Study and development of sensorimotor interfaces for robotic human augmentation

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Alla mia famiglia, amici compresi.

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Abstract

This thesis presents my research contribution to robotics and haptics in the context of human augmentation. In particular, in this document, we are interested in *bodily* or *sensorimotor augmentation*, thus the augmentation of humans by supernumerary robotic limbs (SRL).

The field of sensorimotor augmentation is new in robotics and thanks to the combination with neuroscience, great leaps forward have already been made in the past 10 years. All of the research work I produced during my Ph.D. focused on the development and study of fundamental technology for human augmentation by robotics: the *sensorimotor interface*. This new concept is born to indicate a wearable device which has two main purposes, the first is to extract the input generated by the movement of the user's body, and the second to provide the somatosensory system of the user with an haptic feedback.

This thesis starts with an exploratory study of integration between robotic and haptic devices, intending to combine state-of-the-art devices. This allowed us to realize that we still need to understand how to improve the interface that will allow us to feel the agency when using an augmentative robot.

At this point, the path of this thesis forks into two alternative ways that have been adopted to improve the interaction between the human and the robot. In this regard, the first path we presented tackles two aspects conerning the haptic feedback of sensorimotor interfaces, which are the choice of the positioning and the effectiveness of the discrete haptic feedback. In the second way we attempted to lighten a supernumerary finger, focusing on the agility of use and the lightness of the device.

One of the main findings of this thesis is that haptic feedback is considered to be helpful by stroke patients, but this does not mitigate the fact that the cumbersomeness of the devices is a deterrent to their use. Preliminary results here presented show that both the path we chose to improve sensorimotor augmentation worked: the presence of the haptic feedback improves the performance of sensorimotor interfaces, the co-positioning of haptic feedback and the input taken from the human body can improve the effectiveness of these interfaces, and creating a lightweight version of a SRL is a viable solution for recovering the grasping function.

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Sensorimotor interfaces: new possibilities to augment humans

> *Technology should not aim to replace humans,*
> *rather amplify human capabilities.*
>
> Doug Engelbart

Human augmentation has been done from the beginning of history in its wider definition. Humans have dressed themselves with ornamental clothing to raise their social status, or invented chemicals to improve their physical performance or alter their emotional condition. The reader can interpret *human augmentation* as an abundant variety of concepts. Indeed, if we think about it, from a gnoseological point of view, also spirituality can be seen as a form of enrichment of the human condition. Hereupon, we will narrow its meaning down to the essential perspectives and concepts which form the pillars of this Ph. D. thesis.
The concept of human augmentation includes a very wide group of applications if we take into consideration human-centered technologies developed in robotics. In the list of applications we find exoskeletons, both as a device to rehabilitate and enhance humans. Also Brain-Computer Interfaces (BCIs) augment human capabilities if we consider the possibility of using brain electrical activity to drive a car or write on a computer [1].

In [2], Raisamo *et al.* gave the definition:

> Human augmentation is an interdisciplinary field that addresses methods, technologies and their applications for enhancing sensing, action and/or cognitive abilities of a human. This is achieved through sensing and actuation technologies, fusion and fission of information, and artificial intelligence (AI) methods.

Given that the augmentation of the human is intended or as a physical extension of the body which enables the enhancement of bodily skills or an extension of the set of cognitive possibilities of the person, in this dissertation we focus only on the robotic augmentation of humans by using *Supernumerary Robotic Limbs*, and thus we will consider the *sensorimotor* or *bodily* augmentation [3]. Supernumerary robotic limbs are wearable robots, built to achieve human sensorimotor augmentation, typically legs, arms or fingers, but sometimes also tails or scarfs [4]. The sensorimotor augmentation is the enhancement of human bodily skills by using SRLs, robots which add up with the human body, augmenting its degrees of freedom. This can happen in a seamless way only if the technologies developed to this aim take into account the human *somatosensory system*, which is responsible of informing us about the environment through touch and proprioception but also through the feelings of temperature, pain and many more. Indeed, as we will see in detail in Chapter 1, sensorimotor augmentation can be succesfully achieved by closing the loop between the SRL and the human somatosensory system [3], making the coordination between the human and the robot possible while the human is in control of the robot and moves instinctively. To this end, receiving *feedback* from the robot is key to the sensorimotor augmentation, that is why the ideal *sensorimotor interface* enables the human to control SRLs in a natural fashion both extracting information intelligently from the human body and conveniently stimulating the somatosensory system of the user.

In the rest of this brief introduction, we will see how we humans have come to the idea of augmenting ourselves and how in the modern era we are facing these challenges of humanity's near future, accompanying the reader to take ownership of the fundamental concepts addressed in this dissertation.

Human augmentation brings up to the mind acheiving superhuman possibilities. This is due to the immense popularity of extraordinary stories present in human literature, both classical, if we think to the Ovid's poem *"Metamorphōsēs"* in which Dedalus built wings out of wax and feathers for himself and his son Icarus, and in modern era where there are countless examples of humans with augmenting devices, recalling for example Tony Stark's Iron Man suit. The history of mankind is studded with real life examples where humans tried to increase thier physical abilities by resorting to machines, as outstanding exmaples we can recall Hero of Alexandria which invented the *aeolipile*, considered to be the first recorded steam engine, or Leonardo da Vinci which conceptulized both the bicycle and the helicopter [5]. Generally, all kind of tools that were invented in human history to increase the possibilities of achieving better life conditions can be interpreted as augmenting. Easily, the reader can think at the invention of the wheel as an actuation technology, or agriculture as a fusion of informations, and all these great inventions contributed to revolutionize the way humans lived and evolved on earth. Technology augments human ability by helping individuals do things they could not do before. Ron Fulbright in *"A Brief History of Human Augmentation"* [6] stated: "Humanity creates technology and technology changes the trajectory of human history".

A key example of this phenomenon is the industrial revolution, which contributed to exponentially increase our manufacturing possibilities. During the Renaissance, mechanical technology advanced mainly from a theoretical and mathematical point of view, and consequently manufacturing processes did not allow for the creation of key tools for human augmentation. With the advent of industrialisation of processes and the emergence of electronics and information theory, the idea of extending human capabilities by leveraging these modern technological breakthroughs began to form among scientists across the globe. While for prostheses there are records of devices made in the 16^{th} century [7], for the first example of human augmentation by robots humanity had to wait until 1890, when a patent documented a human-powered exoskeleton [8]. Although the idea of augmenting the human body with a device was born, more consisent examples of human augmentation start to appear in 1960s with Hardiman, the first example of a practical powered exoskeleton co-developed by General Electric and the US Armed forces [9].

In modern times, scientists have begun to categorise different wearable robots by distinguishing them into various macrocategories: rigid and soft, fully actuated and under-actuated, prostheses, exoskeletons and so on [10, 11]. But only in the last 10 years SRLs were properly distinguished from exoskeleton and prosthesis [3, 12].

In the next chapter, we will propose to the reader an overview on the main technologies exploited in this thesis, functioning as the background for the next chapters. Moreover, we will analyze how SRLs found space in the robotics panorama by looking at the state-of-the-art of such new devices. We will also see how, to link the human and the augmenting device (*i.e.* the SRL), we need a third technology: the sensorimotor interface.

Chapter 1

Background

We can never be right, we can only prove we're wrong.
Richard Feynman

The purpose of this first chapter is to provide the reader with a background on the main topics taken into consideration in this thesis: the concept of sensorimotor interface, the role of the sensorymotor system in human augmentation through supernumerary robotic limbs and the somatotopic mapping. Also, an overview of state-of-the-art devices, techniques, and findings is presented as it is instrumental in locating the contribution of this work into the panorama of these fields as it is today.

1.1 Scientific background of sensorimotor augmentation

In this Section we will consider some fundamental concepts of the sensorimotor augmentation, explaining what are supernumerary robotic limb and how they interact with our somatosensory system. In the second part of this section we will present some of the main technologies which are essential in a sensorimotor interface: input and feedback devices.

1.1.1 What is sensorimotor augmentation?

Augmenting the human bodily possibilities is not an easy task, and certainly presents a number of very difficult challenges. Our brain has a very precise idea of what parts contitute our body, and neurologists called this phenomenon *body schema*. This human mechanism gives us a postural model that allows us to know where our limbs are, involving both brain processes, sensory and proprioceptive stimuli, and by integrating them with vision [13]. Some studies suggest the possibility of an extended body schema, which would be the mechanism through which we are able to incorporate external objects to our body schema [14]. A much more solid result we can find in literature is the importance of the body schema in tool use. It has been empirically shown that objects can be integrated into an extended sense of the body by performing experiments on behavioural performance of normal and brain-damaged humans, by experimenting with afterimage of the body, and by single-neuron recordings in the monkey brain [15, 16]. Nowadays, it is widely accepted that the body schema is altered by the use of tools that extend our physical body [17], and these studies are functional to human sensorimotor augmentation by robots, laying the foundation for the integration of robots and the human body. Most of these studies use mechanical grabbers which are directly operated by hands, and it is well known that the hand is both the more dexterous part of our body and one of the most sensitive [18]. If and how a wearable robot could be represented in our body schema is still object of study [19, 20] and scientists are probing this phenomenon by looking at the somatosensory system. As stated in the introduction, sensorimotor augmentation is the enhancement of human bodily skills through the use of SRLs, robots that increase the active degrees of freedom of the human body. SRLs increase the human body workspace, but to achieve the sensorymotor or bodily augmentation, the user must feel the sense of agency, subjective awareness of initiating, executing, and controlling robot actions. In other words, we need more than a robot, otherwise we fall in the realm of human-robot collaboration, and thus, to achieve sensorimotor augmentation, the robot must be "felt" by the human. In order to do that we must add haptic feedback and including the human motion for giving inputs to the SRLs: this is necessary to make our somatosensory system take control of them.

1.1.2 Supernumerary robotic limbs

Wearable robots known as supernumerary robotic limbs are developed to improve humans' sensorimotor abilities. We can say that we augment the human body with a SRL if we increase its degrees of freedom (DoFs), which we could also call supernumerary DoFs. SRLs can be used to substitute for lost functions in patients with motor deficiencies and, more broadly, to enhance humans' sensorimotor ability to interact with their environment. SRLs differ from other augmentative technologies like exoskeletons and prosthetics, since they aim to add extra limbs that can be controlled simultaneously with biological limbs, rather than enhancing existing limbs or replacing missing limbs. Thus we can say that the aim of SRLs is to provide extra outputs to the sensorimotor system of the subject. As we saw, usually tools can increase human performances, for example a screwdriver will enable us to screws very tightly two objects together by applying a reduced torque with our hand. On the other hand, we are limiting some DoFs of our body since we are dedicating them to handling and using a tool. Instead, with SRLs we have more active DoFs than our natural ones, and just with that augmentation we can think of stabilizing and positioning a beam with our human arms and hands *while* the SRL is in charge of screwing the two pieces. This task requires more than two limbs, and its impact is great, considering the savings in terms of human resources.

Now let's see some state-of-the-art SRLs to understand where the research this field is aiming. The main functions of a SRL is augmenting bodily skills, and thus activities like walking, grasping, but also stabilizing and reaching objects, and for that purposes most of them are categorized in review papers as supernumerary legs, arms, hands and fingers [21, 3] or presented as wearable devices mounted on the human arm, forearm, leg and other parts of human limbs [22]. In line with the literature now the reader will be guided among those categories.

Supernumerary legs Supernumerary legs are devices which are made to enhance human walking and balancing and to compensate for missing functions in people with lower limb impairement which retain the whole leg. In [23], Parietti *et al.* built two extremely lightweight extra legs with 2 DoFs each which are controlled by using pectoral and abdominal muscle activities, and saw that linking sEMG to the velocities of such robots is the best among three possible strategies. The same authors in [24, 25] built also a device composed of two extra legs with 3 DoF each with the intent of augment human balance, both in a stance (static) and in a gait (dynamic) situation. Another more complex example of supernumerary leg can be found in [26]. In this work, authors presented a pair of 6 DoFs fully actuated robotic legs, which are copuled with the human on the back of the body and allow the human to assume poses that are impossible to balance with the use of natural limbs alone. This device is weighty but supports its own weight by unloading it onto the ground. It consists of powerful motors that allow the human to lift very heavy loads. Whit a similar function but a much simpler implementation in [27], authors developed chest-mounted extra limbs which acting as two extra legs enable the user to work near the ground avoiding unwanted spinal cord flexions or extensions.

Supernumerary arms Supernumerary arms are built to enhance the workspace of the human and add that complexity which allows humans do things that required more than one person. These devices are equally frequent in the form of a single supernumerary arm or as a pair of robotic arms and they're often worn on the body of the user. Supernumerary arms allow the user to reach areas which otherwise would be unreachable for the human arms, extending the person's workspace. One of the first examples of robotic arm purposely designed to be bodily augmenting was developed by Asada *et al.* [28], in which the coupling between the human operator and the SRL was described using a biomechanical model and results showed a reduction in the force generated by the human. In later publications [29, 30, 31], the concept was further explored in depth, with a focus on the control and dynamic analysis of SRLs. The device they built is a novel type of wearable robot to assist workers in the assembly of an aircraft fuselage. It has two extra arms for holding objects, pressing them to a fixture, and guiding and supporting human hands. The SRL kinematic configuration and joint torques that minimize the human workload were identified using an optimization method. A more recent example of supernumerary robotic arm can be found in [32], where Saraiji *et al.* presented two wearable arms, complete with a robotic hand each. This device is controlled by exploiting the movement of feet, and the possibility of co-operation between the two natuarl limbs and the two supernumerary arms can enable to perform complex tasks that are difficult for a single person to complete. In the same year, Ciullo *et al.* [33] proposed a soft robotic third arm-hand system capable of reducing the overloading and the vibration transmission to the human natural upper limb. Despite the field being new, supernumerary arms are very popular and we can see in literature that also other alternative direction start to arise in their development. Surely one direction which stands out is the implementation of wearable soft robots [34, 35]. This kind of robots continuously deform when actuated and have the big advantages of adapting their shape to environmental constrains and that of being safer for the human [36].

Supernumerary fingers and hands Supernumerary fingers are robotic fingers added to the human hand and usually they are called sixth finger or third thumb, whereas devices including more than one robotic finger are called third hands. Adding a single extra robotic finger to a human hand is the simplest, yet most effective, way to improve its capabilities. The finger can increase the grasping capabilities of the human hand by extending its workspace to a larger volume. In 2014 was published the first paper to introduce the concept of supernumerary finger [37], it was a fully actuated robotic finger with 4 DoFs, three reproducing the flexion and extension of the human phalanges and one for the ab/adduction. The device could be worn as a wrist bracelet by means of an elastic band. In successive iterations, this device underwent several different change of path: it was complicated by the addition of force sensors [38], and then simplified to remove as much complexity as possible [39]. In its more recent iteration, the device developed by Prattichizzo *et al.* is underactuated (tendon driven) and composed of both soft and rigid parts, feature which allowed also the definition of the finger trajectory for a non fully actuated robot [40]. Even though this device was intended to augment the grasping capabilities of healty individuals, it also had a great success in compensating the missing grasping function in stroke patients that experienced hemiparesis [39]. Another notable example of supernumerary finger is the robotic thumb presented in [41], it is worn on the ulnar side of the hand and is shaped like a human thumb, it has two DoFs dedicated to the flexion and one to the ab/adduction. Similar to this device, another third thumb has been designed by Clode [42], and was recently object of neurocognitive studies aiming at manipulation augmentation [43]. In [44], authors presented a wrist worn device composed of two fully actuated fingers which are placed to both the radial and the ulnar side of the wrist. With this device the hand has 7 fingers and this led authors to define bio-artificial grasp synergies. Another notable example of device which added two fingers to the hand if the double soft-sixth finger [45], in which authors underactuated two parallel kinematic structures composed of rigid and flexible materials by means of a differential mechanism. A full antropomprhic robotic hand was presented in [46]. This device called the SoftHand is an underactuated robotic hand with one degree of actuation (one motor) and 19 DoFs, and such an implementation removes the need for complex control algorithms. The supernumerary hand can grasp object autonomously from the human hand, opening to the possibility of functional substitution in subjects with hand control deficiencies. Among all these amazing devices, the Robotic sixth finger developed in Siena is one of the more projected in the field of sensorimotor augmentation: started as a full actuated finger composed of 4 servomotors, nowadays this device has evolved through several iterations, and complexity has been moved from the device to the input interfaces.

Before studying how this new class of devices can be merged with the human body, it is necessary to understand what the somatosensory system is and which role plays in the sensorimotor augmentation.

1.1.3 Somatosensory system

The somatosensory system is a network of neural structures in the brain and body that mediate touch, temperature, body position (proprioception), and pain perception. Without this system our brain could not recieve the information about what we touch, and thus could not integrate this information with those of hearing, sight, smell and taste. This subset of the sensory nervous system is so fundamental for our living that without its proper development we humans could not do almost anything. Without the sense of touch and proprioception, we could not maintain balance, nor standing on two legs, nor on all fours. The skin is undoubtedly the more extended organ in our body, and touch is one of the main means through which our brain exchange information with the environment. The sense of touch is mainly divided in two modalities, the first is the itch, tickle sensations and crude touch, the second is the discriminative touch. Touch, pressure, flutter, and vibration are all types of discriminative touch. Neurons that exhibit modality specificity represent each of these sensations. When a somatosensory neuron is stimulated naturally (for example, by skin warming) or artificially (for example, by electrical stimulation of the neuron), the sensation experienced is specific to the information normally processed by the neuron (i.e., warm skin). As a result, a somatosensory neuron responding to the warm will not respond to skin cooling or a touch stimulus that does not cause the skin to be heated.

