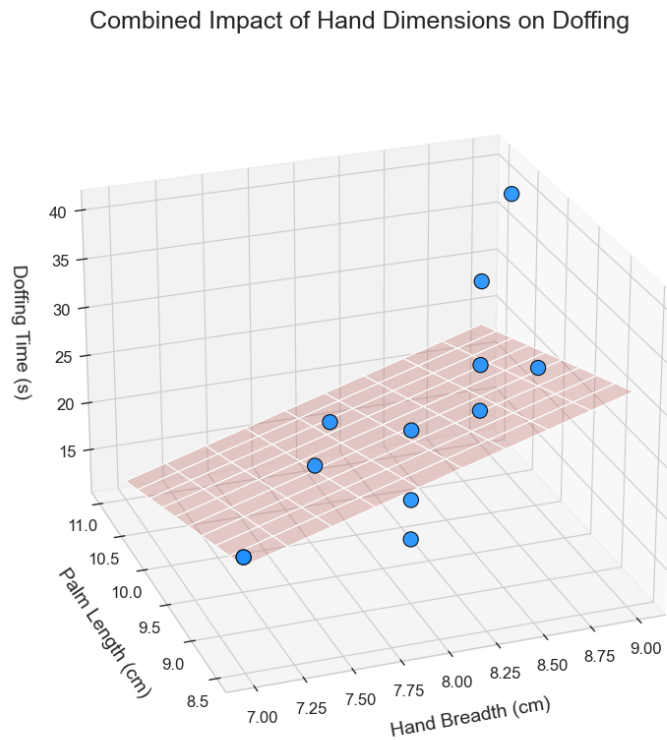


Figure 3.16: Relationship between overall hand dimensions and doffing time for healthy subjects.

Moreover, a positive trend emerged between both forearm ($r = 0.411, p = 0.185$) and upper arm ($r = 0.367, p = 0.240$) circumferences and donning time. While these correlations did not reach the threshold of statistical significance, the consistent positive r -values provide quantitative evidence that the fixed diameter sleeve introduced higher frictional resistance for users with larger arm volumes.

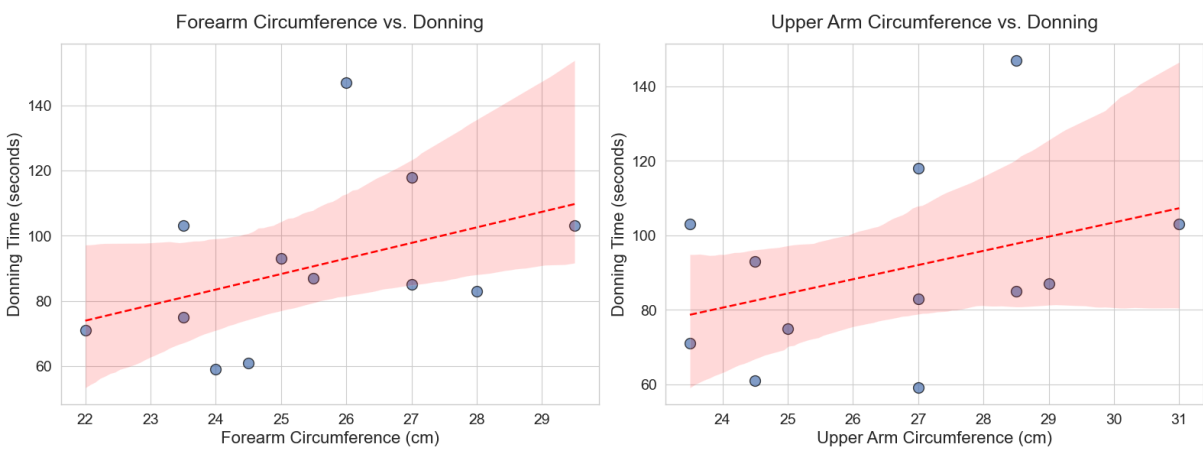


Figure 3.17: Relationship between individual arm dimensions and donning time for healthy subjects.

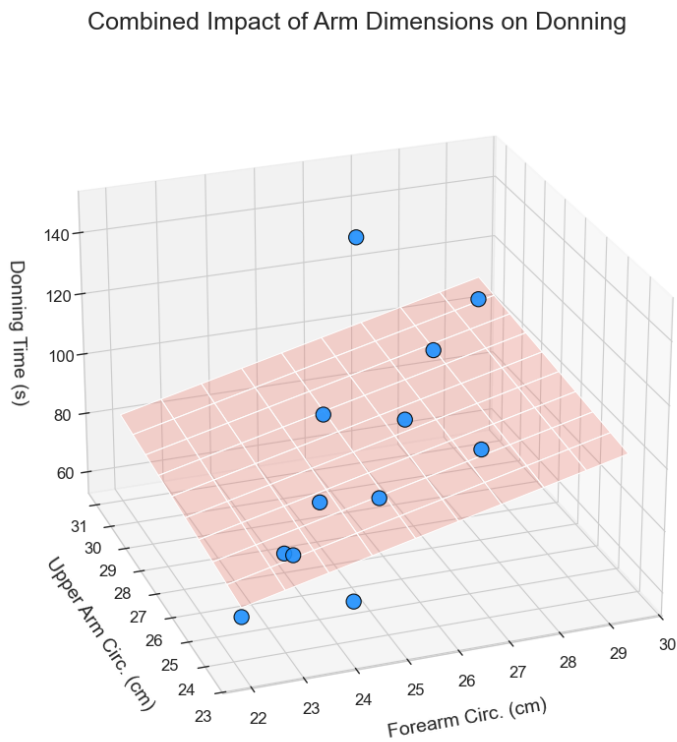


Figure 3.18: Relationship between overall arm dimensions and donning time for healthy subjects.

Obviously, this specific usability limitation could be easily overcome in the future by developing modular exosuit sizes.

3.1.5. HQ - Questionnaire

The subjective user experience was evaluated using a multi domain questionnaire, and the general outcome proved to be highly positive.

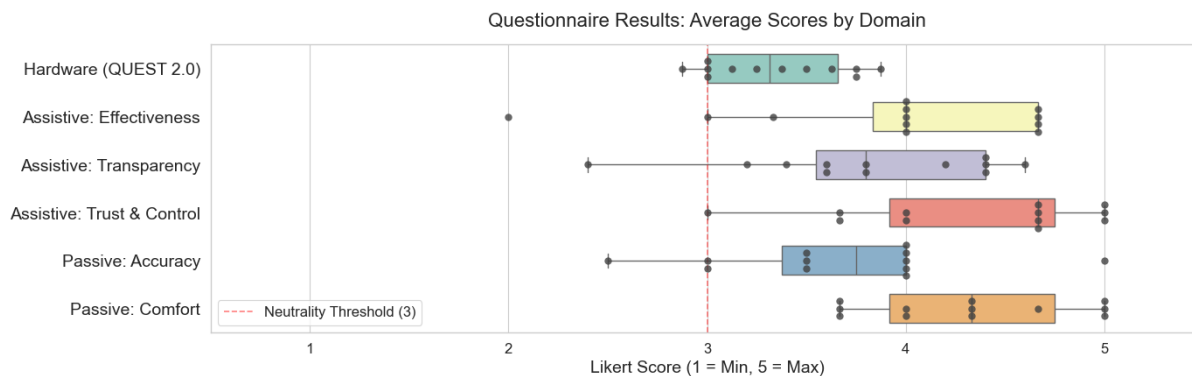


Figure 3.19: Results of the questionnaire for healthy subjects divided by topics.

The hardware characteristics, which were evaluated utilizing the standard QUEST 2.0 metrics, were appreciated more by female subjects than by males. This discrepancy is mainly due to the overall dimensions and the fact that the cable routing was patterned on a female reference user with average arm dimensions. Specifically for pronation and supination, the longer the user's arm, the smaller the useful tangential pulling component becomes, proportionally increasing the useless longitudinal force component.

