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Experimental Study of a Shape Memory Alloy Actuation System for a Novel Prosthetic Hand

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1. Introduction

Recently, the development of compact, light-weight and powerful actuation systems has been in the centre of investigation at many scientific institutions and research groups all over the world. These systems can be used in devices of almost every aspect of modern life and based on their inherent technology they come with certain benefits and costs. One of the most demanding applications field in terms of actuator selection and design is the field of upper-extremity prosthetics. Modern commercial advanced hand and arm prostheses are conventionally actuated by electric servomotors. Although these motors achieve reasonable kinematic performance, they have been proven insufficient in meeting amputees' demands, mainly due to their noisy operation and limited energy density which leads to the use of bulky and heavy driving systems (Herr, 2003). Therefore, an alternative nonconventional actuation technology is requisite in order to overcome these limitations which make a substantial proportion of upper-limb amputees avoiding the use of their prostheses.

One of the most promising actuation technologies is based on Shape Memory Alloys (SMA) and phenomena related to change of their atomic structure. SMA are metallic alloys that can exhibit an actuation mechanism resembling the biological muscle - they contract producing actuation forces. These muscle-like actuators present high power to weight ratio enabling the development of compact, lightweight prosthetic devices without too much compromising power capabilities and eliminating the forced-tradeoff between dexterity and anthropomorphic size, weight and appearance (Bundhoo, 2009). Additional benefits include an inherent position feedback method (given a near linear relationship between ohmic resistance and contraction), silent, smooth and life-like operation, and the lack of requirement for force or motion transmission devices (Kyberd et al., 2001).

During the last two decades SMA have been studied and reviewed as possible actuation technology in prosthetics by many researchers but efficiency and response time are claimed as the most limiting factors (Del Cura et al., 2003). So, in order to render this material appropriate for application in upper-limb prostheses, these impediments must be overcome. Towards this scope, an innovative SMA actuation system for a newly developed prosthetic hand is constructed and studied. The technology applied in this hand offers a series of improvements when compared to current commercial prosthetic devices. Its design

proposes the development of a multifunctional, silent and cosmetic appealing hand that will enable patients to carry out basic daily tasks more easily and reduce the rejection rate of prostheses.

In the next Section, a brief background of this research is provided. The development of the hand prototype along with its SMA actuation system will be analytically described in Section 3. In Section 4, the performance of this actuation system is assessed for a one-finger testbed while in the last section conclusions and future work are discussed.

2. Background

2.1 Prosthetic Hands

The human hand plays a vital role in every aspect of our daily life as it defines our interaction with the material world. Its functional uniqueness arises from the complex geometric arrangement of joints, ligaments and muscles enabling a massive array of movements and uses (manipulation, prehension and exploration) accounting for up to 26% of the human body's movement potential (Brown, 2008a). But because of this complexity, which is the result of evolution's continuous adaptation to the requirements of universal gripping and holding tasks over millions of years, human hand cannot be imitated expediently with the technological means available today. As a consequence, loss of such an important organ could be devastated.

Much work over the recent years have been involved in the creation of an anthropomorphic hand, where degrees of freedom and size of the hand and forearm are designed to be as close as possible to human proportions; some impressive robotic devices can be found in literature such as the NASA's Robonaut Hand, DLR Hand, Gifu Hand and Shadow Hand (Biagiotti, 2002). However, these devices fail to operate as prosthetic hands since this type of application imposes a series of challenging requirements (considerably different for example from the ones imposed in industrial robotics context, where the industrial hands typically operate in a structured environment with predefined tasks and, often, with no considerations to the human interaction). These requirements, as cited by many authors (Kyberd et al., 2001; Schultz et al., 2004) based on surveys about amputees' needs, can be summarized to the following: low weight, noiseless operation, anthropomorphic size and appearance, high functionality and dexterity, compactness, easy controllability, low cost, user safety and reliability.

The majority of users has one good hand (unilateral amputees), and become quite adept at performing tasks one-handedly, with little involvement of the prosthetic hand. So, they need the prosthesis merely as an assistor to their normal arm, and do not want to be bothered with complex devices which need care and maintenance and great concentration and practice in order to gain proficiency. Besides, if the prosthesis does not make life any easier for the user, it is likely to be rejected.

There are a wide variety of prosthetic devices available for upper limb amputees ranging from purely cosmetic ones to advanced myoelectrically controlled bionic hands. The reader can address to (Lake and Dodson, 2006; Trost and Rowe, 1992; Kelly et al., 2009) for an in-depth review of pros and cons of the different types of upper-limb prostheses. Current commercial high-tech prostheses use small electromagnetic motors installed into a human-like terminal device and are typically controlled by one or two surface electromyography (EMG) electrodes that detect muscle contraction signals in the remnant limb of the user. At

present, the best known commercially available prosthetic hand is the i-Limb™ Hand (Fig. 1a). It was first launched in 2007 by Touch Bionics and until now more than 1000 patients all over the world have fitted with it. It has 5 independent digits each controlled by a small conventional electric motor along with a combination of planetary drives, worm gears and cables. The fingers have three phalanges, like the human hand, with a stationary end joint. The knuckle joint is motor driven and motion transmitted to the middle joint by a tooth belt. Hence, a single motor can cause the whole finger to bend. This approach of having less actuators than degrees of freedom (DOF) is said to be an underactuated approach providing compliance to the fingers. The thumb is also controlled by a motor and is able to rotate so that it closes to meet the fingers.

Before i-Limb, amputees' choice was limited in prosthetic hands that behave as simple grippers with one DOF. In these devices, a single motor drives the first and second fingers of the hand in unison with the thumb to produce a tri-digital grip (open or close); the two fingers are actually one rigid link, in opposition to a rigid thumb. The remaining two fingers on the hand are for aesthetic purposes and move passively with the first two fingers. Some examples of these prostheses include the Otto Bock's SensorHand™ Speed (Fig. 1b) and the Motion Control Hand (Fig. 1c). These electric prostheses do not allow adequate encirclement of objects, in comparison with the adaptive fingers of i-Limb Hand, resulting in unstable grasps in presence of external perturbations.

Taking advantage of the significant technical improvements of the last years in various fields, the development of better and more functional prostheses is feasible. Progress in materials science, microcontrollers, batteries and cosmetics products permit the design of advanced prosthetic devices able to help amputees' life to be minimal affected.

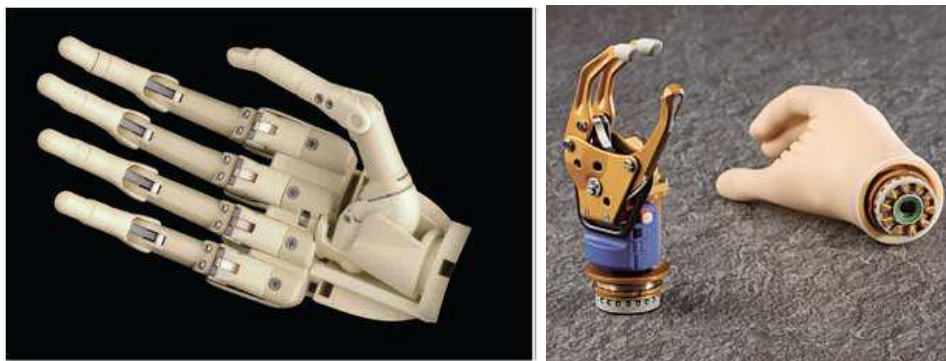


Fig. 1. Commercial prosthetic hands: a) Touch Bionics's i-Limb Hand, b) Otto Bock's SensorHand Speed, c) Motion Control Hand

2.2 Shape Memory Alloys

Shape memory alloy (SMA) is a smart material that has the ability to return to its predetermined dimensional configuration when heated beyond a threshold phase transformation temperature (Mavroidis et al., 2000). This behavior occurs due to a change in the material's crystallographic structure between two phases: the low temperature (martensite) and high temperature (austenite) phases (Fig. 2). Austenite and martensite are identical in chemical composition, but have different crystallographic structures. When an SMA is deformed in martensite, the residual strain can be recovered by heating the material to the austenite phase. This Shape Memory Effect (SME) returns the SMA to its original

shape. Thus this property can be exploited for the design of devices/actuators. This is done by training the material to remember a specific shape in the austenitic condition. When used as linear actuators, SMA are commercially available as pre-strained (trained to remember a shorter than actual length on heating) martensitic wires. Joule heating beyond the transition temperature triggers the phase transformation to austenite where the stretched wire contracts to the pre-strained length. During this transformation, the SMA can yield an extremely large force if it encounters any external resistance. Now, when the contracted wire is cooled, it returns to the martensite state (twinned) where the material is malleable. A reverse-bias force is needed to return the wire to its original length. Bias forces can be created by many methods: gravitational pull, spring, magnetic force, opposing SMA wire.

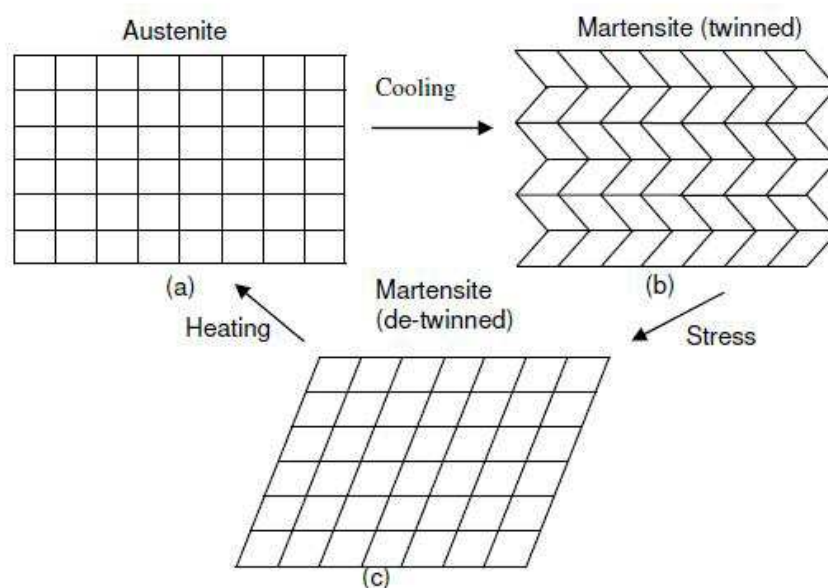


Fig. 2. Grid-like representation of SMA structure

The SME described above, whereby only the parent austenitic phase is remembered by the alloy, is referred to as the one-way SME. It is however possible to make an alloy remember both the parent austenite phase and the martensite shape simultaneously. This is referred to as the two-way SME. In this case, the alloy exhibits two stable phases: a high-temperature austenite phase, apparent on heating and a low-temperature martensite phase, apparent on cooling. Although SMA of two-way SME provides contractive and tensile forces, its tensile force is much smaller than contraction force and recoverable strain normally less than half that of one-way type (Lan et al., 2009). Thus SMA actuators of one-way effect are more attractive in robotic applications and usually preferred.

