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The Development of a Novel Biomechatronic Hand – Ongoing Research and Preliminary Results

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Abstract– An "ideal" upper limb prosthesis should be perceived as part of the natural body by the amputee and should replicate sensory-motor capabilities of the amputated limb. However, such an ideal "cybernetic" prosthesis is still far from reality: current prosthetic hands are simple grippers with one or two degrees of freedom, which barely restore the capability of the thumb-index pinch. This paper describes the design and fabrication of a novel prosthetic hand based on a "biomechatronic" and cybernetic approach. Our approach is aimed at providing "natural" sensory-motor co-ordination to the amputee, by integrating biomimetic mechanisms, sensors, actuators and control, and by interfacing the hand with the peripheral nervous system.

Index Terms—Biomechatronic, Prosthetic hand, Artificial Limbs, Actuators

I. INTRODUCTION

The objective of the work described in this paper is to develop an upper limb prosthesis that can be felt as a part of the body by the amputee (Extended Physiological Proprioception – EPP [1]), and that can substitute the amputated limb by closely replicating its sensory-motor capabilities ("cybernetic" prosthesis [2]).From the user point of view, current commercial prosthetic hands are unable to provide enough grasping functionality and to provide sensory-motor information.

Commercially available prosthetic devices, such as Otto Bock SensorHand™, as well as multifunctional hand designs [3,4,5] are far from providing the manipulation capabilities of the human hand [6]. This is due to many different reasons. For example, in prosthetic hands active bending is restricted to two or three joints, which are actuated by a single motor drive acting simultaneously on the metacarpophalangeal (MP) joints of the thumb, of the index and of the middle finger, while other joints can bend only passively.

The main limitations of commercially available prostheses are the following:

1. reduced grasping capabilities;
2. non cosmetic appearance;
3. lack of sensory information given to the amputee;
4. lack of a "natural" command interface.

In order to overcome these problems, we propose to develop a "cybernetic" prosthesis based on *biomechatronic* approach, i.e. on the design of a mechatronic system which integrates harmoniously artificial and natural components and modules.

The first and the second problems indicated above can be solved by increasing the number of active and passive DOFs. This can be achieved by embedding a higher number of actuators in the hand structure and by designing coupled joints. The third and forth problems can be addressed by developing a "natural" interface between the Peripheral Nervous System (PNS) and the artificial device (i.e., a "natural" Neural Interface (NI)) to record and stimulate the PNS in a selective way. A biocompatible neural interface can restore some sensory feedback to the amputee by stimulating in an appropriate way his/her afferent nerves (third problem), and can allow the motor control of the prosthesis based on ENG-based control (fourth problem). This will be possible by focusing appropriate research efforts on the technological development of the neural interface and on the characterization of the PNS afferent signals in response to mechanical and proprioceptive stimuli.
This approach is illustrated in Fig. 1.

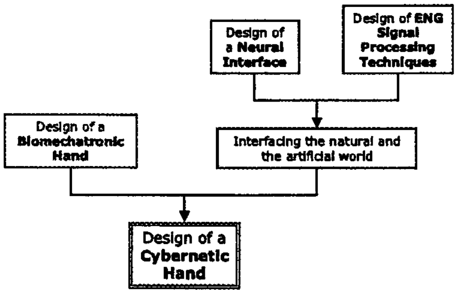


Fig. 1: Approach for the development of a cybernetic hand

In this paper we focus on the first and the second problems. Our approach is aimed at demonstrating that the introduction of a new generation of micro actuators allows to enhance the number of active joints. In order to control the prosthesis, we developed hall-effect position sensors for every active joint. Both sensors and actuators have been integrated inside the structure of the prosthetic hand, keeping the natural size of the human hand. This paper also reports preliminary tests on a first prototype of biomechatronic prosthetic hand.

II. DESIGN OF THE BIOMECHATRONIC HAND

A. Biomechatronic design

The main requirements to be considered since the very beginning of a prosthetic hand design are the following: cosmetics, controllability, noiselessness, lightness and low energy consumption. These requirements can be fulfilled by implementing an integrated design approach aimed at embedding different functions (mechanisms, actuation, sensing and control) within a housing closely replicating the shape, size and appearance of the human hand. This approach can be synthesized by the term: "*biomechatronic*" design [7]. Recently, a mechatronic robotic hand for teleoperation in space applications has been designed and fabricated [8]. An outstanding technological effort has been directed to obtain a polyarticulated hand with high integration of the actuators, sensors and embedded control.

