This article has been accepted for publication in a future issue of this journal, but has not been fully edited. Content may change prior to final publication. Citation information: DOI 10.1109/TNSRE.2018.2829183, IEEE Transactions on Neural Systems and Rehabilitation Engineering



# The *S-Finger*: a synergetic externally powered digit with tactile sensing and feedback

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***Abstract*—Partial hand amputation is by far the most common type of amputation worldwide. Nevertheless, regardless of their potential clinical and socioeconomic impact, battery-powered partial hand prostheses, namely powered digits, have modestly progressed so far and very few clinical solutions are available today. Here we present a mechanical architecture, an alternative to state of the art solutions, which exploits a high efficiency, non-back drivable mechanical transmission based on a face-gear pair and a miniaturized clutch. We took inspiration from the synergetic prehension approach proposed by Childress for whole hand amputation. The finger was equipped with a myoelectric controller and a tactile sensor able to provide users with *discrete events* sensory feedback. Measured speed (90°/s) and force (6.5N) of the newly dubbed *S-Finger* proved comparable with those of clinically available prostheses. The design demonstrated to be compact and rugged enough to undergo a clinical viability test with two partial hand amputees, fitted with custom three-fingered research-prostheses using the *S-Finger*. The subjects successfully completed several dexterity tests and gave relevant feedback for the development of a second generation device. These results contribute to the increasing research endeavors in the field of partial hand amputation.**

***Index Terms*— Partial hand prosthesis, Face-gear mechanism, Prosthetics, Upper limb amputation, Synergetic prehension.**

## I. Introduction

PARTIAL hand amputation is the most common type of amputation injury in the upper extremities, representing 90% of all upper limb amputations [1]. Although its incidence is considerable, the treatment of partial hand amputation has modestly progressed so far, compared to, e.g., the remarkable advances of articulated (full) hands or leg prostheses [2], [3]. The difficulty in prosthetic treatment is due to two principal reasons. Firstly, the wide range of anatomical and functional presentations that makes it difficult to find standardized and scalable solutions. Secondly, the technological complexity in replacing the motor and sensory function of a lost finger in the same size as a digit [4].

Akin to prostheses for higher levels of amputation, partial hand prostheses can be divided into passive and active (powered) devices; the latter can be either body or externally powered (i.e., powered by a battery). This work is focused on battery-powered prostheses, however the interested reader may find a comprehensive review of available options elsewhere [4], [5]. Battery-powered prostheses are motorized devices activated by user-dependent input signals, picked-up by the individual's residual limb and processed to control an intentional opening or closure of the terminal device. State of the art techniques for the detection of voluntary input signals include surface electromyography (EMG) or the recording of subtle pressure forces of the residuum against the socket [6], [7].

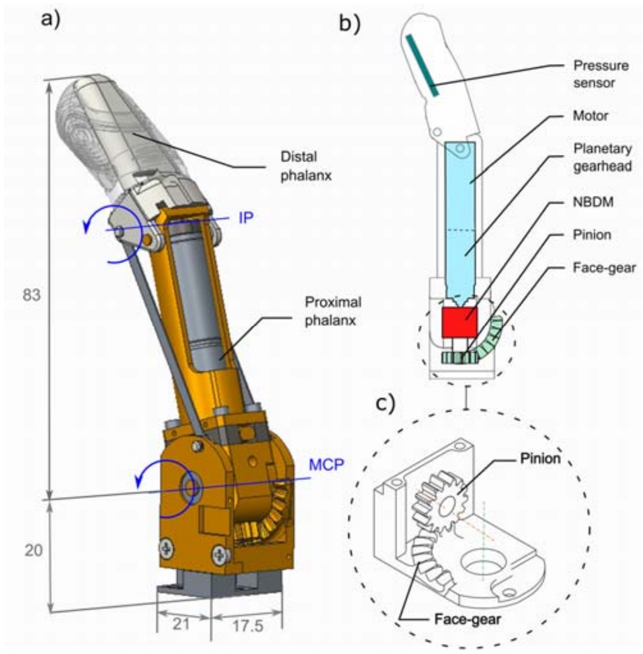


Fig. 1 a) Mechanical architecture and overall dimensions (in mm) of the *S-Finger*. b) Lateral section highlighting the placement of the different components. c) Detailed view of the arrangement of the pinion and of the face-gear. Acronyms: MCP – metacarpal-phalangeal (joint); IP – inter-phalangeal (joint); NBDM – non-back drivable mechanism.

While passive and body-powered prosthetic digits have been widely accessible for many decades [8], battery-powered solutions did not become clinically available until 2007, when Gow and colleagues presented the *i-Limb* digits [9] (Touch Bionics Inc, Livingstone, Scotland). The *i-Limb* digit contains

Manuscript received July 21, 2017. This work was supported by the National Workers' Compensation (INAIL) under the PPR3 grant.

