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TORQUE CONTROLLED EXOSKELETON FOR UPPER LIMB REHABILITATION

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TORQUE CONTROLLED EXOSKELETON FOR UPPER LIMB REHABILITATION

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Stroke commonly results in abnormal muscle synergy, spasticity, muscle weakness and neural couplings. Stroke patients often present with movement deficits with respect to range of motion (ROM), joint coordination, and movement smoothness in the affected arm and hand. Our previous work showed that HandSOME, a spring-powered hand exoskeleton that compensates for flexor tone in the fingers and thumb, improves ROM and function while worn. This study aims to 1) investigate if an independent home therapy program using HandSOME can improve unassisted ROM and functional grasp of the affected hand, and 2) to design a portable and lightweight exoskeleton suitable for home based arm rehabilitation. This dissertation was broken into three components.

1) Design, testing and evaluation of HandSOME II

2) Exoskeleton for Upper Limb Rehabilitation with Series Elastic Actuator and Cable-driven Differential

3) Design and prototyping of a low inertia 3 DOF shoulder exoskeleton for upper limb rehabilitation.
This dissertation by Tianyao Chen fulfills the dissertation requirement for the doctoral degree in biomedical engineering approved by Peter S. Lum, Ph.D., as Director, and by Sang Wook Lee, Ph.D., Gregory P. Behrmann, Ph.D., Arash Massoudieh, Ph.D., and Ozlem Kilic, D.Sc. as Readers.

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3.1 Introduction .......................................................................................................... 36
1 Introduction

There are 800,000 new strokes in the United States each year [1]. According to the Centers for Disease Control, stroke is a leading cause of serious long-term disability in the United States [2]. As the survival rate from stroke has increased each year [2], more and more survivors are in need of rehabilitation. In the year of 2016, the American Heart Association/American Stroke Association (AHA/ASA) issued first-ever stroke rehabilitation guidelines calling for intensive, multidisciplinary treatment [3]. These new guidelines are significant as they show the AHA/ASA recognizes the important of rehabilitation in the period after the initial event. Previously stroke has been considered by many as an acute event. As a matter fact, it is the chronic condition in which patients often live with a disability for the rest of their lives. But these guidelines show this attitude is changing. They represent a huge effort to expand care of stroke patients into the chronic phase, which should help people lead the fullest of lives. The chronic phase is usually considered staring six month after the onset of stroke.

The latest Evidence Based Review of Stroke Rehabilitation (EBRSR) [4] reports that it is more difficult to recover the lost function in the upper extremities than in the lower extremities, and that limited rehabilitation resources, time constraints, and a lack of early motor recovery in the upper extremity deterred researchers to focus therapy on improving balance, gait and general mobility. However, individuals with stroke tend to have long lasting difficulty in performing activities of daily living (ADL) such as reaching, grasping and lifting objects. After their acute phases, the rehabilitation resources and intensity normally drop down for stroke patients as their medical conditions become more stable. The so-called critical window for therapeutic
intervention is gone as motor recovery has reached its plateau. However recent studies of chronic post-stroke motor recovery have reported that the intensive therapeutic training in the chronic phase lead to significant improvement in motor function [5, 6, 7, 8, 9, 10]. Therefore more effort should be put forth to help design therapeutic interventions suitable for chronic stroke patients.

1.1 Stroke Characteristics

Stroke happens when the disruption of normal blood flow to the brain leads to its cell death. There are mainly two types of stroke: ischemic due to lack of blood flow, and hemorrhagic, due to internal bleeding of brain itself. As a result, part of brain does not function properly. Signs of stroke may include movement and sensation deficits, speaking and vision problems. Movement deficits are associated with reduced active range of motion (ROM) when reaching against gravity [11], abnormal muscle coactivation and weakness [12], disrupted interjoint coordination [11], decreased smoothness of movement [13].

1.1.1 Abnormal muscle synergy

Muscle synergy is a term used to describe the neural strategy of simplifying multi-degree-freedom control [14]. To achieve a functional synergy, a single neural command signal produced by the central nervous systems (CNS), recruits a pattern of co-activation muscles. Normal muscle synergy allows for almost instantaneous engagement and dis-engagement of these muscle co-activations. In the case of abnormal synergy, neural coupling between upper limb joints are
formed and reduces the capacity to control the joints independently. It is present in individuals with moderate to severe stroke. Flexion synergy pattern as shown in Figure I.1.11, as the most prevalent movement pattern in chronic stroke, is defined as the coupling of shoulder abduction with elbow, wrist and finger flexion, and forearm supination during movement [15, 16]. Extension synergy pattern is defined as the coupling of scapular protraction, shoulder adduction and internal rotation, elbow extension, forearm pronation and wrist and finger flexion [15]. These couplings reduce patients’ active range of motion when reaching and grasping against gravity.

Figure I.1.11 Image of a chronic stroke subject performing the task of 90 degree shoulder elevation starting from shoulder dropped position in the frontal plane, along with extended elbow and supinated forearm. The flexion synergy pattern present in this patient, includes elbow flexion, shoulder abduction, forearm supination, finger flexion, which can been seen from the left to right image. And this abnormal pattern becomes more pronounced as the shoulder is elevated against gravity.
1.1.2 Spasticity

Spasticity may be a contributor to abnormal synergy patterns across multiple joints, with symptoms such as tight or stiff muscle, an inability to control those muscles, hyperactive reflexes persisting for too long. It occurs when damage to the CNS affects the upper motor neurons [17]. Imbalance in the excitatory and inhibitory input to upper motor neurons leads to increased excessive muscle contraction. In the case of hand, motor neurons cannot properly respond to the excitation and inhibition command from CNS by knowing when to switch between finger flexors and extensors. As a result, hand flexor muscles are often over-excited and stroke patients have difficulty in extending their fingers to grasp a small real-life object such as milk container shown in Figure 1.2. A defining feature of spasticity is that an increased resistance to passive stretch and this resistance is velocity dependent [18,19]. The faster the passive movement the stronger the resistance. Hypertonia is resistance to passive stretch as well and can be the result of neural and non-neural effects, while spasticity is considered a neural phenomenon.
Figure I.2 Image of a chronic stoke patient using his paretic hand to grasp a small milk container. Both of his hands are supported by the table. This patient is not able to open his hand to grasp the container due to spasticity and finger extensor weakness.

1.1.3 Muscle Weakness

Muscle weakness is decreased muscle force and can be due to decreased neural drive and/or muscle atrophy. It leads to strength imbalances across opposing muscle groups on torque production [20]. Work space in the paretic arm of stroke patients is thus reduced. This can be easily demonstrated with a stroke patient using the paretic arm attempts to lift up an object. As the mass of object increases, with weakness of elbow extensors under higher shoulder loading conditions, the patient tends to decrease the height the object lifted and to bring the object closer to the patient’s body. Grip strength can be used as a representative measure of muscle weakness in upper extremity after stroke as grip strength and all arm strength measurements are highly correlated [21].

1.1.4 Difference between stroke and other neurological diseases

There are many neurological conditions and diseases that can affect movement. Parkinson's disease is the result of the loss of dopamine-producing brain cells. This disease can progress aggressively. Multiple sclerosis (MS) is a disabling disease of the brain and spinal cord, in which the immune system attacks the protective sheath (myelin) that covers nerve fibers and destroys the communication networks formed by neurons between the brain and the rest of body. Amyotrophic lateral sclerosis (ALS), also called Lou Gehrig's disease, attacks the nerve cells that
are used in voluntary muscle actions. Causes of ALS can be gene mutation, chemical imbalance, and protein mishandling, etc. Dystonia is a disorder characterized by involuntary muscle contractions that cause slow repetitive movements, which is usually intensified by physical activity. Cerebral palsy (CP) is caused by abnormal development or damage to the parts of the brain that control movement and balance, during pregnancy, child birth, or shortly after birth. On the one hand, Parkinson’s, MS and ALS lead to aggressive and non-irreversible neuron degeneration. On the other hand, Dystonia and CP are often birth related and hereditary, and can lead to permanent motor dysfunction. Different from these movement disorders, stroke is often behavior and environment related, and not associated with neurodegeneration and neurological birth defects. This would put stroke clinicians and researchers in a good situation to devise effective prevention and rehabilitation strategies.

1.2 Rehabilitation Robotics for Upper Extremity

Movement therapy, as a typical form of rehabilitation, is generally recommended to begin as soon as critical conditions are under control and patients are able to resume self-care activities at some level [22]. Training intensity and repetition are key features in movement therapy. The use of robotic devices to treat upper limb movement is increasingly accepted as it offers these features in more precise manner with less therapist involvement. Rehabilitation robots may be classified in two categories: end-effector and exoskeletons, depending how the device is linked to the users. In the first category, users hold the extremity of the mechanical structure of the end-effector. PHANToM, among the first robots, demonstrated the possibility of end-effector for arm rehabilitation [23]. Later on, these studies with using MIT-Manus [24,25] Mirror Image
Movement Enabler system (MIME) [26,27], and HapticMaster based Gentle rehabilitation system [28] showed the robot-assisted movement was safe and well tolerated by patients with stroke and resulted in measurable reductions in impairment. However, these devices lack control over the full arm as their only interaction with the human arm is at the hand or wrist. In the exoskeleton category, the device is attached to several points of the user’s body, and follows his anatomy. Although it brings more complexity to the design, the exoskeleton not only allows the user to know the position of the end-tip, but also different parts of the body, and to apply force and torque control on these contact points.

1.3 Exoskeleton

Exoskeleton allows repetitive training and as well as the retraining of joint coordination [29,30]. Its family may be further broken into two types: active and passive. Active exoskeleton uses DC motor or pneumatic actuators to assist or augment movement, whereas the passive exoskeleton uses spring to power movement.

