The Sparthan 3D Printed Exo-Glove: a Preliminary Evaluation of Performance via Case Study

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This paper describes the development and testing of a low cost 3D printed wearable hand exoskeleton to assist people with limited finger mobility and grip strength. The function of the presented orthosis is to support and enable light intensity activites of daily living (ADLs), and improve the ability to grasp and hold objects. The Sparthan Exoskeleton prototype utilizes a cable-driven design applied to individual digits with motors. The initial prototype is presented in this paper along with a preliminary evaluation of durability and performance efficacy.

KEYWORDS

Additive manufacturing, 3D Printing, Hand Exoskeleton, Assistive Devices

1 Introduction

The functionality of the hand tends to deteriorate with age and is adversely affected by neurodegenerative diseases and pathologies that impact hand structure [1]. Because of this, there is a significant demographic that suffers from a weakened hand grip and release. Based on findings from the National Center for Health Statistics at the Center for Disease Control (CDC), as many as 18% of adults aged 60 and over have weak or intermediate muscle strength, which is defined as having a maximum hand grip strength of 31.9 kilograms or lower for men, and 19.9 kilograms or lower for women [2]. Four prevalent ways in which the grip strength and extension of the fingers can lose their functionality are peripheral neuropathy, stroke, Amyotrophic Lateral Sclerosis (ALS) that is also known as Lou Gehrig's Disease, and structural degeneration. Of the several primary neurological hand disorders, peripheral neuropathy, or damage in the peripheral nerves, can cause numbness or weakness at tar-

get areas where the peripheral nerves interface with the muscles [3]. Another prevalent cause for loss of functionality in the hand is loss of brain function due to stroke. In fact, more than 795,000 people suffer from stroke every year in the United States, with at least 50% of stroke patients suffering from reduced mobility or hemiparesis [4, 5]. Lou Gehrig's Disease, or ALS, is a neurodegenerative disease that is classified as the progressive degeneration of motor neurons, which leads to muscle atrophy due to a decrease in muscle mass, and ultimately results in muscle weakness. [6].

Decreased grip strength and hand mobility is also prevalent in people who suffer from osteoporosis, or the structural degeneration of bones. Low bone mineral density and deterioration of bone tissue are indicative of osteoporosis and increases the risk of bone fracture [7]. Consequently, grip strength and finger extension can be negatively affected because of osteoporosis. It is estimated (by the International Osteoporosis Foundation) that 44 million people aged 50 and over in the USA suffer from osteoporosis.

2 Background

There are several muscles involved in grip strength and hand extension that could be impaired by neurodegenerative or structurally degenerative diseases. Muscles actuate movement of bones through tendons that cross the joints of the bones. Thus, joints and bone structures impact moment arms, tendon balance, and the gripping and extension of the hand [7], while any muscular degeneration in the hand causes hand movement to deteriorate.

There have been two main approaches for designing exoskeletons in the literature. Historically, the classical approach towards building an exoskeleton glove involves using rigid links and joints to apply the desired forces to the joints of the hand, providing assistance with flexion and extension [8–14]. The major drawback of this style of exoskeleton is that they are generally bulky and heavy, which results in increased user fatigue with extended use. Additionally, because there are many mechanical parts involved in rigid hand exoskeleton designs, the exoskeletons are prone to failure.

A comprehensive review of hand exoskeletons by the School of Engineering in Monash University Malaysia compares 45 different hand exoskeletons. According to the review, there has been a transition from rigid hand exoskeleton designs to soft actuation designs to be able to conform to the natural movement of the fingers. Some of the soft actuation designs are created from plastic, fabric, or a silicone-based material, and are designed based on the patient's specific finger measurements [15].

Many of the soft actuation designs rely on cable actuation to move the hand, with motors to facilitate the cable tension to grip or extend the fingers (e.g., see Fig. 1). The Exo-Glove Poly 2.0, which is a cable tension silicone-based exoskeleton, is at the forefront of the soft hand exoskeleton market, allowing for the user to have control over their grip and extension through motor actuation [16]. However, the materials that are required for the fabrication of the device make the fabrication process and materials procurement generally inaccessible to the common person and non-3D printable, as the silicone is embedded with a thermoplastic to mitigate plastic deformation of the silicone. Also, the actuation system for the Exo-Glove Poly 2.0 is bulky and requires for this system to be mounted to a wheelchair or placed on a table.