Next, a brief review of the advantages and limitations of SMA for actuation purposes is presented.

Advantages

- Compact, lightweight with high power/mass ratio and energy density - Comparing a large selection of actuation technologies, SMA actuators feature the highest power to mass ratio at less than 0.45kg masses. The energy (work) density of SMA is also very high; between 5000 to 25000 KJ/m³ when human muscles in comparison exhibit between 40-70 kJ/m³ (Abolfathi, 2007).

- Ease of actuation and low voltage requirement - Various methods can cause thermal activation on the SMA but since these are inter-metallic alloys, they can be easily driven by electrical current via Joule heating. Also, the small voltages that are required to operate make them safe for human-oriented applications.
- Clean, silent and spark free operation - In contrast to many actuators such as electric motors, SMA actuators operate with no friction or vibration allowing extremely silent movements; this is a strong asset for the prosthetic applications where the conventional technology used remains quite annoying and uncomfortable for the users. They are also free of parts such as reduction gears and do not produce dust particles.
- Long actuation life - When used within sustainable strain and stress limits, SMA actuators can be expected to last hundreds of thousands of operation cycles.
- High biocompatibility and excellent corrosion resistance - This enables their use in an environment with high humidity or wet.
- Direct-driven - Using SMA as actuators there is no need for complex and bulky transmission systems. This drastically reduces the complexity of the hand's driving mechanism.

Limitations

- Low displacement levels - Even though SMA exhibit relatively large strains, only a fraction of the net strain can be utilized in order to maximize the actuator lifetime. So, long lengths of SMA wire are required for large strains. However, using appropriate arrangements that convert the small strains into large motions this limitation can be moderated.
- Low power efficiency - SMA operate through heat and as such are limited by the Carnot efficiency model to at most 10%.
- Low operating frequency - The rate at which an SMA actuator can shift to austenite phase and return to martensite phase is limited by the slow heat transfer processes (low thermal conductivity) needed to promote the phase transformation of SMA. Typically, operation is faster for actuation (heating) than it is for relaxation (cooling). So, mostly the level of cycles per minute is dependent upon the rate of cooling of the wire. A variety of methods have been proposed to increase cooling speeds such as water immersion, heat sinking and forced air. However, even if these methods improve the bandwidth, they also cause an increase in power consumption as more heat is required to actuate the wire within the cooling medium.
- Control difficulties - Hysteresis, nonlinearities, parameter uncertainties and un-modeled dynamics introduce difficulties in accurate control of SMA. The SME is not a thermodynamically reversible process. Heat losses during the phase transformation phases (owing to internal friction or structural defects) cause hysteretic behavior of SMA as shown in Figure 3.

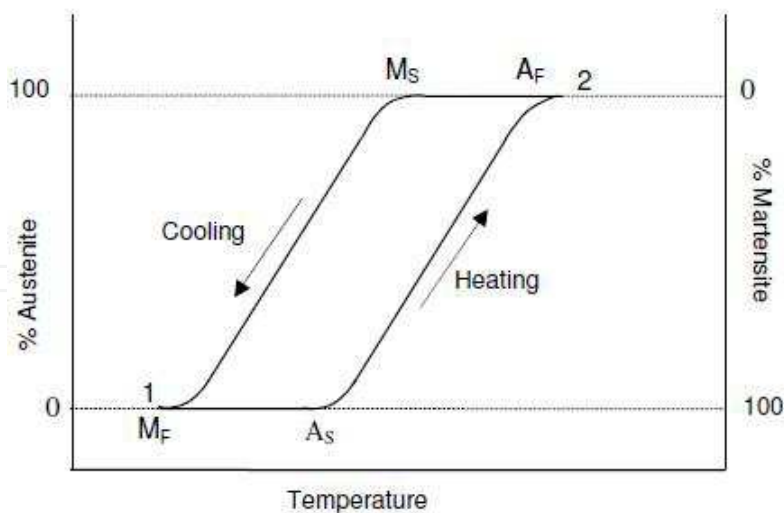


Fig. 3. Hysteresis loop in SMA

Out of all the SMA that have been discovered so far, Nickel-Titanium (NiTi) -well known as “Nitinol” (which stands for Nickel (Ni), Titanium (Ti) and US Naval Ordnance Laboratory (NOL) where the alloy was discovered in the early 1960s by William Beuhler) has proven to be the most flexible and beneficial in engineering applications showing the best combination of properties. The following characteristics of Nitinol make it stand out from the other SMA: greater ductility, more recoverable motion, excellent corrosion resistance, stable transformation temperatures, high biocompatibility and low manufacturing costs (Teh, 2008).

2.3 Related Research

Considerable efforts have already been made from different research groups worldwide to develop articulated robotic hands or finger mechanisms using SMA as the power elements of their actuation system. More specifically, DeLaurentis and Mavroidis developed an aluminum finger with 4 DOF, using bundles of Nitinol wire strands 150 microns in diameter as artificial muscles to move its joints (Fig. 4a). As a continuance of this work, an SMA actuated hand (the Rutgers Hand) was rapid prototyped comprising of 5 fingers and 20 DOF (DeLaurentis and Mavroidis, 2002). Price et al., 2007 showed practical application of shape memory alloys as actuators (Price et al., 2007) in a three-fingered robotic hand (Fig. 4b). Maeno and Hino proposed a miniature five fingered robotic hand for dexterous manipulation of small tissues and parts in medical and industrial fields, driven by SMA wire actuators with diameter of 0.05 mm (Maeno and Hino, 2006). The size of this hand was about one third of the human hands and it had 4 DOF per finger and 20 DOF in total (Fig. 4c). Cho et al. presented a five-fingered robotic hand with 16 controlled DOF and 32 independent SMA axes demonstrating the advantages of joule-heated Segmented Binary Control where each SMA actuator wire is divided into several segments (Fig. 4d). The total weight of this hand system was about 800 grams (Cho et al., 2006b). O’Toole and McGrath with their work (O’Toole and McGrath, 2007) also proposed a prosthetic hand design that incorporates embedded SMA bundle actuators. Bundhoo et al. constructed a 4-DOF (three active and one passive) artificial finger testbed (Fig. 4e) based on the combination of compliant tendon cables and one-way SMA actuators (off-the-shelf products from Miga

Motor Company) in an antagonistic arrangement for the required flexion/extension or abduction/adduction of the finger joints (Bundhoo et al., 2008).

Besides the above research efforts that took place in the last decade, it is worth referring one of the very first attempts to introduce SMA in robotic hands: the Hitachi Hand. This hand caused an understandable sensation with its debut in 1984 claiming a 10:1 reduction in weight as compared to other hand designs. It had three 4-DOF fingers and a thumb. It also possessed a forearm and a pitch-yaw wrist. It used a large number of SMA wires, actuated in parallel. Electrical heating of the wires resulted in their contraction against a force spring. On cooling, the wires returned to their original length. The force generated on contraction was used for joint actuation. The Hitachi Hand used 0.02 mm diameter SMA actuators for the fingers. Each DOF of the wrist, on the other hand, were actuated by 0.035 mm diameter SMA wires, set around pulleys. The above mechanism enabled 90° joint travel. Joint positions were sensed by potentiometers. This four-fingered gripper was 69.85cm long, weighed 4.49kg and was capable of holding a 2kg load capacity (Yang and Wang, 2008).

However, in each of the above cases few experimental results are available, and no complete feasible solution has been proposed to satisfy the requirements of a prosthetic device.

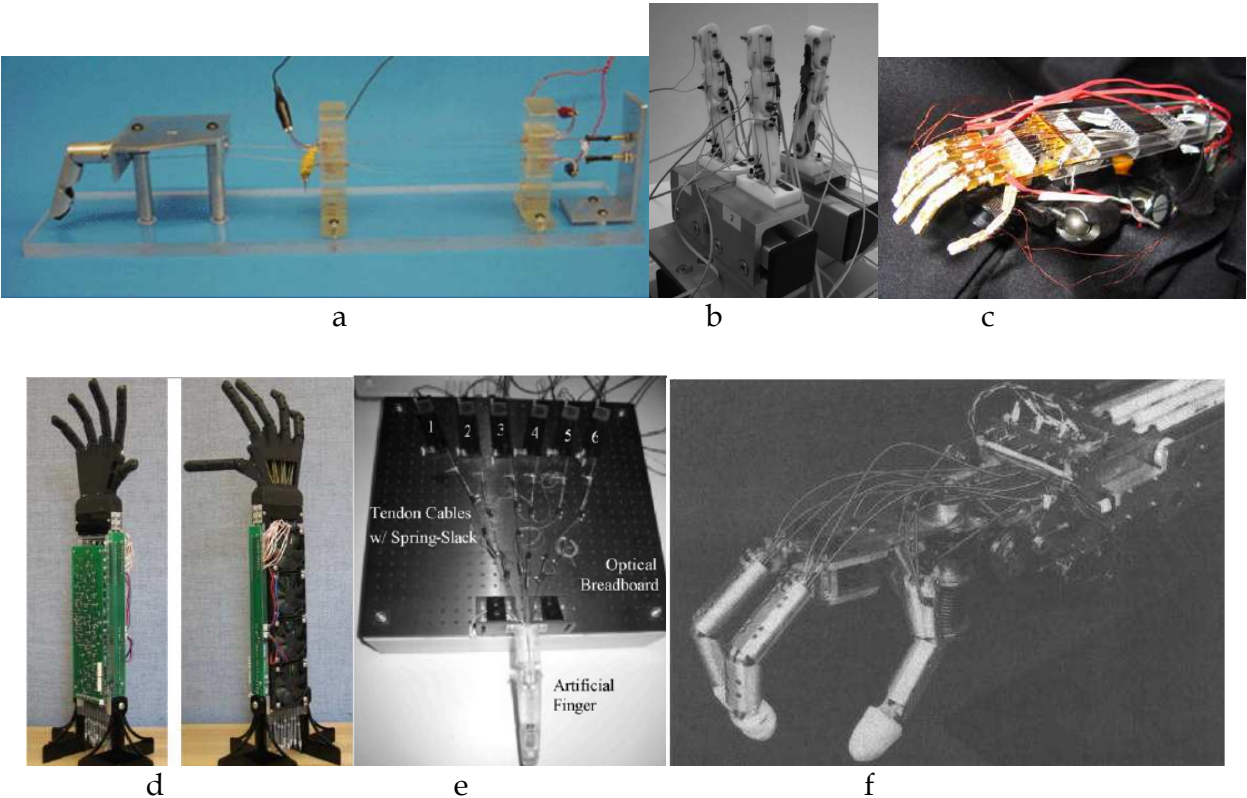


Fig. 4. Various SMA actuated mechanisms

3. Prosthetic Hand Development

3.1 Mechanical Design

Human hand is a biological system that has evolved into a very efficient and effective mechanism after millions years of evolution and as such it is the unchallengeable

benchmark of our research efforts. Its advanced and complex kinematic format gives dexterity, combined with delicacy of movement and capacity for power grasping, and endows humans with incredible powers of maneuverability and capability for interaction through various object grasping and manipulation strategies. At the same time the hand and fingers also have a sensory function fundamental to these exploring purposes. Thus, an extensive study of natural hand biomechanics and anatomical data stands to reason when designing a new prosthetic device.