1.3.1 Active exoskeleton

Many countries have undertaken the development of active exoskeletons suitable for helping stroke patients. The 6-DOF robot ARAMIS (Automatic Recovery Arm Motility Integrated System) by Dolce et al. [31] is able to characterize motor disability and residual function and outcome in a quantitative way. 6-DOF (4 active and 2 passive) ARMin robot developed by Nef et al. [32], addressed interactive forces by using PD and gravity compensation controller. 7-DOF (5 active DOF) arm MGA exoskeleton developed by Carignan et al. [33] considered the motion of the scapulothoracic joint and is able to articulate translation of the shoulder and to resolve the issue of shoulder singularities. A 7DOF cable-driven robot arm prototype by Yang et al. [34], has
a lightweight structure of 1kg. A EMG-based 7-DOF exoskeletal robot developed by Kiguchi and Hayashi [35], can assist 3 DOFs in shoulder (around glenohumeral joint), 1 DOF in elbow, 1 DOF in forearm, 2 DOFs in wrist through dc motors. A 8 DOF electromechanical arm robot (ARMOR) developed by Mayr et al. [36], is capable of moving all joints through complex patterns. A 9-DOF portable arm exoskeleton from Schiele and van der Helm [37] optimized the kinematic structure of human arm interfacing exoskeleton by reducing linear and angular misalignment between human joints and robot joints. Other active exoskeletons such as ViSHaRD [38], the German Aerospace Center (DLR) bimanual interface [39], and the sensoric arm master (SAM) [40] are designed as a master haptic device for teleoperation but has the potential to be modified for stroke rehabilitation.

If DC motor actuation is replaced by pneumatic muscle actuators (pMA), a better power/weight ratio can be achieved to allow a low system weight. This group includes RUPERT [41] which has 4DOF driven by pneumatic muscles on shoulder, elbow and wrist, a “soft-actuated’ 7-DOF upper-limb exoskeleton developed by Tsagarakis and Caldwell [42] which weighs less than 2 kg and uses Pneumatic Muscle Actuators (pMA) for power source, 7 DOF (3 shoulder, 1 elbow, 1 forearm, 2 wrist) wearable robot designed by Jeong et al [43] that used pneumatic actuators with proportional control valve; a 9 DOF exoskeleton robot developed by Chen et al. [44] that used the pMA actuators.
To reduce robot inertia felt by the human arm during rehabilitation, an idea was also proposed to relocate the actuators with Bowden-cables outside the movable exoskeleton structure. This can help exoskeleton achieve high power/weight ratio as well. A cable driven CAREX has DC motors mounted on a large aluminum frame over the chair where patients sit [45]. A tendon driven 5 DOF L-EXOS has all motors located on a fixed frame behind the user [46]. A 6 DOF MEDARM for motor rehabilitation of the shoulder complex and elbow flexion/extension, moved away the DC motors by cable driven mechanisms [47]. The design of 14 DOFs X-Arm-2 relocated the most power-demanding DC actuators to a back orthosis by means of Bowden-cable transmission [48].

The concept of relocating actuation source is appealing but brings more complexity due to the requirement of an annex system or cable routing. In addition, user’s hand often has to hold a handle at the end of these devices. This would prevent the paretic hand from training effectively in task oriented practice. Also most active exoskeletons are not wearable and are not easily portable. Even the SAM and X-Arm2 haptic exoskeleton for teleoperation have very limited portability with dependence on desktop computers and large external hardware for control.

Several active exoskeletons have also been developed that assist movements of the hand isolated from the rest of the limb [49,50,51,52,53,54,55,56,57]. These devices require sitting in the clinic with the arm supported in a pre-defined posture and/or don’t allow interacting with objects. The potential to transfer gains to real life situations could be limited, given growing evidence of
abnormal coupling of proximal and distal control in stroke patients, such that arm posture [58], activation level of proximal muscles [16], and level of arm support [59] can affect control of hand muscles. Examples of hand robots that can be used during ADL practice include the Hand-of-Hope, which is powered by five linear actuators and offers individual control of each digit [60]. The PneuGlove [61] is a pneumatically powered glove that contains air bladders that extend the fingers when inflated. Cybergrasp (Immersion Inc, San Jose, CA) uses cables routed through a linkage mounted to the back of the hand [62]. Extension force in each cable is controlled with five motors located remotely. The X-Glove is a portable device with 5 linear actuators that independently extend the digits [63]. A 3-fingered hand exoskeleton was developed to augment hand strength using linear actuators [64]. These approaches are promising, with the ability to finely control assistance levels to each digit during task practice. However, their tethered designs do not allow for integration with daily activities or easy transition to home-based therapy interventions.

1.3.2 Passive Exoskeleton

The alternative would be exoskeletons equipped with passive power augmentation. It uses a mechanical spring or elastic cord (such as rubber band and bungee cord) as the actuation source, depending on the level of movement assistance needed by the patient’s upper extremity. Comparing to active exoskeleton, this passive approach has the following advantages. It is cost effective. The price of springs are at least one magnitude less than that of motors. It can be very easy to mount and adjust on the device. Rubber band, Bungee cord and mechanical spring often come in a ready-to-mount fashion. Even if coming in as a long string, they can be tied as a closed
loop, or have a hook at its end. Mechanical springs can be contained even in a structure as a pluggable module. The weight of the device can be much less without motors or cable systems. This also helps reduce the effort for maintenance and makes the device more portable and sturdy, which are very important features to motivate stroke patients in home training. In addition, the springs available on the market can cover the torque assistance spectrum required for patient’s upper extremities. It was reported that the distal part of the arm including forearm, wrist and hand usually need less assistance than the proximal part including shoulder, upper arm, and elbow [33]. Wrex developed by Iwamuro et al. [65] is a passive arm orthosis that provides the arm support against gravity through elastic band. It applied the concept of four-bar parallelogram and allowed users with moderate and above severity of impairment to achieve relatively large ROM [65]. AmeroSpring, the commercial version of the upgraded T-Wrex [66], applied the concept of zero-length spring by relocating the spring to be aligned with one link in the parallelogram to achieve better gravity compensation and thus better control of movement. SaeboFlex [67] and SaeboGlove (Saebo Inc., Charlotte NC), used mechanical spring or elastic cord to help extend fingers to grasp the objects.

1.4 Goals of this dissertation

In spite of the small amount of work done on passive exoskeleton, they have shown potentials to help upper limb rehabilitation for stroke patients. The following research further tested usability of a passive hand exoskeleton for home based therapy. This research aims to develop a passive exoskeleton not just for hand, but for the rest of arm at the paretic side to help chronic stroke patients. It can be divided into three sections.
Section 1 covers an independent home therapy program using HandSOME, a spring-powered hand exoskeleton that compensates for flexor tone in the fingers and thumb. Our previous work showed that HandSOME improves range of motion (ROM) and function while worn. This study aims to investigate if a home therapy using HandSOME can improve unassisted ROM and functional grasp of the affected hand. Individuals with chronic stroke completed a 4-week home intervention with a data logger recording the number of movements completed. Outcome measures were collected before and after the intervention and in a 3-month follow up.

Section 2 covers the development of a 5-DOF spring actuated exoskeleton, called SpringWear. This device was developed to increase range of motion and functional performance in the affected arms of stroke patients. Theoretically perfect gravity compensative is provided at the shoulder elevation DOF. Additional torque is provided to assist weakness in forearm pronation/supination and elbow extension. A game interface is also designed to provide a training mode for stroke patients to use SpringWear.

Section 3 covers initial testing of SpringWear in chronic stroke patients. Range of motion and functional ability were tested with and without assistance from SpringWear. Hand biomechanics such as aperture and finger joint angles were examined with the assumption that assistance provided at the proximal part of arm would benefit the distal part of arm.
Section 4 covers the design improvement on HandSOME and SpringWear. Currently the development of HandSOME R3 and SpringWear 2.0 is under way. The design of HandSOME R3 focuses on improving the sensor and datalogger technology, and refining the structure. The design of SpringWear 2.0 focuses on incorporating gear system in ROM measurement, reducing the deformation shoulder elevation mechanism under heavy load.
2 Hand rehabilitation after stroke using a wearable, high DOF, spring powered exoskeleton

There are 800,000 new strokes in the United States each year [1]. Functional recovery in the upper limb after stroke is strongly dependent on motor recovery in the paretic hand [2], however the probability of regaining functional use of the impaired hand is low [3]. Most stroke survivors regain the ability to flex the fingers voluntarily, but extension is limited by inappropriate activity in flexors [4] and inability to activate extensors [5]. Devices that apply extension force to the fingers may facilitate effective task practice, allowing completion of movements that would otherwise be impossible to complete unassisted. The majority of robotic hand devices under development do not allow practice of reach and grasp tasks with real physical objects in real life situations [6,7]. Training the hand in isolation from more proximal arm joints may limit gains in functional recovery, given the abnormal coupling of proximal and distal control in stroke patients, such that arm posture [8] and activation level of proximal muscles [9] can affect control of hand muscles.

However, a few robotic hand devices can be used in whole arm tasks. These include the PneuGlove [10], a pneumatically powered glove that contains air bladders that extend the fingers when inflated, and the X-Glove[11] that has linear actuators that independently extend the digits. Previously we developed the Hand Spring Operated Movement Enhancer (HandSOME), a lightweight wearable exoskeleton with elastic elements that apply extension assistance to the fingers that compensate for the flexor tone [12]. Wearing HandSOME significantly increased range of motion and functional ability in chronic stroke subjects. However, HandSOME only allows movement in 1 DOF (pinch-pad grip). We describe the development and initial testing of
HandSOME II. This device uses the same concepts as the original HandSOME but allows isolated movement and customized extension assistance to 11 finger and thumb DOF, allowing performance of all the grip patterns used in daily activities.

2.1 Mechanical Design

2.1.1 Main design goals.

1) We wanted to achieve individual joint movement and assistance for all finger metacarpophalangeal (MCP) joints. Proximal interphalangeal (PIP) and distal interphalangeal (DIP) joints of each finger are coupled during normal movements, so we opted to couple these DOF together in the device as well. 2) All the mechanical parts need to be on the back of the hand to minimize the interference between fingers, especially when the hand is fully closed. Therefore, none of the device DOF would be aligned with the human finger DOF. 3) At the thumb, we wanted independent carpometacarpal (CMC) add/abduction, CMC flex/extension, and interphalangeal (IP) movement and assistance. 4) To decrease friction in the device, lubricated strengthened acrylic sheet and polished ABS plastic (fused-deposition modeling) are used.
2.1.2 Finger linkage design.

Figure 2.1. Three different design options (described in the text). Each joint consists of a proximal phalange (pink) and a distal phalange (blue). The mechanical parts are strapped to the phalanges with Velcro straps. Mechanical stops limit motion to 90 degrees of flexion.

