HandCARE: A Cable-Actuated Rehabilitation System to Train Hand Function After Stroke

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***Abstract*—We have developed a robotic interface to train hand and finger function. *HandCARE* is a Cable-Actuated REhabilitation system, in which each finger is attached to an instrumented cable loop allowing force control and a predominantly linear displacement. The device, whose designed is based on biomechanical measurements, can assist the subject in opening and closing movements and can be adapted to accommodate various hand shapes and finger sizes. Main features of the interface include a differential sensing system, and a clutch system which allows independent movement of the five fingers with only one actuator. The device is safe, easily transportable, and offers multiple training possibilities. This paper presents the biomechanical measurements carried out to determine the requirements for a finger rehabilitation device, and the design and characterization of the complete system.**

***Index Terms*—Cable system, hand and finger functions, human-oriented design, rehabilitation robotics.**

I. Introduction

POSTSTROKE rehabilitation starts with one-on-one therapy with physiotherapists in acute-care hospitals. To limit the cost of treatment, patients are usually sent back home when they are able to walk, even if they have not recovered complete function of upper limbs, especially of distal parts, i.e., hands and fingers. In most cases, it will take a longer time to recover the functions of extension, abduction, and adduction of the fingers, thereby leaving the fingers in a flexed position, resulting in difficulties with activities of daily living (ADL) such as grooming, dressing, eating, and personal hygiene [1]–[5].

It is, therefore, usual to pursue further rehabilitation at home, with the advantages of practicing skills and developing compensatory strategies in the context of one's own living environment. Stroke patients are generally instructed to perform different exercises with the hand in order to restore physical function and skills, mainly by treating the motor and sensory impairments using simple *nonactuated devices*[1] (see first three rows of Table I).

In recent years, robotic devices and game-like virtual reality exercises have been increasingly used across industrialized countries, and may redefine rehabilitation by motivating people to train more, without clinical supervision. Because these devices can accurately measure variables such as position and force, they can be used for treatment as well as to diagnose and assess motor impairments such as spasticity, muscle tone, and strength with great accuracy.

These devices can automate repetitive tasks and provide *passive movements*, i.e., without voluntary muscular contraction by the individual, or *active movements*, i.e., with voluntary movement of a joint. In addition, they can provide assistance adapted to each subject and degree of recovery. Several studies suggest that robot-assisted therapy has positive effects on the rehabilitation progress of stroke patients [6]–[11]. However, interfaces to train the distal components of the upper limbs, e.g., wrist and hand, have received little attention so far.

Different robots have been developed to provide *continuous passive motion (CPM) of the hand* (see fourth and fifth rows of Table I) helping subjects reduce joint stiffness of the fingers together or individually.[2 3] This type of device offers a versatile, comfortable and portable therapy, but lacks the possibility of performing active finger movements.

Several *active robotic devices*, i.e., with active participation of the patient, have been recently developed to train hand function. They can be divided into four groups (see lower rows of Table I). The first type of device consists of a hand module added to robotic structures used for rehabilitation of the arm. Masia *et al.* have developed a Hand Robot Alpha-Prototype II, which is mounted at the output of their MIT-MANUS system. This device interacts with the palm and can provide high force to train grasp and release, but it may be limited by a small range of motion [12]. Riener *et al.* have also extended their ARMin device to provide exercises for forearm and hand. The distal module is characterised by a semi-exoskeleton structure, with the arm placed inside an orthotic shell [13]. The Gentle/G system, developed by Loureiro *et al.* [14], involves a 6 degree-of-freedom (DOF) hand module with one motor for the thumb metacar-

