Assistive Rehabilitation Robotic Glove

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Abstract

Two novel designs for a rehabilitation robotic glove are detailed. The paper focuses of the design and implementation of these devices. The main aim is to explore lightweight, low profile gloves that can facilitate movement of fingers that have lost strength or nerve control. This has been by achieved by using an exoskeleton to provide the actuation for flexion, thereby limiting the number of active actuators required. Pneumatic muscle actuators were used to provide flexion force, and force sensors used to supply control inputs by the user. Control methods have been implemented that would help lead to the development of rehabilitation programs. The glove can also be used as an assistive device in order to allow the patient to grasp and pick up objects. Prototype I, with 1 DOF, was found to successfully actuate a hand in order to grasp a shape similar in size to a golf ball. Prototype II, with 3 DOF, achieved actuation of a single finger from fully flexed to at rest elongation.

1 Introduction

A healthy, working hand is vital to perform everyday activities. Any loss in the ability of function of the hand to complete these activities can drastically reduce one's independence, often resulting in limitations of work, social and family interactions [Abolfathi, 2008]. Many injuries, illnesses and defects can lead to the loss of hand function, including trauma, peripheral nerve injury, stroke and arthritis. The incidence of the latter of these is highly dependent on age and obesity, and for aging populations such as Australia's, the incidence of hand disability will only increase. Currently 20% of the world population is aged 65 and over, and it is predicted that by 2050, 35% will be over this age [Bogue, 2009]. Given that the incidence of stroke doubles every decade after the age of 55 and that age-related diagnosis including arthritis and orthopaedics account for 70% of physical therapy demand [Miller, 2007], the demand for devices providing rehabilitation and supplying the source of movement in the hand will also increase. With the paradigm of an aging population, assistive robotic devices will become more prevalent in the future as the demand increases. Specifically, devices which augment a user's hand functionality will be required, as many age related diagnosis result in the partial loss of strength or control of the hand.

In the last three decades there has been a shift towards creating assistive robotic hands that can augment a user's motion, or create motion where necessary. These robotic devices are seen as critical technologies that will allow people living with disability to retain their independence and maintain an active lifestyle. There has also been an increasing trend to design rehabilitative devices that help a patient regain their strength and control. The idea behind rehabilitation is to reduce impairment and increase the range of motion by encouraging the body to heal itself. This is often achieved through repetitive motion which ultimately can relieve pain, inflammation and swelling. This joint motion under a specified program with supportive structures implemented as required has been clinically shown to increase the total range of motion [Abolfathi, 2008]. Supportive structures are often required to restore injured nerves by relieving the pressure placed on them [Fu *et al.*, 2007].

Current implementations of rehabilitative robotic devices include those that allow impaired joints to move individually and simultaneously to train for specific movements, [Xiong *et al.*, 2009] for instance to grasp a door handle and eat. These systems require measurements of position and torque of the fingers and often apply resistive torques to increase a patient's strength as the rehabilitation process develops. This operation, known as patient-passive-robot-active rehabilitation needs to take into account physical behaviours, such as spasm, to reduce the chance of further injury. To achieve this, the angular movements and forces produced are kept within safe limits. Using a robot in the rehabilitation process can achieve all of the following;

- Movement with no human interaction (joint mobilization)
- Force-based control (force enhancement, muscle training)
- Mixed force and position control (guided movement)

Being able to use a robot, rather than a human being, to provide these training regimes means independent rehabilitation can easily occur, leading to faster and more productive recoveries as rehabilitation could occur at the user's home in their own time. Rehabilitation in this manner can produce faster results at a lower cost [Wege and Hommel, 2005] as patient's are able re-enter the workforce after less time and do not require constant physiotherapy sessions with a physiotherapist, but instead could use their own assistive robotic device to produce the same results.

A device which could produce patient-passive-robot-active and patient-active-robot-passive control would be useful for patients requiring rehabilitation support and power augmentation. However; few devices exist which can allow a person to augment their complete hand motion during everyday activities. This is due to both limitations in size and weight of the device, as well as problems in achieving the same complex manoeuvres the human hand can exhibit. It is necessary that future implementations of these devices address these problems, in order to provide a rehabilitation platform for the hand, but also act as an adequate source of movement. Coined a power glove, these assistive robotic device will be vital in the future.

The success of a power glove is dependent on the underlying technology it utilises. With advances in material science, dynamic actuators have been realised which realistically mimic the actuation of human muscles when given a stimulus. Combined with sensory and control systems, a lightweight, low profile mechatronic device can be created to assist in hand functions, ideally achieving degrees of freedom of movement similar to that of a human hand.

2 The Human Hand

The human hand is one of the most complex mechanical systems of the body, the fingers and thumb alone account for twenty degrees of freedom, nineteen muscles, seventeen joints and twenty bones [Kolyvas *et al.*, 2005]. It is extremely dexterous, stable and precise but is also strong and flexible and able to complete fast, complex manoeuvres. Roboticians have tried to model the human hand for over five decades by patterning the movement of these devices after the human hand; [Becker *et al.*, 1986] however have been limited in the complexity of motion achieved as well as the devices available to provide the actuation.