Internal forces generated by the position or movement of a body part are known as proprioceptive stimuli. The position of a limb is determined by the static forces on the joints, muscles, and tendons that maintain limb position against gravity. This part of the somatosensory system is mostly internal to the body and is associated with the kinestetic haptic feedback. To experience this kind of feedback usually we have to interact using our limbs to generate forces against the environment or haptic devices which constrain our joints. When developing a sensorimotor interface, an important quality of the device is that does not interfere with the body by limiting its natural DoFs, since their main aim is augmenting human DoFs. The target for feeding back tactile informations from a robot are cutaneous receptors, which are found all over our skin and allow for easy, non-invasive display of tactile feedback.

Skin mechanoreceptors can be classified as encapsulated or unencapsulated. The most common nerve ending in skin is a free nerve ending, which is an unencapsulated dendrite of a sensory neuron. Painful stimuli, heat and cold, and light touch are all perceived by free nerve endings. They are less susceptible to rapid changes in stimulation because they react slowly to a stimulus. Mechanoreceptors sense stimuli also due to physical deformation of their membranes, and the main four are Merkel's disks, Meissner's corpuscles, Ruffini endings, and Pacinian corpuscles. Merkel's disks is the only one of these four mechanoreceptors to be composed of unencapsulated nerve endings, they are slow-adapting and are responsible for feeling light touch. Merkel's disks are part of the Merkel complex, a specialized type of cell capable both of localizing tactile stimuli and percieve the duration of the force applied on the skin. Meissner's corpuscles respond to fine touch, pressure and low-frequency vibration or flutter. Ruffini endings, are responsible for the skin stretch sensing and together with proprioceptors contibute to the sensing of the position of the body. Finally, Pacinian corpuscles are the sensitive receptors of high frequency vibration. Although we have now well identified and know a lot about these subsystems that enable us to perceive the different components of tactile interaction, there are still some uncertainties about how the biopotentials generated by the different mechanoreceptors contribute to the well-known sensations of pressure, touch, vibration [47]. As everything which has to do with human perception, those signals undergo processing and integration of the brain, responsible for all conscious sensations. This work of decoding the human somatosensory system mechanism is already started thanks to neurophysiology studies [48] and psychology studies [49], in the meanwhile in the fields of robotics and haptics, studies mainly focus on the applicative side of stimulating the somatosensory receptors evaluating haptic devices in terms of psychophysical properties performances [50] or performances in collaborative tasks of human-robot systems [51].

1.2 Sensorimotor interfaces and the Somatotopic mapping

As we saw, our external sensing apparatus is the door to access integration with SRLs. We can choose any part of the body that we can move to control a robot, and surely we will find that most of this body part is covered with skin. We will now see how the sensorimotor interface binds our movement to the robot and brings back information by establishing a one loop with the somatosensory system. Finally we will focus on the concept of somatotopic mapping, a key element for understanding how our brain processes the positions of our body sensations by means of the somatosensory system.

1.2.1 The sensorimotor interface: body movements based inputs and haptic feedback

The sensorimotor interface is the wearable device responsible for the connection between the human body of the user and the SRL. Its main parts are an input system capturing the human body movement and a feedback system capable of reproducing tactile sensations which represent information coming from the robot.

In the next paragraphs we will outline the main technologies which are taken into account when developing a sensorimotor interface.

Body movement tracking Extracting inputs from the body is one of the most exciting part of the human robot interaction field. Some technologies such as inertial measurement units (IMUs) and optical tracking systems make robot interaction feel like magic. Engineers developed a lot of methods and devices based on different physical principles to track and take advange of human body movement. In order to use information about the human body measured by a sensor we surely have to process this information, transferring the information from the realm of analog electronics to that of digital electronics. This passage always imply some sort of analog-to-digital conversion, a process that unavoidably reduces the content of information that a sensor can capture. A very smart way to avoid the loss of critical data is to know in advace the frequency content and the resolution of what we have to measure. When dealing with human motion tracking, electrical measurements of mechanical, inertial, acoustic, magnetic, optical, and radio frequency sensors are commonly used [52]. One of the simplest method to track human motion is the mechanical tracking which consists in attaching the object to track to sensors which transduce rotational and linear displacements in an electrical quantity. One simple component that achieves this is the potentiometer, even though is not very used due to its poor mechanical resistance over prolonged physical stress. Such a device is know as electrogoniometer and strain gauge based versions, also known as flexible electrogoniometers are often used in biomechanical studies [55, 56]. Usually combination of mechanics and other physical quantities bring advantages to this approch, for example if we consider optical or magnetic encoders, which can measure absolute rotations, are frictionless and can achieve very high spatial precision. These technologies can be linked to the human body to track certain target joints, often this is achieved by implementing exoskeletons. Exoskeletons allow accurate estimations thanks to their rigid structure and high quality sensors [53, 54], but they are expensive and heavy.

Another staple in motion tracking are optical tracking systems. This system use cameras to capture the position of objects in space. Usually these devices need some type of marker to be placed over the target object to track, and by positioning the marker in a proper way, they can reconstruct the whole body pose in three dimensions [57, 58]. These devices are so accurate and fast that usually are exploited to validate trajectories of robot end effectors or other tracking technologies. In research this kind of technology is often used in biomechanics and outside academia for the Computer-generated imagery (CGI) in films production. One big drawback of using such technologies is the optical occlusion of markers. To avoid this issue often the number of cameras is highly redundant and the cost of those systems becomes very high.

Towards the concept of portability, camera-based tracking algorithms have become a widespread solution due to improvements in computer vision techniques and progressive growth in computers computational capabilities [59], and commercial devices, like the Leap Motion, have gained success for VR applications.

However, camera-based solutions have some limitations: RGB-D cameras might not work properly in an outdoor environment due to the infra-red interference, and, in particular for hand tracking, occlusions of the fingers may cause a poor estimation of the hand pose.

A viable solution consists in using fabric-integrated devices, *e.g.*, datagloves based on piezoresistive, fiberoptic, magnetic, Hall-effect [60], or inertial and magnetic sensors [61]. Based on the latter, our group has recently developed a cost-effective sensing glove based on inertial and magnetic sensors to track the human hand without occlusion problems [62], and a headband for controlling a robot arm [63]. Nowadays, devices such as IMUs or more in general by MARG (Magnetic, Angular Rate, and Gravity) sensors, use Micro Electro-Mechanical Systems technology, which is composed of silicon nano and microstructures which transduce linear and rotational acceleration. By using motion tracking algorithms, we can obtain attitude estimations from those devices. The big problem of IMUs is the gyroscope drift of the measurement, which is caused by thermic effects, electrical noise and construction imperfections [64]. In recent years lots of research focussed on drift errors in pose estimation based on these devices [65, 66, 67]. The MEMS based IMU is one of the most promising technologies of our century. They are compact, fast and accurate and for those reasons their are good candidates to input a system controlling a supernumerary robot.

Surface Electromyography as a movement input There are several research in the subject of biomedical engineering in which engineering approaches are used to medical/biological concerns to solve difficulties of various types. Biosignals monitoring is one of the most important topics in the biomedical sector. In this field, the essential stages are the acquisition execution, the signal processing, and the data interpretation which ultimately lead to body information extraction. When determining whether or not a muscle is active, the surface ElectroMyoGraphic signal (sEMG) is of particular relevance among non-invasive signals. The envelope extraction method is a common way to recognize muscle activity using sEMG. It allows to understand the general trend of the signal amplitude over time, which can be used to estimate muscular activity and thus mechanical behavior. EMG is utilized in kinesiology and biomechanical investigations in relation to the movement of body segments. Furthermore, in ergonomics, this signal can be utilized to assess muscular load during work and monitor fatigue in order to prevent muscular disorders and provide safe work environments. Another frequent application is for patients with amputations or congenitally lacking limbs to control prosthetics, as well as exoskeletons. EMG is a relatively new technology that has the potential to be employed as a control signal for multifunction prostheses or robotic devices in general. Surface EMG technology could be used in the future for human-machine interfaces [68]. Control by physiological impulses is the best option in terms of fluidity and user comfort, as this is the natural way for the nervous system to control muscle fibers. A very popular device which exploited sEMG is the gesture recognition device Myo Armband which has been widely used to realize human robot interaction [69]. Surface EMG is not suitable for accurate joint position tracking, in fact one of its main applications motor intention prediction. In [70] authors reviewed some relevant techniques used for continuous motion prediction from human upper limb, covering all steps of data acquisition and processing, and displaying different method, from regressive models to artificial neural networks. EMG recordings can also be used to control a robotic extra limb. In [39], authors used the EMG in a esay way, by counting the number of consecutive muscular contractions and using these events to drive a finite state machine. The electrodes were embedded in a cap to monitor the frontalis muscle, responsible for the movement of eyebrows. As we saw, the sEMG provides information on muscle activity that has been inspected in numerous application as motor-control studies, muscular fatigue evaluation, and interface/prosthetics control [71]. In these scenarios, sEMG sensing has been often implemented in wearable devices to monitor patients during daily activities and specific tasks [72], or to provide an unobtrusive control of Human-Machine Interfaces [73]. Many wireless sEMG commercial devices have been developed in the recent years, focusing primarily on data logging for sport and monitoring applications, and on gesture recognition for entertainment or remote control purpose.

The haptic feedback Since the sensorimotor interface is both the channel that connects the human body (movement input) to the robot (control unit) and also the robot (its sensor technology) to the human (its somatosensory system), haptic feedback is a fundamental element of this device. Haptic devices can be classified based on whether they are wearable or grounded, if they convey kinestetic or cutaneous feedback and also by counting how many degrees of freedom they can represent. The majority of haptic devices that are currently available on the market cannot be considered wearable[1].

The pursuit of more wearable haptic technologies led to the development of exoskeletons [74], that, however, are often quite heavy and cumbersome, reducing their applicability and effectiveness. Also, grounded devices limit some DoF of the human, and thus are not suitable for sensorimotor augmentation. This is why, in recent years, research in the field of haptics focused on the development of a new generation of wearable haptic interfaces [75]. Haptic thimbles [76, 77], haptic rings [78], and haptic armbands [79], have been successfully applied in different applications, ranging from teleoperation and virtual/augmented reality, to human guidance. Wearable haptic interfaces are designed to provide only cutaneous stimuli usually through vibrations, skin stretch and variation of temperature. These stimuli can be obtained using different type of actuators that can be easily embedded in light and portable devices [75]. Technologies to achieve vibration are usualy based on eccentric rotating mass motors, linear resonant actuators, voice coils or piezoelectric actuators [80]. To achieve the haptic feedback by means of skin stretch various technologies can be exploited, such as servomotors, linear actuators and pneumatic systems. In [81] authors developed a soft haptics actuation system to transmit the sense of touch in a real prosthesis worn by a user. They achieved that by including in the donning part of the prosthesis a small silicon balloon which was inflated to reproduce pressure on the skin. In [82, 83], authors realized fabric based motorized devices to press against the skin or for applying a shear force on it, by ravelling and unravelling a piece of fabric worn around the arm. Instead in [84], a motorized rocker was utilized in contact with the skin to provide the stretch sensation. Another outstanding way to apply pressure on the skin was realized by Mun *et al.* [85], which exploited electro-active polymers embedded in a glove and in a band worn on the forearm to press against the skin.

1.2.2 Somatotopic mapping

While considering the contribution of the haptic feedback in an augmentation device, researchers and developers cannot ignore the neurophysiological principles underlaying the interaction between human senses and the device. As we saw, the somatosensory system plays a central role in acquiring the tactile information that allow us to percieve and interact with the environment and the robots. Aware of this, it is also desirable to understand how our brains store the information gathered from our tactile receptors, mapping them in specific

[1] http://www.forcedimension.com/products, https://www.3dsystems.com/haptics-devices/3d-systems-phantom-premium

areas of the brain. The neural pathways leading to the brain are structured in such a way that information about the physical stimulus's location is preserved. Adjacent neurons in the brain's somatosensory cerebral cortex represent nearby locations on the skin or in the body, forming a map. This map is known as a homunculus [86], since in one of the first studies regarding the discovery of the somatotopic maps, Wilder Penfield and Edwin Boldrey in 1937 published an article in which thay summarize data garthered from electrical stimulation of the brain cortex in 126 operations performed on conscious epileptic patients. The union of the results gave rise to the famous homunculus illustration which depicts the parts of the body ordered by spatial order of appearance in the primary somatosensory cortex and scaled by extension of the brain part representing the body part. Thus was concluded that *somatotopy* is the correlation of a specific spot on the central nervous system to a specific region of the body. Recently, a study confirmed that functional activity inside the sensorimotor cortices is already somatotopically structured in a way comparable to the traditional mature homunculus representation during the human preterm period [87]. Florence *et al.* [88] showed that cutting and then repairing the nerves of the hand in monkeys, so as to destroy the topology of its projections into the brain, can still yield somatotopic maps for the hand, thus demonstrating that despite disordered sensory inputs, there are mechanisms in the developing brain that can create cortical topography. In neurological literature is full of evidence that the somatotopic maps in the brain can self re-arrange after the amputation event [89, 90]. But what happens to the human somatotopy when we augment the human with robotic devices? We still do not know this. While in 2014 Di Pino *et al.* [91] reviewed studies on the brain effects of augmentative technology on humans with an optimistic feel, Dominijanni *et al.* in [92] define the neural resource allocation problem as the possibility of humans adapting their brains to use augmentative robotics trading off the preexisting natural functionalities.

In this thesis, specifically in the chapter 3, I will present my contribution to the study of what happens from an application point of view to the somatotopic function of haptic feedback when we have the input at a fixed body point. In other words, what, if any, is the best point in the body to receive haptic feedback.

1.3 Our contribution

The thesis starts contributing to the field of augmenting technologies, by combining a SRL, a sensorimotor interface and state-of-the-art devices. We present a device that we called the *SoftPro Wearable System*, a device which integrates different wearable technologies with the aim of extending the possible uses of a supernumerary robotic finger. The integration was conceived because the sixth finger is an aid for compensationg missing grasping capabilities of post-stroke patients, and some of those do not recover the possibility of lifting their impaired upper limb against gravity, limiting the range of possible uses of the sixth finger. We enabled also those patients to use this SRL by combining it with a wearable device for gravity compensation of the paretic upper limb. We also included a sensorimotor interface to enhance the ease of use of the device. Although the evaluation gave us positive results, the direct confrontation with the patients led us to highlight the limitations of this device. Two main problems emerged: the poor wearability of such an orthosis for the gravity compensation, and the invasiveness of multiple distributed haptic feedback devices. This was a starting point for the study and the development of the sensorimotor interfaces and for the SRLs, which led us to work on both those two main drawbacks of this device.

From this point, we chose to take two paths to solve the issues: working on the haptic feedback of the sensorimotor interface and minimizing the featuers of a supernumerary limb to ensure that the device is not cumbersome and highly user-friendly.

Examining the role of the haptic feedback of a sensorimotor interface, we firstly analyzed the role of the *positioning* of the haptic feedback which completes the sensorimotor interface. The motivation for this study comes from the necessity of finding a reference position for placing the haptic feedback on the user's body, when using a supernumerary finger. For the experimental setup we used a back brace to position an accelerometer on the top of the shoulder, which allowed us to extract users' shoulder upward movement, trigger of the task. We also used a distributed system of vibrotactile haptic feedback allowed us to elicit the somatosensory system of users with short haptic cues, informing the user of the correct input interpretation of the system. We tested four body positions, shoulder, wrist, hip and ankle by recording and analysing reaction time data to the haptic feedback presented by the system.

Later on, we developed an experiment involving an head worn sensorimotor interface equipped with sEMG sensors and vibrotactile feedback placed on the occipital side of the head. In this experiment, both the input and the feedback apparatus are implementend in the same physical device, an instrumented headband. The experiment was simple: repeating the opening and the closing of a supernumerary robotic finger worn on the wrist. With this experiment we obtained the result that an acknowledgment haptic feedback, a discrete vibration representing the specific event of correctly interpreted input, provided by a sensorimotor interface while using a SRL, enhances performances of the human-robot system, reducing the muscular effort of the subject and shortening the task time.

Lastly, we present the second way we chose, that of working on the supernumerary robotic limb itself. We extended the usage of the sixth finger by removing all of its electronics and keeping only the essential components. We removed both the motor and the electronics, realizing an all mechanical version of the sixth finger. At first, this could seem counterintuitive, but looking at the finel result, we realized a self-contained, very lightweight, cheap and functional additive finger, which has the peculiarity of being so intuitive that the user can easily use it from the very first moment he wears it.

Chapter 2

Integration of state-of-the-art technologies and methodologies for human augmentation

There is no joy equal to that of being able to work for all humanity and doing what you're doing well.

