**Advancing the Functionality and Wearability of Robotic Hand Orthoses
Towards Activities of Daily Living in Stroke Patients**


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# Abstract

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Post stroke rehabilitation is effective when a large number of motor repetitions are provided to patients. However, conventional physical therapy or traditional desktop-size robot aided rehabilitation do not provide sufficient number of repetitions due to cost and logistical barriers. Our vision is to realize a wearable and functional hand orthosis that could be used outside of controlled, clinical settings, thus allowing for more training repetitions. Furthermore, if such a device can prove effective for Activities of Daily Living (ADLs) while actively worn, this can incentivize patients to increase its use, further enhancing rehabilitative effects. However, in order to provide such clinical benefits, the device must be completely wearable without obtrusive features, and intuitive to control even for non-experts. In this thesis, we thus focus on wearability, functionality, and intuitive intent detection technology for a novel hand robot, and assess its performance when used both as a rehabilitative device and an assistive tool.

A fully wearable device must deliver meaningful manipulation capability in small and lightweight package. In this context, we investigate the capability of single-actuator devices to assist whole-hand movement patterns through a network of exotendons. Our prototypes combine a single linear actuator (mounted on a forearm splint) with a network of exotendons (routed on the surface of a soft glove). We investigate two possible tendon network configurations: one that produces full finger extension (overcoming flexor spasticity) and one that combines proximal flexion with distal extension at each finger. In experiments with stroke survivors, we measure the force levels needed to overcome various levels of spasticity and to open the hand for grasping using the first of these configurations, and qualitatively demonstrate the ability to execute fingertip grasps using the second. Our results support the feasibility of developing future wearable devices able to assist a range of manipulation tasks.

In order to further improve the wearability of the device, we propose two designs that provide

effective force transmission by increasing moment arms around finger joints. We evaluate the designs with geometric models and experiment using a 3D-printed artificial finger to find force and joint angle characteristics of the suggested structures. We also perform clinical tests with stroke patients to demonstrate the feasibility of the designs. The testing supports the hypothesis that the proposed designs efficiently elicit extension of the digits in patients with spasticity as compared to existing baselines. With the suggested transmission designs, the device can deliver sufficient extension force even when the users have increased muscle tone due to fatigue.

The vision of an orthotic device used for ADLs can only be realized if the patients are able to operate the device themselves. However, the field is generally lacking effective methods by which the user can operate the device: such controls must be effective, intuitive, and robust to the wide range of possible impairment patterns. The variety of encountered upper limb impairment patterns in stroke patients means that a single sensing modality, such as electromyography, might not be sufficient to enable controls for a broad range of users. To address this significant gap, we introduce a multimodal sensing and interaction paradigm for an active hand orthosis. In our proof-of-concept implementation, EMG is complemented by other sensing modalities, such as finger bend and contact pressure sensors. We propose multimodal interaction methods that utilize this sensory data as input, and show they can enable tasks for stroke survivors who exhibit different impairment patterns.

We then assess the performance of the robotic orthosis for two possible roles: as a therapeutic tool that facilitates device mediated hand exercises to recover neuromuscular function, or as an assistive device for use in everyday activities to aid functional use of the hand. 11 chronic stroke ($>2$ years) patients with moderate muscle tone (Modified Ashworth Scale $\leq 2$ in upper extremity) engage in a month-long training protocol using the orthosis. Individuals are evaluated using standardized outcome measures, both with and without orthosis assistance. The results highlight the potential for wearable and user-driven robotic hand orthoses to extend the use and training of the affected upper limb after stroke.

The advances proposed in this thesis have the potential to enable robotic based hand reha-

bilitation during daily activities (as opposed to isolated hand exercises with limited upper limb engagement) and over extended periods of time, even in a patient's home environment. Numerous challenges must still be overcome in order to achieve this vision, related to design (compact devices with easier donning/doffing), control (robust yet intuitive intent inferral), and effectiveness (improved functionality in a wider range of metrics). However, if these challenges can be addressed, wearable robotic devices have the potential to greatly extend the use and training of the affected upper limb after stroke, and help improve the quality of life for a large patient population.

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# Acknowledgements

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# Chapter 1: Introduction

According to the American Heart Association, more than 795,000 people in the United States have a stroke every year, and approximately 610,000 of these are first or new strokes [96]. Stroke is the leading cause of physical disability and paretic upper limb, which is a common unwanted complication after a stroke. In particular, hand motor impairment significantly deteriorates the ability to independently perform Activities of Daily Living (ADLs), affecting the victims' quality of life [50]. Independence, a sense of control, and freedom are some of the key factors positively correlated with life satisfaction and health status for people with motor impairments [36]. Improving confidence in the ability to undertake daily tasks, many of which involves some form of manipulation, could help provide a greater sense of control over life, which is an outcome positively correlated with better health [83].

Fortunately, growing evidence demonstrates high quality, highly repetitive, and task-specific training is beneficial in upper limb recovery after stroke [48, 70, 82]. However, there are challenges that impede many chronic stroke patients from receiving proper treatment. These include logistical and geographical barriers of visiting therapy clinics, insurance and reimbursement limitations, and insufficient availability of therapists with specialized training [52]. Insufficient numbers of planned training movements of the affected limb are also well documented in conventional occupational therapy [47].

The number of repetitions can be substantially increased by the use of available robotic therapies [45]. However, traditional robotic therapies are provided using workstation-type devices requiring direct therapist supervision in a clinical setting. As a result, therapy is provided in brief sessions; typical protocols aim for only three to five hours per week for only six to twelve weeks. Cost and logistical considerations make significant increases in the amount of use per person un-

likely. Moreover, patients training with these devices must devote substantial blocks of time to a therapeutic activity that competes with social and leisure activities. Block training, as provided by standard and robotic therapies, is less effective for motor learning than distributing training in smaller but more frequent aliquots [84]. Finally, current robotic training takes place in a non-functional context, whereas training actual ADLs with "real world" objects is likely to be more effective and of genuine meaning to stroke survivors [44].

To address this need, we propose to develop a **wearable and functional hand orthosis** to assist with movement, force generation, and ultimately the manipulation component of ADLs. A wearable orthosis can have significant impact for both the ability to live independently (while using the orthosis as an assistive device) and regaining use of one's own limb (in which case the orthosis can be seen as a training and rehabilitation device). A wearable device that can be used outside of controlled clinical settings can represent a therapy paradigm shift by:

- providing assistance with the manipulation components of ADLs, helping to increase independence and a sense of control;

- enabling functional training on real-world manipulation tasks and activities rather than exercises in a non-functional context;

- extending training in the use of the affected limb beyond the relatively small number of sessions that can typically be performed in a clinical setting;

- providing distributed training during the course of daily activities, rather than block training in designated therapeutic sessions.

In order to provide these clinical benefits, the technology we propose must meet two important high-level goals. First, we will aim towards functional orthoses that can be used in daily life; such devices must be **wearable** and non-obtrusive to enable regular use. Second, in order for an orthosis to be used in a care receiver's natural environment without technical supervision, it must be equipped with highly **intuitive control** that is learnable within a short period of time.

Although numerous hand robots, which are reviewed in the next chapter, have been recently developed for patients with incomplete motor recovery, devices that simultaneously demonstrate *functional use* by stroke subjects with needed actuation in *wearable packages* and *intuitive intent detection* methods for robust control are significantly less common. Furthermore, studies on mechanical designs and intuitive intent detection methods of a wearable device are often tested on healthy subjects, as opposed to clinical evaluations for functional use or efficacy. For all of the clinical studies included in this dissertation, we prioritize evaluation of functional assistive capabilities of the device on our target population.

Our technical contributions in this work include underactuation through exotendon networks and efficient transmission mechanisms to provide sufficient force via a small number and size of motors, and multimodal intent inferral utilizing impaired, but natural residual motion of the user. In addition to the technical development, we have also assessed rehabilitative and assistive capabilities and limitations of the device. We list the overview of our proposed solutions to unsolved challenges below.

## 1.1   Underactuation with Exotendon Networks

Development of a wearable assistive devices for the hand is a particularly challenging context: the human hand is highly dexterous, often modeled as having as many as 20 individual joints. This high dimensionality of the joint position space gives rise to an enormous set of possible configurations. A key tenet of our approach is that a hand orthosis can provide meaningful assistance with daily manipulation tasks even when using a number of actuators far smaller than the number of joints in the hand. Though the human hand is highly dexterous, previous research suggests that numerous manipulation tasks are dominated by a smaller number of effective degrees of freedom [78, 91, 89]. To implement this principle, we use a network of exotendons, or tendons routed on the surface of the hand, to initiate and assist movement. In Chapter 3, we investigate the feasibility of providing assistance for whole-hand movement patterns using a single-actuator and a network of exotendons. In the course of this investigation, our main contributions include

following:

- We quantitatively assess the combined actuation force needed for assisting a multi-digit hand movement pattern (hand extension) in stroke patients. It is, to the best of our knowledge, the first time that exotendon force needed to overcome spasticity has been measured and reported. We believe this data will be highly significant as we make progress towards compact and wearable assistive devices for the hand.

- We correlate our results with the spasticity level observed in stroke patients, measured using the Modified Ashworth Scale (MAS), commonly employed in patient assessment for rehabilitation. This type of data will help identify patient populations most suited for using wearable assistive devices for the hand. We also characterize the resistance to movement provided by spastic muscles through the assisted range of motion further informing future designs.

- We provide qualitative results indicating the feasibility of single-actuator assistance for a second movement pattern (fingertip pinch).

## 1.2 Effective Transmission Designs

One of the most critical challenges faced by wearable hand robots for post stroke rehabilitation is increased muscle tone due to spasticity meaning that the user involuntarily resists against actuation. Furthermore, the tone also often increases when the patient is fatigued. This results in a large space requirement due to the increased need for high level of motor force as well as the safety problems associated with it. To address these needs, we introduce designs of mechanical structures on a hand orthosis to enhance force transmission efficiency in a tendon-driven device in Chapter 4. The main contributions of this chapter are the following:

- We propose two mechanical structure designs: one for higher spasticity at the proximal interphalangeal (PIP) joint than the metacarpophalangeal (MCP) joint, and one for equally severe tone on each joint. To the best of our knowledge, this study is the first to present and

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evaluate transmission mechanisms by which exotendons can overcome hand spasticity for functional tasks with low motor forces and no rigid joints.

- Using mathematical models of the two designs, we measure the moment arms around the PIP and MCP joints. Assuming that the PIP and MCP joints move simultaneously, we vary design parameters to see how they affect the moment arms. Also, we compare the two with a baseline design to demonstrate their validity.

- We evaluate the computational results through experiments, quantitatively assessing the joint angles and force on the actuated tendon with a 3D-printed artificial finger, designed to mimic the finger of a patient after stroke.

- We present quantitative results through clinical tests with stroke patients to demonstrate the theoretical computations and outcome of experiments with the artificial finger.

## 1.3 Multimodal Intent Inferral

The key to intuitive, user-driven control for a wearable orthosis lies in the ability to infer the user's intent from sensor data collected by the device. The robotic orthosis thus becomes a sensory platform in addition to an actuation mechanism. Forearm electromyography (EMG) is an attractive sensor data that can feed useful signals into a control algorithm for intuitive operation. While studies [60, 79, 17, 71, 29] have shown that intuitive control with EMG is indeed possible, it has also highlighted numerous challenges. For example, EMG signals are inherently abnormal in hemiparesis and distorted by spasticity and fatigue [12, 64]. If signal patterns drift or change between training and deployment, the control method has no way of coping without new calibration or training data. Physical interaction with the orthosis also alters the signals. In fact, other unimodal interaction methods face similar challenges: if the nature of the impairment, which varies greatly between individuals, is such that the signal exhibits too much or too little variation, the entire device can become unusable. We were thus motivated to research and develop various forms of sensing on a wearable hand orthosis for intent detection where different sensing modalities can

complement and augment each other. In Chapter 5, we introduce development of an active hand orthosis as a multimodal sensory platform as well as an actuation device allowing characterization of physical interaction with the user in novel ways. We refer to controls that use multiple sensor types for input as multimodal controls. Overall, the main contributions of this study are as follows:

- We introduce multimodal control methods for the orthosis using the multiple sensors (EMG, bend, and pressure) as inputs. To the best of our knowledge, we are the first to propose intuitive multimodal control schemes for a hand orthosis which, leverage natural hand movement signals (as opposed to side channels such as voice).

- We incorporate bend and pressure sensors into an exotendon framework with EMG sensing while keeping the orthosis compact and without impacting grasping tasks.

- We show that the various controls can be used with different impairment patterns commonly found in stroke subjects through offline controller accuracy testing and functional tasks performed by stroke patients.

## 1.4 Rehabilitative and Assistive Performance via Long-term Training

It is important for a wearable robot to be tested with target population since such devices can hardly prove its efficacy without clinical evaluation due to unique impairment patterns of each individual with stroke. Although aforementioned studies have established the basic operation principles of the device in limited case series with stroke survivor, the clinical performance of this device and the importance of training effects over longer-term use have not been investigated. In Chapter 6, we present clinical outcomes from a 12 session training program, comprising 3 sessions per week for 4 weeks to assess our hand orthosis as a rehabilitation device and an assistive device. Each session involved 30 minutes of active training time in which participants practiced a variety of grasp and release tasks with everyday objects to simulate ADLs. 11 subjects with chronic stroke completed the protocol and were evaluated with a battery of clinical assessments pre and post-intervention. Overall, the main contributions of this chapter are as follows:

- We present clinical outcomes of the monthly training protocol both with and without robotic assistance. To the best of our knowledge, it is the first time an active wearable hand robot was evaluated in clinical assessments both with and without assistance, following user-driven functional hand exercises over multiple training sessions for chronic stroke patients interacting with real world objects.

- Clinical outcomes suggest that intensive hand functional exercises using our robot may improve motor function in distal segments on the upper limb.

- Clinical outcomes highlight the potential for the assistive capability of the device for stroke patients with lower functional use of their upper limb.

## 1.5 Towards a Wearable and Functional Hand Orthosis

Overall, this dissertation is devoted to development of a wearable hand robot that is effective in functional assistance for independent ADLs. We have introduced novel methods and conducted a number of studies to assess and improve the wearability and intent detection methods of the device. These advances have allowed us, for the first time, to evaluate both the rehabilitative and assistive performance of the device via a long-term functional training protocol. The results support the general feasibility of using the device in both roles, but also highlight needed areas of improvement. We believe these contributions have laid the foundation towards functional assistance for rehabilitation outside clinical settings, and a shift of the post stroke rehabilitation paradigm from passive stretch or exercise in a non-functional context to active interaction with real-world objects outside of clinical settings.

# Chapter 2: Related Work

In the past, main focus of upper limb rehabilitation devices had been on proximal components, such as elbow and shoulder. This type of devices provides repetitive upper limb exercises through large exoskeletal workstations. The Armeo Power by Hocoma, Inc., perhaps the most advanced off-the-rack upper limb training tool, allows gravity and weight offset support to enable training using games and functional movement in a simulated environment. A study with 35 enrolled ischemic stroke patients shows improvement of motor function after 40 training sessions of an hour long exercise [10]. The MIT-MANUS robotic system is another example of commercially available training tool for upper limb motor recovery [27]. In extensive randomized, controlled clinical trials involving 127 chronic stroke patients, robot-assisted therapy outperforms human-delivered usual care and provides similar benefit with intensive treatment by therapist for motor performance after 36 weeks of training [53]. Even though these clinical evidences support that the robot treatment can be effective in restoration of arm functions, improvement only on elbow and shoulder limits the use of upper limb in ADLs since there is little incentive to the arm if the hand is not functional.

Due to highly complex and versatile hand movement as well as broad spectrum of possible impairment patterns, only more recently robotic device development for hand rehabilitation has been proposed [46]. The Amadeo is one of the very few robotic workstations for hand rehabilitation on the market [97]. A randomized controlled trial elicits possibility of superior outcomes in subacute stage stroke rehabilitation after 40 sessions with at least 300 repetitions per each training [65]. However, workstation devices are often bulky, costly and tethered in clinical setup, requiring excessive medical supervision. Wearable robots, on the other hand, promise to enable use outside the hospital, providing the larger number of repetitions and have established themselves as an impor-

tant area of focus for research in robotic rehabilitation [47]. To facilitate these benefits, we focus on wearable assistive devices for the hand in this work.

## 2.1 Passive Devices

Wearable assistive devices for the hand have been proposed in the literature using various actuation methods, transmission mechanisms, and control inputs. Among all kinds, one of the most widely used systems is a passive mechanism. An exemplary product, Saebo Flex, utilizes passive underactuation to provide spring-assisted extension force to help stroke patients who cannot volitionally open their hand. Because the device is affordable and relatively easy to don and doff thanks to its simple structure, it has become one of the most successful off-the-shelf wearable hand devices. A study reports there were meaningful clinical improvements for the majority of participants for the Action Research Arm Test (ARAT) and Upper Limb Motricity Index after a 12 week rehabilitation program [87]. SCRIPT hand orthosis also adopts passive transmission using extension springs to assist a user in hand opening [4]. The prime objective of the project is to deploy the device in home environment for telerehabilitation. In a feasibility study by Nijenhuis et al., seven recruited chronic stroke patients show positive outcome on Fugl-Meyer (FM) and ARAT after 113 minutes of medial weekly training for six weeks [62].

While such passive devices achieve a compact and lightweight design, they inherently interfere with finger flexion due to the constantly applied spring force, which makes small object grasping more difficult. The other passive device, Hand Spring Operated Movement Enhancer(HandSOME), increases range of motion by adjustable finger extension force profiles throughout the joint angles depending on the user's impairment pattern [9]. For patients who have some level of volitional finger extension capability, the device can be tuned to apply almost no torque at fully flexed posture and highest torque at extended position. In the case of patients with stiff joint at flexed posture, a therapist can adjust the shape of assistance profile to the one that can apply more torque in the flexed position. In a follow-up pilot study, seven chronic subjects completed reach and grasp tasks for 1.5 hours per day, five days per week and for four weeks, and show func-

tional improvement on FM and ARAT [13]. However, given that patients with hemiparesis may exhibit severe muscle weakness in grip strength [6], even lower level of mechanical interference with finger flexion via passive mechanisms can adversely affect hand functionality. The mechanical interference with finger flexion present with mechanical finger extension aids can be avoided using active hand orthoses.

## 2.2   Exoskeletal Devices

Linkage driven systems exhibit efficient power transmission and support bidirectional actuation. Jo et al. present a single degree of freedom (DOF) exoskeleton using linkage structures that follow fingertip trajectories found from experiments with a motion capture system [37]. Pu et al. developed five digits actuated device, Exo-finger, based on hand kinematics using a linkage driven system [74]. ExoKab utilizes mechanical transmission components, such as worm gears and sliders with two micro motors for four fingers and one motor for the thumb to assist independent hand movement [77]. While such robots benefit from efficient power transmission, this type of devices often faces the additional challenge of rotational axes misalignment [85]. Also, this misalignment can result in discomfort and even injury [81].

Several methods have been proposed to address this, such as direct matching of joint centers in HANDEXOS [14] and HEXORR [80], remote center of motion mechanism in EHI [23], and serial links chain connected to distal phalanges in HEXOSYS [35]. While effective for rehabilitation exercises, complex linkages also increase size and thus reduce applicability in constrained, cluttered environments typical of daily living. We note that wearable linkages can also take the form of a supernumerary robotic finger [30], a different way of providing assistance without interfering with the natural kinematics of the hand.

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## 2.3    Pneumatic Devices

Unlike exoskeletons, wearable hand orthoses comprising only soft structures produce more compact systems since soft devices do not require appropriate alignment with the biological joints of the user. Hand rehabilitation devices using soft pneumatic actuators keep the advantages of completely soft and flexible robots for better interaction with the human hand. A recent narrative reviews on soft hand orthoses reported that wearable hand robots actuated through pneumatic chambers have become more popular as they offer compliance, light weight hardware, and natural movement assistance [15]. However, pneumatic actuation shows inherent drawbacks which are difficulty in control and expensive components. Zhao et al. developed a low-cost soft orthotic glove that contains integrated optical strain sensors for control to address the challenges [102]. The optical bend sensors provide real time feedback on how each finger moves, which can be of great benefit. A customized inconsistent bending profile with variable stiffness can also be implemented to enhance the usability of this type of device [100]. For a similar benefit, Polygerinos et al. utilizes fiber reinforcements to induce specific bending trajectories of molded elastomeric chambers on a soft glove [72]. This device shows remarkable force generation capabilities combined with the compliance and comfort inherent to a fully soft structure. Nine Spinal Cord Injury (SCI) patients participated in a following study where they showed notable improvement on Toronto Rehabilitation Institute Hand Function Test (TRI-HFT) and lift force using this device [11]. Nevertheless, pneumatic actuators have to be tethered to external air pressure sources during operation, which can reduce wearability without stable grounding, such as a wheelchair.