The research proposed in this paper can be seen as a similar effort in order to apply mechatronic design to prosthetic hands. Actually, the field of prosthetics, compared to the field of space applications, requires to focus the design on different goals:, cosmetic appearance, absence of noise during operation, lightness and low cost. These requirements are related to the acceptability and affordability of the prosthetic hand by the final user who must feel the hand as a part of his/her body.

B. Architecture of the biomechatronic hand

The biomechatronic hand is equipped with three actuator systems to provide a tripod grasping: two identical finger actuator systems and one thumb actuator system (see Fig. 2).

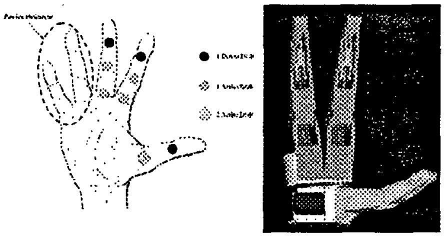


Fig. 2: Architecture of the biomechatronic hand

The finger actuator system is based on two micro-actuators, which drives the MP and the PIP joints respectively; for cosmetic reasons, both actuators are fully integrated in the hand structure: the first in the palm and the second within the proximal phalange. The DIP joint is passively driven by a 4-bars link connected to the PIP joint. The thumb is equipped with two active DOFs in the MP joint and one driven passive DOF in the IP joint.

The grasping task is divided in two subsequent phases in which the two different actuator systems are active:

1) reaching and shape-adapting phases;
2) grasping phase with thumb opposition.

In phase 1) the first actuator system allows the finger to adapt to the morphological characteristics of the grasped object by means of a low output torque motor. In phase 2), the thumb actuator system provides a power opposition force, useful to manage critical grasps, especially in case of heavy or slippery objects.

It is important to point out that the most critical problem of the proposed configuration is related to the strength required to micro-actuators to withstand the high load applied during the grasping phase.

In order to demonstrate the feasibility of the described biomechatronic approach, we started by developing one finger (index or middle), completed with the actuation system for the MP, PIP and DIP joints.

C. The actuation system

The adoption of bulky and heavy actuators, in the design of commercial upper limb prosthesis, lead to an extreme reduction of DOFs. Consequently, a stable grasp can be achieved by means of high grip forces. This design technique can be represented as a loop (see Fig. 3).

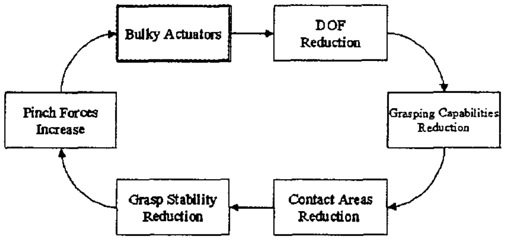


Fig. 3: Standard approach loop

The above schematization shows how this approach led to design hands with a maximum of two DOFs able to provide a pinch force of about 100 N.

The aim of our new approach (see Fig. 4) is to "invert" the previous scheme. This can be done by using micro-actuators, thus focusing on the increase of DOFs.

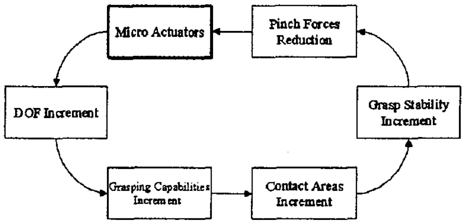


Fig. 4: Novel approach loop

We are developing a prosthetic hand actuated by micro-drives following the novel approach loop described in Fig. 4. Due to its enhanced mobility our hand will be able to increase the contact areas between phalanges and grasped object. According to this concept, a reduction in power actuation can be accepted if the contact area is increased. This also allows to increase grasp stability.

III. DESIGN OF THE FINGER PROTOTYPE

As outlined above, the two DOFs finger prototype is designed by reproducing, as closely as possible, the size and kinematics of a human finger. The finger has three phalanges and a palm housing, which is the part of the palm needed to house the proximal actuator (see Fig. 5).

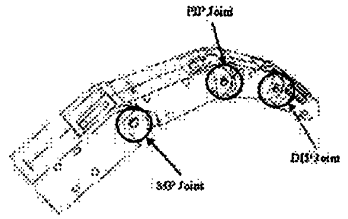


Fig. 5: The finger prototype.

A. Actuator system architecture

In order to match the size of a human finger, two micromotors are incorporated in the palm and in the proximal phalange, respectively. This high integration level is achieved by enclosing the motors in a shell housing, where they are constrained only by friction forces. The shell housing is obtained directly from the structure of the proximal phalange.

The actuator system is based on Smoovy™ (RMB, Eckweg, CH) micro-drivers (5 mm diameter) high precision linear actuators based on bi-directional DC brushless motors with planetary gears. The rotary motion of the shaft is converted to linear motion using lead screw transmission.

