Design and Characterization of a Lightweight and Fully Portable Remote Actuation System for Use With a Hand Exoskeleton

Article *in* IEEE Robotics and Automation Letters · July 2016

DOI: 10.1109/LRA.2016.2528296

Christopher Nycz
Worcester Polytechnic Institute

Tobias Butzer
ETH Zurich

Gregory S Fischer
Worcester Polytechnic Institute

Roger Gassert
ETH Zurich

Design and Characterization of a Lightweight and Fully Portable Remote Actuation System for Use With a Hand Exoskeleton

Christopher J. Nycz, Tobias Bützer, Olivier Lambercy, Jumpei Arata, Gregory S. Fischer, and Roger Gassert

***Abstract*—Enabling individuals who are living with reduced mobility of the hand to utilize portable exoskeletons at home has the potential to deliver rehabilitation therapies with a greater intensity and relevance to activities of daily living. Various hand exoskeleton designs have been explored in the past, however, devices have remained nonportable and cumbersome for the intended users. Here we investigate a remote actuation system for wearable hand exoskeletons, which moves weight from the weakened limb to the shoulders, reducing the burden on the user and improving portability. A push-pull Bowden cable was used to transmit actuator forces from a backpack to the hand with strict attention paid to total system weight, size, and the needs of the target population. We present the design and integration of this system into a previously presented hand exoskeleton, as well as its characterization. Integration of remote actuation reduced the exoskeleton weight by 56% to 113g without adverse effects to functionality. Total actuation system weight was kept to 754g. The loss of positional accuracy inherent with Bowden cable transmissions was compensated for through closed loop positional control of the transmission output. The achieved weight reduction makes hand exoskeletons more suitable to the intended user, which will permit the study of their effectiveness in providing long duration, high intensity, and targeted rehabilitation as well as functional assistance.**

***Index Terms*—Rehabilitation Robotics, Prosthetics and Exoskeletons, Physically Assistive Devices, Medical Robotics and Systems, Bowden Cable Transmission.**

I. Introduction

LOSS of mobility in the hand is a condition faced by many individuals manifested through various mechanisms such as orthopedic or neurological injury, general aging, or chronic disease. The incidence of stroke related disability in particular, with over *665,000* individuals surviving a stroke in the United States each year and *50%* retaining some degree of hemiparesis *6* months post stroke [1], presents a need for expanded rehabilitation and long-term assistance methods. To this end, the use of robotics in rehabilitation has been explored in great depth over the past two decades [2], [3].

Muscle weakness in the arm is present, to varying degrees, in the majority of stroke patients [5], [6], leading to difficulties in lifting the arm against gravity, controlling movements, or grasping and manipulating objects. Much of the initial focus in robotic rehabilitation was on training the proximal joints of the arm, however, improvements in the proximal joints were not seen to generalize to the distal joints [7]. In contrast, studies which have focused on robotic training of the hand and wrist have indicated functional improvements at the distal joints can generalize to the whole arm [8]. Finger muscle weakness, especially in extensor muscles, is further considered as one of the primary causes of hand impairment after stroke [9], [10].

Due to the importance of improving hand function following stroke, a current area of focus has been developing devices to deliver high intensity training of the hand in relevant movements. Many exoskeleton devices have been designed for this purpose with various approaches to the problem [7], [11]. Among these designs there are a few common methods of interacting with the fingers including rigid linkages [12]–[15], soft pneumatic or hydraulic devices [16], [17], and cable actuated gloves [18]–[20].

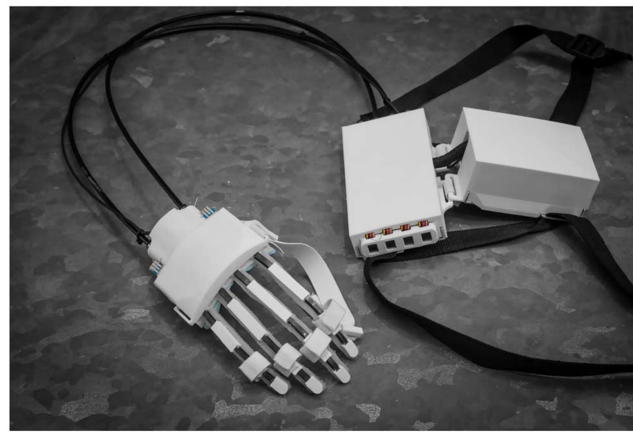

Fig. 1. Proposed remote actuation system (right middle of frame) integrated into Arata et al.'s hand exoskeleton [4] (left forward of frame) connected by 2 push-pull Bowden cables.

Manuscript received August 31, 2015; accepted January 17, 2016. Date of publication February 11, 2016; date of current version March 4, 2016. This paper was recommended for publication by Associate Editor C. Nabeshima and Editor K. Masamune upon evaluation of the reviewers' comments. This work was supported in part by the National Science Foundation under Grant NSF IGERT DGE 1144804 and in part by the Swiss National Science Foundation through the National Center of Competence in Research (NCCR) Robotics. Gregory S. Fischer and Roger Gassert are co-senior authors.

C. J. Nycz is with the Automation and Interventional Medicine (AIM) Laboratory, Worcester Polytechnic Institute, Worcester, MA 01609 USA, and also with the Rehabilitation Engineering Laboratory, ETH Zürich, 8092 Zürich, Switzerland (e-mail: cjnycz@wpi.edu).

T. Bützer, O. Lambercy, and R. Gassert are with the Rehabilitation Engineering Laboratory, ETH Zürich, 8092 Zürich, Switzerland (e-mail: tbuetzer@student.ethz.ch).

J. Arata is with the System Engineering Laboratory, Kyushu University, Fukuoka 819-0395, Japan (e-mail: jumpei@mech.kyushu-u.ac.jp).