Richard Buckminster Fuller

In this chapter, we present the integration of four different state-of-the-art devices which combined can help stroke patients gain back bimanual grasping ability.

The final prototype integrates haptic devices, an EMG-based input/sensing device, a robotic extra finger and an exoskeleton, working together to assist in the grasping and in the lifting the paretic arm of a stroke subject.

The novel system, called the *SoftPro Wearable System* aims to extend the possible uses of an assistive robot that is capable to give back bimanual grasping capabilities, which is the *Robotic Sixth Finger*. As we saw in Sect. 1.1.2, the Sixth Finger's use is to help stroke patients stabilize and thus bimanpulate objects of everyday living. The main drawback of this wearable robot is its weight, since even if the battery pack can be moved away from the forearm, the motor is attached there, and sometimes adding its weight to the weight of the object to grasp is not sustainable for the affected limb. Following the evolution/development of the Sixth Finger, engineers tried to alleviate this burden by using gravity compensating devices such as the SaeboMAS [93], but while solving the problem of weight they hindered the wearability of the device. The first part of the chapter explains how we obtained both a gravity compensated and wearable assistive robot.

The second part of this chapter deals with the limitation of this system, leads us to observe the possible developments of these technologies.

2.1 Sensorimotor interface integration with a robotic supernumerary finger

Upper limb impairments are ever-present in activities of daily living (ADL). As a consequence, people affected by a loss of arm function have to cope with severe limitations. Many of these limitations result from no longer being able to use both hands to operate objects, such as a fork and knife to eat independently. Often, sufferers of hemiplegia completely lose the functional use of their hands. In addition, the muscles of the arm and shoulder also suffer a significant loss of function, further limiting the possibilities of the patient, who can potentially undergo various rehabilitation techniques. In this section, I will present the development of a wearable system that combines several assistive technologies including sensing, haptics, orthotics, and robotics to compensate for the lack of a functional arm and hand. The resulting device helps to lift the forearm by the use of a passive exoskeleton and improves the grasping function of the impaired hand by utilizing a wearable robotic supernumerary finger. A pilot study involving 3 patients, conducted to test the device's ability to assist in performing ADLs, confirmed its usefulness and serves as a first step in investigating new paradigms of robotic assistance devices.

2.1.1 Motivation

Stroke is a leading source of long-term impairment of the upper limb [94]. For those affected, restoration of hand and arm function is critical for performing activities of daily living (ADL). The use of robotic aid tools is promising for post-stroke paretic upper limb recovery. Devices have been developed to provide intensive, supervised rehabilitation training to patients with mild to severe motor disabilities after neurological injuries [95, 96].

The use of robotic devices in rehabilitation and assistance can enable intensive, involving, and goal-oriented treatment of the weak arm and can also serve as a reliable means of monitoring the patient's progress [97]. A comprehensive review on robot-assisted therapy for hand treatment can be found in [98]. In [99], the authors presented a comprehensive survey on hand exoskeleton innovations for rehabilitation and assistance. These systems aim to restore functional movements during the first few months after a stroke when plastic changes in the central nervous system typically occur [100, 101]. For patients with a chronic condition, hardly any design of assistive devices effectively restores hand functions. In the context of the European SoftPro project[1], a consortium of universities and companies have investigated novel solutions for assistive robotic tools to be used at home by chronic stroke patients [102]. As pointed out by [103], exoskeletons can increase the frequency and accessibility of physical therapy. Traditional exoskeleton approaches include specific user interfaces for training [104, 105], and

[1] https://www.softpro.eu

are barely portable [106, 107, 108]. We proposed to use wearable robots as assistive tools for the recovery of grasping capabilities in patients with paretic arms. A prime example of such an assistive wearable robot is the Robotic Sixth Finger [109], a soft robotic finger worn at the wrist of the paretic limb. The sixth finger achieves a stable grasp by combining its flexion capability with the presence of the adjacent paretic hand, which acts as a palm to stabilize the grip. Patients who retain the ability to move the forearm against gravity after stroke can use the robotic sixth finger, allowing them to perform bimanual tasks. In the SoftPro project, we have studied the combination of the Sixth Finger with a passive and lightweight elbow exoskeleton called Assistive Elbow Orthosis, an instrumented cap as a human-robot interface called the e-Cap, and a force feedback device called the CUFF. By combining these devices, we aim to expand the possibilities of using the Sixth Finger. The design proposed in this work is very lightweight and would enable chronic stroke patients to perform ADLs that require a bimanual grasp.

Another essential aspect to consider when designing assistive tools is the user interface. Neurological deficits might occur after the stroke event making it necessary that the interface must be intuitive while remaining highly reliable and robust. We achieved these requirements by designing an sEMG interface embedded in a cap, called e-Cap, for the Sixth Finger control and a wearable haptic interface, called CUFF, for force feedback from the wearable robotic finger.

In this section, we report how we have integrated these components into a single easy-to-wear device, see Fig. 2.1, as well as the results of a pilot study involving three stroke patients.

The following describes the organization of section 2.1. In section 2.1.2 all devices that are involved in the integration are presented and described in detail. In section 2.1.3, we explain how the integrated system works and how we modified each of the components to match the technical requirements and achieve our goal. In sections 2.1.4 and 2.1.5 we report respectively on the pilot study involving three stroke patients and show the preliminary results. The last two sections, 2.1.6 and 2.1.7, respectively present a discussion and outline conclusions.

2.1.2 System components

We aimed to restore the upper limb function that a person loses due to a stroke by integrating four devices. All technologies in this integration have as their focus that of restoring the functionality of the impaired upper limb. More specifically, the intended use of the system is to restore the ability to perform bimanual ADLs, for instance, opening a bottle. Analyzing this simple task in more detail, we identify two concurrent sub-tasks: stabilizing the bottle and unscrewing the cap. The proposed system assists the paretic arm in grasping and thus stabilizing the bottle, whereas the healthy one is supposed to perform the more dexterous sub-task of

Figure 2.1: The SoftPro Wearable System worn by a healthy subject.

unscrewing the bottle cap.

Four research laboratories developed independently from each other the individual system parts, except for the e-Cap and the Sixth Finger, both developed and previously integrated by the University of Siena. The Sixth Finger aims to restore the ability to grasp objects with a paretic hand, allowing bimanual tasks. To intuitively operate the Sixth Finger, University of Siena developed the e-Cap. Before this sensorimotor interface development, the Sixth Finger was used through a ring with buttons on the healthy hand of the stroke patient. Stroke patients reported a lack of usefulness when occupying the healthy hand to operate the Sixth Finger grasping action. The e-Cap overcomes this inadequacy by substituting the function of a button, translating it into the form factor of a regular cap that embeds all necessary electronics. The device recognizes the movement of the eyebrows through real-time sEMG measurement of the frontalis muscle [39, 109]. In [110], the authors have shown the advantages of using vibrotactile feedback at the occipital area of the head as an acknowledgment of the correct processing of the sEMG signal, improving the usability of the interface. In this way, whenever the user moves the eyebrows to control the Sixth Finger, the e-Cap informs the user with a short vibration burst that the Sixth Finger is about to close or open. The integration presented in this section extended the feedback feature, mapping the motor current of the Sixth Finger, an estimate of the force exerted onto the grasped object, to the CUFF [111]. University of Pisa developed The CUFF interface in collaboration with the Istituto Italiano di Tecnologia. It can render real-time force feedback by squeezing the arm through an actuated fabric belt. In addition to discrete vibration feedback of the e-Cap, the integrated device also features continuous force feedback. Furthermore, the two haptic devices use different modalities; the one for input acknowledgment is vibrotactile, while the one for force feedback is skin stretch based.

In addition to wearing both the Sixth Finger and the CUFF on the paretic arm, the patient will also have to wear the battery pack needed to power these devices. This added weight can further reduce the mobility of the limb in some patients. To address the weight issue, we mounted both the Sixth Finger and the CUFF onto the Assistive Elbow Orthosis [112], a passive gravity-balancing device developed by the University of Twente, featuring a rigid-link arm brace [113] and a 3D-printed spring which provides weight compensation to the forearm of the affected limb. The Assistive Elbow Orthosis is a wearable device and by tuning the spring dimensions, the amount of weight compensation can be adjusted to include the weight of the added devices and facilitate elbow flexion.

All the devices embedded in the wearable robotic system will be briefly presented in the following paragraphs.

Robotic Sixth Finger The Robotic Sixth Finger is a wearable supernumerary robotic finger. It acts as a functional replacement for the thumb and has been shown to compensate for lack of grasping ability in stroke patients [109, 114, 39].

As shown in Fig. 2.2, the Robotic Sixth Finger is a modular, underactuated robotic finger consisting of rigid and flexible links driven by a motor via a tendon.

In the SoftPro Wearable System, the Robotic Sixth Finger has the task of enabling the stroke patient to grasp objects in combination with the impaired limb. The device was modified to improve wearability, and operation time and to accommodate a wider variety of graspable objects. For this purpose, it now has three 2000 mAh Li-Po batteries in addition to a Dynamixel MX-28 motor. Stroke patients have difficulty lifting objects since the weight of the

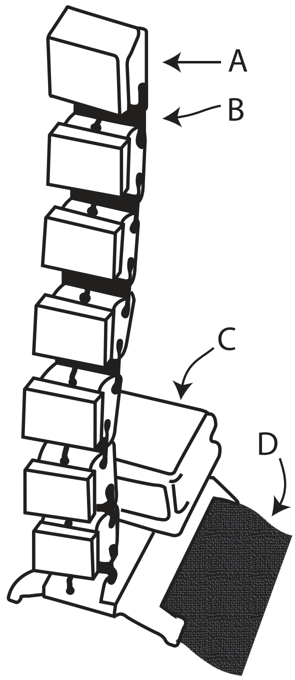

Figure 2.2: The Robotic Sixth Finger. (A) 3D-printed rigid link. (B) 3D-printed flexible link. (C) Dynamixel MX-28 servomotor. (D) Velcro strip mounted on the rigid 3D-printed base.

objects sums up with the weight of Robotic Sixth Finger. To address this issue, the 3D-printed base of the device was modified to be firmly attached to the Assistive Elbow Orthosis, which is described in detail in section 2.1.2. This modification alleviates the problem of lifting the impaired forearm by shifting the weights of the devices to the exoskeleton.

Regarding the software, the code is stored and runs on the Robotic Finger microcontroller, a Teensy 3.2. The Finger's movements are coded in a finite state machine, described in [39]. This part of the code is unchanged. We chose to use the Sixth Finger microcontroller as the master since it was easier to program than the CUFF whereas we chose the CUFF to be the slave device, having a wide set of input commands. We will describe the hardware and software modifications that allow these devices to communicate in section 2.1.3.

e-Cap We redesigned the e-Cap has to integrate it into the novel system. As shown in Fig. 2.3 and 2.4, the e-Cap consists of an sEMG acquisition chain composed of dry electrodes that will be described in section 2.1.2, a commercial instrumentation amplifier, and a Teensy 3.2 microcontroller to sample the analog signal. The e-Cap electronic board has a Bluetooth antenna (RN-42) to stream data to the Sixth Finger. The onboard microcontroller samples the sEMG signal, processes it to extract the command for the Sixth Finger, conveys it to the Sixth Finger and generates an acknowledgment feedback, which is then sent immediately after to a vibromotor placed at the back of the head. Finally, a 3D printed box for the battery pack is positioned on the back of the cap.

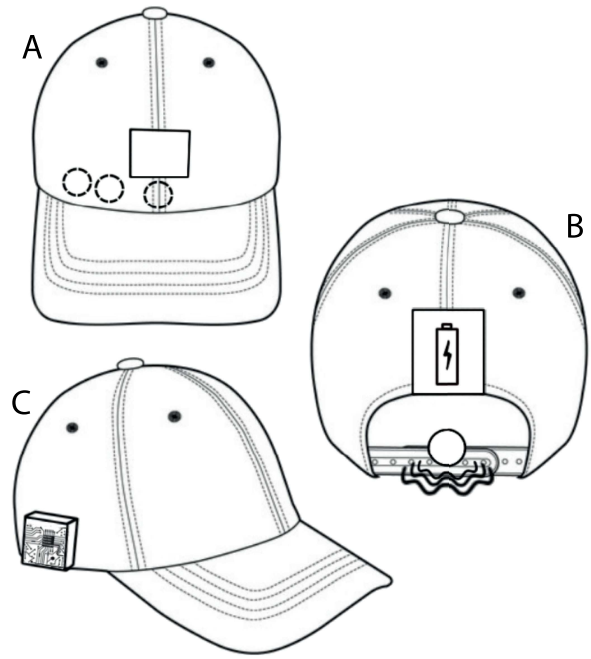

Figure 2.3: The e-Cap. (A) Front view (electrode positions indicated by dashed circles, touch interface by solid rectangle). (B) Rear view (Vibromotor indicated by solid circle and waves, battery by solid rectangle). (C) Side view (showing the electronics box).

Being an sEMG-based interface, it requires calibration before operational use. The trigger for the calibration was upgraded in comparison to previous versions [39]. The physical switch was replaced by a copper pad that acts as a touch button. The calibration procedure is as follows: the touch button toggles the e-Cap from an operational status to a calibration status; once in the calibration status, the user perceives a vibration, and must raise the eyebrows to record the maximum value of the envelope of the sEMG signal; a threshold is automatically set as described in [110].

3D-printed electrodes The most critical part of the e-Cap is the electrode-skin interface. A lot of research exists in the literature that deals with the problem of acquiring sEMG signals reliably [115, 116]. Dry electrodes are preferred due to hygienic considerations and the advantage of being reusable and are better suited for long-term measurements due to their more stable impedance compared to wet electrodes [117]. We propose a solution that combines a reusable sticky plastic tape and non-gelled 3D-printed flexible TPU-based sEMG electrodes developed by the University of Twente [118]. As shown in Fig. 2.4 the sticky tape is transparent and allows easy and accurate electrode positioning. To increase wearability, the electrode interface was electrically and mechanically coupled to the e-Cap, utilizing a magnetic connector.

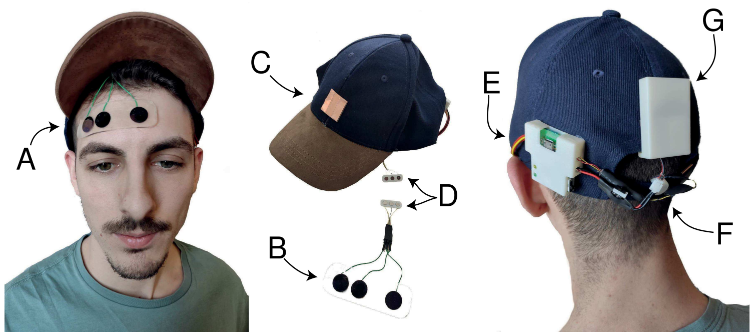

Figure 2.4: The novel e-Cap. (A, B) 3D-printed electrodes by UT combined with the support for easy fixation at the forehead. (C) Touch button for calibration procedure. (D) Magnetic connection for easy connection of the electrodes to the cap. (E) Control board. (G) Power supply. (F) Vibrotactile motor for haptic feedback.

Assistive Elbow Orthosis Tschiersky *et al.* in [112] have developed the Assistive Elbow Orthosis at the University of Twente as a wearable assistive device designed to aid in the lifting of the forearm by providing a gravity-balancing moment to the elbow joint of the wearer. As shown in Fig. 2.5, the device consists of a modified Wilmer elbow orthosis (Ambroise, Enschede, The Netherlands) [113] that acts as the mechanical interface to the wearer, and a stack of nested springs which is mounted laterally onto the orthosis. The spring shape has been optimized to provide an angle-dependent moment, which counteracts the moment caused by gravity acting on the forearm.

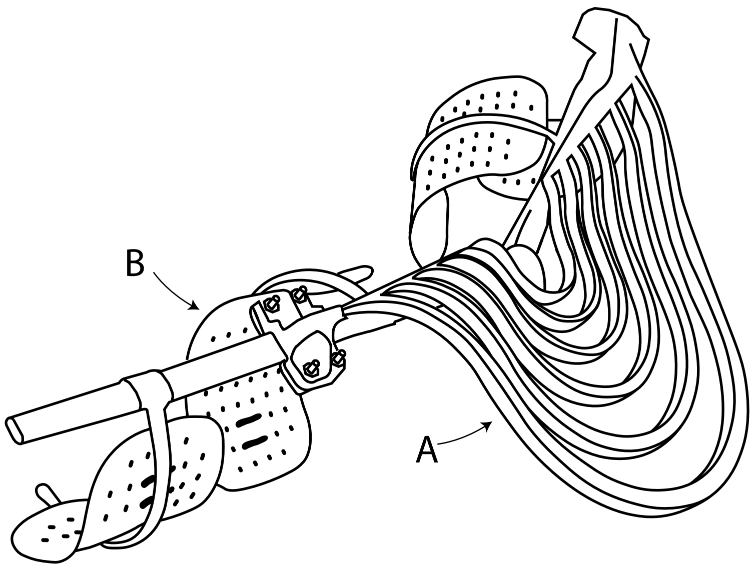

Figure 2.5: The Assistive Elbow Orthosis. (A) Lateral stack of 3D-printed nested gravity-balancing springs. (B) Metal brace with plastic supports (Wilmer elbow orthosis [113]).