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a brushed DC motor with integrated gearhead and a non-back drivable worm gear pair. In prosthetics, non-back drivability is required to maintain a certain grip force without consuming battery power [10], [11]. The *i-Limb* digit includes two phalanxes, one rotating around a metacarpo-phalangeal (MCP) joint (coincident with the worm gear pair) and the other around a distal joint. A tendon transmission links the flexion/extension of the MCP joint to the flexion/extension of the distal joint, following a fixed kinematics that mimics the closure of the human digit. The *Vincent* finger [12] (by Vincent Systems GmbH, Karlsruhe, Germany), represents a clinically viable alternative to the *i-Limb* digit, even though it shares with the latter a similar motor and mechanical transmission. The *Vincent* finger is built around a worm gear pair and a four-bar linkage between the MCP and the distal phalanx. The transmission is partially back drivable: the digit can be repositioned by applying an external force on it [13].

Partial hand prosthetic management represents an exciting frontier in the specialty of upper limb prosthetics [14]. Nonetheless, besides *i-Limb* and *Vincent* fingers, which are very similar, there are no other individually powered digits for amputations at the transmetacarpal level. In addition to that, paradoxically, none of these modern powered digits retain the sensory awareness that the mechanical transmission used to provide to body-powered prosthesis users [4]. Amputees wearing myoelectric prostheses mostly rely on vision or on other sensory cues (e.g., motor sound) to regulate grasp with significant cognitive effort, especially when either the thumb or all fingers are missing. In contrast, it was recently shown that sensory feedback may reduce phantom limb pain [15] while enhancing closed-loop controllability and embodiment of the prosthesis in amputees [16]-[18].

Another basic observation is that, in 95% of the cases, partial hand amputees retain close-to-normal use of either the thumb or of one (or more) of the long fingers [4]. Hence, since the sound part is capable of providing enough force to grasp, the prosthesis should only be able to maintain/hold the grip against it while providing enough grip aperture and speed.

Building on these observations, we designed a novel prosthetic finger, integrated with an unobtrusive vibrotactile sensory feedback system. We explored a mechanical transmission alternative to the already available solutions. In particular we designed a new, compact transmission based on a face-gear pair [19] coupled with a bidirectional non-back drivable roller clutch [11]. The face-gear is a high efficiency, low-ratio (2:1) reduction stage whereas the roller clutch makes the system non-back drivable without decreasing the efficiency of the overall transmission. In this way, the digit exhibits speed comparable to the commercial available prostheses and, although unable to apply large active forces, it is able to sustain large passive loads. This architecture was based on the *synergetic prehension* approach proposed by D. Childress [20], [21] and for that reason the finger was dubbed *S-Finger* (i.e. synergetic finger).

The finger was equipped with a bidirectional human machine interface (HMI) able to control it using EMG signals and to provide users with *discrete events* sensory feedback akin to our previous work [18], [22]. The proposed design proved compact and rugged enough to undergo a clinical viability test with two partial hand amputees, fitted with custom three-fingered research-prostheses using the *S-Finger*. Our results demonstrate the feasibility of building an effective prosthetic digit based on the synergetic prehensor and contributes to the increasing research endeavors in the field of partial hand amputation.

## II. Requirements for a powered digit

The main goal of a prosthetic digit is to restore an acceptable level of motor function, typically the restoration of the flexion/extension movement to support opposition. As a target for the performance, we took inspiration from the natural hand: average physiological speeds for everyday pick-and-place tasks are in the range of 3 to 4 rad/s (170-230 °/s), while most ADLs require prehension forces in the range of 0-65 N [23]. The experience gained in the field has taught us that these figures may be revised in order to comply with the other important requirement, namely the size of the digit. Indeed matching the anatomical size and proportions of the human finger is extremely important in order to fit a wide range of users. Based on this we chose the 35th percentile female index finger as the target size [24]. Finally, the digit