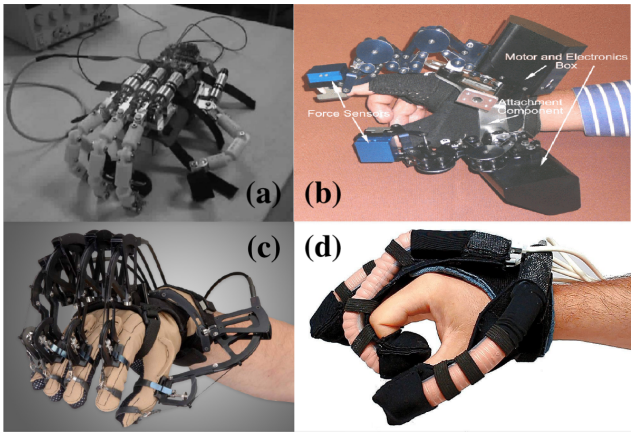

Fig. 1. Previously developed wearable exoskeleton systems for the hand [16–19]

3 The Sparthan Exoskeleton

Assistive devices require customizability to meet the needs of their users. This is particularly true for the hand, which maintains high variability in size and shape of the fingers between individuals [20]. There is a need for an open-source, 3D printable soft exoskeleton solution that is functional, can be easily assembled, and generally accessible. We propose a hand exoskeleton that incorporates the Sparthan module, which is a compact electronics module from a team from the Hong Kong University of Science and Technology that provides motor actuation and intuitive control for the user. The Sparthan Exoskeleton is designed to assist in light intensity activities of daily living (ADL). Examples of such activities include holding onto regularly shaped objects (bottles, cans, book bags, etc.), household maintenance, cooking, eating, dressing, personal hygiene, and opening doors with handles or round knobs. The exoskeleton itself can be 3D printed from NinjaFlex and/or TPS, which are 3D printable thermoplastic elastomer filaments that are flexible, durable, and lightweight.

The flexible exoskeleton consists of three connected ring structures for each finger, or two of these structures for the thumb when necessary. In the case study presented, no thumb actuation was used. Unlike the hinge joints of the fingers, the thumb joint is an example of a saddle joint, and permits the movement in two separate planes. The subject in this study had no control over the movement of their thumb, and currently relies on a stiff brace to constrain their thumb in a strong static position to assist in grasping with the fingers. Therefore, instead of actuating the thumb, the subject used their stiff brace in conjunction with the finger actuators for the tests. This also illustrates the capability for modular-

ity of the Sparthan to be adjusted and optimized for patient specific needs.

In general, the rings are positioned above and below the distal interphalangeal (DIP) and proximal Interphalangeal (PIP) joints for each of the fingers, while the thumb has rings positioned above and below the interphalangeal joint. The rings are connected to each other through multiple bands of NinjaFlex to allow for ease of wearing. Each band is also adjustable to accommodate different finger circumferences. A 3D printed palm piece and Sparthan module mount are also included for the exoskeleton design. Figure 2 depicts the Sparthan module components and the complete exoskeleton. The exoskeleton is actuated by a cable system, which is routed through channels along the dorsal side of the fingers to the Sparthan module. On the palmar side of the exoskeleton, there are also cables that keep the hand at a naturally closed position. The strings are connected to elastic bands, to keep tension in the string and avoid tangling, that route onto a 3D printed PLA plate, which serves as the mount for the elastic bands.

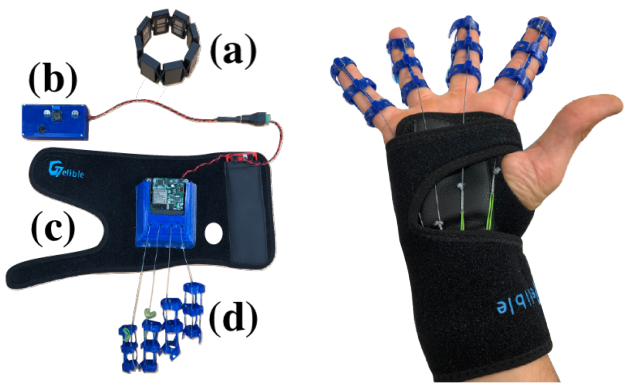

Fig. 2. Myo Armband (a) and Sparthan Exoskeleton components (b-d) (left) and exoskeleton displayed on a hand (right)

3.1 Sparthan Module

Project Sparthan is an initiative aimed at providing makers and engineers with a self-contained, programmable module that is capable of actuating five independent tendons using micro DC brush-less motors. The project was started at the Hong Kong University of Science and Technology (HKUST) and is currently being developed by a joint effort between U.C. Berkeley and HKUST. Developing prosthetic and orthotic devices often requires a multidisciplinary skill-set, including mechanical, electrical and bioengineering components. Project Sparthan aims at lowering barriers to entry in prosthetics and orthotics development by providing user-friendly, EMG-controlled mechatronic modules that can be easily integrated in such projects.

One of the key differentiating features of the Sparthan module consists in its ability to electronically control each finger independently, with incorporated current detection for feedback control. Sensing the current to each individual motor allows the module's micro controller unit to know exactly when the finger is touching the object and the motor is stalling, thanks to the peak in current draw. This peak is limited by the over-current protection and can be used to detect exactly when to stop the motor's motion. The module is rated for a peak power consumption of 25W, which is rarely reached during standard operation. Standard power consumption is around 0.2W during motor resting state and between 2.5W-3W while performing motion, depending on input voltage and object resistance. Each motors current consumption is limited in software to 0.1A while input voltage to the motors can be set by the user (range 9-24V). In this configuration the worst-case scenario peak power consumption is around 12W at 24V. However, the modules current limiting feature can be disabled by the user, up to the hardware limited threshold of 0.2A per motor. With the software limiting disabled the module reaches its maximum rated power consumption of 25W at 24V.