CUFF The CUFF is a wearable haptic device that can provide both pressure and skin stretch information to the user's arm. The device is composed of a structural frame, two mechanical actuation units, and the feedback interface. Each actuation unit is powered by a Maxon DCX16S motor and equipped with a two-stage planetary gear-head with a gear ratio of 44:1. The maximum continuous power of each motor is 2.5 W. In Fig. 2.6 a side view of the device is shown. The fabric band is attached to both motors, in such a way that, when actuated in a counter-rotating motion, the length of the tissue band is reduced, ultimately squeezing the arm. In the integrated device, the purpose of the CUFF is to provide force feedback proportional to the load on the Sixth Finger.

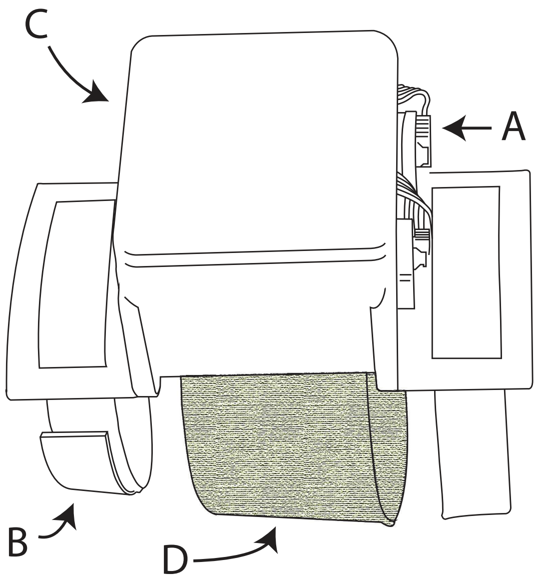

Figure 2.6: The CUFF. (A) Exposed electronics (motor encoders and connection plugs). (B) Velcro straps used to wear the device. (C) 3D-printed box enclosing the electronic board. (D) Tissue band attached to the motors for squeezing the arm.

2.1.3 System integration

In this section, we provide an explanation of how every component works within the system, and how we implemented hardware and software changes to build the device. In Fig. 2.7 we show the setup worn and ready to be used. When the patient raises the eyebrows once, the e-Cap recognizes the gesture by sEMG real-time processing and triggers the Sixth Finger to close in order to perform a grasp.

This trigger signal is sent via a Bluetooth antenna to the Robotic Sixth Finger microcontroller, that starts closing itself until it reaches contact with an object. The contact sensing is implemented by setting a threshold on the Robotic Finger motor current. The input given by the patient is also acknowledged by the system, by providing a short vibration burst at the back of the head.

Once the Sixth Finger is in contact with the object, the patient can decide to increase the strength of the grasp by simply keeping the eyebrows raised. The amount of exerted force is proportional to the time the patient held the eyebrows up, and the value of the Sixth Finger current is sent in real-time to the CUFF device via the RS-485 interface. The value of the current is scaled to match the input range of the CUFF and then sent to the CUFF which renders the force feedback in real-time.

To extend the finger, the patient has to raise the eyebrows twice consecutively. This movement is acknowledged by the e-Cap by a double vibration on the occipital area of the user's head. The integration process required modifications to all components. The Assistive Elbow Orthosis device is essential to help the stroke patients to use the system. As stated before, all devices are indirectly attached to the arm via the exoskeleton. To increase wearing comfort, two out of four plastic ergonomic supports have been removed and replaced by the Sixth Finger and the CUFF using custom 3D-printed mechanical interfaces (see Fig. 2.7). Furthermore, the dimensions of the spring were adjusted to accommodate for the increased weight due to the added devices.

Customized code was implemented to map the sensed load of the Sixth Finger motor to the CUFF device. This task was challenging since the devices were developed independently from each other and the CUFF features a propietary firmware, loaded onto its control unit. Moreover, the CUFF interface code, available at [119], was written to control the device from a personal computer. In order to run on a microcontroller, the control library was modified to use an UART port instead of a USB port.

To use the UART port, additional hardware was necessary to allow for communication via the RS-485 protocol used by the CUFF device. To this end a MAX3485 chip was used to connect the Robotic Sixth Finger and CUFF communication pins. This was a suitable solution, since both devices use a 3.3 V power supply and support high data transmission rates, up to 10 Mbps. A small PCB was used to interpose the additional electronics between the Sixth Finger and the CUFF. With these modifications, the entire control library could be utilized to control the CUFF movements by the Sixth Finger microcontroller.

The CUFF was modified to increase wearability in the integrated system. To provide feedback the device has a fabric strip that goes around the arm. Patients, however, can have difficulties while wearing the CUFF, because the fabric strip can get stuck on the arm. Thus, the fabric strip has been modified by cutting it at the middle and sewing big velcro pads to both ends of the fabric. The CUFF is powered by the same 12 V battery pack as the Sixth Finger.

A button was added to be used as an additional redundant control interface, since previously users reported problems using the e-Cap.

2.1.4 Pilot studies

We have conducted a pilot study on the usability of the SoftPro Wearable System involving three patients (all male, average age 64.4). Two subjects taking part in the experiment were

Figure 2.7: The Robotic Sixth Finger (A) and its power supply and control system (B), developed by US, has been integrated with the Assistive Elbow Orthosis (C), developed by UT, and the CUFF (D) from UP. The CUFF has been modified to be completely wearable. Motion of the finger is controlled via a new e-Cap version (E) developed by US with novel 3D-printed electrodes by UT

in acute phase (they have been affected by stroke no more than three months before the test) and one subject was in chronic state. The device can be used by subjects showing a residual mobility of the arm. For being included in the pilot experiment, patients had to score ≤ 2 when their motor function was tested according to the National Institute of Health Stroke Scale (NIHSS), item 5 "paretic arm". Moreover, the patients showed the following characteristics: normal consciousness (NIHSS, item 1a, 1b, 1c = 0), absence of conjugate eyes deviation (NIHSS, item 2 = 0), absence of complete hemianopia (NIHSS, item 3 ≤ 1), absence of ataxia (NIHSS, item 7 = 0), absence of completely sensory loss (NIHSS, item 8 ≤ 1), absence of aphasia (NIHSS, item 9 = 0), absence of profound extinction and inattention (NIHSS, item 11 ≤ 1). Patients wore the system on the paretic upper limb, the left arm for two subjects and the right one for the other. Due to the design of the device, the same prototype can be worn on either the right or the left arm. Written informed consent was obtained from all participants. The procedures were in accordance with the Declaration of Helsinki.

Patients were asked to wear the system, familiarize with the controller and then use the system to execute a series of bimanual tasks, representing common ADLs: opening a bottle (see Fig. 2.8), removing the cap from a jar, and peeling an apple. After 30 minutes of use, we asked the participants to answer the ten questions of the system usability scale (SUS) [120] . The SUS is used to evaluate subjective assessments of usability. SUS yields a single number that represents a composite measure of the overall usability of the system being studied. It is a Likert scale where each item can be given a mark ranging from 1 “strongly disagree” to 5 “strongly agree”. SUS scores range from 0 to 100, where 0 means “awful” and 100 represents “excellent”. A SUS score above a 68 is considered above average. Details on how to compute the final mark can be found in [120].

Figure 2.8: A stroke patient opening a bottle with the help of the SoftPro Wearable System.

2.1.5 Results

The obtained SUS scores are 70, 95 and 90. This means that the system was deemed useful and very easy to use. We also collected some suggestions for further improvements of the system. One patient suggested to provide the ability of adapting the finger length depending on the task. Two patients perceived the haptic feedback from the CUFF as not very useful during operations and suggested to reduce the bulkiness of the feedback device. They also suggested to limit the force feedback to the grasping phase and remove it once a stable grasp is achieved. One patient preferred the push button for control and stated that a cap could not be comfortably used indoors.

Throughout all experiments, the Assistive Elbow Orthosis was instrumental in allowing the patients to complete their tasks. Even though they criticized the bulkiness of the 3D-printed springs, patients always struggled raising the paretic forearm with the Sixth Finger when not wearing the orthosis.

Finally one patient suggested to add the ability of regulating the closing velocity and applied force through knobs embedded in the control box at the forearm.

2.1.6 Discussion

Patients with reduced mobility of the hand often stop moving the affected limb, loosing the muscular tone recovered during the rehabilitation period. The compensation offered by using the SoftPro system motivates the patient to use her or his muscles by encouraging the patients to use their residual abilities effectively, instead of being solely dependent on the motion of a robotic device. The advantage of the proposed system compared with the constraint-induced movement therapy – a rehabilitative approach characterized by the restrain of the healthy upper limb accompanied by the shaping and repetitive task-oriented training of more affected upper extremity, with the purposes of overcoming the learned nonuse phenomenon of the hemiplegic upper extremity [121] – is that there is no need to immobilize or restrain the healthy limb to encourage the use of the paretic hand. Moreover, being the system self-contained and wearable, patients will not struggle to wear it and operate it autonomously.

2.1.7 Conclusion

The aim of this study is to present a novel device for grasping compensation in motor-impaired subjects. The SoftPro Wearable System is the result of the integration of different technologies, such as a supernumerary robotic finger, an sEMG input device using 3D-printed electrodes, a haptic feedback device and a gravity-balancing arm orthosis.

A pilot study was conducted involving three stroke patients. The positive feedback from the subjects confirmed the need for technological advances and novel concepts in the field of assistive devices. The first impressions of the users, collected by the authors, will serve as guidance for subsequent development of portable assistive devices. The proposed technical solution could also be used by spinal cord injury patients, considering that this category is also subject to upper-limb paresis.

2.2 Issues of state-of-the-art systems for human augmentation

The final outcome of the study showed that, although the device had a very positive effect on patients who reported that they would like to use it in daily life, some crucial points need to be reviewed and evaluated.

One unanswered question concerns the actual need for such a complex system versus a more simple solution, e. g., the Robotic Sixth Finger without any other device connected. We should evaluate whether patients would prefer to use only the Sixth Finger due to the lower encumbrance, or if it is preferable to use the entire system which offers more arm mobility and more fine control, including feedback. All three patients who tested the system reported that, although the functionality of the device helped to move and use the upper limb, its size may reduce its usability. An aspect to be investigated in future studies is the usability of the system in domestic contexts with different usage conditions, e.g. sitting and standing.

The control interface is another crucial aspect to be considered for assistive devices designed for long-term use. The subjects who used our system reported that they were interested in testing different input devices for the integrated system, since they were not always comfortable using the e-Cap. Previous studies reported that some patients prefer to use the e-Cap because they perceive the interface at the frontalis muscle as very intuitive [109], whereas some other patients prefer a button to trigger the opening and closing of the Sixth Finger, together a knob to control its grasp strength. In the future, the 3D-printed sEMG electrodes described in section 2.1.2 could be integrated into the e-Cap and be printed in one go. Furthermore, 3D-printing potentially allows for easy adaptations, improving customizability.

The pilot study involving patients also offered the opportunity to collect important suggestions from potential final users that can guide the future development of the device. For instance, one patient asked for a solution to turn the pages of a book, since this activity of daily living is very important to him. His suggestion was to reduce the length of the Sixth Finger, which led us to the conclusion that adding a length adjustment capability to Sixth Finger could greatly expand its possible uses. Finally, all patients underlined the need for customization and reduction of encumbrance. To this end, the Assistive Elbow Orthosis could be modified to reduce the size of the spring and the CUFF could also be reduced in size and placed at the forearm to make the design more compact.

Chapter 3

Somatotopic mapping and the use of haptics for SMIs

Automation is good,
so long as you know exactly where to put the machine.

Eliyahu Goldratt

In this chapter, we present two works aiming the enhancement of the user experience but also bringing an overall increase in performances of a human-robot system. We can approach this challenge by working on the sensorimotor interface. As described in Chapter 1, when we use sensorimotor interfaces, understanding the role of the haptic feedback is a key aspect. The haptic feedback can be categorized in two macro categories, discrete-type and continuous-type haptic feedback. As a logic consequence the discrete-type of haptic feedback helps the user in tasks where we have distinct important events (e.g. the beginning of the robotic task or movement), or the notification/acknowledgment of a recognized contact between the robot and an element of the environment. The same applies to the continuous-type haptic feedback, which is easily associable to physical variables measured in time using sensors. For example the force sensed by the end-effector, by undergoing an appropriate mapping, can be scaled to match the ranges of human tactile receptive channels.

Another fundamental choice that has to be taken when developing a sensorimotor interface is the positioning of the interface itself on the human body. If the robot is wearable and substituting a missing part of the body, it is a natural choice to place the haptic feedback in the body place where the stump is [81]. But which placement choice is better for the haptic feedback, when the input for the robot is placed on the body of the human, and the robot is worn on another body part, is still a non-answered research question. In the first part of this chapter we deal with the problem of locating the best position to convey the haptic feedback when using a sensorimotor interface, while in the second part of it, we show how the presence of a discrete-type haptic feedback enhances the performances of a sensorimotor interface.

3.1 Somatotopic mapping of haptic feedback from robotic supernumerary limbs

SRLs represent a new class of wearable robots that can augment human manipulation capabilities. SRL can be controlled through input interfaces worn on the user body and can interact with the environment. Such interaction can be measured and feedback to the human wearer through wearable haptic interfaces. However, human somatotopic arrangement on the central nervous system lacks a location for artificially added limbs. Where is the best location for feedback coming from a robot not directly associated with a part of the wearer's body?

This section sheds light on the problem of the best body location for the feedback coming from an SRL as well as on the relation between the position of the input interface and the haptic interface. We have tested four different body locations - shoulder, wrist, hip, and ankle - for vibrotactile feedback coming from the simulated interaction with a robotic extra limb activated using an interface consisting of an accelerometer worn on the user's shoulder. Results from the experiment involving 14 participants demonstrated that the ankle feedback position led to significantly worse performances when having inputs from the shoulder, whereas the other three locations led to comparable results.

3.1.1 Motivation

Wearable robots are usually described as mechatronic systems designed around the human body, with segments and joints matching those of the person it is externally coupled with [122]. This definition perfectly fits all the exoskeletons that have been developed in the last couple of decades. However, in the last few years, a novel generation of wearable robots has been designed not to empower the human joints, but to augment human body functions. These Supernumerary Robotic Limbs [3] are designed to be grounded on the human body, but with their own kinematic structures that do not always resemble that of human limbs. Beside the mechatronic challenges in the design of light and portable SRL, there are other two interesting issues to be addressed: how the human can control the SRL motion and how the SRL can feedback to human important task execution information, *e.g.*, the forces exchanged with the environment. The joint action of interface and feedback is the key toward the usability of this additional limbs [92]. Concerning the input of the interface, several solutions have been proposes ranging from EMG interfaces [123] to measurement of human body motion through, for instance, accelerometers [124].

The haptic feedback is another interesting challenge for SRL. In fact, this wearable robot do not have a direct association with the human body. They are grounded on the body, and so they exchange forces with the human. These forces can be interpreted by the human that can have a proprioceptive information, for instance about the load carried by an extra arm [125].

But there are many other signals, *e.g.* the grasp tightness of a gripper used as end-effector for a SRL that do not have a direct match with the human body. Is it better to display this force where the robot is grounded in the body? Or it is better to display this in another location? Our investigation starts from the hypothesis that when a muscle is activated to start the motion of the supernumerary limb, in our case the shoulder motion that trigger the flexion/extension of the sixth finger, the reaction time to a haptic stimulus is shorter in the body location close to the activated muscle. This hypothesis is sustained by the physiology of the muscles since when a muscle is active, it signals constantly its proprioceptive kinaesthetic feedback both to the cerebellum for movement correction and to the primary somatosensory cortex by means of very fast conduction fibres. Also cutaneous feedback is provided by skin stretch arising from muscle activation [131].

In this Section, we investigated which is the best location for a feedback signal coming from a robotic extra finger controlled by a motion reading interface placed on the shoulder. By lifting up/down the shoulder, it is possible to control flexion/extension of the extra finger. We considered four possible locations for the feedback: the shoulder where the interface is placed, the wrist where the extra finger is body grounded, the hip and the ankle. We measured the reaction time after a vibration burst that was randomly provided in one of the four locations after the activation of the finger motion through the interface. Fourteen participants took part to the experiment. We demonstrated that only the ankle has a statistically significant worsening of the results among the locations.

The rest of this section is organised as it follows. Section 3.1.2 deals with the experimental setup used and the results obtained. Finally, in Section 3.1.3 a discussion on experiment relevance and limitation is proposed, whereas in Section 3.1.4 conclusion and future work are outlined.