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[1]http://www.rehabmart.com

[2]http://www.vqorthocare.com/Products/CPM/index.php

[3]http://www.tyromotion.com/index.php?id=8

TABLE I
REVIEW OF DEVICES FOR HAND AND FINGER REHABILITATION

		Interface	Movement	DOF	Output Force, Torque	Workspace
Non actuated		Thera-Band Hand Exerciser [6]	hand opening/closing (fingers move together)	1○	45N	∞
		Digi-Flex [6]	extension/flexion of the five fingers (fingers move individually)	5○	40N / finger	30–60°
		Power-Web [6]	any movements of the five fingers (fingers move individually)	∞○	*	∞
CPM		Hand 8091 by VQ OrthoCare [13]	extension/flexion of the five fingers (fingers move together)	1⊕	*	0–90°
		Amadeo System by Tyromotion [14]	extension/flexion of the five fingers (fingers move individually)	5⊕	*	0–70°
Active robotic devices	Additional hand module	Hand Robot Alpha-Prototype II [15]	extension/flexion of the five fingers (fingers move together)	1●	120N	-45–90°
		ARMin forearm extension [16]	forearm pronation/supination wrist extension/flexion	2●	4Nm 3Nm	±70° -30–75°
		Gentle/G Grasp Robot [17]	extension/flexion of four fingers (2 DOF) (fingers move together) thumb extension/flexion (1 DOF)	3● 3○	18N 12N	0–70° -10–60°
	Robot dedicated to hand function	Rehabilitation Haptic Knob [18]–[20]	forearm pronation/supination extension/flexion of five fingers (fingers move together)	2●	1.5Nm 50N	±180° 0–60°
		HWARD [21]	extension/flexion of four fingers (fingers move together) thumb extension/flexion wrist extension/flexion	3●	15N	25–90° 0–60° 0–20°
	Glove and exoskeleton for individual finger	Rutgers Master II [22], [23]	extension/flexion of four fingers, without little finger (fingers move individually)	4●	16.4N / finger	0–40°
		CyberGrasp, Immersion [24]	extension/flexion of five fingers (fingers move individually)	5●	12N / finger	-45–85°
		Gifu Haptic Interface [25], [26]	extension/flexion of each finger (2 DOF) abduction/adduction of each finger (1 DOF) thumb extension/flexion (3 DOF) thumb abduction/adduction (1 DOF) forearm pronation/supination wrist extension/flexion	18●	5N 5N 5N 5N 3Nm 1.3Nm	0–90° 0–45° 0–80° 0–60° ±90° ±80°
	Robot for 1-2 fingers	SPIDAR [27], [28]	extension/flexion of two fingers	6●	4N / finger	-45–90°
		HIFE [29]	extension/flexion of one finger	2●	10N / finger	-45–90°

○ non actuated DOF ⊕ actuated and passive DOF ● actuated and active DOF
$*$ not available

pophalangeal (MCP) joint and two actuated phalanges for the opposing fingers, and free orientation in roll, pitch, yaw. This module is connected to a HapticMaster robot providing three DOF movement, such that subjects can train to grasp and move objects in space.

The second group of active robotic devices has been developed to train specific hand functions. We have recently created a two DOF Haptic Knob to train opening and closing movements of the hand, as well as pronation and supination of the forearm, so as to simulate interaction with objects [15]–[17]. The HWARD (Hand-Wrist Assisting Robotic Device) [18] is a three DOF pneumatically-actuated robotic device that assists the impaired hand in grasping and releasing movements. Advantages of these two robots include their large ranges of motion and force, although it is not possible to train each finger independently.

The third group consists of gloves and exoskeletons focusing on finger function. Burdea *et al.* introduced the Rutgers Master II, a haptic glove which serves as an instrumented interface to sample hand positions and provide suitable resistive forces [19], [20]. However, the limited workspace and the difficulty that patients with spasticity may have in slipping on this type of glove may limit its therapeutic use to laboratory and clinical settings. Exoskeletons are also being used for hand and finger rehabilitation. The CyberGrasp Exoskeleton developed by Immersion[4] allows full range of motion of the hand without obstructing movements. A novel exoskeleton, the Gifu Haptic Interface, has been developed by Kawasaki *et al.* to provide a self-training rehabilitation system, allowing patients to perform rehabilitation exercises by themselves [21], [22]. However, the limited range of force (5 N), that may not be appropriate for patients with severe spasticity, and considerable friction interfering with smooth movement are significant drawbacks of this type of system.

Different types of robotic devices have been developed for dedicated finger exercises. The SPIDAR (SPace Interface Device for Artificial Reality) system uses a different approach consisting of a rigid cubic frame and several motors with pulleys attached to each corner of the frame. Strings span from each motor-pulley unit to the thumb and index finger of the subject to allow different finger movements and grips [23], [24]. The low range of force that can be applied at the output may limit the

[4] http://www.immersion.com/ 3d/products/cyber_grasp.php

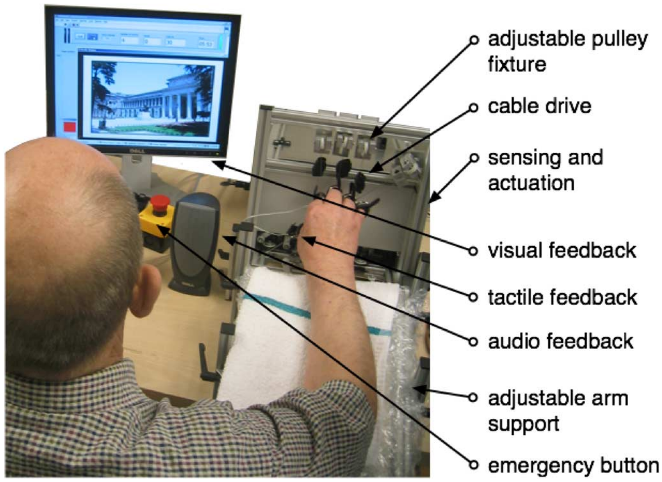

Fig. 1. View of the *HandCARE* system used by a stroke patient. The main part of the interface is the clutch and control box, which includes the motor, the control card, the clutch systems as well as the sensing systems. Five adjustable pulley fixtures allow the direction of the movements to be modified. Visual, tactile, and audio feedback are implemented to keep the subject informed during the training. The dimensions of the interface are $60 \times 30 \times 30\ \mathrm{cm}^3$ (arm support included).