Each motor is also equipped with Hall effect magnetic sensors that act as encoders for the motor's rotation. The exact rotational position of the motors is regulated by a closed-loop, position PID controller. The controller takes as input a target position and subtracts the encoder count to calculate the error. To avoid instability in the output of the PID controller due to integrator windup the integral term is reset when the output of the controller enters a window around the target value [21]. The magnitude of such window is set by multiplying the minimal output value of the controller by a constant (empirically set at 3.6). The minimal output value of the controller is defined as the lower signal that will cause a rotation in the magnetic encoders.

While the module itself provides a EMG classifier with up to five options, our exoskeleton design uses only a binary subsets, which can differentiate between flexion and extension of the wrist. While the hand pose is held, the target position fed to the motor PID controllers is incremented or decremented linearly, based on which gesture was performed. This behaviour allows the user to have fine control over the grip by assessing its position visually.

3.2 Actuation Method for the Sparthan Exoskeleton

The motors were programmed to be actuated by two different methods: The first method is based on the use of the Myo Armband, which is an EMG sensing armband that can control based on different user hand gestures [22]. The Myo Armband is placed on the opposing forearm of the exoskeleton, and then the user can use two different hand gestures to control the counterclockwise or clockwise motion of the motors in the Sparthan module. The Myo armband actuation method is suitable for rehabilitation of hemiparesis patients by assisting the hand with the exoskeleton in gripping and extension. Hence, this exoskeleton can be particularly useful in stroke patients with hemiparesis. However, the Myo armband actuation method may not be ideal for patients with muscle atrophy that affects the wrist, as the Myo armband relies on the wrist to contract forearm muscles. To accommodate for muscle atrophy affecting the wrist, the user could use finger gestures to actuate the

motors instead. The second method to actuate the motors is through flex sensors. Flex sensors measure the amount of bending from an object. By detecting the degree of bending from the flex sensor, the motors actuate to assist the user when gripping or extending their hand.

Sparthan Exoskeleton Exclusion Population

The population that would not be able to use the Sparthan Exoskeleton are patients with Osteoarthritis. Osteoarthritis occurs when there is depletion of cartilage on the ends of the bones, where the joints are located. As a result, the friction between the bones cause for the individual to experience pain and swelling. [23] Since the Sparthan Exoskeleton relies on the movement of joints for gripping and releasing, the Sparthan Exoskeleton is not suitable for people with osteoarthritis, as the implementation of the exoskeleton could cause pain for the user.

3.3 Accessibility

One of the goals for the Sparthan Exoskeleton is to generate more accessible research around orthodic devices by providing a low-cost, reproducible template for building a hand exoskeleton.

The average cost of raw 3D printing materials, excluding the overhead of the printer itself, is estimated to be around $4.55, based on a conservative estimate of the cost of TPU filament (0.11$/g). The exoskeleton can be fully 3D printed, but optional components like a wrist brace can be added to ensure comfort. The standard settings available on Ultimakers Cura software (Open Source) for TPU polymers are very effective for this specific application and do not require extensive modification by novice users. However, in case of repeated failure the most relevant setting to adjust s printing speed. Usually, printing at or below 70% of the default print speed results in a substantial minimization of print failures. Including the cost of these components we estimate that replicating our mechanical design will cost around $15.00.

The Sparthan module (both the modular electronics kit, and 3D printed exoglove components) will be available for purchase as a kit on the Sparthan Github (https://github.com/project-sparthan). However, the designs will also be made open-source for anyone with the means to manufacture themselves. This model of untraditional dispersion of technology is based on, and has been shown effective by the E-nable community [24].

It is acknowledged that 3D printing is not accessible for patients and caregivers directly, but making it open-source will allow for communities such as E-nable to utilize the electronics and the exoskeleton design for those who need them. Basically, tapping into infrastructures with the capability to manufacture and assemble the framework around the electronics (i.e. Open Bionics, E-nable community, and makerspaces across the country) that act as intermediaries to our technology is our strategy for accessibility.

4 Experimental Methods and Results

To evaluate the Sparthan Exoskeleton, several methods were used to assess the performance and durability of the device. For the performance assessment, a 30 year old male user with limited hand mobility tested the device in a series of two tests, a maximum force grip test and a hanging weight grasp test. Since these tests were performed for a single case subject, no Institutional Review Board (IRB) approval was required. However, the IRB was contacted and informed of our study, and confirmed that the subject was enrolled ethically, and given informed consent and media release information prior to testing. For the durability assessment, two tests were performed on the actuation channels of the exoskeleton; an accelerated lifecycle abrasion test, and a load to failure test.

