Natural grasping

Design and evaluation of a voluntary closing adaptive hand prosthesis

G. Smit

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Summary

Replacement of a missing hand by an artificial alternative remains one of the biggest challenges in rehabilitation. Although many different terminal devices are available, around 27% of the amputees does not actively use their device and 20% totally refrains from wearing it. There are various reasons for prosthesis abandonment, e.g. wearing discomfort (too heavy, too hot), too little added functionality, difficult or tiring to use, lack of sensory feedback. User studies identified multiple aspects of the prostheses that need improvement, in order to meet the user demands. Mass reduction was identified as the most important design priority. In general the user demands can be summarised by the three C's: Cosmesis, Comfort, and Control. The prosthesis should be beautiful to look at, comfortable to wear, and easy to operate.

The goal of this thesis was to design and test a new lightweight and efficient body-powered hand prosthesis with articulating fingers. A low mass will increase wearing comfort. Mechanical efficiency will decrease the required actuation force, which will lead to an increased control comfort. It will also enable the hand to produce a higher pinch force, which will increase the functionality of the hand. The articulating fingers of the hand will enable both power and pinch grip. This enables the grasping and holding of a broad range of different objects and enhances natural cosmesis.

The first step of the study was to determine the state-of-the-art in body-powered prostheses. Chapter 2 describes the testing of voluntary closing devices and Chapter 3 the testing of voluntary opening devices. The mechanical performance of the hooks was better than that of the hands. The hands required a high actuation force and energy (1058-2292 Nmm). They dissipated a large part of the actuation energy and produced only a low pinch force (~15 N). The mass of the hands was high (~423 gram). Comparison with data of a study from 1987 showed no improvement in the mechanical performance of the terminal devices over the last decades. In order to meet the user demands, a new hand design should have a lower mass, require less actuation energy, dissipate less energy and should be able to produce a higher pinch force.

Chapter 4 describes the design and testing of two underactuated finger prototypes. One finger had a pulley cable transmission, the other a hydraulic cylinder transmission. The fingers were optimized for application in a finger of a cosmetic glove of a prosthetic hand. The fingers had identical dimensions and they had a very low mass. Quantitative mechanical tests were performed to select the most efficient way of transmission. The pulley finger required 35-74% more energy for various tasks than the hydraulic finger. Based on the results the hydraulic finger was selected as the most suitable for application in a prosthetic hand, as it had a higher energy efficiency than the pulley finger. Furthermore the hydraulic transmission offers an additional improvement of efficiency of 10-40% of the entire system, when hydraulics is used to replace the Bowden-cable in the shoulder harness. Therefore the hydraulic transmission was chosen to be used in the new hand prototype.

Chapter 5 describes the mechanical comparison of silicone and PVC cosmetic gloves. Both types of gloves can be used for a prosthesis. The tests were performed to select the most energy efficient cosmetic glove. The tested silicone gloves had a 2.5-4.5 lower stiffness than the PVC glove, required 1.8 to 3.8 times less actuation energy and dissipated 1.7 to 3.4 times less energy. Therefore for the new hand prototype a silicone glove was used.

Chapter 6 describes the design and testing of a glove compensation mechanism. This mechanism, which fit inside a finger, had a negative stiffness which compensates the undesired positive stiffness of a cosmetic glove. The negative stiffness of the mechanism reduced the required input torque range by 58% for the PVC glove and by 52% for the silicone glove. A negative stiffness mechanism was applied to the new hand prototype, in order to reduce the actuation effort for the user.

The final step of the study was the design, and testing of a new hand prototype, described in Chapter 7. The new hand prototype, the Delft Cylinder Hand, has underactuated articulating fingers which adapt to the grasped object. It has voluntary closing body-powered control and it has a hydraulic cylinder transmission. The hand was subjected to various mechanical and functional tests. Chapter 8 describes the comparison of the performance of the hand to current available hands. Through the application of a hydraulic transmission, the hand requires 49-162% less energy from the user when compared to commercially available body-powered hands and it has a higher maximum pinch force (30-60 N). In functional tests the hand scored similar to current myoelectric hands. Yet its mass (152 gram without glove; 217 gram with glove) is 68% lower than the lightest available articulating myoelectric hand and 55% less than the lightest body-powered hand of similar size. Functional tests showed that The 'Delft Cylinder hand' provides the amputee with a level of function that is at least comparably to contemporary hands, at a cost (mass and actuation effort) which is much lower than that of all currently available hands.