## 2.4    Tendon driven Devices

Tendon driven devices have advantages in terms of wearability since actuators can be remotely positioned and structures located on the hand only require a few small anchor points, which make the system more suitable for underactuation. Also, this approach simplifies the construction of underactuated kinematics as tendons can cross multiple joints for intra-finger coupling and bifur-

cate for inter-finger coupling. Compact and lightweight, BiomHED, for example, exploits artificial exotendons to mimic the geometry of hand muscle-tendon units [51]. Experiments established the ability of the device to generate fingertip motion and increase finger workspace in stroke survivors, highly encouraging for the area of active hand orthoses. Biggar and Yao built a tendon driven robotic glove with suction cups on an inner glove as a cable guide using vacuum pressure [5]. Exo-Glove utilizes exotendon driven system on the surface of a glove adopting a differential mechanism to actuate a multiple fingers with a single motor [33]. In an effort to reduce control inaccuracy caused by compliance in fabric-based gloves, Exo-Glove Poly has been developed [41]. This device allows better fit through various adaptable features rather than relying on the compliance of soft material. Also, the polymer based structures are completely washable and water resistant, which makes the device much more hygienic than any other wearable devices. We note that hygiene is an important feature in a practical point of view.

The tendon driven devices described above deploy either multiple motors closely located to the transmission [51], [5] or a distally mounted single strong motor connected to the end effector via a Bowden cable [33], [41]. However, installation of many motors for the assistance of one movement pattern is redundant, and Bowden cables introduce unnecessary friction. Although there has been a study on transmission mechanisms that provide a natural joint extension sequence using a tendon-driven orthoses [43], no prior study has systematically examined effective transmission methods for tendon driven devices to the best of our knowledge. Our approach introduced in Chapter 4 seeks to achieve efficient transmission on a tendon-driven system while maintaining a compact and lightweight design for optimal wearability.

## 2.5 EMG Sensorized Devices

EMG is one of the most dominant user intent detection modalities for robotic hand orthoses as it requires relatively simple algorithms and enables intuitive operation. Most commonly, sensors are attached to the flexor and extensor muscles of the impaired arm and an open-loop control opens and closes the hand when EMG exceeds a threshold [71, 29]. EMG pattern recognition

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algorithms are also becoming more popular as they can classify multiple hand postures (as many as eight hand motions in stroke patients, if used with a select population because of abnormal muscle activation [54]) and enable the use of commodity EMG armbands [60].

However, these algorithms often only work on a subset of stroke population due to abnormal muscle activation [64]. Several strategies have been developed to adapt to these irregular EMG patterns. One strategy is to place the sensors on muscles that retain healthy EMG patterns. For example, stroke subjects can utilize the contralateral upper extremity [55] or facial expressions [31] to trigger EMG-based controls. Both of these methods require learning a control which uses muscles unrelated to the desired task.

An alternative strategy is to develop a multimodal control that uses EMG in addition to a more robust sensing modality. The VAEDA glove uses voice recognition to specify the control mode, and EMG signals to trigger commands [90]. Voice recognition is robust in ideal conditions, but sensitive to noise. Radio frequency identification (RFID) tags on objects can serve as non-biological switches to identify desired hand postures, again using EMG as a trigger to execute these postures [99]. RFID tags predetermine which objects the subject can interact with, which limits their utility in real world environments. The SPAR Glove for SCI patients is integrated with flex sensors on the wrist and an EMG commodity band on the forearm [76]. The flex sensors are used to control the device in the same manner as the tenodesis grasp. This control mechanism has the potential to be highly intuitive for patients as it leverages a movement already used for natural control of the paretic hand; however, no data is yet available from clinical tests with patients from the target population. Fusing mechanomyography (MMG) and EMG for prosthetic controls has been studied [25, 98]. MMG is more robust to noise than EMG, but its use for individuals with neurological impairment is largely unexplored [32].

## 2.6 Non-EMG Sensorized Devices

Some studies have developed controls which rely on types of sensors other than EMG to avoid intrinsic drawbacks of the EMG control. Some of these controls are unimodal; they trigger the de-

vice using a simple analog button [41], a bend sensor on the wrist [33], body-powered motions [56], or force myography [101]. The Soft Extra Muscle Glove uses force sensitive resistors (FSRs) as a control because they provide useful information when subjects interact with objects [63]. Zhao et al. integrated optical strain sensors into a rehabilitation device based on pneumatic actuation in order to provide accurate position feedback for control and motion analysis [102]. However, unimodal controls have not yet been shown to be robust for long-term operation, and often rely on external cues, instead of natural hand motions.

Some devices use sensors not as control inputs, but as tools to analyze hand movement. The SCRIPT passive orthosis is equipped with multimodal sensors to estimate joint rotations and torques. These sensing capabilities enable interactive rehabilitation games for users [4, 1], but use in real-life tasks or ADLs has not been attempted.

## 2.7  Trends in Upper Limb Robotic Rehabilitation

Overall, the focus of upper limb robotic support has been altering from proximal components (using simple and big machines) to distal components (using small and more complex tools) of the upper limb. As noted in an extensive review paper [46], only more recently, hand rehabilitation tools have been introduced and tested for efficacy for stroke patients. There have been a number of robot-assisted rehabilitation tools for hand function, and the number is rapidly growing [26, 95, 58, 7]. Yet, only few of the existing hand rehabilitation devices have been evaluated with end users for functional feasibility, and even fewer assessed for clinical efficacy in their target populations [46]. The field is looking for more evidence to confirm clinical benefits of robot-assisted rehabilitation, and we believe our project introduced in this thesis aligns with this trend and addresses these needs.

# Chapter 3: Exotendon Networks for Whole Hand Movement Patterns

As discussed in Chapter 1, wearable devices have established themselves as an important area of focus for research in robotic rehabilitation, as opposed to traditional robot-assisted therapy with desktop-sized (or larger) machines. However, wearable devices designed for the hand quickly encounter a challenge in the extensive articulation of the hand itself, which is commonly modeled as having between 20 and 24 degrees of freedom. With current actuator technology, a device with a similarly large number of degrees of actuation cannot achieve the packaging requirements needed for wearability while simultaneously delivering the needed levels of actuation force.

We thus begin our project towards a wearable and functional hand robot with an investigation of whether a hand device can provide *meaningful assistance with daily manipulation tasks even when using a number of actuators far smaller than the number of joints in the hand*. Even though the human hand is highly dexterous, previous research suggests that numerous manipulation tasks are dominated by a smaller number of effective degrees of freedom [78, 91, 89]. Previous work has shown that this result translates to artificial hands as well [16], where it is often implemented using
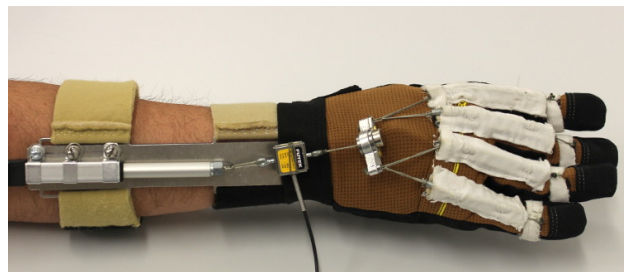


Figure 3.1: Whole-hand movement pattern implemented with single-actuator exotendon network.

the key principles of underactuation and passive compliance. In addition, In et al. [34] have shown how underactuation can also be applied to our area of interest, namely tendon-driven assistive devices for the human hand. Overall, it seems that using a relatively small number of motors is the key in achieving a compact and lightweight wearable device.

To implement these principles, we used a network of exotendons, or tendons routed on the surface of the hand, used to initiate and assist movement. The tendons form a network providing both intra- and inter-finger underactuation; a subset of them are connected to actuators mounted on the forearm (Fig. 3.1). However, before such devices become practical, key questions still need to be addressed. *First, can a device using few and relatively small motors reach the force levels needed for meaningful assistance?* This is a particularly important question given that a common stroke aftereffect is hand spasticity, with permanent involuntary flexion. *Second, can we hope to achieve the dexterity levels needed to enable varied and useful manipulation*, across a wide range of patients exhibiting different impairment patterns?

In this chapter, we investigate the feasibility of a single-actuator device to assist whole-hand movement patterns through a network of exotendons. We propose two possible tendon network configurations to assist full hand extension and fingertip pinch, which many stroke patients cannot actively perform due to the disease. Based on the proposed exotendon designs, we build prototypes. Then, we conduct experiments with stroke survivors to measure the force needed to overcome various levels of spasticity and open the hand for grasping using the first configuration. In addition, we qualitatively demonstrate the ability to execute fingertip grasps using the second configuration.

## 3.1   Movement Patterns and Tendon Networks

Stroke survivors experience a broad range of hand impairments, ranging from barely perceptible slowing of fine finger movements to complete loss of all voluntary movement. It is unlikely, for now, that a single device can effectively address all these impairment types. We have thus chosen to focus on certain patterns of impairments that are particularly common and challenging from a rehabilitation perspective, selected based on the clinical experience of our team. We describe these

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(a)



(b)
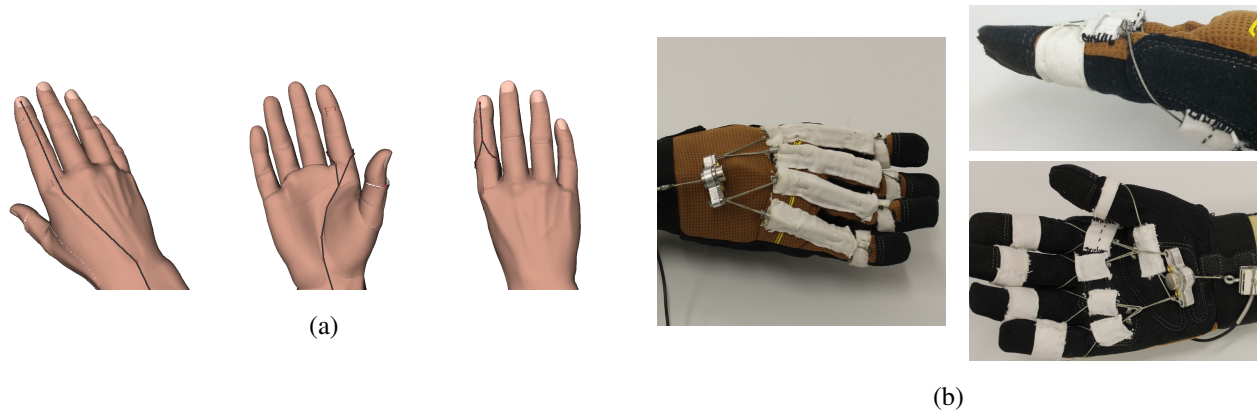
Figure 3.2: (a) Example exotendon routes for achieving different movement patterns illustrated for index finger. (b) Hand exotendon configurations for eliciting desired movement patterns. Left: tendon configuration 1 (hand extension), dorsal view. Right: tendon configuration 2 (MCP flexion / IP extension), lateral and palmar views.

below, noting again that there are a wide range of other motor and functional impairments that affect stroke survivors.

- Pattern A: These individuals are able to form a gross grasp with the hand moving all digits in coordinated movement pattern, but lack sufficient finger extension to actively open the hand after grasping. This pattern is particularly challenging as it commonly includes spasticity, where the hand is in a persistently flexed pose and digit extensors are unable to overcome ongoing involuntary contraction of the flexors. These individuals also typically lack individuated finger movements.

- Pattern B: Another common pattern is the ability to move all digits of the hand, but to have limited individuation, and for the movements to be slow, lacking in dexterity, and of diminished force. Such an individual may be able to oppose the thumb to each of the other digits in sequence, but only slowly and with considerable effort. The ability to manipulate objects is limited.

Here, we report on two exotendon network configurations informed by these patterns. Each of these configurations is designed to be driven by a single-actuator, with multiple joints moving in synergy. For initial study and assessment, we have implemented each configuration separately,

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in a dedicated prototype. However, combined versions able to produce multiple movement patterns with few actuators will be a promising direction for future research. Both configurations are illustrated in Fig. 3.2b.

**Tendon configuration 1**: hand extension. In this configuration, one motor assists extension for all digits. From a clinical perspective, this configuration addresses Pattern A described earlier, where a person lacks sufficient finger extension to overcome spasticity and actively open the hand. Combined finger extension is amenable to direct implementation through a single motor, since a tendon can be routed on the dorsal side of all joints all the way to the phalanges. These routes also allow the tendons to be neutral with regard to abduction/adduction motion of the fingers. Simulations carried out using the human hand model included with the *GraspIt!* simulator for robotic grasping [61] have shown that complete range of motion of all the joints of the index finger requires 57mm of travel of the tendon. This matches the specifications of the off-the-shelf linear actuators we use, noting that functional use of the hand for common tasks is unlikely to require full simultaneous extension of all joints.

The implementation we report on addresses all digits except for the thumb. The trapeziometacarpal joint is significantly more complex than finger carpometacarpal joints; *GraspIt!* simulations based on the common model with two non-perpendicular axes of rotation indicate that most extensor exotendon routes will also have a limited but non-zero effect on thumb abduction. Later in this proposal, we present our solution to this thumb conundrum.

**Tendon configuration 2**: MCP flexion / interphalangeal (IP) extension. This pattern assumes opposite motion at the MCP joints versus the IP joints for each finger. Functionally, this pattern can allow transition between enveloping postures and fingertip opposition postures. For stroke patients exhibiting pattern B described earlier, this could increase the range of grasps that can be executed. From an implementation perspective, it is achieved through a single tendon for each finger, routed on the palmar side of the MCP joint then wrapping around the finger to the dorsal side of the proximal and distal IP joints. The tendon bifurcates to wrap around both sides of the finger in order to obtain a neutral effect on finger adduction. Our implementation of this pattern

addresses all digits, including the thumb, where the role of MCP flexion is instead played by adduction.

## 3.2    Prototype Design and Fabrication

Our overall design is illustrated in Fig. 3.3b. To facilitate donning the device, we split it into two modules: a forearm piece with actuation, and a glove with the tendon network. The two components are connected via mechanical features that automatically detach before potentially dangerous forces are reached. This mechanical coupling includes a permanent magnet connecting the motor and exotendons. We use permanent magnets capable of a pull force of either 34N (D73, KJ Magnetics Inc.) or 41N (D73-N52, KJ Magnetics Inc.), both cylindrical with a diameter of 11 mm and thickness of 4.7 mm. Connector pieces with different tendon lengths allow us to adjust the device to the subject, such that, with all digits fully flexed and the actuator in the fully extended position, we remove all tendon slack up to a few millimeters.

A linear actuator with a 50 mm stroke length, a 5 mm/s maximum speed of travel, and a 50 N peak force (Firgelli, L12-P-50-210-12) is mounted on the forearm piece. A 50mm stroke length suffices for the expected range of motion, and 5mm/s as maximum speed is slow enough to prevent any hazardous circumstances. The peak force of 50N is above that of the breakaway magnetic coupling, so it was never reached in our experiments.

One of the main roles of the forearm piece is to constrain the wrist joint (Fig 3.3a). Splinting the wrist is important in our mechanism as it helps extend the fingers without hyperextending the wrist. Furthermore, the splint is designed to maintain a wrist extension angle of 30°, considered a functional wrist pose [49]. This design also reduces distal migration, or the phenomenon where an entire orthotic device slowly slides towards the distal end of the arm while in use. To reduce pressure, which might cause pain on the hand, soft materials, such as moleskin, are attached underneath the splint.

The tendon networks described in the previous section are implemented on the glove component of the device. In each case, one tendon connected to the actuator bifurcates into a network
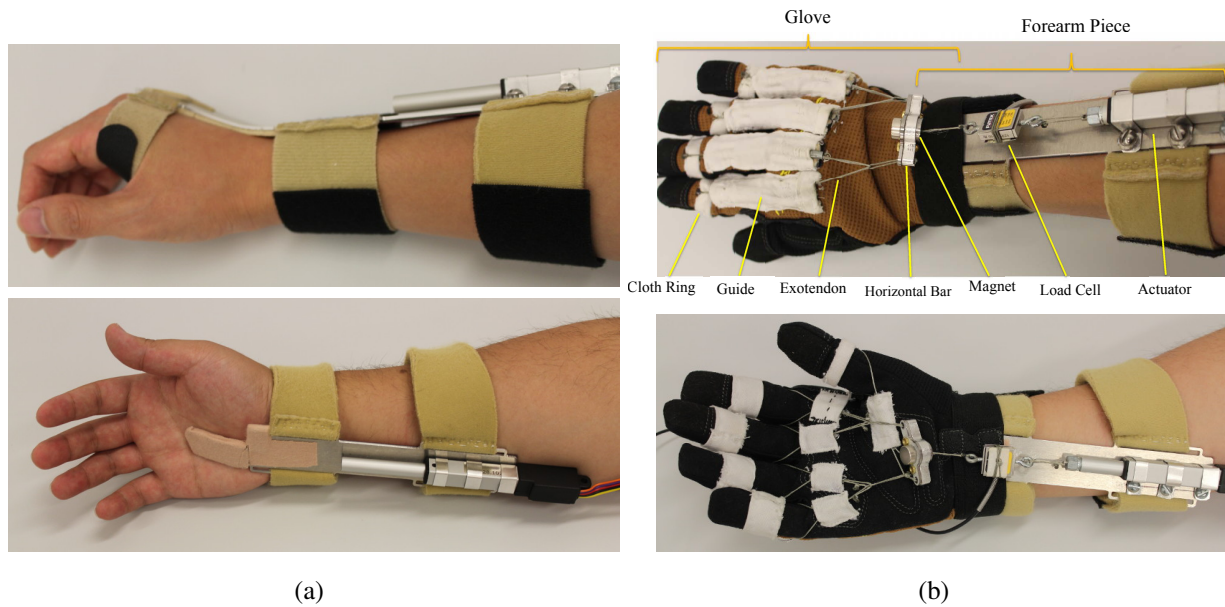
Figure 3.3: (a) Forearm splint components for devices with tendon configuration 1 (top) and tendon configuration 2 (bottom). (b) Prototype hand orthotic devices. Top: tendon configuration 1. Bottom: tendon configuration 2. Both devices comprise a forearm splint with a mounted actuator and a glove implementing the desired tendon network.

that actuates each finger. All bifurcation points are rigid, with no differential mechanism installed to distribute loads. A number of existing cable-driven soft wearable hand devices have tendon attachment points on the fingertip of the glove [34, 19]. We have found that this design can produce finger hyperextension at the DIP joint. To alleviate this problem, the tendons attach on each finger to a cloth ring on the middle phalanx. Through IP joint coupling, this produces both PIP and DIP extention, without causing hyperextension.

On the dorsal side of each finger, raised tendon guides sitting on top of multiple layers of fabric are used to increase the moment arm of the extensor tendons around the joints. The increased moment arms allow us to reduce the linear forces applied to the tendons. For the index finger (and representative for the other fingers), the raised pathways have height of 8.5mm above the MCP joint and 7.5mm above the PIP joint; the cloth ring only protrudes 1.5mm above skin. For tendons on the palmar side of the joints, we have found that such increased moment arms are not necessary.

A load cell (Futek, FSH00097) is installed between the actuator and the magnet piece to measure the tension of the actuated tendon. The sensor has been calibrated to have a resolution of

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0.196N and can measure up to 50N.

## 3.3 Experiments

Our experiments were designed to provide initial validation of the approach with the intended target population of stroke patients. In particular, we aimed to verify the capability of the device to produce the expected patterns and ranges of motion, and to characterize the forces encountered, especially when assisting patients exhibiting various levels of spasticity.

Testing was performed with five stroke survivors, three female and two male. All testing was approved by the Columbia University Internal Review Board, and performed in a clinical setting under the supervision of Physical and/or Occupational Therapists. All subjects displayed right side hemiparesis following a stroke event; in all cases, experiments took place more than 6 months after the stroke. Subjects also exhibited different spasticity levels, ranging between 1 and 3 on the MAS.

The first step in the experimental procedure consisted in measuring the patient's range of motion in all digits as well as the wrist, and assessing the spasticity level on the MAS. The next step consisted of donning the orthotic device, consisting of the forearm splint and the exotendon glove. After donning, the motor on the forearm splint was connected to the tendon network via a breakaway magnetic mechanism as described earlier.

Starting with the linear actuator at full extension, we define one trial as one excursion of the actuator to the completely retracted position. Depending on the tendon network being used (tendon configurations 1 or 2 described above), this produced a given movement pattern of the subject's hand. Throughout each trial, we recorded both the actuator position and the tendon force levels reported by the load cell; both measurements were taken at a frequency of 100 Hz. When tendon forces exceeded the maximum load supported by the magnet, the breakaway mechanism disengaged and the actuator retraction completed without exerting any forces to the subject.