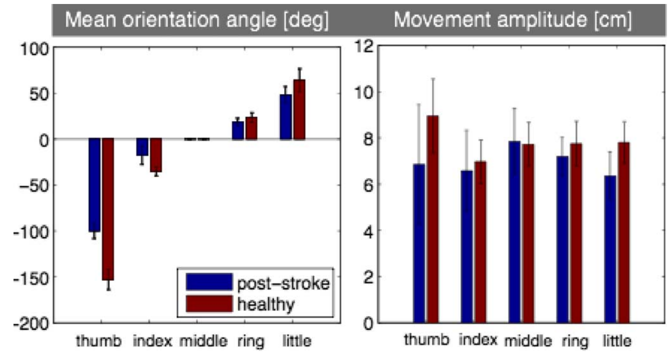

Fig. 2. Mean orientation μ_n with middle finger as reference (A) and amplitude (B) of the five fingertip trajectories for eight healthy and three poststroke subjects. The error-bars indicate the standard deviations.

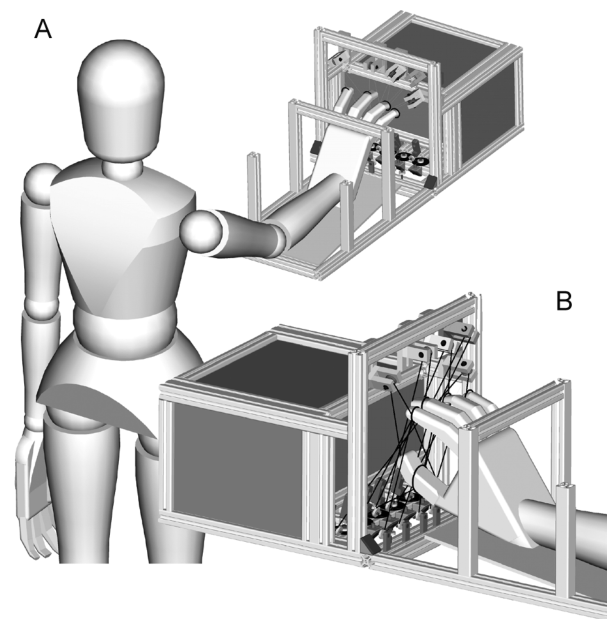

Fig. 3. Drawing of the *HandCARE* (A) with expanded view of the finger–robot interaction (B).

use of this system for hand rehabilitation. The HIFE-Haptic Interface, a device based on a tendon-driven transmission system, has been developed to train extension/flexion movements of one finger. The low friction and large range of motion and force make this system well adapted for finger exercises. However, the use of this system is limited to one finger only [25].

Table I summarizes the main characteristics of the robotic interfaces described above, namely the specific *movement* that can be trained with the robot, the number of *DOF*, the maximal *force/torque* that can be applied at the output of the system and the functional *workspace* within which the hand or the finger can move (finger extension/flexion angle is evaluated at the MCP joint, where the origin is defined for a proximal phalange aligned with its corresponding metacarpal, negative values represent extension and positive values are for flexion).

This table suggests that the challenges of recovering fine motor control of the fingers have yet to be addressed by suitable interfaces. Current *active robotic devices* for hand rehabilitation are often too large to be used at home, have too limited range of force or do not offer the possibility for training each finger individually. On the other hand, *nonactuated devices* cannot control the force, while *CPM interfaces* do not train active movements.

Our objective is to develop an interface to train distal segments of the upper limbs with the advantages of both nonactuated interfaces and active robotic devices. This interface should be safe and compact while producing adequate forces within a large workspace. It should enable poststroke patients to train at home or in decentralized rehabilitation centers by performing motivating virtual reality (VR) game-like exercises. Furthermore, the interface should be flexible such that it can be adapted to patients' biomechanics, provide comfortable interaction, and be cost effective [26].

This paper presents a Cable-Actuated REhabilitation System to train Hand functions (*HandCARE*) (Fig. 1) addressing these requirements. We first examined the different tasks for which this interface is to be used, and measured corresponding biomechanical parameters (Section II). The design, determined by these specifications, is described in Section III. Material choice, the actuation system, and the implemented control schemes are described in Section IV. Experiments were conducted with the interface to evaluate its performance (Section V).