The muscles in the hands can be loosely divided into two groups, intrinsic and extrinsic. The intrinsic muscles are located around the palm and fingers, and are responsible for fine movements. The extrinsic muscles are remotely located in the wrist and forearm, and are primarily responsible for the grasp and pinch actions which require more strength. Fingers consist of three joints. From knuckle to finger-tip these are the metacarpophalangeal (MCP) joint, proximal interphalangeal (PIP) joint and the distal interphalangeal (DIP) joint. Each of these is capable of extension and flexion and the MCP joint also exhibits ad/abduction DOF. Specifically, the extension and flexion of the fingers are achieved by a 'co-joined tendinous structure' [Brown *et al.*, 1993] that involve cooperative interactions of intrinsic and extrinsic muscles involving multiple tendons. Every human hand exhibits tremor, though the type and severity may differ.

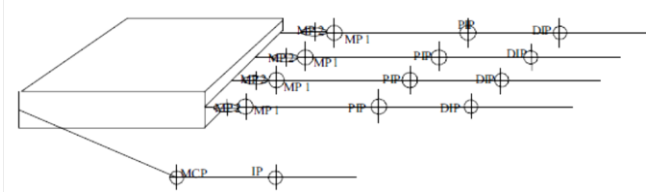


Figure 1: Human hand physiology [Fu *et al.*, 2007]

The range of each of the joint angles involved in extension/flexion in a healthy hand is limited to 0-90 degrees for the MCP and DIP joints, and 0-110 degrees for the PIP joint. For the abduction/adduction motion, the MCP joint can range between 0-15 degrees and is approximately 0 degrees for the thumb and middle finger [Saliba and Axiak, 2007]. The forces that can be applied by these joints are summarized in the following table, and have been evaluated from average data gathered from healthy male hands at the midpoint of each finger section between two joints [Brown *et al.*, 1993].

Motion	**Force (N)**
Finger DIP joint	3.92
Finger MCP and PIP joint	8.82
Thumb PIP joint	7.84
Thumb MCP joint	5.39
Thumb CMC joint	6.86

Table 1: Average force outputs for male hand [Brown *et al.*, 1993]

3 Existing Solutions

Many robotic hands have been developed that display similar DOF to human hands. Notable examples include the Shadow Hand which is a compliant, 25 DOF hand and wrist that uses pneumatic muscles as actuators [Kochan, 2005], the Otto Bock Healthcare prosthetic which uses nerves in the chest as control inputs, the MPL Device known as "Luke" Arm which has 22 DOF and control via a neural interface [Pavlou and Conyers, 2010] and the iLimb hand, which uses electric motors to individually power each with control from myoelectric signals [Bogue, 2009]. Shape memory alloys, pneumatic muscles actuators and electric motors are examples of actuators that have been used to successfully implement these fully actuated hands.

Whilst many exoskeleton assistive robotic gloves have been developed over the last four decades, none have achieved independent motion of every DOF on each finger and thumb in a physical package that would be easy for a user to carry around and use for everyday tasks. There are two main reasons for this; the first is that the complete motion of a human hand is very complex. To achieve the bidirectional motion of each joint in the hand, a minimum of sixteen actuators would be required, assuming that the flexion/extension DOF of each of the finger's PIP and DIP joints are coupled. The second reason is that the physical space the actuators would require is very large, even if remotely located, and so limits the weight and profile targets that can be achieved. However, many devices have achieved successful systems by simplifying the DOF required, whilst still being able to reproduce much of the hand's functionality and not impinge too greatly on the user's hand profile.

3.1 Choice of Actuators

Exoskeletons using electric motors as actuators are a common method of mechanical implementation. This technology is well proven and reliable however generally results in bulky solutions that use pulleys and levers to transfer the actuator movement to the fingers.

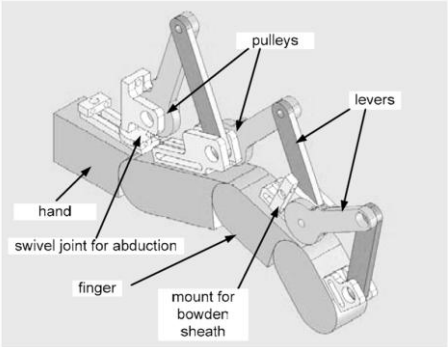


Figure 2: Mechanical finger glove using pulleys, levers and electric motors [Fontana *et al.*, 2009]

Recently there has been a trend to use smarter materials as actuators, as these can often provide more lightweight solutions. A two DOF finger glove using shape memory alloy (SMA) actuators was theoretically developed and achieved a low mass solution [Kolyvas *et al.*, 2005]. When using this method of actuation, isolation of thermal and electrical conduits is vital to allow for safe operation for a human-robotic device. These actuators are extremely light, small and can have high power to mass ratios. One major disadvantage of their current implementation is the hysteresis shown in their operation. For rehabilitation, where predictability of movement is vital for controlled movements, this can be a major problem.

Pneumatic muscle actuators (PMA) are another relatively new technology that has been used in applications where a high power to mass ratio is required. They have added benefits of safe and quiet operation, low maintenance and can be used in aquatic or liquid environments which may prove vital when creating a device that can be worn continuously by the user. Another advantage for the use of PMAs in human-robotic device is that they act in an analogous manner to a human tendon. [Xiong *et al.*, 2009**].** PMAs generally have 30% contraction, which is typical of human muscles. However, the contractile force for a given cross-sectional area can be over 300N/cm^2 for PMA compared to 20-40 N/cm^2 for human muscle, meaning they can be used to create power assistive devices [Tsagarakis *et al.*, 1999].