4.1 Load to Failure

A major design concern in this device is the durability and effect of wear at the point of contact with the string. We determined this through visual inspection during preliminary testing of the device in use. We noticed that throughout the fingers range of motion, the string made the most contact with the outer ends of the actuation channels. Therefore, these actuation channels of the exoskeleton (see Figure 3) are expected to experience high friction and forces. An attribute of regular PLA is that it is self lubricating, which makes it good for use in applications that require low friction. However, components printed with softer PLA, TPS and/or Ninjaflex filaments have higher friction, making them more susceptible to wear damage [25]. Both the abrasion and load to failure tests use Spectra Brand braided fishing line, because of its low cost and ease of accessibility for potential users, and test the durability of the actuation channels of the string. Throughout this study, we use 0.4mm braided fishing line rated up to 80lbs. We use this because it is strong, durable, and affordable, and similar line has been used in other tendon cable driven upper-limb assistive devices as well [24].

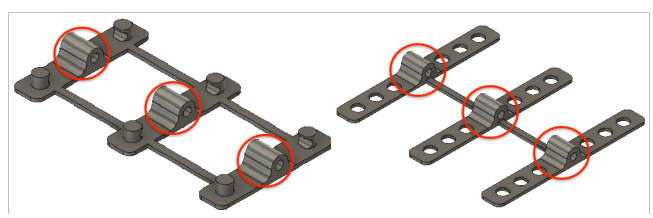

Fig. 3. CAD image of the bottom (left) and top (right) finger rings of the Sparthan Exoskeleton with actuation channels shown in red

Abrasive Wear

A testing mechanism was designed and built to approximate the lifetime of actuation channels due to abrasive wear. An actuation channel was printed on an Ultimaker 3 with a 0.4mm nozzle with TPS flexible filament; the ring was then loaded with 500 grams and supported by a string attached to a servo motor as shown in Figure 4. Physiologically, 500

grams is approximately the weight of a 16 ounce water bottle; a relevant object size and weight that this device is designed to assist in manipulating. In practice however, when holding an object of this size, the load would be distributed throughout the fingers and individual rings. Therefore, a weight of 500 grams directly on a single actuation channel is in excess of what the device would typically undergo, and accordingly testing with this loading will provide a conservative estimate for abrasive wear.

The servo motor actuated a 50 mm diameter wheel cyclically between 0°-180°, moving a string at approximately 100mm/s across the actuation channels. A cycle was considered to be one back and forth motion of the wheel, moving approximately 160mm of string across each channel at an angle of about 60°. This experiment was also conservative in the sense that abrasive wear was simulated at a high angle between the actuation channel and the string; in practice, this angle remains relatively small throughout the motion of the finger, until the very end of the closure of the hand.

Depending on finger size, the amount of string required to actuate each digit with the exoskeleton ranges from about 50 - 100mm. Therefore, each cycle of the abrasion test can be treated as a conservative approximation of the grasping cycle of a hand. The mechanism ran continuously until either string or actuation channel failure occurred, triggering a switch and saving the number of cycles. Failure for the string and failure for the actuation channel were defined as a completely severed string and actuation channel respectively.

During the testing period, which we observed a total of 3 failures, all of which were due to string failure. These occurred after 140,245, 197,128, and 154,720 cycles. The string was replaced after each failure and the test was continued on the same actuation channel and was not replaced. Failure of the actuation channel was not observed after a total of 492,093 cycles, with only mild surface wear and no indication the actuation channel was near failure.

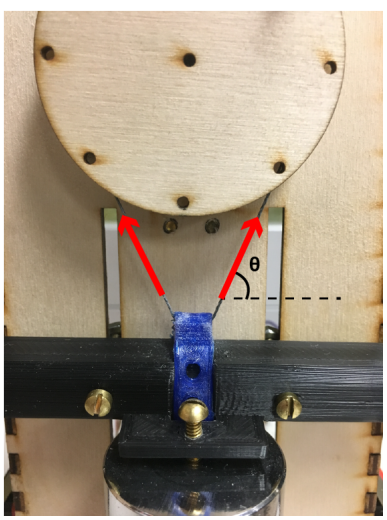

Fig. 4. Experimental abrasion test setup

Load to Failure (Ring)

The purpose of the load to failure test was to approximate the maximum load the actuation channel of the exoskeleton can withstand. To do this, each ring was held down with a clamp on either side, and a braided fishing line was tied in a double knot around the actuation channel. The fishing line was then pulled parallel to the opening of the actuation channel as shown in Figure 5 until failure, and the maximum force was recorded with a CAMRY digital dynamometer. A total of 60 actuation channels were tested.