G. S. Fischer is with the Automation and Interventional Medicine (AIM) Laboratory, Worcester Polytechnic Institute, Worcester, MA 01609 USA.

Digital Object Identifier 10.1109/LRA.2016.2528296

Most previous exoskeletons have demonstrated sufficient force and correct kinematic alignment with the hand, however, have neglected other design aspects required for daily use. The impracticality of current designs can limit their capability of being tested on stroke populations and has proven a hindrance to evaluating the effectiveness of exoskeletons in rehabilitation. In particular, weight and bulk placed on the hand are important considerations in device design. In addition to weakness, the abnormal synergies observed in elbow and shoulder joints after stroke are known to affect active arm movements and reachable workspace when the load on the arm is increased [21], [22]. This highlights the need for minimizing the weight of components which are mounted directly on the hand and arm of stroke survivors. A few groups have tried to identify acceptable device weights based on anatomical comparisons with the hand and forearm, resulting in claims that the threshold for acceptable weight on the hand is in the region of *400g* to *500g* [23], [24]. These anatomic comparisons add context to a device's weight, however, a definitive value of "light" is mostly patient specific. Considering well-developed devices which report the amount of weight placed on the hand, most tend to be in the range of *250g* to *500g* [4], [16], [23]–[25].

In an effort to improve upon the weight and size limitations of current devices, Arata et al. developed a low profile hand exoskeleton by implementing a novel sliding spring mechanism described in [4]. This specific device was designed to assist stroke patients with muscle weakness in opening and closing the hand, and generating the grasping forces required to manipulate objects in activities of daily living (typically below 20N) [26]. Arata's device was minimally actuated with 2 DC electric motors mounted on the exoskeleton, resulting in a weight of *250g* placed on the hand. The exoskeleton could also easily be donned by a stroke patient, and had a low profile over the fingers of just *6 mm*. While light and compact compared to many existing designs, patients in small scale studies all raised concern over the burden created by having this amount of weight placed at the end of their weakened arm. Discussions with rehabilitation therapists also raised concerns about the weight and where it was placed on the individual. The actuators mounted on Arata's exoskeleton accounted directly for approximately one third of its total weight and as such their removal would greatly reduce the burden on the user.

To limit the weight placed on the weakened hand, several devices incorporate remote actuation, typically through cable-drive, hydraulic, or pneumatic systems [13], [16], [18], [19], [25], [27]. Some remote actuation units are stationary, with long transmission lines to allow sufficient mobility of the exoskeleton for specific tasks [13], [16], [18]. Others have developed portable actuation systems which allow the user to be physically untethered from any external device. These portable devices have enough control electronics and on-board power to execute movement commands for a reasonable time period and also have components of a complete design such as protective covers and mounting hardware. Upon review of these portable devices, typical weight is in the range of *3 kg* to *5 kg* [19], [27], [28]. While portable, reduction in the weight of these systems is needed for practical use by a wider stroke population.

Improvements in portability are needed for exoskeletons to see regular use for in-home therapy and functional assistance. A lightweight remote actuation system which is fully stand-alone and capable of applying enough force and independent degrees of freedom (DOF) for functional grasps has the potential to improve the user acceptance of these devices. Furthermore, making the device modular and easily integrated with various types of exoskeletons would work to make it adaptable to patient needs. To this end, we have developed a remote actuation system with a reduction in size and weight compared to previous efforts. The functionality was characterized and ultimately tested by integrating it with the exoskeleton previously presented by Arata et al. [4].

II. Design Review and Requirements

In general, a remote actuation system which could reliably set a linear position at it's output was desired. The design would have to minimize the drawbacks of remote actuation, namely transmission loss and the reduction of positional accuracy. For a remotely actuated system, direct control of force at the output is challenging due to transmission losses which will change depending on the orientation of the system. Implementation of force control through a series elastic element and positional control of the output would be possible with the proposed architecture. Several designs also use linear actuators for their control input allowing for simple integration with existing devices [4], [18], [29], [30].

A. Actuation Unit Output Force

A maximum transmission output force of between *25N* and *35N* was chosen as a design criteria. This would closely match the input forces required by previous works such as the Gloreha's *12N* [18], Arata et al.'s *35N* [4], and Ho et al.'s *30N* [29].

B. Transmission Type

Remote actuation transmissions utilized in prior work were found to fall within three general types: pneumatic, hydraulic, and cable drive. Backpack mounted hydraulic systems have seen implementation on several lower limb exoskeletons [31] and a belt mounted hydraulic system has been implemented for use with a hand exoskeleton [28]. Pneumatic solutions for hand exoskeletons implementing air pistons or other, more novel actuators, have been explored by several groups although few attempts have been made to develop these into fully portable systems [11], [16], [17], [32]. Pneumatic and hydraulic designs can benefit from energy dense actuators, however, the total system typically requires a multitude of components usually including a pump, reservoir, regulator, valves, hoses, connectors, actuators, and a power source [28], [31], [32]. These components can contribute to a larger size, more weight, and a greater cost.

Cable drive systems have also been commonly implemented on exoskeleton devices [11], [12], [18], [20], [27], [33]. These devices have proven some degree of commercial viability with the Gloreha [18] and CyberGrasp [27]. Bowden cables are desirable due to their inexpensive components, simple construction, and ease of interfacing with DC motors. As a drawback, transmission losses (friction), backlash, and other positional inaccuracies are expected [34], [35].

C. Anatomical Positioning

Actuator placement requires a balance between transmission efficiency and user comfort. Three possible locations were considered and discussed with rehabilitation therapists; these were the upper arm, the back, and the waist.