II. Hand Biomechanics Requirements

Different dysfunctions such as muscle weakness, spasticity, and compulsory co-activation of antagonistic muscles at multiple joints, contribute to impairment of finger and hand function after stroke. Due to extensor muscle weakness, the fingers are often locked in a flexed position and stroke patients are not able to control finger motion. Thus, the first function the robotic interface should train is finger extension. Then, finger flexion should be trained to strengthen weak muscles and reduce the effects of synergies. Sufficient versatility of the robot is required to allow individual movements for each finger, grasping with all

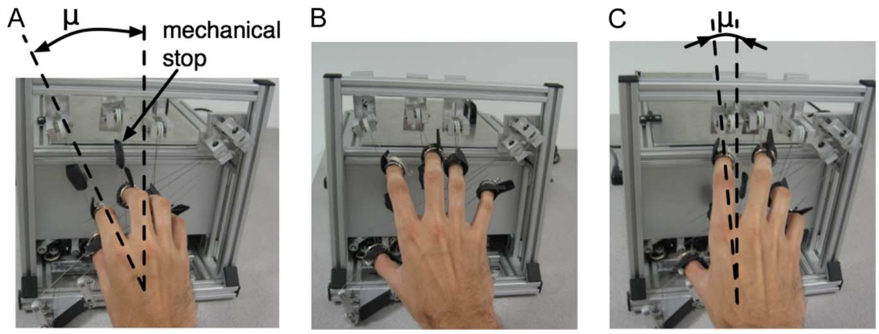

Fig. 4. Hand in closed (A) and open position (B). The fixtures can be adjusted to change the orientation μ of finger trajectories (C). Different movements can be trained, e.g., opening/closing with five fingers (A), (B) or with tripod thumb-index-middle fingers (C).

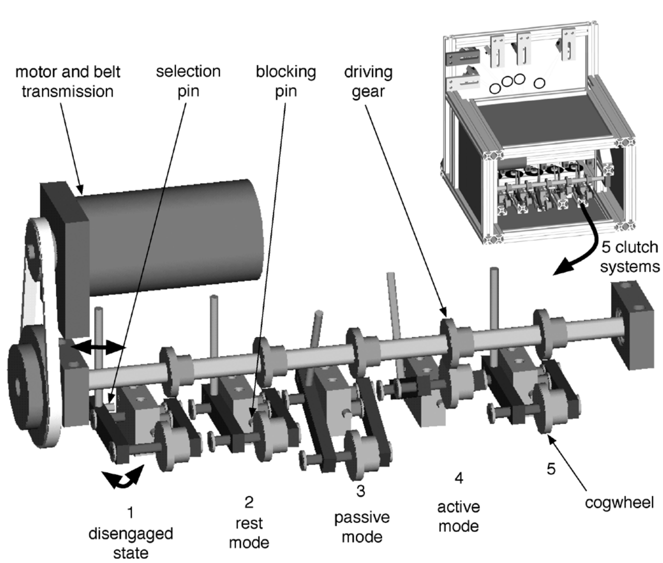

Fig. 5. A 5-clutch system is used to train the five fingers independently. The clutch mechanism allows three operation modes for each individual finger: i) rest mode—the cogwheel and the cable are blocked by a pin and the finger cannot move (clutches 2 and 5), ii) passive mode—the cogwheel is free to rotate so the finger can move freely (clutch 3), and iii) active mode—the cogwheel is driven by the motor, which moves the finger (clutch 4). In order to select the mode, a pin is engaged in one of three positions corresponding to the described modes.

five fingers, or more precise functions such as pinching between two fingertips or tripod pinch.

A simple experiment was performed to identify the biomechanical constraints of the human hand. Eight healthy subjects between 21 and 32 years of age, all right-handed, as well as five chronic stroke patients participated in this experiment [27]. These patients, two females and three males, were between 54 and 91 years of age, all right-handed with right hemiplegia. Subjects were first asked to open the hand until the fingers were maximally extended at the MCP joint and then to close the hand until the fingertip of the thumb touched the fingertips of the four opposing fingers. The movement of the fingertips was constrained to a plane.

To determine the natural orientation μ_n and amplitude of finger movements, measurements were made when the five fingers were at the extreme open and closed positions. Fig. 2 presents the orientations as well as the amplitudes of the finger trajectories. The orientation of the fingers during movement is different for healthy and poststroke subjects because of limited finger abduction of the latter. In particular, the orientation angle of the thumb is significantly smaller. Due to joint stiffness, muscle contracture, flexor synergy, or spasticity, the stroke patients were all unable to place the thumb in opposition to the other four fingers.

The five patients had difficulty in opening the hand, but in terms of passive range of motion, there was no notable difference with the healthy subjects. The averaged maximal grasping force for poststroke subjects was 240 N for male and 120 N for female, and, respectively, 450 N and 300 N for healthy subjects.

III. Design of the Rehabilitation System

Cable interface designs such as the SPIDAR (Table I) or the Mantis Workstation developed by Mimic[5], which have shown the high potential of cable-based haptic interfaces, attracted our interest and served as a starting point for our design. The *HandCARE* is also a cable-driven robotic tool, where each finger is attached to a cable loop allowing predominantly linear displacement approximately equal to the measurements presented in Section II (Fig. 2). The interface can assist or resist the subject in opening and closing movements. The device consists of four main parts.

- A cable-driven system with a frame and pulleys that convey the cable (Figs. 3 and 4).
- A clutch actuation system providing assistive or resistive forces to the fingers (Fig. 5).
- A sensing system to measure interaction between the subject and the interface (Fig. 6).
- An arm support, i.e., a versatile system to support the forearm of the subject (Fig. 1).