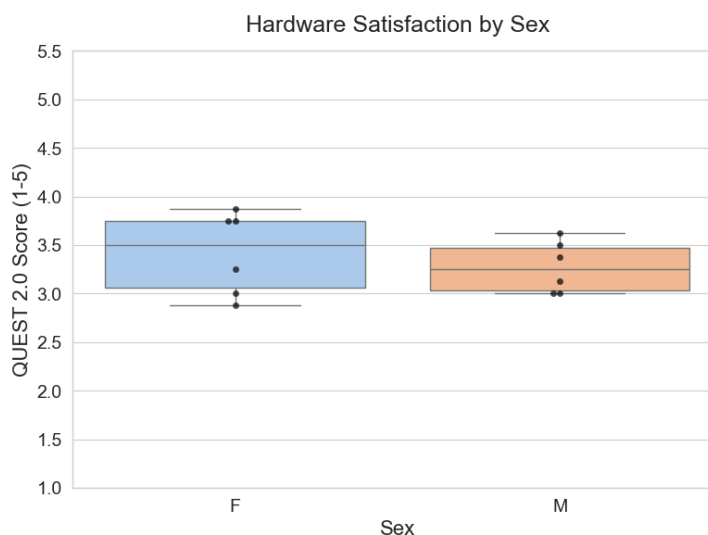


Figure 3.20: Hardware satisfaction scores of healthy subjects categorized by sex.

The general control preferences expressed by the users are summarized in Figure 3.21.

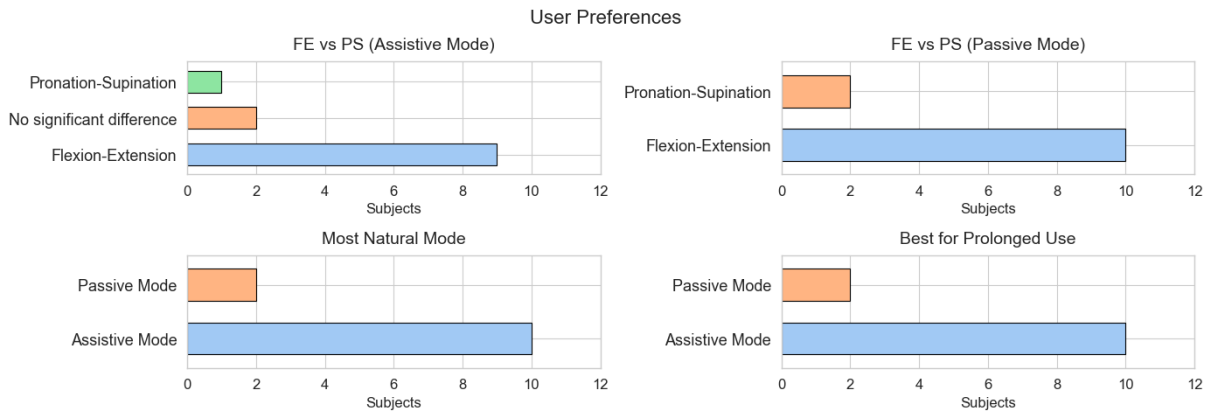


Figure 3.21: Mode preferences of healthy subjects.

Finally, the considerations provided by the participants in the open feedback section corroborated the quantitative results. Users consistently praised the intuitive nature of the active assistive mode, while clearly identifying the mechanical slippage that occurred during passive trials. Furthermore, the most recurring qualitative drawbacks cited by the participants included the restrictive sizing of the device and the difficulty of fastening the closures using only one hand.

3.1.6. Tests on patients

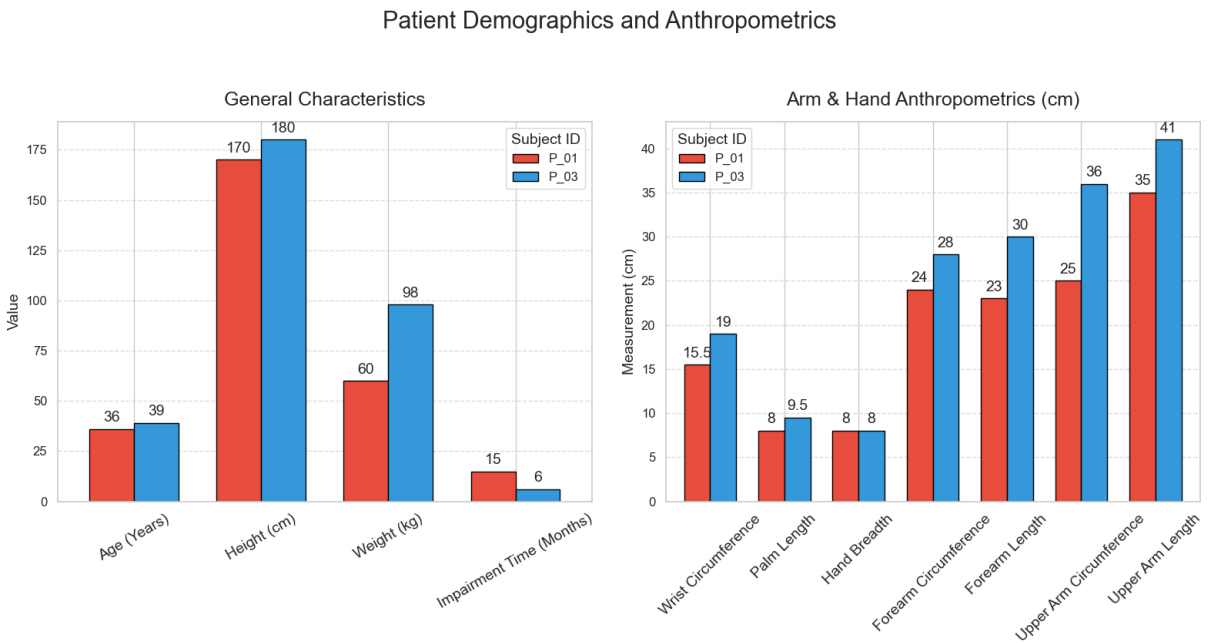


Figure 3.22: Demographic and anthropometric characteristics of the patients.

Figure 3.22 reports the demographic data and anthropometric dimensions of the participants. It is important to note that user P01⁴ presents arm dimensions that are significantly closer to the reference measurements utilized during the mechanical design phase. Consequently, as extensively discussed in previous chapters, subject P01 inherently benefits from a mechanical advantage regarding device fitting and force transmission efficiency compared to user P03, who exhibits larger overall anatomical dimensions.

P1 - RoM in different conditions

The initial analysis focuses on the maximum Range of Motion achieved by the patients under various conditions.

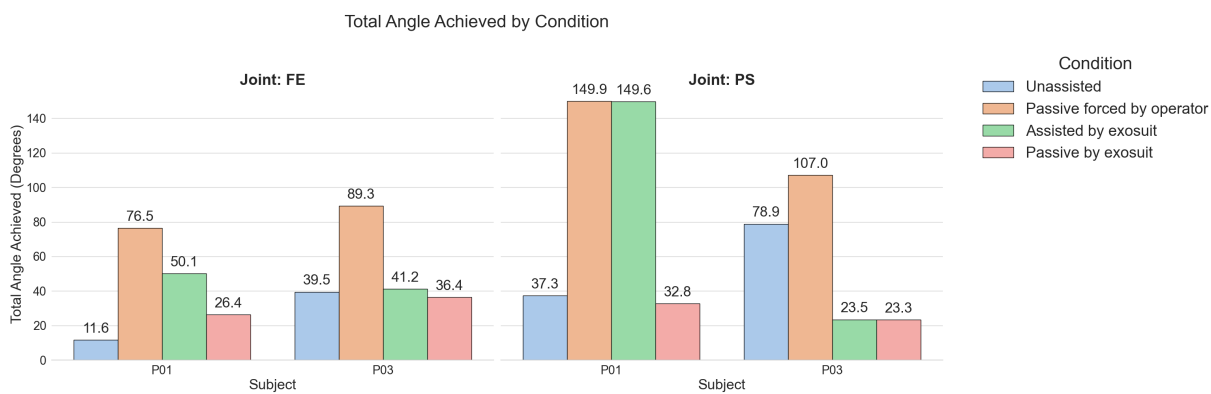


Figure 3.23: Range of Motion achieved by the patients in different conditions.