3.1 Choice of Sensors

There are many sensors available to measure the tactile action of the human hand and hence gain command inputs for the operation of the power glove in order to parallel the natural movement. The existing solutions to this sensor mainly fall within two categories: sensors that measure the pressure or force directly off the fingers and those that measure bioelectric potential through surface electrodes, giving electromyography (EMG) signals.

Direct measure of force or pressure uses contact between the human finger and a contact surface to produce a resistance that is a function of the applied force or pressure [Xiong *et al.*, 2009; Shields *et al.*, 1997; Wang *et al.*, 2009]. These sensors are generally in the shape of thin films which can run along the finger and take inputs from a small and specified contact patch, usually coincident with the fingertip [Kawabuchi, 2007]. Other implementations include those of rubber-based transducers, consisting of pressure sensitive conductive rubber or piezo-rubber, [Yamada *et al.*, 2005] or alternatively strain gauges can be implemented [Shields *et al.*, 1997]. In the same manner, these generate a changing resistance based on the applied force. All these tactile readings have been shown to be unreliable when based purely on a small contact patch, but reliable enough to measure dynamic changes [Wege and Hommel. 2005]. The accuracy of the system can also be improved by fusing this data with that gained from the tension of transmission cables or the current and torque of the motors (if used) in order to estimate the applied force at the finger [Wege and Hommel. 2005].

Measuring the EMG signals of a part of the body means inherent movement commands of the human can be intercepted and used to control motion of the robotic hand. This has two main advantages, the first being that the same commands sent to move a specific part of the hand can be used to command the robotic counterpart, producing parallel sensory inputs [Bogue, 2009; Hasegawa *et al.*, 2008]. The second is that if movement of the hand is severely hindered and if nerve damage has occurred, a separate part of the body can be monitored in order to measure human based control commands. Most disabilities caused by stroke are hemiplegic, only affecting one side of the brain, so a normal healthy hand generally exists [Kawasaki *et al.*, 2007]. This could also be true of force and pressure sensors and is the basis behind haptic devices. The disadvantage of a system such as this is that it can be harder for the patient to learn to control the hand especially as the only feedback is from the user changing their EMG signals; it is not a closed loop system [DiCicco *et al.*, 2004].

4 Structural Design

An assistive power glove, used for rehabilitation or everyday wear, has to be able to supply adequate DOF in order to augment a user's motion and act safely within known limitations of the workspace. As well as this, the structure needs to have dimensions suitable for various size hands, be mounted on the dorsal (back) side of the palm in order to avoid interference with palm space and be worn comfortably and easily. An ideal system would be modular and able to be adapted and attached to deformed or scarred hands, in order to suit the individual patients' needs. Rehabilitation can be undertaken for seriously to minimally affected hands and fingers, and so the force a device exerts needs to be tailored for the individual's situation.

There are many safety considerations that should be incorporated into a successful power glove design. These include mechanical stops to limit the angular range of motion of each joint, emergency cut off switches and software limits for hand positions and forces. Given that the user will be wearing the exoskeleton over their limb, the structure also needs to be designed to be safe if the user falls over and impacts the glove.

For all these reasons pneumatic muscle actuators were chosen. These actuators look promising in their use for systems that mimic the natural movement of hands. Whilst they may need to be quite long in order to achieve high enough contractile lengths, the simple nature of their

actuation means they can be placed remotely off the user's hand, minimising the mass on the user and the total profile of the glove.

PMA actuators are uni-directional in their movement, they can only pull, not push. They were used to open the user's fingers by supplying the extension DOF of the joints. In order to keep the palmer side of the patient's hand free, this meant a restorative force was required for the flexion DOF. If this force could be provided by the mechanical structure of the exoskeleton itself, it would halve the required number of actuators, greatly decreasing the mass and profile of the system. Both designs presented in this paper use PMAs for the extension force of the finger joints. The first prototype was designed to analyse whether a moulded composite material would be adequate in supplying the flexion force, and the second prototype tested whether restorative torsion springs would be feasible. In both designs, flexible force sensors were used to gather control inputs from the user and gain tactile information. These sensors are ultra thin and flexible printed circuits that produce a resistance that is proportional to the contact force applied to the collection pad. Force sensors capable of measuring a maximum contact force of 1 pound were used, and with integration into an amplification circuit, gave output readings between 0 and 5 Volts.

4.1 Prototype I

The critical design point for the exoskeleton structure was the ability of the material to have a sufficiently high yield strength to accommodate the elongation needed for a fist to change from a fully extended to fully flexed position. The yield strength represents the point at which a material acts in plastic deformation rather than elastic deformation. For an exoskeleton to have high enough fatigue strength to perform a useful number of actuation cycles, it must remain in the elastic deformation region.

The required elasticity would differ for different dimensioned hands. The engineering strain required in the structure to accommodate position changes for the test subject was found by measuring the total distance from fingertip to wrist along the back of the hand. It was found a 9.7% elongation for a change from a fully extended to fully flexed position was required, and 4.1% for a change from an at rest position to fully flexed position. For this last change, a minimum of 8mm of length displacement from wrist to fingertip was necessary. These hand positions can be seen in Figure 3. This elasticity would need to be able to be provided by the exoskeleton structure itself; and this property would be determined by the composition and thickness of the structure.