3.1.2 Experimental setup

The goal of this study is to understand if there is a part of the body where it is preferable to display haptic feedback from a supernumerary robotic finger and if there is a relation with the position of the input interface. Recent studies started investigating how neural body representation is changed when performing task with an augmented hand with an additional robotic finger [126, 43]. However, the role of haptic feedback is still under-explored and one of the first issues to be faced is the body location of this feedback. As a first step toward the study of somatotopic mapping for SRL, we design a simplified experiment involving an interface for extra finger control, the Robotic Sixth Finger [127] and four haptic interfaces located at shoulder, hip, wrist and ankle. In the following we will describe each component in details. The locations were chosen for the following reasons. The shoulder is the place where the input device is located and it is interesting to evaluate if a co-location of input and feedback device may be beneficial. The wrist was the location where the sixth finger was physically grounded on the subject body. The hip and the ankle were chosen for two main reasons: they represent a medium and a long distance from the input device and they have a bony prominence that may be exploited for better transmission of the vibrations.

The simplified task consisted in moving up the shoulder as if we would like to start finger flexion. After the shoulder gesture is performed, the system generates a vibration feedback, acknowledging the correct shoulder motion, in one of the four locations. As soon as the subject feels the vibration, he/she has to press a button to confirm the feedback perception. The performance metric used in this work is the measure of the voluntary reaction time after the haptic stimulus is sensed by the user. In other words, we evaluated if the perception of the vibration was faster in one of the locations. Moreover, we asked to participants to express their feedback location preference by means of a 7-point Likert scale.

Input device, command extraction algorithm As an initial choiche for the input device we thought of using a frontalis muscle developed in our laboratory. Even though we developed and upgraded our frontalis muscle sEMG interface [63, 127, 110, 128] electrode contact stability is still an issue since in prolonged time of use, the user can sweat of undergo muscle fatigue. Moreover, during experimentations we noticed that some user have difficulties in voluntarily move the frontalis muscle with the sufficient dexterity and in a repeatable fashion. We overcame these problems by developing a new interface for these experiments, a postural back brace equipped with accelerometer and Bluetooth which can remotely give a real-time estimate of the shoulder inclination. The human input to the system is obtained by a wearable device that recognizes the shoulder upward movement. We developed this interface by envisioning an augmentative scenario for humans, in which we try to exploit the kinematic redundancy of the human body. The shoulder is an optimal point in this sense, as it can be moved even when both hands are occupied. The device consists of a commercial back brace posture corrector, upon which it was placed an ADXL362 accelerometer, a Teensy 3.2 microcontroller and a RN42 Bluetooth antenna. The microcontroller sampled the three axes of the accelerometer every 13 ms ($\approx$ 77 Hz sampling frequency) and low-pass filtered each channel with a moving average (cut-off frequency 6.8 Hz). Then the filtered data were sent to the computer through the Bluetooth antenna. We calibrated the accelerometer by following the procedure described in [129], which consisted in fitting a 3D ellipsoid to data acquired from the accelerometer which has been rotated with respect to all three axes.

The pre-processed accelerometer signals were buffered and further processed in LabVIEW 2019. The mean of the signal was subtracted to the channels and then was applied an algorithm to detect the upward shoulder movement.

The algorithm is based on the assumption that every subject is able to generate the same wavelet-like waveform. This wavelet signal can be recognized by finding the peak-valley-peak pattern that characterizes it, shown in Fig. 3.1.

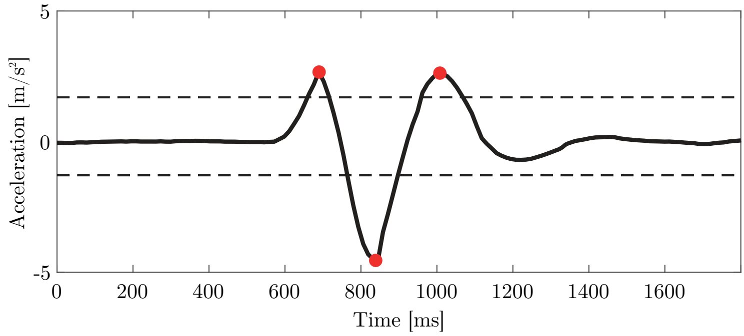

Figure 3.1: The solid line represents the signal generated on the x-axis of the accelerometer when the shoulder upward movement is performed. The dashed lines are the thresholds manually selected to find the peaks and the valley, represented by red dots.

The algorithm takes a moving window of 150 samples (≈ 2 s) and checks if there is this peak-valley-peak pattern in the axes of the accelerometer. The event recognition is performed by calibrating the system as described in Appendix (Algorithm for shoulder event based on real-time accelerometer data).

Feedback system The haptic feedback was generated by using four Precision Microdrives cylindrical vibromotors. Each of these vibration feedback devices was positioned in a different place of the body: on the shoulder where we placed the input of the system, and in three body places with bony prominences, since the vibration feedback could be clearly sensed, on the styloid process of the ulna (wrist), on the anterior iliac crest (hip), and on the malleolus (ankle).

All the feedback location were selected on the same side of the body where the input system was located (right side of the body) so that the readiness of the patient was not influenced by the button press task, in fact the button is held on the other part of the body as shown in Fig. 3.3.

Task protocol After the event recognition – a shoulder upward movement – a random delay is introduced to avoid learning effect. The delay can be between 100 and 300 ms or between 1 and 3 s or be absent. After the delay, the system actuates one of the vibromotors, giving the acknowledge feedback in one of the four selected body spots. The user are asked to press the button as fast as possible once the haptic feedback is received. An explanatory representation of the timing of the experiment is shown in Fig. 3.2. We decided to test the subjects responsiveness both with and without this random delay in between the event recognition and the feedback, since it has been found that variability in the fore-period delay influences the reaction time [130].

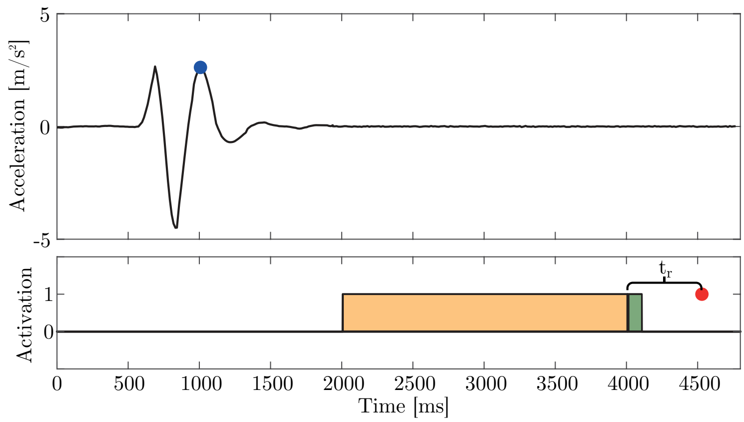

Figure 3.2: Timings of the experiment. The upper plot represents the accelerometric signal (solid line) and the movement detection time instant (blue dot). The lower plot represents timings of the system: the orange part represents the time interval in which the feedback could be randomly given, the green part is the vibromotor action and the red dot represents the time in which the button is pressed by the subject. The reaction time t_r is the time that passes from the vibromotor onset time to the button activation time.

The haptic stimulus consists in the actuation of a vibromotor for 100 ms. At the same time of the vibration triggering, the microcontroller starts to count time with a millisecond resolution. When the user presses the button, an external interrupt is generated and the microcontroller stops to count. The result in milliseconds is then sent back to LabVIEW where is stored and labelled according to the vibromotor that was activated after the shoulder upward movement detection. The use of the external interrupt let us take full advantage of the timing precision of the microcontroller, avoiding to rely on a time measured on Windows operative system. During the experiment the subject worn headphones reproducing pink noise to avoid the acoustic feedback of vibromotors. At the end of the experiment we asked to the subject to fill a 7-point Likert scale questionnaire, reporting the sentence “I felt the location very effective for haptic feedback”. Each of the four feedback location had on the side the 7 options, ranging from “Strongly disagree” to “Strongly agree”.

Data collection and results We collected the reaction time, computed as the time which passes in between the events of motor vibration onset and the voluntary press of the button.

Fourteen subjects aged between 20 and 35 participated voluntarily to the experiment. Each

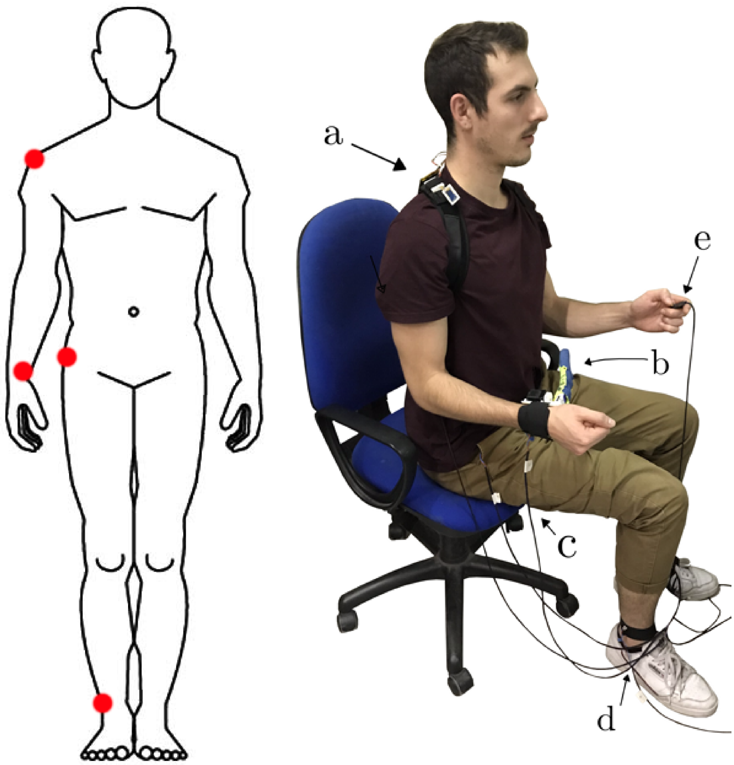

Figure 3.3: In the left part of the figure red dots represent the location selected to place the vibromotors. In the right part of the figure is shown the hardware setup mounted on a subject. It is shown by the arrows: the input device and the shoulder vibromotor (a); the output system, the Robotic Sixth Finger, and the wrist vibromotor (b); the vibromotor placed on the hip (c); the vibromotor placed on the ankle (d); the button to measure the reaction of the subject (e).

subject contributed to gather 20 reaction time measurements for each vibromotor, for a total of 280 reaction times per feedback location. The experiment was repeated with three delay conditions obtaining $14 \cdot 20 \cdot 4 \cdot 3 = 3360$ reaction time measurements.

The reaction times data distributions passed the Shapiro-Wilk normality test. We ran a one-way ANOVA for each delay condition. The one-way ANOVA did reveal statistically significant differences between feedback conditions as shown in Fig. 3.4.

It is worth noting that for all conditions the shoulder positioning performed better than the ankle positioning ($p < 0.02$, $p < 0.0007$, $p < 0.0001$). Another important result is that

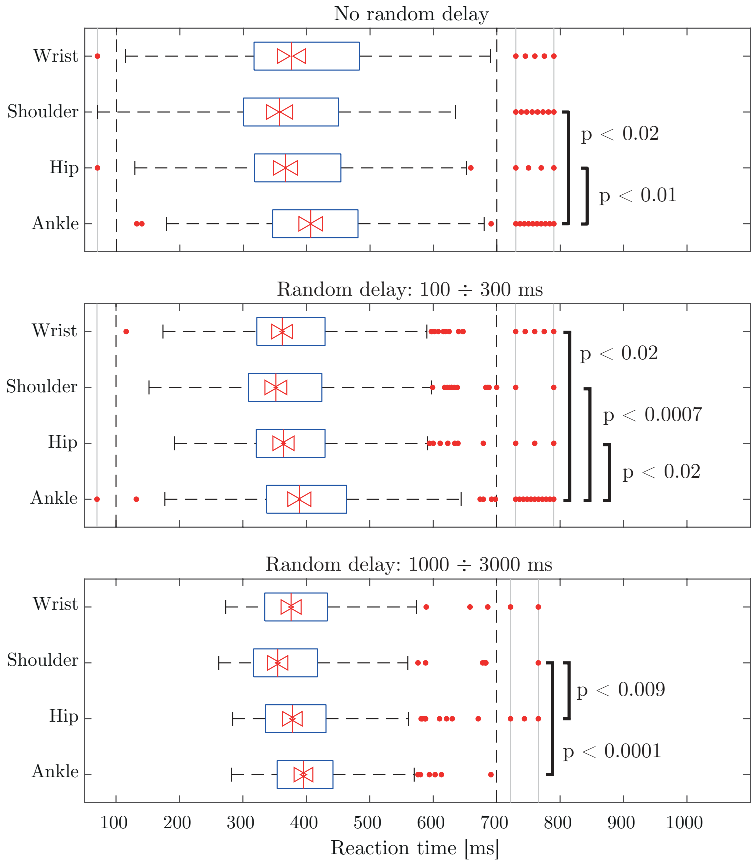

Figure 3.4: Results of the experiment under the three delay conditions: without a feedback foreperiod (top), with a variable delay in the range 100 to 300 ms (middle), or with a variable delay in the range 1000 to 3000 ms (bottom).

the ankle positioning statistically worsened the performance with respect to other conditions.

Also the 7-point Likert scale questionnaire confirmed the result that the ankle positioning is the worst performing, as shown in Fig. 3.5.

3.1.3 Discussion

The aim of this work is to make a first step toward the understanding of the best positioning of haptic feedback coming from a robotic supernumerary limb that misses a direct and clear somatotopic mapping. In our experiment, our initial hypothesis –co-locating input and

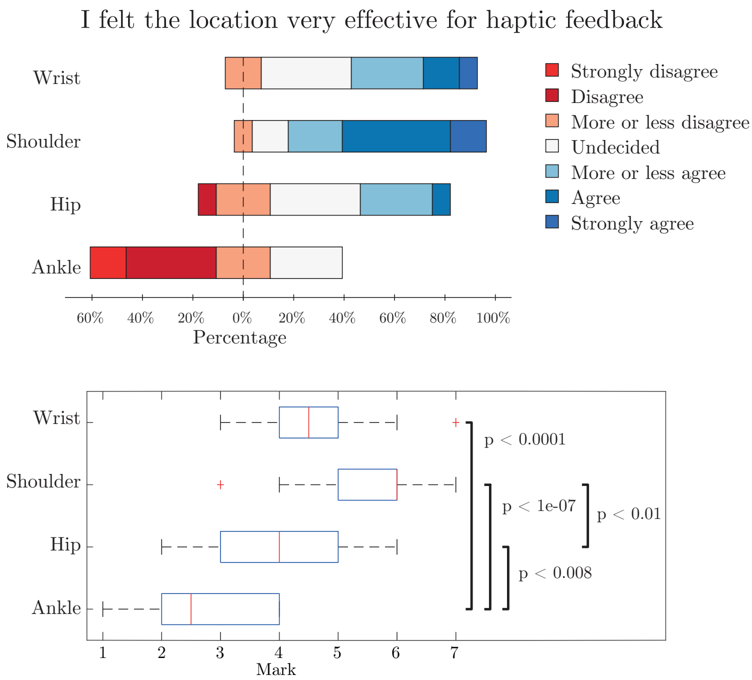

Figure 3.5: Results of the 7-point scale Likert questionnaire completed by subjects after the experiment.

feedback– was not confirmed and, apart the ankle, no statistically significant difference has been measured. One possible explanation to this result is related to the type of feedback. It is possible that a vibration burst interpretable as a discrete event is not enough to elicit the aforementioned neurological signalling that can be considered as a controller for movement correction. A second possible explanation is related to skin physiology. In fact, taking into account that tactile innervation densities are more or less equal in the places where feedback was provided [18], it is possible that the user performs the same in each part of the body, unless the feedback is located very distant in the body, such as the ankle with respect to the shoulder input.

To probe the possibility that the reaction time to the acknowledgement feedback is shorter when the feedback is located in the same body part of the input a different experiment must be adopted. The timing of the subject pressing a button greatly increases the variability of the measurements and more importantly, they are not comparable to bio-potential signals velocity. One possible experiment would be to perform the measurement of the reaction time in a much more invasive way, by using microneedle electrodes to probe muscular onset of muscles responsible for button press [132]. Another, more viable solution would be to evaluate the performances with a different strategy, by leveraging the continuous haptic feedback. This alternative would be non-invasive and of easy implementation, since there are multiple scenarios in which a robot can sense continuous time-varying variables from the environment (*e.g.* feedback proportional to the sensed force of a robot). The subjects that gave an opinion to the experimental setup, claimed that wherever they received the haptic stimulus was not a determining factor. Since the subjects were focused to perform the reaction task of pressing the button, they expected to receive a vibration cue, and, even if the haptic cue was given after a random amount of time, they were somehow ready to react to it.

3.1.4 Conclusions

We concluded that in developing a human-robot interface whose input is placed on the shoulder, physically positioning a haptic feedback channel on the ankle is not a suitable solution. The ankle positioning is inconvenient for a manufacturer since these body parts are far from each other. From an applicative point of view, the ankle is not a practical solution since the user must wear two devices to complete this bidirectional connection with the robot. Data gathered for this experiment did not prove that any of the four locations where we placed the haptic feedback is better performing by examining reaction times to an acknowledgement. However, our results do not exclude the possibility that the haptic feedback could be more useful if placed in one body part rather than another one. The hypothesis we formulated, which would locate the haptic feedback close to the input, was not rejected. Our intent for the future is to keep on exploring this scientific question by providing a continuous haptic feedback to the subject. Such an implementation is non-invasive and maintains the applicative aspect of the research question.