To actuate all five fingers, the system would in principle require five motors. However, a clutch system was developed to allow training of different movements with only one single actuator. With the five clutches, it is possible to train grasping and pinching as well as independent movements of each finger using the single motor.

A. Cable Driven System

Five adjustable pulley support fixtures guide the cables which move the fingers. Fig. 4 illustrates how the pulley fixtures can

[5]http://www.mimic.ws

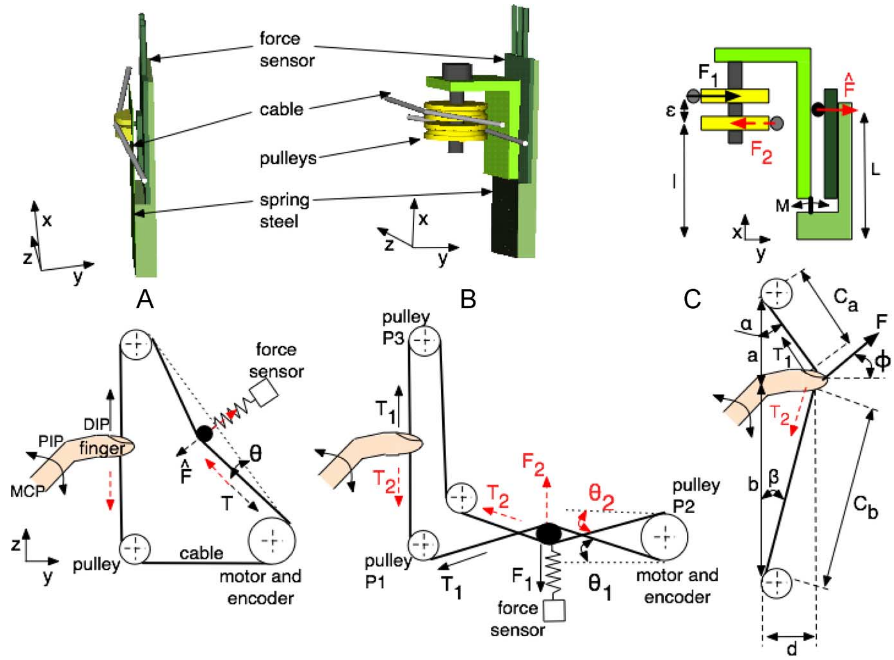

Fig. 6. Methods to measure cable tension. (A) Implementation and diagram of the three-roller principle. This method allows tension in the cable to be measured with a static force sensor while the cable is moving. (B) Implementation and diagram of a differential method to compensate the cable deflection when the force applied is not colinear with the direction of the cable (C).

slide along the frame for adjusting the orientation of the cable μ to fit the natural orientation of finger trajectories μ_n, defined in Section II. The cables cross within the workspace and so precautions must be taken to avoid interference.

B. Clutch System

One clutch is used for each finger, and can be manually switched between three different modes (Fig. 5).

- *Fixed mode*: The driving cogwheel is fixed and the cable blocked. The finger is thus maintained at a fixed position to allow for training of isometric force tasks.
- *Free mode*: The driving cogwheel is free to move. The finger can move without restriction along the path defined by the cable.
- *Active mode*: The driving cogwheel is engaged with the motor shaft and the torque generated by the motor is applied to the finger.

The clutch system allows the subject to train a variety of combinations of finger movements, e.g., with five fingers [Fig. 4(A) and (B)] or with the tripod thumb-index-middle [Fig. 4(C)]. While switching between the various clutch modes is performed manually in this interface, it can easily be automated using simple and cheap servomotors, as has been implemented in the second generation of *HandCARE*.

C. Differential Force Sensing

One limitation of this cable driven system is that any noncolinear force will perturb the tension in the cable. A conventional implementation of force measurement, for instance, the use of a force sensor at the output, has the disadvantage of being sensitive to these noncolinear forces, thereby causing the measurement of finger force to be biased. Therefore, a differential sensing system has been developed, which is based on the three-roller principle, consisting of an external elastic component that measures cable tension [Fig. 6(A)]. Fig. 6(B) illustrates how the differential method compensates for the effect of noncolinear force by mechanically compensating for the tension in the two cable strands attached to the finger.

The key elements of the differential systems are the five "MilliNewton" 2 N force sensors[6] used to determine the force applied by each finger. These sensors use the piezoresistive properties of thick films. The sensing element is an alumina cantilever with a thick-film piezoresistive Wheatstone bridge and is soldered onto a thick alumina base, which contains the (thick-film) conditioning circuit [28], [29]. This construction allows batch fabrication of simple yet fully amplified and calibrated sensors. The practical measuring range of the cantilever geometry, up to ca. 2 N, limited by the strength of the cantilever and the solder joint, is sufficient to measure the cable tension, provided that its deflection angle is relatively small.