Human fingers are a 3 DOF serial linkage. We assume DIP, PIP and MCP joints are all 1 DOF rotary joints. Lateral range of motion is neglected. Considering design goal #2, the simplest linkage design will be to add two bars on top of the finger and form a four-bar linkage structure, as shown in Figure 1A. The distal end of the bar linkage has two DOFs, therefore forces applied to the distal phalange are not orthogonal to the phalange. Forces can have a component parallel to the phalange, causing the distal attachment point to slide along the skin. Additionally, the attachment pad can rotate relative to distal phalange causing one edge of the pad to dig into the skin.

By adding extra links (Figure 1B) we can remove 1 device DOF. This parallelogram keeps the end point on a circular trajectory and by aligning the center of rotation with the human joint, only forces orthogonal to the distal phalange are applied. However, the distal finger pad is still
free to rotate. Since this pad must be the ground point for the linkage for the next finger DOF, reaction forces from that linkage will rotate this finger pad, causing discomfort.

Our final design is to add one more link to constrain the orientation of the finger pad (Figure 1C). Not only will the finger pad travel around a circular trajectory, it will also rotate around the center of rotation maintaining perfect alignment with the distal phalange. All extra DOF are eliminated and the linkage and human finger joint together form a structure with only one DOF. The applied force direction is identical to the more common design of aligning the exoskeleton DOF with the human DOF.

Figure 2.2 Thumb CMC joint angle definitions.
2.1.3 Thumb mechanical parts design.

The thumb CMC joint has three DOFS: Abd/adduction, flex/extension and oppo/reposition (Figure 2). Our design focuses on the former two. We used Euler angles to represent abd/adduction and flex/extension angles. Coordinate system XYZ (red) is global and fixed to the back of hand with its origin at CMC joint. Coordinate system X”Y”Z” (blue) is the local coordinate system attached to the thumb metacarpal bone. We used sequence Z-Y‘-X” to represent the rotation from XYZ to X”Y”Z”. Angle α is the first Euler angle rotation around the Z axis and represents the CMC extension angle. CMC abduction is represented as β, the second Euler angle rotation around the Y’ axis. The third Euler angle is rotation around the metcarpal bone and was not considered.

The thumb’s metacarpal bone, is not easily accessible for attachment. Considering that the thumb MCP joint has a relatively small range of motion, we decided to simply jump over the thumb MCP joint and strap to the thumb proximal phalange.

2.2 Torque profile design

2.2.1 Design principles

In previous work, we found that a torque profile that decreases as the finger flexes maximizes available range of motion [12]. We decided to use this same principle for assistance. To incorporate different options for the slope or rate of decline in torque, we modeled and designed
different anchor points and wrapping options to allow the user to easily change the applied torque profile.

For a steep torque profile that drops to zero when fully closed, we only use two anchors for the spring (Figure 3). As the finger closes, the spring path gets closer to the rotation center of the linkage. Hence the torque gets smaller due to decreasing lever arm length, despite the fact that spring force is increasing as the finger closes. For a slower torque decline, we add a pin(s) that the spring wraps around as the finger closes, which results in a slower decline in lever arm length.

Figure 2.3. Method for adjusting spring paths for the MCP (A) and PIP joint (B). Without the extra wrapping pin, the spring path is shown in red. Blue and green lines shows how the path is altered with the addition of wrapping pins.
2.2.2 MCP joint torque profile

Figure 2.4. Spring path geometry for torque modeling. ‘O’ represents rotary joint. Blue lines represent the spring path. Green lines are auxiliary lines. P1 and P_base are two necessary anchors for the spring. P2 is additional wrapping pin. L,R,S,l,r represent line lengths in the picture. \( \gamma \) and \( \theta \) represent angles defined in the picture.

In an ideal case, spring force \( F \) is proportional to its length, \( F = KL \) (zero rest length and constant elastic coefficient \( K \)). The torque applied around the center of rotation \( O \) is \( M = FSSin(\gamma) = KLSsin(\gamma) = 2KA \), with \( A \) being the total area of triangle P1-P_base-O, as shown in Figure 4A. From here we know that the torque is proportional to the triangle area. The maximum area happens when \( \theta = 90^\circ \). As \( \theta \) increases above 90°, the area and applied torque drops. We can use this as a guide to design the position of pins and the spring paths.

For the steepest decline in the torque profile, no extra pins are used, as shown in Figure 4A.

\[
L_{\text{spring}} = L_{p1,p\_base} = \sqrt{R^2 + S^2 - 2RS\cos(\theta)} \quad (2.1)
\]
To calculate spring force $F$, we used a Matlab 3rd order polynomial regression to model the length-force relationship of the small rubber bands we used for springs. Empirical data were collected and we used the curve fitting function to determine force, given a certain length. Spring force is given by:

$$F = f(L, n) \quad (2.2)$$

with $L$ being the length of the rubber band and $n$ being the number of rubber bands. The moment around point $O$ is:

$$M = F\sin(\gamma) = f(L_{P1,P_{base}}, n)\sin(\gamma) \quad (2.3)$$

Using law of sines, $\frac{\sin(\gamma)}{R} = \frac{\sin(\theta)}{L_{P1,P_{base}}}$, the moment is:

$$M = f(L_{spring}, n)SR\sin(\theta)/L_{P1,P_{base}} \quad (2.4)$$

For a slower torque drop, an extra pin ($P2$), is added. For small flexion angles near full extension, $P2$ is not engaged and does not affect the spring path and the torque is the same as in equation (iv). For larger flexion angles, $P2$ engages and changes the spring path (Figure 4B):
\[ L_{spring} = L_{P1, P2} + L_{P2, P_{base}} = l + \sqrt{r^2 + S^2 - 2rS \cos(\theta)} \]

If we assume no friction between the rubber band and pin, the force in the spring is the same on either side of the pin. Using equation (ii) to get spring force \( F \):

\[ M = f(L_{spring}, n)S \sin(\theta) / L_{P2, P_{base}} \]  \hspace{1cm} (2.5)

The calculation is similar when the second extra pin, P3 is added to further slow the decline in torque as the finger flexes. In the previous steps we assumed the tension in the spring is the same on either side of the pin. This is the ideal case when the spring and the pin have zero friction. However we did also consider the other extreme case of infinite friction. Under this assumption, the moment the rubber band touches pin #1 or #2, it will stick to the pin and not slide over the pin. The calculated torque did not differ dramatically from the zero friction assumption.

A testbed with a one DOF mechanical human joint was built to test the torque applied to the finger. We used an encoder (GL 60, Contelec, Biel, Switzerland) and loadcell (MLP 50, Transducer Techniques, Temecula, CA, USA) to record the angle vs torque phase plot. The linkage was attached to the mechanical finger and the mechanical finger was moved very slowly, around 0.3 rad/s, through its range of motion to mimic a quasi-static movement. The theoretical and measured profiles agreed reasonably well (Figure 5).
Figure 2.5. Calculated and measured torque profiles with different spring path options. Full extension is 0 degrees and the x-axis represents flexion angle. Darker lines are for 8 rubber bands and the lighter lines are for 6, 4, 2 rubber bands respectively. Only the 8 and 2 rubber band cases are shown in the measured torque profiles. The hysteresis in the empirical data is due to friction in the linkages. The rapid rise in the measured torque profile between 0 and 5 degrees is an artifact due to contact with a mechanical stop on the test apparatus which caused the sensor to measure only a portion of the applied torque in this range.

2.2.3 PIP and DIP coupling

The linkage for the PIP joint is similar to the MCP design. However, because of space limitations, we only had the option for one extra pin to change the torque profile slope. The measured PIP torque is shown in Figure 2.6.
Figure 2.6. Measured torque profile for PIP joint. Darker lines are with 4 rubber bands and lighter lines are for 2 rubber bands.

In subjects who have limited DIP range of motion, DIP assistance is also required, and an optional gear mechanism can be added with a fingertip cap. As noted before, only in very rare situations do the PIP and DIP move independently. Figure 7 shows the gear design to couple PIP and DIP joints. We chose dimensions to align the virtual centers of rotation with the human joints. Given that the device is strapped firmly to the finger, we assumed the shaded triangle is a rigid object. With a 4:5 gear ratio, $\Delta \varphi : \Delta \psi = 4:5$, DIP and PIP also moves in a 4:5 ratio, since all the linkages are based on virtual parallelograms. Now that the DIP and PIP are coupled, they will share the torque provided by the rubber bands. The exact sharing ratio is indeterminant.
Figure 2.7. Laser cut spur gears that couples PIP and DIP joint.

2.2.4 Thumb torque profile

The calculated thumb flex/extension and add/abduction torque profiles are shown in Figure 8. The flex/extension spring is located on the back of hand. The additional space available allowed for more spring path options and more torque profiles can be achieved. However, the add/abduction spring path options were limited because of space constraints, and so less options were available for torque profiles.

Figure 2.8. Achievable torque profiles for thumb extension and abduction. Different colors represent different spring path options. Lighter colors are for smaller numbers of attached springs.
Figure 2.9. HandSOME II on a stroke patient’s hand. This subject required DIP assistance in the index finger and was fitted with the gear and fingertip cap. The other fingers did not require DIP assistance.

2.3 Testing

We used a motion capture system (Osprey, Motion Analysis Corporation, Santa Rosa CA, USA) to test range of motion with and without assistance from HandSOME II (Figure 9). Markers were placed on the fingers or the exoskeleton to measure movements of the index finger and thumb. Eleven subjects with chronic stroke were tested, among them 10 of them completed
range of motion data. Subjects closed the hand as far as possible and then opened the hand as far as possible. Test results are shown in Table 2.1.

<table>
<thead>
<tr>
<th>Subject</th>
<th>MCP</th>
<th>PIP</th>
<th>DIP</th>
<th>MCP</th>
<th>PIP</th>
<th>DIP</th>
<th>Extension</th>
<th>Abduction</th>
<th>Extension</th>
<th>Abduction</th>
</tr>
</thead>
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<td>135</td>
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<td>145</td>
<td>180</td>
</tr>
</tbody>
</table>

Table 2.1. Improvements when wearing the HandSOME II. Range of motion is max extension minus max flexion.
Figure 2.10. Max extension angle of finger joints.
Figure 2.11. Range of motion comparison of finger joints.

Figure 2.12. CMC joint angle comparison of thumb.
Besides range of motion tasks, subjects were also required to complete 7 picking tasks. For each task, subjects were asked to pick up a daily object on the table with their impaired hand, and drop it into a bin next to it. The seven daily objects include: 1 inch wood block, 2 inch wood block, 3 inch wood block, baseball, marble, pen, and key. The task is successful if they were able to pick the object up and drop it into the bin within 30 seconds. We recorded the number success for each patient with and without the device. The result is shown below:

Figure 2.13. Picking task result.