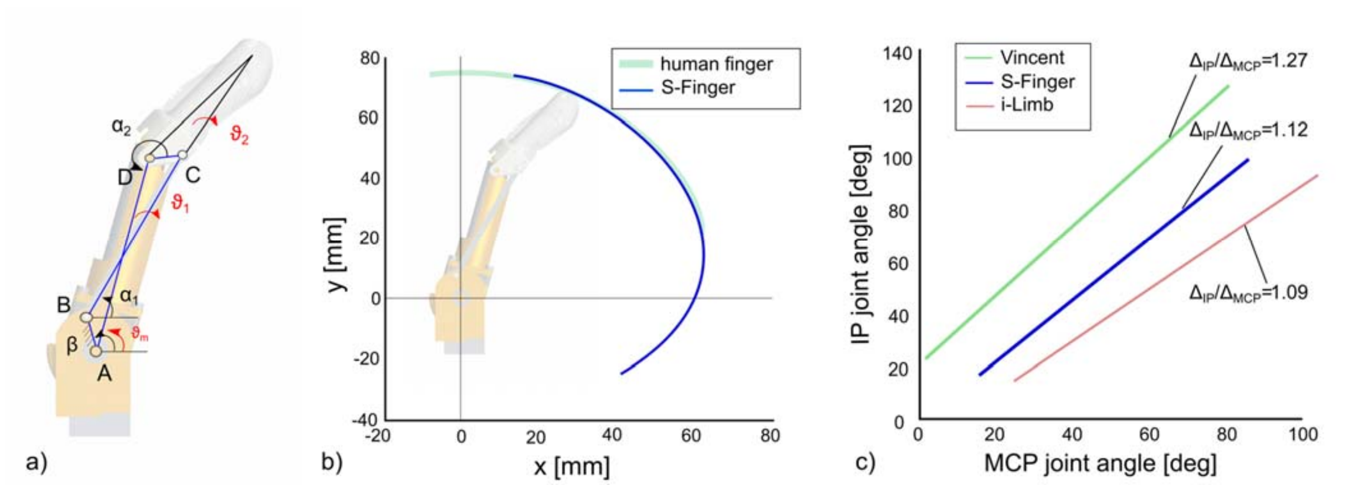


Fig. 2 Kinematics of the *S-Finger*. a) Scheme of the inverse four-bar mechanism. b) Trajectory of the fingertip compared with human data (adapted from [28]). c) Relationship between the IP (inter-phalangeal) and MCP (metacarpal-phalangeal) joints of the *S-Finger* (red) and other commercial prostheses (as calculated by Belter et al. [2]). The figure next to each line is the slope of the curve.

should be thought of as a modular unit of a partial hand prosthesis and should be able to cover different levels of amputation ranging from the transmetacarpal distal to the transmetacarpal proximal.

To comply with the target features (i.e. size and mechanical power) using off-the-shelf, low power, miniaturized motors the mechanical efficiency of the transmission must be maximized. In fact an efficient prosthesis requires smaller and lighter batteries, thus reducing the weight of the system. Nevertheless the battery should ensure a full-day operation without being recharged.

Our design considered the prevalent case of amputations with either a retained thumb or retained long fingers. In this case the prosthesis may simply counteract the forces produced by the sound digit(s) by opposing them. Hence a non-back drivable mechanism is required.

With regards to sensing and to the possibility of conveying sensory feedback, our starting point was the *Discrete Event driven Sensory feedback Control* (DESC) policy [25]. The latter was demonstrated effective in enhancing closed loop control in complete and partial hand amputees wearing a prosthesis with non-invasive vibrotactile feedback [22], [18].

## III. Design of the *S-Finger*

The *S-Finger* is comprised of two joints: a proximal joint (equivalent to the MCP in humans) and a distal joint, here referred to as inter-phalangeal joint (IP) (Fig 1a). The MCP joint is directly actuated and coupled to the IP joint through an inverse four-bar linkage mechanism. The MCP joint is actuated by a brushless DC motor (EC8, 2W, Maxon Motor, Switzerland), with integrated planetary gearhead (GP8, 1024:1 reduction ratio, Maxon Motor, Switzerland). A non-back drivable mechanism (NBDM), designed following the approach described in [26], is attached to the output of the gearhead (Fig 1b). The output of the NBDM is coupled with a face-gear pair [19] (2:1 reduction ratio) to the frame supporting the MCP joint axis. The distal phalanx of the prosthesis embeds a pressure sensor in order to detect contact forces of the fingertip. The finger is controlled in real time by a custom driver that processes EMG signals picked up from the residual limb. The driver also processes fingertip sensory signals in order to deliver sensory feedback to the user following the DESC approach [25], [22].

### A. Kinematics of the finger

In the four-bar mechanism (ABCD in Fig 2.a) the rotation of a proximal phalanx (AD) around the MCP joint (pivoted in A) is coupled with the rotation of the distal phalanx (integral with CD) around the IP joint (in D). This architecture is well-known for being robust and has been used in several state of the art powered digits [2], [4] and prosthetic hands [27]. The kinematics of the mechanism, which describes the relation between the input rotation ($\theta_m$) and the rotations of the proximal ($\theta_1$) and distal ($\theta_2$) phalanxes, was derived in [27] and not reported here. The kinematics was used to simulate the motion of the finger (using different lengths of the links and different parameters, cf. Fig 2) in order to find the optimal configuration that would (i) be compatible with the target size and (ii) mimic the trajectory of the human fingertip during grasping (Fig 2b) [28]. The chosen configuration replicates the kinematics of the human finger (Fig 2b), akin to other state of the art prostheses (Fig 2c).