As shown in Figure 3.23, the active assistive mode for flexion/extension (FE) enabled both patients to increase their RoM from initial baseline values below the target necessary for Activities of Daily Living (50° , as defined in Table 2.2) to values close to or exceeding this functional threshold. This represents a remarkable outcome, demonstrating that the exosuit successfully fulfilled its primary objective for the FE degree of freedom.

Regarding the active assistive mode for pronation/supination (PS), the results varied significantly between the two participants. For user P01, the performance was excellent: the device enabled the forearm to achieve a kinematic range comparable to the passive RoM forced by the clinical operator, reaching an increase of four times over the unassisted baseline. Conversely, an anomalous behavior was recorded for user P03, whose assisted RoM was paradoxically lower than the unassisted one. This reduction clearly indicates the onset of severe physical fatigue between the first and second measurements; the pa-

⁴The participants are identified as P01 and P03. A second subject, initially designated as P02, was recruited but ultimately excluded from the testing phase because their motor impairment affected the left hand, whereas the current exosuit prototype is exclusively designed for the right arm.

tient's voluntary movement was subsequently insufficient to effectively trigger the device's assistance.

While the numerical values for the passive RoM might initially appear limited, a detailed kinematic analysis reveals a highly performative system behavior.

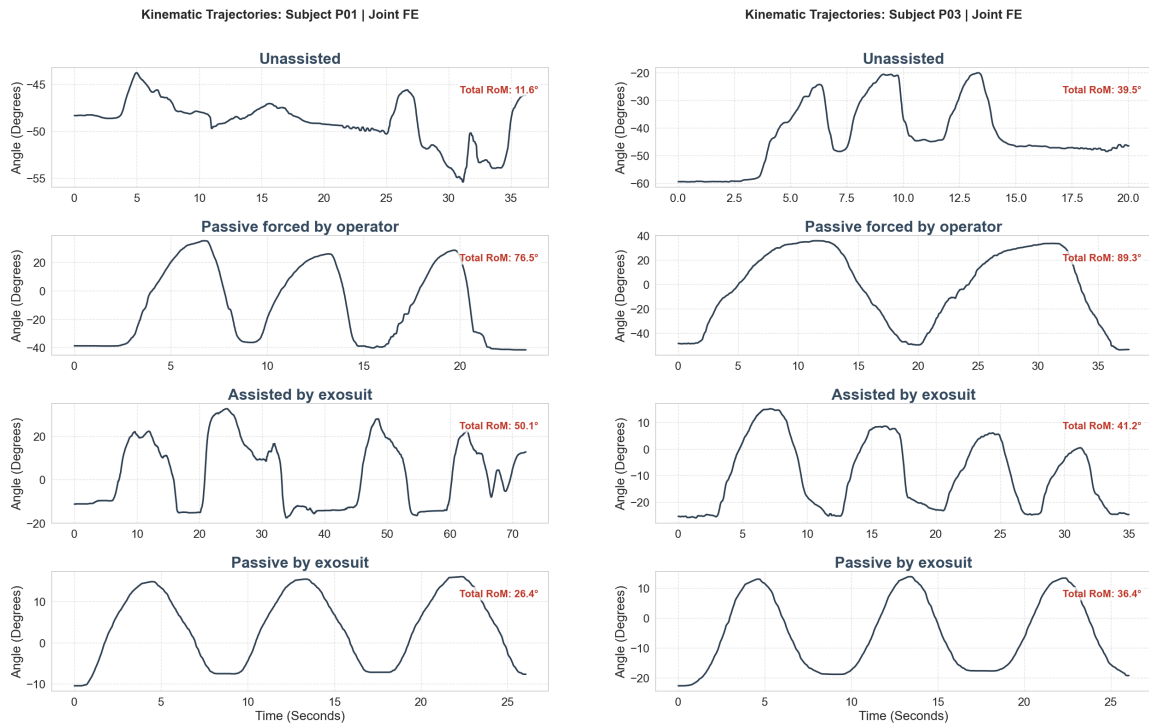


Figure 3.24: Kinematic trajectories of the wrist joint angle during the flexion/extension Range of Motion trials for both patients.

Observing the kinematic curves in Figures 3.24 and 3.25, the passive trajectories are remarkably clean. Therefore, even though the absolute RoM is not exceptionally large, due to the structural and dynamic reasons previously discussed in the trajectory tracking tests on healthy subjects, the device proves to be highly effective for executing rehabilitation trajectories and for smoothly mobilizing the patients' joints to maintain active flexibility.

As expected, the active assistive trajectories do not exhibit the same level of smoothness as the purely passive ones; however, they still display significantly less jerky behavior compared to the completely unassisted movements. The impact of muscular exhaustion is particularly evident in the assisted plot of user P03 in Figure 3.24, where the kinematic peaks become progressively smaller over time. This macroscopic and extremely rapid performance drop, explicitly corroborated by the patient's own oral feedback during the trial, is the direct cause of the already mentioned limited displacement in the subsequent tests, particularly the assisted pronation/supination and the functional activities.

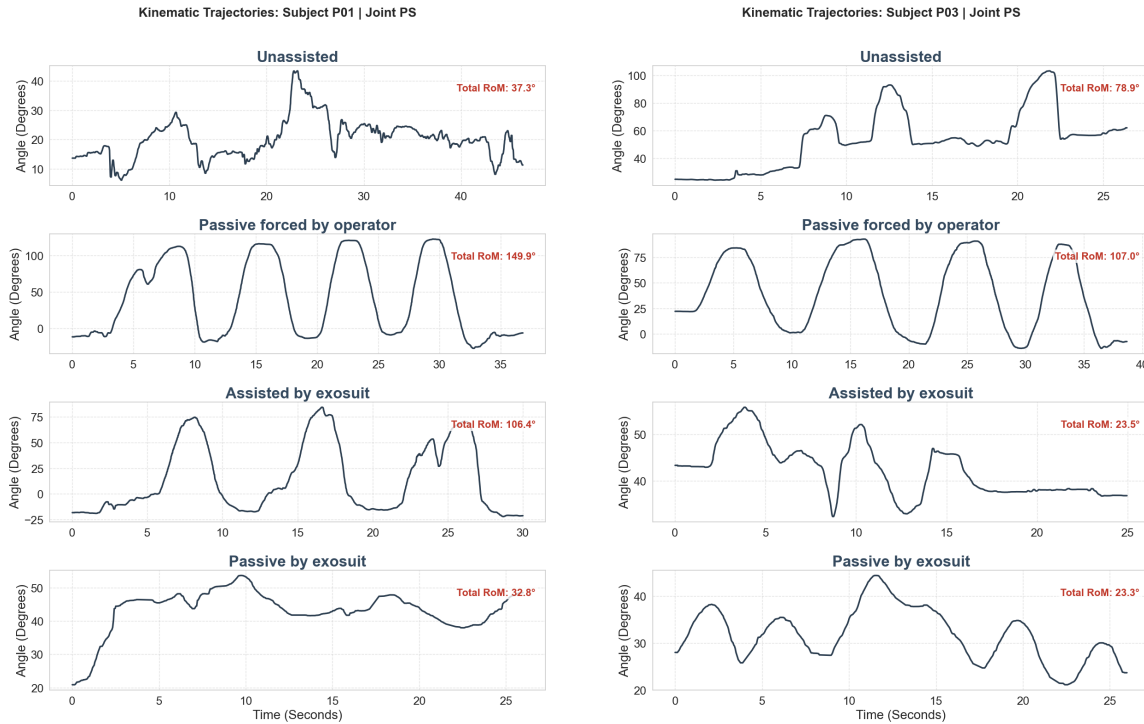


Figure 3.25: Kinematic trajectories of the wrist joint angle during the pronation/supination Range of Motion trials for both patients.

Finally, the passive PS curves appear slightly less clean than the passive FE curves. This discrepancy is attributable to the inherently less direct force transmission of the wrapping actuator geometry compared to the longitudinal FE routing, yet it still yields a sufficiently smooth overall rotation.