Among the eleven subjects who completed this test, six of them showed a significant improve with the HandSOME than without it. The other five subjects either can complete all the task with and without the device, or have 1 less success while using the device.
2.4  Conclusion

We have developed an exoskeleton that allows movement at 15 hand DOF. At 11 DOF separate elastic elements can be added to each DOF to customize the extension assistance for individual subjects. In testing with 7 chronic stroke subjects with impaired finger movement, both range of motion and max extension angles were significantly increased when wearing the device. While it is not surprising that applied extension torque would increase max extension angles, the ability to apply torque profiles that decrease with increasing flexion allowed the range of motion to also increase and the grip force to only decrease by 9.2%. The high DOF of this device will theoretically allow a large range of grasp tasks and individual finger movements. Use of the device would allow practice of object manipulation tasks even in subjects with poor hand extension ability, and may be more effective than unassisted task practice.

2.5 Design iteration

In our previous design, we used rubber band to drive the exoskeleton. The adjustment of assistance is achieved by using different type and number of rubber band. However, it can be very hard for stroke patients to handle such tiny rubber band by themselves. Another downside of rubber band is its short lifespan. For these reasons, we are motivated to improve the design to make the prototype function better. We replaced rubber bands with steel coil tension spring (Lee Spring Co.). And we designed a string-knob mechanism so that the user can easily change the pretension of the spring, thus change assistance level. Before the finalized string-knob design, we have tried many approaches to adjust the string length. Due to constrained space, the mechanism need to be very small. We have tried the ratchet mechanism, push and release mechanism and
even copied the design of retractable paper knife. They all failed to meet our requirement, either because it’s too difficult to fabricate, or it’s too hard for a patient to operate with one hand.

Figure 2.14. Design iteration with steel coil springs and thumb knob.

Just like the previous design, we hope MCP torque and PIP/DIP torque to be independent. To achieve this, we have two springs actuating the two DOF separately. MCP spring is relatively straightforward. We only use one pin to change the direction of the string so that the string is perpendicular to the screw axis, thus can wrap around the thumb screw as expected. To actuate PIP/DIP joint without affecting MCP torque, we have a piece of string running through the linkages and wrap around all the pins. In an ideal case where the string passes the center of the pins, zero torque will be applied to MCP joint because the lever arm is zero. With a pin of 0.75 mm in radius, the lever arm is very small so the undesired torque around MCP joint can be ignored. We therefore regard the two DOFs as independent from each other.
The string tip is super-glued on to the screw. With several revolutions, the friction of the string accumulates, and the stress on the string tip is relived. And the wrapping of the string on the thumb screw is designed to follow right hand rule. The string should follow the screw thread, which makes the wrapping resistant to tangling. Turning the knob clockwise makes the string wrap more on the screw and the spring will thus be pre-stretched. A tighter spring will thus provide higher torque to the subject. This design is also intuitive to the patients and therapist because it’s consistent to the saying ‘left loopy righty tighty’.

The tension in the string have the tendency to backdrive the thumb screw, which leads to loss of tension or decrease of assistant torque. To prevent this, we tried several method. We tried to use smaller screws (M2), the torque on the screw would be smaller due to smaller screw radius, but the backdrive torque of the screw is also smaller due to smaller radius. We also tried to rely on self-locking mechanism. The string tension will want to tilt the screw, which lead to higher friction in the thread. And the higher the tension, the higher the friction. However, it didn’t work as expected, probably due to insufficient friction. Finally, we decided to use an off-the-shelf lock-nut to increase friction. It greatly prevent backdrive. And the downside is it takes more effort to adjust the screw knob.
We also improved strapping method so that the subject can put on and take off the device all by themselves. Flexible and thin cable tie is used as strap. The strap is actually a hook and loop tape available widely at hardware stores. For thumb and wrist strapping, it is necessary to use hook and loop, because the sticky surface is needed to tighten the strap. For fingers, however, we
insert straps into a narrow slots and rely on friction to hold the strap. Hook and loop tape may not
be necessary here. Better material is open for upgrade in the future.

The improved design is more robust and more user friendly. We are now confident to send
out the prototype to patients for their home training.

2.6 Data logging

We hope to obtain range of motion data when the subjects use HandSOME II in their home
training. We thus propose a data logging system to record finger motion. Unlike HandSOME I,
there is not enough space to mount an encoder. And string-based potentiometer is not desirable
because it adds unexpected friction and resistance. We then came up with proposal of using a
magnetometer and permanent magnet, and measure and record their relative motion.

A disc-shape permanent is bonded to the finger structure, and a magnetometer is located at
the back of the hand. The movement of finger will change the magnet field around the
magnetometer, thus the relative angle between the finger and the back of hand is measurable.
And to eliminate the effect of earth magnetic field around the magnetometer, a second
magnetometer is added. The two magnetometers are both located on the back of hand but are
widely spaced. We assume the further-back magnetometer is free from the magnet’s effect. Thus
the difference between the two magnetometers are purely due to the rotation of the magnet.
Battery, mini SD card, switch, and mini processor is also included and these electronics are
enclosed in a box, which is bolted down on to the back of hand.
3 Exoskeleton for Upper Limb Rehabilitation with Series Elastic Actuator and Cable-driven Differential

3.1 Introduction

Neurological disorders such as stroke are one of the leading causes of long term disability worldwide and negatively affects quality of life [68]. Stroke survivors often experience movement impairments related to inability to activate muscles appropriately for the task [69,70] and abnormal synergies between different muscle groups [71]. Current approaches to neurorehabilitation focus on task-specific repetitive movement practice and there is growing interest in robotic approaches that assist completion of movements and can enhance the effectiveness of the movement practice [72]. A large meta-analysis of 34 studies reported that use of upper extremity robots had advantages over other interventions in promoting recover of strength, arm function and ability to perform activities of daily living [73].

One of the unsolved engineering challenges in the field of robotic exoskeleton design is developing a device that can deliver accurate torque control over an appropriate range. For this application, the device must simulate a wide range of impedances, from highly transparent modes where the torques applied are negligible during “free” movements to higher stiffness modes for assisting movements. In the case of robots for planar end point movements, one approach is using large direct drive motors that drive a 2-DOF linkage [74]. However there are many challenges to achieving adequate performance in upper extremity exoskeletons that typically have 5-DOF. Achieving low impedance in “free” modes is particularly challenging where the mass and inertia of the robot links are often much larger than the mass and inertia of the human upper extremity.
Gravity compensation is straightforward, but the dynamics of robot structure and actuators can only be partially cancelled by active compensation schemes \[^{75}\]. Forcing the patient to adapt to the large inertias inherent in many exoskeletons may limit the effectiveness of the device as a training tool, when the goal is to improve the patient’s ability outside of the robot.

Additionally, the potential for a wearable exoskeleton is severely limited by the mass of many current designs. Exoskeletons commercially available or under development vary widely in terms of kinematics, redundancy level and actuation methods. Limpact utilizes hydraulic actuators, has a max torque of 36 Nm and weighs 8 kg \[^{76}\]. The Armin III robot \[^{77}\] uses DC motors and harmonic drives, and has a maximum moving mass of 7.06 kg. The Caden-7 weighs 6.8 kg, uses a combination of gears and cable drive transmissions to provide 62 Nm of peak torque \[^{78}\]. These robots all have masses that far exceed the mass of the human upper extremity, which is approximately 5\% of body-weight (~3.5 kg) \[^{79}\].

In this chapter we describe the design and testing of the forearm component of an upper extremity robot called Ultra Low Impedance eXoskeleton (ULIX). This component actuates elbow flexion/extension (FE) and forearm supination/pronation (SP). The actuation of SP is particularly challenging because the axis of rotation is along the long axis of the forearm, so a mechanism is needed that allows rotation along this axis while avoiding the human forearm volume. Circular tracks are the most common solution, but actuation often leads to bulky structures and high friction. Our solution uses a novel cable driven differential that couples the elbow FE and SP joints, allowing load sharing of actuators, a compact design and application of pure SP torques, reducing loads and the size of structural members.
We adopted a second mass reduction method commonly used by wearable robotics devices that separate actuation hardware from worn elements through use of flexible Bowden cable transmissions. This approach provides a low worn mass, high torque capacity solution, which is ideal for our application. However, the friction in the cable can be problematic. To minimize the effects of Bowden cable friction and reflected inertia from gearing speed reduction, we used series elastic elements (SEE) to reduce impedance and mask the dynamics of the actuators from the user. However, maximum controllable stiffness is limited by the stiffness of the SEE. Bowden cables and SEE have been used effectively in other rehabilitation robots \[^{80}\], however the combination of these elements with a differential mechanism is novel. Performance of the system is described in bench-top testing and the effects of critical design choices will be discussed.
A. Tethered system  
B. CAD illustration  
C. Human arm in prototype

**Fig. 3.1.** Mechanical design of the tethered system. A. The system consists of off-board actuation and control hardware, flexible Bowden cable, and end-effector worn by the subject. B. The end-effector has two independent series elastic actuator (SEA) on each side of the elbow. Their coordination provides the desired torque for elbow flexion/extension and supination/pronation. The two SEAs are connected together through the forearm ring, which is free to rotate relative to the frame. Carbon fiber extension and handle are bolted to the ring. C. Prototype of exoskeleton.

### 3.2 Design

We designed and constructed a novel 2-DOF exoskeleton for torque control at elbow flexion/extension (FE) and supination/pronation (SP). We characterized system performance in benchtop testing with measurement of torque accuracy, bandwidth, step response time, impedance, max stiffness and 2-DOF coordination. Figure 1 shows CAD models and a picture of the final
prototype. The design specifications were maximum continuous torque of 10 Nm and 5 Nm of continuous torque in FE and SP, free mode impedance of .02 Nm/deg at 1 Hz, and maximum stiffness of .5 Nm/deg. The peak torque specifications are comparable to what was implemented in our prior clinical trial using the ARMin robot [81], but the target free mode impedance is significantly lower.

3.2.1 Mechanical Design

To reduce the moving mass of the exoskeleton, we decided to use a cable transmission to locate the motors remotely, a strategy used by many wearable robotic devices [80,82]. A novel cable-driven differential is integrated to couple the two DOFs and allows load sharing at each DOF. Each motor and gear can provide up to 9 Nm continuous torque, giving a maximum flexion/extension torque of 18 Nm and maximum supination/pronation torque of 31 Nm (due to additional speed reduction through the differential).