The human hand is a very articulated structure and the most dexterous part of the human body. It consists of 27 bones, 8 of which are located in the wrist. The other 19 constitute the palm and fingers as shown in Figure 5. The bones in the skeleton form a system of rigid bodies connected together by joints with one or more DOF for rotation. Joints between the bones are named according to their location on the hand as metacarpophalangeal (MCP) (i.e., joining fingers to the palm), interphalangeal (IP - the proximal interphalangeal (PIP) and the distal interphalangeal (DIP) joints) (i.e., joining finger segments) and carpometacarpal (CMC) (i.e., connecting the metacarpal bones to the wrist). The nine IP joints can be accurately described as having only one DOF, flexion/extension. All five MCP joints, however, are described in the literature as saddle joints with two DOF: abduction/adduction (i.e., spreading fingers apart) in the plane defined by the palm, and flexion/extension. The CMC of the index and middle fingers are static while the CMC of the pinky and the ring finger have limited motion capability reflecting palm folding or curving, which is often discarded yielding a rigid palm. The CMC of the thumb, which is also called trapeziometacarpal (TM), is the most difficult to model. Biomechanical studies have shown that the TM joint has two non-orthogonal and non-intersecting rotation axes. The two DOF saddle joint is a restrictive model but it has been used in many studies (Albrecht et al., 2003). The structural limits of these joints are dictated by the anatomy and physiological make-up of the hand. Though the hand has a high number of freedoms, some of these degrees are interdependent owing to the muscular inter-connections of the hand. Fingers motion is performed by contracting muscles in the forearm and palm which are attached to the finger bones through a convoluted network of interdependently acting tendons and ligaments.

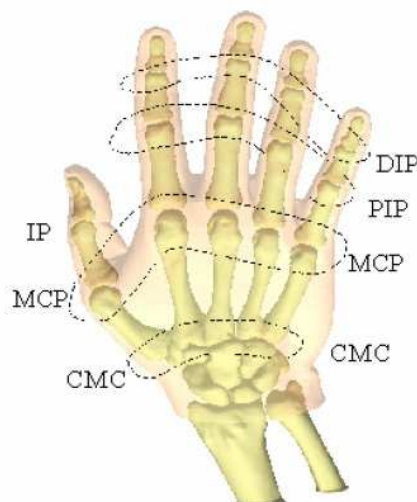


Fig. 5. Human hand joints

So, inspired by the nature's elegant and effective solution to the problem of actuation and exploiting nonconventional biomimetic forms of actuation such as SMA we proceed with the development of a novel prosthetic device that mimics human hand in appearance, size and performance as much as possible. Many design approaches were evaluated experimentally or via intense graphically simulations and eventually an optimized CAD model was created, illustrated in Figure 6. Its geometry is based on anthropometric measurements and is equivalent to the hand size of the average percentile American female (almost equal to the 5th percentile American male). It is designed along with a small forearm stump of 15cm long needed to accommodate all its necessary components such as its actuators and electronics. In this way, a very compact design is achieved allowing for the fitting of a high number of upper limb-deficient persons.

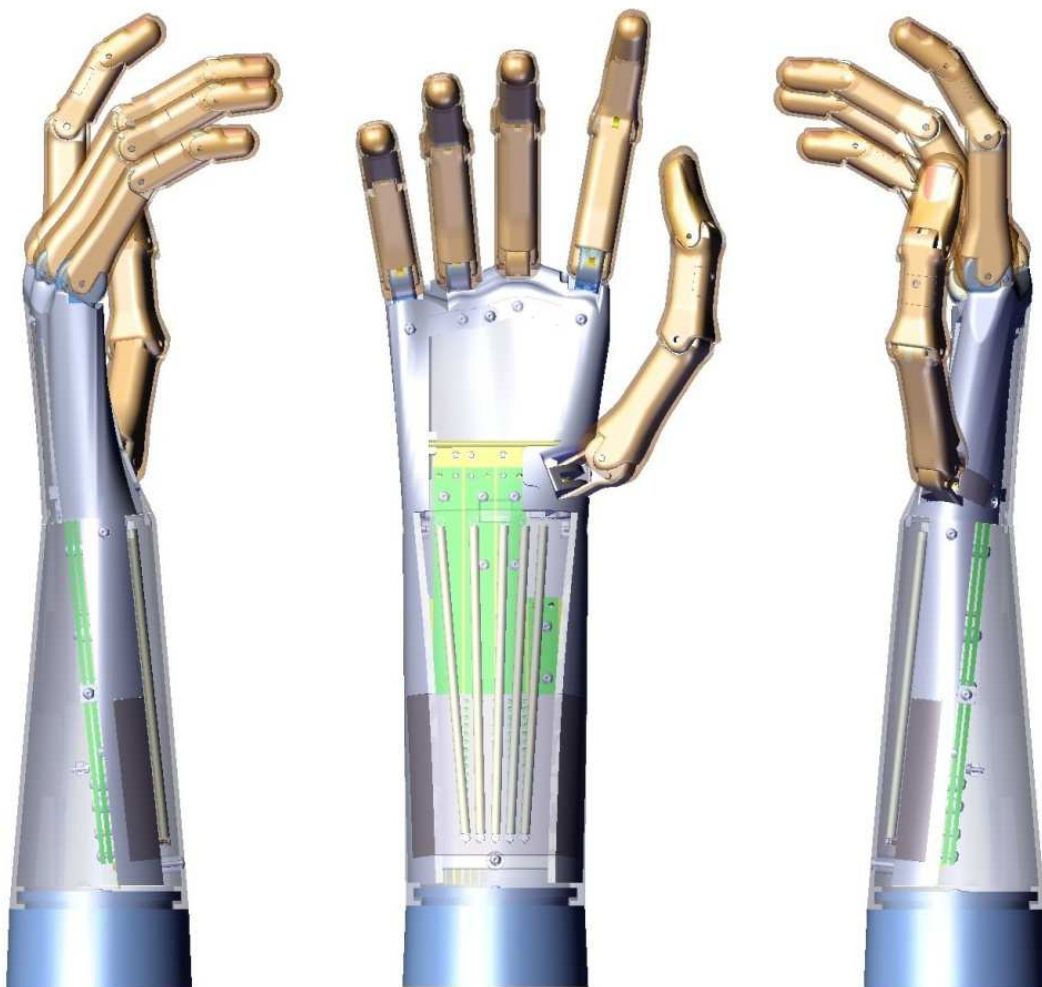


Fig. 6. CAD model of the prosthetic hand

It consists of five digits: a thumb and four fingers. In order to enhance its dexterity, a high number of DOF is requisite. Towards this goal, SMA due to their compactness can outmatch the traditional electro-magnetic actuators. However, since in myoelectric prosthetic hands only few control signals are available from the EMG control interface, it is not possible for the amputee to control many actuators (Carrozza et al., 2003). Therefore, an underactuated approach is adopted whereby a large number of DOF is controlled with a limited number of

actuators. In this way, fingers can conform to the shape of an object during grasping (adaptive grasp). The geometric configuration of each finger and the orientation of each phalange are automatically determined by the external constraints imposed by the object, so that active coordination of the phalanges is not needed. The dimensions of the links, the configuration of the fingers, and the position of the contact points will determine the distribution of forces between the phalanges and the grasped object. In general, low contact forces are needed for grasping (below 10N for most manipulations) and contact forces are distributed similarly to those of natural hands.

The kinematic architecture is shown in Figure 7. These 19 DOF are controlled via 8 SMA Actuation Units (AU). More specifically, index and middle fingers utilize two AU per finger while one AU is granted to ring and pinkie. The last two are used for the actuation of thumb. Thus, all digits can move independently from each other. Finally, an extra DOF concerning forearm rotation is independently actuated via a high torque servomotor installed in the back of the forearm housing. As it is claimed by (Almstrom et al., 1981) this movement is considered to be the most useful after the grip for amputees.



Fig. 7. Kinematic architecture of the proposed prosthesis

With this architecture, the formation of all the basic types of hand grasping as described by Schlesinger in 1919 and shown in Figure 8 is permitted. As it has been stated, 90% of all common dexterous actions are covered with these six basic grasp primitives. Thus, it is expected common activities of daily living to be easily performed using this device by amputees.