P2 and P3 - Functional activities

During the functional activity tests, user P01 successfully completed all tasks under all provided conditions. However, the execution relied heavily on compensatory strategies, indicating that the wrist was not the primary source of the overall kinematic displacement. Conversely, user P03 was unable to complete all the functional tasks in every condition. By this stage of the experimental protocol, fatigue had reached a critical level, preventing the execution of even slight voluntary movements and consequently failing to trigger the device's active assistance.

Despite these limitations, analyzing the recorded kinematic curves provides valuable insights into the device's impact on motion quality.

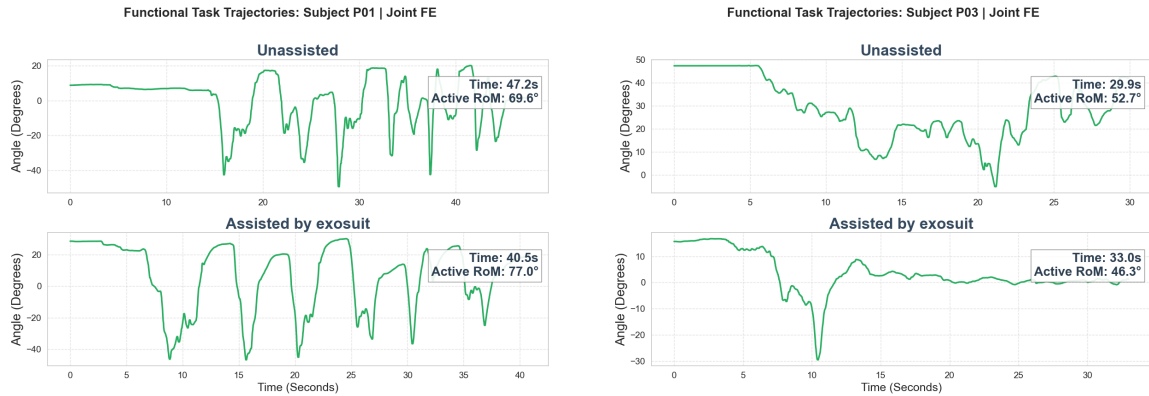


Figure 3.26: Kinematic trajectories of the wrist joint angle during the flexion/extension functional activity (button pressing) for both patients.

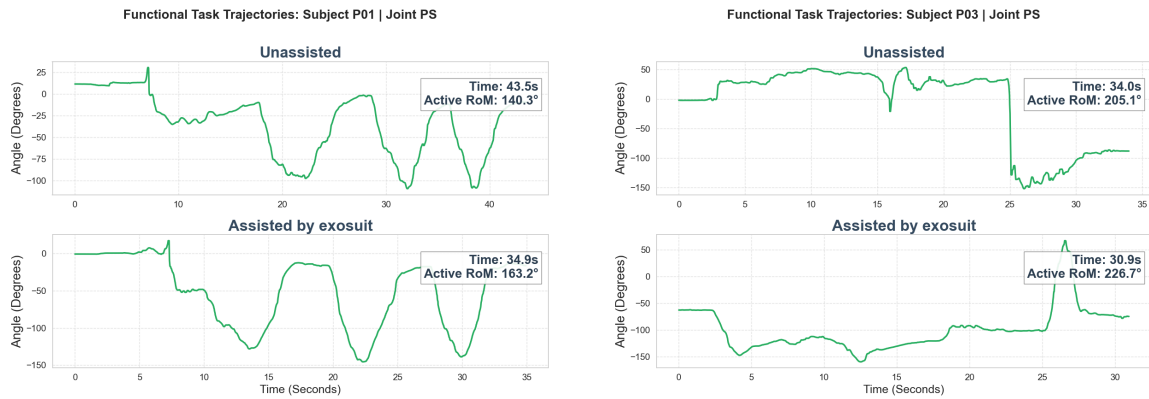


Figure 3.27: Kinematic trajectories of the wrist joint angle during the pronation/supination functional activity (eating with a fork simulation) for both patients.

For subject P03, the curves remain largely flat and irrelevant, directly reflecting the absence of triggered assistance. For subject P01, however, the data clearly demonstrate the benefits of the robotic intervention. The application of the active assistance rendered the functional movements noticeably smoother and more pseudo-periodic, while simultaneously reducing the overall time required to complete the three consecutive task repetitions.

3.2. Requirements and tests

Table 3.1 reports in detail whether each requirement was satisfied by the tests.

Table 3.1: Requirement verification summary.

Requirement	Metric	Target	Measured Value	Verification	Status
F1 - Assisted movements	Operating range	FE & PS motion supported	Functional tasks completed by patients	P2/P3 functional tests	Pass
F2 - Operating modes	Mode availability	Active assistive & passive modes	Both modes implemented	By construction	Pass
F3 - User interaction	Control interface	Start/Stop	GUI with hardware stop	By construction	Pass
N1 - Range of motion	Achieved RoM	$\geq 50^\circ$	50° target set in trajectory game	H3 trajectory tracking	Pass
N2 - Assistance capability	Passive RoM completion; active assistive compensation	100% passive; $\geq 50\%$ active assistive	$84.5\% \pm 18.6\%^a$ passive completion; active assistive verified by design	H3 trajectory tracking; by construction	Partial ^b
N3 - Controllability	Active assistive RMSE	$< 6.0^\circ$	$2.01^\circ \pm 0.80^\circ$	H3 trajectory tracking	Pass
N4 - Speed	Peak actuation rate	Muscle-like ($\approx 500\%/s$)	Limited by servo bandwidth	By construction	Partial ^c
N5 - Transparency	Unpowered residual RoM	No resistance	FE: $133.9^\circ \pm 11.7^\circ$ ($88.7\% \pm 9.2\%$ retained); PS: $151.2^\circ \pm 18.2^\circ$ ($86.0\% \pm 6.9\%$ retained)	H1 transparency test	Pass
N6 - Comfort of wearing	QUEST 2.0 mean score	> 3.0 (neutral)	$3.3 \pm 0.4/5.0$	HQ questionnaire	Pass
N7 - Ease of operating	Mean donning time	≤ 120 s	90.4 s ± 25.0 s	H4 donning/doffing	Pass
	Mean doffing time	≤ 60 s	23.4 s ± 3.8 s	H4 donning/doffing	Pass
N8 - Portability and weight	Collocated mass	< 300 g	394 g (0 g uncollocated mass)	Physical measurement	Partial ^d
N9 - Adaptability	User cohort diversity	Multiple anatomies	$N = 12$ healthy (6F/6M), $N = 2$ patients (1F/1M)	Demographic sizing	Partial ^e

Continued on next page...

Requirement	Metric	Target	Measured Value	Verification	Status
N10 - Emergency stop	System halt	Immediate power cut	Hardware emergency stop operational	By construction	Pass
N11 - Electrical safety	User insulation	Isolated power path	USB-C breakout separates motor power from logic	By construction + commercial servos are safe	Pass

^a Pooled across FE and PS trials (not average of averages).

^b Limited by fabric compliance and tendon slippage in the passive modality.

^c Constrained by standard servo motor dynamics; the theoretical HASEL/HALVE design is projected to satisfy this criterion by design.

^d The 394 g collocated mass slightly exceeds the 300 g target; however, the system carries zero uncollocated weight. Most tendon driven exosuits achieve low collocated mass at the expense of significant uncollocated hardware (motors, controllers, power supply). Here all components are body-mounted, making the total system weight comparable to or lower than the literature when evaluated on a whole system basis. The 300 g target was aspirational rather than a hard functional constraint.

^e A single prototype size was produced; full satisfaction would require manufacturing multiple sizes.

4 | Discussion

The experimental validation presented in the previous chapter demonstrates that the proposed soft exosuit successfully achieves its primary clinical objective: providing measurable and perceptible assistive support for wrist flexion/extension and pronation/supination in both healthy subjects and target patients. At the same time, the results reveal a set of mechanical and functional limitations that are inherent to the current prototype stage and that motivate a clear path for future development. This chapter discusses the key findings in terms of overall system performance, the asymmetry between the two degrees of freedom, the behavior of the different control modes, the challenges specific to the patient population, and the broader engineering and clinical implications of the design.