3.2 Improvement of the performances in using a sensorimotor interface by haptic acknowledgment

This work presents a study on the effectiveness of tactile feedback for the acknowledgement of a correct command detection in an EMG-based interface for the frontalis muscle. EMG interfaces are increasingly used in assistive robotics to control robots exploiting the repeatability and robustness of the electromyographic signal. However, in many application a feedback about the correct detection of an input is often missed and the user has to wait for the device motion in order to understand if his/her will has been correctly detected by the system. We demonstrate with a user study involving fifteen subjects, that a simple vibrotactile feedback can reduce the muscular effort and the time needed to execute a sequence of action commanded by an EMG device. As a case study, an EMG interface for the frontalis muscle has been used, however proposed results could be extended to EMG interfaces designed for other muscles, e.g., for prosthesis or exoskeleton control.

3.2.1 Motivation

Assistive robotics is gaining an increasing importance due to novel technological and scientific progresses. Alongside novel and sophisticated robots that can assist physically impaired subjects, several solutions for Human-Machine (HMI) and Human-Computer Interfaces (HCI) have been developed. Such interfaces usually exploit bio-signals that can be voluntary controlled by the user. Muscular contraction measured through EMG activity of skeletal muscles is a classic example of bio-signal used for interfaces. The technology necessary to process such bio-signal is becoming smaller and more powerful. This allows to develop wearable interfaces which interact with the human body.

Among the body muscles, the frontalis muscle has been selected by several research groups to build a HMI or a HCI, for example to realize a computer pointing system [133] [134] [135], to recognize facial gestures [136] [137], or to move a wheelchair [138]. One of the main reason for choosing this muscle is that it is always spared in case of a motor stroke either of the left or of the right hemisphere due to its bilateral cortical representation and it is usually usable by tetraplegic patients.

Our research group developed an EMG interface called eCap for the control of an assistive device called the Robotic Sixth Finger. This assistive device is a supernumerary robotic finger that allows hemiparetic chronic stroke patients to compensate for the missed grasping capability and to perform again a set of bimanual tasks typical of activities of daily living [109, 139]. The robotic extra finger can be worn at the patient paretic forearm and can restrain the motion of a desired object by wrapping around the object and pushing it toward the paretic arm realising a hybrid human-robot grasp. The eCap was a wearable wireless EMG interface where electrodes, acquisition and signal conditioning boards were embedded in a cap. For chronic patients it is generally difficult to generate repeatable EMG patterns in their paretic upper limb due to the weakness in muscle contraction control. For this reason, we coupled the flexion/extension motion of the robotic device with the contraction of the frontalis muscle. The activity of the muscle was recognized filtering the raw EMG signal and setting a threshold as a percentage of the maximum voluntary contraction of the muscle. The user could contract this muscle by moving the eyebrows upwards. The movement of the robotic device was then controlled by using a Finite State Machine (FSM) whose states were spanned through a trigger signal obtained from the eCap interface. The outputs of the FSM were predefined commands based on sequences of input signals. We considered a finite number of states, transition between those states, and commands. States represented predefined motion commands for the robotic device and transition actions were associated with contractions of the frontalis muscle. The FSM was necessary to keep as simple as possible and low cognitively demanding the muscular activity requested to the patients to trigger a certain event. We successfully started using this interface in pilot studies involving chronic stroke patients. However, we quickly recognize the need for a feedback to warn the user about the current FSM status and to inform about the correct detection of a command. Our first solution was a visual feedback consisting in a LEDs board which represented different states, see in [140] [139]. However, this kind of feedback resulted cognitively demanding because, while the user was moving the affected arm to center the object to grasp in the workspace of the robotic finger, he/she had also to check the LEDs to be sure that the eCap received the correct input. Furthermore, such a visual feedback suffers from possible visual obstruction in case the user wears something that could cover the LEDs.

In this work, the visual feedback has been substituted with a tactile feedback, implemented by a vibromotor which gives cues on the head. This solution does not overload the user with additional visual informations and is more intimate, because the user is the only person that can notice the feedback. The aim of this study is to demonstrate that a tactile feedback of the correct detection of a command can improve the usability of an EMG interface. The haptic feedback has been usually used in literature when the EMG interfaces have been used to control prostheses [141] [84] [142] [143] or exoskeletons [144] [145]. The feedback is thus contributing to gather information about the interaction of the device with the environment. To the best of our knowledge, little studies exists on the importance of command acknowledge through tactile feedback. We used the haptic feedback to inform the user about the correct detection of a control input to the system. We proved that the acknowledge of the input is very useful to the user since it reduces the cognitive delay between the intended action, the muscle movement and the actual robot action. We demonstrated that haptic feedback improve the usability of the interface and help in exploiting the functionalities of the Robotic Sixth Finger through a user study involving fifteen subjects.

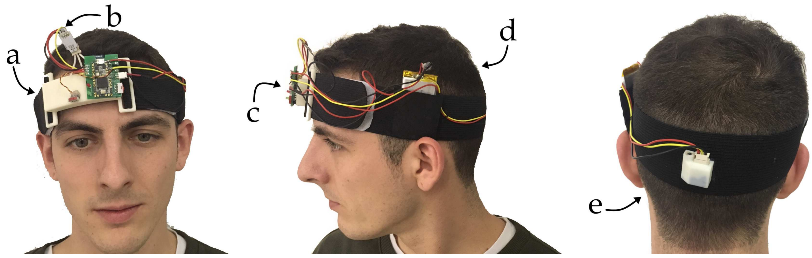

Figure 3.6: The Frontalis muscle interface front, side and back view. Arrows indicate: (a), 3D printed electrodes socket with loops for elastic band; (b), EMG conditioning board; (c), sampling and data processing board with Bluetooth module mounted on a custom PCB; (d), Li-Po battery; (e), vibration motor (ERM) for the haptic feedback, embedded in a 3D printed socket.

The rest of the Section is organised as follows. In Section 3.2.2, a detailed description of the EMG interface is provided. In Section 3.2.3, we describe the experimental setup used to evaluate the tactile feedback, whereas the obtained results are reported in Section 3.2.4. In Section 3.2.5, a discussion on the proposed results is reported, whereas in Section 3.2.6 conclusion and future work are outlined.

3.2.2 The frontalis muscle interface

In this section, we describe in detail the frontalis muscle interface. The device consists of a headband that provides sEMG signal acquisition, on-board data processing, Bluetooth connection, and haptic/vibration feedback, see Fig. 3.6.

The headband is made out of an elastic band with sewed velcro strips to adapt the band length to the user head size. Non-gelled electrodes are used since they result more easy to wear and to be adjusted on the user front to improve the comfort. Non-gelled electrodes result also less expensive on a long term used since they have not to be replaced after few time of usage. The housing for surface EMG electrodes is 3D printed with two lateral belt loop so that is easy to tighten and lock the elastic band.

The electrode housing is a critical aspect of the device in terms of mechanical coupling since it has to correctly place the electrode surfaces over the frontalis muscle getting rid of the different head curvature among different users. The same part has to retain comfort for prolonged use, but at the same time, it has to exert the correct pressure against the skin to guarantee enough stability to the skin-electrode interface since the electrodes used are not gelled. To address these issues, the curved rigid 3D printed part tied to the elastic band has been designed with embedding a compliant housing for the electrodes. A set of springs push a

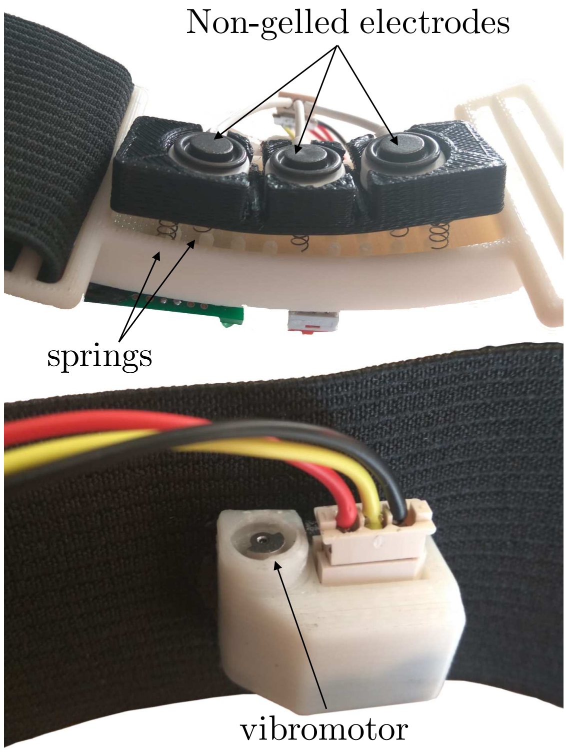

Figure 3.7: Details of 3D printed parts of the frontalis muscle interface. Top part shows EMG electrodes housing with compliant connection through springs to the rigid case made of ABS material. The black part where electrodes are fixed is 3D printed with flexible thermoplastic polyurethane. On the bottom part, the vibromotor case made of ABS that also embeds a current amplifier.

3D printed flexible socket for the electrodes allowing the electrode plane both to tilt, aligning with the head surface, and provide pressure to the electrodes. Details of the obtained electrodes housing are reported in Fig. 3.7. Three reusable non-gelled electrodes coated with Ag/AgCl polymer 10 mm in diameter and 2 mm thick have been used. The EMG signal is amplified with an instrumentation amplifier and passes through an analogic filter both embedded in a compact board. Both electrodes and amplifier (BITalino, Portugal) are designed for EMG acquisition, with a gain of 1009, input range of ± 1.65 mV, filter bandwidth $25 \div 482$ Hz, common mode rejection ratio of 80 dB, input impedance of 10 GΩ.

The haptic feedback is realised through a vibrotactile motor. A rigid 3D printed case containing an Eccentric Rotating Mass (ERM) motor (Precision MicroDrives, United Kingdom) is fixed to the elestic band in the occipital position, see right hand side of Fig. 3.6 and the bottom part of Fig. 3.7. Two main reasons motivate this decision for the vibromotor positioning. Firstly, Jesus Oliveira et al. proved in [146], that the ear mid-line area has to be avoided for placing vibration cues since vibration can result in undesirable disturbing sounds for the user. Secondly, the positioning in the head frontal region cannot be considered since the front is occupied by the acquisition stage and vibrations could affect the EMG recording during operation.

A Lithium-Polymer battery was used to power the system since all circuitry are supplied with 3.3 V. It was secured to the headband by sewing a pocket directly on the elastic headband, see the center of Fig. 3.6. A switch button is used to interrupt battery connection.

The sampling and data processing of the EMG signal is performed with the Teensy 3.2 (PJRC, USA), a cheap yet powerful micro-controller which mounts a 32-bit ARM Cortex-M4 module. It can sample signals up to 16-bit resolution, and it is over-clockable up to 96 MHz, allowing fast/real-time data processing.

The headband is interfaced with the Sixth Finger through Bluetooth 2.1, implemented in the RN-42 module of Roving Networks.

A switch was added to the headband to swap between a calibration mode and an operation mode.

A custom printed circuit board was realized to connect all the electronics hereupon described, avoiding most of the messy connections and providing the hardware with reliability and robustness.

3.2.3 Experimental Setup

In this section, we describe the experimental setup used to test the effect of haptic feedback on the use of the EMG interface. We selected a possible task to be executed with the help of the interface that consists on controlling the flexion/extension of the Robotic Sixth Finger. We already proved in [140, 139, 109] the importance of an EMG interface for controlling grasp assistive tools. Readers are referred to those references also for further details on the Robotic Sixth Finger functioning and realization.

The robotic finger and the interface are connected through Bluetooth antennas. An additional antenna connects the Sixth Finger to a laptop to gather processed data and to visualize the raw EMG signal during the calibration operation, see Fig. 3.8. The real-time signal processing runs at the rate of 1 kHz as well as the signal sampling rate. At each new sample, the raw EMG signal processing consists of the computation of the zero mean signal, subtracting the mean of a sufficiently long buffer of the raw signal to the new incoming sample. It was empirically determined that 350 samples long buffer, was enough to ensure a stable estimate of the raw signal mean value.

Then, the Teager-Kaiser operator [147], a non-linear energy operator is applied to the last 3 samples, enhancing the signal to noise ratio. This is a discrete operator defined in time domain as

$$\Psi_d[x(n)] = x^2(n) - x(n+1)x(n-1) \tag{3.1}$$

and has been proven to enhance performances of data processing algorithm when dealing with EMG onset identification [148, 149]. A buffer of 350 samples of the energy of the signal is stored and the envelope of the signal is obtained by weighting the resulting 350 samples with a Gaussian window of weights $w_G[n]$ and summing them up together. The Gaussian weights were defined as

$$w_G[n] = \exp\Big(\frac{-1}{2}\Big(\frac{\alpha n}{\frac{N-1}{2}}\Big)^2\Big), \tag{3.2}$$

with $-\frac{N-1}{2} \leq n \leq \frac{N-1}{2}$, $\alpha = 2.5$ and $N = 350$.

The obtained envelope is the convolution between the energy of the signal and a window of Gaussian weights, thus can be synthetically described as a modified moving average of the energy of the EMG signal. The envelope value E_{EMG} at the time instant t_0 is obtained as

$$E_{EMG}(t_0) = \sum_{n=0}^{N}\Big(\Psi_d[x(n)]\Big)^2 w_G(n + \tfrac{N-1}{2}). \tag{3.3}$$

This envelope represents the instantaneous power exerted by the muscle and will be taken as an index of performance as reported in Section 3.2.4.

A calibration procedure is run for each subject before starting to execute the task. The calibration is needed to select the correct threshold for acknowledging an intentional contraction of the frontalis muscle. We used the Maximum Voluntary Contraction (MVC) technique for the threshold computation [150, 151]. The system records the highest value of the envelope during a single upright movement of the eyebrows in which the users slowly start increasing the contraction of the forehead muscle to reach their maximum effort. The single-threshold value defined is then defined as the 50% of the MVC, a level that is repeatable and sustainable for the subject without producing noticeable fatigue during the use of the device. This procedure limits the problems related to the high influence of detection condition on EMG signal amplitude that can change between electrode sites, subjects and even day to day measures of the same muscle site. During the calibration phase, the raw signal and the computed envelope are streamed via Bluetooth to the laptop. An operator checks whether the signal is correctly measured and not affected by artefacts due to electrodes' sliding over the skin of the user. After the calibration procedure, when the EMG envelope amplitude overcomes the threshold a counter verifies that the contraction persists in time for at least 50 ms. This second threshold ensures that noise or short involuntary contraction are not assumed as a voluntary muscular activation.

Once a voluntary contraction is recognized by the device, the rising and falling edges of the muscle activation, are sent to the Sixth Finger through the Bluetooth connection. The microcontroller of the Sixth Finger receives the rising and falling edges and counts how much time elapses between them. A single activation is considered when a rising and a falling edge are received within a predefined time window. If two rising edges and falling edges are received inside the time window, then a double activation is identified. This is used to switch among states in the Finite State Machine (FSM) that runs in the the Sixth Finger microntroller. A detailed description of the FSM running on the Robotic Sixth Finger microcontroller can be found in [140]. In this work, we used a simplified version that is briefly summarized in the following. According to the current state of the robotic finger, the state of the FSM can change to

- “finger flexion" when a single contraction is detected, only if the finger is already in the completely open state;
- “finger extension" when two contractions are detected, only if the finger is already in the completely closed state;
- “null” the input is ignored.

In other words, if the eyebrows movement is performed one time in the time window the finger start to flex, if the eyebrows are moved upwards twice, the finger start to extend.

The task to perform is to open and close for two consequently times the robotic finger by moving upward the eyebrows, for a total of four commands to be interpreted by the interface. Two feedback conditions were considered:

- *Haptic feedback* (CF): A vibration burst acknowledges for the contraction detected contraction of the muscle. If there is a single muscle activation, the haptic feedback consists of a single vibration burst. If a second contraction is detected inside the given time window, a second and longer vibration is provided to the user. In other words, a single vibration burst correspond to the detection of the command “close the finger”, whereas a vibration burst followed by a longer vibration correspond to the detection of the command “open the finger”.
- *No haptic feedback considered* (N).