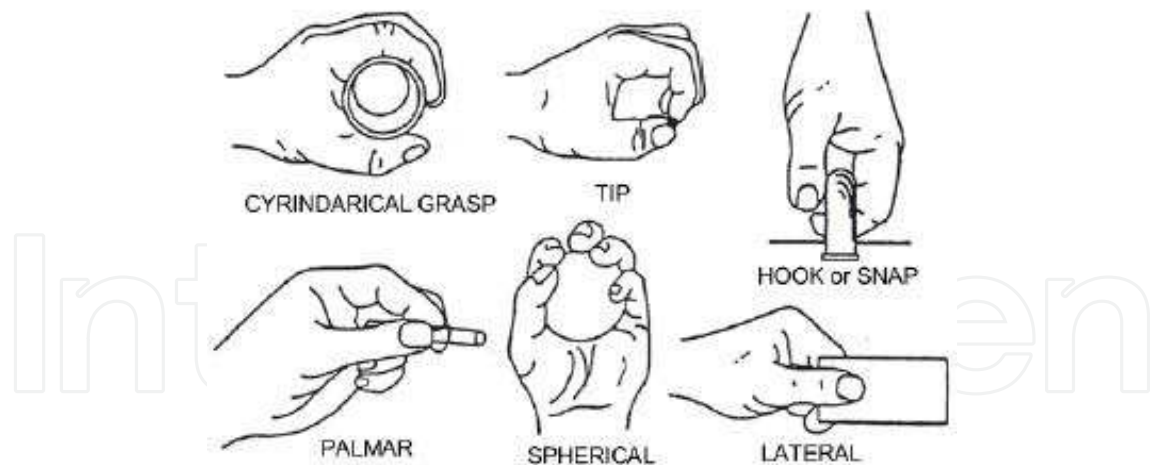


Fig. 8. Basic grasp primitives according to Schlesinger

3.2 Actuation Mechanism

SMA actuators are the key elements for the development of a lightweight, noiseless and multifunctional prosthesis. In this research, specially processed Nitinol wires, commercially available under the trademark “Flexinol” by Dynalloy Corporation, are used (Brown, 2008b). These wires are one-way actuators that when heated contract about 3-4% in length exerting significant stress ($\sim 200\text{MPa}$). Their contraction-heating can be easily accomplished using an electrical current in less than one second. On the contrary, the cooling phase is much slower and a bias force is needed in order to have the wire returned to its relaxed uncontracted state.

Flexinol wires are available in various diameter sizes (ranging from 0.25 to 0.375 mm). Smaller diameter results in quicker cooling time but smaller force output. In fact, there is a trade-off between speed and force; while cooling time has an inversely proportional relationship with cross-sectional diameter, force output is inversely proportional to its square. There are also other factors that affect the cooling time of the SMA wire, such as the method of heating, the medium surrounding the wire while also there are methods that can increase force output such as bundling parallel wires or force-multiplying arrangements.

After intensive experimentation with these wires and different setups, we develop modular and compact AU, illustrated in Figure 9a, that have a high force output and fast time responses (heating and cooling times are discussed analytically in Section 4). For each AU, one Flexinol wire of 0.25mm diameter is used; based on the supplier this wire can last for more than 1,000,000 life cycles, exert forces up to 9N and needs 3.5 seconds for cooling. Bending this wire to the middle doubles the provided force to 18N (another way to augment the force output of each AU concerns the bundling of several parallel wires; however, it is proven in practice that implementation of this method encounters many difficulties). Also, given the limited strain capabilities during contraction and prosthetic hand's tight dimensions a special pattern for the Flexinol wire has been applied to all the AU (Fig. 9b). More specifically, the wire is wrapped around pulleys of 5.5mm diameter permitting an about 10% contraction in regards to AU's length. Moreover, in order to increase actuation bandwidth, Flexinol wires run through a small diameter rubber tube which acts as a heat sink helping them cool faster.

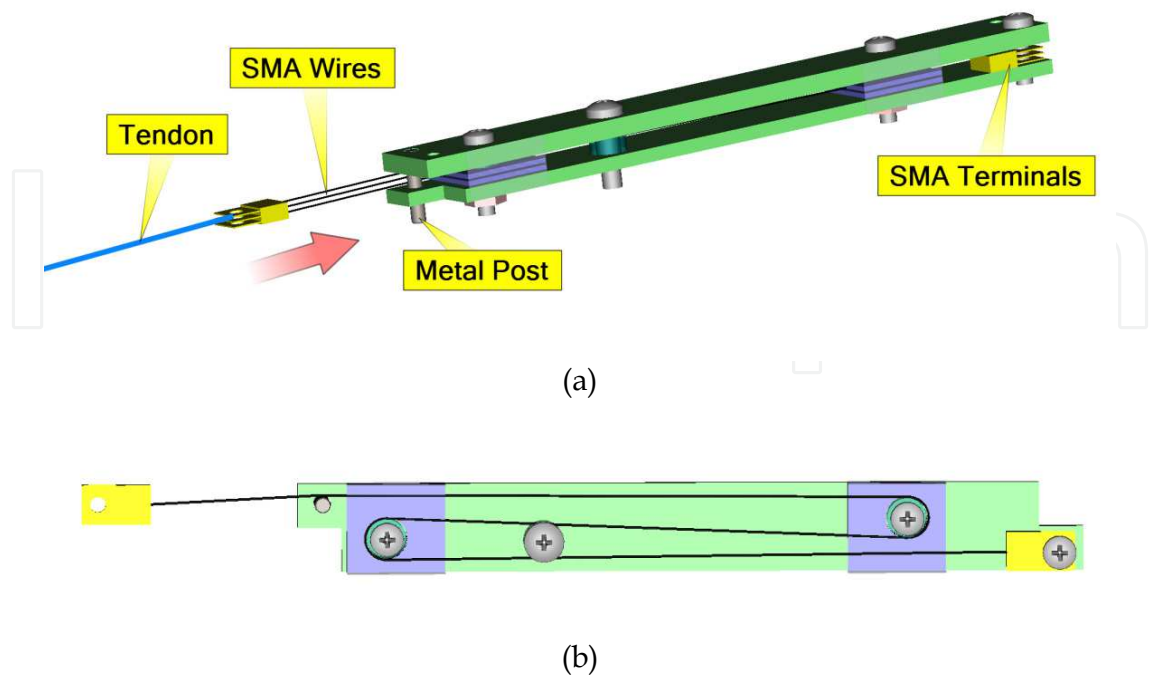


Fig. 9. An actuation unit and its a) working principle, b) SMA wire arrangement

As it is already mentioned, an underactuation mechanism is chosen allowing the digits to move independently. Their motion is controlled by 8 AU in total located intrinsically of the hand structure as shown in Figure 10 and rested on a printed circuit board (PCB) that brings all the necessary wirings.

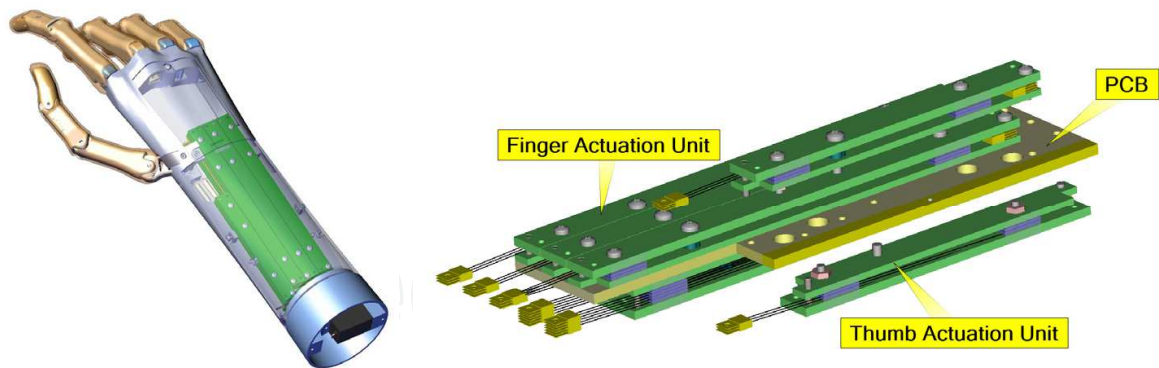


Fig. 10. Integration of an SMA actuation system in the structure of the prosthetic hand

In order to configure each AU's assignment for optimizing hand function, the following must be taken into account. Since SMA actuators have a very low power efficiency (<10%) and active hand prostheses are typically powered by rechargeable batteries of limited capacity, there is a need to reduce the power requirements of the hand as much as possible. Towards this goal, a voluntary opening mechanism is applied to our prosthesis. This means that the device is closed at rest and electrical power is required to open the hand. Pre-tensioned extension springs act antagonistically to the AU (which are controlling the opening of the hand) and flex all the digits providing a passive adaptive grasp. Apart from

the pre-tensioned springs, in case that higher pinch forces are requisite for a grasp, two additional AU give extra flexion force to the index and middle finger. In the thumb, two AU are responsible for its motion: one for extension and the other for adduction; extension springs are used for passive flexion and abduction.

A tendon transmission system is used in order to transmit force in the fingers. The tension of the tendons generates a flexing torque around finger joints, by means of small pulleys, allowing hand motion. Whereas biological tendons are made up of dense connective tissue which is elastic, flexible, and strong, in our design nylon-coated stainless steel wires are used as mechanical counterparts. They have their one end terminated into the middle phalange of each digit and the other end attached to an AU or extension spring. This implies that all the distal (DIP) joints of the digits are not actively actuated; in fact they are passively flexed about 25 degrees by means of a contraction spring between the last two phalanges.

3.3 Prototype Fabrication

The overall hand design consists of many different CAD-generated solid parts assembled in a manner that forms the suggested prosthesis structure. Using these three-dimensional models and a Rapid Prototyping (RP) technique (SLS process), a hand prototype has been fabricated and assembled as shown in Figure 11. For this purpose, a fiber-reinforced plastic composite with high strength and stiffness is used.



Fig. 11. Prosthetic hand prototype

Its weight is less than 200 grams and uniformly distributed along its length. With the integration of electronics and battery, it is estimated to weigh significantly less (about one

third) than the current commercial prosthetic hands. Besides, as it is already mentioned, the mass of a prosthetic hand must be as low as possible in order to be accepted as a technical aid by the amputee; as it is worn on the end of a closely fitting external socket, its weight bears directly onto the skin of the stump. The lever-arm created is therefore large and the weight can obstruct blood flow in the underlying skin and results in symptoms ranging from discomfort to skin breakdown (Kyberd et al., 2001).