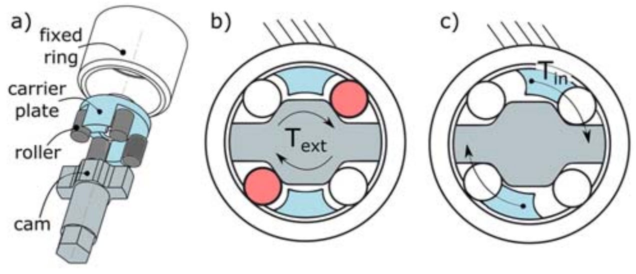


Fig. 3 The non-back drivable mechanism of the *S-Finger*: a) Exploded view. b) Cross section showing the locking condition: the rollers (colored in red) are wedged in between the outer ring and the cam due to the action of the external torque $T_{ext}$. c) Cross section in the unlocking condition: the motor torque $T_{in}$ applied through the carrier plate to the rollers unlocks the mechanism, thus dragging the cam.

### B. Actuation

An off-the-shelf motor (EC8, 2W Maxon Motor, Switzerland) with integrated planetary gearhead (1024:1 reduction ratio) was used to actuate the finger. Unlike the commercial digits available we chose a brushless motor. Although they require more complex control electronics, brushless motors exhibit a greater power density with respect to brushed ones. Because of this it was possible to choose a smaller/lighter motor compared to existing digits (Ø 8 mm, instead of Ø 10 mm).

### C. Non-back drivable mechanism (NBDM)

The requirements of non-back drivability and high efficiency in the direct motion were matched by designing a bidirectional NBDM based on wedge phenomena [10], assembled at the output of the motor [11], [26]. This mechanism is composed of a fixed ring fastened to the frame, a shaft/carrier-plate shaped with a two-tooth extrusion (input axis), a cam (output axis) and four rollers (Fig.3a). It is housed in the proximal joint, connected to the motor shaft by means of a form-locking groove in the input side, and to the pinion of the face-gear pair through the output side. When an external torque is applied to the output, the cam wedges the rollers between itself and the ring, thus locking (Fig.3b). Driving the mechanism from the input axis instead, causes the teeth of the carrier plate to move the rollers away from the wedging position and to disengage the mechanism (transmission permitted). The efficiency of the direct motion of the NBDM is ~95% as shown in our previous work [11].

### D. Face-gear pair

The face-gear is a well-known back-drivable, compact, robust and efficient transmission (with efficiency $\eta_{\mathrm{FG}} \approx 0.96$) [19]. It is composed by a spur cylindrical pinion (input) that meshes with a conjugated plane gear with frontal teeth (output) [19]. From a kinematic point of view, it transmits the motion between orthogonal and incident axes, representing a particular case of the bevel gear set [29]. The main advantage is that the axial motion of the pinion does not affect the contact pattern and the backlash. This facilitates the assembly because it requires neither the pinion to be assembled with strict tolerances nor to place thrust bearings, since no axial force is generated [19], [30].

In the *S-Finger* the pinion is coaxial with the NBDM output

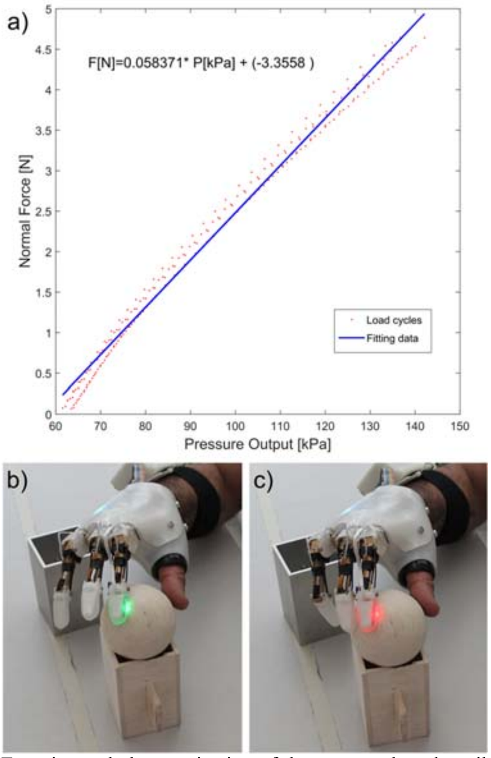


Fig. 4 a) Experimental characterization of the pressure based tactile sensor using an automatic positioning platform (VT-80, Pi miCos GmbH, Germany) instrumented with tri-axial load cell (NANO17, ATI Industrial Automation, USA) as in [31]. b-c) Demonstration of the LED feedback: before contact the LED is green (b), at contact the LED turns red (c).

and is orthogonal with the axis of the MCP joint. The plane gear is integral with the frame, which also embeds the MCP rotational joint (Fig 1c). The face-gear pair was designed with a 2:1 reduction ratio, which combined with that of the motor gearhead (1024:1), produces a global ratio of 2048:1.

The efficiency of the whole transmission $\eta_{TOT}$ (i.e. NBDM and face-gear) can be estimated to ~91%. This is a significant enhancement when compared with state of art powered digits using worm gear pairs ($\eta \leq 0.5$) [11].