Mounting on the upper arm offered the most efficient location for power transmission due to its short, arm posture independent, path length and single bend at the elbow. Mounting of actuators on the arm has been implemented in the past [33]. The location, however, places weight on the affected arm, albeit in a place which would reduce the effort required by the user to lift it.

Mounting around the waist was also considered since weight here would be more bearable and, if mounted anteriorly, would not impede sitting. This placement has also been implemented in the past [28]. There were concerns, however, with routing a power transmission between hand and waist as the distance varies greatly throughout the range of motion of the arm. This makes a direct connection potentially obtrusive due to the need for a considerable amount of slack in the transmission when the hand is lowered to prevent constraining raising of the arm. Routing across the shoulder and elbow joints would maintain a relatively constant path length, minimizing slack. However, routing up to the shoulder and back down to the hand would lead to higher transmission losses and potentially a more complicated donning process.

The back represents a compromise between the two previously described locations. The shoulders would feel less burdened by the weight than would the hand or upper arm and the transmission would maintain a fairly constant path length, traversing bends at the shoulder and elbow. This position was chosen as the one to implement. The main concern raised by rehabilitation therapists about mounting here was the user's comfort while seated in a chair. With this feedback, a requirement of making the design flat to the body where possible and mounted high up on the shoulders was set.

D. Actuation System Size, Weight, and Ergonomics

To make the system bearable to a wider stroke population, a target weight of under *1 kg* on the back was set. The amount of weight bearable by an individual will be highly patient specific and a definitive value is difficult to determine. Keeping the design to under *1 kg* however, would represent a substantial improvement over the typical *3 kg* to *5 kg* actuation units of previous designs [18], [28]. Ultimately, patients will need to be selected based on the ability to bear this weight.

The size was also determined to be a critical factor to portability. In general, a slimmer device mounted higher up the back was desired. A goal of limiting the amount which the device protrudes posteriorly to *25 mm* was targeted. Constraining the pack to not extend below the middle of the back (beyond vertebrae T7) was also desired. The shape should also be comfortable when worn, it should conform to the back's shape and not cause pressure points or chafing.

III. Design and Construction

Based on our established criteria, an actuation system was designed and constructed. This system was then integrated into a version of Arata et al.'s hand exoskeleton with the necessary modifications to Arata's original design presented.

A. Power Transmission Design

We explored a push-pull Bowden cable transmission with DC motors as actuators, which have benefits in size, cost, energy storage, and control. This design can deliver tensile and compressive forces allowing for finger flexion and extension, dependent upon the exoskeleton design it is paired with. To determine the Bowden cable's expected efficiency an estimation of transmission loss was made.

Bowden cable efficiency is typically modeled using the capstan equation [34], [35]. The model, however, neglects losses other than those due to cable bend. We assumed that there would also be some constant amount of resistance due to factors such as pressure between the cable ends and their guiding structure. As such, a modified version of the capstan equation was used and is shown in Eq. 1 with F_{out} equal to the Bowden cable output force, F_{in} the Bowden cable input force, μ_s the static friction coefficient of the Bowden cable, θ the sum of all bending angles in the cable, and F_{static} equal to the amount of force needed to induce motion on a straight cable.

$$\frac{F_{out}}{F_{in}} = e^{-\mu_s \theta} - \frac{F_{static}}{F_{in}} \tag{1}$$

Eq. 1 was evaluated for a Teflon-coated cable inside a housing with a greased nylon liner. This combination was expected to result in a friction coefficient of *0.052* [35]. It was assumed that a value of F_{static} would be around *5%* of the max input force, or about *2N*. This value was not rigorously vetted but was chosen as a means of providing a more realistic estimation of efficiency. A resistance of up to *5%* of the actuator force could be expected, however resistance greater than this would seem indicative of poor tolerances. The operational range of the Bowden cable bend angle was determined to be 0° to 180°, allowing for 90° bends at the shoulder and elbow from a back mounted device. The typical operational range of the actuators was expected to be *10* to *40N*. From these parameters a range of expected efficiency was modeled. For all operational conditions, the efficiency was expected to be greater than *65%*, with most operation being *70%* to *80%* efficient. This was deemed acceptable and as such the Bowden cable design was implemented.

A *5 mm OD* Bowden cable housing with a *2 mm ID* greased nylon liner guides the motion of a *1.5 mm* Teflon-coated braided steel cable. *M3* threaded terminals are soldered onto the cable ends. Clevises are threaded onto the input side to allow for pinning to the actuators. Cable terminals have *15 mm* long,

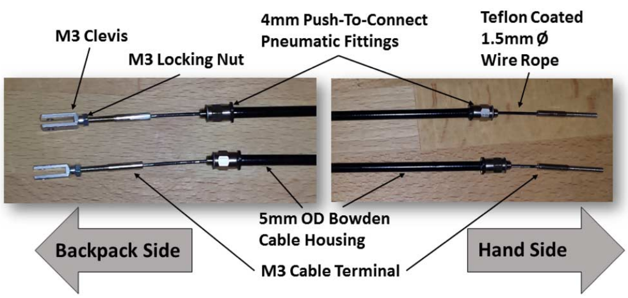

Fig. 2. View of Bowden cables removed from UHMWPE guide blocks. Bowden Cable constructed of commonly available low-friction housing and cable materials.