4. Experimental Study

4.1 Testbed

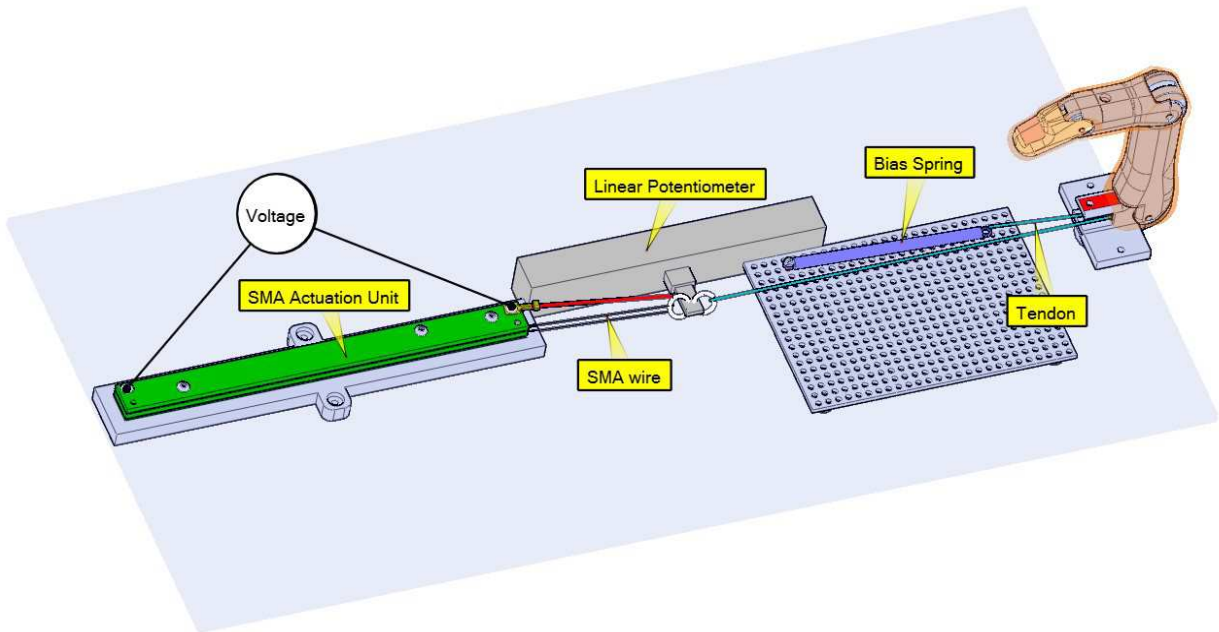
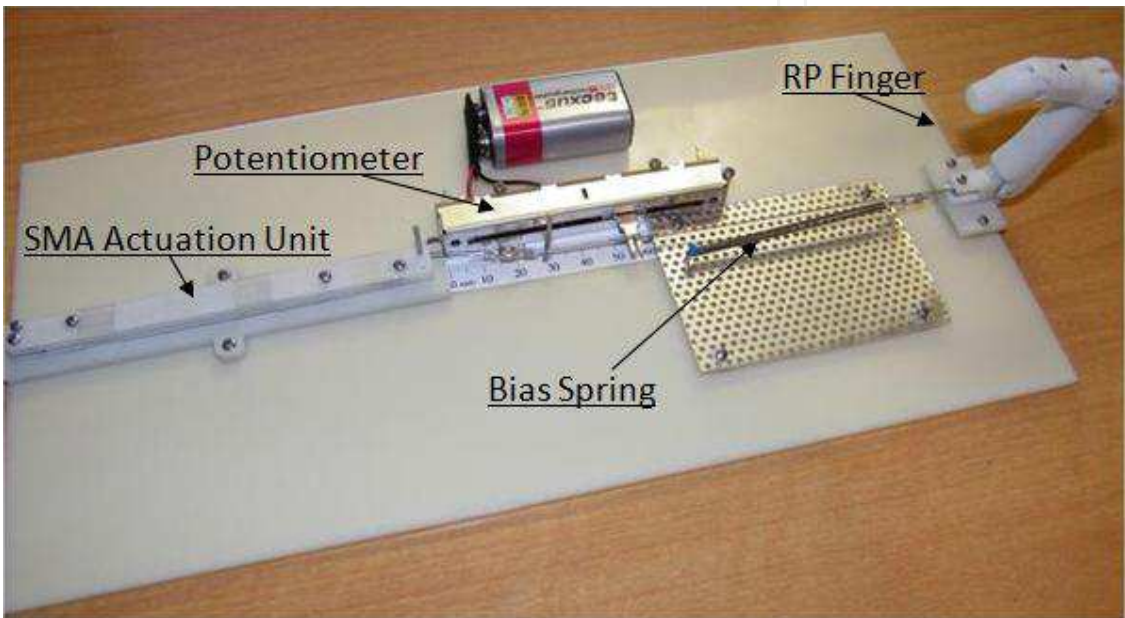


Fig. 12. Artificial Finger Testbed

As our prosthetic hand consists of five individual digits, the first phase of experimentation concerns a one-finger testbed developed as shown in Figure 12. It follows the aforementioned activation pattern: when AU is powered, the finger extends (opens) while an extension spring acts antagonistically to this motion in order to flex the finger back to its rest position. A linear potentiometer is used in order to measure the displacement generated by the AU. Additionally, miniature thermocouples are attached to the Flexinol wire of the AU for temperature monitoring purposes. A personal computer (PC) interfaces with the testbed through a PCI-6229 data acquisition card manufactured by National Instruments.

Actuation is controlled by electrical heating of the Flexinol wire. Though this can be done using a direct current (DC), Pulse Width Modulated (PWM) technique is preferred; PWM activation enables a more uniform heating of the SMA wires as compared to joule heating with a DC current (Bundhoo, 2009). Thus, it can improve the overall energy efficiency of our actuation system. As only a single voltage power supply is required, the need for additional voltage amplification equipment is eliminated. Moreover, PWM is easily implemented using microprocessors.

A low-cost development board (STM-H103) interfaced via a USB port on the PC and based on an ARM 32-bit Cortex™-M3 CPU is used for generating the PWM signals. The duty cycle (τ) and frequency of the PWM can be specified by writing to specific hardware registers. The PWM signals have a value ranging from 0 to 4095 (based on a 12-bit signal) resulting in PWM duty cycle ranging from 0% to 100% (Fig. 13). An external power supply provides the necessary voltage (V_o) and custom-made electronics are used to feed Flexinol wire with the desired current. Supervision and experimentation with this testbed is carried out through a graphical LabVIEW environment.

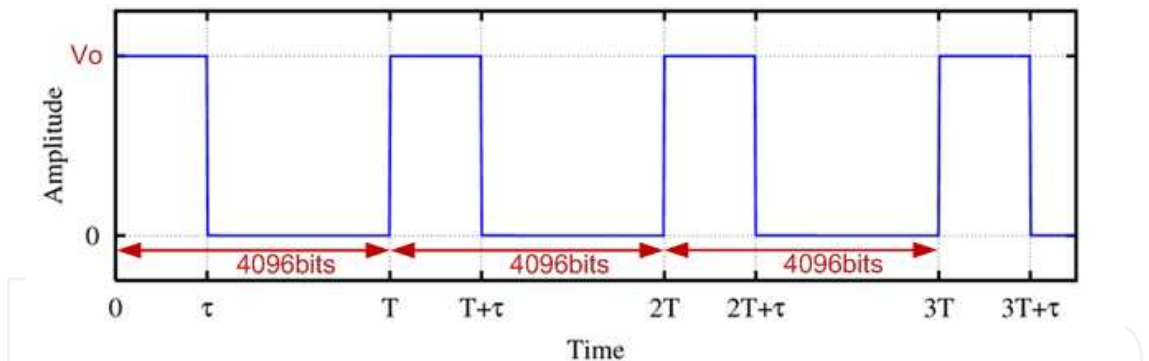


Fig. 13. A PWM output signal waveform

4.2 Results & Discussion

Open-loop Experiments:
As stated previously, the AU imparts motion to the finger via a tendon transmission system; a flexing torque is generated around the finger joints by means of 3mm radius pulleys. The range of motion for the controlled joints of our finger is: 90 and 75 degrees for the MCP and PIP joint respectively. So, the contraction length needed for extending fully the finger can be approximated by the following equation:

$$\Delta l_i = \sum_i R_i \theta_i = R_{MCP} \theta_{MCP} + R_{PIP} \theta_{PIP} = 3 \cdot \frac{90\pi}{180} + 3 \cdot \frac{75\pi}{180} = 8.64mm$$

The AU has greater strain capabilities than the ones needed. It can provide a 12mm displacement which is about 10% of its overall length. This means that the Flexinol wire will have quicker responses since it undergoes only a partial phase transformation in order to bring the finger in an open position.

Using the setup described above, experimental studies of open-loop control are conducted. During each experimental run, the following data is recorded: SMA voltage drop, current, contraction and temperature at the surface of the SMA. A third-order Butterworth filter with a 20 Hz cutoff frequency was used to smooth noise in the current data.

Towards evaluating finger's response, different electrical currents are applied to the SMA wire for an initial load of 5.32N (generated by a pre-tensioned extension spring with a constant $k=0.28\text{N/mm}$); when the displacement-contraction level measured by the linear potentiometer equals to 8mm (i.e. finger in open position), power is off letting unhindered relaxation-cooling of AU flexing back the finger. The resulting plots are depicted in Figures 14-17. From their examination, one sees that an increase of the amperage has drastic reduction in time needed for a full actuation cycle. Comparing the temperature curves (Fig. 15-16), a much higher temperature peak is observed for the lower current input. However, this higher temperature did not result in increased contraction as better shown in Figure 17. In other words, a significant amount of thermal energy was wasted since it was not converted into mechanical work.