### *E. Sensory system*

The fingertip was equipped with a tactile force sensor which can be used to trigger sensory feedback to the user. The tactile sensor was based on a pressure sensor integrated circuit (MPL115A2, Freescale, USA) embedded into a compliant silicone-rubber matrix similar to the sensor developed by Jentoft et al. [32]. The relationship between the pressure and the force on the fingertip was found to be linear in the range 0-5 N (Fig.4-a). This was true for a sensitive area not smaller than a square of 6x6 mm$^2$ centered in the middle of the fingertip. Besides the tactile sensor the finger was equipped with a current sensor and a Hall effect limit switch to detect the homing position.

### *F. Embedded bidirectional controller*

The *S-Finger* is controlled by an 8-bit microcontroller board able to: (i) recognize external commands (on a serial bus) or

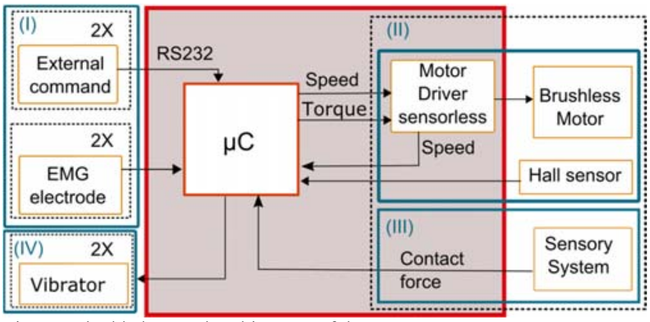


Fig. 5 Embedded control architecture of the *S-Finger*.

acquire two EMG signals, (ii) drive multiple independent fingers, (iii) acquire/process in real time the fingertip sensors, and (iv) provide sensory feedback (Fig.5).

As this work focused on the design of the digit, we did not implement so far state of the art controllers, like those currently being explored by other groups [33], [34]. Instead a two-state amplitude modulated controller, acquiring a differential pair of EMG sensors, was implemented [35], [36]. The controller drives multiple fingers in parallel albeit it may stop them, individually when contacting an object, in order to conform the grasp to its shape. In particular each finger is driven until its motion is hindered for a certain time (~0.5 s). Akin to commercial prostheses, the current (and thus the torque provided by the finger) is delivered in a series of *bursts* until it overcomes a pre-programmed threshold (the so called *pulse mode* in [2]).

The controller acquires the tactile sensor to detect salient events like *touch* and *breaking of touch* in order to produce DESC sensory feedback [25], [22] using a miniature vibrating motor (to be placed on the skin of the amputee). The sensor is also used to drive a colored LED embedded in the fingertip, as additional visual feedback, akin to [37]. The LED is normally green and turns red when the fingertip gets in contact; the intensity of the red light is modulated by the fingertip force (Fig.4 b-c). This additional feedback was included not only for investigating scientific questions pertaining to sensorimotor integration In fact, the intensity of the light (i.e. the fingertip force) could help the prosthetist in aligning the digits while manufacturing the socket [4].

## IV. Prototype Evaluation

The finger components were manufactured using technologies and materials based on their function and shape. Complex geometries (i.e., gears and carter) were manufactured in brass using lost wax casting. The distal phalanx frame was 3D-printed in nylon (PA2200); the latter was chosen due to its good mechanical properties (tensile strength $S_u$=48 MPa, density ρ=0.93 g/cm$^3$, Young's modulus E=1700 MPa) and simple manufacturability. The fingertip was molded in silicone rubber (shore 40A, SORTAClear, Smooth-On Inc. USA). The developed prototype weighs 70 g, excluding the electronics and the batteries (normally housed in the prosthetic socket and in a wrist bracelet, respectively).

Important parameters of the prototype including: power consumption, speed, grip force and opposition strength were assessed and compared with those of clinically available fingers (Table 1). Furthermore the ability of the *S-Finger* to oppose to unimpaired long fingers and grasp (relatively) heavy loads was emulated and assessed. Finally, a clinical viability test with two partial-hand amputees was conducted.

### A. Power Consumption

The power consumption of the embedded controller is roughly 0.5 W (55 mA @ 12 V), which coincides with the instantaneous power required to maintain the finger in a fixed posture. The 12 Volts version of the EC8 motor was preferred to the 6 V version because of its better efficiency; while requires lower currents it provides higher stall torque. The no-load consumption at 100% speed for a complete closing is 2.39 J, whereas 2.6 J are required to grasp an object at 80% of full speed. Based on these measurements, it was estimated that a 12V battery with 1 Ah capacity (20 x 34 x 50 mm, ~60 g) could ensure ~5000 grasps [38].

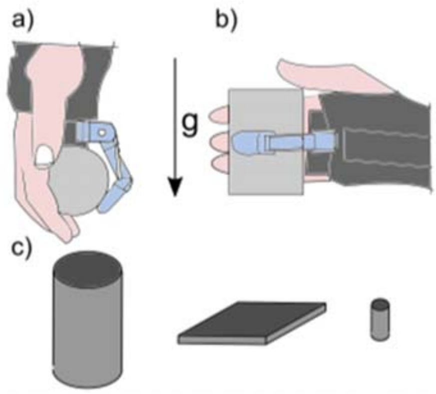


Fig. 6 Experimental orientations during the emulated grasping assessment. a) Normal orientation. b) Transversal orientation. c) Lifted objects.