With each subject, we performed the following set of trials:

- 1 trial where we asked the subject to relax their hand and not apply any voluntary forces;
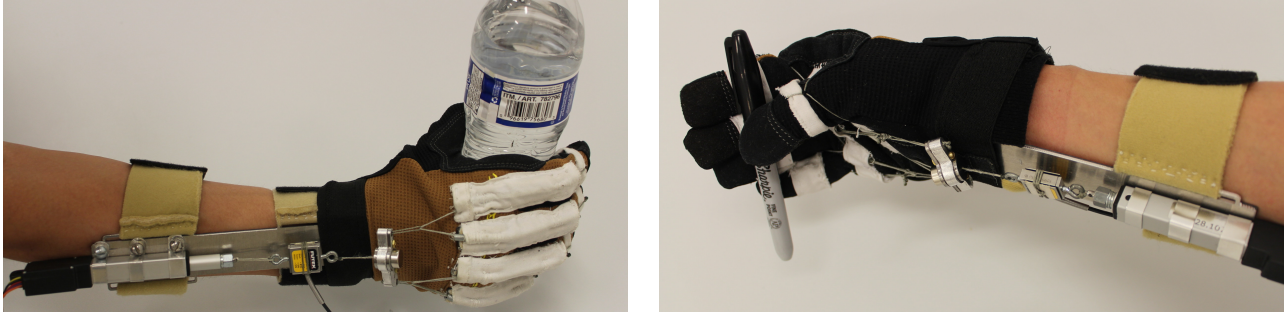
Figure 3.4: Illustrations of functional experiments with tendon configuration 1 (left) and tendon configuration 2 (right).

- 2-3 trials where we asked the subject to voluntarily assist the device in producing the intended movement pattern, to the best of their abilities;

- (for Configuration 1) 2-3 trials where the subject attempted to grasp an object (soda can). Starting from the subject's rest pose, the exotendon was engaged by retracting the linear actuator, providing finger extension. Once functional extension was achieved, the hand was positioned around the object and the exotendon was released allowing the subject to flex the fingers (illustrated by an able-bodied user in Fig. 3.4). If needed, the subject was assisted by the experimenter in positioning the arm such that the hand would be able to execute the grasp.

- (for Configuration 2) 2-3 trials where the subject attempted to execute a pinch grasp of an object (highlighter pen). Starting from the subject's rest pose, the exotendon was engaged by retracting the linear actuator, placing the hand in a pose appropriate for fingertip grasping a given object (illustrated by an able-bodied in Fig. 3.4). If needed, the object was positioned by the experimenter such that the hand would be able to execute the grasp.

After the completion of the procedure, subjects were asked to describe their impressions of wearing the device, any discomfort or pain produced by it, and any suggestions for improvement.

## 3.4 Results

One of the main objectives of our set of experiments was to determine the actuation forces needed to achieve functional hand extension in stroke patients exhibiting various levels of spasticity. Fig. 3.5 summarizes our results measuring actuation forces during trials with five stroke patients using tendon configuration 1 (full extension). We present one representative trial per subject; surprisingly, we found very little variation between trials where the subject was asked to relax and trials where the subject was asked to actively assist the device or to attempt a grasp. Throughout the trials, we recorded the applied force levels as a function of the position of the actuator. Hand opening in response to the device was observed by the experimenter and rated as functional (sufficient to grasp an object of approximately 55mm in diameter) or not; however, quantitative data for joint angles was not recorded.

The assistive hand device was able to achieve functional hand extension for 4/5 patients. In 2/5 cases the force level led to the breakaway mechanism disengaging; however, in one of those cases this occurred after functional hand extension was achieved. Overall, we were able to achieve functional hand extension for all patients with MAS spasticity levels of 1 and 2. Breakaway occurred after achieving functional extension for one patient with MAS spasticity at level 2, and without achieving functional extension for one patient with MAS spasticity at level 3. In all cases, the subject was able to complete an enveloping grasp of the target object, as described in the previous section.

The maximum level of recorded force varied between subjects: between 15-20 N for Subjects 1 and 3, between 25-30 N for Subject 2, and exceeding 35 N (and thus leading to breakaway) for Subjects 4 and 5. These results suggest that an exotendon assistive glove with a single-actuator able to apply up to 40 N to a tendon network similar to the one used here will succeed in generating functional hand extension for most patients with spasticity levels 1 and 2, but will not be strong enough to be used by patients with spasticity level 3.

An interesting finding concerns the observed relationship between force and position, used here

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as a proxy for hand pose. In all observed cases, the relationship was highly linear. To quantify this phenomenon, we first normalized the data as follows. First, as each trial generally begins with a small amount of slack in the tendon that is picked up as the actuator retracts, we removed all data points until the force first first reached a threshold of 3N. For cases where the trial ended by a disconnect of the breakaway mechanism, we also removed data points starting at the breakaway moment (observed as a sudden drop in force levels all the way to 0). Finally, we normalized remaining force and position values by dividing with the maximum observed value, and measured the correlation coefficient between the resulting data series.

For all the trials shown in Fig. 3.5, we obtained correlation coefficients ranging between 0.97 (Subject #2) and 1.00 (Subject #4). Fig. 3.6 shows the normalized data, along with the linear fit, for the trials with the highest and lowest correlation coefficient. These results suggest that the spastic muscle does not oppose movement with a fixed force level. Rather, it behaves in spring-like fashion, with resistance increasing linearly along with elongation.

We also carried out experiments with two stroke patients using tendon configuration 2 (MCP flexion / IP extension). Force vs. position data for these experiments is less relevant, since, in these cases the assistive device does not need to overcome involuntary forces in the opposite direction. Rather, we were interested in the functional aspect: does the assistive device in this configuration enable stable fingertip grasping. This was quantified as the ability to hold the object using such a grasp without external support while the assistive device was engaged and applying tendon forces. One of the subjects displayed this ability without the use of an assistive device; however, both subjects were able to perform this task with the use of the device.

Throughout the experiments, none of the subjects reported any pain or discomfort from using the device. However, subjective feedback repeatedly included suggestions to make doning the glove component of the device easier. We address the issue of donning in the next chapter.

## 3.5 Summary

In this chapter, we have proposed two tendon network configurations to assist full hand extension and fingertip pinch, each elicited by a single motor. In experiments with stroke survivors, we measured the force levels needed to overcome various levels of spasticity and open the hand for grasping using the first of these configurations, and qualitatively demonstrated the ability to execute fingertip grasps using the second.

The force measurements suggest a single linear actuator applying a total force below 40N can overcome hand spasticity and produced desired movement patterns for stroke survivors with MAS levels 1 and 2. This implies that lightweight wearable devices (we used an actuator with a total weight of 40g) can be effective, from a force generation perspective, for a significant range of the population affected by hand impairments as a result of stroke.

Our results also suggest that spastic muscles oppose movement in a spring-like fashion, with forces increasing linearly with elongation. In turn, this suggests that selection of actuators (and implicitly force levels) for assistive devices must take into consideration how applied forces will vary throughout the expected range of motion.

In addition, we demonstrated that multiple functional whole-hand movement patterns can be produced using a single-actuator for each. In particular, we showed two possible patterns: full extension (which combines with voluntary flexion to produce enveloping grasps) and MCP flexion / IP extension (which produces fingertip grasps). These results suggest that a single device with a small number of actuators can combine multiple such patterns, producing a wider range of manipulation capabilities.

Overall, we have verified that a small motor force can enable functional finger extension using our exotendon network design, which is highly significant from wearability perspective. This means *a wearable device within a small and lightweight package for assisting a multi-digit hand movement pattern* is feasible for stroke patients with moderate muscle tone. Also, the measured actuation force to overcome spasticity will be valuable data as we make progress towards a compact

wearable device.
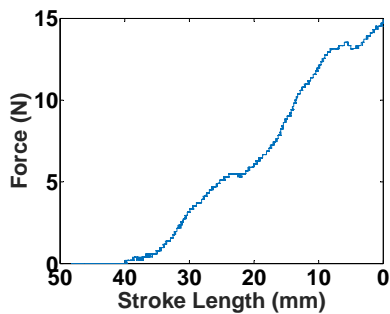
Subject #1. Functional extension was achieved. Spasticity level: 2.
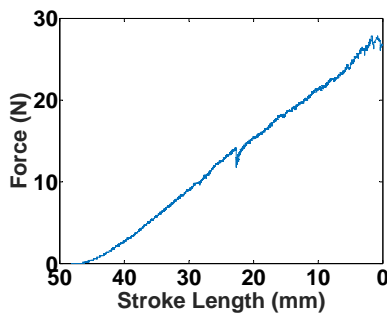
Subject #2. Functional extension was achieved. Spasticity level: 1.
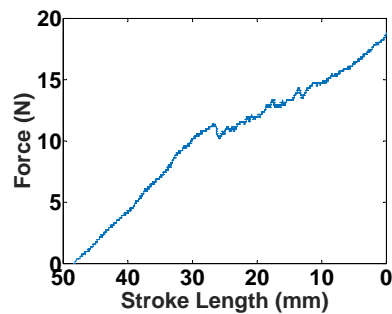
Subject #3. Functional extension was achieved. Spasticity level: 1 (except for index finger, rated at 2).
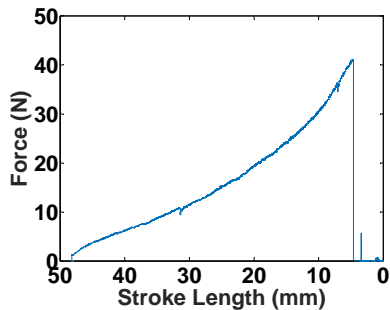
Subject #4. Functional extension was not achieved. Spasticity level: 3.
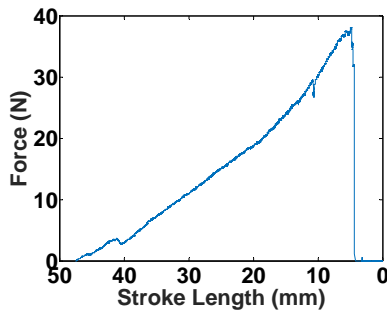
Subject #5. Functional extension was achieved. Spasticity level: 2.

Figure 3.5: Characterization of hand extension trials: force vs. position data. Each plot shows, for one trial, the relationship between the measured force in the actuated tendon and the linear position of the actuator. Note that each trial begins with the actuator fully elongated (50 mm actuator position) and slack tendon network (0N force). As the actuator retracts (left-to-right movement on the plots), we measure the force applied to the tendon network. If the force exceeds the maximum load supported by the magnet, the mechanism disengages producing a sudden drop in the force profile. This plot shows one representative trial from each of the 5 subjects tested using this movement pattern. For each trial, we also indicate whether functional extension (defined as sufficient hand opening to grasp an object of approximately 55mm in diameter) was achieved before the mechanism achieved maximum retraction or disengaged.
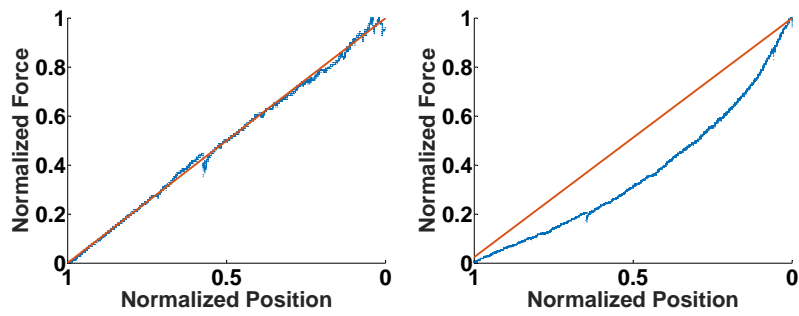
Figure 3.6: Normalized force vs. position data (blue) and superimposed linear fit (red) for trials exhibiting the highest (left, 1.00) and lowest (right, 0.97) correlation coefficient between these two variables.

# Chapter 4: Effective Transmission Mechanisms

In the previous chapter, we measured the level of force needed to overcome spasticity and open the hand for grasping through preliminary experiments. However, we observed that muscle tone of stroke patients can drastically increase during repeated use of the upper limb. Also, fatigue can influence the degree of spasticity in various levels. In this chapter, we focus on addressing this issue of increased stiffness on the finger joints such that the user can achieve functional hand extension.

Simply increasing the tendon force to overcome spasticity was not our prime option since it raises the chance of injury in case of malfunction and the size and weight of the motor. Efficient transmission, on the other hand, allows a reduction in the tendon force required to achieve functional finger extension. Lower tendon forces also reduce the unwanted phenomenon of distal migration, where due to the applied forces, the motor component of the devices slides on the forearm towards the hand. Both of these characteristics can lead to more wearable devices.

We propose two designs that provide effective force transmission by increasing moment arms around finger joints. We evaluate the designs with geometric models and experiment using a 3D-printed artificial finger to find force and joint angle characteristics of the suggested structures. We also perform clinical tests with stroke patients to demonstrate the feasibility of the designs.

|                     |                     |                     |                     |
| Traditional Design | Baseline Design | Proposed Design A | Proposed Design B |
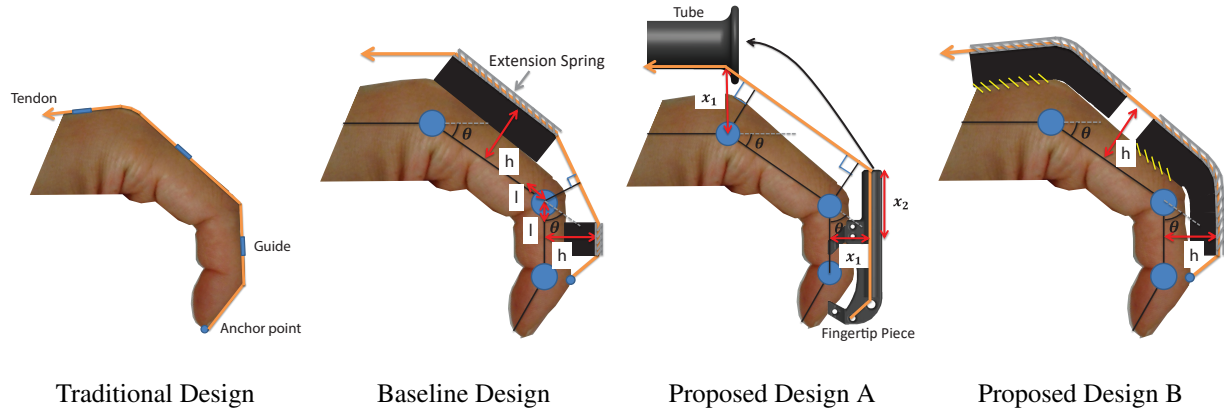
Figure 4.1: **Traditional Design** : Tubes or rings are installed on the surface of a hand as cable guides, and the fixed point is often located at the finger tip. **Baseline Design** : Raised pathways are attached on each phalanx, and the fixed point is at the head of middle phalanx. **Proposed Design A** : A 3D printed part on back of the hand works as an anchor point and another part is attached on the distal and middle phalanx. **Proposed Design B** : Two raised pathways are used, one between the palm to the proximal phalanx and one between the proximal and middle phalanges. Yellow lines indicate where each pathway segment attaches with the glove. The distal ends of the pathways hang freely, to avoid hindering finger flexion.

## 4.1 Design Criteria

Here, we outline the goals that drive our exotendon device development, in a manner independent of specific design choices. In the following, we present and compare several designs intended to achieve these goals.

*1) Achieve Functional Finger Extension*: Impaired finger extension is one of the most common after-effects in stroke patients. Since finger extension plays an essential role in functional grasp, this impairment adversely affects the quality of life. However, many individuals can still form a grasp in coordinated movement pattern. Given the volitional finger flexion capability, we require that our assistive device help the user achieve functional finger extension.

*2) Efficient Transmission*: Given the first requirement above, it follows that exotendons should apply significant extension torques around the MCP and IP joints to overcome spasticity. Use of a strong motor to achieve large extension torques is undesirable as it requires sizable motors and causes distal migration. Increasing tendon moment arms around the joints is an attractive alternative which avoids such unwanted effects. We thus look for effective transmission mechanisms that

increase torque for a fixed tendon force.

*3) Effective Torque Distribution*: In experiments with stroke survivors Cruz and Kamper [18] conducted, proportions of constant extension torques applied to keep the joints in the neutral position were approximately 0.03:0.66:0.46 for the distal interphalangeal (DIP), PIP, and MCP joint respectively. This means that the PIP joint typically exhibits higher tone than the other joints. Therefore, it is important to distribute proper amount of torques translated from a motor force to each joint for some patients.

*4) Wearability*: For a device to be used in the home environment, it has to be kept compact and lightweight while delivering meaningful assistance. Designing such hand devices is especially challenging, as available space on a hand is limited. To conform with this constraint, the designs presented here elicit movement using a single motor. In previous work, we have shown that a single motor can elicit the desired movement patterns [68, 60], but did not consider the effects of increased spasticity during functional tasks due to repetition and fatigue.

## 4.2 Transmission Designs

### 4.2.1 Designs

In this section, we introduce a number of possible device designs, which we will later compare and contrast from the perspective of our requirements. All the designs discussed here are illustrated in Fig. 4.1.

*1) Baseline Design*: The starting point is the simplest design where the tendon is simply routed on the dorsal side of the finger (Traditional Design in Fig. 4.1). With no moment arm increase, however, this is an ineffective way to achieve the torque levels needed to overcome spasticity and is included here only as a reference starting point. Furthermore, anchoring the tendon at the fingertip is likely to cause hyperextension of a DIP joint unless the range of motion is perfectly fitted with the user. For all other designs, we attach the tendon to the distal end of the middle phalanx.

The most direct way to increase the moment arm is to install a raised pathway for the entire
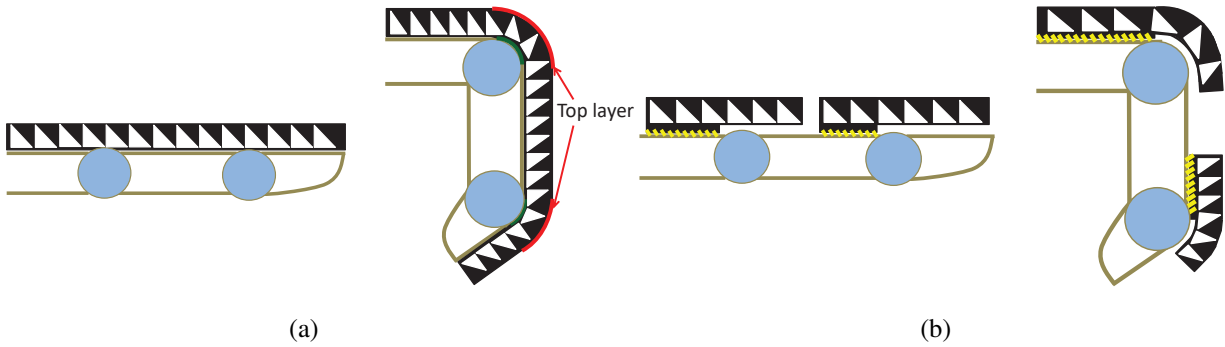
Figure 4.2: (a) Example of a raised pathway attached on a finger in extension (left) and flexion (right). Note that insufficient expansion of the top layer of the pathway hinders finger flexion. (b) Simple illustration of Design B on a finger in extension (left) and in flexion (right). Yellow lines show the attachment areas between the glove and the raised pathway. Since the pathway does not elongate in flexion, finger movement is not hindered.

tendon route. However, such a pathway must elongate to support finger flexion, creating elastic effects that hinder motion. This phenomenon increases with the height of the pathway, as the top layer must elongate even further. This behavior is illustrated in Fig. 4.2a.

Our Baseline Design thus consists of raised pathways separated section-by-section to avoid the interruption of finger flexion. Although moment arms around the joints are increased with this design, the cable takes a shortcut between the pathways leading to a shorter moment arm when the finger is not fully extended (illustrated in Fig. 4.1). In this study, we use this structure as a baseline to make comparisons among envisioned designs.

*2) Design A*: The main goal of Design A is to achieve a greater moment arm around the PIP joint, which typically exhibits the strongest spasticity among the three joints of the finger. In this design, we implement two 3D-printed parts as cable guides (Fig. 4.1).

A fingertip piece is mainly used to increase a moment arm around the PIP joint. This component also mechanically prevents hyperextension of the DIP joint while assisting finger extension. Fabric straps are secured around the finger using velcro to maintain the position of the device on the hand as depicted in Fig. 4.4. A funnel shaped tube is installed on the dorsal side of the hand to increase the moment arm around the MCP joint. However, for a small handed person, this palmar component may collide with the fingertip component in full finger extension. Therefore, the fun-

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nel tube is designed to allow the fingertip component to be inserted into the dorsal component to enable full range of motion.

Two parameters, $x_1$, the normal length between the center of the PIP joint and tendon location and $x_2$, the length between a support of the fingertip piece and the end of the fingertip piece, determine the moment arm around the PIP joint depending on the joint angle $\theta$.

To learn how the two parameters contribute to the geometric characteristics, we have recorded the moment arm around the PIP joint. For simplicity, we assume that the PIP and MCP joints are simultaneously moving with the joint angle $\theta$, and the range of motion for both joints is from $-90\,^\circ$ (fully flexed) to $0\,^\circ$ (fully extended). Also, we limit the range of $x_1$ and $x_2$ to avoid designs that are either too bulky or ineffective.

The results indicate that $x_1$ is more responsible for torque generation when the finger is extended, whereas $x_2$ is more influential for flexed positions, as shown in Fig. 4.3-(a), (b). With chosen parameters, the PIP and MCP joints are recorded across finger motions with results plotted in Fig. 4.3-(c), and it shows that the moment arm around the PIP joint is longer than around the MCP joint.