4.1. Overall performance and positioning in the literature

To contextualize the performance of the proposed exosuit within the broader landscape of soft wrist rehabilitation devices, Figure 4.1 presents a comparative spider plot against three representative devices, one for each of the main actuating technologies identified in the state of the art. The selected comparators are Bartlett et al. [5] for pneumatic actuation, Jeong et al. [30] for shape memory alloy actuation, and Choi et al. [13] for cable driven actuation. These three devices were chosen because they represent the best reported performance within their respective categories among the works surveyed in Table A.1.

Eight axes were selected to capture both the hardware properties and the functional capabilities that are most relevant to the clinical application:

- **DoF coverage:** the number of independently or jointly assisted wrist degrees of freedom;
- **Control modes:** the sophistication of the human-robot interaction strategy, from passive motion to proportional active assistive control;

- **Collocated weight:** the mass of the components worn directly on the limb, where a lower mass corresponds to a higher score;
- **Total portability:** the degree to which the system can be used without external tethered hardware, including compressors, pumps, or remote motor units;
- **Clinical validation:** the extent and rigor of the experimental evaluation on target patient populations;
- **Cost and affordability:** the accessibility of the system in terms of component cost and manufacturing complexity;
- **Actuation speed:** the bandwidth of the actuation system relative to the speed requirements of Activities of Daily Living;
- **Max torque output:** the peak assistive torque relative to the torque required for standard ADLs.

All scores follow a 1 to 5 scale, where 1 indicates poor or absent performance and 5 indicates top performance within the comparison group.

The proposed exosuit presents a unique combination of features that is not fully captured by any single comparator across all evaluated dimensions. Its most notable strengths are control mode sophistication, cost and affordability, and the breadth of clinical validation. The axes where it does not lead the comparison are collocated weight, where the SWA achieves a lighter wrist mounted footprint thanks to the low mass of SMA wire, and max torque output, where all comparators perform comparably.

The total portability score of the proposed device deserves a careful reading. The current prototype requires a wall power outlet to run the servo motors and a USB connection to a PC to operate the graphical user interface. It is therefore not fully portable in its present form. However, this is a deliberate engineering simplification of the prototype stage, not a structural limitation of the architecture: every component is collocated on the body, and a future version equipped with a small battery pack, possibly on the upper arm, and a compact embedded display would achieve complete autonomy with minimal redesign. This path to full portability is far more straightforward than for pneumatic systems, which require a compressor and valves that are fundamentally incompatible with portability, or for the SMA based SWA from Jeong et al., which requires an active heating circuit and a liquid cooling system to achieve frequencies comparable to ADLs.

The most important distinction between the proposed device and all three comparators is the combination of multi-DoF coverage with proportional active assistive control and

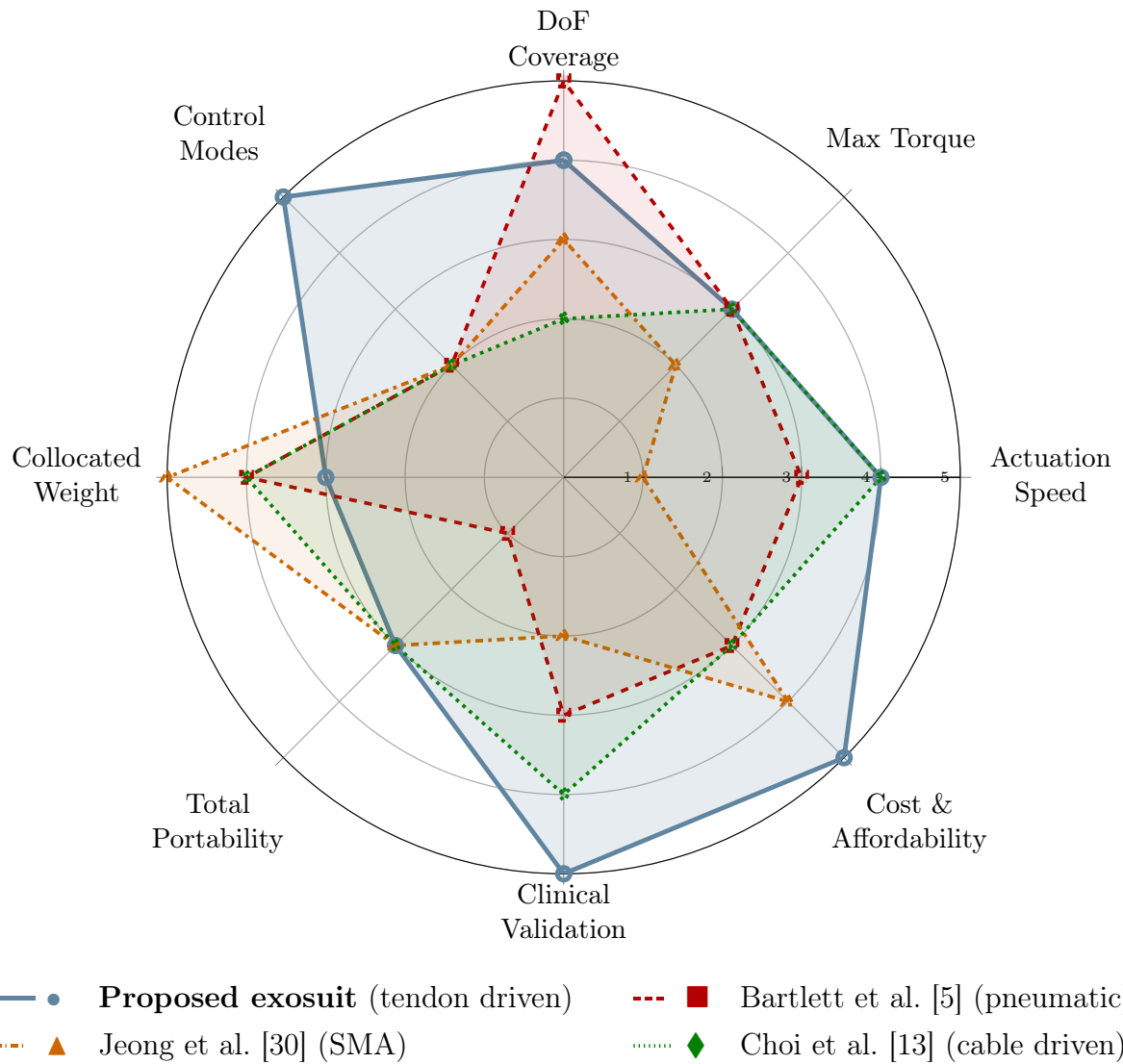


Figure 4.1: Spider plot comparing the proposed exosuit against three representative soft wrist devices, one per main actuating technology. Scores on a 1-5 scale (5 = best within the comparison group). Collocated weight is scored inversely to limb-mounted mass; total portability is scored by independence from tethered external hardware.

clinical pilot validation. Bartlett et al. achieve the broadest DoF coverage but rely entirely on open loop continuous passive motion, with no intention detection; the device cannot respond to the user's voluntary effort. The SWA by Jeong et al. is exceptionally lightweight and self contained, but its thermal actuation dynamics limit the operating frequency to approximately 0.5 Hz ¹, which is insufficient for dynamic daily assistance. The Exo-Wrist by Choi et al. shares the same class of electric motors and achieves real-time bandwidth, but it is designed exclusively for the oblique dart-throwing motion and does not address pronation and supination.

The proposed exosuit is the only device in this comparison that simultaneously covers independent FE and PS, implements proportional active assistive control responding continuously to the user's voluntary motion as well as feedback passive control, and has been pilot tested on patients with neurological impairment.

Taken as a whole, the prototype successfully satisfies the large majority of the defined requirements, with the notable exceptions of the collocated weight target and the actuation speed criterion, both of which are directly linked to the choice of servo motors as the actuating technology. Regarding the weight, the 394 g value slightly exceeds the 300 g target; however, most tendon driven and cable driven wrist exosuits reported in the literature achieve a low collocated mass at the expense of a significant non collocated one, including motors, gearboxes, controllers, and power supplies mounted remotely [12, 13]. In the present design, every component is body-mounted, meaning the total system mass is represented entirely by the 394 g figure. When evaluated on a whole-system basis, this value is in line with or lower than comparable devices in the literature [5, 13], making the 300 g target aspirational rather than a hard functional constraint.