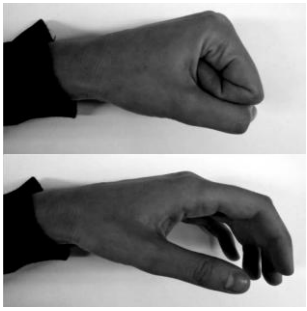


Figure 3: Fully flexed (top), at rest (middle) and fully extended (bottom) positions of test subject's hand

When manufacturing a composite structure, different layers of composite material are joined together using an epoxy resin. In isolation, these resins can only act in elastic deformation for up to 6% strain [Andrews and Stevenson, 1978] and so introduced the first limiting factor to a composite design exoskeleton. However, the movement from fully flexed to the natural rest position of the hand can still be achieved by an epoxy resin structure, and this was adequate to test the worthiness of the composite design. When performing simple manoeuvres such as picking up the handle of a mug or pen, grasping a door knob or holding a bar, the hand does not extend open more than the at-rest position, and if the power glove could assist a user to perform these tasks, then it still has relevance as a useful tool.

Fibreglass, Kevlar and carbon fibre composite materials were used to create exoskeletons in the shape of a cylinder or triangle surrounding the joint geometry of a human fist. Multiple structures were made with combinations of different numbers of layers, and epoxy was used to create the matrix. These are shown in Figure 4.

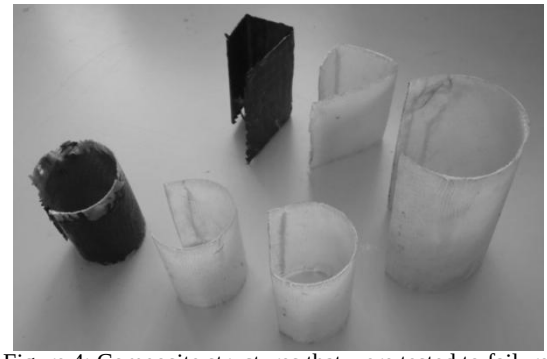


Figure 4: Composite structures that were tested to failure

Each was tested by stretching open the cylinder until failure occurred in order to demonstrate the forces that could be sustained and to see what displacement this occurred at. These are summarised in the table below.

Shell Structure	**Max. force at deformation (N)**	**Displacement (mm)**
3 layers of fibreglass (small radius)	19.6	20
6 layers of fibreglass (small radius)	34.3	44
6 layers fibreglass (large radius)	49.0	10
Carbon/Kevlar/carbon structure (small radius)	34.3	5
4 layers of fibreglass (triangular shell)	14.7	8
4 layers of carbon fibre (triangular	34.3	4

shell)		

Table 2: Results of the composite structure testing

Based on these results, a rounded 5 layer fibreglass composite, with a thickness of 1.5mm, was chosen as the exoskeleton structure. Fibreglass is a cheap composite material that is able to be easily utilised to manufacture various products. This was an advantage, as it meant prototypes could be readily made, and if introduced as a product, would be cheap to produce. Another advantage was the behaviour of fibreglass when it was subject to an impact force. Unlike carbon fibre which can shatter, producing splinters on impact, fibreglass structures tend to fracture within the epoxy layers, keeping the user safe from any sharp points on the surface. Using fibreglass satisfied the design criteria that the exoskeleton must be safe to wear in the event that the user falls over.

A CAD model of the device was developed and depicted in Figure 5. Separate sections were used for the forearm, on which the actuators, feedback sensors and other hardware were located, and the glove over the hand which was uninhibited except for the Bowden cable and flexible force sensors. A plastic bearing was used to join the two sections in order for the user to maintain the natural motion of their wrist. A linear variable voltage transducer (LVDT) was used as the feedback sensor for the position, hence the displacement, of the PMA. One DOF was utilised in this prototype, i.e. supplying flexion and extension for all the fingers together, as this was adequate to test the concept.

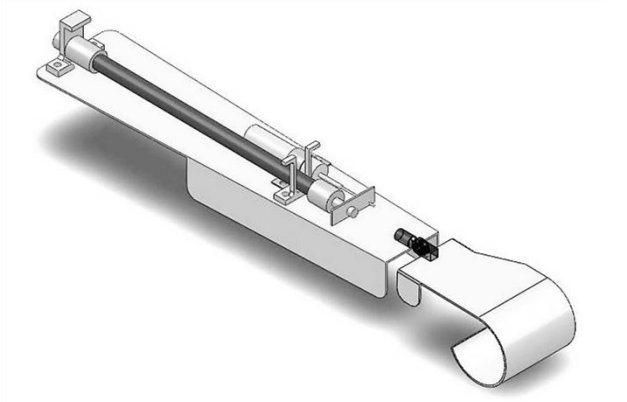


Figure 5: Prototype I CAD model. Bowden cable and force sensors not depicted

4.2 Prototype II

Instead of using a composite exoskeleton to provide flexion force, this device tested whether restorative torsion springs could be used. The design was extended to three DOF; extension and flexion of the MCP and couple DIP/PIP joints of one finger. The extension force of each of these was provided by two PMAs. These were located above each other, so that there was little angular difference between the movement of the PMA and the extension of the exoskeleton, and remotely from the palm of the hand in order to minimise the mass. The CAD model of this prototype can be seen in Figure 6. The wrist DOF has been removed from this design in order to isolate and test only the DOF of the fingers.