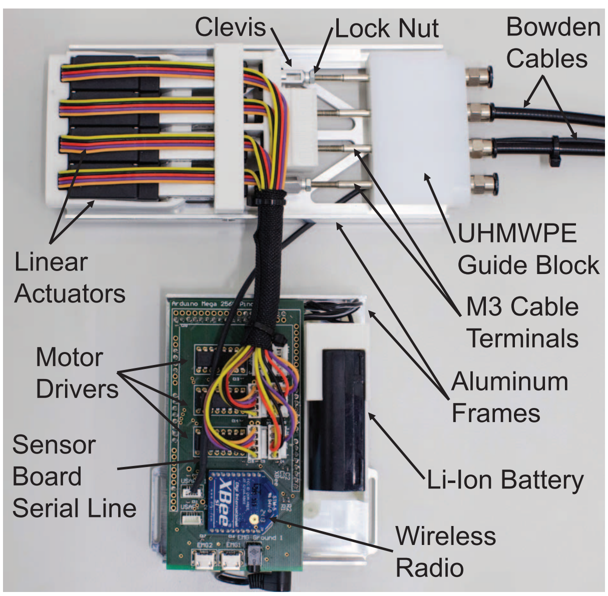

Fig. 3. Construction of actuator and electronics packs with protective covers removed. The actuator module is positioned above the electronics module with a single cable bundle connecting the two. Two of four possible Bowden cables are attached in this configuration.

3.5 mmø shanks which are inserted into ultra-high-molecular-weight polyethylene (UHMWPE) guide blocks mounted in the backpack and exoskeleton. The shanks slide inside these guide blocks, providing linear motion and preventing buckling of the cable end under compression. A pneumatic quick connect placed on either end of the Bowden cable assembly allows for threading the housing into Helicoil® inserts on the guide blocks.

A diagram of the Bowden cables, removed from their guide blocks, is shown in Fig. 2. The actuator pack guide block can be seen in Fig. 3 and the hand side guide blocks are shown in Fig. 4.

B. Remote Actuation Pack Design

The remote actuation pack is split into two sections, an actuation module and an electronics module, allowing for simpler construction and greater modularity. The actuation module mounts four DC linear actuators (L12-30-100-6-P, Figelli Technologies Inc., Victoria BC, Canada) and Bowden cable UHMWPE guide block on a *2 mm* thick aluminum frame. The motors were chosen for their light *35g* weight, compact $74 \times 15 \times 18$ *mm* size, *40N* output force, and *6 mm/s* speed at peak power. These motors allow for an exoskeleton like Arata's to fully open or close, against *60%* of it's maximum force, in less than *1.5s*. Printed PLA plastic mountings hold the motors in place with a PLA cover protecting internal mechanics. The actuation module, with cover removed, is shown in Fig. 3. The electronics module follows a similar construction, with *2 mm* thick aluminum frame and printed PLA cover, and is shown in the bottom half of Fig. 3.

C. Electronics Design

The electronics design consists of a main control board, a peripheral sensor board, and an interface device. A commercially available microcontroller board (Mega2560 R3, Arduino LLC) is used for the main control board. This board handles wireless communication with the interface device as well as UART communication with the peripheral sensor board. The main control board also handles running motor control loops and setting inputs on up to three dual H-bridge L293B motor drivers. Control electronics and motors are powered from a *7.2 V*, *1700 mAh* Li-ion battery (PS-BLM1, Olympus Corporation, Shinjuku, Tokyo, Japan). Under heavy use (maximum motor force with a *50%* duty cycle), the average current draw of our final device was *500 mAh*, conservatively allowing for a battery life of about *2.5 hrs*.

A custom peripheral sensor board uses an 8-bit microprocessor (ATTINY828R, ATMEL Corporation, San Jose, California) to read 18 analog input channels, serialize the data, and send it via UART to the main control board with a bit rate of 250kbps. The abundance of analog channels was included to allow for the addition of sensors to meet application need. The board footprint was 22×22 *mm* making it possible to embed in the exoskeleton without increasing the overall size.

For testing, a PC is used for interfacing between user and device. Sensor values from the peripheral board, EMG inputs, and actuator positions are relayed to the computer for storage or viewing. The PC can also directly issue motion commands to the device, this control was used for all testing. Communication between the PC and main board is handled with a wireless radio transmitter (XBEE S1, Digi International, Minnetonka, MN, USA) operating at a bit rate of 57.6 kbps. While a PC was used during testing, a portable device such as a tablet could be substituted as PC functions require minimal processing power.

D. Hand Exoskeleton Integration

Finally, the actuation unit was integrated into the hand exoskeleton of Arata et al. [4]. Two of the four available actuators are utilized providing coupled movement of the index and middle fingers and of the ring and little fingers.

Finger actuation on Arata's exoskeleton is accomplished with a *3* layer sliding spring design depicted in Fig. 4. The layer closest to the fingers is fixed while the center layer is actively pushed and pulled. A top layer freely moves and functions to constrain the motion of the center spring. Pushing the center spring results in a bending motion in the direction of finger flexion. The mechanism was designed to create a range of motion