The vibration are generated by controlling the vibrotactile motor at full intensity for 100 ms in the case of a short burst, and for 300 ms in the case of the longer vibration. This is useful for the user since functionally differentiated pulse quality allows the user to clearly understand whether they managed to perform two movements in a row or failed in this task. Moreover,

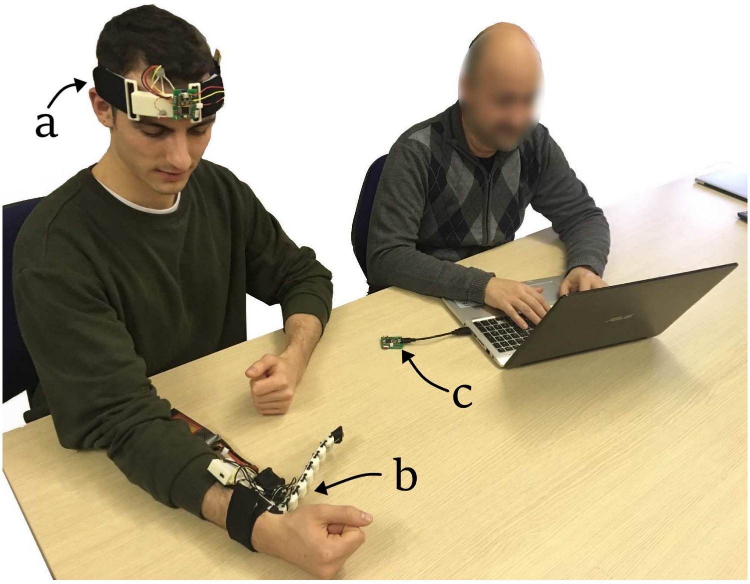

Figure 3.8: Experimental setup. The user wears both the frontalis muscle interface (a) and the Robotic Sixth Finger (b). Data is received through a Bluetooth receiver (c)

this differentiation in pulse length increases the confidence of the user in operating the robotic finger, because with this kind of feedback it is acknowledged that the system recognized the input. The acknowledge is faster than the actual movement of the robotic finger, in this way the user does not have to wait for it.

Three different lengths for the time windows were randomly used during the experiments so to limit possible learning effect. In fact, once the user familiarizes with the control interface, learns how much time available there is to complete the movements. The possible time window lengths were 800, 1000 and 1200 ms.

3.2.4 Results

Fifteen subjects (8 females, average age 28.5 ± 6.8 years) participated to the experiment. Seven of them had previous experience with haptic interfaces. None of the participants reported any deficiencies in their perception abilities and they were all naive as to the purpose of the study.

Each user repeated an open/close command for all combinations two times, with two feedback conditions, and with three possible time window lengths. This led to twelve consecutive open/close per experiment. The experiment is repeated twice gathering a total of twenty-four open/close commands per subject. We define as a task the sequence of two open/close com-

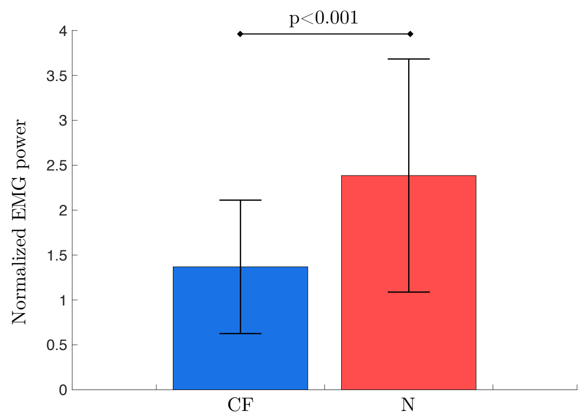

Figure 3.9: Normalized MEI. Mean and standard deviation are plotted for the two considered conditions.

mands. We considered two measures to evaluate the effectiveness of tactile feedback for the EMG interface: we defined a Muscle Effort Index (MEI) to quantify the effort needed to accomplish a task with or without haptic feedback and, we considered the time need to accomplish a task with and without haptic feedback. Moreover, we also measured users' experience through a questionnaire.

Muscle effort index The goal of this evaluation is to determine whether the use of tactile feedback can reduce the muscular effort to execute a sequence of commands. To this aim, we defined the MEI as the measure of the total effort needed by each subject to accomplish the open/close task. The MEI is computed as follows. Every second we acquire 1000 EMG samples @ 1 kHz. For each sample it is possible to compute the envelope contributions as defined in Eq. (3.3). Summing all the thousand sample contributions we obtain an estimation of the EMG power for the considered period of one second. If we normalize according to the highest EMG power measured, we obtain a series EMG power estimation ranging from 0 to 1. This normalization is necessary since different muscle sizes and electrode positioning may result in different maximum values when computing the task total effort. We define the MEI for a task as the sum of all the normalized EMG power. These values represent the total effort to complete a single task. Fig. 3.9 shows mean and standard deviation of the MEI for each feedback condition. A paired T-test analysis revealed statistically significant difference between the tactile feedback and no feedback condition ($p < 0.001$).

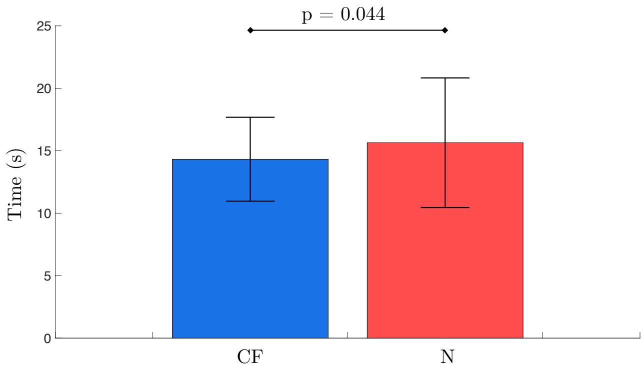

Figure 3.10: Completion time. Means and standard deviations are plotted for the two considered conditions.

Completion time The goal of this evaluation is to verify whether the use tactile feedback can reduce the time necessary to execute a series of commands. Fig. 3.10 reports the mean and standard deviation of time values obtained by taking into account how many seconds elapsed between the first detected muscle contraction and the detection of a opening command, i.e. two muscle contraction in the time window. For each task this operation is repeated twice thus the reported times are the sum of the two open/close commands. A paired T-test analysis revealed statistically significant difference between the tactile feedback and no feedback condition ($p = 0.044$).

Perceived effectiveness At the end of the experiment, subjects were asked to rate, on a slider going from 0 to 10, the effectiveness of each feedback condition in completing the given task. Fig. A paired T-test analysis revealed statistically significant difference between the marks assigned to tactile feedback and to no feedback condition ($p < 0.001$).

3.2.5 Discussion

As we mentioned in the introduction, this work had been inspired by the real need of providing a non visual feedback to the patients using the Robotic Sixth Finger controlled through an EMG-based interface for the frontalis muscle. During the pilot studies performed, most of the patients asked for a possible acknowledge of the correct detection of a command gave by moving the eyebrows upward. The introduction of tactile feedback directly on the head of the user seems to be a promising way to solve this request. Our expectation was that the MEI

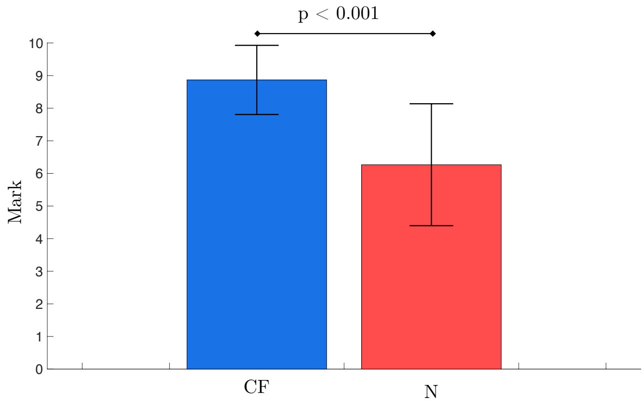

Figure 3.11: Perceived effectiveness. Means and standard deviations of the marks are plotted for the two considered conditions.

value is expected to be higher when no haptic feedback is provided because in this case, the user has to wait for the movement of the robotic finger, that occurs at the end of the decision time window. In the elapsed time, usually the user continues to keep the eyebrows up when is no longer necessary. Warning the user with the described haptic cues before the finger action and immediately after the activation recognition, reduces the muscular effort needed to operate the robotic finger, diminishes the fatigue risk related to the use of the interface for a prolonged time and helps to reduce errors during the robotic finger control. Moreover, the presence of haptic feedback enhances the awareness of the user in using the system so that it becomes faster to execute the task of opening and closing the robotic finger.

We noticed that when no feedback is provided, is not easy to understand if the finger is actually receiving the opening trigger. This often leads users to try to repeat the eyebrow movement three or more times, losing trust in the interface or in their own ability to use it. Indeed, these ambiguities are strongly reduced when tactile feedback is given, since all the user were able to understand whether the interface recognized the double movement or not.

Some participants reported that the experience with the interface was uncomfortable after the whole experiment, finding the cause of that both in the excessive pressure that electrodes exerted on the forehead and the repetition of the same task that involves a muscle that is not typically so often used voluntarily. Another issue to tackle in a future version of the device is the electrodes positioning. The frontalis muscle is not centred in the forehead, but is composed of two symmetrical muscles placed over the eyebrows. To have repeatability of EMG recording the mechanical coupling could be revisited, implementing an active adaptation to head curvature. Moreover, a second EMG channel could be added, monitoring the EMG activity of both frontalis muscles. Different actuators could be used to provide tactile cues, such as linear resonant actuator or piezoelectric elements.

Finally, in this work we only validated the effectiveness of the interface itself, but we did not consider that typical applications of EMG interfaces for assistive robotics may also require a feedback from the interaction of the device with the environment. This open interesting questions on the possible location of the device/devices providing these two distinguished feedbacks (acknowledge and interaction feedback). As reported by the users in the questionnaire, the haptic acknowledgement here studied is an element that enhance the usability of the device. This could be true not only for this interface/system but also in other prosthetic devices/orthosis which rely upon EMG inputs and haptic feedbacks. We point out that the different quality of the acknowledgement would not influence the force feedback, since is perceived as less natural by the user, yet very useful. With this in mind, we are willing to investigate whether that input acknowledgement haptic feedback could be implemented using the same actuators implied in the force feedback without any detrimental effect on it.

3.2.6 Conclusion

In this work, we analysed the contribution of tactile feedback on a EMG interface for the frontalis muscle. In particular, we focused on the evaluation of the importance of the acknowledge of the correct detection of a command through vibrations on the occipital side of the head. We demonstrated that a simple vibrotactile feedback can reduce the time to execute a series of commands. Moreover, also the muscular effort required to execute the series of commands is significantly reduced by the haptic feedback. All the fifteen subjects participating to the experiment indicate the tactile feedback as important add on for the EMG interface. We are currently testing the tactile feedback also with patients using the Robotic Sixth Finger. As a future work, we would like to test the vibrotactile command acknowledge also for different EMG interfaces, including EMG interfaces for prosthesis and exoskeleton.

Chapter 4

Implementation of simple and lightweight supernumerary limbs

Simplicity is the ultimate sophistication.

Clare Boothe Luce

In this chapter we propose another possible solution for building a SRL, but opposite in the approach. The MASF, acronym of Manually Actuated Sixth Finger is the all-mechanical version of the Robotic Sixth Finger. Here we aim to the simplification of a wearable robot reducing to the bare minimum both size, weigth and features, without compromising the functionality of it.

The prototype serves as an aid to gain back the bimanual grasp capability, in a similar way as in the recent scientific history of the Robotic Sixth Finger.

This version of the Sixth Finger solves two main problems. The first problem is that until now the Sixth Finger was a mechatronic device only, and thus the patient had to take care of recharging its batteries and operating it. These action can be as simple as we do everyday with our smartphones, but this level of technology readiness has not been reached yet. The second problem that has been solved is the weight of the device, which in its last update [93] relies on a powerful and not lightweight servomotor.

As an advantage, the removal of the motor made this device also non expensive at all, which increase accessibility of its functions to a wider group of users.

4.1 A Manually Actuated Robotic Supernumerary Finger To Recover Grasping Capabilities

The Robotic Sixth Finger is a wearable aid that has been developed to recover grasping capabilities in patients with a reduced mobility of the hand, usually after a stroke. The finger and the paretic arm act as the two jaws of a gripper to restrain the motion of an object.

In this work, we present a novel version of the Sixth Finger that is completely free of any electronic component and that can be manually actuated. This solution reduces drastically the cost of production while increasing the intuitiveness of the control and use. The prototype includes an a 3D-printed strong and compact ratchet mechanism. We also explore the possibility of using the Sixth Finger to assist not only stroke patients but, more in general, everyone that have lost grasping capability to one hand, such as amputees or spinal cord injury patients. We propose this simple and low cost prototype as an alternative to prosthesis or as an aid for activities of daily living in developing countries and in world zones where classical prosthesis or rehab facilities are rarely accessible. To demonstrate the usability of the device, ten subjects simulating an amputation or a deficit in hand motion tried the supernumerary finger by completing a modified version of the box-and-block test.

4.1.1 Objectives

Stroke is a leading cause of long-term disabilities, which are often associated with persistent impairment of an upper limb [152]. In developed countries where access to rehab facilities can be offered to the majority of population it has been estimated that 33% to 66% of patients do not fully recover the use of the upper limb [153, 154]. These number are reasonably higher in developing countries with a significant higher number of fatal case of stroke. The other two significant reasons for the lost of grasping and manipulation capabilities are represented by spinal cord injuriesand limb amputation [155]. As reported in [156]: "*The proportion of traumatic spinal cord injuries from land transport is decreasing/stable in developed but increasing in developing countries due to trends in transport mode (transition to motorised transport), poor infrastructure and regulatory challenges*." This suggests an increasing trend of possible user of physical aids. For what concerns amputees, despite the great research and the important advancement in the design and diffusion of prosthesis [157], low-cost aids and easily spreadable technologies are still missed. A recent study using 2017 data [155] estimated that a large number of amputation are happening in low income countries, most of them require a prosthetic aid. Higher prevalence was found in South and East Asia.

In [158], authors found, by providing an anonymous online survey, that prosthesis abandonment is significantly less when the device respect some requirements such as appearance,

Figure 4.1: The manually activated Sixth Finger grasping a bottle.

comfort, function, ease of control, reliability, and cost. Although cost is not listed as a primary factor in the study, the population has been recruited from developed countries such as Canada, United States and Netherland. Such a study would be very hard to replicate in third-world countries, and the perspective of the population would shift giving the cost of the prosthesis a much relevant weight.

An example of low-cost and body-powered device to recover bimanual grasp for single sided hand absence was implemented in [159, 160], and a commercial ratchet mechanism was used to regulate the tension of the cables connected to the fingers of a 3D-printed hand. This design is easy to operate, but presents two major issues which are non-negligible: to properly fit the prosthetic device a non-trivial process of scaling must be done since the device fit is not universal, and the functionality of grasping is limited to subjects which retain the wrist flexion and extension. Moreover, there is a large population of subjects which looses the hand functionality without the limb removal, i.e. stroke and spinal cord injury patients. In a meta-analysis performed in 2015 [161], authors report how stroke prevalence in low-income countries increased by 14.3% annually.

With the aim of developing a wearable and portable aid for compensating hand grasping capabilities, our research group has started investigating a device called the Robotic Sixth Finger (RSF) to assist stroke patients [109]. All the proposed versions of the RSF share the same possibility to be worn at the wrist so to achieve hybrid grasps of small and medium sized objects by opposing to the human hand. Different designs have been proposed including a tendon driven soft version that has been widely tested with patients [114]. A crucial aspect for the use of the RSF is represented by the interfaced used by the patient to control the RSF flexion/extension.

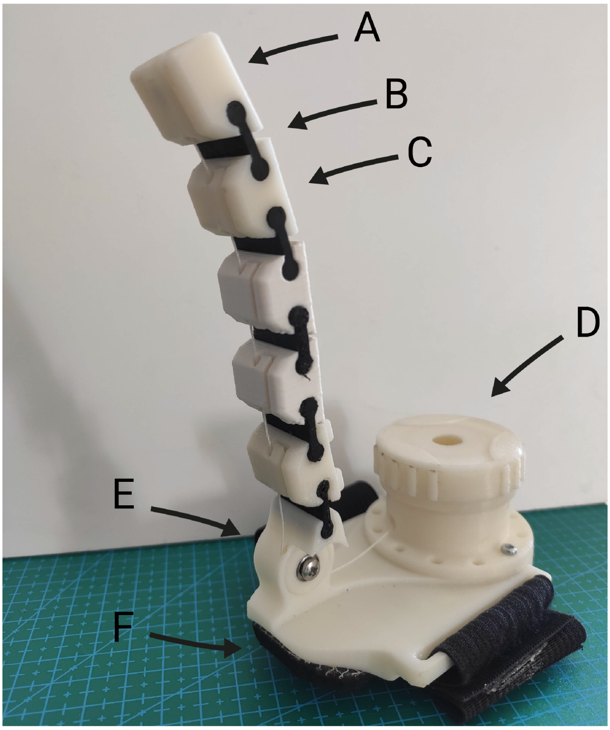

Figure 4.2: The manually activated Sixth Finger with exposed 3D-printed parts. A) End module of the finger. B) Flexible joint. C) 3D-printed module of the finger. D) Ratchet mechanism with pull release for the actuation. E) Ball bearing insert for tendon redirection. F) Base of the MASF.

With several iterations of its implementation, the human-robot interface needed to use the robotic finger has been changed and improved.Initially, the device was controlled via buttons placed on rings worn on the healthy hand [162], later on, shifting paradigm of its interface, we succeed in controlling the robotic extra finger by means of an instrumented cap, called the e-Cap which is capable to recognize through sEMG and in real-time if the user is moving the eyebrows upward [163]. Even though the robotic finger has been proven to be intuitive and helpful [109]. often the patients dislike the idea of wearing a ring with buttons to actuate the robotic finger or the need of prolonged training to use sEMG interfaces. Moreover, the motor used to actuate the finger, the necessary electronic control boards, the batteries as well as the interface have been often perceived as an annoying weight to carry for the users.