The transmission is compact and has the potential of becoming part of a wearable, mobile exoskeleton. Series elastic elements are included in the transmission to facilitate high accuracy torque control as well as low impedance. Most structural parts are aluminum alloy and made with computer numerical control (CNC) machining. Stress analysis and finite element analysis (FEA) simulations were done as part of the design process. Manual optimization was done to decrease weight. “I” shaped cantilever structures were widely used. To further reduce inertia and maintain structure rigidity, carbon fiber material was used for distal components supporting the handle. The overall weight is less than 0.9 kg.
Mechanical design of the modularized series elastic actuator. A. The series elastic element is a two-part assembly, consisting of a large wheel (blue) and a small wheel (green). These two wheels can rotate relative to each other within ±12° and they are connected by tension springs in a circular pattern, forming a custom torsion spring assembly. There are two digital encoders measuring the position of the big wheel and small wheel. On the motor end, a two-part motor pulley keeps the tension in the fishing line. The arched slots on the motor pulley ensure that the two pulley discs can be clamped down at any relative position. Idler rollers are used to regulate the winding of the fishing line on the pulley as well as the angle to enter the Bowden cable tether. B. A close up picture of the series elastic element.

3.2.1.1 Series elastic actuator module

Series elastic elements can be used to partially mask the reflected inertia and friction in motors and transmissions, allowing accurate torque control and low impedance \[^{83, 84, 85}\]. We used two identical SEA modules (Figure 2). A DC motor and gear head (RE40, GP42C, Maxon Motor AG,
Sachsein, Switzerland) are selected to power each SEA. The torque of each motor is transmitted to the end-effector through a flexible Bowden cables (Lexco Cable Mfg., Norridge, IL) and high stiffness fishing line with 300 lbs capacity (Hercules Pro, LA, CA). We chose fishing line because it has high strength-to-diameter ratio and near zero stretch within the expected operating forces. Also, from a practical standpoint, it is easy to tie knots on fishing line. For each SEA module, two Bowden cables and two lengths of fishing line are needed. One challenge is maintaining the tension in the fishing line and while keeping the friction low. At the actuator end, we designed a two-part motor pulley and each length of fishing line is firmly tied to one side of the pulley. By adjusting the relative rotary position of the two sides of the pulley, we can adjust the tension in the fishing line. Idler rollers are positioned so that the fishing lines wrap around nearly the entire circumference of the pulley, so that the radial force applied to the pulley shaft by the fishing line is minimized.

On the end-effector end, a two-part torsion spring, made of aluminum 6061, acts as the elastic element. The component has two concentric wheels. The larger wheel is attached to the distal sections of the Bowden cables and converts the force in the Bowden cable to torque. The smaller wheel is attached to the distal forearm components. Linear springs connect the 2 wheels and provide a torsional stiffness between the 2 wheels. To measure the rotary deflection of the wheels relative to ground, we used spur gears laser cut from Delrin. Two incremental encoders (E4T, US Digital, Vancouver, WA) measure these rotations. Each encoder has a resolution of 0.25 degree, and the gear ratio further increases the resolution by 184:19 and 140:19, respectively, yielding a final resolution of 0.03 degree. The relative rotation of these 2 wheels represents the rotary deflection of the torsion spring.
The torsion spring can hold up to 12 tension springs and a variety of stiffness values can be achieved by choosing different type and number of spring. To ensure balanced forces, springs should be evenly spaced. Smaller spring free lengths and higher initial tension result in a more linear stiffness. For our application, however, an exponential-shaped profile was selected. This will be discussed below.

### 3.2.1.2 Cable driven differential

Mechanical differentials can be found in robotic joints with gimble-like kinematics. The human elbow flexion/extension and forearm supination/pronation joints move as a gimble and motivated the use of a mechanical differential. There are several advantages of this approach compared to using two independent actuators. A more compact and lightweight design could be developed because of more balanced forces applied to the structure. Also, the load sharing by the 2 motors theoretically reduces the reflected inertia of the motor/gear by 50%. If the desired peak torque requires a gear reduction of 100/1, two load sharing motors can be used with a reduction of 50/1 for each motor. The scaling factor for the reflected inertia of the one motor system would be $100^2$, while the scaling factor in the two motor system would be $2\times50^2$, a reduction of 50%.

Classical differentials usually contain bevel gears since right angle transmission is needed. Bevel gears have high weight, bulkiness, backlash and high customization cost. A novel cable-driven differential was presented in [86]. The structure is very compact and requires minimal mechanical parts. By carefully designing the path of the cable and its pre-tension, a low friction, small backlash right angle transmission can be achieved. We designed a similar mechanism and constructed a
cable driven differential with 4 additional rollers to regulate the path of the cable, as shown in Figure 3.

For each side of the differential, two lengths of cable tie the SEA to the forearm ring, forming a closed loop. When the two SEAs rotate in the same direction, the two SEAs are locked to each other through the ring, and the exoskeleton will rotate around FE axis with no SP rotation. When the two SEA rotate in opposite direction, the forearm ring will rotate around SP axis with no FE rotation. It is clear that undesirable coupling between DOF is possible if control of the SEA units is not coordinated.

Supination/pronation range of motion is designed to be at least 180°, or 90° for supination and 90° for pronation. Considering a gear ratio of $R=1.78$, each SEA needs to be able to rotate ± 159°. Flexion/extension range of motion is designed to be at least 135°. Each SEA is required to rotate at least 135°. The overall range of motion of each SEA is 0± 159° to 135± 159°, which is -159° to 295°, equivalent to 0 to 454°. The two cables are tied to each SEA and forearm ring with enough revolutions to satisfy range of motion requirements.

Torque generated by the two SEA modules, $T_1$ and $T_2$, contribute to FE torque and SP torque in the following manner, where $T_{FE}$ is FE torque, $T_{SP}$ is SP torque, and $R = 1.78$ is the gear ratio between the forearm ring and SEA.

<table>
<thead>
<tr>
<th>Equation</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>$T_{FE} = T_1 + T_2$</td>
<td>3.1</td>
</tr>
<tr>
<td>$T_{SP} = R \times (T_1 - T_2)$</td>
<td>3.2</td>
</tr>
</tbody>
</table>
The forearm ring is a custom 4-point-contact ball bearing. The outer ring of the bearing is fixed to the forearm frame and the inner ring can rotate freely around SP axis and is connected to the handle. The tolerance for the ball-contact surface can be adjusted by adding or removing paper-thin shims. To reduce weight, the 4-piece bearing was CNC milled from aluminum 7075. And the ball-contact surface was polished by hand. The balls, however, are hardened alloy steel. An aluminum-steel contact is not the best option but so far we haven’t seen any wear on the bearing. We used undersized plastic balls in between every two steel balls to further reduce weight and keep steel balls evenly spaced. The two pieces of cable wind around the inner ring of the bearing in a double helix. Not only does this minimizes the overall width of the winding, but also ensures that the pretension in the two lengths of cable are perfectly balanced, causing no moment or force on the bearing.
Fig. 3.4. Diagram of physical model of series elastic actuator. Physical model can be simplified by the assumption that cable transmission is rigid.

3.3 Control

A physical model of the system is shown in Figure 4. The physical model consists of three mass-spring-damper modules. The motor pulley is connected to the series elastic element through Bowden cables, and the series elastic element is connected to the handle through the cables in the differential mechanism. Since we used closed cable loops and pre-tensioned high stiffness cable, we assumed the cable transmission is rigid, meaning that spring stiffness of $k_1$ and $k_2$ are infinite. The two encoders measure the angular displacement of the series elastic element. Their difference is the spring deflection, which we used to calculate spring torque. The SEA encoders can be used to measure FE and SP angles for feedback control. We added a 5th encoder to measure the angle of the forearm ring directly, providing a more accurate measurement of SP angle.
**Fig. 3.5.** Diagram of physical model of series elastic actuator. Physical model can be simplified by the assumption that cable transmission is rigid.
Fig. 3.6. Control diagram illustrating torque control approach. Current spring deflections (SD1 and SD2) are the difference between encoders on each SEA. They are converted to current torque T1 and T2 through the SEE stiffness k. Flexion-extension torque and supination-pronation torque T_FE and T_SP are calculated according to Equation 1 and Equation 2. Ref_FE and Ref_SP represents reference torque in flexion-extension and supination-pronation. The flexion-extension torque error, e_FE, is feed into both controllers with the same sign so that both SEA can work together to decrease e_FE. The supination-pronation error, e_SP, is fed in to the controllers with opposite sign. At the output stage, control voltages voltage, u1 and u2, are sent to the motor amplifiers.
<table>
<thead>
<tr>
<th>A. SEA stiffness</th>
<th>B. Impedance plot</th>
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<td><img src="image" alt="Series elastic element stiffness" /></td>
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<tr>
<th>C. Step response</th>
<th>D. Bode plot</th>
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<td><img src="image" alt="Step response" /></td>
<td><img src="image" alt="Bode plot" /></td>
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**Fig. 3.7.** Benchtop test on system dynamics and performance. A). Series elastic element stiffness. Experimental data vs simulation data. B). Experimental data on system impedance in free mode. End-effector was moved by human in different frequency and reactive force and torque is measured with ATI Gamma torque sensor. C). System step response on F.E. and S.P. D). System dynamic response bode plot. Single SEA module was measured independently. F.E. and S.P. performance is measured afterward.
We used traditional PID control and tuned the gains of each SEA separately using Ziegler-Nichols method \cite{87}. Reference torque signal is the input and is converted into a command spring deflection angle. We also added a feed forward term associated with the torque/current constant of the motor and gearing ratio to improve performance.

We designed and implemented different control modes, including a) free mode, b) wall mode and c) torque control mode. In free mode, the exoskeleton follows the movement of the user with minimal residual force felt by the user. Reference torque is zero and thus the motors work to keep spring deflection zero. For wall mode, we allow free mode within a certain angular range, and outside the range, the motor regulates spring length so that the user feels resistance to movement similar to hitting a wall of predefined stiffness. For torque control mode, the desired torque profile is converted into a desired spring deflection pattern.

3.4 Experimental Method and Results

We conducted benchtop testing and quantified the performance in terms of the stiffness profile estimation, single SEA module bandwidth, flexion/extension and supination pronation step response, torque accuracy bandwidth, impedance in free mode, max wall stiffness and 2-DOF coordination. Bowden cables were set at a 90 degree angle, which represents the worst case scenario when friction is the highest \cite{88}.