*3) Design B*: Design B aims to maintain moment arms around the PIP and MCP joint at preset length $h$ and avoid interference with grasps during hand motions. The design consists of a glove, extension springs, and two raised pathways on each finger (Fig. 4.1). The pathways are placed on top of the middle and proximal phalanges, and back of the hand covering the PIP and MCP joint. This prevents the tendon from taking a shortcut that reduces the moment arms around the two joints. In order to avoid hindering finger flexion, raised pathways are rigidly secured proximally to the PIP and MCP joint while the other ends distally located to the joint are free to slide without direct attachment(Fig. 4.2b). A cloth cover is sewn on top of the pathway to prevent it from drifting laterally during use.
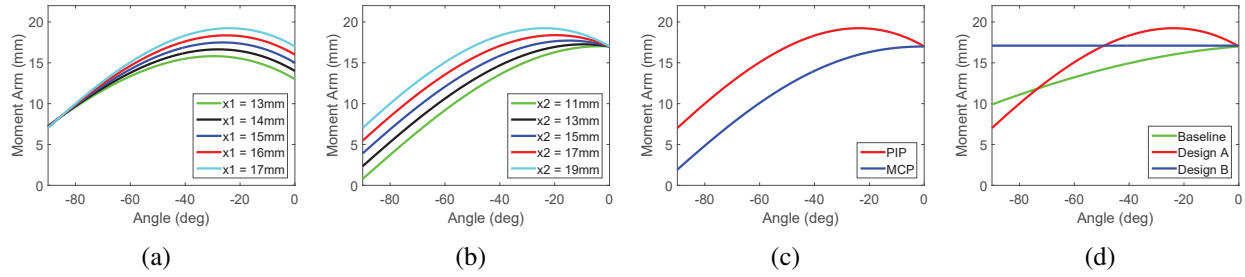
Figure 4.3: (a) Moment arm around the PIP joint vs. joint angle for different $x_1$ where $x_2$ is fixed at 19mm. (b) Moment arm around the PIP joint vs. joint angle for different $x_2$ where $x_1$ is fixed at 17mm. (c) Moment arm around the PIP and MCP joint vs. joint angle for Design A. (d) Moment arm around the PIP joint vs. joint angle for Baseline design, Design A, and Design B.

### 4.2.2 Design Comparison

Fig. 4.3-(d) shows a moment arm around the PIP joint versus joint angle $\theta$ for Baseline design, Design A, and Design B. The length between the center of each joint and the tendon is set to 17mm at fully extended position for all three designs.

The result indicates that Design A and Design B generate a larger moment arm around the PIP joint than Baseline design throughout hand motions. The moment arm with Design A in an extended finger position is notably larger than the others, which is beneficial considering that a proportional increase in extension torque is required as the finger is in a more extended position [68]. Design B also creates a larger moment arm than Baseline design.

From the outcomes, one can assume that the force level required to extend the finger by Baseline design would be the greatest. For Design A, the extension would require relatively lower force level on the PIP joint than the MCP joint as the moment arm around the PIP joint is larger. Finally, in Design B, the PIP and MCP joints would need similar level of force to execute full extension because the geometry of the design for both joints is relatively similar.

### 4.2.3 Exotendon Device

The designs described above are used in a combination with the exotendon device previously used in the previous chapter with little modifications. Mechanical components of the device are

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composed of a forearm piece with actuation and a structure based on the two designs (Fig. 4.4). This structure engages the impaired hand with a motor on the forearm piece through a tendon network. An S-hook connects the tendon network from the end effector with the motor to facilitate the donning process.

The forearm piece works as both an anchor point to stabilize a base of the motor and a splint that constrains the wrist movement to efficiently transmit the motor force to the end effector. A DC motor (Pololu corporation, 47:1 Medium-Power 25D Metal Gearmotor) with a 100N peak tendon force is mounted on the forearm piece. The motor is driven by Proportional-Integral-Derivative(PID) position controller, and the range of motion is determined at the clinical test after fitting the device. A simple push button is implemented to trigger finger extension. While pushing the button, the motor stalls when the applied motor force reaches its maximum level or the fingers arrive at the fully extended position. Releasing the button allows the fingers to flex and the hand to grasp. The DC motor applies an extension force to all four fingers except for the thumb.

## 4.3   Experiments and Results

In order to evaluate the theoretical results shown in the previous section, we have conducted experiments using a 3D-printed artificial finger to find a relation between joint angles and applied force. In addition, we performed clinical tests with stroke patients to provide validity of the results from simulations and experiments with an artificial finger. In the experiments, subjects wore a device with each of Baseline design, Design A, and Design B at a time, and the joint angles of an index finger were measured while the device was assisting finger extension. We provide a comparison of the range of finger extension elicited by the three designs.

**Testing with Artificial Finger**

The artificial finger consists of 3D printed parts, torsion springs, and encoders(Fig. 4.5a). Torsion springs are installed on all three joints to mimic hand spasticity. A proportion of the spring constants is 3.5:76.9:54.9 for the DIP, PIP, and, MCP joint respectively, which is similar

Figure 4.4: Prototypes of the hand orthosis transmission mechanisms (top: Design A, bottom: Design B) proposed in this study.

to 0.03:0.66:0.46 from Cruz and Kamper's work [18]. Hall effect rotary encoders(AS5600) are placed on the side of the PIP and MCP joint. The DIP joint, which is coupled with the PIP joint is excluded from the measurements for the sake of simplicity. Since four fingers other than the thumb have similar structures and exhibit identical movements experiments with one fingered device should suffice.

The main objective of this experiment is to measure force characteristics throughout an entire hand motion assisted by the device with Baseline design, Design A, and Design B. To measure a tension on the actuated tendon, a load cell(Futek, FSH00097) is installed in series with a motor and the tendon network. The force is recorded at 100Hz while the motor applies extension torques to the fully flexed artificial finger ($-90\,°$) till the finger is fully extended ($0\,°$).

For reliability, the measurements have been taken 50 times for each design. The average values of force vs. joint angles are shown in Fig. 4.5b. The result shows higher force is required for Baseline design than the others to achieve finger extension. Also, note that differences in force requirement are more prominent in flexed positions as simulation results from the previous section suggest. For Design A, the PIP joint requires less force than the MCP joint across hand motions as evidenced from Fig. 4.3-(c).

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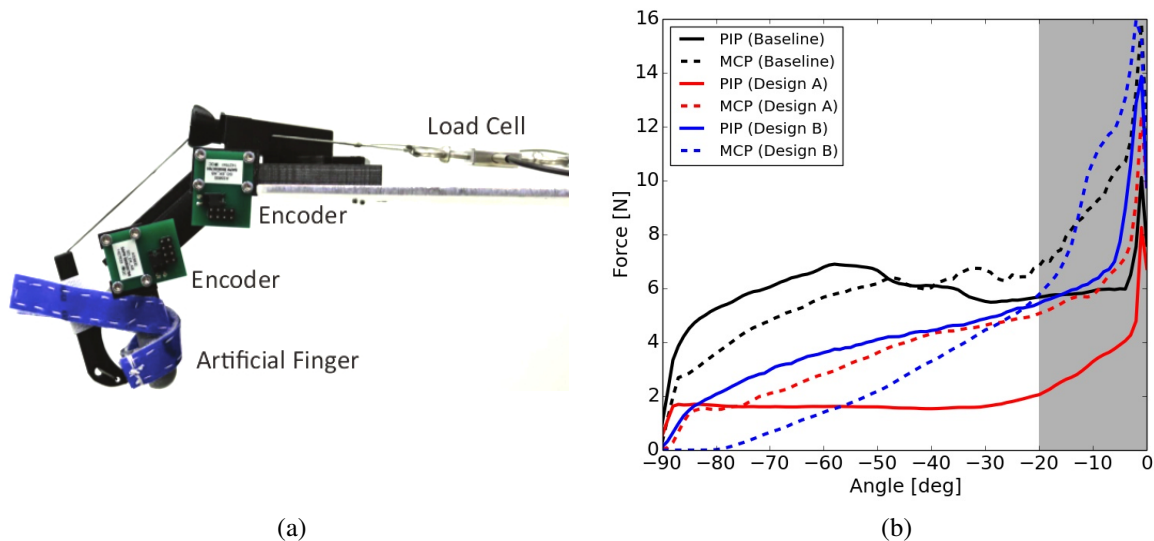
Figure 4.5: (a) Experimental set-up with an artificial finger. (b) Force vs. PIP (solid) and MCP (dashed) joint angles from artificial finger experiments. Black, blue, and red colors indicate force needed to reach certain joint angles with Baseline design, Design A, and Design B respectively. Particular ranges of our interest span from $-90°$ to $-20°$ as this region encapsulates the necessary functional movements.

### Testing with Stroke Patients

Four participants, one female and three male, with right side hemiparesis and limited mobility following a stroke event at least 6 months prior were recruited from a voluntary research registry of individuals who have survived stroke. Testing was approved by the Columbia University Institutional Review Board and took place in a clinical setting under the supervision of licensed physical and/or occupational therapists. Upper limb spasticity measurements were between 0 and 2 on the MAS for all participants (Table 4.1).

Each testing session was performed over the course of one visit. Spasticity scores at the elbow, wrist, and digits were assessed using the MAS before and after testing. Subjects were then fitted with the exotendon orthotic device and guided through the following procedure with each version of the device.

- The subject opens the hand using the orthosis. Extension of the PIP and MCP joints of the index finger are measured using a goniometer. The index finger was selected for measure-

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| | MAS Extensor Score | | | MAS Flexor Score | | |
|---|---|---|---|---|---|---|
| Subject | Elbow | Wrist | Finger | Elbow | Wrist | Finger |
| A | 1+ | 0 | 0 | 1 | 0 | 1 |
| B | 2 | 0 | 0 | 2 | 1 | 1+ |
| C | 1 | 0 | 1 | 1+ | 1 | 1+ |
| D | 0 | 0 | 0 | 0 | 0 | 1 |

Table 4.1: Subject clinical information

ment as it was most accessible with the device in place; it was also qualitatively observed to be representative of the four fingers.

- The subject attempts to grasp and release 15 times to induce fatigue. The device is triggered to assist hand opening using a button at the point of maximal effort. On the 15th repetition, the joint angles of the index finger are measured again. In general, fatigue increases tone in hand movement, and this measurement is taken to see if one can still achieve functional hand extension in this condition.

- The subject takes a rest for five minutes to reduce the impact of fatigue. Then, the last measurements of the joint angles of the index finger are recorded while the device is assisting.

- To avoid effects of fatigue carrying over to the next trial, the subject rests for ten minutes between trials with different devices (including time spent on doffing and donning the devices).

Fig. 4.6 shows the average of measured PIP joint angles with all participants. Since the MCP joint was fully extended for every patient, only the PIP joint angle was measured.

The result demonstrates a comparative advantage of Design A and Design B over Baseline design. In particular, as patients became fatigued, the Baseline design generally failed to elicit functional extension, whereas Design A and Design B were less vulnerable to increased tone after activities. This result also suggests that assessing the feasibility of a hand device through range of motion measurements without the integration of functional tasks may not be representative of real life use. For a hand device to allow repetitive exercises, post-fatigue evaluation should also confirm the effectiveness of the device.
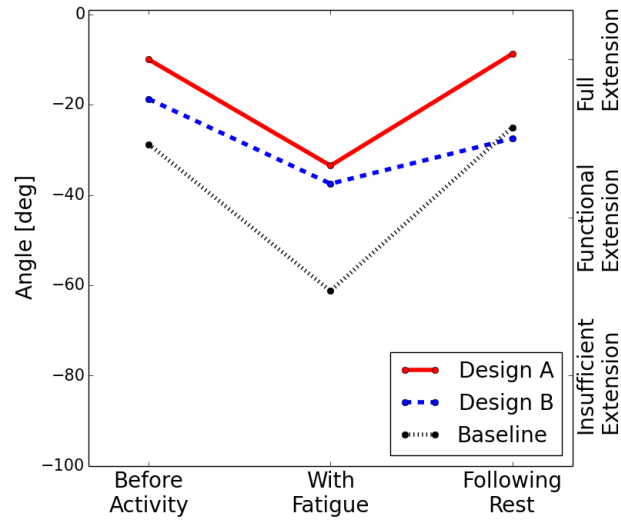
Figure 4.6: Joint angles of the index finger measured with stroke patients while the device with Baseline design(black dotted line), design A(red solid line), and design B(blue dashed line) is assisting finger extension.

| Version | Before Activity | With Fatigue | Following Rest |
|---|---|---|---|
| Design A | -10.0(±5.7) | -33.5(±6.4) | -8.8(±5.9) |
| Design B | -18.8(±11.9) | -37.5(±20.5) | -27.5(±16.0) |
| Baseline Design | -28.8(±12.6) | -61.3(±18.1) | -25.0(±16.0) |

Table 4.2: Joint angles of the index finger measured with stroke patients(mean± standard error)

## 4.4 Summary

In this chapter, we have proposed distal structures of a tendon driven hand assistive orthosis for efficient force transmission. In order to evaluate the designs, we ran simulations with mathematical models and conducted experiments using a 3D-printed artificial finger. We also performed clinical tests with stroke patients to study real-life applicability.

In the geometric model analysis, the two designs we propose enabled torque generations with large moment arms around joints of a finger, compared to traditional devices. The advantages of the large moment arm were demonstrated by the experiments with an artificial finger. The result suggests that the force required by proposed structures to extend a finger was lower than Baseline design. Clinical trials with four stroke patients where we measured joint angles with the device assisting finger extension also supported the feasibility of the effective mechanism. In the experiments, all four participants attained a functional finger range of motion even when fatigued with the assistance of Design A and Design B.

Efficient transmission mechanisms in a wearable tendon driven device offer several advantages. Since the required level of force is lower, the size of the device can be reduced, and the likelihood of injury decreased. It means that the operation is more robust because *the patients can use the device even when their muscle tone is higher than usual due to fatigue or abnormal muscle synergy.* Also, small and light actuators can be placed closer to the affected hand, further increasing wearability. Finally, a low tendon force leads to reduced distal migration, which is a limitation of active devices.

# Chapter 5: Multimodal Intent Inferral

With our development in the previous chapters, we now have a hardware capable of providing needed actuation in a compact, wearable package, allowing greater freedom and flexibility than their workstation-like counterparts. Such wearable devices could allow use beyond the confines of a therapist's office. However, the vision of a wearable orthotic device used for ADLs can only be realized if the patients are able to operate the device themselves. Control methods must be effective and intuitive, robust to long term operation, and cannot impose significant cognitive load. These algorithms must also cope with a wide range of impairment levels and abilities in the target population. While the actuation abilities of robotic hand orthoses have made great strides, control algorithms have not made similar progress in addressing these challenges.

In an effort to realize the intuitive control, we have previously developed an intent detection method using an EMG commodity band [60]. However, the variety of encountered upper limb impairment patterns in stroke patients means that a single sensing modality, such as EMG, might not be sufficient to enable controls for a broad range of users. In this chapter, we introduce a multimodal sensing and interaction paradigm for an active hand orthosis to address this significant gap. In our proof-of-concept implementation, EMG is complemented by other sensing modalities, such as finger bend and contact pressure sensors. We propose multimodal interaction methods that utilize this sensory data as input, and show they can enable tasks for stroke survivors who exhibit different impairment patterns.

To equip a hand orthosis with multimodal sensing, we expand upon our work with exotendon hand device (Configuration 1 in Chapter 3 and Design A in Chapter 4). See Fig. 5.1a and Fig. 5.1b for actual implementation. Tendon driven systems require less space than linkage-based exoskeletons, as they utilize few, small anchoring structures. Therefore, they are well-suited for sensor
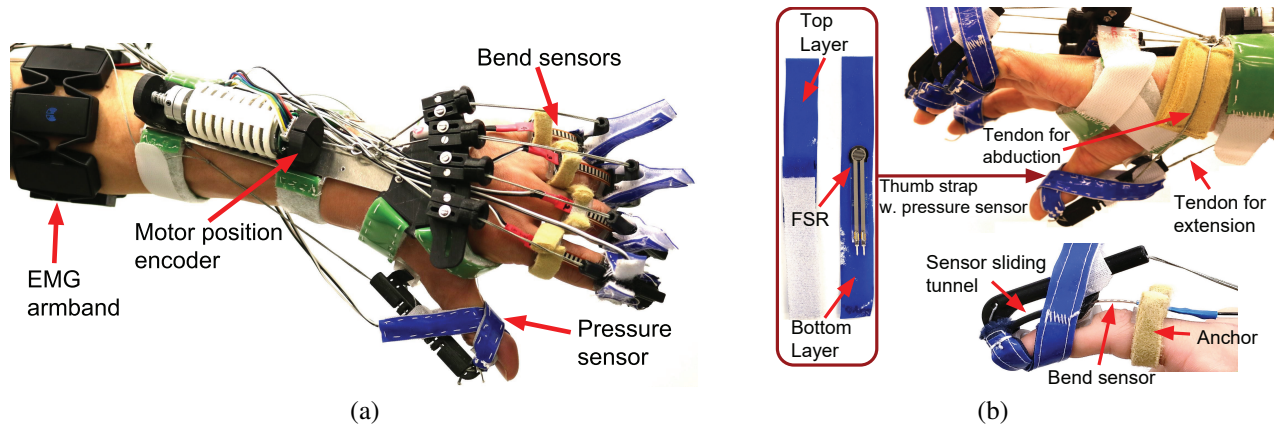
Figure 5.1: (a) Prototype of a hand orthotic device with multimodal sensors. (b) Integration of a pressure sensor into a thumb strap (left), tendon routing for thumb extension and abduction (top) and bend sensor implementation (bottom).

implementation.

The 3D-printed fingertip components are secured to the fingertips using Velcro straps. The underside of the strap is rubber to prevent distal migration. The components route the exotendons through raised pathways that enhance force transmission by increasing the moment arm around the PIP joints. In addition, the components prevent hyper-extension of the DIP joint and serve as an anchoring point for the tendons.

The thumb moves differently than the other four fingers, and therefore requires different routing. As long as the four fingers are sufficiently extended, we can enable grasping tasks by simply splinting the thumb in a stationary, opposed position [2]. We splint the thumb using two tendon routes, which adjust the thumb's abduction and extension (Fig. 5.1b).

## 5.1 Multimodal Sensing

While existing work has focused primarily on robotic hand orthoses as actuation devices, we envision future devices serving an equally important role as sensory platforms, equipped to characterize physical interaction with the user. Numerous sensing modalities can be envisioned, focusing on tendons, joints, contacts, etc. In this context, we have developed a multisensory platform prototype, combining sensors for the following: forearm EMG, motor position, fingertip pressure, and

joint angles (Fig. 5.1a, 5.1b). We describe these sensing modalities and their integration with the orthosis next.

### 5.1.1 Forearm EMG

EMG is one of the most common orthotic controls because it is intuitive. EMG sensors are low profile, and commercial devices, like the one used in this work, are easy to don and doff. With relatively simple algorithms, EMG sensors can be used to identify a variety of different hand poses.

We use the Myo Armband from Thalmic labs for our EMG sensing. The armband consists of eight EMG sensors and is placed on the subject's forearm, proximal to the splint. Our pattern recognition algorithm (Section 5.2.3) uses the EMG sensors to predict the user's intended hand state. Fig. 5.2a shows an example of EMG activation patterns as a subject attempts to open and close their affected hand. In this figure, the EMG activation for open and close is distinct; however, these patterns will change over time as the subject fatigues.

### 5.1.2 Motor Position

Motor position sensing is commonplace in robotic devices, and we include its description here for completeness. The motor encoder provides high-resolution position feedback, which enables us to control the actuator with position control and determine the current state of the orthosis. Because our tendon network is underactuated, this feedback does not provide information about individual finger behaviors, but their combined movement pattern.

### 5.1.3 Finger Joint Angles

Joint angles can serve as cues to determine patient intent. One typical pattern is partial voluntary movement, where patients try to open their hand and some fingers partially extend. Another, abnormal, movement pattern from which the sensing modality can potentially benefit is overactive stretch response [40], which exhibits finger flexion when patients try to extend. By measuring PIP joint angles with bend sensors, both movement patterns can give us information about user intent.