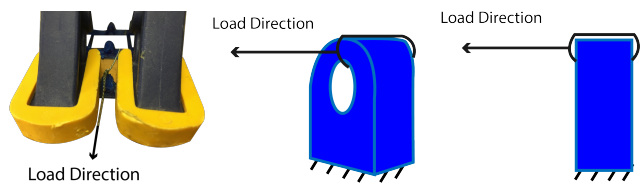

Fig. 5. Image of the experimental setup for the load to failure test of the actuation channels (left) and schematics of isometric (middle) and side views (right) of the actuation channel loading.

The mean failure load of the 60 samples was 7.1 kg with a standard deviation of 1.02. The maximum and minimum loads to failure recorded were 11 and 5.4 kg respectively (see Figure 7).

Two types of failure modes were observed; the most common failure mode was a tear through the actuation channel, and the other was the peeling of the entire actuation channel off of the base layer of the ring (see Figure 6). Of the 60 samples tested, only two samples exhibited layer peeling. These failures occurred at loads of 6.7 and 8.0 kg, both within the standard deviation of the mean.

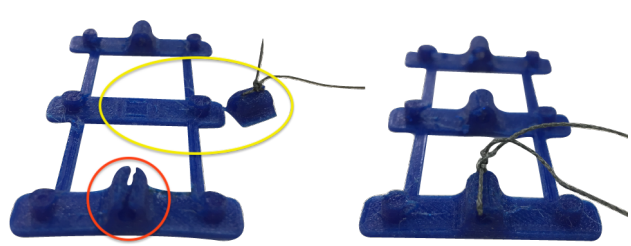

Fig. 6. Example of the two failure modes (left); a tear through actuation channel (shown in red) and peeling of the actuation channel off of the base layer (shown in yellow). And, an example of a double knotted fishing line tied to actuation channels for the load to failure tests (right).

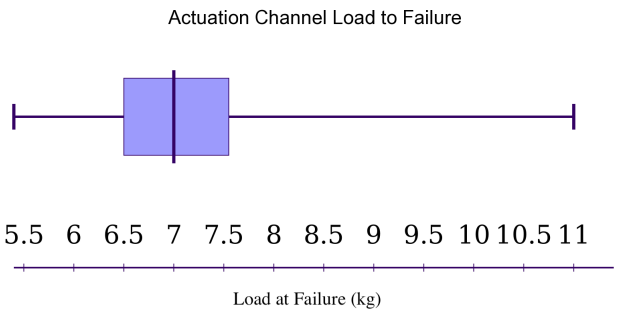

Fig. 7. Results of the load to failure tests for 60 actuation channels

4.2 Preliminary User Testing

To test the capabilities of the exoskeleton, a user with limited hand mobility tested the performance of the device with a maximum grip force and hanging hold test, and also provided qualitative feedback about the comfort and practical use of the device.

Maximum Grip Force

In the maximum grip force test, we used a CAMRY Digital Hand dynamometer to test maximum grip strength of the subject with and without the Sparthan Exoskeleton. The subject, who has maintained a spinal cord injury, has impaired active control of grasp function. However, because of preserved wrist extension, the subject is able to use tenodesis as a replacement of grasp function. During wrist extension, tenodesis causes passive tendon shortening leading to passive finger-to-palm flexion and thumb-to-index lateral face adduction [26]. The maximum grip strength procedure used is outlined in detail below.

Maximum Grip Strength Procedure:

A. Have participant sit comfortably with the shoulder adducted and neutrally rotated, with the elbow towards/against the body and flexed at 90 degrees, and the forearm and wrist initially in a neutral position.

B. Place the hand dynamometer in the participants hand, and let the participant arrange the instrument so that it fits comfortably in the hand. Adjust the handle if necessary for a comfortable grip.

C. Reset the dynamometer by pressing the start button.

D. Request that the participant squeeze with maximum strength. The dynamometer will automatically record the highest force exerted, and display it on the digital screen.

This test was performed 3 times by the participant to obtain a range of values for the maximum force for tenodesis and the exoskeleton. Sufficient resting time (at least one minute of rest) was given between the tests to decrease the effect of fatigue on the results [27]. Results of the grip force test (cf. Fig. 8), as seen in Table 1, show slight improvement in grip strength for the subject, but do not show significant change in maximum grip force strength relative to healthy individuals.

Table 1. Maximum grip force test results with tendonesis and with the Sparthan Exoskeleton along with the approximate healthy baseline grip strength for adult males between the ages of 20-40 from literature. [28–30].

	Tenodesis	Exoskeleton	Healthy Baseline
Max. Force	0.2 - 0.4 kg	1.3 - 1.5 kg	35 - 55 kg

Fig. 8. Grip force measurement using digital hand dynamometer for tenodesis (left) and with the exoskeleton (right)

Hanging Hold

The subject tested the hanging hold strength of the exoskeleton glove with several different shaped handles and weights ranging between 0.5 - 2.5 kg in 0.5 kg increments [31]. The goal of the test was to determine the ability of the exoskeleton in assisting the user to hold a variety of different object shapes. The subject was instructed to hold the bar with their palm facing towards their body. A successful hold was defined as holding the bar securely for at least 10 seconds. The test was repeated for a total of 6 different types of grip (cf. Fig. 9), with 5 different sized grip handles whose dimensions are listed in Table 3. The testing order of the grips was randomized, and a rest period between each test was given to the subject to avoid fatigue. This rest period was typically between 1-3 minutes; the subject was asked between tests if they felt any fatigue, and were given as much time as needed before continuing the test. The results of the grip tests are shown in Table 2.