The main mechanical characteristics of the linear actuators as declared by manufacturer are listed below (see Table 1).

Gear stages	3
Transmission rate	1:125
Maximum load radial	25 N
Maximum load axial	40 N
Maximum speed	200 mm/min
Nominal force	12 N
Weight	3.2 g

Table 1: Main characteristics of the Smoovy™ (RMB, Eckweg, CH) micro drivers (5 mm diameter).

In principle, the selected actuator fulfils almost all the specifications for application in the prosthetic finger: small size, low weight and high bandwidth. The main problem encountered is related to noise, which in present implementation turns out to be too high to be tolerated by prosthesis users. Despite of this limitation, we decided to proceed with the application of the linear actuator in order to investigate integration problems and global performance. One possible solution for reducing noise caused by motors activation is to adjust the acoustical impedance of the motors housing and of the external palm/finger structure.

The output force resulting from motor activation is sufficient to move the phalanges for achieving adaptive grasp. In addition, the shell housing provides adequate mechanical resistance of the shaft to both axial and radial loads. This turns out to be essential during grasping tasks, where the loads generated by the thumb opposition act both on the actuator system and on the whole finger structure.

B. Kinematics architecture

The kinematics of each finger joint is described in the following subsections.

1) MP Joint

The proximal actuator is integrated in the palm and transmits the mechanical power through a slider crank mechanism to the proximal phalange providing flexion/extension movement. The slider is driven by the lead screw transmission directly mounted on the motor shaft.

2) PIP joint

The same mechanism used for the MP moves the PIP joint. Only the geometrical features are varied in order to fit the space available according to the specifications of the biomechatronic hand.

High friction forces occur, because of the low pitch of the threaded shafts, during mechanism movement. For this reason the two lead screw transmissions are non backdrivable. This choice turns out to be useful for ensuring that grasping forces are kept even without power supplying.

3) DIP joint

A 4-bars link has been adopted for the DIP joint and its geometrical features have been designed in order to reproduce as closely as possible natural DIP joint flexion. According to the three prescribed positions method [9] we synthesized the mechanism.

C. Fabrication of the finger prototype

A first prototype of the finger was fabricated using the Fused Deposition Modeling [FDM] process. This process allows to fabricate, in a single process, three-dimensional objects made out of acrylonitrile/butadiene/styrene [ABS] resin, directly from CAD-generated solid models.

IV. FINGERTIP FORCE CHARACTERIZATION

A first set of experimental tests has been performed in order to measure the force that the finger is able to make on an object [10]. To this aim we have measured the force resulting when the finger is pressing directly on a force sensor (3-axial piezoelectric load cell 9251 A, Piezo-Instrumentation KISTLER, Kiwag, CH), corresponding to different configurations of the joints.
Two "pressing" tasks were identified in order to evaluate separately and independently force obtained by the two actuators incorporated in the finger:

- TASK 1: the pushing action was exerted only by the distal actuator.

- TASK 2: the pushing action was exerted only by the proximal actuator.

For each task, two subtasks were identified according to the position of the non-active joint (extended, flexed). The different values of joint rotation angles corresponding to each subtask are illustrated in Fig. 6.

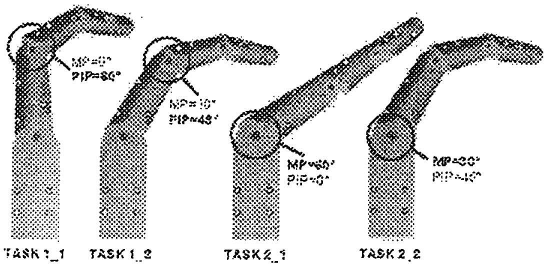


Fig. 6: Positions of finger joints for each task. The active joint for each task and position is indicated by a small circle.

During the force characterization the fingertip pushed on the force sensor. The Z force component was recorded, the X and Y outputs of the load cell were simply monitored. This was obtained by adjusting the finger position in order to obtain a force parallel to the Z-axis of the load cell. A first set of experimental tests was performed with the finger prototype, with the aim of evaluating how much force the finger is able to apply to an object.

A. Results and discussion

Ten tests were performed for each subtask. The results obtained are illustrated in Fig. 7. These force values are comparable with force exerted by "natural" human finger during fine manipulation, thus demonstrating the feasibility of the biomechatronic approach, at least for this class of manipulation tasks. The output force resulting from motor activation is sufficient to move the phalanges for achieving adaptive grasp [11].