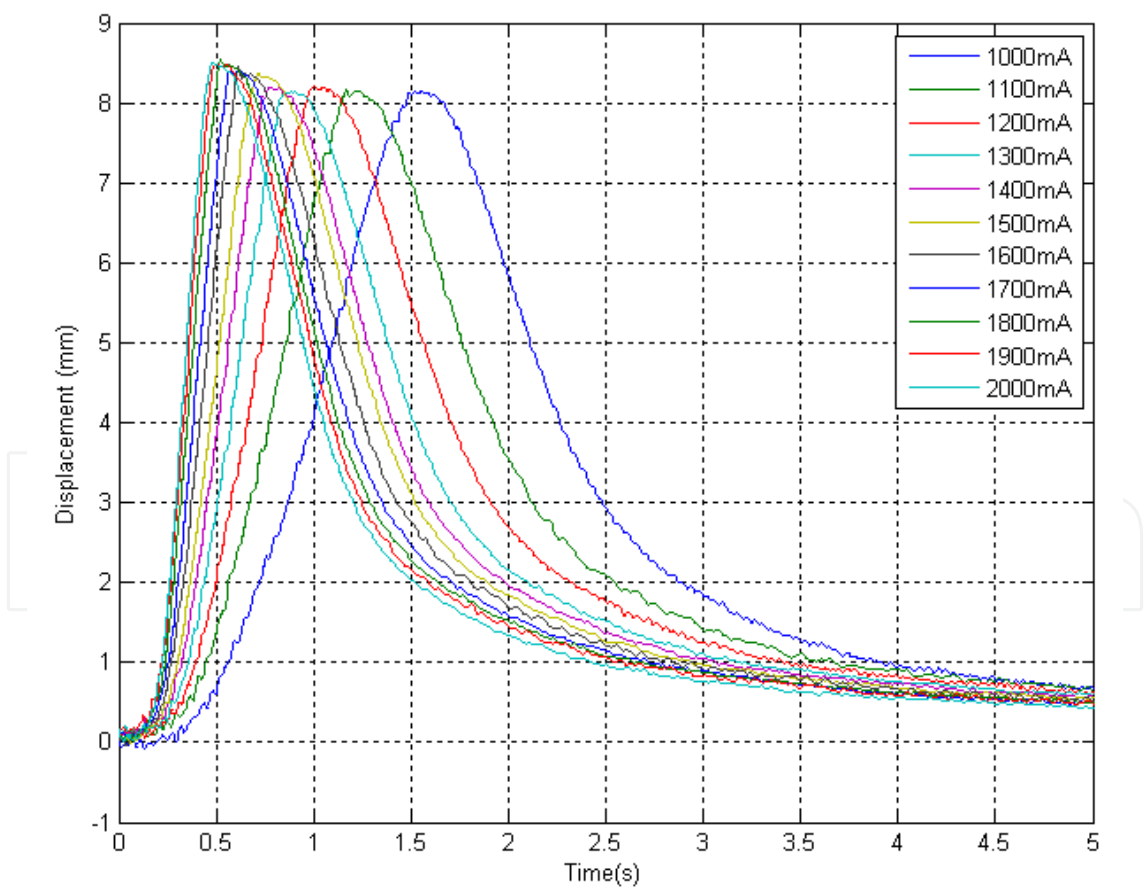


Fig. 14. Motion Profile for different electrical current input signals

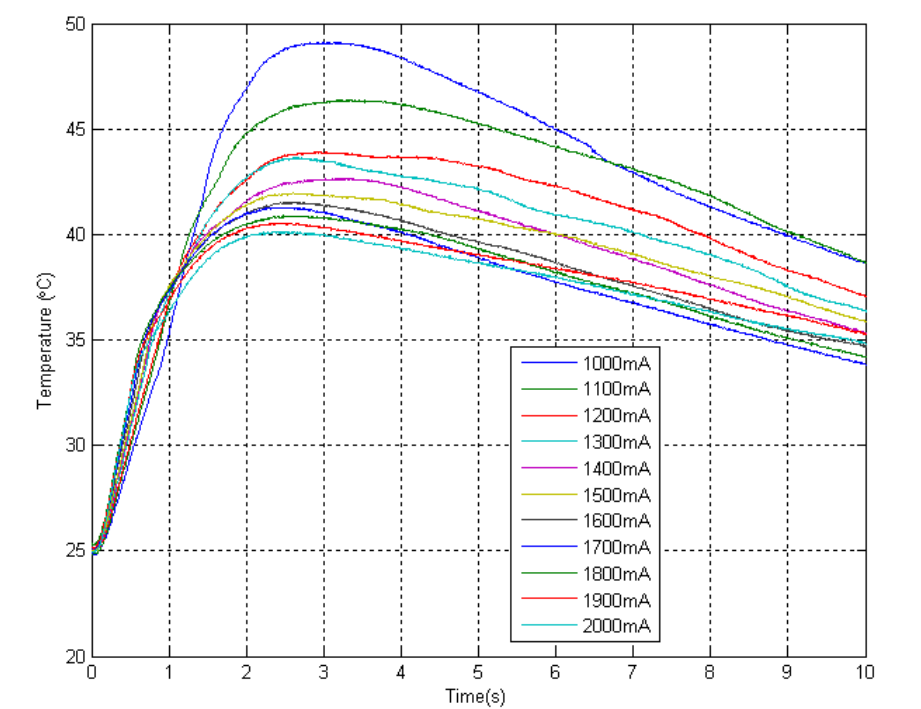


Fig. 15. Temperature Profile for different electrical current input signals (t=10s)

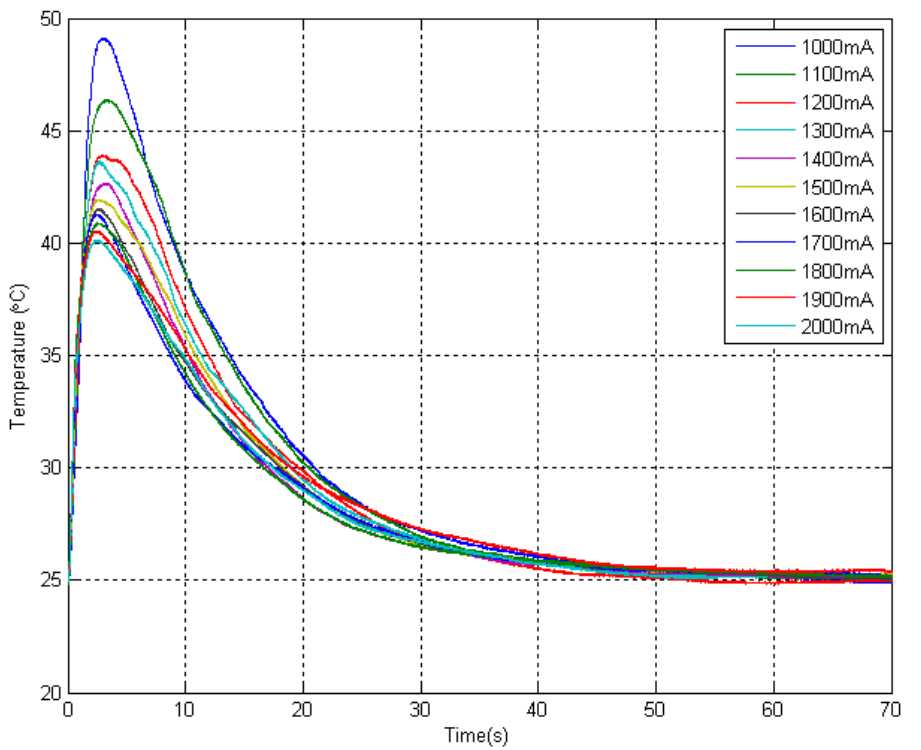


Fig. 16. Temperature Profile for different electrical current input signals (t=70s)

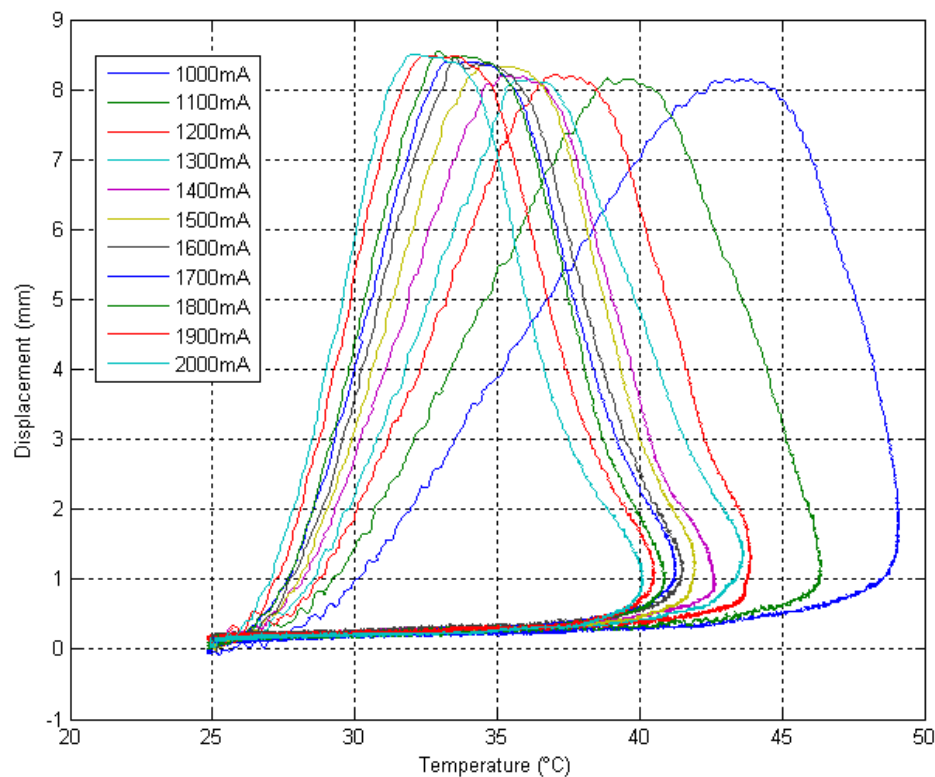


Fig. 17. Displacement vs. Temperature for different electrical current input signals

Next, the performance of the AU is studied for different loads (Fig 18). Supplying the same amperage ($I_{rms} = 1400\text{mA}$), measurements are taken for three different bias loads of the extension spring: Low Load = 5.32N, Average Load = 7N, High Load = 8.68N. From the plots shown below, one may notice a slight faster response for the lighter load. This can be easily explained considering the elasticity of the SMA wire. The elastic strain acts antagonistically to the shape memory strain. So, for a larger load, there is more elastic strain and thus more shape memory strain is needed to cancel it. As a result, there is a longer delay before the load starts to move.

The speed performance of the finger can also be evaluated from this plot. The time needed for extension of the finger is less than 0.9 seconds. Passive return to the rest position is a more slow process: about one second after power is off an approximately 70% of the displacement is restored, while with a 2.5 seconds of cooling the finger flexes about 90% of its maximum range.

As it has been already observed (Teh, 2008), the electrical resistance of the SMA wire varies directly with its length. Indeed, our experiments (Fig. 19) confirm that during its heating phase the resistance drops about 15% with an almost linear relationship with contraction. Thus, measuring the resistance value, position feedback can be provided eliminating the need for additional position sensors.