Fig. 6(B) shows how the pulleys $P1$ and $P2$ must be placed to compensate for noncolinear forces. The force $\widehat{F}$ measured by the sensor is

$$\widehat{F} = \frac{M}{L} \tag{1}$$

where the moment M depends on the forces F_1 and F_2 resulting from the tensions T_1 and T_2 in the two cable strands

$$M = F_1(l+\epsilon) - F_2 l \tag{2}$$

$$F_i = T_i \sin\theta_i, \quad i = 1,2. \tag{3}$$

The distances ϵ and l are defined in Fig. 6(C). The objective is that the measured force $\widehat{F}$ equals zero when a force normal to the cable is applied by the subject: $F = 0$ when $T_1 = T_2 >$

[6]http://lpm.epfl.ch

0. Equations (1)–(3) leads to the following condition for the placement of the pulleys $P1$ and $P2$:

$$\sin\theta_2 = \sin\theta_1 \frac{l+\epsilon}{l}. \tag{4}$$

The force $\widehat{F}$ measured by the sensor can vary with two factors.

- The direction ϕ of the applied force F [Fig. 6(C)], assuming there is negligible deflection $(d \cong 0)$

$$\widehat{F} = F\sin\phi. \tag{5}$$

- The position a of the finger along the cable, due to the cable deflection d [Fig. 6(C)]. Assuming that the force F is normal to the cable $(\phi = 0°)$ and the distance ϵ is small $(\theta = \theta_1 = \theta_2)$, the contributions of the forces F_1 and F_2 to the force $\widehat{F}$ are thus identical. From relations (1)–(3)

$$\widehat{F} = (T_1 - T_2)\sin\theta \frac{l}{L} \tag{6}$$

where

$$T_1\cos\alpha = T_2\cos\beta \tag{7}$$

and

$$\cos\alpha = \frac{a}{C_a} \quad \text{and} \quad \cos\beta = \frac{b}{C_b}. \tag{8}$$

The distances C_a and C_b are

$$C_a = \sqrt{a^2 + d^2} \tag{9}$$
$$C_b = \sqrt{b^2 + d^2} \tag{10}$$

and Hooke's law

$$F = k\left[(C_a + C_b) - (a+b)\right] \tag{11}$$

completes the set of equations. The coefficient k of 40 N/mm is the compliance of the system. The force $\widehat{F}$ is significantly biased when the force F is applied near one of the pulleys $P1$ and $P3$. Therefore, the workspace is constrained to a central interval, i.e., $3.5 < a < 21.5$ cm, where the variation of the force represents less than 3% of the nominal force. The variation is linear within this interval, i.e., the correlation coefficient with the linear fit is 0.96.

This sensing system is suitable for our purpose as only forces parallel to the cable are measured.

D. Arm and Finger Fixation

The support for the forearm and elbow was designed to provide comfort, while mechanically isolating the hand from other body movements. The support can be adjusted to change the position (±10 cm) and the orientation (±30°) of the forearm (Fig. 1).

Different techniques for finger attachment are possible [30]. The use of gloves is precluded in our application as stroke patients with spasticity may have difficulty in donning them. A first method was tested where the subjects inserted their fingers into sewing thimbles. The size of the thimbles was not adjustable and subjects reported discomfort due to perspiration after using them for more than 20 min. A better finger fixture consisted of a Velcro loop within a metal ring to which the two ends of the cable were attached (Fig. 4). The finger can be inserted partially or completely into the Velcro loop, i.e., just the fingertip (distal phalange), or as far as the intermediate or the proximal phalange. Insertion of just the fingertip was mostly used for the feasibility study with stroke patients.

TABLE II
FEATURES OF THE *HandCARE*

linear motion of each finger	8.5 *cm*
flexion of each finger	0 - 70 *deg*
maximal opening (between thumb and opposing fingers)	18.5 *cm*
minimal opening (between thumb and opposing fingers)	1.5 *cm*
maximal/minimal force generated at the ouput	± 75 *N* ± 15 *N* for each finger
force measuring range	± 15 *N* for each finger
force sensitivity	0.2 *N*
control frequency	100 *Hz*
sensor sampling frequency	1000 *Hz*
weight (with motor and control system)	5 *kg*
external dimensions	60 x 30 x 30 cm^3

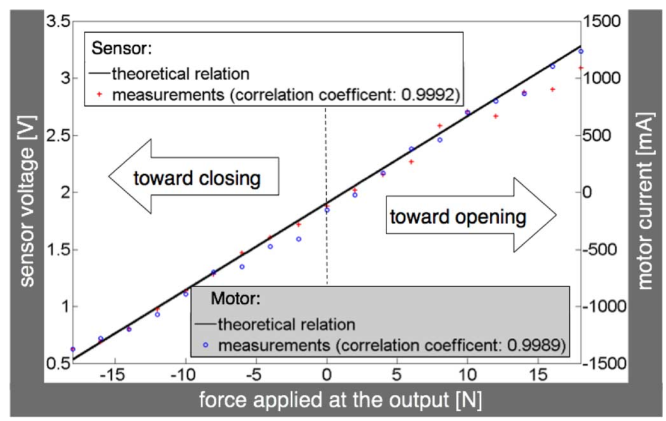

Fig. 7. Measurement of one of the five force sensors (+) and motor current (∘) required to maintain a constant position of one finger (middle finger) when different forces are applied at the output. The theoretical relations are superimposed.