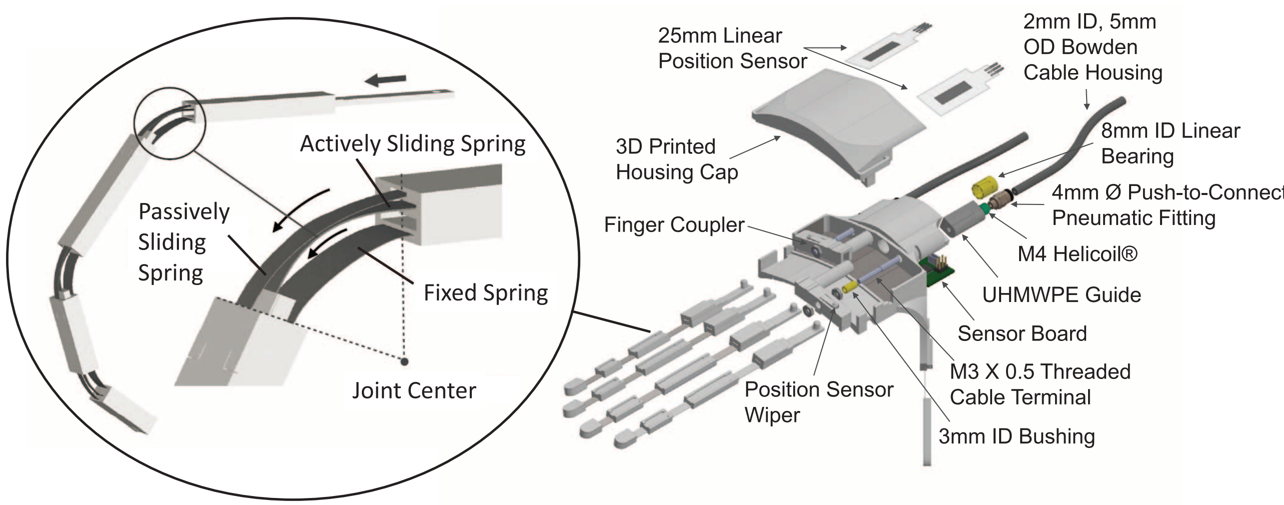

Fig. 4. Exploded view of Arata's exoskeleton modified for remote actuation (right) with detail of Arata's finger mechanism (inset left). Details of the finger mechanism design can be found in the work of Arata et al. [4].

relevant to power grasps as well as finger-thumb opposition. A detailed description of the finger mechanism and its capabilities can be found in [4].

Arata's design was modified only where needed and improved only where possible due to the implementation of remote actuation. Improvements made were the removal of the mounting structure for the motors and shortening the amount that the exoskeleton extends past the wrist.

Necessary modifications were the inclusion of two UHMWPE Bowden cable guide blocks as well as modification of Arata's finger coupling pieces to interface with the Bowden cable ends. Two guide blocks are pressed into the back of the exoskeleton housing and retained by a clip added to the housing design. The couplers have through-holes added for fixation to the threaded Bowden cable terminals. Shafts added to the back of the couplers slide inside linear bearings. The couplers also have small wipers added for "Soft Pot" linear position sensors (P-L-25-103-3%-ST,Spectra Symbol, Salt Lake City, Utah, United States) bonded to the Housing's cap. An exploded view of the modified exoskeleton is shown in Fig. 4.

E. System Overview

An image of the complete system, with hand exoskeleton attached, is shown in Fig. 1. The actuation module and electronics module are joined with fabric allowing for the pack to bend and form better to the user's back. Fig. 5 shows an individual wearing the final device. The actuator module is centered on the back, mounted above the scapula (roughly between vertebrae T2-T4) while the electronics module is mounted between the scapula (roughly vertebrae T4-T7). This design orients the Bowden cables for efficient operation and minimally interferes with sitting and arm movement.

Fig. 5. User wearing the complete system. The actuation pack is mounted high on the user's back with a modified version of Arata's exoskeleton on the user's hand.

IV. Testing and Characterization

The goal of the system was to improve portability by being lightweight and non-cumbersome for a person with reduced mobility. The final system meets our definition of portable in that all actuation, control, and energy storage are handled by a wearable and sealed package. The device is untethered and has all components necessary for daily use. The design also sits high on the back while taking up the amount of space of a small backpack. This device was fully tested and characterized to determine if the design goals had been quantitatively met.

TABLE I
COMPARISON OF ARATA ET AL.'S ORIGINAL EXOSKELETON AND OUR REMOTELY ACTUATED (RA) DESIGNS

	Arata et al. [4]	**RA**
Exoskeleton Weight (g)	256	113
Remote Weight (g)	201	754
Actuator Pack Weight (g)	N/A	346
Electronics Pack Weight (g)	201	286
Transmission Weight (g)	N/A	61
————	———	———
Actuated DOF	2	4
————	———	———
Exoskeleton Size (mm)	130 X 84 X 32	98 X 84 X 18
Actuation Pack Size (mm)	N/A	175 X 79 X 26
Electronics Pack Size (mm)	88 X 103 X 49	83 X 114 X 46
Transmission Size (mm)	N/A	5⌀X 750
————	———	———
Battery Capacity (Wh)	10.4	12.2

A. Weight and Dimension

Table I compares the exoskeleton previously presented by Arata et al. alongside the updated remotely actuated version. Dimensions are listed as $L \times W \times H$ unless otherwise noted where H is the maximum perpendicular distance the component rises above the body. The actuation system weight of *754g* is below the targeted *1 kg* value.

Weight of the exoskeleton was decreased by 56% through the implementation of remote actuation and the height of the exoskeleton was reduced by 63%. Total system weight increased from *457g* to *867g*. This added weight stems from the transmission weight and from the two additional actuated degrees of freedom. The additional motors were included in all measurements although their function was not utilized by this exoskeleton.

B. Transmission Efficiency

To evaluate the transmission efficiency, Bowden cable input and output forces were measured for a range of bend angles and input forces. The output-side of the Bowden cable was held in place with the cable terminal allowed to push freely against a *100N* force sensor (AFG 100N, Mecmesin Limited, Slinfold, West Sussex, UK). The input-side of the Bowden cable was moved to various positions, wrapping around a *100 mm* radius block, to test efficiency at bend angles from *0°* to *180°* at *45°* intervals. Force was applied to the input side of the Bowden cable by pushing on a *50N* force sensor (Type 9205, Kistler Holding AG, Winterthur, Switzerland). Tests were conducted under a low load input condition of *10N* and a maximum load input condition of *40N*. Force was applied slowly to avoid any dynamic effects until the desired measurement force was reached. Applied forces were kept to within *+2N* of the desired setpoint.