### B. Flexion/Extension Speed

The maximum flexion/extension speed was assessed by means of video analysis. The finger moves from the extended to the flexed position within 0.7 seconds. As the angular span around the MCP is 75° this gives an average maximum speed of the MCP joint of ~90 °/s.

### C. Grip Force and critical load of the NBDM

The maximum active force produced by the *S-Finger* and the maximum passive force it is able to sustain were assessed. Noteworthy, the digit was conceived to ensure prehension in synergy with and in opposition to the remnant biological fingers. Hence the passive force is more relevant than the active one.

TABLE 1
PERFORMANCE OF POWERED DIGITS (ADAPTED FROM [14])

| Name (year) [developer] | $V_{max}$ (°/s) | F (N) | $F_o$ (N) | weight (g) |
|---|---|---|---|---|
| ***S-Finger*** **(2017) [SSSA]** | 90 | 4.3 (6.5*) | 24.2 | 70 |
| ***Vincent*** **(2010) [Vincent System]** | 103 | 4.82 (8.44*) | - | 40 |
| ***i-Limb*** **(2009) [Touch Bionics]** | 95 | 5.39 (6.54*) | - | 50 |

*Values obtained by controlling the finger using the pulse mode as in [2].

The maximum active force applied by the fingertip, *F*, was measured following the procedure used by Belter et al. [2] for a direct comparison with other powered digits (Table 1). The finger, in fully extended position, was secured to the basement of an instrumented platform. Then it was driven to flex at full torque and speed against a multi-axis load-cell (NANO17, ATI Industrial Automation Inc., USA) rigidly fixed on the frame and facing the fingertip. In this condition, as the moment arm is maximum, the force transferred to the fingertip is the lowest possible. The force after the impact (*F*) was recorded, and this procedure was repeated 15 times in order to get an average value. The average holding force is 4.3 N that can be increased to about 6.5 N using pulse mode akin to the commercial prosthetic digits [2]. The maximum passive (opposing) force, $F_o$, depends on the critical load (CL) of the NBDM [11], which is defined as the maximum torque sustained by the NBDM. The CL experimentally assessed proved to be 0.89 Nm. This means that the finger can withstand forces (at the fingertip) up to 24.2 N during a grasp.

### D. Emulated Grasping Assessment

The functionality of the *S-Finger* was assessed in tests aimed to evaluate its ability to oppose to unimpaired long fingers, in different conditions. For this purpose the finger was fixed to a wrist orthosis in opposition to the long fingers, simulating an opposed thumb (Fig. 6). An individual (non-amputee) was asked to control the finger using the sEMG interface to repeatedly lift objects wearing the prosthesis. Three grasp types (and objects) were tested: power (grasping a plastic cylinder, Ø 65 mm), pinch (grasping a small tube, Ø 14 mm) and extension (grasping an aluminum plate 7 mm thick). The objects were connected to weights, starting from 0.5 kg, and increased in steps of 0.5 kg if at the previous step no slippage had occurred. This series continued until the object started to slip and thus the grasp failed. The test was performed in two different orientations of the hand (Fig. 8) so that the weight of the object forced the hand to open along the normal (Fig. 8a) or transversal (Fig. 8b) directions. Normal and transversal direction were referred to the direction of the gravity force with respect to the finger sagittal plane. Normal force was applied in the sagittal plane of the finger while transversal direction was orthogonal to that plane.

Weights up to 3.5 kg for the cylinder, 3 kg for the tube, and 2 kg for the plate, were successfully lifted in the normal configuration (Table 2). In the transversal condition, we recorded 3.0 kg for the cylinder, 2.0 kg for the plate, and 1 kg for the tube (Table 2).

TABLE 2
LIFTED WEIGHTS IN THE EMULATED GRASPING ASSESSMENT

| Object | Grasp type | $m_n$ (kg) | $m_t$ (kg) |
|---|---|---|---|
| *cylinder* | power | 3.5 | 3.0 |
| *plate* | extension | 2.0 | 2.0 |
| *tube* | pinch | 3.0 | 1.0 |

### E. Clinical Viability Test

The clinical viability of the device while performing conventional dexterity tests was investigated. Notably, rather than providing measurable and comparable outcomes, this pilot served to collect feedback and suggestions from target patients while using the *S-Finger*.

Two subjects were recruited by the INAIL Prosthetic Centre (Budrio, BO, Italy). The inclusion criteria were: trans-metacarpal amputation, myoelectric prosthesis user, physically and mentally healthy. Informed consent in accordance with the Declaration of Helsinki was obtained before conducting the tests from each subject. The study was approved by the local Ethical Committee of Bologna-Imola (CE-BI), Italy, and was carried out in accordance with the approved guidelines.