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(a) Forearm EMG activation



(b) Bend sensor data and its time derivative



(c) Fingertip pressure and the time derivative of the pressure
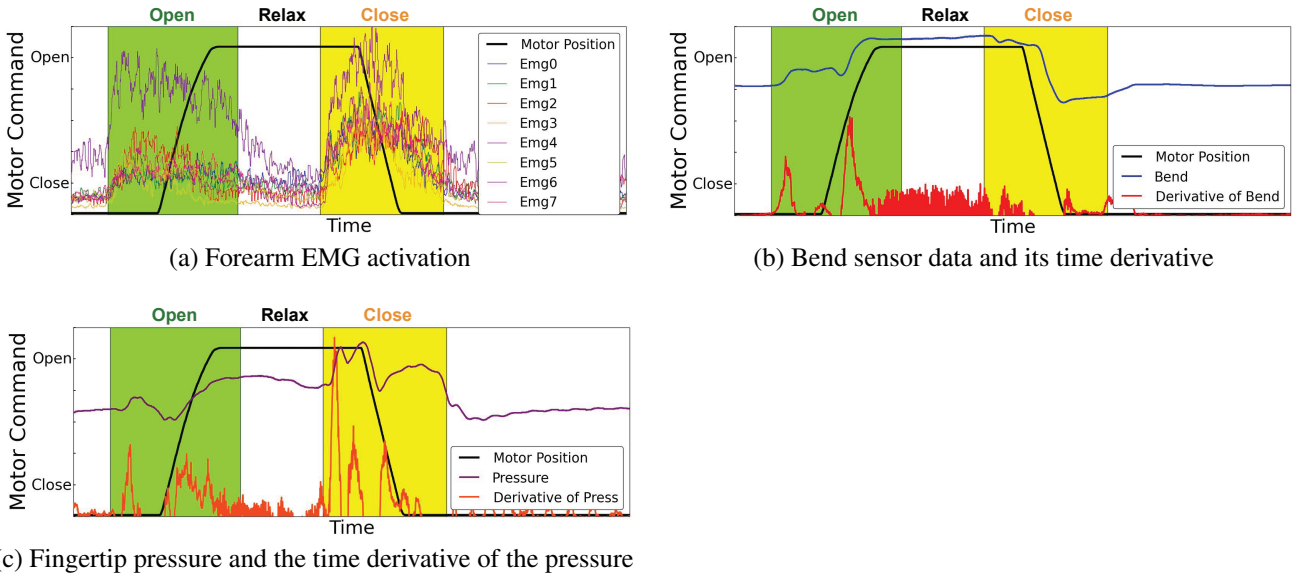
Figure 5.2: Sensor data recorded while a user with stroke opens (green) and closes (yellow) their affected hand during training. The plots depict the following sequence (verbal instructions are given to the user throughout): the user, with hand at rest, tries to open. After a few seconds, the device is actuated to assist the open. The user relaxes once the hand is open (tendon fully retracted). Then, the user tries to close, despite resistance from the orthosis. Shortly after, the device extends, allowing the hand to close.

We use a bend-sensitive resistor on each finger to measure joint flexion of the PIP joint (Fig. 5.1b). We assume residual movement of the PIP joint is greater than the MCP joint and therefore only deploy sensors on the PIP. For each finger, the proximal side of the sensor is anchored to the subject's proximal finger link by a strap. The distal side of the bend sensor is fed through a flat hole in the bottom of the fingertip component to keep it close to the distal link of the finger.

We found that using a simple threshold on the raw bend sensor data to trigger an open command was limited as a control because motor position and the size of the objects with which the user interacts both dramatically affect the raw data values. Fig. 5.2b shows the raw bend data and bend derivative during an example open-close motion. Note that the bend derivative peaks soon after the subject is asked to open. The next notable maximum is caused by the device extending the fingers.

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### 5.1.4 Fingertip Pressure

Pressure sensors on the fingertips serve a dual role: since the digit straps are the conduit by which exotendons apply force to the fingers, the pressure sensor can record the level of force between the hand and the device. When the user is performing a grasp, the pressure sensors will also record the contact force between the hand and the object. In this way, pressure sensing allows us to paint a complete picture of force transmission, from the orthosis to the patient's hand, and from the hand to the environment.

Fingertip pressure increases when the subject is either interacting with an object or trying to close the hand while the device is open. Though we cannot differentiate between the two actions, the increase in pressure gives us useful information about when the user intends to close their hand, especially when the user cannot maintain the muscle activation necessary for detection via EMG.

Again, we use the time derivative of the pressure data rather than the raw data. As shown in Fig. 5.2c, both the raw data and the pressure derivative increase soon after the subject is asked to close their hand, but the derivative provides more robust cues because the raw data alters over time due to fatigue and irregular tone.

We fit our exotendon device with pressure sensing using force sensitive resistors (FSRs). FSRs are compact enough for integration inside the digit straps which attach the 3D printed fingertip components to the subject's fingers. Fig. 5.1b shows how the FSRs are placed inside the digit straps.

For simplicity, we integrate pressure sensing only on the thumb because it is the finger which generates the greatest force when the subject tries to close their hand [42]. The thumb is also used in all gross grasping, ensuring we will see interactions between the subject and any grasped objects.

## 5.2 EMG Control

Even in a multimodal context, EMG is still a key sensing modality. We describe here our EMG-based control intent method, developed initially as a single-modality control [60] and then updated to be used as one modality among many. For additional details and experiments regarding EMG as a single-modality control, we refer the reader to the companion publication [60].

One of the key tenets of the approach is to rely on signals from a multitude of sensors placed around the forearm. Unlike simple intensity thresholding, which is effective for a single sensor precisely located on a specific muscle, pattern classification identifies patterns in the complete set of signals from the sensors. This approach has three main benefits:

1. It enables the use of commodity sensors. Even though the quality of the EMG signal from commodity sensors is lower than medical grade sensors, we compensate for signal quality with sensor quantity. Pattern classification provides an image of the overall EMG signal in the entire forearm instead of trying to isolate a high quality signal from specific muscles.

2. It eliminates the need to search for specific muscles with exact sensor placement. Pattern recognition examines EMG signals from the entire forearm. Studies have suggested that when electrodes are placed around the entire forearm, targeted and untargeted placement of EMG electrodes result in similar classification accuracies [22]. Throughout our experiments, the only effort to position our EMG sensors was placing one of the sensors on the dorsal side of the arm. Even with this untargeted approach, we were still able to use pattern classification with good accuracy. The flexibility in sensor placement means that donning our control unit does not require a therapist, or even a basic understanding of forearm anatomy. For a device that is designed for take-home use in mind, this is an extremely desirable quality.

3. It allows for the possibility of an orthosis with more DOFs. Current orthoses look at two specific muscles, a flexor and an extensor. The flexor controls the close motion of the orthosis and the the extensor controls the open motion. Pattern classification allows for the

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recognition of more complex muscle motions, which could control different DOFs of the orthosis [73].

To acquire the EMG signal, we use the Myo Armband from Thalmic Labs. It has 8 EMG sensors and 8 IMUs, which can indicate the orientation and acceleration of the device. In this study, we only use the EMG sensors.

### 5.2.1 Pattern Classification

Our pattern classification algorithm seeks to take the 8-dimensional raw EMG data from the 8 Myo sensors and identify patterns that correspond to certain desired hand motions. The current algorithm only identifies hand opening and closing.

We collect raw EMG data from the Myo Armband at a rate of 50Hz. At time $t$, we collect the EMG signals $e_j^t$ from the sensors and assemble them into a data vector $\boldsymbol{x}_t$:

$$\boldsymbol{x}_t = (e_1^t \ldots e_8^t) \tag{5.1}$$

We aim to predict two possible user intention: to open the hand (*Intent=Open*) and to close the hand (*Intent=Closed*). While training, ground truth data is provided by the experimenter who gives the subject verbal commands to open or close the hand. The training period is around 45 seconds - allowing the experimenter to command the user to try to open and close the hand twice. Although this training time is short, we receive a large quantity of data points ($\sim$2,400) which we use to establish patterns in the EMG with our classifier.

Our first order goal is to predict user intention based on EMG signals from the sensors. We use a random forest classifier trained on the ground truth data described above to make this prediction. A random forest classifier is an ensemble machine learning method created from a combination of tree predictors [8]. Because of the random nature of the bootstrap sampling used to create our classifier, the number of decision trees in the forest classifier and the decision trees themselves change with every training iteration. Despite the underlying randomness, our classifiers for all

subjects still achieve high accuracy.

We denote the random forest classifier function as:

$$CLAS(\boldsymbol{x}_t) = p_t^O \in [0, 1] \qquad (5.2)$$

where $p_t^O$ is the probability of *Intent=Open* at time t. The converse probability that the user's intent is to close the hand is simply $p_t^C = 1 - p_t^O$. We filter and use this result as described in the following.

### 5.2.2 Output Processing

We collect raw EMG data $\boldsymbol{x}_t$ at a rate of 50Hz. However, the time scale for hand opening and closing and for pick and place tasks is much lower frequency than the rate at which data is collected, so classifying individual data points correctly is not as crucial as correctly identifying a hand motion. To identify these motions, we assume hand posture does not change with high frequency, which allows us to filter and process the probabilities returned by the classifier.

While filtering raw EMG signals is a common technique, we chose instead to apply our filter to the results of the classifier. We compute filtered probabilities at time $\hat{T}$ as:

$$\hat{p}_{\hat{T}}^O = \texttt{MEDIAN}(p_t^O), t \in [\hat{T} - 0.5s, \hat{T}] \qquad (5.3)$$

$$\hat{p}_{\hat{T}}^C = \texttt{MEDIAN}(p_t^C), t \in [\hat{T} - 0.5s, \hat{T}] \qquad (5.4)$$

The 0.5s median filter increases transition delays, but helps eliminate spikes and spurious predictions. 0.5s was chosen because shorter filters resulted in spurious classification errors. Despite the delay, our subjects reported no noticeable delay between intention initiation and device movement. We note that, as a result of filtering, generally $\hat{p}_{\hat{T}}^O + \hat{p}_{\hat{T}}^C \neq 1$.

To produce the final output for our control, we compare $\hat{p}_{\hat{T}}^O$ and $\hat{p}_{\hat{T}}^C$ against two threshold levels, $L^O$ and $L^C$ respectively. If $\hat{p}_{\hat{T}}^O \geq L^O$, then the controller issues an *Device=Open* (retract the tendon). If $\hat{p}_{\hat{T}}^C \geq L^C$, then the controller issues a *Device=Closed* (extend the tendon). If

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neither condition is met, no new command is issued and the orthosis continues executing the command from the previous step. The values of $L^O$ and $L^C$ are set manually by the experimenter for each subject after completing training data collection, then kept constant throughout all tests. The thresholds are set with subject feedback such that the control is responsive, but there are no spurious errors during sustained hand commands.

### 5.2.3   Integration in Multimodal Framework

As described so far, the EMG control can be used standalone; additional details on its performance with stroke patients can be found in the study by Meeker et al. [60]. However, for the purpose of multimodal control, we introduce a number of changes. We use the same eight sensor EMG armband (Myo) and a similar pattern recognition algorithm as previous work introduced above. The main difference in this work is that we aim to predict three possible user intentions rather than two: to open the hand (*Intent=Open*), to close the hand (*Intent=Closed*), and to relax (*Intent=Relaxed* - newly introduced here). The addition of the *Intent=Relaxed* class allows the user to open the hand using the exotendon device, and then relax their hand while they are positioning their arm, for example in order to execute a pick and place task, without having to continue to exert effort to keep the hand open. We believe this approach can help avoid muscle fatigue.

To classify user intent at a given time, we input the EMG signals collected at that time into a random forest classifier. The classifier outputs three values, each being the probability that the EMG signals belong to a corresponding intent class. These three probabilities are put through a median filter (0.5 s window) in order to eliminate spurious predictions. Finally, we compare the output probabilities from the median filter to three manually set thresholds. If the probability for a class exceeds the threshold, we classify the end result as belonging to that class. The end-result belongs to either the *Intent=Open*, *Intent=Relaxed*, or *Intent=Closed* class. We assign thresholds such that only one class can exceed a threshold at a time. If none of the thresholds are exceeded, the intent remains the same as at the last time step.

The EMG control can then issue motor commands to the exotendon device based on the pre-

dicted user intention. If the EMG control predicts that the user's intention is *Intent=Open*, the device is commanded to open (retract the tendon, thus extending the fingers). If the user's intention is *Intent=Closed*, the device is commanded to close (extend the tendon, thus allowing the user to flex the fingers). If the predicted user intention is *Intent=Relaxed*, we continue to send the previous motor command to the device.

## 5.3  Multimodal Control

We propose two types of multimodal control. Subjects in our target population display a wide range of impairment patterns. Some cannot maintain a "close" EMG signal, and others have more voluntary finger extension. A single sensing modality is limited due to the various impairment patterns; similarly, multimodal sensing is limited if it does not fit the subject's impairment pattern.

We propose one kind of multimodal control where bend sensors detect the user's intention to open the hand, and EMG sensors detect the user's intent to close the hand. The other multimodal approach uses pressure sensors to detect the user's intent to close the hand, and EMG sensors to detect the user's intent to open the hand. The multimodal approach used for each of our subjects was chosen based on a qualitative analysis of their abilities, such as range of voluntary finger extension, and ability to maintain EMG signals.

**Bend to Open, EMG to Close**: The first multimodal control uses bend sensors to determine when the exotendon device should open, and EMG sensors determine when the device should close. Subjects who use this control would typically have the ability to initiate finger extension, but be unable to achieve functional extension and have difficulty maintaining an "open" signal for EMG.

To determine user intent based on voluntary extension, we collect data from the four bend sensors built into the orthosis. In the current version, the therapist determines which of the subject's fingers has the greatest range of voluntary motion and we focus on bend data from that specific digit; in the future, we plan to integrate the data from all four sensors. Bend data is then passed through a moving mean filter with a window size of 0.25 s. We take the derivative of the resulting

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signal, which we refer to as $\frac{\partial b^i}{\partial t}$ (where the subscript $i$ denotes the digit found to have the highest voluntary range of motion). Motor commands are sent as follows:

- When the orthosis is in the *Device=Closed* position (tendon extended allowing fingers to flex) and $\frac{\partial b^i}{\partial t}$ exceeds a given threshold $L^B$, the device is commanded to open (retract the tendon).

- When the orthosis is in the *Device=Open* position (tendon fully retracted, or motor stalled) and the EMG classifier predicts *Intent=Closed*, the device is commanded to close (extend the tendon).

- If neither of the above conditions are met, we continue to send the previous motor command to the device.

The threshold $L^B$ is determined based on the training data collected in the procedure described in Section 5.3.1. For the training dataset, we find the local maxima of $\frac{\partial b^i}{\partial t}$ while we ask the subject to try to open. We select the smallest value between the local maxima as $L^B$. If necessary, the experimenter will manually tune the threshold so the control can enable tasks. After the threshold is set, it is kept constant throughout all tests performed by the subject.

When the device is in the *Device=Closed* position, EMG signals are ignored, as are bend signals when the device is in the *Device=Open* position. Furthermore, our control will not switch motor commands while the device is transitioning from *Device=Open* to *Device=Closed* or from *Device=Closed* to *Device=Open*. We note that although this consideration can reduce rapid oscillations in the motor command, it is limiting if the subject only wants to open their hand halfway and then close again, for example, when grasping small objects. If the subject starts closing their hand before the motor is done transitioning, they will encounter resistance from the orthosis until the transition finishes and the control issues another command to the motor.

**EMG to Open, Pressure to Close**: For the second kind of multimodal control, EMG sensors determine when the exotendon device should open and the pressure sensors determine when the device should close. Subjects who use this control typically have a clear EMG muscle pattern for "open" and difficulty maintaining a "close" signal for EMG.

To implement this control, we use data from the thumb pressure sensor. As with bend data, the raw signal is first passed through a moving average filter with window size 0.25 s; we then compute the derivative of the output $\frac{\partial p}{\partial t}$. Motor commands are sent as follows:

- When the orthosis is in the *Device=Closed* position (tendon extended allowing fingers to flex) and the EMG classifier predicts *Intent=Open*, the device is commanded to open (retract the tendon).

- When the orthosis is the *Device=Open* position (tendon fully retracted, or motor stalled) and $\frac{\partial p}{\partial t}$ exceeds threshold $L^P$, the device is commanded to close (extend the tendon).

- If neither of the above conditions are met, we continue to send the previous motor command to the device.

The threshold $L^P$ is set with a procedure similar to the one previously described for the bend threshold $L^B$ this time using training data while the subject is being asked to try to close. Again, we do not issue new commands while the device is transitioning between states.

### 5.3.1 Training with the Exotendon Device

Stroke subjects often produce EMG patterns which change dramatically depending on arm position, even if the subject's intention to open, relax or close the hand remains the same. These EMG patterns are further changed by the hand's physical interaction with the exotendon device. We therefore train the subjects with their arms in different positions and the exotendon device in different states.

We design our training protocol as follows: the exotendon device starts in the closed state (tendon is fully extended) and the subject is asked to relax. Then the subject is asked to try to open their hand. The experimenter waits three seconds, and as the user continues to try to open, the experimenter opens the exotendon device (retracts the tendon) to extend the subject's fingers. The subject continues to try to open for three seconds after the exotendon device is fully opened and is then relaxes. Next, the subject is instructed to close their hand. The experimenter waits three seconds and then closes the device. The subject continues to try and close for three seconds after

the device has fully closed and then is instructed to relax. During training, subject intent, or ground truth, is given to the program by the experimenter as they simultaneously provide participants with verbal commands.

The subject repeats the above procedure five times. The first two times, the subject's arm rests on the table, and the next three times, the subject raises their arm off the table.

## 5.4 Experiments

We evaluate the feasibility of our multimodal controllers when used by subjects with different impairment patterns, using EMG control as a baseline. We selected patients whose EMG patterns showed signs of being abnormal, affected by fatigue and interaction with the orthosis (which, in our experience, is commonplace), but who were still able to complete pick and place tasks using EMG control. Our multimodal control is designed to be robust to different impairments, so we chose subjects with distinct patterns.

We note that, in this current version of the study, the experimenter plays the important role of selecting the appropriate control mode for a patient. We believe this approach serves to establish the feasibility of multimodal sensing in our context, but is also applicable to real-life scenarios, where an experienced clinician can make similar decisions based on patient observations. Nevertheless, we hope to automate this aspect of the procedure in future work.

Testing was performed on four chronic subjects with a spasticity level of two or less on the MAS. Subject clinical information can be found in Table 5.1. Participants had prior experience with the exotendon device, in varying capacities. Subjects gave informed consent and all testing was approved by the Columbia University Internal Review Board, and performed in a clinical setting under the supervision of an occupational therapist.

The experimenter placed the orthosis on the subject's hand and made any necessary sizing adjustments. The subjects were trained using the protocol described in Section 5.3.1.

After training, we asked the subject to perform two types of testing. The first one, designed to isolate the effects of the chosen control method, consists exclusively of performing open-close

53

| Subject | Gender | Affected Limb | Fugl Meyer UE | Box and Blocks |
|---------|--------|---------------|---------------|----------------|
| A | F | Left | 26 | 0 |
| B | F | Right | 26 | 0 |
| C | M | Right | 25 | 6 |
| D | M | Right | 23 | 0 |

Table 5.1: Subject clinical information

hand motions. We refer to these as *Controller Accuracy* experiments. In these tests, we asked the subject to perform several open and close motions in order to compare the accuracy of the baseline and proposed controls. The experimenter verbally cued the subjects to open and close their hand while providing the program with ground truth for the desired motor command.

The second type of test is designed to verify that the multimodal sensory platform we have developed can be used in a functional context. We refer to these as *Pick and Place* experiments. Here, five blocks (1" square cubes) were placed in a square pan on a table in front of the subject. The subject was required to start with their hand in a relaxed state, grasp a block, transport it over the median with control and release it onto the tabletop. The task was considered complete when the subject activated the device to extend the digits and released the block. The therapist timed how long it took the subject to pick and place each of the five blocks. For each condition, the subject moved all five blocks three times. Patients were given sufficient time between testing procedures such that order effects which might have been induced by fatigue were negligible. Each subject was given three minutes of play time to acclimate to each control.

While we designed our pick and place task to minimize the impact of external factors on performance, the nature of functional tasks renders them replete with factors that impact performance. Even such a simple task reflects an individual's shoulder strength, residual fingertip sensation, and grip strength and is not a pure measure of controller efficacy. The number of clinical tests needed to average out the significant effects of all of these compounding factors is beyond the scope of this paper. We therefore rely on Controller Accuracy to evaluate the proposed controls isolated from other factors, and use Pick and Place experiments simply to illustrate their feasibility in a functional context. For this reason, all subjects completed Controller Accuracy testing, but only

| Condition | Control Type | Global Accuracy | PPV | NPV | Transitions Correct | False |
|---|---|---|---|---|---|---|
| Regular | EMG | 77.9% | 77.1% | 78.8% | 4/7 | **0.5** |
| | Multimodal | **83.4%** | **81.6%** | **84.9%** | **6/7** | 1.5 |
| With arm support | EMG | 85.2% | 85.6% | 84.8% | 8/9 | 2 |
| | Multimodal | **86.0%** | **86.6%** | **85.6%** | 8/9 | **0** |

Table 5.2: Results for controller accuracy

Subjects A and B completed Pick and Place testing.

In stroke subjects, fatigue and abnormal coactivation [20] can cause EMG patterns to change over time. To study these effects, we also asked the subjects showing most pronounced effects of fatigue and abnormal co-activation (subjects B and D, as observed by the experimenter) to perform all experiments a second time, in a different condition: wearing an arm support system which aids arm movement through gravity compensation. Testing with and without the arm support system helps us evaluate when the multimodal control is most effective.

During testing, subjects were unaware of the control mode they were using. The controls should be intuitive, so subjects were merely instructed to try to open and close their hand.

## 5.5 Results

For Controller Accuracy testing, we compare the output of the EMG and multimodal controls to the ground truth provided by the experimenter. At each time point, the controls can correctly predict an open (true positive), correctly predict a close (true negative), incorrectly predict an open (false positive) or incorrectly predict a close (false negative). We report the global accuracy, the positive predictive value (PPV), and the negative predictive value (NPV) for our classifiers [66]. Global accuracy is the number of true predictions (positive or negative), divided by the number of total predictions. PPV is the number of true positives divided by the number of all positive predictions (whether true or false). NPV is the number of true negatives divided by the number of all negative predictions.