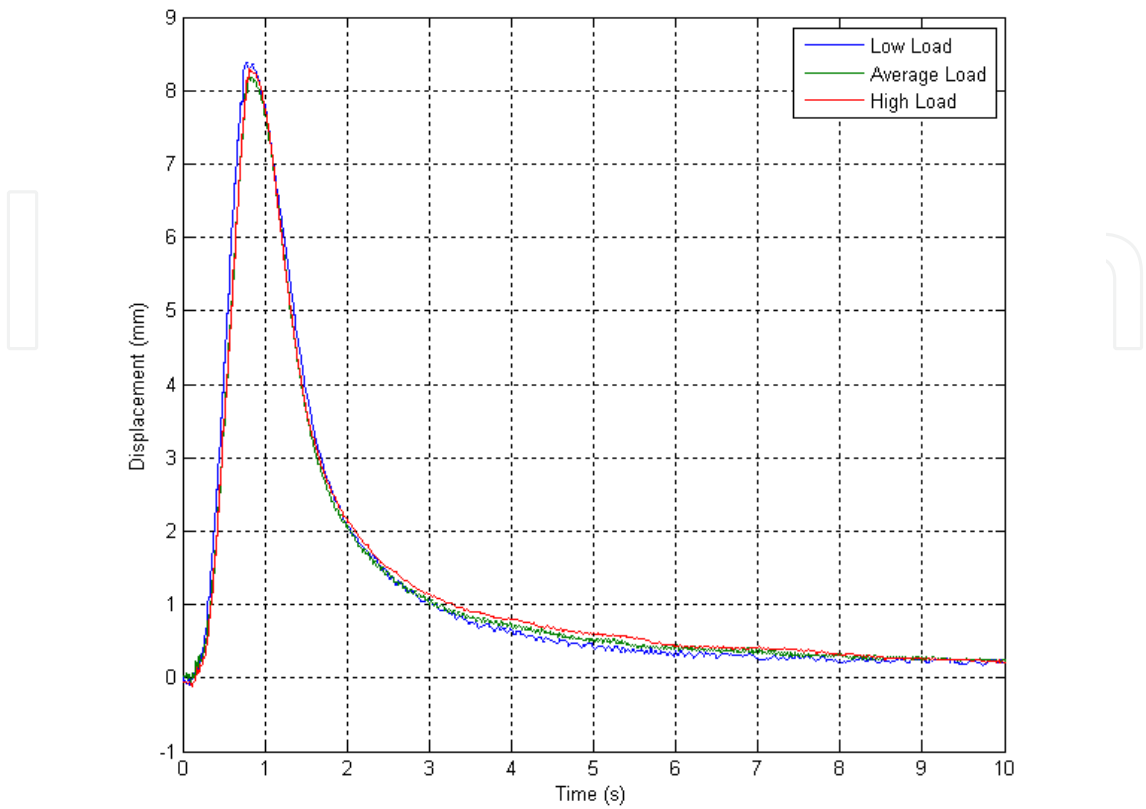


Fig. 18. Motion Profile for different loads

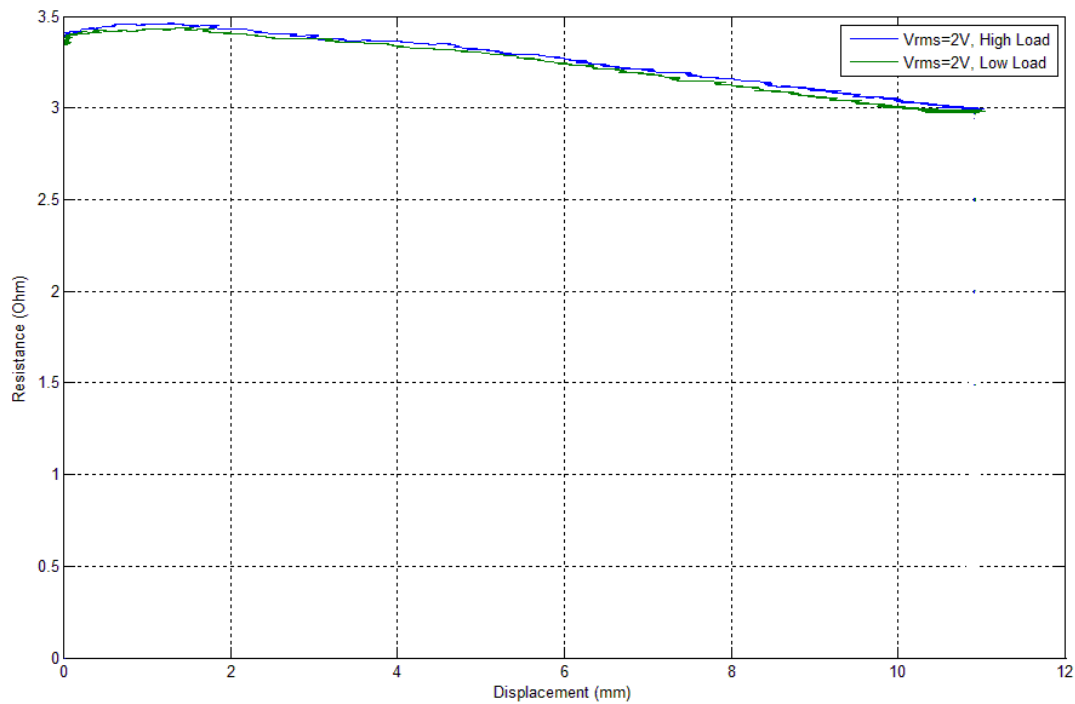


Fig. 19. Resistance vs. Displacement for different loads

Closed-loop Experiments:

In order to assess the performance of a position feedback control to the system, tests are carried out using the control infrastructure shown in Figure 20.

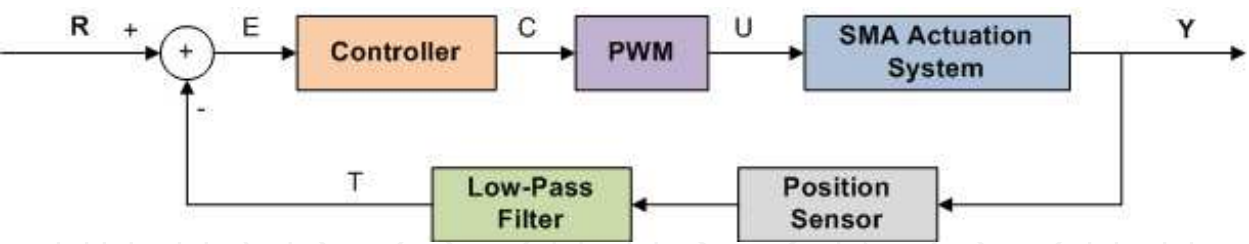


Fig. 20. Control infrastructure

In literature, various control strategies have been developed to cope with the nonlinearities and hysteresis phenomena of the SMA. A thorough review of these strategies can be found in (Rezaeeian et al., 2008; Cocard et al., 2006; Ashrafiuon et al., 2006). Briefly, researchers have explored linear controllers as well as nonlinear control schemes including fuzzy logic, neural networks, feedback linearization, optimal control and variable structure control. However, the large range of factors affecting SMA behavior (e.g ambient settings, stress, strain and material fatigue) hinder the robustness of the majority of controllers.

As it is underlined in (Cho and Asada, 2006a), due to the bi-stable nature of SMA’s phase transition, an on-off type actuation is more suitable than a proportional control. Thus, in our research, an on-off controller is initially tested and its results are presented below. With this type of controller, an electrical current of $I_{rms}=1400mA$ is applied to SMA wire when there is a negative position error ($E = R-T$) otherwise current is zero. Results for an incremental step input are shown in Figure 21 indicating serious overshoot problems and intense fluctuations around steady-state values. Consequently, it is clear that an ON-OFF controller is incompetent to provide satisfactory responses for our system.

Therefore, other control approaches were examined for our system. After experimentation with various control schemes, a gain-scheduling controller was selected. This is one of the simplest and most intuitive forms of adaptive control. It is an approach to control of non-linear systems using a family of linear controllers, each of which provides satisfactory control for a different operating point of the system. One or more observable variables, called the scheduling variables, are used to determine what operating region the system is currently in and to enable the appropriate linear controller. In our case, position error is the only scheduling variable. The operating regions and gains are tuned empirically.

Experimental results of Figures 22-25 indicate the efficacy of the suggested controller. In Figure 22, finger demonstrates a significant improvement to its performance compared to the ON-OFF controller. Overshoot problems have been eliminated and finger’s stability around different set-points has turned to be more robust. In Figures 23-24, the responses for input signal tracking using sinusoidal command signals of 0.15 and 0.05 Hz are shown respectively. In this case, the reference signal is selected regarding the speed performance computed before in open-loop experimentation. Finally, a qualitative study of the robustness of this controller is provided in Figure 25 where several external disturbances are imposed to the system.

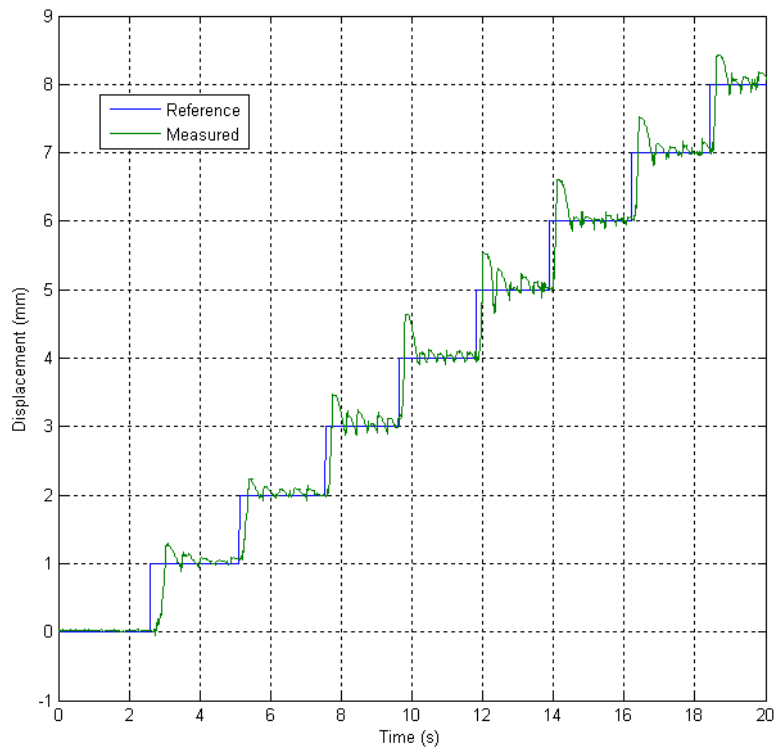


Fig. 21. Position Response to an incremental step input with an ON-OFF controller