Regarding cost, although affordability was not explicitly stated as a system requirement, it emerges naturally from the chosen components. The entire bill of materials relies on commercial servo motors, a widely available microcontroller, standard IMU breakout boards, and off-the-shelf textile components, resulting in a total hardware cost that is a fraction of that of clinical-grade rehabilitation devices, which typically reach several thousands of euros and are therefore confined to specialized hospital settings [63]. This cost profile makes the proposed platform particularly attractive for home rehabilitation and for contexts where continuous daily use is required, where expensive clinical devices are simply not a viable option.

¹Note that 0.5 Hz reflects the operating frequency of the actuator in isolation. The bandwidth of the fully integrated exosuit is unreported but it is presumably lower.

4.2. Asymmetry between flexion/extension and pronation/supination

A consistent pattern that emerges across all tests is the difference in performance between the flexion/extension (FE) and pronation/supination (PS) degrees of freedom, with FE systematically outperforming PS in terms of both perceived assistance intensity and trajectory tracking accuracy.

This asymmetry is rooted in the mechanical geometry of the two actuation strategies. The FE tendon routing is direct and largely longitudinal: the motor pulls a tendon anchored to the dorsal side of the hand, generating a moment arm that increases monotonically toward the middle of the extension range. The effect is immediate, powerful, and intuitive. The PS actuation, conversely, relies on a helical wrapping geometry that distributes the pulling force along the forearm. While this approach is biomechanically motivated by the distributed nature of pronosupination, which involves the rotation of the radius around the ulna along the entire forearm length, it is also geometrically sensitive to the fit of the garment on the user's arm. In particular, the longer the user's forearm, the smaller the useful tangential force component becomes relative to the useless longitudinal one, which explains the lower hardware satisfaction scores observed in male subjects and the qualitatively different PS performance between the two patients.

In the H3 trajectory tracking test on healthy subjects, this force asymmetry manifested in an unexpected way: the extension upward phase was the only movement direction where the assisted condition produced a larger tracking error than the unassisted baseline. Because FE assistance is strong, healthy users who applied the same voluntary muscular effort as in the unassisted trials ended up overshooting the target angle during the extension peak. This weakness of the control strategy indicates that the current architecture could be further optimized to seamlessly accommodate subjects with milder motor impairments. However, it is reasonable to expect that dedicated user training would mitigate this limitation in the long term, as individuals learn to subconsciously modulate their voluntary effort in response to the robotic intervention. Furthermore, in patients with genuinely reduced motor function, this overshooting effect does not occur, and the strong FE assistance is precisely what enabled P01 to exceed the ADL threshold of 50°.

One practical implication of this geometry-driven PS limitation is that the wrapping tendon sometimes produces a squeezing sensation along the forearm rather than a clear rotational pull, particularly in users with larger arm circumferences. A possible improvement, as can be inferred from the analogy with pneumatic contracting actuators, would

be to route the PS tendon inside a soft sheath anchored proximally and distally to the forearm. In this configuration, the tendon would pull against the sheath walls rather than against the skin, effectively reproducing a pseudo-contractile effect and concentrating the friction loss inside the sheath rather than at the tissue interface. This intermediate solution would partially bridge the performance gap without requiring a full transition to electrohydraulic actuation.

4.3. Control modes: strengths and limitations

4.3.1. Active assistive mode

The active assistive mode proved to be the most successful control strategy across the entire experimental campaign. The achieved RMSE of $2.01^\circ \pm 0.80^\circ$, well below the 6° design threshold, indicates that the kinematic control architecture is precise and that the power curve and velocity based intention detection effectively amplify voluntary motion without fighting the user. The clear preference for this mode expressed by healthy subjects in the questionnaire, both in terms of perceived naturalness and suitability for prolonged use, further corroborates this outcome.

An important and somewhat non obvious finding from the directional analysis is that the active assistive mode improved tracking performance in all movement phases except the extension upward phase, including the static horizontal holding phase and all PS directions. This is a nontrivial result for a wrist, where FE and PS are mechanically coupled through the shared joint. It validates the complete kinematic tracking strategy: while utilizing two independent IMUs is essential to correctly capture the complex spatial orientation of the hand, it is the subsequent Gram-Schmidt orthogonalization that mathematically decouples these measurements. Consequently, when the user exerts effort in one anatomical direction, the system effectively prevents cross-talk, ensuring that assistance is correctly isolated to that specific degree of freedom without introducing spurious commands to the other servo.

4.3.2. Passive mode

The passive mode results are more nuanced. The kinematic trajectories recorded during both healthy subject and patient trials are qualitatively smooth, which suggests that the underlying PI controller and tendon driven transmission are well-suited for slow, periodic rehabilitation movements. However, two mechanical challenges affected the quantitative completion metrics: fabric compliance, which prevents the tendon from fully translating

servo rotation into hand displacement, and mechanical slippage of the garment, which introduces trajectory offsets that the current PI controller is not designed to correct.

The PI controller design was intentionally conservative, with a capped integral windup and a moderate proportional gain, to prevent dangerous overcorrection on a compliant, wearable device. This is the correct choice for a first prototype used on impaired users. However, it reveals a structural limitation: standard linear controllers are poorly suited to handle large, sudden disturbances in a mechanically indeterminate soft system. A slippage detection mechanism, for instance based on a sudden discontinuity in the IMU signal, combined with a dedicated corrective re-tensioning phase, would be a meaningful improvement. Alternatively, a more aggressive integral gain could be permitted during a brief recovery window following the detection of a large offset, while keeping the conservative gains during normal operation.

An additional consideration is that the automatic pre-tensioning of the FE tendon during PS passive trajectories, which keeps the hand roughly horizontal during pronosupination cycles, is an example of how the coupling between the two degrees of freedom can be exploited rather than fought. This design choice is also reflected in the passive mode mapping, where the FE servo maintains a neutral holding position while the PS servo executes its trajectory, preventing the wrist from drifting into flexion due to gravity. This FE tensioning proved particularly important for patients also during unassisted PS movements, as it provided the necessary stabilization to hold the hand in a horizontal position, effectively acting as a supportive aid for their reduced motor control.

4.3.3. Teach and Play mode

The passive mode trajectory can be defined either via a mathematically imposed reference or through the Teach and Play mode. The Teach and Play modality was implemented and tested functionally but was not included in the formal experimental protocol, for reasons that are themselves worth discussing. Testing it on healthy subjects would have been methodologically problematic: healthy users naturally tend to perform at least part of the movement themselves during the Play phase, reducing the device's contribution and making the comparison with a purely passive trajectory unfair. Testing it on patients was not possible within the experimental sessions, as both participants showed significant fatigue by the time the passive RoM and functional activity assessments were completed, as clearly visible from the progressive reduction of kinematic peaks in the patient P03 plots.

This untested status reflects a broader challenge in rehabilitation robotics: the most

clinically relevant mode for severely impaired patients, who cannot provide a voluntary kinematic input and therefore cannot use the active assistive mode, is also the hardest to validate in a short experimental session. A dedicated clinical study, conducted in a structured rehabilitation setting over multiple sessions, would be the appropriate framework to evaluate Teach and Play, and the results on patient P01 during the functional activities provide a preliminary indication that the underlying trajectory execution infrastructure is solid.

4.4. Patient-specific considerations

4.4.1. Kinematic intention detection and its limitations

The active assistive mode requires the user to initiate a voluntary wrist movement that the device then amplifies. This kinematic based intention detection strategy works well for users with reduced but residual motor function, as demonstrated by patient P01. However, it is inherently circular for patients with more severe impairment: producing enough movement to trigger assistance requires precisely the motor capability that the device itself is meant to compensate. This limitation became apparent with patient P03, whose voluntary motion was insufficient to consistently trigger the device after the initial fatigue onset.