TABLE III
FRICTION COEFFICIENTS

	Coefficient		
	Coulomb [N]	viscous [N/rpm]	correlation of linear fit with data
0 clutch	0.07	0.008	0.77
1 clutch	0.11	0.010	0.71
2 clutches	0.16	0.011	0.87
3 clutches	0.23	0.010	0.79
4 clutches	0.30	0.013	0.85
5 clutches	0.41	0.013	0.87

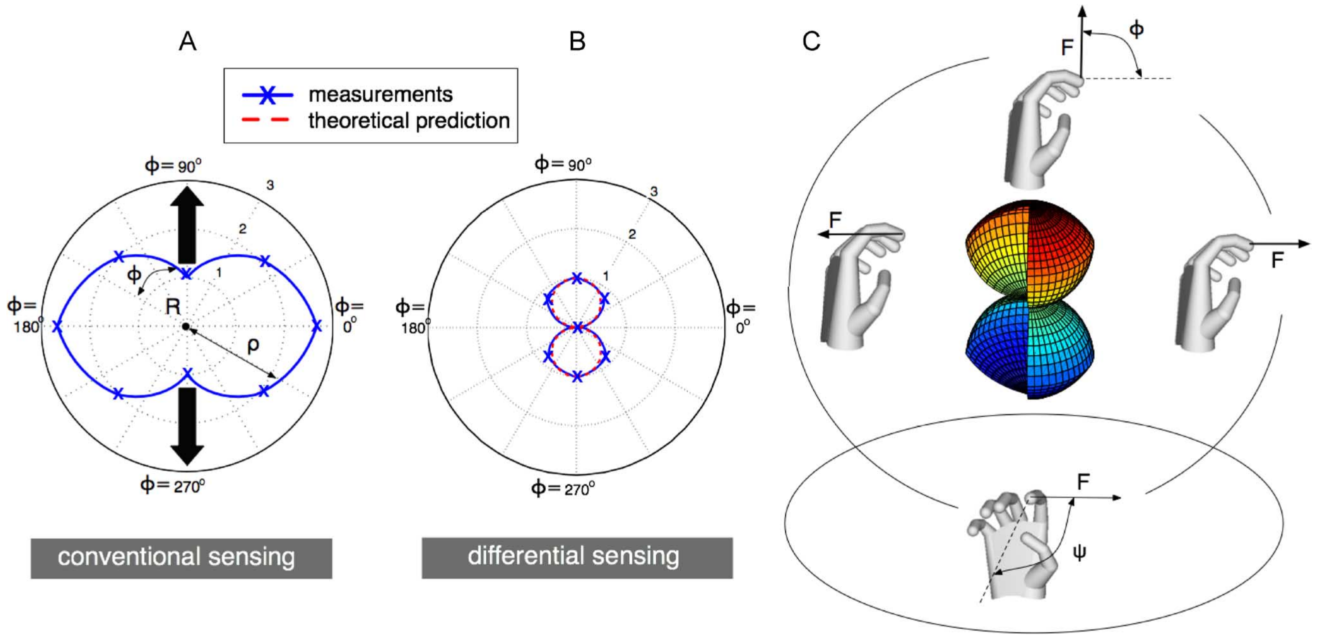

Fig. 8. Measurement of the sensitivity to cable deflection for a conventional system (A) and the compensated system (B), (C). The direction of forces applied at the output can be defined by two angles, ϕ (corresponds to finger extension/flexion) and ψ. The force is applied by the finger at reference point R and the direction of the cable is represented by the black arrows. The crosses represent the measurements. The coefficient ρ is the ratio between the force $\widehat{F}$ measured by the sensor and the force F applied at the output.

IV. System Implementation

A. Materials and Components

Materials for the cable were compared according to their compliance, friction around a pulley, breaking strength and creep. Polyester, polyester reinforced with carbon fibers, polyethylene fiber, and steel wire were tested and it was found that steel cable (diameter of 0.5 mm) had the most suitable combination of these factors. The 30 pulleys used to guide the cables are made of POM (polyoxymethylene, or Delrin) and are mounted on standard ball bearings.

Cogwheels, which are used for the clutch system, made from POM or steel were tested for durability and transmission smoothness. Plastic cogwheels wore out quickly, therefore, steel cogwheels were more suitable for our purpose.