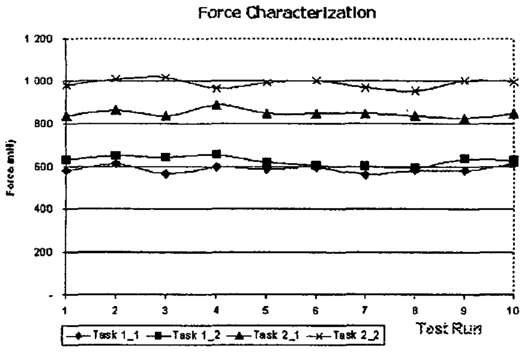
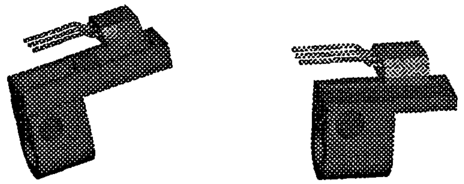


Fig. 7: Experimental results on finger force.

V. POSITION SENSORS

In order to control the position of the two active phalanges of the finger, the angular position of the two active joints (MP and PIP joint) should me . To this aim, two position sensor based on Hall-effect sensors (SS495A, Honeywell, USA) were integrated in the structure of the hand. The main advantages of Hall-effect sensors are small size and contactless working principle, a very important feature useful to decrease friction.

As illustrated in Fig. 8, the Hall sensors are fixed to the palm and to the proximal phalange, respectively, whereas the magnets are mounted on the sliders of the two joints.



(a) Slider for the MP joint. (b) Slider for the DIP joint.

Fig. 8: Drawings of the two position sensors. Size of (a) is 12x4x8 mm^3, and size of (b) is 8.7x4x6 mm^3).

In this configuration the sensor measures the linear movement of the slider, that is related to the angular position of the joint. In the MP joint, the sensor measure a linear movement of 5.2mm; in the PIP joint, the linear movement is 8mm.

In order to find the optimal configuration for the sensors, the distance between the magnets and the number of the magnets were varied. Then the covered range (expressed as the distance between the minimum and the maximum in the output curve) and the linearity, expressed as:

$$linearity = \frac{E}{|V_n - V_1|} \cdot 100 \text{ with } E = \sqrt{\frac{\sum_{i=1}^{n}(V_i - L_i)^2}{n}},$$

where n is the number of the measures, V_i is the voltage measured in the ith position, and L_i is the linearized voltage in the same position, were measured. The linearized voltage is a straight line from the minimum to the maximum of the output voltage.

Using a micrometric translator stage two optimal configurations were identified. The first one comprises two magnets are used at a distance of 3.5mm. This configuration provides an output range of 3.78V, a covered range of 5.4mm, and a linearity of about 5%. Therefore, this configuration is suitable for the PIP joint.

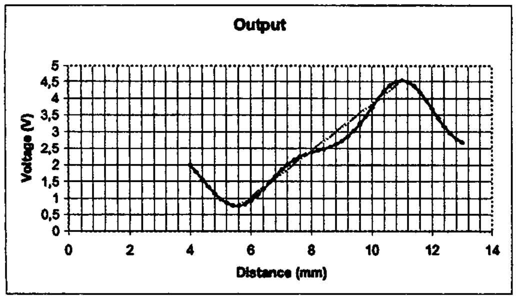


Fig. 9: Output voltage of the Hall-effect sensor in the first configuration used for the PIP joint.

The second configuration uses 6 magnets. The covered range is 8.4 mm with a linearity better than 4% f.s.o., as shown in Fig. 10. So this configuration is suitable to be mounted in the MP joint.

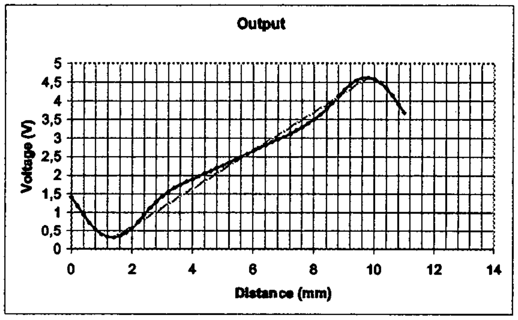


Fig. 10: Output voltage of the Hall-effect sensor in the second configuration used for the MP joint.

The sensors have been integrated in a new finger prototype which is depicted in Fig. 11.

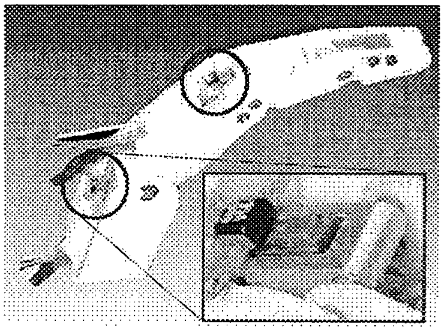


Fig. 11: Finger prototype with two Hall sensors integrated in it.