This is a well recognized challenge in assistive robotics. Control strategies that rely on the kinematic effects of motion (such as velocity derived from IMUs or encoders) inherently suffer from time delays, requiring the user to exert significant effort simply to initiate the movement [12, 53]. To break this circularity, the literature widely proposes the integration of surface electromyography (sEMG) signals as a more robust intention detection mechanism [38, 51, 73]. Because sEMG activity captures the electrical muscle spikes that precede actual joint movement, it relates directly to user motion intention [53]. This allows the device's controller to anticipate and trigger proportional mechanical support before kinematic motion is even observable, bypassing the physical effort required to overcome initial inertia [38, 51]. Integrating at least a surface EMG channel, often used in parallel with IMUs to evaluate spatial orientation [38], would be the most practical and clinically established extension of the current sensing architecture.

4.4.2. The role of fatigue and protocol sequencing

The patient results highlight the critical importance of experimental protocol design in clinical robotics validation. In the current protocol, the FE and PS RoM assessments

in multiple conditions preceded the functional activity tests, resulting in a progressive fatigue accumulation that strongly affected the later measures. The performance drop observed in patient P03, which was explicitly corroborated by the patient's own oral feedback during the trial, propagated into all subsequent tests and effectively precluded a meaningful evaluation of the functional activity outcomes for that subject.

For future clinical trials, the protocol should include mandatory rest intervals between test blocks and should consider counterbalancing the order of conditions across participants to decouple fatigue effects from the specific tasks being assessed. Shorter, more focused sessions spread over multiple days would also reduce the confounding effect of physical exhaustion in a patient population whose motor capacity is already limited.

4.4.3. Donning and doffing for patients

While healthy subjects successfully donned and doffed the device within the defined time thresholds, patients did not attempt to perform this operation autonomously. The donning procedure, which requires wearing the sleeve, the glove, and securing multiple closures, was challenging even for a trained external operator due to the finger spasticity that caused the fingers to resist extension during glove insertion. This represents a significant usability barrier for real world deployment, where the device would ideally be managed by the patient themselves or a non specialist caregiver.

Two complementary solutions could address this problem. From a hardware perspective, a glove that opens completely flat along one side and closes around the hand like a book, secured with a single large closure, would minimize the manipulation required during donning. Regarding the fastening mechanism, several healthy subjects also reported difficulty operating the buckle closures with their non dominant hand and suggested replacing them with hook and loop (Velcro®) fasteners or magnetic closures, which are easier to engage with one hand.

Additionally, a clinical operator from the partner rehabilitation center suggested that finger separators, similar to those commonly used in nail care, could be employed to passively extend the fingers during donning and even during operation. This suggestion is biomechanically grounded in the *tenodesis effect*: because the extrinsic finger flexor tendons are polyarticular and cross the wrist joint, passively extending the fingers increases the tension along these tendons, which in turn facilitates wrist extension and reduces the resistance that the closed fist spastic posture exerts against glove insertion [3]. Beyond donning, this principle suggests that integrating a simple finger separation element into the glove itself could reduce the overall extension resistance experienced during FE

assistance, potentially improving the RoM outcomes in spastic patients.

4.5. Design lessons and generalizability

4.5.1. The effect of anthropometry on performance

One of the clearest lessons from the experimental campaign is that the performance of a single size soft exosuit is inherently influenced by the anthropometry of the user relative to the reference design. The prototype was patterned on a female user with average arm dimensions, and the experimental results appear to reflect this alignment: female subjects generally reported higher hardware satisfaction scores, while empirical trends suggest that the PS tangential force efficiency may degrade with increasing forearm length, and donning times appear to increase with arm circumference. While these observations did not consistently reach the threshold of statistical significance in the current cohort, they represent meaningful indications of the challenges inherent in fitting a garment-based device to a diverse user population that deviates from the design reference.

The observed trends between hand size and doffing time, as well as between arm circumference and donning time, highlight the potential benefits of size grading in future iterations. Developing at least three sizes, for instance targeting small, medium, and large hand and forearm dimensions, would likely improve the performance distribution across the user population and expand the range of subjects for whom the device is clinically and operationally effective.

4.5.2. Acoustic noise as an unvalidated but relevant limitation

The use of servo motors introduces audible acoustic noise during operation. For a device intended for continuous daily use or for home rehabilitation, acoustic noise carries practical consequences in terms of social discretion and, for some users, cognitive comfort. This limitation is absent from electrohydraulic actuators, which operate silently, and should be acknowledged as an additional motivating factor for the technology transition explored in the feasibility study.

4.5.3. The absence of direct force measurement

Because the assistance capability requirement (N2) served primarily as a baseline for the mechanical design, its validation relied on the analytical torque model rather than direct measurement of the torque delivered to the wrist. While valid for a first generation

prototype, this leaves the biomechanical model experimentally unverified. Instrumented validation, such as routing an in-series load cell, would provide direct evidence of the actual assistive torque and allow the power curve exponent and velocity gain to be tuned against a ground-truth signal rather than empirical observation.

4.5.4. Mechanical transmission refinements

As a general consideration for future hardware iterations, the mechanical transmission could benefit from minor structural refinements. Adding targeted reinforcements to the textile architecture could help better distribute the applied loads across the forearm and hand. Furthermore, replacing the current servo arms with circular pulleys would eliminate the non linear tendon excursion caused by the varying lever arm, which currently degrades following a sinusoidal profile as it rotates away from the perpendicular optimal position. While these modifications would not fundamentally alter the system's dynamics or resolve the primary challenges inherent to compliant wearable robotics, they represent straightforward improvements to guarantee a constant transmission radius and slightly optimize overall mechanical efficiency.

5 | Conclusions and Future Developments

5.1. Summary of the research

This thesis presented the complete mechanical design, software development, and experimental validation of a novel tendon driven soft exosuit tailored for wrist assistance and rehabilitation. The wrist is a fundamentally crucial joint for the execution of Activities of Daily Living, as it is responsible for orienting the hand, transmitting loads along the arm, and stabilizing the fingers during grasping and manipulation tasks. Despite this clinical importance, the wrist remains severely understudied in the field of collaborative robotics compared to the broader arm or the hand alone. This discrepancy arises primarily from the biomechanical difficulty of securely anchoring a device to the forearm without impeding natural grasping capabilities, coupled with the fact that wrist impairments are frequently accompanied by broader upper limb deficits that complicate isolated study. Consequently, a rigorous review of the state of the art revealed a profound technological gap in the current landscape of wearable robotics. While rigid rehabilitative exoskeletons are well documented, they inherently suffer from excessive weight, kinematic misalignment with human joints, and poor user comfort. Conversely, the emerging field of purely soft wearable devices has largely been restricted to single degree of freedom systems, typically focusing only on wrist flexion and extension. This severe limitation prevents the execution of complex Activities of Daily Living, which naturally require the simultaneous coordination of multiple anatomical rotations.

The research conducted in this thesis addressed this gap by engineering a multi-DoF soft prototype capable of supporting both wrist flexion/extension, and forearm pronation/supination. By leveraging a tendon driven architecture and a custom electronic control unit based on two Inertial Measurement Units (IMUs), the project demonstrated the feasibility of creating a lightweight, comfortable, and articulated textile device.

The experimental validation demonstrated the functional capabilities of the system. Trials

conducted on healthy subjects verified the kinematic transparency of the garment, indicating that the unpowered suit imposes minimal restriction on natural movement, while also demonstrating precise trajectory tracking in the active assistive mode. Importantly, subsequent pilot tests involving neurologically impaired patients provided preliminary evidence of the device's clinical potential. The active assistive control strategy effectively managed the initial muscular rigidity of the subjects, assisting in reaching a functional range of motion that was otherwise unachievable. Furthermore, the passive modality showed promise for rehabilitation purposes. Visual inspection of the kinematic results highlighted the qualitative smoothness of the guided movements and the ability of the system to gently compensate for slight trajectory deviations. Because the controller was designed to be delicate, it prioritized safety and avoided applying excessive force to the impaired limb. This passive control was structured into two distinct modes: the passive trajectory mode, which is suitable for executing predefined rehabilitation paths, and the Teach and Play mode, which allows for a customizable therapeutic approach. Ultimately, this pilot work indicates that a multi-DoF soft exosuit is mechanically viable and holds promising clinical utility, offering a novel approach to address the identified gap in current rehabilitation technology.