3.4.1 Stiffness profile estimate

The stiffness of the SEA was characterized by applying known loads and measuring deflection angles (Fig. 6A). This setup contains 6 tension springs with a stiffness of 9.57 N/mm and rest length of 15.75 mm. A theoretical model stiffness was calculated based on the geometry of the
SEA and known spring parameters. Comparison between experimental and theoretical results found a 0.05 Nm RMS error, which is less than 2.5% of applied load. The equation below is a 3rd order polynomial fit for the experimental profile with RMS error of 0.08 Nm.

\[
T = 0.0013\alpha^3 + 0.1685\alpha
\]

Spring deflection is \(\alpha\) and \(T\) is spring torque. The close correlation between experimental data and theoretical predictions shows we can accurately predict stiffness profiles with different spring set ups in the future without the need to confirm experimentally.

### 3.4.2 FE and SP step response

Step response tests were performed for both flexion/extension and supination/pronation. For flexion/extension, a 6 Nm torque step was commanded, with each SEA providing 3 Nm torque, which required around 10 degree of spring deflection angle. For supination/pronation, 4.6 Nm torque was commanded, with each SEA providing 2.3 Nm in torque and 6 degree spring deflection angle. Five trials were done for each setup.

Data is presented in Figure 6C. For flexion/extension, the 90% rise and fall times were 0.110 ± 0.002 sec (mean ± s.d.) and 0.090 ± 0.008 sec, with -0.3 % and 9.1 % overshoot, respectively. For supination/pronation, the 90% rise and fall times were 0.098 ± 0.003 seconds and 0.072 ± 0.001 seconds, with 0.3 % and 4.3 % overshoot, respectively.

### 3.4.3 SEA module bandwidth

Open loop bandwidth tests were done to each SEA independently with 5 trials. Experimental spring deflection data were collected from the open loop system with an input chirp signal and the
output side of the SEA grounded. The open loop system can be modeled as a mass spring damper system and the open loop transfer function from spring deflection angle (SD) to command voltage (u) is:

$$H = \frac{SD}{u} = K_m \frac{1}{Ms^2 + ds + k}$$

in which $K_m$ is motor constant times amplifier gain and SEE stiffness, $d$ is damping ratio and $k$ is spring stiffness and M is the mass of the system, including the reflected inertia from the motor-gear. Using Matlab System Identification Toolbox, with 2 poles and 0 zero, the open loop transfer function, $H_{estimate}$, was determined. The variance-accounted-for (VAF) factor of this estimation was 77.11%.

The closed loop transfer function with a controller C and feed forward gain G has a transfer function of

$$H_{CL} = \frac{SD}{SD_{ref}} = \frac{H(G + C)}{1 + HC}$$

With a known controller C (from Ziegler-Nichols method) and feed forward compensation G, the closed loop transfer function is known and thus its dynamic response can be predicted.

The predicted and experimental Bode plots are shown in Figure 6D. With a desired spring oscillation amplitude of ±6 degree, corresponding to ±1.15 Nm, the SEA’s -3 dB magnitude crossover frequency is 9.8 Hz. The 45 degree phase margin crossover frequency is 9.6 Hz.
3.4.4 FE and SP bandwidth

Both flexion/extension and supination/pronation torque bandwidth was measured. SEA output torque can be estimated by measuring spring deflection angle. However, many factors can result in a different torque applied to the user, such as inertia, friction and springiness in the differential transmission. We used an ATI Gamma 6-axis torque sensor (ATI Industrial and Automation, Apex, NC) to measure applied torque in both flexion/extension and supination/pronation. One side of the torque sensor was fixed to a rigid work bench and the other side was bolted to the forearm ring of the exoskeleton. The torque measured by the torque sensor is the torque applied to the user.

A torque chirp signal with frequency from 0.01 to 15 Hz was applied as input. The amplitudes were 2.6 Nm in FE and 2.7 Nm in SP. This corresponded to spring deflection angles of 6 degree for FE and 4 degree for SP. For each setup, 5 trials were collected. For FE, the -3 dB magnitude crossover frequency was 4.45±0.04 Hz; 45° phase margin crossover frequency was 7.58 ± 0.03 Hz. For supination pronation, -3 dB magnitude crossover frequency was 4.42 ± 0.09 Hz and the 45° phase margin crossover frequency was 7.69 ± 0.08 Hz.

3.4.5 Impedance measurement

In a free mode, the exoskeleton should be able to follow the user’s arm movement with minimal resistance felt by the user. Spring deflection angle is regulated to zero. ATI Mini40 6-axis torque sensor was installed at the base of the handle to measure interaction force and torque between the user and the exoskeleton, while the user moves the robot in sinusoidal patterns at different frequencies. The user only focused on either flexion/extension or supination/pronation movements for any given trial.
An added velocity feedback term was used to decrease the effect of stiction in the differential on SP impedance. The 5th encoder at the forearm ring detects SP movement. This signal is differentiated and compared to the velocity of the large wheel of the SEE. This error signal drives the velocity control feedback loop. This control loop parallels the velocity feedback loop in the PID controller (Fig. 5). However the PID controller uses the small wheel angle as reference signal, but is less effective because movement of the small wheel must overcome stiction in the differential. Movement of the 5th encoder is less restrained by stiction because of elasticity in the differential cables. We have also tried to add a proportional controller driven by the 5th encoder signal, but within stability, the gain was not enough to overcome stiction, making it less effective.

Measured torque $T$ is the output and joint displacement angle $\theta$ is the input. Experimental impedance can be expressed as:

$$Z = \frac{T}{\theta}$$

3.6

Ideally, impedance is small, but will increase with frequency of imposed movement. Theoretically at high frequencies, the impedance will plateau at the inherent stiffness of the SEA, corresponding to the case where the large wheel of the SEA stops rotating and the movement frequency is too high for the controller to follow.

Measured torque $T$ for both flexion/extension and supination/pronation remained small at low frequency. As the frequency increased, impedance $Z$ showed some increase but never reached SEA stiffness. For flexion/extension, impedance stayed around 0.03 Nm/deg at frequencies less than 1 Hz. Max impedance of 0.12 Nm/deg occurred at 3.5 Hz. Maximum residual torque for all
trials was 0.5 Nm. For SP, impedance stayed below 0.02 Nm/deg at frequencies less than 1.6 Hz. Max impedance of 0.09 Nm/deg occurred at 3.1 Hz. Max residual torque over all trials was 0.2 Nm.

### 3.4.6 Wall Mode

Haptic interfaces require simulation of walls or a physical stiffness. The end-effector operates in free mode within a certain range of motion. When exceeding the range, the end-effector should behave as if hitting a rigid wall. The goal of this testing was to determine the maximum possible wall stiffness that can be simulated before instability. The FE wall was set so that below 10 degree, the system runs free mode. Above 10 degree, the system runs wall mode. With an angle of 9 degree into the wall, the measured wall stiffness for flexion/extension reached 0.45 Nm/deg. At a max SEE deflection angle of 12 degree, the theoretical push back torque is around 9 Nm according to Matlab simulations. When the SEE reaches its max deflection, the two moving parts bottoms out and the SEE behaves like a rigid disc, providing infinite stiffness. During our benchtop test, however, we never reached the 12-degree limit. According to [83], the max stiffness cannot exceed inherent stiffness. Our experimental data is consistent with this statement as shown in Figure 7. Due to a gear ratio $R$ of 1.78 in the differential, stiffness for SP is higher than FE stiffness by a ratio of $R^2$. 
Fig. 3.8. Experimental wall stiffness vs. inherent wall stiffness. Below 10 degree, the system runs free mode. The residual force felt by the subject is around 0.2 Nm or -0.2 Nm. Beyond 10 degree, the interactive force increases, as if hitting into a wall. Inherent stiffness becomes infinite at 12 degrees of spring deflection. The experimental wall stiffness is close to the inherent SEE stiffness.

3.4.7 Two DOF coordination

We measured the undesirable coupling between FE and SP by simultaneously applying different command torque profiles to each DOF (Fig. 8). Output torque is measured with the ATI Gamma torque sensor and the forearm ring of the exoskeleton was grounded. FE and SP torques were commanded to the end-effector in the following three scenarios: A) 2 Nm ramp torque on FE and 1.5 Nm sinusoid SP torque, B) 1 Nm sinusoid FE torque and 2 Nm ramp SP torque, C) random FE and SP torques. In scenario A, the RMS error was 0.08 Nm and 0.15 Nm for FE and SP. In
scenario B, RMS error was 0.10 Nm for both DOFs. In scenario C, RMS error was 0.16 Nm for FE and 0.19 Nm for SP.

We also measured the coordination of two DOFs in free mode, which is equivalent to setting both commanded torques to zero. The user grabbed the handle and moved randomly and the residual torque was less than 0.17 Nm for FE and 0.2 Nm for SP. RMS error was 0.10 Nm for FE and 0.07 Nm for SP. The residual torques and joint angle in both DOFs are shown in Figure 9.
Fig. 3.9. Flexion/extension and supination/pronation coordination. Scenario A. 2 Nm ramp command on F.E. and 1.5 Nm sinusoid command on S.P. Scenario B. 1 Nm sinusoid command on F.E and 2 Nm ramp command on S.P. Scenario C. Random sinusoid torque with max torque of 3 Nm on FE and 2 Nm on SP.
3.5 DISCUSSION

The additional gear ratio in SP built in the differential had large implications that should be considered for future designs. Recall that from equation (1) and (2), there is torque amplification in the SP DOF due to a gear ratio of R=1.78. The designed SP torque capacity is 50% of FE torque. Thus, theoretically a gear ratio of 0.5 would seem more logical, but we chose to use a larger ratio, partly to allow a compact design. A larger gear ratio does provide a larger range of operation. Based on equation (1) and (2), the achievable FE and SP torque range has a diamond shape, as shown in Figure 10. With R=1.78, the operational range is larger when compared to a smaller gear ratio and both SEAs work under smaller output load with the higher gear ratio.
A downside of choosing this gear ratio for SP is that the physical stiffness in SP is increased by a factor equal to the gear ratio squared. Therefore the SP stiffness is 3.2 times stiffer than FE stiffness. For elbow FE, since there’s no additional gear ratio, the stiffness is given by:

\[
K_{FE} = 2K
\]

Where \( K \) is the stiffness of each series elastic element. Stiffness in elbow SP is given by:

\[
K_{SP} = 2R^2K
\]

Therefore, errors in positioning of the SEAs are amplified in the SP DOF, and torque errors will be larger in SP than FE. This is most apparent in free mode, where higher stiffness increases the residual force or impedance felt by the user.
Fig. 3.11. The achievable torque in both FE and SP forms a diamond area, as marked by the dashed line. Each corner of the diamond represents when both SEAs are working at 100% torque output. On each edge of the diamond, one SEA is providing 100% output. During normal operation, max SP torque is designed to be 50% of max FE torque. Therefore, the operation range is smaller, as marked by the colored range. The left figure shows our selection of R=1.78. The right figure shows a smaller gear ratio. Operation area is larger for larger gear ratios.