Also measured was the value of F_{static} from Eq. 1. This was done by straightening the Bowden cable and removing the force sensor from in front of the output. Force was slowly applied until motion was induced, the value of this force was then recorded. The average value of F_{static} from *15* trials was determined to be *1.7N*.

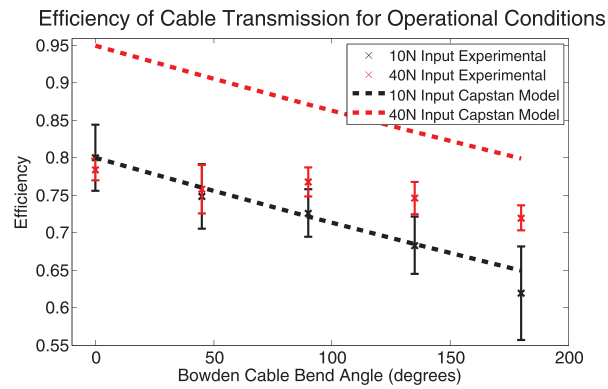

Fig. 6. Result of experimental efficiency tests compared with the capstan model. *70–80%* efficiency is observed for most test conditions. The capstan model is seen to be a good predictor at low loads but break down under high compressive loads.

The results of the efficiency tests are shown in Fig. 6 alongside the capstan model of Eq. 1. For the *10N* trial, the model can be seen as a reasonable predictor of transmission loss. The model, however, was no longer accurate at the *40N* input case. During testing, buckling of the inner cable was observed under larger compressive loads, leading to greater contact between cable and housing and a breakdown of the model. Regardless of this effect, reliable operation was observed and efficiency for all inputs was above 60% with most tested conditions above 70% efficient. The analysis of efficiency was not intended to determine a precise model but rather guide design decisions such as motor selection and cable path.

C. Output Position Compensation

While an ideal Bowden cable would maintain a constant path length independent of bend angle, this is not observed experimentally. To measure the effect that bending the Bowden cable has on its output position, tests were conducted with the remote actuation system integrated with Arata's modified hand exoskeleton.

To test the effect of using an open loop position controller, which sets motor position and assumes Bowden cable output follows, the Bowden cable was bent from *0°* to *180°* at *45°* intervals around a *100 mm* radius. Static measurements of finger pose were taken at each of these positions. Motion capture markers were placed on the palm and fingertip as well as the centers of the metacarpophalangeal (MCP), proximal inter-phalangeal (PIP), and distal inter-phalangeal (DIP) joints. Five motion capture cameras (Optitrack, NaturalPoint, Inc., Corvallis, Oregon, United States) were used to measure the pose of the finger with results shown in Table II.

The optical tracking test revealed that, starting at an initially neutral pose, bending the Bowden cable housing *180°* varied total finger flexion by *49.4°*. This change in finger flexion resulted in a *53.0 mm* movement of the fingertip position. These results verified that there is a need to compensate for transmission-induced errors.

We implemented closed loop position control of Bowden cable output and tested the ability of the actuators and

TABLE II
ANGULAR CHANGE IN FINGER FLEXION (SUM OF MCP, PIP, AND DIP JOINTS) AND CORRESPONDING POSITIONAL CHANGE OF THE EXOSKELETON FINGERTIP CAUSED BY BENDING THE BOWDEN CABLE

Cable Bend (°)	Angular Change (°)	Positional Change (mm)
0	0	0
45	1.9	2.9
90	27.9	30.2
135	32.4	34.7
180	49.4	53.0

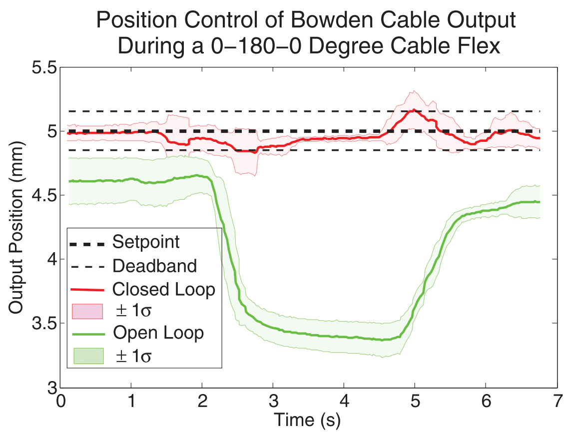

Fig. 7. Compensation of Bowden cable output position for bending motion from 0° - 180° - 0° with and without feedback. Each test was run 10 times with mean value and standard deviation shown. A constant output position setpoint of 5mm was explicitly commanded for both trials. Closed loop control was accomplished with a Bang-Bang controller.

embedded position sensors to compensate for dynamic bending motions. To do this, the cables were straightened and an output position of *5 mm* was set. Open loop control was implemented as the Bowden cable was bent from *0*° to *180*° and then from *180*° back to *0*°. Each movement lasted about *1s* with a *2s* pause between motions. The test was repeated ten times with actuator output position logged from the internal sensors, with results plotted in Fig. 7. This procedure was then followed while a bang-bang controller was used to regulate Bowden cable position around the desired set-point. This bang-bang controller has a *0.15 mm* deadband allowing positions near the setpoint to be held against the back-drive force of the motors, saving power. The results of closed loop position control, with setpoint and deadband labeled, are shown in Fig. 7.

Addition of closed-loop feedback control compensated for much of the deviation created from the cable bend. Deviation around the setpoint was kept to within ± *0.3 mm*. This positional deviation approximately correlates to the error seen in the *45*° cable bend from Table II, resulting in a fingertip deflection of only about *3 mm*. Since the selected motor gearing results in a maximum motor speed which closely matches the speed of desired movements, bang-bang control is effective at setting position for this exoskeleton.