Global accuracy can be misleading for EMG pattern recognition controls [66], so we believe

| Condition | Control Type | Each Block | Total |
|---|---|---|---|
| Regular | EMG | **13.4±1.4** | **66.8±5.6** |
| | Multimodal | 15.7±2.8 | 78.4±8.4 |
| With arm support | EMG | 30.1±6.5 | 150.7±54.6 |
| | Multimodal | **24.3±3.8** | **121.5±32.4** |

Table 5.3: Results for pick and place tasks

another important metric is the ability to correctly identify transitions between motor commands. A transition is defined as a change in motor command, and a correctly identified transition means a predicted transition which occurs within 1.5 seconds of the ground truth transition. 1.5 seconds allows enough time for the experimenter to give the verbal command and for the subject to react and start performing the motion. (We found that the subjects would often raise their arm off the table before they attempted the instructed hand motion, which increased reaction time.) The correct transitions are reported with the total number of ground truth transitions. Success for this metric is a number of correctly identified transitions that is close or equal to the total number of ground truth transitions. We also report the number of false transitions, or transitions which do not have a corresponding ground truth transition. These transitions cause motor oscillations, confusing and frustrating the user. Success for this metric is a number of false transitions close to zero.

The results for the Controller Accuracy experiments are shown in Table 5.2. We averaged across subjects and show results for experiments performed with arm support (Subjects B and D) and without arm support (all participants). For the Pick and Place testing, we report the time to pick each block and the total time to pick all five blocks, averaged across three trials, and standard error (Table 5.3). We show results for subjects with arm support (Subject B) and the average result without arm support (Subjects A and B).

## 5.6 Summary

In this chapter, we incorporated EMG, bend, and pressure sensors into an exotendon framework to create a multimodal sensing and interaction platform for a hand orthosis. We believe the future of robust controls for orthoses involves multiple sensing modalities which complement each other

to inform controls. The bend and pressure sensors give us information about user intent if subjects display certain impairment patterns we have observed in many stroke patients.

We proposed two multimodal control modes, tailored to the different impairment patterns we have observed. Controls that can cope with many impairment patterns are necessary because these patterns vary across subjects; one patient could even display several patterns as they undergo therapy post-stroke. This is a preliminary study with a limited sample size; however, our results showed that *multimodal controls can be adapted to different impairment patterns and can help functional tasks.* This is the first step towards the development of robust, flexible controls, which could play an important part in deploying robotic rehabilitation to a large population of stroke patients.

# Chapter 6: Rehabilitative and Assistive Performance via Long-term Training

Our studies described in the previous chapters have established the basic operation principles of the device: in limited case series with stroke survivors, we have shown that our exotendon network can facilitate finger extension to enable gross grasp/release (Chapter 3) via a small, wearable motor relying on effective force transmission mechanisms (Chapter 4), and that, for a subset of patients, we can infer the intent to open and close the hand when the orthosis is used in conjunction with a commodity EMG armband [60]. However, clinical performance of this device and the importance of training effects over longer-term use have yet to be investigated. This is an important step for the device because not only does it have to prove its robustness and safety for a long term use, but we can also learn capabilities and limitations of the device as a rehabilitative and assistive tool in a functional context.

In this chapter, we present clinical outcomes of chronic stroke patients using our hand orthosis for a monthly training. 11 chronic stroke patients with moderate muscle tone (MAS $\leq$ 2 in upper extremity) participated in a 12 training program, comprising 3 sessions per week for 4 weeks, and each subject practiced a variety of grasp and release tasks with every day object to simulate ADLs. We evaluated their performance with a battery of clinical assessments pre- and post-intervention FM for upper extremity, ARAT, and BBT.

In order to determine the efficacy of the orthosis as a *rehabilitative device*, we compare clinical outcomes both pre- and post-intervention *without device assistance*. To distinguish between recovery throughout the entire upper limb and more localized improvement of the distal segments, we subdivide FM into shoulder/elbow (FM-proximal) and wrist/hand (FM-distal). To determine
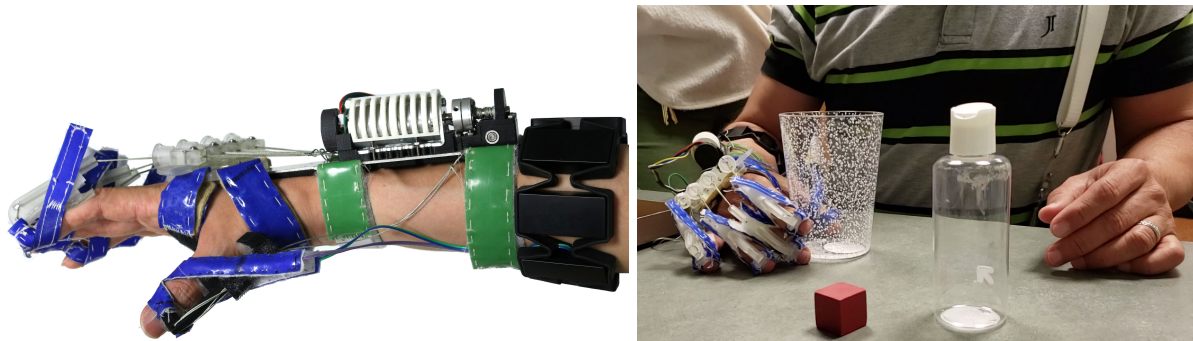
Figure 6.1: Left : Exotendon device and EMG armband used in this study. Note that the device is the same as the one (without bend and pressure sensors) used for the multimodal intent detection study. Right : Functional movement training with a wearable hand orthosis.

efficacy when used as an *assistive device*, we compare baseline performance pre-intervention with post-intervention performance while wearing the device. Our goal is to determine if increased competence using the device from a month-long training program leads to increased performance in tasks requiring grasp, transport, and release of objects while wearing the orthosis. We examine the assistive capability through ARAT and BBT scores measured while the users are wearing the robot. Finally, we compare performance for our two intent inferral methods, namely EMG and shoulder harness controls, used to drive the hand device during both training and post-testing.

## 6.1 Exotendon Device and Intent Detection

As discussed previously, chronic stroke patients with hemiparesis often experience functional disuse of their hand. In this chapter, we focus on the most common impairment patterns, a flexor synergy pattern where individuals may be able to actively flex their fingers to form a gross grasp, but are unable to volitionally extend their fingers to release the grasp. By assisting finger extension, our device enables users to harness their residual function and incorporate their impaired hand into functional grasp and release tasks. We utilize the same exotendon device without bend and pressure sensors introduced in Chapter 5. Note the device specification is listed in Table 6.1. Next, we explain the intent detection methods we used for the protocol.

Our goal for this robotic device is for the patient to initiate robotic assistance by signaling the

| | |
|---|---|
| Weight | 365 g |
| Motor gear ratio | 47:1 |
| Extension/retraction time | 1.8 sec |
| Maximum extension force | 100 N |
| Donning time | 15 mins |
| Doffing time | 1 min |

Table 6.1: Device specification.

intent to open the hand (when the motor retracts, providing assistance for finger extension), or to close (when the motor extends, allowing the fingers to flex). We provide two methods for detecting the intent of the patient, and compare them here.

### 6.1.1 Intent Detection via Ipsilateral EMG Signals

The first method utilizes ipsilateral forearm surface EMG control described in Section 5.3. This approach relies on a pattern recognition algorithm to detect user intention based on data collected by commercial armband (Myo by Thalmic Labs) equipped with eight sensors, and does not require precise sensor placement. The armband is placed approximately one inch proximal to the splint to avoid contact with the motor.

While previous work has shown that intuitive control is indeed possible, it has also highlighted a number of challenges. EMG signals are inherently abnormal in patients with hemiparesis and can be distorted by spasticity and fatigue [12, 64]. As a result, when working with stroke patients while engaged in functional tasks, we found our current EMG-based intent detection method to be effective only for a subset of patients.

### 6.1.2 Intent Detection via Contralateral Shoulder Movement

To account for this phenomenon, we introduce here a second intent inferral method, using contralateral shoulder movement. This approach, often used for body powered upper limb prostheses [94], provides a more robust control compared to EMG, as it relies on the unaffected side. Additionally, compared to other non-EMG control methods, such as a button switch [41], it enables bimanual tasks since the unaffected hand is free. Thus, the focus with this method was on restora-
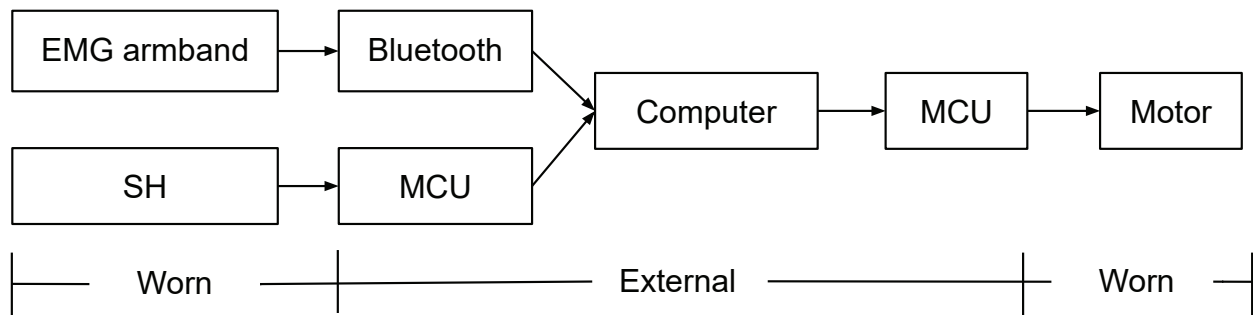
Figure 6.2: EMG armband or SH, depending on the assigned group, send biophysical data to a computer through bluetooth or microcontroller unit (MCU). In the computer, intent detection algorithm classifies the intention and generate a motor command. Then, the command is transmitted to the motor through the MCU.

tion of functionality, rather than neurorecovery. It has the disadvantage of requiring the patient to engage in additional movement (elevating the contralateral shoulder) with the only purpose of providing a signal for our device. Such movement can be unintuitive, and of limited rehabilitative value. However, it may follow that improvements in functional use result in improved proximal strength and bilateral integration.

In our work, a shoulder harness is worn on the unimpaired upper extremity and used to detect shoulder movement. When shoulder depression is detected, the device retracts to trigger hand opening through finger extension (Fig. 6.3). Shoulder flexion was ruled out to control release, as it promoted a flexor synergy in the affected limb, while shoulder depression (often coupled with relaxation/exhalation) appeared more favorable to facilitate release. Conversely, when shoulder elevation (shrug) is detected, the device extends to allow hand closure via finger flexion.

A load cell (Futek, FSH00097) is installed in series with a suspender to measure the tension in the harness, and an extension spring which eases discomfort caused by the tension connects a belt and the suspender. Two different thresholds on the load cell signal are used to detect shoulder elevation and depression in order to prevent unnecessary motor oscillation. The thresholds are manually calibrated at the beginning of each session. In the rest of this study, we will refer to this intent inferral method as SH, shorthand for shoulder harness.

The main advantage of the contralateral SH intent inferral method over EMG is its robustness
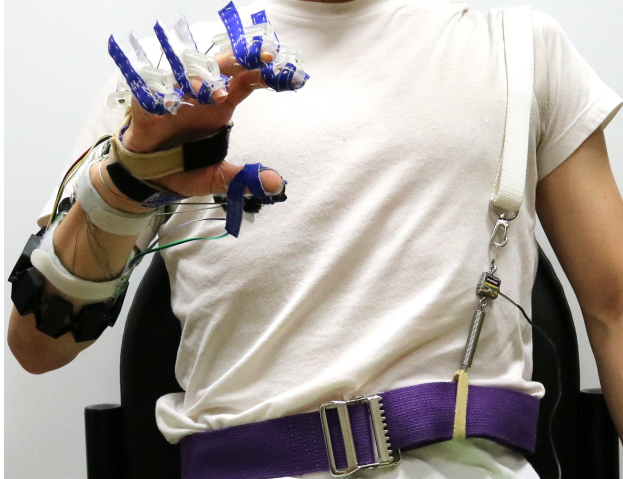
Figure 6.3: Exotendon device with SH method.

Figure 6.4: Real world objects used for treatment.

to differences in impairment patterns, since it relies exclusively on the unimpaired side. Still, we believe that the potential advantages of ipsilateral EMG control (more intuitive motor commands, closing the loop on the affected side) outweigh the SH robustness advantage, as long as EMG control is applicable.

## 6.2 Clinical Intervention

In order to evaluate our active hand orthosis, we performed a clinical study aiming to quantify its performance as either a rehabilitative or assistive device. The three main characteristics of the study included the following. First, each training session consisted of user-controlled functional interaction with everyday objects and simulated ADLs. This is intended to emulate use of the impaired upper limb outside of clinical settings, which is our directional goal for the project. Second, each patient underwent twelve 30 minutes training sessions, distributed over the course of one month. The relatively large number of sessions (compared to our previous feasibility studies) was required both to study rehabilitative effects, and to allow patients to develop familiarity with the device and its controls, in order to quantify performance as an assistive device. Third, our outcome measures post-intervention included clinical assessments performed both assisted (with the device on, in order to study assistive performance) and unassisted (without the device, in order to study

rehabilitation effects). Results were compared to baseline measures without using the device. We present the details of our clinical intervention next.

### 6.2.1 Participants

Total twelve community-dwelling individuals with chronic stroke volunteered to participate in the study and met inclusion criteria. Inclusion criteria were:

- Stroke diagnosis at least 6 months prior to start of study

- Passive range of motion: Wrist to neutral, Digits within normal limits

- Moderate muscle tone, i.e., Modified Ashworth Scale (MAS) $\leq 2$ in digits, wrist, and elbow

- Active range of motion: At least 30 degrees shoulder flexion, 20 degrees shoulder abduction, 20 degrees elbow flexion, finger flexion within functional limits

- Strength: At least trace palpable finger extension

- Able to successfully flex the fingers to form a grasp

- Unable to extend the fingers fully without assistance

- Intact cognition to provide informed written consent

Exclusion criteria were:

- Concurrent participation in another study

- Comorbid orthopedic condition/pain limiting functional use of the impaired upper extremity

- History or neurological disorder other than stroke

- Excessive spasticity (MAS $> 2$)

- Recent botox injection to the affected limb ($< 13$ weeks)

5 participants had prior experience with the exotendon device in varying capacities, but not within 6 months before start of the protocol. All subjects gave informed written consent to participate and the protocol was approved by the Columbia University Medical Center Institutional Review Board. Participants were primarily recruited through a pre-existing, IRB-approved, research registry of stroke patients. Additionally, physiatrists in our clinic referred some of their patients. The trial was registered on ClinicalTrials.gov (NCT03767894). All training and testing sessions were performed under the supervision of an occupational or physical therapist.

### 6.2.2 Outcome Measures

**Baseline Assessments**

All clinical assessments were performed by an occupational therapist who was not involved in the training protocol, though blinding was not possible in this study design. For all testing sessions, the MAS was performed first since other measurement tools can cause fatigue. After the MAS, the FM, ARAT, and BBT, were administered in a randomized order to limit order effect. FM for upper limb is an impairment level measure of body structures that evaluates reflexes, motor function, and joint range of motion of the upper limb [24], ARAT is an activity level assessment that involves specific grasp, grip, pinch, and gross motor tasks for the upper limb [57], and BBT is an activity level assessment that tests unilateral pinch and manual dexterity in a timed manner [59]. The maximum score on the FM for upper limb is 66, and the FM can be subscaled into FM-proximal (42/66), and FM-distal (24/66). Its estimated Minimal Clinically Important Difference (MCID) ranges from 4.25 to 7.25 [67]. The maximum score on the ARAT is 57, and anchor-based MCID of the ARAT for chronic stroke is 5.7 [93].

**Post Assessments**

To evaluate rehabilitative and assistive effects of the training, participants completed post-testing assessments, split over the course of two sessions to avoid fatigue. One session involved administration of the FM, ARAT, and BBT without robotic assistance, while the other session in-

volved administration of the ARAT and BBT with robotic assistance. The order of post-testing days was also randomized. The FM was only performed at post-testing without robotic assistance because the FM assesses capacity of the arm primarily through gross motor tasks, and comparatively few grasping and pinching tasks. We thus presumed that robotic assistance would have minimal influence on FM scores.

Post-testing without the device assesses motor recovery after robot-assisted training whereas post-testing with robotic assistance evaluates the assistive aspects of the device. We assumed that the proposed intent detection methods, particularly EMG-based, will take time for users to learn, so the clinical assessments are performed after 12 training sessions in order to allow competent use of the device.

**Statistical Analysis**

FM and ARAT can be sub-scaled. The FM has two sub-scales: FM-proximal evaluates the shoulder and elbow, while FM-distal evaluates the wrist and hand. The ARAT has 4 sub-scales: grasp, grip, pinch, and gross movement. In addition to total scores, we were interested in evaluating grasp components as those are our targeted areas of intervention. To determine statistical significance, we provided p-values for gains between baseline and post-testing scores, in both conditions as applicable.

Our comparisons include baseline performance against post-testing without robotic assistance for rehabilitative effects, and baseline against post-testing with robotic assistance for assistive effects. One might argue that comparisons of baseline and post-testing with robotic assistance are coupling the rehabilitative and assistive effects of the device after 12 training sessions. In this line of reasoning, purely assistive effects would be highlighted by comparing post-testing assisted vs. post-testing unassisted results. However, we consider that the training component needed to get competent of the device is an intrinsic part of its use as an assistive tool. We therefore chose to use baseline against post-testing with robotic assistance results in our discussion of assistive effects. For completeness, however, we also include the comparison of post-testing assisted vs. post-testing

unassisted in our report.

We tested all the clinical outcome data and gains for normal distribution based on the residuals of our dependent variables with Shapiro-Wilk and visually with Q-Q plots. We also tested the homogeneity of variance using the Levene test. FM and its subsections, ARAT, ARAT-grasp, and ARAT-grip passed the normality and homogeneity test, and we applied paired sample t-tests. On the contrary, ARAT-pinch, ARAT-gross movement, and BBT failed the normality test. Therefore, we applied a nonparametric paired sample Wilcoxon test which does not make assumptions of normality or homogeneity of variance. Statistical significance was determined at $p < 0.05$.

Even with multiple number of comparisons we made, the reason why we did not apply analysis of variance (ANOVA) is because we did not consider the outcomes as repeated measures. Rather, we consider that the post-testings with robotic assistance and without assistance were assessed simultaneously and independently even though we had to separate the sessions in order to avoid fatigue effects. A similar approach can often be found in the literature, most directly related to us in a study of a soft wearable hand robot [75]. The authors also included multiple comparisons and applied paired sample t-tests and Wilcoxon tests for non-repeated measures. ANOVA and Friedman tests with corrections were applied only for repeated measures. In addition, we compare between the two intent detection methods based on gains, but this was not the primary objective of this work; we do not draw any general conclusions regarding the suitability of one control method over the other for hand orthoses. Given the exploratory nature of this pilot study, we chose to report the results here without Bonferroni correction applied to the statistical significance thresholds for non-repeated measures, an approach widely supported in the literature [3, 69, 86].

### 6.2.3 Protocol

After participating in the informed consent process, all participants were screened for inclusion. Those who were included then performed baseline measurements. During the next screening visit, each participant was fitted with the exotendon device to avoid hyper-extension and was screened with the EMG classifier to determine which control method they would use for the duration of the

study. During the control screening, a classifier was trained with the subject's EMG signals, and the user was instructed on how to use the control. The user was asked to open, relax, and close their hand three times each, with the forearm on and off the table. If our EMG method was able to classify the user intention correctly under all conditions, and the user could maintain each signal for at least 2 seconds on every attempts, the user was assigned to the EMG group. Otherwise, the participant was assigned to the SH group.

We allowed the use of a mobile arm support (Saebo MAS) for participants who were clinically observed to have significant difficulty performing the training protocol even with frequent rest. Criteria for use of the arm support included weakness (2-/5 to 3-/5 muscle grades for elbow flexion and shoulder flexion, abduction, and rotation) and significant fatigue limiting functional performance as observed by the therapist. The subjects who met the criteria used the arm support for the duration of training at a set level of support. The level of support was customized for each subject by the therapist in order to optimize their ability to perform functional tasks and limit the impact of shoulder fatigue on grasp training.

Each participant completed 12 training sessions, three times per week for four weeks. Each training session was between 60-90 minutes including time for set up, system classification, donning/doffing the device, and rest breaks as needed. Participants completed 30 minutes of active training during each visit. After the 12 sessions, participants completed two days of post-testing as described above as well as a questionnaire for subjective feedback on their experience.

### 6.2.4 Training

The series of selected tasks reflect best-practice in upper limb prosthetic training (controls training, repetitive drills, and bimanual functional skills training) [38]. During controls training, participants were educated on the device operation and demonstrated proficiency in doing so absent any objects or functional tasks. Participants then advanced to repetitive drills, which incorporated an array of objects of various shapes, sizes, and densities (Fig. 6.4).