The exoskeleton is constructed from four sections. These are the plate from wrist to MCP joint, structure from MCP-PIP joint, structure from PIP-DIP joint and structure from DIP-fingertip. Each was connected by a torsion spring. Instead of using a steel Bowden cable to connect the PMAs to the exoskeleton, twisted Kevlar thread was used. Kevlar thread is very light but has a high tensile strength, making its use desirable in a hand exoskeleton. It was necessary to encase the torsion springs in a plastic bearing in order for the Kevlar thread to run easily over the top of the joints.

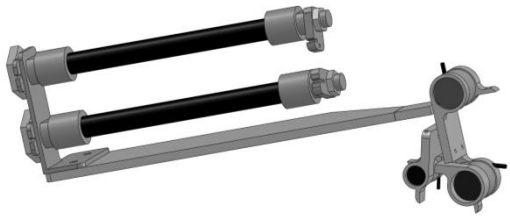


Figure 6: CAD model of prototype two showing three torsion springs and two PMAs used as the actuators. Bowden cables and sensors are not depicted.

The exoskeleton was manufactured from mild steel using a CNC machine in order to satisfy the tight tolerances of fits for the torsion spring placements. The springs were also used as the hinges between joint sections, and the end rods of each spring were held in place by securing holes. Mechanical stops were placed on each section, in order to create a physical barrier to eliminate overextending the fingers. The geometry of this design can be seen in Figures 7 and 8.

Figure 7: Components of the exoskeleton finger. Three torsion springs hold each section together. They are supported by inner spaces (shown above springs) and are encased by plastic bearing sleeves

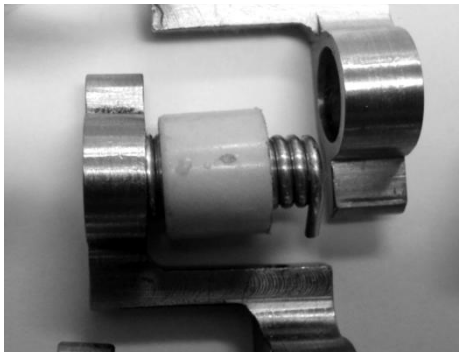


Figure 8: Close up of DIP joint showing spring held by PIP-DIP exoskeleton structure (left) and separate from DIP-finger joint structure (right)

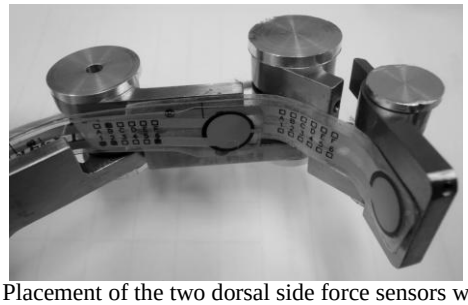


Figure 9: Placement of the two dorsal side force sensors which sit along the back of the finger.

Flexible force sensors were used as the control sensors in this device. Two were used for each joint, one to control flexion and the other two provide extension. This meant one sensor was placed under the fingertip and MCP-PIP joint, on the palmer side of the user's hand, which was not ideal. However; as these sensors are very light, small and thin, it was considered not to be a major hindrance to any task the user may do with their hands. For feedback on the length and state of the PMAs, flexible conductive rubber was used, replacing the much heavier and bulky LVDT used in Prototype I. These sensors produce a resistance linearly dependent on their length, and so can were connected to the two ends of the PMA in order to provide displacement information.

5 Control Algorithm

For rehabilitation devices, the control is vital to ensure successful programs can be implemented, and to make sure no harm comes to the user from wearing the device. One major issue that needs to be taken into account when designing the control system is the user's tremor. In case of disability or injury, hand tremors can be magnified and can often cause spasmodic events. A control system needs to be able to handle these, and to have a method of filtering tremor forces from command forces.

There are two main methods of controlling such devices. The first is through force and position control, which may include the use of a PID controller on the actuators used. This force and position control can also be used to monitor spasmodic events that are both instant and sustained and act on them in a suitable fashion. For example, a controller can measure when the torque at any of the joints exceeds the torque spasm limit that was previously set, and time its duration. If the duration of the spasm is less than a threshold of two seconds, the hand can wait in its current position until the spasm stops. However, as soon as the time threshold is exceeded, the exoskeleton hand can rest, allowing normal human motion to occur [Xiong *et al.*, 2009]. This is a safety feature that will limit the chance of further harm coming to the user.

The next trend is to use a binary controller to actuate the system as required until the input signal indicates a change of state of the operation. This is well suited to EMG control signals, and can also be implemented for force control signals. The algorithm used for both prototypes presented in this paper was a state-based system where the system states were defined by all the actions the glove could perform. Specifically these were;

- Hold (steady pressure)
- Flex (release pressure)
- Extend (increase pressure)
- Stop (release all pressure)

The states were defined by the sensor readings for force inputs and feedback from the position of the PMAs, and the active state defined the input command to the pressure regulator. Figure 10 shows the hardware diagram for the device.