V. DISCUSSION

We have presented the design of a remote actuation system which is lighter weight than other, similar devices, identified in past work [18], [28]. While our design put priority on weight reduction, it still provides sufficient output force, control capabilities, and battery life for reasonable use. Integration of the device with an existing exoskeleton demonstrated these abilities as well as the benefits of using remote actuation in reducing the weight placed on the hand and arm. Furthermore, the system has shown itself to be capable of compensating for deficiencies in positional control introduced by the remote actuation system. Closed-loop position control reasonably maintains positional accuracy against errors induced by bending the Bowden cable.

Our analysis and measurement of transmission loss demonstrated a reasonably efficient system and revealed that a capstan model, which has been used to model Bowden cable transmissions previously [35], can break down under large compression loads. This analysis was conducted to guide design choices and our findings were included to help shape expectations and design decisions of other groups looking to implement a similar type of system.

The modular approach taken in creating separate actuation and electronics modules, as well as the incorporation of additional actuation and sensor inputs, has allowed for a device which can be reconfigured, re-positioned, and expanded upon to meet user needs and adapt to advances in exoskeleton design. This actuation system will allow for the addition of thumb movement, with a mechanism such as the one described in Lambercy et al. [30], or added dexterity to the index and middle finger in future revisions. Actuation of the wrist could also be incorporated with the included motors.

The weight realized by the remote actuation system combined with the reduction in weight of the exoskeleton has lead to a practical and portable device weighing only *867g* in total with only *113g* on the weakened limb. A well developed prototype is critical to study in the field, allowing for patient trials. As a next step, clinical evaluation will determine if the weight reduction was sufficient, and patients feel comfortable wearing the remote actuation system for extended periods. Future evaluation of the exoskeleton will inform on its ability to assist stroke survivors in performing activities of daily living. While patients suffering from muscle weakness are the primary target group for the proposed device, the force output of the current prototype might limit the applicability for stroke survivors suffering from hypertonicity of the finger muscles, which affects about *30–40%* of patients [36]. Further development of the exoskeleton will investigate the possibility of adjusting parameters of the spring blades (e.g. thickness) and selection of motors to increase the force output without compromising the weight of the device.

REFERENCES

[1] A. S. Go *et al.*, "Heart disease and stroke statistics—2014 update: A report from the American Heart Association," *Circulation*, vol. 129, no. 3, p. e28, 2014.
[2] A. C. Lo *et al.*, "Robot-assisted therapy for long-term upper-limb impairment after stroke," *New Engl. J. Med.*, vol. 362, no. 19, pp. 1772–1783, 2010.
[3] J. Mehrholz, T. Platz, J. Kugler, and M. Pohl, "Electromechanical and robot-assisted arm training for improving arm function and activities of daily living after stroke," *Cochrane Lib.*, no. 4, Article no CD006876, 2008.
[4] J. Arata, K. Ohmoto, R. Gassert, O. Lambercy, H. Fujimoto, and I. Wada, "A new hand exoskeleton device for rehabilitation using a three-layered sliding spring mechanism," in *Proc. IEEE Int. Conf. Robot. Autom. (ICRA)*, 2013, pp. 3902–3907.

[5] J. Colebatch and S. Gandevia, “The distribution of muscular weakness in upper motor neuron lesions affecting the arm,” *Brain*, vol. 112, no. 3, pp. 749–763, 1989.

[6] L. Ada, C. G. Canning, and S.-L. Low, “Stroke patients have selective muscle weakness in shortened range,” *Brain*, vol. 126, no. 3, pp. 724–731, 2003.

[7] S. Balasubramanian, J. Klein, and E. Burdet, “Robot-assisted rehabilitation of hand function,” *Curr. Opin. Neurol.*, vol. 23, no. 6, pp. 661–670, 2010.

[8] O. Lambercy *et al.*, “Effects of a robot-assisted training of grasp and pronation/supination in chronic stroke: A pilot study,” *J. Neuroeng. Rehabil.*, vol. 8, no. 1, p. 63, 2011.

[9] D. G. Kamper, H. C. Fischer, E. G. Cruz, and W. Z. Rymer, “Weakness is the primary contributor to finger impairment in chronic stroke,” *Arch. Phys. Med. Rehabil.*, vol. 87, no. 9, pp. 1262–1269, 2006.

[10] E. Cruz, H. Waldinger, and D. Kamper, “Kinetic and kinematic workspaces of the index finger following stroke,” *Brain*, vol. 128, no. 5, pp. 1112–1121, 2005.

[11] P. Heo, G. M. Gu, S.-J. Lee, K. Rhee, and J. Kim, “Current hand exoskeleton technologies for rehabilitation and assistive engineering,” *Int. J. Precis. Eng. Manuf.*, vol. 13, no. 5, pp. 807–824, 2012.

[12] M. Cempini *et al.*, “Kinematics and design of a portable and wearable exoskeleton for hand rehabilitation,” in *Proc. IEEE Int. Conf. Rehabil. Robot. (ICORR)*, 2013, pp. 1–6.

[13] A. Wege and G. Hommel, “Development and control of a hand exoskeleton for rehabilitation of hand injuries,” in *Proc. IEEE/RSJ Int. Conf. Intell. Robots Syst. (IROS)*, 2005, pp. 3046–3051.

[14] M. A. Rahman and A. Al-Jumaily, “Design and development of a hand exoskeleton for rehabilitation following stroke,” *Proc. Eng.*, vol. 41, pp. 1028–1034, 2012.

[15] J. Iqbal, H. Khan, N. G. Tsagarakis, and D. G. Caldwell, “A novel exoskeleton robotic system for hand rehabilitation–conceptualization to prototyping,” *Biocybern. Biomed. Eng.*, vol. 34, no. 2, pp. 79–89, 2014.