A. Characterization of the sensors

The sensors integrated in the finger has been characterized experimentally by means of a vision system and image analysis.

1) Experimental setup

A Nikon Coolpix 950 digital camera was mounted on a tripod in order to obtain a stable position perpendicular to the plane of movement of the finger, which was fixed in vertical position. The movement of each Smoovy actuator was driven by a CCS00001 Controller (RMB, CH). Each controller has a power supply of 11V, while each sensor was supplied with 6V.

100 different frames were acquired for each active joint, 50 for the flexion and 50 for the extension. For each frame the output value of the sensor was measured with a digital multimeter and recorded, and the relative position of the joint was measured using the module *Measure Tool* of Adobe PhotoShop 5.5, with a precision of 0.1°, according to the angles shown in Fig. 12. The mark points are the center of the brass pivots of the DIP and MP joint, respectively.

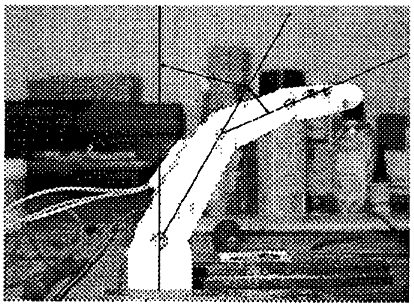


Fig. 12: The angles of the MP and PIP joints are measured using the brass pivots as mark points.

2) Results and discussion

Results are presented in Fig. 13 and in Fig. 14, respectively for the sensor in the MP joint and in the PIP joint. The flexion phase is indicated with small light circles, while the extension is indicated with small dark squares.

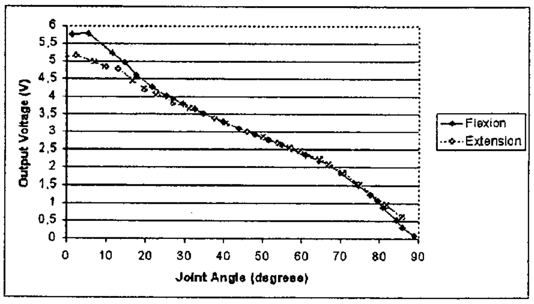


Fig. 13: Response curve for the MP joint.

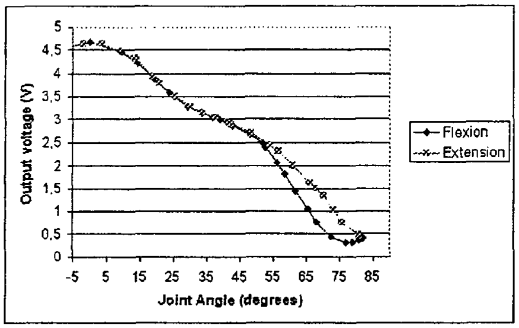


Fig. 14: Response curve for the PIP joint.

It is important to point out that both curves show low hysteresis. The difference between the flexion and the extension curve is due to the mechanical clearance of the sensorized slider, clearance that was quite pronounced during the change of direction of the joint (respectively occurring between 0 and 15 deg. for the MP joint, and between 55 and 90 deg. for the PIP joint).

VI. FUTURE IMPROVEMENTS

The experimental tests showed promising results, but there is still room for improvement. First of all, the movements of human fingers during grasping activities will be further investigated in order to achieve a truly "human-like" behaviour of the prosthetic finger. A micro-sensory system, incorporating also multi-component force sensors, will be integrated in the mechanical structure of the finger in order to sense incipient slippage and to obtain force sensing abilities. Finally, suitable control strategies will be investigated and applied in order to develop a smart and user friendly control interface for the prosthetic hand. In the future extensive experiments will be carried out in order to interface the artificial prosthesis with the peripheral neural system to extract sensory information [12] and to stimulate the nerves [13].

VII. CONCLUSIONS

A novel approach to the design and fabrication of innovative prosthetic hands, called *biomechatronic* approach, has been presented. This approach implies integrating together multiple degrees of freedom, multiple sensing capabilities, and distributed control in order to obtain "graceful" human-like appearance, simple and direct controllability, low weight, low energy consumption and noiselessness, which are important for the ultimate acceptability of the prosthetic hand.

Following this type of approach a first prototype of an active finger with two DOFs has been designed and fabricated. In this paper we focused our attention on the design and development of a first implementation of an innovative hand, based on the biomechatronic design.

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