The Delft Cylinder Hand has articulating fingers and is anthropomorphic, slender, fast, efficient and silent. The hand mass is much lower than the lightest commercially available hand. The hand therefore meets one of the most important user demands in upper limb prosthetics, which is a low hand mass. The hand can pinch harder (>30 N) at a lower user effort.

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Introduction

1.1 Upper limb deficiency

Upper limb deficiency is a condition in which a part of the upper limb is missing. Upper limb deficiency can have different causes. In the first place there is the congenital limb deficiency. In children with a congenital limb deficiency, the deficient limb has not fully developed during pregnancy. Although congenital defects can be caused by drug use during pregnancy [1, 2], or by syndromes and genetic defects [3], the cause of a congenital limb deficiency is in many cases unknown. The second cause of limb deficiency is the acquired limb deficiency as a result of an amputation. There can be various reasons why a limb needs to be amputated. Common causes can be health conditions e.g. dysvascular conditions, cancer [4], or traumatic causes e.g. physical and thermal injuries [5], infections after injury [6, 7], or war related injury [8].

The prevalence of upper limb deficiency is relatively low. It is estimated that there are about 3,750 persons with an upper limb deficiency in the Netherlands [9], of which 1350 have a congenital deficiency (prevalence of 0.8 per 10.000 inhabitants) and 2400 an acquired deficiency (prevalence of 1.5 per 10.000 inhabitants). For the entire US it is estimated that there are 41,000 persons with an acquired major upper limb deficiency [10], which gives a prevalence of 1.4 per 10.000 inhabitants. The prevalence of various congenital upper limb deficiencies in the US ranges from 2.8 to 5.0 per 10,000 births [11]. This number might however include deficiencies located distally from the wrist. Although the prevalence of upper limb deficiency is relatively low, the impact of missing an upper limb can have a significant impact to the individual amputee [12-15]. A prosthetic hand or arm can restore some of the functions of a missing limb and help the user in performing activities of daily living.

1.2 Upper limb prostheses

Throughout the ages many different prosthetic hands and functional replacements have been developed to restore some of the function of the missing limb. Overviews of the range of current available upper limb prosthetic devices can be found in [16-19]. The main components of a prosthetic arm are: the terminal device, the arm socket and the shaft. The terminal device is the part which replaces the function of the hand. It can be a prosthetic hand or hook, or other device. Depending on its functionality, a terminal device can be used to grasp, pinch, fixate or support objects. The socket is the parts that is fitted around the residual arm. The socket forms the interface between prosthesis and residual limb. The prosthetist fits the socket to the individual patient, as the shape of the residual limb is different for each individual amputee. The shaft, which replaces the arm, connects the socket to the terminal device. Beside these main components there can be other parts, depending on the way of control and the level of amputation. Upper limb prosthetic devices can globally be divided into four categories, based on the type of terminal device and the way the device is controlled:

- *Tools and aids.* These devices are developed to assist the amputee in specific tasks, like self-care, recreational or occupational activities. Examples include tools like: a hammer, pliers or cutlery, or recreational terminal devices like: a swimming fin, a bicycle handlebar adapter or an adapter for playing a musical instrument [20-23].
- *Passive and cosmetic devices.* The main goal of these devices is to replace the appearance of the missing hand. Furthermore a cosmetic hand can be used to clamp, push, fixate and support objects [24, 25]. Also passive controlled devices are available. In these devices the sound hand controls the prosthetic hand and provides the energy to open and close the device. The passive controlled device can be used to hold and carry objects. Cosmetic hands are usually covered by a cosmetic glove, made of PVC or silicone [26, 27].