Table 2. Hanging hold test table for various grip shapes; A check-mark (✓) represents a successful hold, an x-mark (✗) represents a failure, and both represent a hang hold with unstable or questionable success (✓/✗)

Type of Grip	0.5 kg	1.0 kg	1.5 kg	2.0 kg	2.5 kg
(a) Small Cylinder	✓	✓	✓	✓	✓
(b) Large Cylinder	✓	✓/✗	✗	✗	✗
(c) Spherical	✗	✗	✗	✗	✗
(d) Rectangular Bar	✓	✓	✓	✓	✗
(e) Variable (wide near thumb)	✓	✓	✓	✓	✓
(f) Variable (narrow near thumb)	✓	✓/✗	✗	✗	✗

Table 3. List and dimensions of the hand grips

Type of Grip	Width
Small Cylinder	20 mm
Large Cylinder	50 mm
Spherical	60 mm
Rectangular	20 mm
Variable	20-50 mm

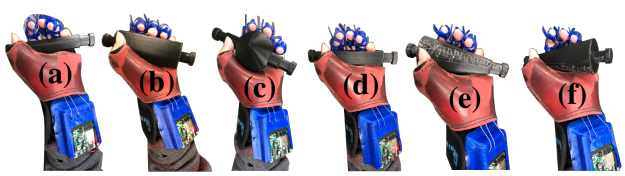

Fig. 9. Various hand grips used in the hanging hold experiment marked (a)-(f) corresponding with the type of grip in Table 2

5 Discussion and Conclusions

In this paper, we presented a 3D printed soft robotic glove that is intended to provide assistance for activities of daily living and rehabilitation for individuals with hand disabilities. The presented work is a preliminary evaluation of performance and a basis for future developments in 3D printable hand exoskeleton designs. The Sparthan Exoskeleton was developed considering the needs of people with hand strength and mobility issues, keeping in mind affordability and customization.

The Sparthan Exoskeleton has features that are designed to be convenient for both the manufacturers and users of the exoskeleton. The finger attachments are designed in such a way that they could be 3D printed in a lay-flat configuration. This enables for the finger attachments to be fabricated quickly with minimal support material, and allows for the manufacturer to 3D print multiple finger attachments simultaneously. The adjustable finger attachments also enable for the user to have a comfortable fit for the individual rings on each finger. The EMG sensors on the Myo Armband enable for the user to control the grip and release of the exoskeleton, and can be configured to have different hand gestures control individual finger movement.

There are aspects of the Sparthan Exoskeleton that need to be improved. Currently, the Sparthan Exoskeleton does not incorporate haptic feedback to enable for the user to experience tactile sensation. For future iterations, the Sparthan Exoskeleton could incorporate vibration sensors to simulate tactile sensation.

The method of using the Myo Armband to actuate the grip and extension of the fingers is, in practice, not a viable method to actuate the exoskeleton. This is mainly because the Myo Armband was discontinued by Thalmic Labs, who are the manufacturers of the armband. However, we are planning to develop a less expensive alternative to the Myo Armband that achieves the same functionality.

The Sparthan Exoskeleton has many separate components that make the exoskeleton difficult to assemble. The strings have to be routed through the finger attachments, and then tied to the motors. In addition, the rubber bands have to be tied to the other end of the strings and then attached to a rubber band mount. Future iterations of the exoskeleton will work to improve and simplify the number of parts that the exoskeleton contains.

Compared to many exoskeletons, such as the ones pictured in Figure 1, which are rigid hand exoskeletons (besides Figure 1d), the Sparthan Exoskeleton is more form-fitting and are less bulky. In addition, the Sparthan module incorporates readily accessible, open-sourced technology that is intuitive for the user to configure gestures to actuate the exoskeleton. However, compared to many of the soft actuation exoskeleton designs, such as the aforementioned Exo-Glove Poly 2.0, the Sparthan Exoskeleton is difficult to put on. A video that demonstrates the donning process of the Exo-Glove Poly 2.0 takes around a minute, while the Sparthan Exoskeleton takes at least two minutes [32]. To improve on the donning process, future iterations of the Sparthan Exoskeleton will be simplified with fewer separate pieces to be quicker and easier to insert the fingers into the attachments.