Both subjects had the four long fingers of their right hand amputated and unimpaired thumb. They had been fitted 2-3 years before this study with a prosthesis employing i-Limb digits, albeit they were not frequent users. Each subject was fitted with a custom three-fingered research-prosthesis using the *S-Fingers*, in order to reduce the weight of the device and facilitate the tests (Fig. 7). One EMG sensor (MyoBock 13E200, Otto Bock, Germany) was placed on the thenar muscle (close signal), and the other on the hypothenar muscle (open signal). The index finger included the tactile sensor which was used to provide supplementary visual (LED) and DESC feedback (by means of a vibrator hosted in the bracelet on the forearm). In the same day the subjects received the prosthesis, learned how to use it and performed two standardized dexterity tests (the *Box and Block test (BBT) [39]and the Nine Hole Peg Test (NHPT) [40])*, with and without the supplementary sensory feedback. The participants also completed a short evaluation questionnaire aimed to evaluate the functionality and satisfaction of the device.

The *S-Fingers* proved reliable and allowed both subjects to successfully complete the different dexterity tests (Fig. 7). To illustrate this, we provided the results in Table 3 and one video clip as supplementary material, showing the subjects while using the device. Both subjects reported that the prosthesis was useful during the tests and perceived an improved stability, with respect to their own usual prostheses. Both participants appreciated the sensory feedback and one reported that it was useful to complete the dexterity tests. Noteworthy, both subjects described the noise produced by the *S-Finger* lower than that of their own prosthesis. One subject reported his willingness to use the *S-Finger* also after the end of the study.

TABLE 3
DEXTERITY TESTS PILOT RESULTS

| | ***Subject 1*** | | ***Subject 2*** | |
|---|---|---|---|---|
| ***Session*** | ***BBT*** | ***NHPT*** | ***BBT*** | ***NHPT*** |
| | *Transferred Blocks (#)* | *Time (s)* | *Transferred Blocks (#)* | *Time (s)* |
| ***Session 1*** | 9 | 153 | 25 | 239 |
| ***Session 2*** | 10 | 111 | 19 | 220 |

Acronyms: BBT - Box and Block Test; NHPT - Nine Hole Peg Test.

## V. DISCUSSION

We described the work done towards the design and development of a sensorized powered digit based on the *synergetic prehension* approach. The latter was proposed by Childress starting from the observation that limited work is required during grasping [4], [20]. In the synergetic prehensor one finger provides high speed and excursion but little force (fast side), while the other (in opposition) produces high force with low speed and excursion (force side). In this way, opposed digits work in synergy to boost the overall speed-force performance. We propose the *S-Finger* as an effective solution for the fast side of a synergetic prehensor in a partial hand prosthesis in which the force side is accomplished by the sound digit(s). This was obtained by developing a miniaturized non-back drivable clutch that resists against external loads and exhibits higher efficiency compared to conventional worm gear transmissions.

The embedded controller was designed as a trade-off between flexibility and size. Akin to the other clinically available powered digits, it was deemed unnecessary for the control to coordinate the overall posture of the fingers. Hence