In general, we would like to keep the stiffness low at small SEA deflections to decrease free mode impedance, while maintaining a higher stiffness overall for wall mode, especially for larger deflections. This motivated our choice of an exponential-shaped stiffness profile. We initially used a nearly linear stiffness profiles in the SEE, which resulted in a peak residual torque of 0.3 Nm in free mode testing, as measured by the torque sensor. However, when interacting with the exoskeleton, the human has a much smaller moment arm for SP movement than FE. This 0.3 Nm residual force was still noticeable in free mode. To further decrease residual torque, we implemented the nonlinear spring profile and an additional velocity feedback loop using the added
extra encoder to the forearm ring to measure SP rotation directly, instead of indirectly through the SEA encoders. With these modifications, residual torque was decreased to below 0.2 Nm up to 8 Hz.

Single SEA has a bandwidth of 9.6 Hz, limited by 45 degree phase margin. The motor was operated well under its current limit. The performance is limited by the low stiffness of the SEE and the reflected inertia of the motor and gear. In Figure 6D, the bandwidth measured for each DOF using the ATI Gamma torque sensor is around half of the single SEA’s bandwidth. This can be explained by the extra springiness in the cable transmission of the differential. Even though the cables are pre-tensioned, their springiness still played a role in buffering the torque from the SEA, resulting in a lower bandwidth.

Gains in the PID controller were limited by instability, resulting in a conservative position controller. Static friction in the Bowden cables played a significant role. The motors must work to overcome stiction and initiate the movement. At low frequency when the Bowden cable moves slowly, it’s more subject to stiction, especially when SEA deflection changes direction. Therefore, low-frequency performance is compromised to some extent.

Bi-directional Bowden cable transmission has its own pros and cons. One drawback is the maintenance of tension in the cable. Proper tension is necessary to decrease backlash. On the other hand, pretension also introduces friction. Because SEA deflection is measured directly at the SEE, we are less concerned with backlash in the Bowden cable. Lower pretension in the Bowden cable is acceptable to decrease friction, as long as the pretension is enough to ensure a proper winding of the cable. One method to achieve a lower friction and zero backlash is a hydraulic or pneumatic
transmission [89]. In terms of friction and stiffness, this method generally performs better than cable transmissions. This should be considered for future work.

ULIX has the potential to be significantly lighter than traditional arm exoskeleton robots that have masses of 6.8-8 kg [76,77,78refs]. Our forearm device weighs 0.9kg and we anticipate the moving mass of ULIX will be around 2kg after extending the device to include 3DOF of shoulder motion. However, the cable driven approach used by CAREX has a mass of 1.35kg [90], even lower than our estimate for ULIX. While the moving mass of CAREX is lower, ULIX has the advantage of potentially evolving into a lightweight wearable device, which is our ultimate goal, and not possible with the CAREX approach.

The use of SEA in ULIX effectively reduces the effect of static friction in the actuators and Bowden cables. Residual torque to overcome static friction and initiate movement was less than 0.2 Nm in both DOF. This breakaway torque is significantly higher in robots that rigidly couple the actuators and the joint. For example, the static breakaway torques reported for ARMin III are 0.95 Nm or higher [91].

The combination of a SEA and Bowden cable drive has been studied previously for the LOPES [80] and NEUROExos exoskeletons [85]. They reported torque control bandwidths of 10-11 Hz. Our single SEA bandwidth was similar at 9.8 Hz, but reduced to 4.4 Hz when measured by a sensor at the end point. Nevertheless, our bandwidth is line with the 3-4 Hz bandwidth reported by other arm exoskeletons [92, 93]. Bandwidths as high as 35Hz have been reported for the LIMPACT exoskeleton [94], which also uses SEA, but this required large and expensive hydraulic actuators, which are not compatible with our long term goal of a wearable device.
The LOPES and NEUROExos reported impedance values that increased with frequency of movement up to 0.17 Nm/deg at 4 Hz and 3.2 Hz respectively. The impedance of LIMPACT appears higher, reaching 0.17 Nm/deg at only 2 Hz. Our impedance was less than 0.03 for lower frequencies, increasing to 0.12 Nm/deg at 3.5 Hz for FE and 0.09 Nm/deg at 3.1 Hz for SP. It appears our impedances are lower, but direct comparison is difficult as the testing conditions were slightly different. It is notable that low impedances were achieved in ULIX with potentially portable actuation. We used small motors with a gear reduction of 72/1, while LOPES uses very large motors with 8/1 speed reduction, and NEUROExos and LIMPACT use hydraulic actuation.

3.6 Conclusion

We have described the design of an exoskeleton with series elastic elements and a cable driven differential. The results of benchtop testing provide a detailed characterization of the system dynamics performance. The combination of series elastic elements and cable driven differential resulted in overall satisfying performance. Performance can be further improved with better controller tuning, mechanical design and part selection. The use of exponential-shaped SEE stiffness allowed simulation of an adequate wall stiffness while low stiffness at low SEE angles allowed low impedance values in free mode. Performance overall was adequate for our application of robotic therapy for stroke impaired individuals. More aggressive application, such as for motion augmentation in healthy individuals, are possible if SEE stiffness is increased. Future work includes building a shoulder module with similar features, and integrating a finger exoskeleton \cite{95} to build a whole arm exoskeleton.
4 Design and prototyping of a 3 DOF shoulder exoskeleton

4.1 Introduction

In this chapter, we will present a 3 DOF shoulder exoskeleton prototype. The shoulder exoskeleton is designed to be grounded. Subject will be able to receive assistance in shoulder abduction/adduction, flexion/extension, as well as upper arm rotation.

In chapter 3, we described a cable driven two-DOF elbow exoskeleton (ULIX). The combination of Bowden cable driven SEA and cable driven differential makes the end-effector mobile in space. However it’s not designed to be a portable exoskeleton. Instead, it’s designed to be integrated with a shoulder exoskeleton. This chapter describes a 3 DOF shoulder exoskeleton module with direct drive of SEAs.

4.2 Shoulder exoskeleton kinematics

From a kinematics point of view, there are 3 DOFs on the exoskeleton to allow for shoulder abduction/adduction, shoulder flexion/extension and upper arm rotation. The kinematics is like the shoulder design of [96]. The first moving DOF (M1) next to the ground is shoulder abduction/adduction. Axis of rotation is always perpendicular to the ground. Then a bracket is attached to M1 axis to support M2 Axis. The second axis of DOF (M2) is for shoulder flexion/extension. Axis of rotation is always parallel to the ground. The third axis (M3) of rotation is for upper arm rotation, and the axis is parallel to human’s upper arm.
Figure 4.1. Kinematics of shoulder exoskeleton. Actuator M1 moves shoulder abd/adduction. M2 moves flex/extension. M3 moves upper arm rotation.

The exoskeleton has very large range of motion to accommodate for activities of daily living (ADL). As for the current design, there is 135 degree ROM for shoulder flex/extension, 90 degree upper arm rotation, and 90 degree for abduction. The following figure shows the detail. Upper arm rotation ROM is configurable by clocking the mechanical stop. Abduction ROM is also configurable by different shape of mechanical stop. Flexion/extension ROM, however, is hard to adjust due to structural robustness.
The motivation for using cable drive for ULIX is to remotely locate the actuators to decrease moving weight and inertia, which is significant for distal segments. Now that the shoulder module will be grounded, it’s no longer significant to remotely locate the actuators. The actuators are directly attached next to the joint.

Figure 4.3 below shows the overall mechanical CAD model of the exoskeleton. The kinematics is the same as the figure above. Three DC motors are used to actuate the device. Motor 1 actuates shoulder abd/adduction. Motor 2 actuates shoulder flex/extension. Motor 3 actuates upper arm rotation. Each motor has an incremental encoder integrated at the back. And at each of the motor output shaft there is an elastic element, making each actuator a series elastic actuator (SEA). The shape and coupling of each series elastic element (SEE) is different and is designed to fit in its surroundings. To measure the deflection of the SEE, a second encoder is needed. The motor output shaft is on one end of the SEE, and the structure is on the other end of the SEE. The motor output angle can be found directly with the integrated encoder. We therefore need extra encoders to measure the position of the structure. We included an optical incremental encoder for each SEA,
as illustrated explicitly in the figure below. The position of each encoder its installation is different. Encoder 1 is used to measure abd/adduction angle, and is installed inside an enclosed structural component. SEE angle is measured through a couple of spur gear with a gear ratio of 4:1, meaning the resolution of the encoder is amplified by a factor of 4. Encoder 2 measures shoulder flex/extension angle, and is installed on the side face of the shoulder structure. The angle is measured through a parallelogram linkage, which also serve as gravity compensation component. Encoder 3 measures upper arm rotation angle. It is located coaxially with the cylindrical SEE described later in the next chapter.

Figure 4.3. CAD model of the exoskeleton.
4.3 Mechanical design

4.3.1 Disc-shape SEA design

For shoulder abd/adduction and flex/extension, we used a custom series elastic elements (SEE) of similar setup, as shown below. The SEE consists of a small wheel and a big wheel, their radius are r and R. At initial position, when the two wheels are in phase, there’s no torque because the force in the spring all passes through the center. At a given deflection angle, the force in each spring will apply torque.