Preliminary analysis demonstrates the potential of the design to be durable, and resistant to effects of abrasive wear. In a test of 60 actuation channels, they were shown to support an average load of 7.1 kg per channel with a standard deviation of 1.01, and observed maximum and minimum loads of 11.0 and 5.4 kg respectively. Potential cause for this spread in force to failure could be due to a number of factors. One possible cause is slight variations in layer adhesion and layer height of the printed component. The rings were printed 10 at a time on a single Ultimaker 3D printer. One thing we noticed is that there are often slight changes in height from one end of the plate to the other, causing extruded filament to either be compressed more, or less to the previous layer. Care was taken to level the plate prior to printing, however this is a point at which human error may be introduced. Print-

ing the components one by one, in the same location on the plate could ensure less deviation due to differences in topical layer adhesion. However, this would require significantly more time and labor to print the components.

In the wear test, a single actuation channel was able to withstand approximately 500,000 cycles without signs of failure. Unfortunately, the sample size of actuation channel wear test was only one, due to the significant amount of time required to run the abrasive wear test (about a month) on a single actuation channel. This test simply suggests that the longevity of the 3D printed exoskeleton rings is likely sufficient to meet the use requirements of light intensity ADLs, without the need of frequent re-printing.

Based on this case study, our device is still lacking in certain functionality; the maximum grip strength shows little improvement from natural tenodesis grip strength of the subject, and still remains significantly less than normal grip strength. Nevertheless, the added benefit of the exoskeleton is that it allows the user to grasp and hold without any muscle action, potentially reducing user fatigue. The holding pose can also be maintained indefinitely, as it does not require active battery power, allowing users to statically hold onto objects whose weight can far exceed the grasping force generated by the motors.

Results also show that the device was not able to effectively grasp many of the wider shapes in the grip test. Poor performance was primarily due to the gripping area that is lost due to the placement of the routing channels. As observed by the user, the routing channels interfered with the ability for the user to grip these particular objects. This feedback has since been incorporated in the subsequent design iterations. Routing channels will now be placed on the side of the fingers, rather than having a complete ring design as presented in this study.

Future work addresses the addition of thumb actuation for relevant case studies, and programming the actuators to act individually. Presently, we are able to actuate five motors using the module. Therefore, it is possible to make the thumb move with a motor if necessary for unique individual cases. Revisions of the exoskeleton design available on tne Sparthan Github constrains the thumb and allows it to act as an anchor point for grabbing objects. This makes it such that people who do not have control of their thumb could still use this device to grip objects.

This study and further evaluations of the Sparthan exoskeleton would benefit greatly from inclusion of a larger set of participants. Since the target user group is within a vulnerable population, next steps would involve obtaining Institutional Review Board approval for a larger, more generalized study.

References

[1] E. Carmeli, H. Patish, and R. Coleman, "The aging hand," *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, vol. 58, no. 2, pp. M146–M152, 2003.

[2] A. C. Looker, "Prevalence of Reduced Muscle Strength in Older U.S. Adults: United States, 20112012," no. 179, p. 8, 2015.

[3] NIH, "Peripheral neuropathy fact sheet." www.ninds.nih.gov/Disorders/Patient-Caregiver-Education/Fact-Sheets/Peripheral-Neuropathy-Fact-Sheet, 2018. Online: 2018-09-10.

[4] CDC, "Stroke facts." www.cdc.gov/stroke/facts.htm, 2018. Online: 2018-11-10.

[5] N. Scherbakov and W. Doehner, "Sarcopenia in stroke-facts and numbers on muscle loss accounting for disability after stroke," *Journal of Cachexia, Sarcopenia, and Muscle*, vol. 2, pp. 5–8, 2011.

[6] L. Kinsley and T. Siddique, "Amyotrophic lateral sclerosis overview," in *GeneReviews®[Internet]*, University of Washington, Seattle, 2015.

[7] J. A. Kanis, L. J. Melton III, C. Christiansen, C. C. Johnston, and N. Khaltaev, "The diagnosis of osteoporosis," *Journal of bone and mineral research*, vol. 9, no. 8, pp. 1137–1141, 1994.

[8] Z. Ma and P. Ben-Tzvi, "An admittance type haptic device: Rml glove," in *ASME 2011 international mechanical engineering congress and exposition*, pp. 1219–1225, American Society of Mechanical Engineers, 2011.

[9] Z. Ma and P. Ben-Tzvi, "Design and optimization of a five-finger haptic glove mechanism," *Journal of Mechanisms and Robotics*, vol. 7, no. 4, p. 041008, 2015.

[10] P. Ben-Tzvi, J. Danoff, and Z. Ma, "The design evolution of a sensing and force-feedback exoskeleton robotic glove for hand rehabilitation application," *Journal of Mechanisms and Robotics*, vol. 8, no. 5, p. 051019, 2016.

[11] M. Fontana, S. Fabio, S. Marcheschi, and M. Bergamasco, "Haptic hand exoskeleton for precision grasp simulation," *Journal of Mechanisms and Robotics*, vol. 5, no. 4, p. 041014, 2013.