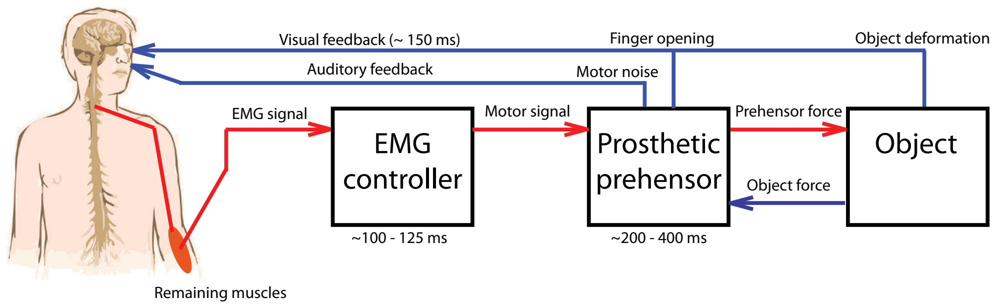

Figure 1.1 The principle of myoelectric control. A skin electrode picks up the EMG signal form the muscle. The controller uses the signal to control the motor. The amputee receives visual and auditory feedback.

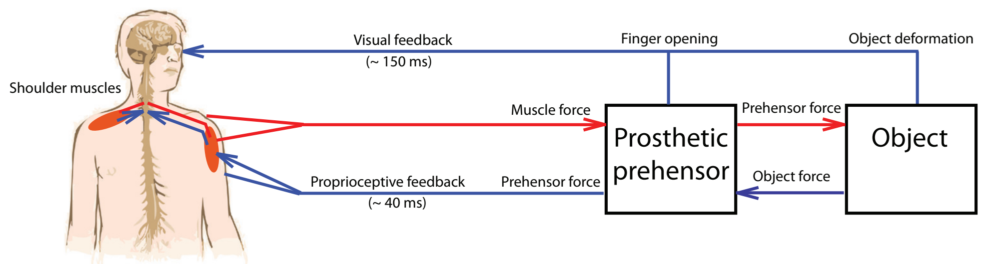

Figure 1.2 The principle of body powered control. The amputee exerts forces to the prehensors by means of the shoulder harness. The amputee receives proprioceptive force and position feedback, as well as visual feedback.

- *Externally powered devices* are active devices that are opened and closed by an actuator that is powered by a portable power source, usually an electric battery [28]. Another power source, although currently not used anymore, is compressed carbon dioxide in gas powered devices [29]. The portable power source supplies the energy to actuate the device, the user provides the signal to control the device. Common used control signals for externally powered devices are: myoelectric signals, control switches and transducers [30]. Myoelectric signals are detected by skin-electrodes placed on the skin of the residual arm. The electrodes detect small potential differences, due to voluntary contraction of the remaining muscles in the residual arm. In this way the amputee can control the opening and closing of the terminal device (Figure 1.1). The amputee receives mainly visual feedback, and some auditory feedback, to determine the opening of the prehensors and to estimate the exerted amount of force.

- *Body powered devices.* These active devices are driven by the body movements of the amputee. In body powered (BP) control the user provides both the control signal as well as the energy required to actuate the device, usually by pulling a control cable. Body powered devices can be subdivided in voluntary opening (VO) and voluntary closing (VC) devices. A VO device opens when the cable is pulled. When the cable is released a spring closes the device and provides the grip force. The working principle of the VC device is opposite to that of the VO device. In the VC device the user closes the prehensors and provides the pinch force by pulling the control cable. An opening spring opens the device, when the control cable is released. The user receives proprioceptive force and position feedback trough the control cable, as well as visual feedback from the prehensor (Figure 1.2). The end of the control cable is most commonly attached to a shoulder strap or harness (Figure 1.3). The control cable can be pulled by upper arm flexion, upper arm extension and by scapular abduction [31]. Another way of body powered control is elbow control [32, 33]. In this type of control the cable is attached to an elbow lever. This lever enables pulling of the cable by elbow extension. Less commonly used principles of body powered control are forearm pronation/supination [34] and wrist flexion/extension [35].

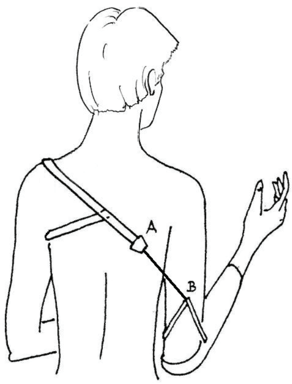

Figure 1.3 A shoulder controlled prosthetic hand is controlled by a strap around the contralateral shoulder. By increasing the distance between A and B, the control cable is pulled and the hand is actuated (adapted from [18]).