The force in the spring is given by:

\[ F = k(L - l_0) \]  \hspace{1cm} (4.1)

According to rule of cosine, spring length L is given by:

\[ L = \sqrt{r^2 + R^2 - 2rR\cos(\theta)} \]  \hspace{1cm} (4.2)

Therefore \( F \) becomes:

\[ F = k(\sqrt{r^2 + R^2 - 2rR\cos(\theta)} - l_0) \]  \hspace{1cm} (4.3)

Next we need to figure out the effective moment arm, which is \( h \) in the figure. Using the area of this triangle,
\[ h = \frac{rR\sin(\theta)}{L} = \frac{rR\sin(\theta)}{\sqrt{r^2 + R^2 - 2rR\cos(\theta)}} \]

Torque on each spring will be:

\[ T = Fh = k(\sqrt{r^2 + R^2 - 2rR\cos(\theta)} - l_0) \frac{rR\sin(\theta)}{\sqrt{r^2 + R^2 - 2rR\cos(\theta)}} \]

With different angle, we can then plot different torque, and can further derive stiffness of SEE.

Figure 4.4. Geometry of disc-shape series elastic element.
4.3.2 Cylindrical SEA design

Upper arm rotation DOF is harder to accommodate because the upper arm rotates around its long axis. There needs to be a transmission to deliver rotational movement to this DOF. [84] uses a linkage design while [armin] utilizes a circular track and belt transmission. We have decided to use close loop ball bearing and a belt transmission. Major advantages of our approach over others are: 1. More compact structure, 2. Lower friction, and 3. Larger range of motion and 4. easy to switch.

The following figures shows our cylindrical SEE design. The general idea is similar to the disc-shape SEE. Because of constraint in space, we can’t make the diameter too big. So we have decided to take advantage of the space in axial direction. There are two coaxial discs with anchor points at around the outer diameter which the springs are attached to. At zero position, all the springs are parallel to the axis and there’s no torque at all. When given a certain deflection angle, the springs will form a helical shape, and each of them will provide torque. The figure below shows the detail.

![Figure IV.5. 3D geometry of the cylindrical series elastic element](image_url)
Spring length $L$ is given by:

$$ L = \sqrt{H^2 + (2R \sin(\frac{\theta}{2}))^2} $$ \hspace{1cm} (4.6)

Spring force $F$ is given by

$$ F = k(L - L_0) = k(\sqrt{H^2 + \left(2R \sin\left(\frac{\theta}{2}\right)\right)^2} - L_0) $$ \hspace{1cm} (4.7)

To calculate torque around the real axis of rotation, we need to find the projection of $F$, $f$, on the disc plane.

$$ f = F \frac{2R \sin\left(\frac{\theta}{2}\right)}{L} = k(\sqrt{H^2 + \left(2R \sin\left(\frac{\theta}{2}\right)\right)^2} - L_0) \frac{2R \sin\left(\frac{\theta}{2}\right)}{L \sqrt{H^2 + (2R \sin\left(\frac{\theta}{2}\right))^2}} $$ \hspace{1cm} (4.8)

And the effective moment arm $h$ is given by

$$ h = R \cos\left(\frac{\theta}{2}\right) $$ \hspace{1cm} (4.9)
The torque for each spring is then given by:

\[
T = fh = F \frac{2R \sin(\frac{\theta}{2})}{L} = k \left( \sqrt{H^2 + \left( 2R \sin(\frac{\theta}{2}) \right)^2} - L_0 \right) \frac{2R \sin(\frac{\theta}{2})}{L \sqrt{H^2 + (2R \sin(\frac{\theta}{2}))^2}}
\]

To achieve desired and adequate SEE stiffness, as well as to make the design more compact, we have designed a gear ratio of \( R = 4.18:1 \). The following figure (figure 4.6) shows the cable transmission. The left circular track is integrated with the upper arm bearing and when in use, subject’s upper arm will pass through the bearing. The small pulley on the right represents the SEE output. The teeth ratio is 92:22.

![Figure 4.6. Geometry of pulley and belt system for upper arm rotation.](image)

With this gear ratio \( R \), the torque in the SEE \( (T_{SEE}) \) and the output torque felt by the human \( (T_{out}) \) has a relationship of

\[
\]
The deflection angle of SEE ($\theta_{SEE}$) and output deflection angle $\theta_{out}$ has a relationship of

\[ \theta_{out} = \frac{1}{R} \theta_{SEE} \]  

4.12

Stiffness of the SEE ($k_{SEE}$) and output stiffness ($k_{out}$) has a relationship of

\[ k_{out} = R^2 k_{SEE} \]  

4.13

From the equations above, the overall stiffness is amplified by a factor of $R^2$. This makes it less demanding when designing the cylindrical SEE. However, the deflection angle $\theta$ is decreased by a factor of $R$, and we then need to make sure the SEE has enough deflection angle.

The following figure illustrate the SEE stiffness and output stiffness. The targeting output torque is larger than 5 Nm, with a deflection angle of 7 deg. So the SEE has to have a max deflection angle of around 30 deg. For the SEE, if the deflection angle is too big, the spring will start to tangle, which may result different torque profile or a permanent deformation of spring. 30 deg deflection is safe in our prototype.
4.3.3 Gravity compensation

Human arm’s weight, as well as the structures weight will be applied to shoulder flex/extension axis. We came up with a gravity compensation mechanism to counter balance the effect of structural weight.
Figure 4.8. Gravity compensation iteration of shoulder flex/extension joint.

To balance a single 1 DOF joint, the simplest design will be to add a zero rest-length between the upper arm frame and the shoulder frame\(^97\), as in figure a.

In this scenario, the potential energy of the system will be:

\[
U = mg\sin(\vartheta) + \frac{1}{2}k[a^2 + b^2 - 2absin(\vartheta)]
\]

Where \(k\) is the spring stiffness, \(m\) is the overall mass, \(L\) is the length from center of mass to the joint, \(a\) and \(b\) are linkage length. A very important assumption here is that the spring is a zero rest-length spring.

If we carefully choose \(k\) such that \(mgL = kab\), then the above equation will always hold, regardless of \(\vartheta\).

With the hypothesis above, we can then work on the geometry. Figure a is not applicable to our design because of parts and human interference. A second iteration will be to extend the upper arm frame backward and create a parallelogram with two more extra linkages, as depicted in figure b. One problem with this design is the difficulty to achieve a zero rest-length spring. Also, another problem with this design is insufficient torque capacity. We hope to make the design clean and compact so the linkage length is limited. With this constraint, we will have to use a super stiff extension spring, which is not available. Alternative design is in figure c, which accommodates
for easier strategy to achieve zero rest-length, but still has the same problem of insufficient torque. The next iteration is shown in figure d, which is similar to figure c, but the spring is relocated in the upper space. The tension in the cable is transmitted to the parallelogram via a pulley. This way a zero rest-length spring is available, and there is enough space to accommodate the elongation of the spring.

The d configuration is perfect if all the forces can pass the center of each pulley and joint. However, it is not possible due to the diameter of the pulley. No matter how small the pulley is, there will always be a remaining torque due to the diameter. This motivated me to come up with a modified string path, as shown in the figure below. Compared to figure d, there are two extra bearing/pulley in this configuration. Given that bearing cannot take moment, I assume the tension in the string is the same everywhere. The combined force will then pass through the center of each pulley and joint, making it a perfect gravity balanced system.

![Diagram](image)

Figure 4.9. Finalized gravity compensation mechanism.
4.4 Prototyping

All the custom parts are CNC machined with 7075 aluminum. Secondary process is necessary for some parts, such as bearing press fit and shaft fit. Gears for encoder are laser cut from 3mm thick Delrin sheet.

Some custom shafts are ordered through Misumi USA. All the fasteners and bearings are off-the-shelf parts.

For the cable transmission, High Torque Drive (HTD) timing belt is used and the teeth pitch is 5 mm. The distance between the two wheels has an optimal value, based on the compactness. The length of the timing belt is chosen such that there’s a little bit slackness in the transmission. In our initial assembly, the slackness seems neglect able. If in the future teeth jump occurs, we have the option to add idle wheels to increase the tension in the belt as well as wrapping angle on the small timing pulley. Shaft mount slot is already machined on the part and installing a shaft and idle pulley can be done easily.

Alternative to timing belt transmission, we have also considered gear transmission. Gear transmission can be more compact and robust. Gear on the SEE end is available in variable sizes. The problem is with the gear on the bearing end. A custom gear with large cutout and bolt hole is desired. Such a bearing is hard to machine and several manufacturer have difficulty machine it. A timing belt pulley, however, is easier to machine. To accommodate for CNC small radii requirement, the teeth on the pulley can’t be too small. Thus a pitch length of 5mm is chose.

For upper arm rotation bearing, we used similar concept as described in chapter 3. Custom bearing inner ring and outer ring are CNC milled with aluminum. And to bear larger load, the
bearing consists two rows of balls. Undersized plastic balls are used as spacers in between every two steel balls. The assembly of the bearing requires the use of gel-like grease. The grease not only serve as lubrication agent in the operation, but also function as sticky gel to help keep the balls in place. This is practical during assembly and disassembly. Shielding may be necessary in the future to resist dirt, but it is a minor concern.

Figure 4.10. Assembly of custom angular contact ball bearing.
4.5 Future work and conclusion

For this shoulder module, future work includes tuning the controller, bench top testing, strap design, mounting frame design, etc. Also, to make the shoulder module and elbow module work together, an interface part need to be designed.
5 Conclusion

In the dissertation, three types of exoskeleton is presented: a high DOF passive hand exoskeleton (HandSOME II), a 2 DOF active elbow exoskeleton (ULIX), and a 3 DOF active shoulder exoskeleton.

HandSOME II showed a positive effect on subject’s finger range of motion and enhanced their performance in their picking tasks. After iterations, a more user friendly and robust HandSOME II is developed. Data logger is also added to it. HandSOME II is now being given away to patients for their home training. And their finger movement data will be studied later.

A two DOF elbow exoskeleton (ULIX) is designed and prototyped. The novelty include a Bowden cable driven series elastic actuator and a cable driven mechanical differential. A controller was designed and tuned. Bench top test was performed. The overall performance is satisfactory. A shoulder exoskeleton will need to be designed and built to work with ULIX.

A three DOF shoulder exoskeleton is designed and prototyped. The actuators are directly located beside the actuated joints. Series elastic element is used to obtain a backdrivable actuator with accurate torque control capability.

Both elbow and shoulder exoskeleton utilized a custom SEE. The SEE’s stiffness is adjustable by using different types and number of tension spring. The SEE stiffness can be predicted given the tension spring’s parameter. The adjustability of our SEE allows us to try different kind of stiffness when tuning the physical human-machine interface.

Future work includes combining the three exoskeletons described above.
6 References


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