Before donning the device, the therapist performed 5 minutes of passive range of motion to all

joints of the upper limb to help mitigate fluctuations in tone across sessions. Then, the patient wore the device. During each session, participants completed 30 minutes of active training. Occasionally, breaks were provided if a patient requested due to fatigue or if any technical issues arose with the prototype device that required an adjustment be made during the session.

The training protocol was always carried out in the same order, with basic tasks first, progressing to more complex tasks. We describe the list of tasks performed by subjects during each session next.

The following tasks were repeated 5 times with the participant's forearm resting on the table top (supported reach) and then 5 additional repetitions with their forearm lifted off the table (unsupported reach) to simulate functional reach. Participants were asked to pick up, briefly hold, and release the following objects:

- 2.5 cm wooden cube

- 5 cm wooden cube

- tennis ball

- 4 cm diameter toiletry bottle

- 13 cm tall, tapered, hard plastic cup

These 5 items were positioned in various locations to optimize the functional envelope. The items were then arranged on a tray with 1 inch raised lip and participants removed all 5 items from the tray twice, and then replaced all items onto the tray twice.

Next, participants picked up and released the following irregularly shaped objects twice each:

- cotton ball

- 1 inch rubber ball

- washcloth

Lastly, participants advanced to bimanual functional skills training to best simulate real-life conditions and the additional physical and cognitive challenge of operating the device without exclusive attention to its performance. The tasks were completed twice each with participants using their affected hand to stabilize and their unaffected hand to perform manipulation:

- removing and replacing the cap from a broad line marker

- removing and replacing the cap from a standard tube of toothpaste (screw off)

- removing and replacing the cap from a screw top beverage bottle

- removing and replacing the wide mouth screw cap from large coffee container

- using a wooden spoon to stir in a small bowl for 10 seconds (affected hand stabilized bowl)

- using a butter knife to make 2 cuts in a 'log' of theraputty to mimic cutting food (affected hand stabilized theraputty)

- opening a lock with a key (affected hand held lock)

- opening a sealed sandwich-size ziploc bag.

Some participants were not able to complete the full training protocol during each session. In that case, they stopped after 30 minutes of training and the last completed task was recorded. Some participants would complete the full protocol in less than 30 minutes. In that case, they continued working on grasp, transport, and release tasks of their choosing (to be client-centered) with oversight from the therapist for the duration of the 30 minutes.

## 6.3 Results

Among 12 enrolled individuals, 11 subjects (6 males and 5 females) completed the training and evaluations. One participant dropped out prior to the first training session due to a medical issue unrelated to the study, therefore all analysis is of 11 subjects. Years since stroke ranged from 2 to 22 years. 9 patients had ischemic, 1 had hemorrogic, and 1 had unknown type of stroke. There were 4 left affected and 7 right affected patients. 10 patients were right hand dominants and 1 was left hand dominant. 6 participants were screened to use EMG method and 5 subjects were assigned in SH group. Note that the subjects screened to the EMG group tended to have more functional use of their impaired upper limb at baseline, as noted by pre-testing scores compared to those assigned to the SH group (Table 6.2). We examine the difference in clinical outcome between the

69

| Assessment | EMG group (n=6) | SH group (n=5) | All subjects (n=11) |
|---|---|---|---|
| FM-Distal | 4.5 | 2 | 3.4 |
| FM-Proximal | 23.2 | 20.6 | 22 |
| FM-Total | 27.7 | 22.6 | 25.4 |
| ARAT | 16.5 | 9.8 | 13.5 |
| BBT | 7.2 | 0.8 | 4.3 |

Table 6.2: Baseline characteristics

two groups in this section as well. One subject used the arm support for training.

### 6.3.1  Fugl-Meyer Upper Extremity Scale

The complete results for FM are shown in Table 6.3. We note that, at baseline, ten subjects had 'no to poor' upper limb capacity ($<31$) and one subject had 'limited capacity' (between 32 and 47) as defined in [28] on the FM. After the treatment, 8 subjects showed improvement, but 3 subjects did not; individual subject results are shown in Fig. 6.5. Overall, the participants showed positive trends with a mean gain of 2.64 points (t-test, n = 11, p-value = 0.026). EMG group had a mean gain of 2.67 points (t-test, n = 6, p-value = 0.114) and SH group improved by 2.6 points (t-test, n = 5, p-value = 0.076) without statistical significance.

To further understand the implications of the results, we analyze the FM into two subtests, FM-distal and FM-proximal. FM-distal, hand and wrist segments, improved significantly with a mean gain of 2.27 points (t-test, n = 11, p-value = 0.001) while significant improvement was not achieved in FM-proximal (t-test, n = 11, p-value = 0.372), shoulder and elbow segments. 86% of the total mean gain (2.64 points) was attributed from the FM-distal. This is particularly notable given that the overall FM is more heavily weighted proximally, with more items in the FM-proximal (42/66) compared to FM-distal (24/66).

### 6.3.2  Action Research Arm Test

Complete ARAT results are shown in Table 6.4. Positive mean gains of 1.33 points (t-test, n = 6, p-value = 0.111) in EMG group and 1.4 points (t-test, n = 5, p-value = 0.286) in SH group
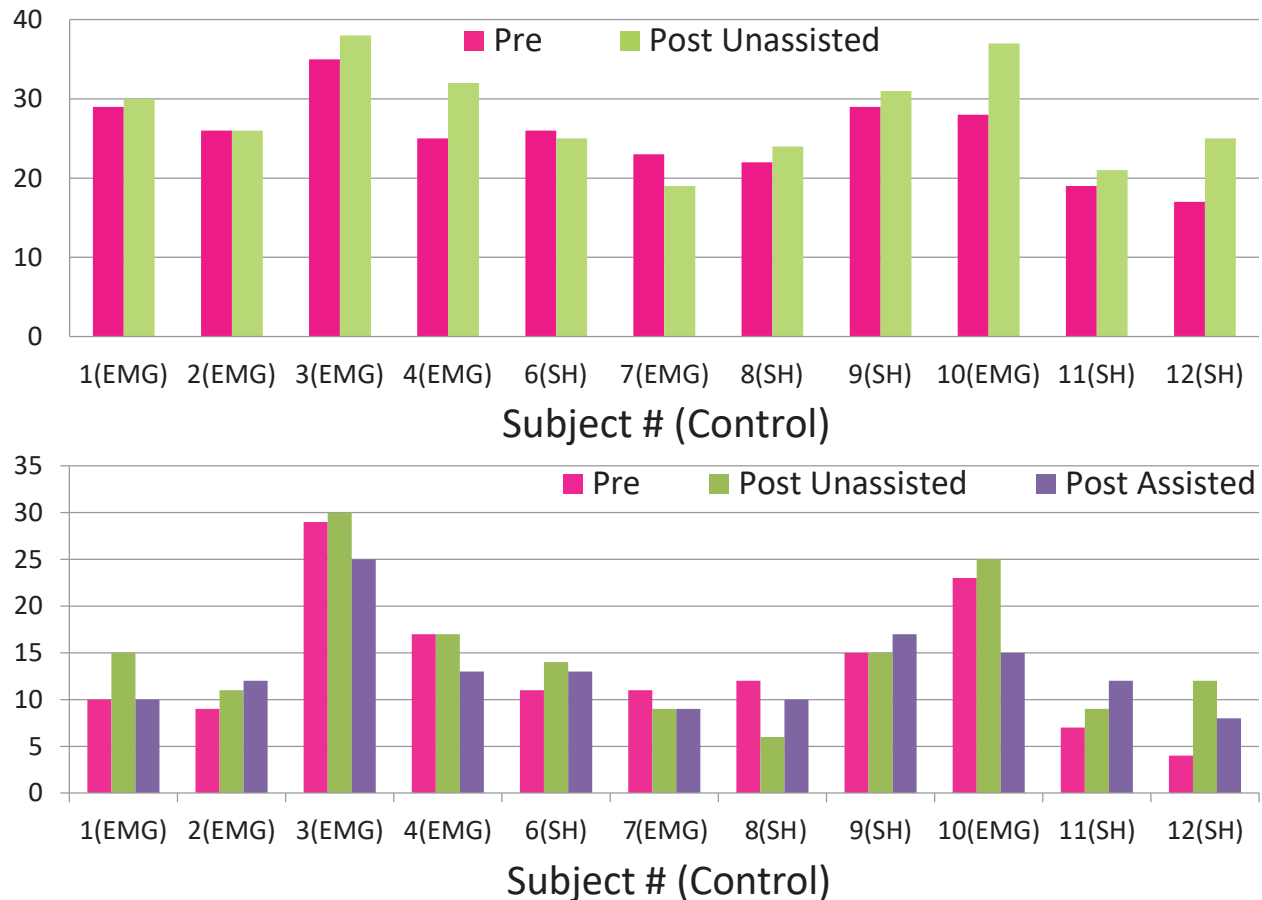
Figure 6.5: FM(left) and ARAT(right) scores. Subject 5 dropped out due to a medical issue unrelated to the study.

were achieved without statistical significance when ARAT was tested without robotic assistance post treatment. For post-testing with robotic assistance, the EMG group performed worse by 2.5 points (t-test, n = 6, p-value = 0.083) whereas the SH group showed mean gain of 2.2 points (t-test, n = 5, p-value = 0.07) (Table 6.4). The most notable improvement in a subcategory was observed in the Grasp category, with mean gain of 2.4 points (t-test, n = 5, p-value = 0.026) in the SH group and 1.17 points (t-test, n = 6, p-value = 0.055) in the EMG group. Mean gains were negative in all other categories (Table 6.4). We also provided ARAT scores post-training with robotic assistance compared to post-training unassisted score in Table 6.4 to examine rehabilitation-assistance coupling effect, but there was no noticeable coupling effect.

| Category | EMG group (n=6) | SH group (n=5) | All subjects (n=11) |
|---|---|---|---|
| Distal | **3 (0.007)** | **1.4 (0.026)** | **2.27 (0.001)** |
| Proximal | -0.33 (0.415) | 1.2 (0.258) | 0.36 (0.372) |
| FM-Total | 2.67 (0.114) | 2.6 (0.076) | **2.64 (0.026)** |

Table 6.3: Mean gains (P-values) without robotic assistance from FM post-intervention. Bold data are statistically significant (P<0.05).

| Category | Robotic Assistance | EMG group (n=6) | SH group (n=5) | All subjects (n=11) |
|---|---|---|---|---|
| Grasp | (A) | 0.67 (0.143) | -0.6 (0.323) | 0.09 (0.444) |
|  | (B) | 1.17 (0.055) | **2.4 (0.026)** | **1.73 (0.004)** |
|  | (C) | 0.5 (0.228) | **3 (0.001)** | **1.64 (0.007)** |
| Grip | (A) | 0.33 (0.182) | 1.4 (0.103) | 0.818 (0.054) |
|  | (B) | **-1.8 (0.029)** | 0.8 (0.169) | -0.64 (0.176) |
|  | (C) | **-2.17 (0.011)** | -0.6 (0.104) | **-1.46 (0.001)** |
| Pinch | (A) | -0.17 (0.564) | 0.4 (0.317) | 0.09 (0.705) |
|  | (B) | -1.83 (0.068) | 0.2 (0.317) | -0.91 (0.104) |
|  | (C) | -1.67 (0.102) | -0.2 (0.655) | -1 (0.077) |
| Gross Movement | (A) | 0.5 (0.317) | 0.2 (0.655) | 0.37 (0.271) |
|  | (B) | 0 (1) | -1.2 (0.109) | -0.55 (0.161) |
|  | (C) | -0.5 (0.45) | -1.4 (0.102) | -0.91 (0.079) |
| Total ARAT | (A) | 1.33 (0.111) | 1.4 (0.286) | 1.36 (0.12) |
|  | (B) | -2.5 (0.083) | 2.2 (0.07) | -0.36 (0.385) |
|  | (C) | **-3.83 (0.032)** | 0.8 (0.307) | -1.72 (0.103) |

Table 6.4: Mean gains (P-values) from ARAT post-intervention. Bold data are statistically significant (P<0.05). (A):Post-training unassisted compared to pre-training, (B):Post-training assisted compared to pre-training, (C):Post-training assisted compared to post-training unassisted.

### 6.3.3 Box and Block Test

BBT results are shown in Table 6.5. At baseline, 7 participants scored 0 point, while 4 subjects scored between 4 to 23 points. The 7 participants without any baseline function were able to improve with mean gains of 2 points (Wilcoxon, n = 7, p-value = 0.026) when assisted by the robot, but more functional subjects performed worse both with and without the device. For the subjects without the ability to pick up any blocks at baseline, the robot served as an assistive device to improve function. BBT scores decreased in the EMG group both with (-4.3 points, Wilcoxon, n = 6, p-value = 0.4) and without (-1.8 points, Wilcoxon, n = 6, p-value = 0.285) robotic assis-

|  | (A) | (B) | (C) |
|---|---|---|---|
| Non functional (n=7) | 0.29 (0.317) | **2 (0.026)** | 1.71 (0.072) |
| Functional (n=4) | -3.5 (0.141) | -9.25 (0.068) | -5.75 (0.285) |
| EMG group (n=6) | -1.84 (0.285) | -4.34 (0.4) | -2.5 (0.916) |
| SH group (n=5) | -0.2 (0.655) | 0.6 (0.461) | 0.8 (0.414) |

Table 6.5: Mean gains (P-values) from BBT post-intervention. Bold data are statistically significant ($P < 0.05$). (A):Post-training unassisted compared to pre-training, (B):Post-training assisted compared to pre-training, (C):Post-training assisted compared to post-training unassisted.

tance without statistical significance, while SH group did not show significant difference neither in assisted score nor unassisted score. We also provided BBT scores post-training with robotic assistance compared to post-training unassisted score in Table 6.5 to examine rehabilitation-assistance coupling effect, but there was no notable coupling effect.

### 6.3.4  Survey

After post-testing, participants provided qualitative feedback through an open-ended survey. Participants generally reported enjoyment, functional improvements, and a desire to continue using the device. "It encourages me to use my hand more. It gave me the feeling of freedom to use my hand again." "I am able to fold and wring out a washcloth." "If I could, I would wear it at home for most of the day for everything." Participants also offered feedback for device improvements such as reduced wiring, less bulk around finger tips, increased training intensity (time and duration), and actuation of the thumb tendon for powered pinch.

## 6.4  Discussion

Overall, we identified trends in the data that suggest this device might serve two distinct purposes for different subsets of the stroke population - namely as a rehabilitation or assistive device. But, the results also highlight limitations of the device, and point towards possible areas for future improvements.

### 6.4.1   An Active Hand Orthosis for Rehabilitative Effects

From a rehabilitation perspective, we discuss results obtained post-intervention without using the device, and compared to baseline performance. Positive gains noted on the FM suggest that training with the device can serve as a rehabilitative tool to remediate some functional use in the affected upper limb, especially for participants with some degree of baseline functionality. The gains were largely attained in the FM-distal subtest, indicating that our functional robotic treatment was beneficial in improving hand functionality as intended. The magnitude of gains on the FM-distal was larger in the EMG group than the SH group, suggesting that the restorative training effect may be greater in participants with some residual baseline functioning.

Based on the observation of a positive trend on the ARAT, we posit that increasing the intensity and duration of the intervention in future studies may lead to increased gains quantified using this measure. For example, small gains captured on the FM (e.g. ability to actively flex or extend the fingers) may not be captured on the ARAT because the improvement in range of motion was not sufficient to translate into increased functional ability (e.g. ability to pick up a small object).

### 6.4.2   An Active Hand Orthosis for Assistive Effects

We focus here on performance measured post-intervention with the participants actively using the device, and compare against baseline. ARAT and BBT results suggest that for participants with minimal to no upper limb baseline capacity, using the robot as an assistive device for long term compensation to increase functionality in daily life may be feasible.

In the case of ARAT results, the improvement in Grasp category from both the EMG and SH groups was expected as the device specifically assists with grasping tasks. We speculate that the differences between groups is related to baseline differences in hand functioning. EMG group participants tended to have more residual functioning at baseline, often employing compensatory patterns to achieve grasp and/or pinch, whereas when wearing the robotic device, the bulky finger components may have impaired their performance by making it more difficult to pick up and place objects in tight spaces. In contrast, those in the SH group had less functionality at baseline, and

therefore the device assisted their ability to pick up objects, though they still had similar difficulty placing objects in tight spaces.

The most encouraging improvement in BBT was observed in participants with no functionality at baseline. We speculate that the negative gain by participants with non-zero baseline functionality occurred because these participants were employing functional compensatory pinch patterns at baseline. Based on our observations, however, training with the device discouraged compensatory patterns and forced users to grasp and pinch with typical patterns (e.g. finger pad to pad pinching), leading to poorer performance as participants did not master the new pinch pattern within the timeframe of the study. We believe additional training time may help, however, it is important to understand that some patients may be satisfied with their compensatory patterns if they are able to participate in their meaningful daily tasks.

The outcome measures highlighted a few limitations of the device. Improvement in Grasp category of the ARAT and negative mean gains in all other categories implies that the device facilitated grasping of mid-size objects, but not small objects which require pinching. We speculate this was because the thumb was splinted into a stable opposition position against the other digits. This had the advantage of reducing actuator load, and we found this pose to be effective when grasping mid-size objects. However, a static thumb also made it difficult for subjects to stably hold small objects in a pinching pattern. This behavior likely affected the results in both the ARAT and BBT tests. We discuss this issue more in Chapter 7.

Another limitation was poor performance of the EMG method due to abnormal muscle synergy in unregistered upper limb postures. The BBT score drop in the EMG group with robotic assistance was likely because of unstable EMG classification when the user had to lift the arm higher due to the partition and height of the box. We note that the training sessions contained no action item that involved lifting an object higher than the height of the partition (15.2 cm), thus the participants did not get an opportunity to practice their proximal muscles or learn to control the device while lifting the arm high in our protocol.

### 6.4.3 Limitations

It is important to point out a number of limitations related to study design in this pilot case series. In particular, we did not use a control group consisting of patient receiving treatment of similar intensity and duration, but without the use of robotic assistance. However, we note that meaningful motor recovery with traditional physical therapy for chronic stroke patients with moderate to severe motor impairments is considered rare [21, 39]. In addition, our robotic assistance enabled training tasks that were not possible for most of the participants otherwise, and thus can not be replicated with traditional therapy. Furthermore, our study did not comprise a follow up assessment to observe the durability of gains. We plan to address both of these limitations in the future.

We also note that assignments between the EMG and SH groups was not performed in randomized fashion, but rather based on the ability of our intent inferral algorithm to classify EMG signals. As a result, we noticed systematic differences between the groups, with SH participants generally having lower baseline functionality. This limits our ability to interpret differences in the results obtained by the two groups.

In addition, we note that rehabilitation studies strive to assess progress using outcome measures that are at the participation-level, observing and rating how participants perform tasks in real-life environments. However, the FM measure, considered as the gold-standard in stroke research due to well established psychometric properties and MCID [67], only assesses body structures at the impairment level, focusing on the capacity of the upper limb to move. ARAT and BBT are activity-level assessments that involve observing and scoring participants performing simulated functional tasks that are shorter in duration and more highly scripted compared to ADLs. We believe customized outcome measures that capture higher task variation and allow longer completion times might be better suited for capturing progress when using robotic devices in an assistive fashion.

While it is important for patients to be able to don and doff a wearable device without assistance, we are still far from quantifying this characteristic. With the current prototype, supervision is required as it would be for traditional therapy, but in the future as the device is further refined,

it is hoped that patients will be able to use the device independently at home after initial training with a therapist.

Finally, we did not conduct any structured interviews or ask the users to rate the device using standard usability scales. However, participants were asked to provide open-ended, qualitative feedback about their experience. Furthermore, a trained occupational therapist supervised each training session and thoroughly monitored for pressure, redness, pain, and any other signs of distress throughout the study. There have been no complaints from patients reported in this study. As the system is further refined, we plan to use standard questionnaires, such as System Usability Scale (SUS) or Likert scale in the future.

## 6.5 Summary

In this work, we presented clinical outcomes after 12 training sessions for a study using a robotic hand orthosis. Our device is designed to assist the paretic hand after stroke, focusing primarily on an impairment pattern characterized by difficulty with active finger extension. Two main design goals for our device are wearability and user-driven operation: we use two different methods to infer the intent of the user, and thus to determine when to provide assistance.

Post-intervention FM sub-scores suggest *the grasp exercises using a robotic device helped improve distal movements of upper limb* whereas it did not have a significant impact on proximal segments. This result suggests the possibility for using our orthosis as a rehabilitative device for the hand. Assisted ARAT scores showed that *the device can indeed function in an assistive role for participants with minimal functional use of their hand at baseline.* However, the results should be cautiously interpreted because of the limited sample size and the fact that the outcomes did not meet MCID for either FM or ARAT. We are planning to address this aspect through both improvements to the device and extended training periods in future work.

Our study also underscored limitations of the device. In particular, the device disrupts compensatory grasp patterns developed by stroke survivors, leading to an immediate decrease in functionality when the device is removed. It is likely that the 12 sessions were not long enough to enable

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learning of new grasp patterns for participants. Furthermore, our current design relies on a static, passively splinted thumb, which enables gross grasp but is not suited for pinching smaller objects.