Current active prostheses

Since the introduction of the electric prosthesis in the 70's [28], there have been minor changes in the range of commercially available upper limb devices. The majority of the devices which are currently available to the user, have already been available for the past decades. A recent development was the introduction of the i-limb by Touch Bionics [36] and the bebionic by RSL Steeper [37]. These myoelectric hands have articulating fingers and also provide the user various grasp modes, e.g. pinch grip, tripod grip, power grip or typing finger configuration. Another recently introduced hand is the Michelangelo hand by Otto Bock [38]. Although this hand does not have articulating fingers, it has more functions than the standard myoelectric hand, e.g. lateral pinch grip and finger ab- and adduction. Although many research projects have been performed on the development of all kind of robotic and artificial hands [39, 40], these projects did not result in new available hands for the amputee. Research projects that did result in new available terminal devices, specifically aimed for the development of prosthetic devices from the beginning [32, 41-43]. In the past decades nearly all research in upper limb prosthetics has been focused on myoelectric hands [44]. For the BP devices there has been very little development in the past decades. Referring to the state of the art in body powered prosthetics, M.A. LeBlanc stated in an article [45] in Clinical Prosthetics and Orthotics in 1985: *"If one looks at the Manual of Upper Extremity Prosthetics first edition (1952) [46] and the Orthopaedic Appliance Atlas—Artificial Limbs first edition (1960) [47] compared with 1985 state of the art, one will not find a great deal of change."* Although he made this statement in 1985, little has changed since. Whereas in electric hands with articulating fingers have become available, BP hands still have stiff fingers. The only exception on this is the VO Becker hand [48], which has joint articulation. This hand was developed in the nineteen thirties [49-51] and has been on the market ever since.

Articulating fingers and underactuation

Every joint in the human hand is controlled by agonist and antagonist muscles. Except from the distal interphalangeal (DIP) joints, the finger joints can be controlled independently. After an amputation the amount of control signals is usually very limited. Typically one to three signals are available. Most current hands have one degree of freedom (DoF), which is controlled by one control signals. The addition of more joints to the hand, will increase the number of DoF's. As the number of available control signals does not increase, one control signal has to control multiple DoF's. This can be achieved by using the principle of 'underactuation'. A mechanism is by definition underactuated, when it has more DoF's than actuators [52, 53]. Grasp configuration of the fingers of an underactuated mechanism is dependent of the actuator force, the mechanism design and the external forces acting on the fingers, and the shape of the grasped objects. The fingers of the hand adapt to the shape of the object. The new prosthetic hand has to be designed in such a way that the different basic grasps, precision grip and power grip [54] can be performed, without ejecting the grasped object out of the hand [55].

Use of prostheses

Despite the developments made in upper limb prosthetics, user studies show high rejection rates among users of upper limb prosthesis [56]. Around 27% of the users does not actively use its active device [56] and around 20% stops wearing it all [57]. When looking specifically to body powered hands, studies show rejection rates of

80% and higher [58, 59]. Hooks are in general better accepted. This can possibly be declared by their lower mass, and their lower required user effort [59]. In general hooks have a much higher mechanical efficiency than hands [60]. To help increase user acceptance of body powered hands, the body powered hand should become more efficient and much lighter [61].

1.3 Problem statement

Current body powered devices are heavy, which causes a reduced wearing comfort. Controlling the BP prostheses requires a high actuation force from the amputee, which causes a reduced control comfort. They can only produce limited pinch forces and they do not have articulating fingers, so the fingers do not adapt to the grasped object. The low pinch force and the stiff fingers both limit the grasping ability of the hand. New myoelectric hands do have articulating fingers, which adapt to the grasped object. However, myoelectric hands are even heavier than BP hands, due to the use of motors and batteries. Their high mass causes a reduced wearing comfort. Furthermore myoelectric hands do not provide proprioceptive feedback, and they do have controller delays. This reduces the control speed, and the accuracy of the force and position control. Therefore myoelectric prostheses do not solve the problems of BP hands.