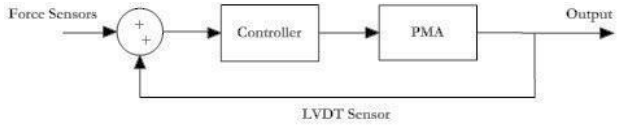
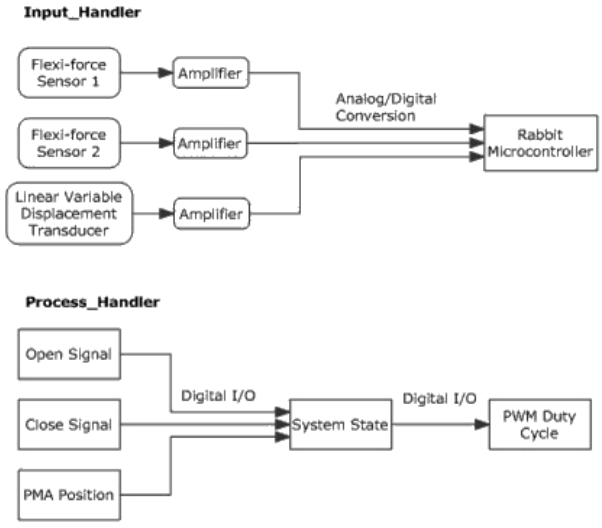


Figure 10: State controller for Prototype I

The following is an example of how the control system operated. For an extend state to be triggered, the user was required to apply a force on the 'extend' force sensor (placed on the back of the dorsal side of the hand) that exceeded a threshold set by the user initially with calibration of the system. This force needed to be applied for a specific amount of time, also set by the user, in order for the system to recognise it as a control signal, and not the product of a tremor. As a safety measure, software stops were placed in the program to ensure no command was activated to actuate the fingers beyond the safe working extension of the finger. Having these forces and limits as calibration variables meant that programs could be updated and changed as the patient's situation changed. This is required for a rehabilitation device, as a physiotherapist may implement different programs that required different levels of movements as rehabilitation progressed. When connected to a PC and acting in factory mode, a menu would prompt the user for calibration if required, and allow a therapist to adjust modes. Figure 11 shows the complete program flow of the device.



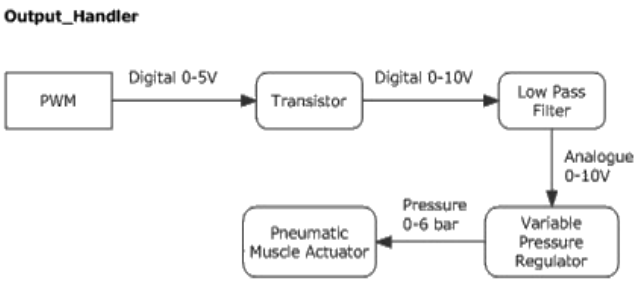


Figure 11: Program flow for prototype I showing inputs from the force and feedback sensors (top), processing of signals to determine system state (middle) and output signal to the PMA in order to produce the required movement (bottom)

The PMAs were controlled by a digital variable pressure regulator (VPR) that increased or decreased the working pressure of the PMA, which was supplied by a paintball air bottle. The control signal to the VPR was filtered in order to ensure smooth operation of the PMA, as any unexpected changes in the control signal would produce unexpected movements of the exoskeleton. A control bandwidth of 40Hz was achieved for the PMA actuation, which was sufficient to create predictable movements for the user at a fast enough pace to allow for user reactions to be recognised and acted upon.

6 Test Results

6.1 Ergonomics

Rehabilitation devices need to be comfortable for a user to wear, especially as it may be acting on sensitive or damaged areas. One way to increase comfort is to have a good correlation between the centres of rotations of both the exoskeleton and the human finger. This was not perfectly achieved with either prototype. Given that prototype I only had 1 DOF it was not possible to match each centre of rotation. For prototype II this was also a problem, though to a lesser extent as more DOF were present. The torsion springs were placed directly on top of the finger joints, and so with non-concentric centres of rotation, parts of the finger could be uncomfortably stretched upon extension. To match the centre of rotation of the torsion springs with those of the finger joints, thin torsion springs could be placed on either side of the joints. This would also act to decrease the vertical profile of the exoskeleton, whilst still maintaining restorative flexion force. This method could not be adopted for prototype I. The problem could be minimised with more DOF, and with compliance placed at the joint locations, allowing more flex in the fibreglass at those points. Structurally, this would be difficult to achieve without increasing the profile of the glove.

Future tests will need to be undertaken with the involvement of physiotherapists in order to determine the comfort of the glove when tested of users with various levels of disability.

Both structures exhibited a low profile over the fingers they were actuating. The profiles for each are shown in Figures 12 and 13.

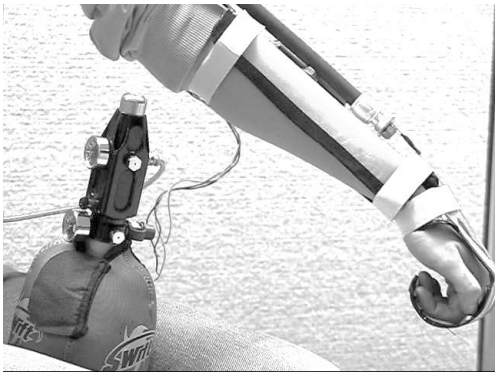


Figure 12: Profile of prototype I. The profile on the user's hand is very low, due to the remote placing of the PMA and LVDT

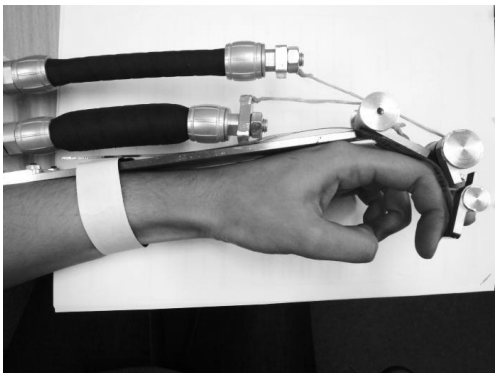


Figure 13: Profile of prototype II. As the wrist DOF has been removed, the profile on the dorsal side of the palm is larger than that of prototype I, yet the profile over the user's fingers is still quite small.