[12] E. Refour, B. Sebastian, and P. Ben-Tzvi, "Design and implementation of an exoskeleton glove for infant medical rehabilitation," in *ASME 2018 International Design Engineering Technical Conferences and Computers and Information in Engineering Conference*, pp. V05AT07A062–V05AT07A062, American Society of Mechanical Engineers, 2018.

[13] E. Refour, B. Sebastian, and P. Ben-Tzvi, "Two-digit robotic exoskeleton glove mechanism: Design and integration," *Journal of Mechanisms and Robotics*, vol. 10, no. 2, p. 025002, 2018.

[14] M. Haghshenas-Jaryani, C. Nothnagle, R. M. Patterson, N. Bugnariu, and M. B. Wijesundara, "Soft robotic rehabilitation exoskeleton (rehab glove) for hand therapy," in *ASME 2017 International Design Engineering Technical Conferences and Computers and Information in Engineering Conference*, pp. V003T13A005–V003T13A005, American Society of Mechanical Engineers, 2017.

[15] T. Shahid, D. Gouwanda, S. Nurzaman, *et al.*, "Moving toward soft robotics: A decade review of the design of

hand exoskeletons," *Biomimetics*, vol. 3, no. 3, p. 17, 2018.

[16] H. Lee, B. B. Kang, H. In, and K.-J. Cho, "Design improvement of a polymer-based tendon-driven wearable robotic hand (exo-glove poly)," in *Wearable Robotics: Challenges and Trends*, pp. 95–99, Springer, 2017.

[17] J. Iqbal, O. Ahmad, and A. Malik, "Hexosys ii-towards realization of light mass robotics for the hand," in *Multitopic Conference (INMIC), 2011 IEEE 14th International*, pp. 115–119, IEEE, 2011.

[18] M. Fontana, A. Dettori, F. Salsedo, and M. Bergamasco, "Mechanical design of a novel hand exoskeleton for accurate force displaying," in *Robotics and Automation, 2009. ICRA'09. IEEE International Conference on*, pp. 1704–1709, IEEE, 2009.

[19] CyberGloveSystems.com, "Cyber glove systems," 2016.

[20] D. Lim, T. Georgiou, A. Bhardwaj, G. D. O'Connell, and A. M. Agogino, "Customization of a 3d printed prosthetic finger using parametric modeling," *Configurations*, vol. 42, no. 80, p. 13, 2018.

[21] K. J. Astrom and L. Rundqwist, "Integrator windup and how to avoid it," in *1989 American Control Conference*, pp. 1693–1698, IEEE, 1989.

[22] M. Sathiyanarayanan and S. Rajan, "Myo armband for physiotherapy healthcare: A case study using gesture recognition application," in *Communication Systems and Networks (COMSNETS), 2016 8th International Conference on*, pp. 1–6, IEEE, 2016.

[23] J. Peter, C. M. Pearson, and L. Marmor, "Erosive osteoarthritis of the hands," *Arthritis & Rheumatism: Official Journal of the American College of Rheumatology*, vol. 9, no. 3, pp. 365–388, 1966.

[24] E-Nable, "Enabling the future," 2018.

[25] J. M. G. Farstad, Ø. Netland, and T. Welo, "Surface friction of rapidly prototyped wheels from 3d-printed thermoplastic elastomers: An experimental study," *Procedia CIRP*, vol. 60, pp. 247–252, 2017.

[26] S. Mateo, P. Revol, M. Fourtassi, Y. Rossetti, C. Collet, and G. Rode, "Kinematic characteristics of tenodesis grasp in c6 quadriplegia," *Spinal cord*, vol. 51, no. 2, p. 144, 2013.

[27] H. C. Roberts, H. J. Denison, H. J. Martin, H. P. Patel, H. Syddall, C. Cooper, and A. A. Sayer, "A review of the measurement of grip strength in clinical and epidemiological studies: towards a standardised approach," *Age and ageing*, vol. 40, no. 4, pp. 423–429, 2011.

[28] E. Innes, "Handgrip strength testing: a review of the literature," *Australian Occupational Therapy Journal*, vol. 46, no. 3, pp. 120–140, 1999.

[29] V. Mathiowetz, K. Weber, G. Volland, and N. Kashman, "Reliability and validity of grip and pinch strength evaluations," *The Journal of hand surgery*, vol. 9, no. 2, pp. 222–226, 1984.

[30] A. B. Swanson, C. Göran-Hagert, and G. de Groot Swanson, "Evaluation of impairment in the upper extremity," *The Journal of hand surgery*, vol. 12, no. 5, pp. 896–926, 1987.

[31] J. W. Harris, *Optimization of Prosthetic Hands: Utilizing Modularity to Improve Grip Force, Grasp, and Versatility*. PhD thesis, UNLV, 2017.

[32] SNU BioRobotics Lab, "Exo-Glove Poly (Extended) (Seoul National University)."