B. Actuation and Control

The interface is actuated by one brushed dc motor (Maxon motor, Switzerland; RE40, 150 W; encoder 500 counts/rev; control card EPOS 24/5). The gear ratio between the motor shaft and the clutch is 2. The interface is controlled by a program written in Labview 8.2 (National Instruments) that runs on a PC (Pentium 4,4 GB RAM, 233 MHz). The main program is divided into subtasks to separate control, display, and data acquisition, and thus distributes the tasks and allows faster control. Data from the EPOS controllers of the motor (positions, velocities and currents) are read at a frequency of 100 Hz and transferred to the main program using an RS232 protocol. Data from force sensors are sampled at 1000 Hz by a data acquisition card (USB-6211, National Instruments). The main control program analyzes position and force inputs and calculates commands to send to the motor at 100 Hz. This frequency is sufficient for control because friction in the system provides stability and because human motion is characterized by a low bandwidth [31]. Indeed, the mechanical bandwidth of human movement is around 7 Hz (2 Hz for normal speed movements) [32]. A display loop has been implemented to provide visual feedback at a refresh rate of 20 Hz.

C. Safety and Psychological Factors

Safety is the first requirement for an interface that physically interacts with humans. To prevent any harm or damage, both software and hardware emergency systems are implemented as described in [33]. Five mechanical stops have thus been installed (Fig. 4) and an emergency switch actuated by a technician or a physiotherapist can stop the motor anytime. Moreover, a safety pneumatic switch is held by the subject during the experiment and stops the system if squeezed.

Psychological factors related to the design are important for the comfort and security of the patient and the therapist. To this end, all electronics and the drivetrain have been enclosed in a box and external parts have been rounded for safety.

V. System Performance

A. General Features

Table II summarizes the characteristics of the completed prototype. The workspace consists of five linear paths of 8 cm length corresponding to a finger extension/flexion angle range of 0°–70° at the MCP joint (for a finger length of 9 cm). The maximal opening is 19 cm and the minimal closing is 1.5 cm between thumb and the opposing fingers. The maximal continuous force that can be generated is ± 15 N per finger, while inherent friction is less than 0.8 N in any position of the workspace.

The system is backdrivable and Fig. 7 shows the current generated by the motor to maintain a position and the force $\widehat{F}$

measured by the sensor when a force F, parallel to the cable ($\phi = 90^\circ$ toward opening and $\phi = 270^\circ$ toward closing movement), is applied at the output. The theoretical sensor voltage V is deduced from relations (1)–(3)

$$V = c\widehat{F} + V_o = cF \sin\theta \frac{l}{L} + V_o \tag{12}$$

where $c = 0.2$ V/N is the calibration coefficient and $V_o = 1.9$ V is the offset of the differential sensing system. The theoretical current of the dc motor is proportional to the sensor voltage. Sensor and motor measurements are highly linear and are close to the theoretical prediction within the range of $[-15, 15]$ N.

B. Friction Identification

Friction was identified from the output force for different velocities and configurations, i.e., while varying the number of clutches engaged with the motor. To obtain consistent results, the measurement was performed three times and a linear fit was used to determine the viscous friction (slope of the line) and the Coulomb coefficients (output force for a velocity equal to zero). Table III summarizes these coefficients and shows that Coulomb friction increases with the number of fingers actuated by the motor, while the viscous friction is similar for any configuration (except a lower coefficient when no clutch is engaged).

C. Force Sensing

Fig. 8 compares the ratio

$$\rho(\phi, \psi) = \frac{|\widehat{F}|}{|F|} \tag{13}$$

of the two sensing principles described in Section III-C, when a force F is applied at reference point R, i.e., at the center of the cable-loop. In (13), ϕ and ψ define the orientation of the applied force F. This figure shows that the conventional method without compensation is highly sensitive to forces applied perpendicularly to the cable ($\phi = 0^\circ$ and $\phi = 180^\circ$), which creates a deflection increasing the overall cable tension. For instance, this experiment shows that forces measured by the force sensors are three times higher than the actual forces applied at the output when the direction of this force is perpendicular to the cable [Fig. 8(A)]. The differential system [Fig. 8(B) and (C)] markedly reduces the effect of noncolinear forces and is sensitive mainly to forces generated along the cable, thus offering a solution adapted to our purpose. The measurements are close to the theoretical curve based on (5).

VI. Conclusion

A new interface for hand and finger rehabilitation has been developed, based on patients' requirements in terms of biomechanics, comfort, and safety. By providing movements of the five fingers with large range of motion and force, the device can help patients train functions such as finger flexion and extension, coordination between the fingers, and independence of each finger, which are necessary for most activities performed with the hand. The *HandCARE* has been designed to be adaptable to most hands and was successfully tested with healthy and disabled subjects.

The design consists of a frame that guides five adjustable cables on which the fingers are positioned. The interface is compact and can be transported and placed on a table. Main features of the interface are the clutch system, which allows independent movement of the five fingers with only one actuator, and the differential force sensing system, that is used to provide feedback to the patient. The encoder and force sensors allow the patient's progress to be monitored during training sessions.

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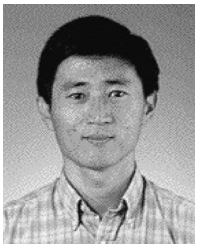

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