Besides having a small profile, it is necessary that these devices have low mass in order to be worn easily, especially as they may have lost strength in their hand. The masses of each device have been summarised in Table 3.

	Mass on Hand (kg)	**Total Mass (kg)**
Prototype I	0.24	2.17
Prototype II	0.55	2.70

Table 3: Mass of each prototype

Prototype I has a much lower mass than prototype II, an advantage of using fibreglass over a metal exoskeleton. The majority of the mass of each system was comprised of the air bottle and pressure regulators. The system design allows these components to be placed remotely off the user's arm. A future development may be to place these in a backpack or bag that could be worn by the user with minimal hindrance to their arm and hand. Mild steel was chosen as the material for the exoskeleton of prototype II to ensure there was no flex in the structure. A large factor of safety was used in the design-theoretically the exoskeleton was able to sustain much more force than was applied on it before deformation. However; this allowed the effectiveness of the torsion springs as a method of supplying flexion force to be analysed independently of any system deformation. Future prototypes could be made from composite materials, or a lighter metal such as aluminium, which would decrease the mass directly placed on the user's hand.

6.2 Tremor Control

Force from tremors exhibited by the users were measured at averaged over different times of day as a user's physical condition can have a marked affect on their tremor. The maximum exhibited tremor was found to be 0.5 volts (V), corresponding to force of 0.45N using the 1 pound force sensors. For the controller to work adequately, this signal needed to be smaller than the control signals or last for a shorter amount of time then the calibrated period. Average readings from the force sensors on both sides of the user's hand have been summarised in Table 4.

State	Palmer Sensor		Dorsal Sensor	
	Volts	Newton	Volts	Newtons
Open	0.21	0.19	2.95	2.63
Close	1.11	0.99	0.10	0.09
Rest	0.41	0.36	1.09	0.97

Table 4: Average force control signals from the flexible force sensors

The minimum difference in sensor readings from any one state to the other was for the palmer sensor between the 'close' and 'rest' states. This c0difference of 0.2V was greater than the magnitude of the test subject's tremor, and so their tremor would not be a strong enough signal to cause an unexpected state change in the control algorithm. A switch between states for the 'close' and 'open' states was dependent on both force sensor reading values, introducing another precaution to stop false state changes as each sensor would have to register high noise signals in order for the change to occur.

6.3 Force Results and Fatigue Life

The effectiveness of the gloves was tested by seeing the range of motion of each device, and the forces that were able to be produced. The limits of flexion and extension of prototype I are seen in Figure 14.

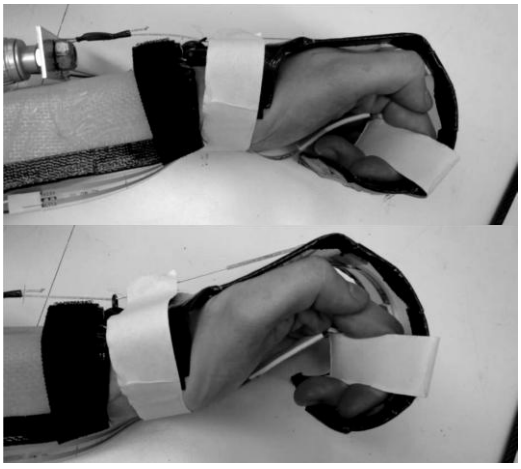


Figure 14: Prototype I at full flexion (top) and full extension (bottom)

The extension of the device was not satisfactory for it to be a useful tool. One mechanical design limitation was the positioning and attachment of the Bowden cable. Ideally friction losses across the cable would be a minimum, however in this design friction has a large impact on the operation of the glove. The cable itself first coincided with the exoskeleton at the knuckles of the hand, and much of the force of the actuator was transmitted to this point, acting to pull the glove back rather than pull the fingers open. The attachment of the glove was also quite close to the end of the exoskeleton curved plate, meaning the cable was likely to bind against the fibreglass in the same manner as a rope wrapped around the trunk of a tree binds into the wood. These characteristics caused much of the actuation force to be transmitted to parts of the exoskeleton other than the desired application point, meaning the glove could not be properly extended. Graphite lubricant did little to address this problem- future developments would require a sheath or cable runner. The final position of the exoskeleton resulted in the user being able to successfully pick up objects with radius of 43mm or less, a shape roughly the size of a golf ball or small door handle.

To test the fatigue life the hand plate was flexed 200 times to the degree stated above without failure occurring, though a small amount of plastic deformation was seen. The magnitude of the plastic deformation was 1.5mm, or 0.8% of the nominal length at full flex position. Slight cracking of the top epoxy layer at this point occurred. It was estimated that with longer fatigue testing failure would occur, and before this happened the magnitude of plastic deformation would increase, making the glove ineffective for continuous use.

Testing of prototype II was met with more success. There were some mechanical elements to the design that needed improvement, however the device was able to successfully actuate the finger from a fully flexed to at rest position. The Kevlar thread did not show any signs of elongation, and there was no fraying or signs of shearing. The positions achieved by this device can be seen in Figure 15 and 16.