5.2. Hardware and software improvements

While the current tendon driven prototype robustly satisfied the primary functional requirements, the empirical data and user feedback gathered during the experimental trials highlighted a few crucial areas for engineering optimization.

The most pressing hardware improvement concerns the power supply and system portability. Currently, the actuation unit relies on a connection to a standard wall power outlet, tethering the user to a fixed location. This is acceptable for rehabilitation but severely limits the possibility of assistance in ADLs. To transform the exosuit into a fully portable, completely standalone device, a lightweight battery pack should be introduced. An optimal anatomical placement for this power source is the upper arm. Since the exosuit already features an upper arm anchoring sleeve that is largely unoccupied, integrating the battery there would distribute the mass proximally. This would help reduce the perceived distal weight on the forearm, which helps prevent user fatigue over extended periods of operation.

To achieve full independence from external workstations, future developments should implement a hybrid control interface. Essential tasks like spatial calibration and operating mode toggling could be managed directly via a streamlined onboard interface on the

central control box. Conversely, more complex configurations and detailed data logging would be most efficiently handled through a dedicated mobile application, offering a user-friendly solution without the need for tethered hardware.

Regarding the control algorithms, the current active assistive mode relies on a kinematic trigger to detect the movement intention of the user. While effective for healthy individuals, patients affected by severe spasticity or muscular weakness often struggled to generate the initial physical displacement required to trigger the motor assistance. This limitation can be addressed by integrating surface electromyography sensors. By reading the electrical neural drive directly from the forearm muscles, the system could anticipate the motor intention of the patient milliseconds before any physical movement occurs, guaranteeing a much more seamless and responsive assistive intervention.

Finally, future hardware iterations should focus on some key mechanical refinements. Structural robustness can be improved by adding targeted reinforcements to the textile architecture to better distribute loads across the forearm and hand, alongside the adoption of stiffer, embedded routing channels to mitigate tendon slippage during high load tasks. To ensure a constant transmission radius and optimize mechanical efficiency, the current servo arms should be replaced with circular pulleys, eliminating non-linear tendon excursion. Furthermore, to enhance usability and clinical applicability, a modular sizing strategy should replace the one-size-fits-all approach, and donning should be facilitated through a fully openable glove design and magnetic or Velcro® closures instead of buckles. Lastly, the integration of finger separators is highly recommended to improve flexion/extension performance in subjects with spasticity, ensuring the device remains effective across a diverse patient population.

5.3. Future developments: towards soft actuation and clinical scaling

The tendon driven architecture successfully served as a robust platform to validate the proposed control strategies and biomechanical textile interfaces. However, the reliance on traditional rotary servo motors imposes inherent limits on peak speed, acoustic discretion, and structural compliance. While these motors cannot fully match the strain rates of biological muscle, they have provided the necessary foundation to prove the device's functional utility.

Consequently, future efforts could be structured along two parallel development tracks.

The first would be a translational effort: the pre-industrialization of the existing tendon-

driven prototype. This should involve refining the structural reliability of the closures and the textile harness, optimizing the tendon routing components, and streamlining the electronic housing to evolve the device from a laboratory grade prototype into a more durable, fully portable, user friendly instrument. Parallel to this, a larger scale clinical study with a broader patient cohort would be essential to statistically confirm the observed benefits, particularly in managing chronic fatigue and ensuring consistent performance across diverse neurological impairments.

The second track could focus on the longer term paradigm shift: the transition to electrohydraulic actuation. The theoretical feasibility study presented in this thesis regarding the implementation of electrohydraulic actuation introduced a highly promising alternative to traditional actuation. By analyzing the integration of Hydraulically Amplified Self-healing ELectrostatic (HASEL) actuators and Hydraulically Amplified Low-Voltage Electrostatic (HALVE) actuators, this research explored new, viable pathways for the future design of upper limb wearable robotics. The potential advantages of adopting such an electrohydraulic system are substantial. These actuators remove the reliance on rigid electromagnetic motors, rotating spools, and complex tendon transmissions. Instead, they use fluid filled flexible pouches that contract under high voltage, offering silent operation and an intrinsic structural compliance that closely mimics the biomechanical properties of human muscle tissue. Furthermore, these smart materials operate at strain rates that far exceed those of traditional servos, directly addressing the speed limitations encountered in the current prototype. The feasibility study indicated that routing these actuators along the exact anatomical lines of the wrist would result in a highly conformal, low profile garment. Translating this feasibility study into a physical electrohydraulic prototype is the ultimate future development of this project.

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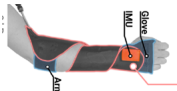
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DoF notation: FE (Flexion/Extension), RU (Radial/Ulnar Deviation), PS (Pronation/Supination), AA (Abduction, Adduction); Combined (C) in case of different DoF acting together, Independent (I) in case of DoF acting one at a time, (-) if not clearly mentioned.

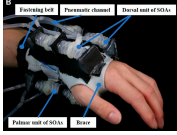
Control indicates Active (A), Passive (P), Active Assistive (AA), Resistive (R), or Not Specified (-).

Sensor refers only to those used for control: the sensors used in the tests are not considered.

Author et al. [Ref]	Picture	DoF	Control	Actuators	Sensors
Pneumatic Actuators					
Bartlett et al. [5]		FE, RU, PS (I)	P, AA	McKibben (contractive)	Pressure
Ferroni et al. [21]		PS	AA	Pneumatic fabric-based twisting (con- tractive)	Pressure, IMU

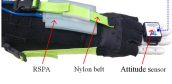
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Table A.1 – *continued*

Author et al. [Ref]	Picture	DoF	Control	Actuation Type	Sensors
Liu et al. [43]		FE, RU (-)	P	Soft origami-patterned actuators (SOAs) (expansive)	Pressure, IMU
Realmuto et al. [57]		PS	P, AA	Twisting pneumatic (twisting)	Rotary Hall effect sensor
Sasaki et al. [59]		FE	AA	Rotary soft pneumatic (expansive)	Flex sensors (wrist angle), pressure
Zhu et al. [76]		FE	AA	Thermoplastic airbag actuators (expansive)	Pressure, IMU
Schaffer et al. [60]		FE, RU (C)	P	Fabric pneumatic artificial muscles (fPAMs) (contractive)	IMU, Pressure

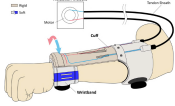
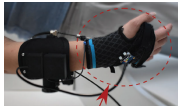
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Table A.1 – *continued*

Author et al. [Ref]	Picture	DoF	Control	Actuation Type	Sensors
Chen et al. [11]		FE, RU, elbow FE (-)	P	Reinforced soft pneu- matic (con- tractive)	Attitude sensor, thin film force sen- sors
Swagata et al. [16]		FE, PS (-)	AA, R	PAMs	Stretch sensors, pressure
Shape Memory Alloy (SMA) Actuators					
Jeong et al. [30]		FE, RU (C)	AA	SMA coil spring ac- tuators	Encoders
Jeong et al. [31]		Elbow FE, PS (C)	P, AA	SMA coil spring bun- dles + coolant vessel	Load cell, absolute encoder
Wahid et al. [70]		FE, RU (I)	P	SMA coil spring ac- tuators	IMU

Cable / Tendon driven*Continued on next page*

Table A.1 – *continued*

Author et al. [Ref]	Picture	DoF	Control	Actuation Type	Sensors
Lessard et al. [40]		PS, Elbow FE, Shoulder FE and AA (C)	P, AA	Micromotors + spectra fishing lines (Bowden cables)	IMUs
Su et al. [68]		PS	P	EC motor + braided wires	Encoder, Torque sensor
Alapati et al. [2]		FE, RU, PS, Elbow FE (C)	P, AA	7 tendons	-
Chiaradia et al. [12]		FE	P, AA	Brushless motor + Bowden cables	IMU, Load cell, En- coder
Choi et al. [13]		FE, RU (C)	P, AA	DC motor + tendon + active anchor	Load cell

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