Nevertheless, we believe that this work can highlight the potential and feasibility for wearable and user-driven robotic hand orthoses. Such devices may enable robotic based-hand rehabilitation during daily activities (as opposed to isolated hand exercises with limited upper limb engagement) and over extended periods of time, even in a patient's home environment. Numerous challenges must still be overcome in order to achieve this vision, related to design (compact devices with easier donning/doffing), control (robust yet intuitive intent inferral), and effectiveness (improved functionality in a wider range of metrics). However, if these challenges can be addressed, wearable robotic devices have the potential to greatly extend the use and training of the affected upper extremity after stroke, and help improve the quality of life for a large patient population.

# Chapter 7: A Preliminary Study of Independent Thumb Actuation

The focus of our research on the device described in the previous chapters has been mostly on the power grasp pattern. While it is true that the ability to gross grasp enables the majority of pick and place tasks for ADLs, we hope to achieve the dexterity levels needed to enable varied and useful manipulation. However, dexterity has not been adequately researched although we briefly investigated the feasibility of single-actuator assistance for a fingertip pinch in Chapter 3.

Results of the study on rehabilitative and assistive performance via long-term training in Chapter 6 capture a limitation of our device in this context. It highlights that the device could facilitate grasping of mid-size objects, but not small objects, which require a pinching pattern. We speculate this is because the thumb is splinted in opposition against the other four digits limiting the range of motion (Fig. 7.1). Since stroke patients often cannot control finger joints independently due to involuntary muscle synergy, if a target object is smaller than the gap in the 2nd picture the user would not be able to grasp it. In order to address this problem, we suggest an additional actuation for pinching and evaluate the design with clinical experiments with stroke patients.

For stroke patients, dexterous manipulation implies that users can be involved in more task-oriented training with various objects, not just repetitive training with one object in massed practice sessions in rehabilitation perspective. Also, users would be encouraged to use the affected hand more often as ADLs become easier with versatile hand functions when used as an assistive device. Overall, assistance for dexterous manipulation would fulfill our ultimate goal of building a take-home device for increased motor repetitions.
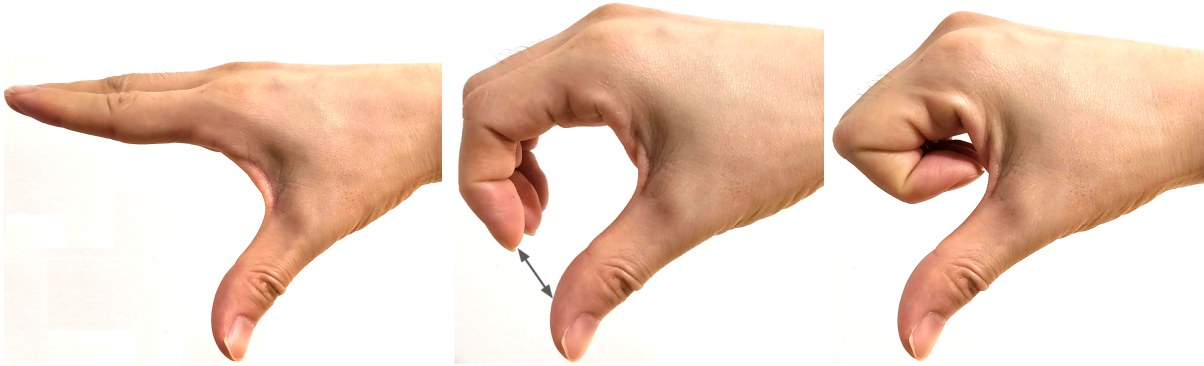
Figure 7.1: Example of a stroke patient's finger motion when using the four fingered exotendon device with the thumb splinted in a fixed position. If the object size is smaller than the gap between the index finger and the thumb (in the 2nd picture), the subject cannot grasp the object.
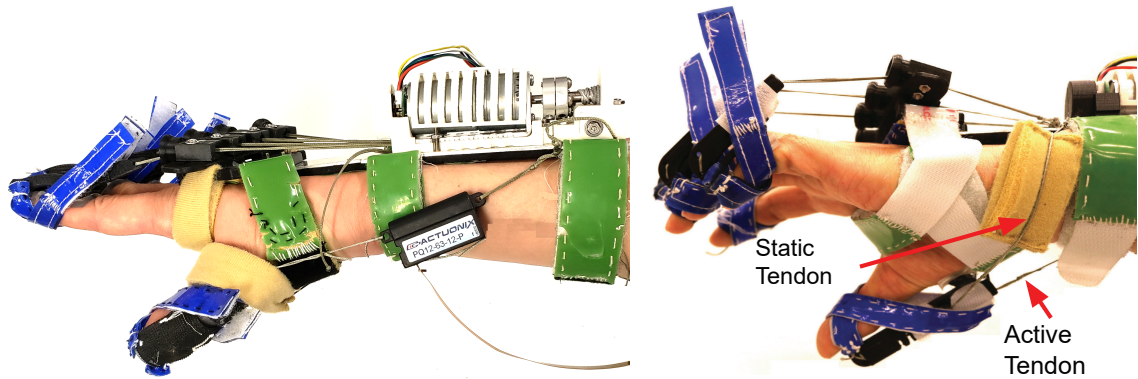


Figure 7.2: Exotendon device with additional actuation for the thumb (left) and tendon routes for the thumb (right).

## 7.1 Designs

Anatomical and kinematic complexity of the thumb movement is high, which makes choice of design for wearable thumb robots difficult. Also, there is no clear consensus about the thumb movement mechanism [88]. Therefore, it is important to understand impairment patterns of the post stroke thumb and focus on a simple, robust, and functional design.

After stroke, the impaired thumb often is flexed inside the palm and other four flexed fingers cover over the thumb [40]. Furthermore, weakness of stroke patients who only can exert 9% of flexion strength also makes ADLs more difficult or impossible [92]. In order to address these impairment patterns commonly found in stroke patients, the assistance should bring the thumb in

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the functional resting posture without limiting the flexion strength of the thumb. In this chapter, we focus on actuation design for opposition-reposition pattern of the thumb that avoids weakening the thumb and enables fine grasp activities.

We utilize two exotendons which are routed in the same manner as previous chapters (Fig. 7.2 - right). We include an actuator (Actuonix PQ12-P, 12V, 63:1) with 20mm of stroke in series with the tendon for extension, which we refer to as "active" tendon here (Fig. 7.2 - left). By actuating the "active" tendon, we are able to approximate opposition-reposition trajectory. To open the hand, the "active" tendon is retracted to assist reposition of the thumb, bringing the thumb back to the resting posture, while the tendon for abduction, which we refer to as "static" tendon here, remains in tension. To close the hand for grasping, the "active" tendon extends while the "static" tendon remains in tension, following opposition trajectory of the thumb. Note this design enables more stable grasp as compared to the previous design since the new design does not limit the grip strength by allowing longer range of motion.

## 7.2 Preliminary Experiments

In order to assess benefits of the second actuation for the hand orthosis, we performed a preliminary clinical study quantifying its assistive performance for dexterous manipulation. Throughout the experiments, we aim to demonstrate the benefits of the thumb routes that reinforce the opposition-reposition trajectory, quantify maximum grip strengths for stable grasp, and evaluate the ability to assist pinching a small object. We note that our experimental plan was disrupted due to external factors, as human subjects work was temporarily paused by the Covid-19 pandemic. As a result, we can only report here our preliminary results, but a more extensive evaluation is planned for the near future.

We recruited two subjects who had prior experience with the exotendon device for the experiments. They had had difficulty with pinching a small object when assisted by the device. The participants had moderate muscle tone (Modified Ashworth Scale $\leq 2$ in upper extremity) and 'no to poor' upper limb capacity. One subject had right sided impairment and another had left

Figure 7.3: Weight measurement setup (left) and torque measurement setup (right).

sided impairment. Testing was approved by the Columbia University Institutional Review Board and took place in a clinical setting under the supervision of licensed physical and/or occupational therapists.

For each subject, we evaluated grasp performance of various object sets under three conditions: without robotic-assistance, with the four fingers assisted device with statically splinted thumb, and with all five fingers actively assisted device. We hypothesize that the subjects would be able to pinch small objects with the bare hand and with the five fingered device because the thumb range of motion is not restricted. Also, participants would be able to grasp large objects better when assisted by the robot regardless of the thumb actuation than without assistance since the subjects cannot achieve functional extension without assistance.

Under each condition, participants were asked to perform two different tasks to assess grip force and versatility. The first task emulated weight variation of a cube (mid-size object) and a sharpening stone (small object) to measure how heavy object a user could lift. The second task emulated bimanual manipulations to measure stabilization force by affected hand while the unaffected hand applied torques to open the cap of a water bottle (large cylinder) and a sharpie (small cylinder). In addition, subject 2 also completed torque measurement using a pill bottle to see how performance would vary for grasping medium cylinder.

For the weight measurement, the objects were connected to a stiff extension spring, which served as a noise filter, and then, to a loadcell, which measured the force (Fig. 7.3 - left). We asked a subject to pick up each object and raise it as high as possible. The higher the subject raised, the more weight the subject could lift with more stable grasp and stronger grip force. As compensatory

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palm grasp, a common pattern for chronic stroke patients, could cloud the result with raising the grip force using muscle contraction, we only allowed pad to pad opposition. If the subject was unable to grasp, lift, and hold the object off the table for at least 5 seconds, the measurement was 0.

For the torque measurement, we used a custom made tool which contained a torsion spring and an encoder (Fig. 7.3 - right). The subject held and stabilized the object with the impaired hand while the unaffected hand twisted the top. The subject could keep on trying during the given time (one minute) for both the weight measurement and torque measurement. Ample rest breaks were provided between the each task.

## 7.3  Results and Discussion

The results are shown in Fig. 7.4. Note that the results with a value of 0 (Subject 1 - baseline performance for cube and sharpening stone tasks and Subject 2 - baseline condition for cube and sharpening stone and static thumb condition) mean the subject was not able to grasp the object.

Both subjects were not able to lift the cube or sharpening stone without robotic assistance in weight measurement experiments. Subject 1 could pick up the sharpening stone while wearing the device, but the thumb actuation enabled higher grip strength and stable grasp. Subject 2 was not able to pick up the sharpening stone at all without the thumb actuation. This suggests the participants outperformed both baseline and splinted thumb conditions in unilateral manipulation tasks using the thumb actuated device.

During torque measurements experiments, the subjects were able to perform equally well or slightly better using robotic assistance compared to baseline condition. However, robotic assisted performances were not evidently better because the subjects could use the unaffected hand to initiate the bimanual tasks as well as used the "saddle area" between the thumb and the index finger instead of pad to pad opposition for bimanual manipulation with a small object (Marker). Therefore, the benefits of the device was washed out, especially for small object grasp.

We also noted limitations of the current assessment design. During the torque measurement,
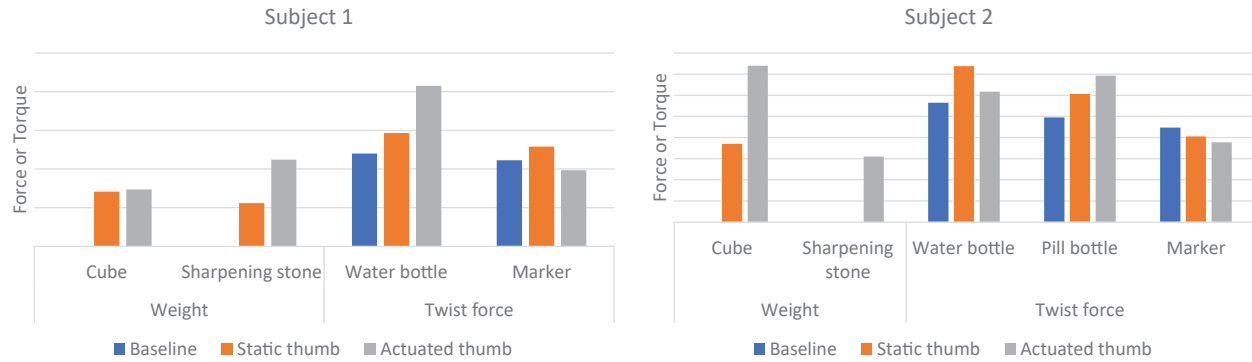
Figure 7.4: Results of maximum weight and torque measurement experiments.



Figure 7.5: Torque measurement tool mounted with a marker. The subjects took advantage of the highlighted portion of the tool to increase stabilization force.

the range of motion was determined by the unaffected hand rather than the grasp stability of the affected hand. Therefore, once the range of motion exceeded a certain angle, the result might be fairly arbitrary. Furthermore, the subjects shoved the affected hand all the way to the highlighted portion of the torque measurement tool in Fig. 7.5 to increase the resistive force. Since this is not a fair measurement, we are planning to verbally instruct and not allow this pattern during the experiment in future study, and disregard data collected when a subject uses this pattern.

This study was a preliminary investigation for future studies with more subjects and a controlled experiment design across all the participants. For more reliable results, we will need more data for each task and proper statistical analysis of it. With more tasks to be performed by subjects, fatigue should be taken into account, and thus the protocol can split over to multiple sessions in which the testing is performed for different conditions (baseline, splinted thumb, and all fingers actuated).

## 7.4  Summary

In this chapter, we proposed an additional thumb actuation design on the existing platform in order to enhance dexterity of user's impaired hand. Preliminary experiments with two stroke patients suggested that independent actuation for the thumb can improve an ability to pinch small object and grasp stability with higher grip strength. While significant additional investigation is needed, we believe that dexterous and stable grasp assistance means the device promotes the use of impaired hand for more tasks with less frustration, subsequently increasing dose of therapy with potentially better outcomes.

# Chapter 8: Conclusions

## 8.1    Summary of Contributions

In this thesis, we have introduced a wearable and functional hand orthosis for chronic stroke patients and evaluated its performance through a number of clinical assessments. Key requirements to make the device operable without medical supervision are wearability and intuitive control methods. Also, functional assistance in ADLs should be effective to better promote movement of the impaired hand. Thus, we focus on wearability, intuitive control methods, and ability to effectively assist in daily activities for technical and clinical contributions.

A key tenet of our approach is that a hand orthosis can provide meaningful assistance with daily manipulation tasks within a small and lightweight package. To implement this principle, we used a network of exotendons or tendons routed on the surface of the hand, used to initiate and assist movement. Using the exotendon device, we measured the actuation force needed to assist a whole hand extension in stroke patients. To the best of ourknowledge, this was the first time that exotendon force needed to overcome spasticity has been measured and reported. Our results also showed that single-actuator devices are able to elicit multiple full-hand movement patterns relevant to functional tasks.

During functional activities, stroke patients can exhibit drastically increased muscle tone after repeated use of their hand and fatigued. To address this problem, we proposed the first transmission mechanisms by which exotendons can overcome hand spasticity for functional tasks with low motor forces and no rigid joints, by increasing moment arm around the finger joints throughout the entire range of motion. We evaluated these mechanisms using mathematical models, artificial fingers, and tests with subjects from our target population, and confirmed the increased ability to

generate assistive join torques, and to assist patients in achieving functional hand extension even when fatigued.

For mechanical hardware to effectively assist in ADLs, the users must be able to operate the device themselves without distracting from the task at hand. Control methods must be intuitive, effective, and robust to long-term operations. This is challenging in a functional context because physical interaction with the impaired hand during complex activities and gives rise to complex, incompletely understood phenomena. In addition, undesired motor activity typically found in stroke patients, such as muscle spasticity, abnormal muscle synergies, and post-stroke dystonias, complicates the reduction in motor control and coordination. To address these challenges, we developed the first multimodal control schemes for a hand orthosis which leverage natural hand movement signals (as opposed to side channels such as voice). Our experiments showed that multimodal control can be adapted to various impairment patterns, and can also be integrated in fully functional systems, laying the foundation for further development in this direction.

The technical advancements described above all build towards our directional goal of a device capable of providing assistance for functional tasks, thus incentivizing long-term use with potential rehabilitative effects. Along this direction, we studied the rehabilitative and assistive performance of the device via long-term training in order to identify its capabilities and limitations. To the best of our knowledge, this was the first time an active wearable hand robot was evaluated in clinical assessments as both an assistive and a rehabilitative device, following user-driven functional hand exercises over multiple training sessions for chronic stroke patients interacting with real world objects. 11 chronic stroke patients with moderate muscle tone participated in 12 training sessions using the orthosis. The subjects were evaluated with a battery of standardized clinical assessments pre- and post-intervention with and without the device. The results suggested the grasp exercises using the device helped improve distal movements of upper limb whereas proximal segments did not show noticeable improvement. Furthermore, the device can indeed function in an assistive role for participants with minimal functional use of their hand at baseline. Finally, based on lessons learned in this study, we introduced an additional actuation design for thumb movement. In a

preliminary study disrupted by external factors (human subjects research limited due to pandemic), we demonstrated that this feature could make grasping more stable and assist in more dexterous manipulation.

Taken together, we believe these improvements have advanced wearable hand robot technology towards a rehabilitative and assistive device that is effective in assisting manipulation component of ADLs outside of clinical settings. Exotendon network and transmission designs allow a small and lightweight platform to assist functional tasks effectively and robustly for users. Multimodal intent inferral provides evidence that customized control for each patient, taking into account different impairment patterns, could be more effective than single modal methods. The study of rehabilitative and assistive performance via long-term training suggested that functional training for robot assisted rehabilitation can enhance distal components of the upper extremity, and the device can also provide functional assistance to patients with severe motor deficit.

## 8.2    Remaining Problems and Future Studies

The work presented here has opened a number of opportunities for immediate, follow-up work on concrete issues. For a start, we would like to finish the thumb actuation study with a complete and rigorous experiment design. The present study introduced in Chapter 7 is not complete and needs more evidence to prove efficacy of the suggested design. We would like to recruit more stroke patients who have difficulties in the thumb mobility and collect more data per each subject.

A problem with the current prototype is that our device does not address proximal components of upper limb, such as forearm pronation/supination, elbow, or shoulder movement. Many stroke subjects present with abnormal and poor control of proximal movement, and therefore they are unable to achieve proper pre-grasp postures. For example, if subjects exhibit abnormal default forearm pronation, they would be disadvantaged from grasping cylindrical objects like cups or bottles properly. In this case, dexterous manipulation capability would be severely affected. For a hand device to be meaningful for wide variety of population, this device would need be integrated with additional proximal assistance. For these additions, we would also have to account for patient

comfort, system complexity, and intent-detection methods as well.

From a usability perspective, this device is not yet ready to be used as a take-home apparatus because it is not equipped with sufficient self-fitting features. Fingertip components in particular would be difficult to don without assistance, particularly more so for patients with severe muscle tone. As a qualitative evaluation, we believe that with some training a subset of patients with low tone could don the device themselves, but this is not a claim we can formally make at the moment. We believe our device is easier to wear than traditional glove-type counterparts, but this aspect still requires further improvement.

Finally, a major challenge for development of wearable hand robot is that the mechanism of spasticity is not completely understood. A well known clinical characteristics of spasticity is increased muscle tone and velocity dependent resistant to movement of joints, but there has not been an attempt to quantify difference between increased muscle tone by passive stretch and muscle contractions by active intent to move. It is important to characterize active muscle tone since a device must operate upon a user's intention to move with muscle activation if we want a user to engage in manipulation tasks. It would be of great benefit to future development of a wearable device if we study relevant spasticity mechanism using a spasticity sensing platform, which measures the interplay between extension/flexion force and joint velocity.

## 8.3   The Future of Post-stroke Hand Rehabilitation

In our study on rehabilitative and assistive performance, the results showed rehabilitative improvements as well as assistive effects to some level. One might wonder whether the future of robotic hand orthoses is as rehabilitative or assistive tools. We believe it is both.

Rehabilitative tools must enable movement and guarantee reliability without a necessary emphasis on portability, whereas assistive tools must be easy to wear and use without medical or technical supervision. It is true that attempting to produce a device that is simultaneously assistive and rehabilitative runs a risk of poor usability; given conflicting requirements, a device might fail to satisfy either. However, the end-goal is very compelling. Assistance in the affected hand for

ADLs could encourage stroke patients to use the hand more on a daily basis with the potential to significantly increase the number of motor repetitions and thus the intensity of training, in turn leading to a better rehabilitative outcome as well. For this reasons, rehabilitative and assistive effects of a wearable device are not clearly separable. A future wearable device that is simultaneously functionally effective, repeatable, wearable, and easy to use, would fulfill this ambitious mission.

What will the next generation of a wearable hand robot look like? Hand, wrist, forearm, elbow, and shoulder assistive devices are all separately developed at the moment. But one can expect to see integration of multiple devices for improved coordinated movement training for manipulation tasks in the foreseeable future. Also, rapid development of machine learning and artificial intelligence will have a great impact on intent detection and intuitive control methods. With related technological growth, such as wearable sensors, actuators, material, transmission mechanisms, and control algorithms, we hope to take a step closer to a wearable and functional take-home device.

With these advancements, stroke rehabilitation can take place with repeatable and extended therapeutic exercises that are distributed in frequent aliquots, enabling functional training on real-world manipulation tasks and activities rather than exercises in a non-functional context. Ultimately, patients with permanent impairment after stroke will develop a greater sense of independence and control within their daily activities.

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