In this Section, we present a Manually Actuated version of Sixth Finger (MASF) which is shown in Fig. 4.1 and 4.2. The finger is built using a modular structure that alternates rigid 3D-printed links with soft conjunctions that act as soft joints. A ratchet system can be rotated to wound a tendon that runs through the finger so to flex the whole structure. With this manually actuated version of the Sixth Finger we avoid the issue of having a second device for the control, being this time a self-contained device. Moreover, the absence of batteries, DC motors and electronics, drastically reduce the maintenance and the need for assistance. The whole structure is scalable and completely 3D-printed favouring a quick and easy fabrication as well as adaptation to different arm sizes. The devise can be work thanks to an elastic band. This solution allows to wear the device indifferently on the left or right arm and to easily adjust the device position according to the user requirements. This feature can be exploited to cover different types of amputation or to better exploit residual mobility of the limb in stroke or spinal cord injures patients. With this light, strong and low-cost device, we aim to provide an easy and universal solution to people in low-income and third world countries which loose the hand functionality.

In Section 4.1.2 we describe the implementation of the device, in Section 4.1.3 we characterize the device, in Section 4.1.3 we describe the experimental setup, whereas in Sections 4.1.4 and 4.1.5 we report results and discuss the usefulness of the device.

4.1.2 Methodology

The MASF shares the finger modular structure with the previous versions. The extra finger is underactuated through a tendon. To avoid any electronics, we substituted the motor to drive the tendon with a 3D-printed ratchet mechanism, see Fig. 4.2. We also redesigned the rigid modules of the MASF so to have a fast way to change the number of modules. This was made also since in a previous study of ours, a stroke patient suggested to provide the ability of adapting the finger length depending on the task [128].

Manual actuation mechanism integration The actuation system is realized by a ratchet system, which can pull a tendon by rotating a knob, which can be released by simply pulling the actuation knob. This mechanism is part of an open-source project of the French team Gre-nable, which used this mechanism in their 3D-printed prosthetic hand to fit the device on the stump of amputees [1]. We downloaded the 3D CAD files from Thingiverse [2] and the parts were printed with a Stratasys F170 3D-printer.

The device requires little to no training since both the wearing is easy, achieved by elastic band and velcro, and actuation is simple since the mechanism require the user to only push and rotate the knob to close the underactuated finger. To see how simple it is to wear and use the

[1] https://www.gre-nable.fr/en/electrically-assisted-flexibone-hand/

[2] https://www.thingiverse.com/thing:3662612

MASF, we remind the reader to the following video in which a subject performed successfully these tasks at the first time he used it https://www.youtube.com/watch?v=_JVAUE0I6JM. In Fig. 4.3 it is reported the procedure for finger flexion. Finger extension is achieved using the energy stored by the soft joints during flexion and can be obtained simply moving up the ratchet handle so to unfasten the mechanism.

Fast length-adaptable Module Design As originally designed, the Sixth Finger is meant to be modular, in a way that it was enough to repeat the flexible tendon / rigid phalanx to obtain a longer finger, hence a greater workspace. With respect to the other design this version is easy to disassemble and reassemble, and these operations could be done to include other modules in the supernumerary finger. In the previous versions, once the Sixth Finger was assembled was not easy to thread the tendon inside the structure and around the motor horn.

A physical open channel was embedded in the design of the finger modules, with the result that the tendon assembly is not any more in series, but in parallel. Both the module design and the process of length adaptation is shown in Fig. 4.4.

4.1.3 Device Characterization

In this section, we will show the main mechanical performance of the device, in particular the flexion force and the payload force. In the final part, we will describe the experiment we performed to asses the usability of the device.

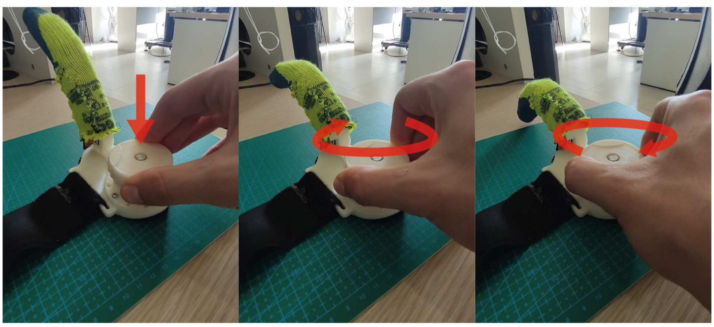

Figure 4.3: In the left image the red arrow indicates the force to lock the ratchet in closing mode, in the center and right indicates the rotation of the knob to cause the finger closure.

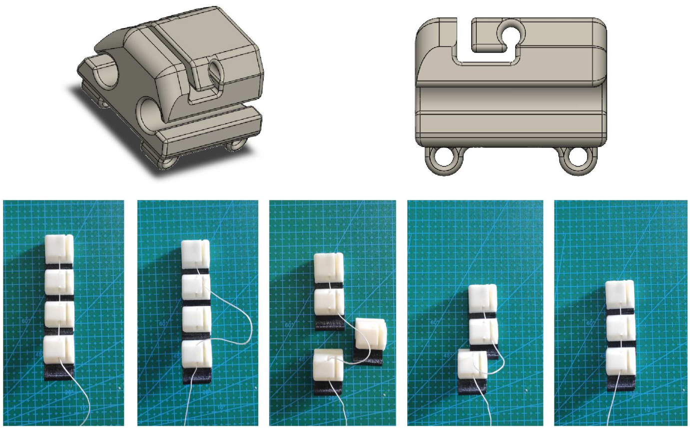

Figure 4.4: The process of removing a module of the finger is shown by following the images from left to right. The process of module addition is the reverse, and it is represented by the same images from right to left.

Maximum flexion force To quantify the traction limit of the MASF, we measured the maximum force that can be exerted by holding the device on the table and closing it as much as possible. The fingertip was tied by a cable to the force measurement device, a Vernier Dual-Range force sensor[3] with a full scale of $\pm 50N$. We repeated the measurement 30 times, saving for each test the maximum recorded force. A video of this test can be found at https://www.youtube.com/shorts/le6WGpzEgdY.

Maximum payload To measure the payload of the Sixth Finger we grasped a cylinder with a diameter of 35 mm and a length of 20 cm which was hang to the same force sensor we used in the flexion force test, by means of a cable. Also this measurement was repeated 30 times. A video of this test can be found at https://www.youtube.com/shorts/g3-bAxTc038.

User Experiment: the Box and Block test The Box and Block test measures unilateral gross manual dexterity [164]. We modified the test in order to meet the simulation of hand loss. We removed one of the two wooden compartments of classic setup leaving the cubes free on the table. The experiment was performed by 9 healthy subjects (2 female, average age 27.3), which had the MASF in their dominant hand (7 right hand). To simulate the inability to use the hand, we immobilized their dominant hand by wrapping it closed in a fist with an elastic band. The subjects had 5 minutes to familiarize with the device and do free trials to understand how to position the MASF with respect to the wrist. The two main positions chosen are the frontal

[3] https://www.vernier.com/product/dual-range-force-sensor/

Table 4.1: Comparative table of the MASF versus the Soft Sixth-Finger

Comparative table		
Feature	MASF	Soft Sixth-Finger
Total Weigth	96 g	140 g
Module Weigth	3.2 g	3.6 g
Max Payload	41 N	2.4 kg
Actuation	Body-powered	Motorized

positioning as shown in Fig. 4.1 or radial. When the frontal positioning was chosen the finger performed an hybrid grasp with the palm side of the wrist, whereas if the radial positioning was chosen the finger would oppose the base of the thumb as can be seen in Fig. 4.5.

4.1.4 Results and Discussion

The experiment performed in Section 4.1.3 gave as a result a mean force of 15.9 N with a standard deviation of 1.6 N. The experiment performed in Section 4.1.3 gave as a result a mean force of 41.1 N with a standard deviation of 4.1 N. Other characteristics of the device are reported in the comparative table 4.1. Moreover, we estimated the cost of the MASF both with our 3D-printer which result in a total cost of $\approx$ 26.5$ and with a conventional 3D-printer, which reduces the cost of the device to $\approx$ 5.2$.

Figure 4.5: A subject grasping a wooden cube during the Box and Block test. This subject chosen the radial positioning during the familiarization phase.

The Box and block experiment described in section 4.1.3 and shown in Fig.4.6, gave promising results. The number of blocks per minute which grew from the first attempt which gave as a result an average of 13.4 blocks per minute with a standard deviation of 4.15, to the third and last attempt in which users were more confident and transferred into the box an average of 16.3 blocks per minute with a standard deviation of 4.06.

Subjects involved in the experiment performed worse than healthy people which are reported to transfer around 75 blocks per minute in average depending on sex and age [164].

4.1.5 Conclusion and Recommendations

The production of the manually activated Sixth Finger is low cost and easily scalable. This would result in a possible wearable aid that can be quickly used by a very large population of subjects with hand impairment due to stroke, amputation, spinal cord injury or any other possible cause. This concept is further substained by the easy to wear and the great possibility of reconfiguration given by the device design. Patient with different amputation can adjust more or less proximally the MASF using the elastic band. The very light weight and the easy of use could also help in promoting a continuous use during the day that is a usually an issue for cosmetic of articulated prosthesis. Benefits for stroke patients have been largely proved by our group with the motor actuated Sixth Finger and are completely hold with the MASF. In addition, the interface is even more intuitive and could promote a more intensive use. Although the experiment gave good results, the sample size is too small (9) to obtain statistically relevant results, and we plan to recruit a larger number of subjects including in the study both amputees and stroke patients. Finally, we believe that by producing parts with

Figure 4.6: A subject performing the Box and Block test.

injection molding techniques the price of the MASF could be lower than the 3D-printing cost estimate of $\approx 5.2\$$. The low price of such a convenient and simple aid for those who lost the function of one hand, would allow extremely poor rural areas to obtain a economically viable solution.

Conclusions and Future Work

If ease of use was the only requirement, everybody would still be riding tricycles

Doug Engelbart

This thesis presents my contribution to the field of *human augmentation*, collecting all the work I have done from 2018 to 2021 toward my Ph.D. degree.

In these years, our work contributed to start addressing the challenge of augmenting humans by means of device integration and prototyping with haptics, for the augmentation of healthy individuals and for compensation of motor abilities in impaired people. We developed wearable devices to compensate missing grasping capabilities of stroke patients, and studied specific open research questions on haptics in relation to the emerging field of sensorimotor interfaces.

Chapter 1 introduces the state-of-the-art and the technical aspects to have in mind when dealing with sensorimotor augmentation. Several concepts such as sensorymotor system somatotopic mapping and SRLs were investigated.

In Chapter 2, we start to look in detail how this thesis contributes to the field of sensorimotor augmentation. We started introducing the SoftPro Wearable System [128], an integration of state-of-the-art technologies composed of an instrumented cap and an instrumented upper-limb orthosis. The device proposed in this Chapter solves the issue of using a supernumerary robotic finger, the sixth finger, for patients that cannot lift their impaired limb against gravity. Previously a grounded solution was utilized to allow all patients to use the robotic aid, the gravity compensator SaeboMAS, creating the problem of non-portability. In this integration the sixth finger works together with a novel wearable orthosis with 3D-printed springs, with an advanced sensorimotor interface exploiting sEMG activity of the frontalis muscle and a device for continuous force feeback. The pilot study we conducted with patients was succesfull, nonetheless brought us to question the need of this large quantity of devices on patients, which confirmed us that something of this device has to be changed.

In Chapter 3, we presented how a key element of the sensorimotor interface, the haptic feedback, plays a fundamental role in the human-robot system, constituting the missing link in the ideal interface. In the Sect. 3.1 we explored the positioning of the haptic feedback with respect to the SRL and to the input interface, creating the basis for the analysis of somatotopic mapping of haptic feedback in a situation where the body position of the haptic feedback is not easy to decide. An experiment consisting of an input interface capable of recognizing the upward movement of the shoulder and a series of vibrotactile feedback devices were used to test reaction time of users to the acknowledgment of the correctly recieved input. We concluded that the ankle is a bad candidate for positioning the vibrotactile feedback if the input is placed on the shoulder, whereas the shoulder positioning gave us good results and was also more appreciated by users. This suggests that co-positioning the input system and the feedback system could be the way to go for future sensorimotor devices.

We also contributed to the integration of haptics in sensorimotor interfaces in Sect. 3.2, in which we present a study on performances of humans using a supernumerary finger, by means of a sensorimotor interface. The interface for this work is an sEMG based headband which recognized the contraction of the frontalis muscle [110]. We tested the performances of users while operating the supernumerary finger, with and without the acknowldgment haptic feedback that informed users of the correct recognition of their input movement. We defined a muscle effort index to evaluate the performances and together with the task completion time and the percieved effectiveness we observed only advanges in having this kind of discrete feedback.

In the last part of the thesis, we took another path to solve the issue of encumbrance and excess of devices that arose in Chapter 2. In Chapter 4 we present a supernumerary finger devoided of motor and electronics [165]. This alternative approach, proposes to reduce the supernumerary finger to an essential minimum, focusing prototyping efforts on the device's ease of use, compactness and high wearability. The manually actuated sixth finger mantains the original functionality of the soft sixth finger of enabling stroke patients to perform bimanual grasps, extending its use to amputees thanks to its high adaptability. The intuitiveness of use achieved by reducing the actuation to the turn of a mechanical knob, allows subjects to immediately understand how to use it. Moreover, being made out of plastic, this device can cost very little (few dollars) and thus could be useful in poor rural parts of the world. We probed the usefulness of the device by simulating the loss of function to one hand of healty subjects, which performed the blocks and box test confirming that, even without training, users could fastly manipulate objects.

With this thesis we contributed to the research on SRLs and of the sompatotopic mapping of the haptic feedback. This field is new and needs attention from both the haptics, robotics and neuroscience communities. In future work, we will include continuous haptic feedback in our experiments to deeply understand what happens to the somatotopic mapping when the user controls a SRL by means of a sensorimotor interface. Also, throughout my journey, the most difficult part was often task selection, since SRLs are a new class of devices and there are not yet many examples in everyday use. Another feature to include to develop advanced experimental tasks is the robot autonomy in interpreting human actions. By including this element in future experiments we could think to much more complex actions that can be performed together with a SRL, and exploit the continuous feedback to understand how and how much it can augment the human experience with robots.

Appendix A

Algorithm for shoulder event based on real-time accelerometer data

To calibrate the system to recognize the input signals of each subject we took the time deltas which occur in the peak-valley-peak signal that characterizes our task for the x and y channels of the accelerometer, and computed a two-dimensional gaussian distributed confidence ellipse with the first 30 movements of the user.

Let the vector Δ be:

$$\begin{aligned} \boldsymbol{\Delta} &= [\boldsymbol{\delta}_{PV} \quad \boldsymbol{\delta}_{VP}] \quad \text{where} \\ \boldsymbol{\delta}_{PV} &= \delta_{PV_1}, \delta_{PV_2}, \cdots, \delta_{PV_n} \\ \boldsymbol{\delta}_{VP} &= \delta_{VP_1}, \delta_{VP_2}, \cdots, \delta_{VP_n} \end{aligned} \tag{A.1}$$

where $\boldsymbol{\delta}_{PV}$ and $\boldsymbol{\delta}_{VP}$ are the vectors of time deltas which occur between the first peak and the first valley and between the first valley and the second peak respectively. Given that these vectors are Gaussian distributed the ellipse will follow a χ^2 distribution, and for the Wilks' theorem we can scale the ellipse by using the factor:

$$s = -2\log(1-p) \tag{A.2}$$

where p is the probability value of cointaining all sample points and ranges 0 to 1, and we chose a value of 0.95, meaning that the ellipse will contain 95% of the data points. Given those assumptions we computed the radii $r_{1,2}$ of the confidence ellipse as the eigenvalues of the scaled covariance matrix:

$$r_{1,2} = \sqrt{\lambda[s \cdot \mathrm{Cov}(\boldsymbol{\Delta})]} \tag{A.3}$$

where the operator λ stands for the eigenvalues and Cov means the covariance matrix. The coordinates for the ellipse center was computed as the mean values:

$$\overline{\boldsymbol{\Delta}} = \left[\frac{1}{n}\sum_{i=1}^{n} \delta_{PV_i} \quad \frac{1}{n}\sum_{i=1}^{n} \delta_{VP_i} \right] \tag{A.4}$$

The rotation of the confidence ellipse was computed as:

$$\theta = \mathrm{atan2}(\vec{v}[s \cdot \mathrm{Cov}(\mathbf{\Delta})]) \tag{A.5}$$

where $\vec{v}$ represents the eigenvector elements. Once we obtained the confidence ellipse parameters, the system was able to identify the shoulder upper movement of users, by checking if the new couple of values δ_{PV_i} and δ_{VP_i} form a point that lies inside the confidence ellipse obtained in the calibration phase.

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