[16] P. Polygerinos *et al.*, “Towards a soft pneumatic glove for hand rehabilitation,” in *Proc. IEEE/RSJ Int. Conf. Intell. Robots Syst. (IROS)*, 2013, pp. 1512–1517.

[17] L. Connelly, Y. Jia, M. L. Toro, M. E. Stoykov, R. V. Kenyon, and D. G. Kamper, “A pneumatic glove and immersive virtual reality environment for hand rehabilitative training after stroke,” *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 18, no. 5, pp. 551–559, Oct. 2010.

[18] Gloreha Light. *Idrogenet srl* [Online]. Available: http://www.gloreha.com/index.php/en/versions/gloreha-lite01, accessed on Aug. 21, 2015.

[19] M. Delph *et al.*, “A soft robotic exomusculature glove with integrated sEMG sensing for hand rehabilitation,” in *Proc. IEEE Int. Conf. Rehabil. Robot. (ICORR)*, 2013, pp. 1–7.

[20] H. In, B. B. Kang, M. Sin, and K.-J. Cho, “Exo-glove: A soft wearable robot for the hand with a soft tendon routing system,” *IEEE Robot. Autom. Mag.*, vol. 22, no. 1, pp. 97–105, Mar. 2015.

[21] R. F. Beer, M. D. Ellis, B. G. Holubar, and J. Dewald, “Impact of gravity loading on post-stroke reaching and its relationship to weakness,” *Muscle Nerve*, vol. 36, no. 2, pp. 242–250, 2007.

[22] R. F. Beer, J. D. Given, and J. P. Dewald, “Task-dependent weakness at the elbow in patients with hemiparesis,” *Arch. Phys. Med. Rehabil.*, vol. 80, no. 7, pp. 766–772, 1999.

[23] B. Gasser and M. Goldfarb, “Design and performance characterization of a hand orthosis prototype to aid activities of daily living in a post-stroke population,” in *Proc. 37th Annu. Int. Conf. IEEE Eng. Med. Biol. Soc.*, 2015, pp. 3877–3880.

[24] P. Aubin, K. Petersen, H. Sallum, C. Walsh, A. Correia, and L. Stirling, “A pediatric robotic thumb exoskeleton for at-home rehabilitation: The isolated orthosis for thumb actuation (IOTA),” *Int. J. Intell. Comput. Cybern.*, vol. 7, no. 3, pp. 233–252, 2014.

[25] M. Cempini, M. Cortese, and N. Vitiello, “A powered finger–thumb wearable hand exoskeleton with self-aligning joint axes,” *IEEE/ASME Trans. Mechatron.*, vol. 20, no. 2, pp. 705–716, Apr. 2015.

[26] N. Smaby, M. E. Johanson, B. Baker, D. E. Kenney, W. M. Murray, and V. R. Hentz, “Identification of key pinch forces required to complete functional tasks,” *J. Rehabil. Res. Develop.*, vol. 41, no. 2, pp. 215–224, 2004.

[27] CyberGlove Systems Inc. (2015). *Cybergrasp* [Online]. Available: http://www.cyberglovesystems.com/cybergrasp, accessed on Dec. 18, 2015.

[28] P. Polygerinos, Z. Wang, K. C. Galloway, R. J. Wood, and C. J. Walsh, “Soft robotic glove for combined assistance and at-home rehabilitation,” in *Proc. Robot. Auton. Syst.*, 2014, pp. 135–143.

[29] N. Ho *et al.*, “An EMG-driven exoskeleton hand robotic training device on chronic stroke subjects: Task training system for stroke rehabilitation,” in *Proc. IEEE Int. Conf. Rehabil. Robot. (ICORR)*, 2011, pp. 1–5.

[30] O. Lambercy, D. Schröder, S. Zwicker, and R. Gassert, “Design of a thumb exoskeleton for hand rehabilitation,” in *Proc. 7th Int. Conv. Rehabil. Eng. Assistive Technol.*, 2013, p. 41.

[31] A. B. Zoss, H. Kazerooni, and A. Chu, “Biomechanical design of the Berkeley lower extremity exoskeleton (BLEEX),” *IEEE/ASME Trans. Mechatron.*, vol. 11, no. 2, pp. 128–138, Apr. 2006.

[32] H. K. Yap, J. H. Lim, F. Nasrallah, J. Goh, and R. Yeow, “A soft exoskeleton for hand assistive and rehabilitation application using pneumatic actuators with variable stiffness,” in *Proc. IEEE Int. Conf. Robot. Autom. (ICRA)*, May 2015, pp. 4967–4972.

[33] F. Zhang, L. Hua, Y. Fu, H. Chen, and S. Wang, “Design and development of a hand exoskeleton for rehabilitation of hand injuries,” *Mech. Mach. Theory*, vol. 73, pp. 103–116, 2014.

[34] A. Schiele, P. Letier, R. van der Linde, and F. van der Helm, “Bowden cable actuator for force-feedback exoskeletons,” in *Proc. IEEE/RSJ Int. Conf. Intell. Robots Syst.*, 2006, pp. 3599–3604.

[35] L. E. Carlson, B. D. Veatch, and D. D. Frey, “Efficiency of prosthetic cable and housing,” *J. Prosthetics Orthotics*, vol. 7, no. 3, pp. 96–99, 1995.

[36] A. Thibaut, C. Chatelle, E. Ziegler, M.-A. Bruno, S. Laureys, and O. Gosseries, “Spasticity after stroke: Physiology, assessment and treatment,” *Brain Injury*, vol. 27, no. 10, pp. 1093–1105, 2013.