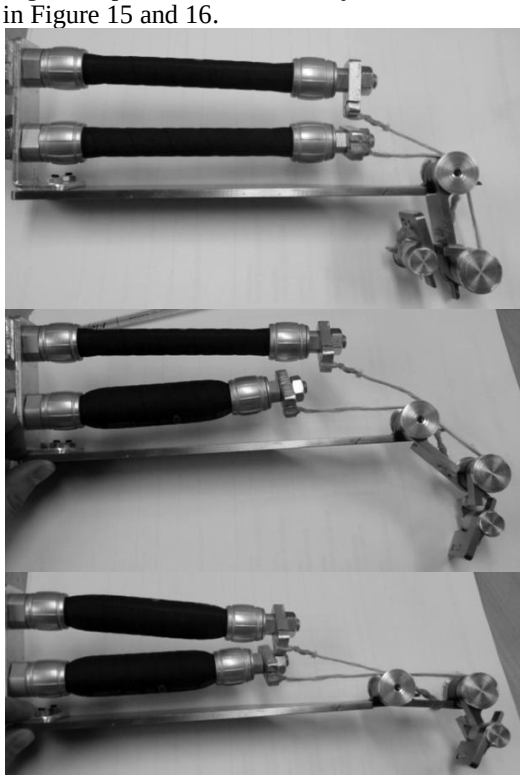


Figure 15: Prototype II at fully flexed position (top), with the couple PIP/PIP DOF fully extended (middle) and with all DOF fully extended (bottom)

In order to successfully couple the movements of the DIP and PIP joints, the Kevlar thread was run underneath the DIP joint then over the PIP joint. This resulted in a required displacement of more than 25mm to fully extend the finger-beyond the limit of the 100mm PMA used, resulting in less than 100% extension. Another limitation that can be seen in the bottom figure of Figure 16 is that the finger joints do not match the centres of rotation of the exoskeleton joints-this is especially prevalent when the hand in fully flexed.

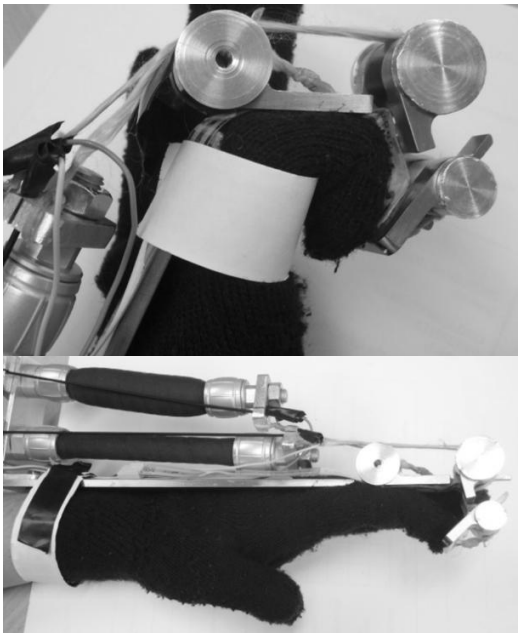


Figure 16: Gloved finger when flexed (top) and when extended (bottom). The force sensors were situated under the exoskeleton and inside the glove.

The restorative flexion force provided by the torsion springs was 1.8Ncm at the MCP joint and 3.5Ncm at the coupled DIP/PIP joint. This was required to initially start extension movement, the torque applied by the exoskeleton structure on the finger increased as the radius of the torsion springs decreased. The user described the force when at full flexion as being similar to what she could exert with a clenched fist.

7 Conclusion

Both prototypes achieved lightweight, low profile solutions that were able to augment the natural motion of the extension and flexion DOF of the fingers. Whilst simple designs that nowhere near met the complexity of movement that is possible by a human hand, each prototype was successfully able to test the concept of incorporating flexion force into the exoskeleton.

It was found that the fibreglass exoskeleton did not have adequate extension in this particular implementation, though theoretically it should be able to extend a hand from fully flexed to at rest position. If this avenue of actuation were to be followed, it is recommended that the design include more DOF and include compliance in the joint locations that allow for more flexibility. Using moulded plastics instead of fibreglass could be one way to achieve this whilst still maintaining the low mass of the device. Prototype II showed that using torsion springs to provide restorative flexion force was adequate in providing extension and flexion of the MCP and coupled PIP/DIP joints of the user's index finger. The finger was able to extend from a fully flexed to a beyond-rest extended position, though full extension was not achieved. Adapting the length of the PMAs used could solve this problem.

The control algorithm worked well for patient-active-robot-passive mode, which is required for a device that augments human motion. Patient-passive-robot-active control was also possible using the system, as defining finger positions under timed movements was possible. Future work should be done in developing generic programs that a physiotherapist could choose for specific afflictions and levels of ability. Whilst not explored in this project, the design can be extended to a haptic device if muscle signals are too weak to be able to control the damaged fingers.

Pneumatic muscle actuators were highly successful in creating actuation that was smooth and of sufficient pace whilst being quiet to operate and simple to implement and control. Remotely locating such hardware largely contributed to the low profiles achieved, and mimicked the placement of large extensor muscles located the forearm that produce grasp and pinch actions of the hand.

These two prototypes demonstrate the effectiveness of using an exoskeleton to provide DOF for the motion can simplify the device, decrease the number of actuators required and minimise size and mass.

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