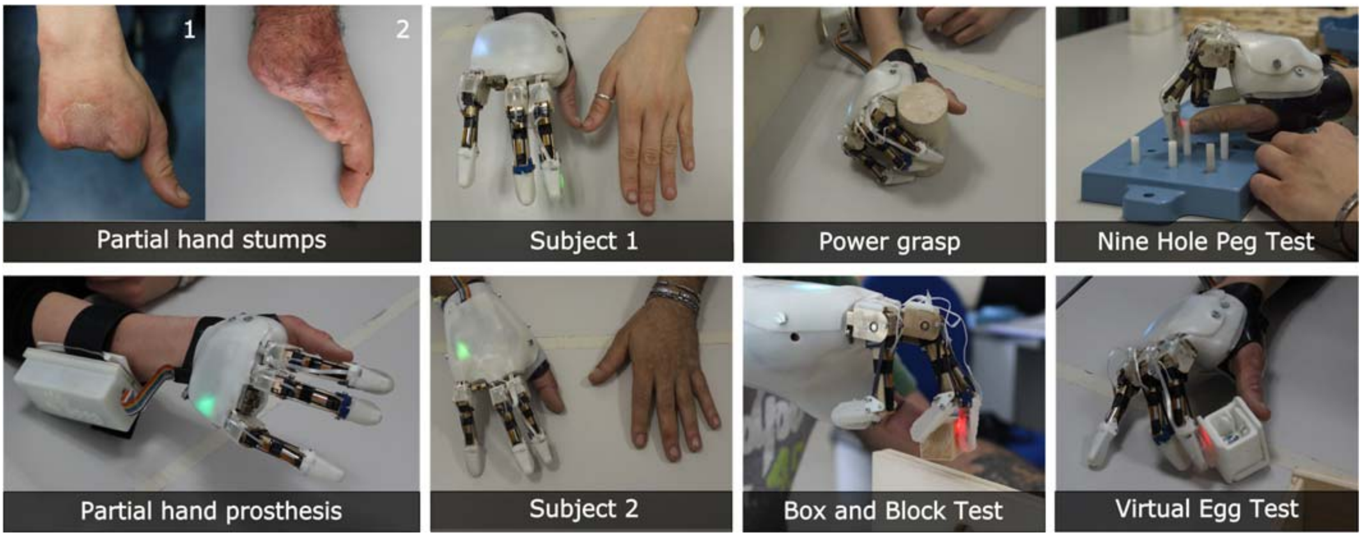


Fig. 7 Three fingered partial hand prostheses fitted to two subjects. The pictures depict scenes from the clinical viability test.

position control was not necessary. This allowed to employ a single microcontroller and thus to reduce the complexity and size of the control board. In addition, since the position control was not required, the motor could be controlled using a *sensorless* driver. In that way the number of wires required to drive each finger was reduced to three (instead of eight as required in a *sensored* brushless architecture). This choice allowed to increase the reliability of the finger, minimizing the probability of failures that may occur due to the moving cables.

We considered of primary interest to provide supplementary sensory feedback to the subject. In this framework, we implemented a vibrotactile and a visual feedback. Interestingly, the latter could also aid the prosthetist in aligning the digits on the socket in order to ensure appropriate opposition to the sound finger(s). This could effectively reduce the time and costs needed to manufacture partial hand prostheses [4], [5].

The weight of the prototype is similar to that of commercial digits, albeit our design and manufacturing techniques were not optimized for this purpose. We estimate that using lighter materials and different manufacturing methods, that are cost-affordable only for mass production, may allow reducing the weight of the finger by 40%. Notably, the *S-Finger* requires a finger absence of 20 mm proximal to the MCP joint (Fig. 1a); both the *i-Limb* and the *Vincent Finger* require an absence of 9÷12 mm, depending on the fitting assembly arrangement [5].

The *S-Finger* showed performance comparable to commercially available prostheses (Table 1): our design produced a slightly lower active force due to the integrated planetary gearhead used. In particular, the input speed-torque of the gearhead, recommended by the manufacturer, was lower than the maximum speed-torque produced by the motor. Hence, we believe that a custom design of the gearhead could allow to exploit the full potential of the brushless motor and of the efficient transmission. Regardless this limitation, the weights lifted using the *S-Finger* as an opposition post were remarkably high (Table 2). The lesson learnt is that when a sound finger actively grasps, what is of most importance is the capability of the powered digit to resist (i.e. to oppose). It should be noted that in the emulated grasping assessment the difference between the performance in the normal and transversal directions was likely due to the relative placement of the artificial finger with respect to the object. In the normal direction the prosthesis sustained a part of the load with both phalanxes while in the transversal direction all the load was applied on the fingertip. Hence the grip was ensured only by the friction of the silicone fingertip. The emulated grasping assessment was performed during the development of the *S-Finger* (ahead of the clinical viability test). This early test prevented partial hand amputees to perform heavy duty tests, while allowing us to safely assess the limits of our design, in the laboratory setting.

The clinical viability test confirmed the robustness of the prototypes. In particular, the two amputees were fitted with the same three fingers. The control strategy was easy enough that the subjects became familiar with within few minutes. Both subjects agreed that the S-Finger improved the grasp stability during the execution of the tasks. This fact was probably due to the compliance of the fingertip of the prototype, which increased stability during precision grasps. The questionnaire also evidenced that the supplementary sensory feedback increased the perceived stability of the grasp. Although largely subjective (the positive feedback by the users is a common outcome from these studies [41], [42]), the experienced improvement may be attributed to the efficacy of the DESC approach [25], [22], [18]. The tests highlighted that there is room for improvements. The force sensor based on a single barometer represented a robust and cost effective trade-off but the sensitive area proved too small. In particular, the sensor was shown to be efficient only for force applied along the sagittal plane, as in the precision or power grasps. Anecdotally during lateral grasps the sensor frequently failed to trigger the contact. Hence one possibility is to use multiple sensors in order to expand the sensitive area of the fingertip as well as its directionality (e.g. for the lateral grasp).

A weakness of the present work is that only two individuals with partial amputations participated in the study. However, we were constrained by a lack of access to such individuals in the time frame available to us. So we chose to conduct this exploratory study by enrolling the only two available people. Nevertheless, our results demonstrate the feasibility of building an effective partial hand prosthesis based on the *synergetic prehensor* idea. While we deem the architecture promising, there is still work to be done in order to reduce the weight and size of the prosthesis for an improved autonomy.

## Acknowledgment

The authors would like to thank M. D'Alonzo for his precious help, P. Randi for the support in the tests, C. Stagni and G. Maldina for building the prosthetic sockets, as well as the volunteers that took part in the tests.

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