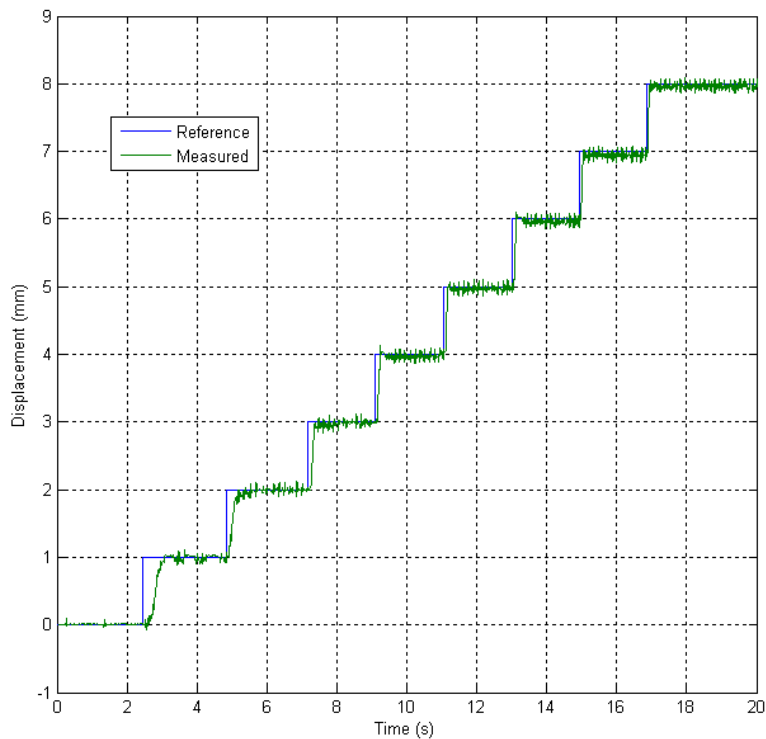


Fig. 22. Position Response to an incremental step input using a gain-scheduled controller

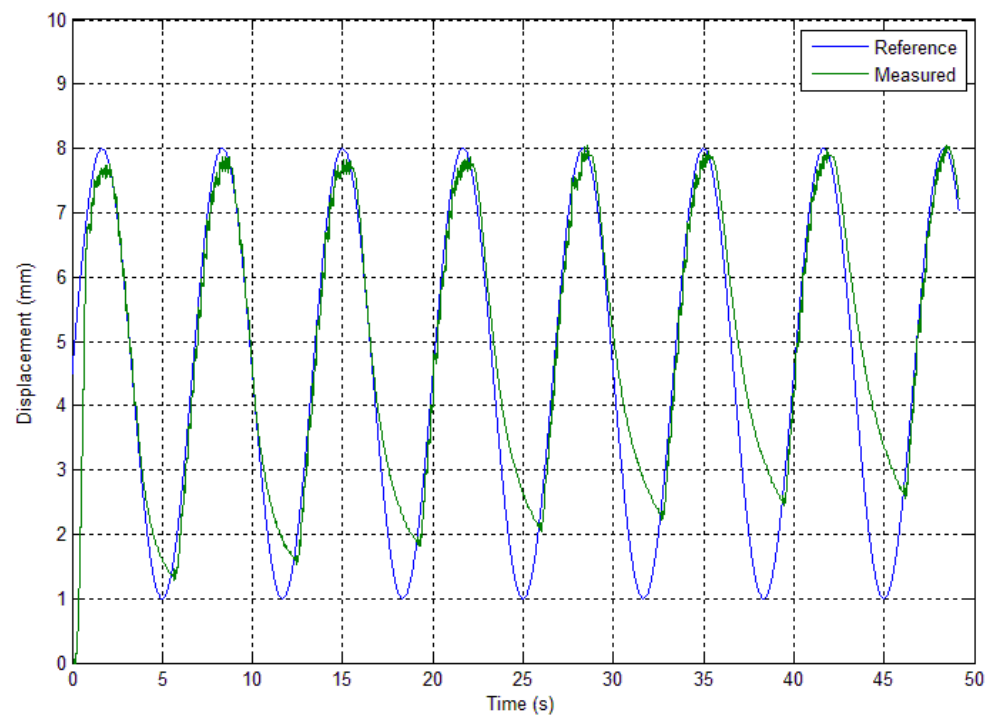


Fig. 23. Position Response to a 0.15Hz sinusoidal input using a gain-scheduled controller

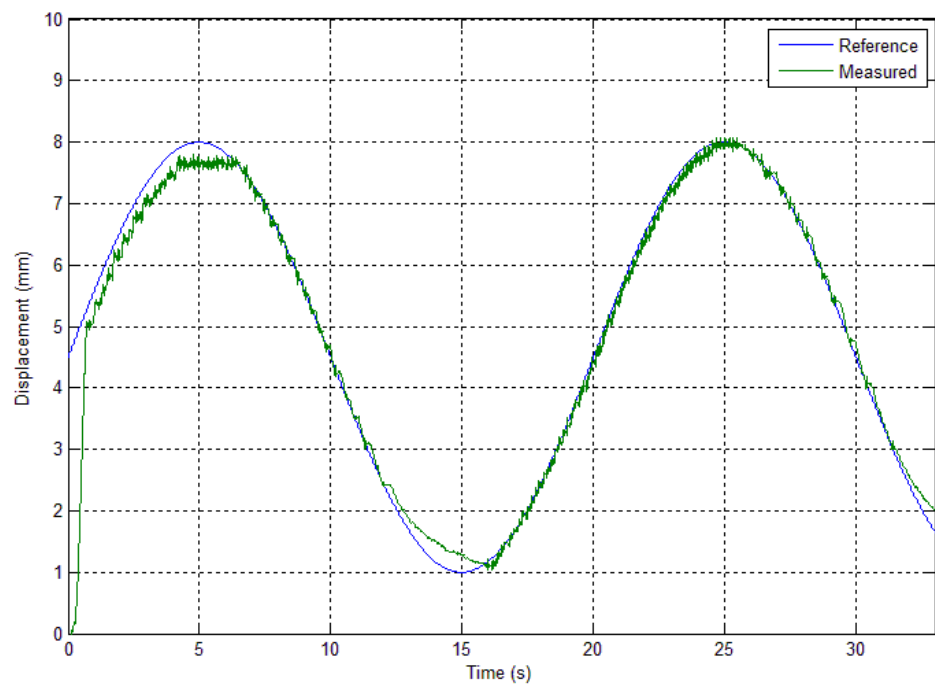


Fig. 24. Position Response to a 0.05Hz sinusoidal input using a gain-scheduled controller

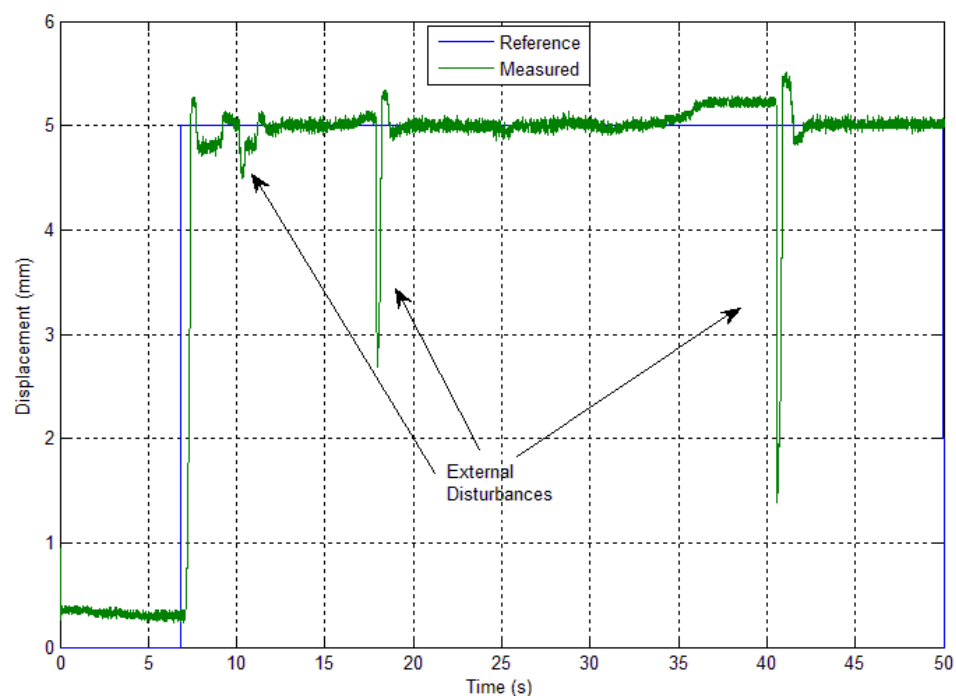


Fig. 25. Finger’s Position Response to a step input with external disturbances using a gain-scheduled controller

5. Conclusions & Future Work

It is a common belief that improvement of actuation systems will have a crucial impact on the development of more anthropomorphic and functional prosthetic hands. One of the emerging technologies in the actuation field concerns the utilization of novel, smart material based SMA wires. Their incredibly high power to weight ratio along with their ability to operate with a biologically similar motion suggest that they could be an ideal solution to the size and weight constraints of prosthetics.

In this research, an SMA actuation system integrated in a prosthetic hand was presented. Owing to the lightweight and silent nature of the SMA, this hand can enhance acceptance by the amputees. Performance of one finger was assessed by open and closed loop experiments. The obtained results encourage future work with the hand prototype. Taking advantage of a voluntary opening mechanism in combination with an underactuation approach, the power needed to perform and maintain a grasp is reduced allowing a rechargeable battery source to operate the prosthesis. Moreover, the response time proved to be satisfactory for manipulative tasks since a finger can open and close with almost the average human speed.

Future work will focus on the prosthetic hand’s performance. Electronics will be embedded in structure and different control strategies will be evaluated on the base of resistance feedback. Reliability tests of the proposed prosthesis are also planned to be conducted for both static and dynamic grasps, with its compliance and conformability to warrant high grip functionality.

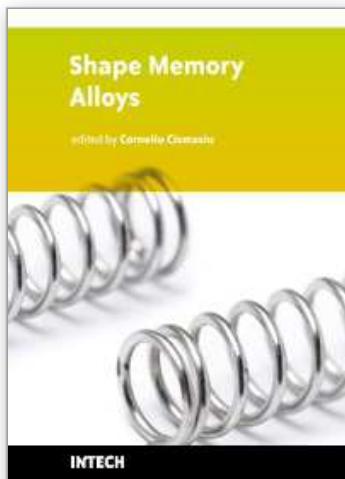
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In the last decades, the Shape Memory Alloys, with their peculiar thermo-mechanical properties, high corrosion and extraordinary fatigue resistance, have become more popular in research and engineering applications. This book contains a number of relevant international contributions related to their properties, constitutive models and numerical simulation, medical and civil engineering applications, as well as aspects related to their processing.

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