1.4 Goal

The goal of this thesis is to design and test a new lightweight and efficient body powered hand prosthesis. A low mass will increase wearing comfort. Mechanical efficiency will decrease the required actuation force, which will lead to an increased control comfort. It will also enable the hand to produce a higher pinch force, which will increase the functionality of the hand. The new developed hand will have articulating fingers, which enable both power and pinch grip. This enables the grasping and holding of a broad range of different objects.

1.5 Research approach

The hand should have a low mass, as this is indicated to be currently one of the most important design priorities in upper limb prosthetics [61]. A low hand mass will make wearing the hand more comfortable to the amputee. A light hand can potentially be used by more people than a heavier hand, as amputees with short stumps or high level amputations generally have problems with the mass of current prosthesis. A low hand mass can be realized by designing thin constructions and by using materials with a high specific stiffness.

The new hand prostheses will be body powered. Body powered control offers many potential advantages. Because of the availability of direct proprioceptive force and position feedback, body powered systems can be intuitively controlled without the need of constant visual attention [62, 63]. A BP system can have a low mass, as no motors or power sources are needed. BP control can be faster than myoelectric control, as there are no controller delays [64, 65] and because the direct

proprioceptive feedback pathway is much faster than the visual feedback pathway (Figure 1.1, Figure 1.2). Despite its advantages, the development of BP prostheses has received very little attention since WWII, when compared to the development of externally powered prostheses. Although nowadays many users use BP devices [56], various studies indicate that there is much room for improvement of the current BP prosthesis [61]. This makes BP control a promising field of further research.
The type of BP control that will be used to control the hand, will be shoulder control. Shoulder control can be used for almost every amputation level, in contrary to principles like forearm pronation and elbow flexion. The use of shoulder control will enable the use of one control signal. The new prosthetic hand will use voluntary closing (VC) operation. Although there is no scientific evidence available [66, 67], voluntary closing control is believed to be more intuitive than voluntary opening control, as in VC control there is a positive relation between the actuation force and the grasp force [68, 69]. Furthermore the maximum grasp force in VC control is not dependent of the maximum spring force in the terminal device.
As there will be only one shoulder control signal available to control the multiple joints in the new articulating hand, the principle of underactuation will be used. The hand has to be able to stably perform the two basic grasp patterns: precision grip and power grip [54].
To enable easy and comfortable operation the new designed had should require a low actuation effort from its user. The user effort can be reduced by minimizing the required energy input and by increasing the energy efficiency. The energy input is minimized by making the system stiff. Increasing the energy efficiency can be realized by reducing the energy dissipation due to friction in the joints and glove hysteresis.
It is important that every new prosthesis is properly tested and evaluated. In the first place a prosthesis should be mechanically tested, to evaluate its mechanical performance and durability. Secondly it should be tested by a group of healthy subjects, to evaluate its functional performance, without bothering amputees. In the third stage the prosthesis should be tested by a small group of amputees, and finally by a large group. This thesis will focus on the first two test stages.

1.6 Thesis outline

Chapter 2 and 3 describe the testing of current state-of-the-art body powered terminal devices. Chapter 2 describes the testing of voluntary closing devices and Chapter 3 the testing of voluntary opening devices. The mechanical performance of current terminal devices was measured to get an impression of the current state-of-the-art of body powered devices and to serve as a guideline for the design of an improved hand prosthesis.
Chapter 4 describes the design and testing of two underactuated finger prototypes. One finger has a pulley cable transmission, the other a hydraulic transmission. Tests were performed to select the most efficient way of transmission. The selected finger prototype formed the basis of the design of a new hand prosthesis with underactuated fingers.

Chapter 5 describes the mechanic comparison of a silicone and a PVC cosmetic glove. Both gloves can be used at a prosthesis. Except for their material, the gloves were identical. The test were performed to select the most energy efficient cosmetic glove. The development of the new hand prosthesis did not include redesigning of a cosmetic glove. Therefore it was important to select an available cosmetic glove that required a low amount of energy during operation, in order to reduce the actuation effort for the user.

Chapter 6 describes the design and testing of a compensation mechanism. This mechanism, which fits inside a finger, has an negative stiffness which compensates the undesired positive stiffness of a cosmetic glove. This mechanism was designed and evaluated to be used in the new hand prosthesis, in order to reduce the actuation effort for the user.

Chapter 7 describes the design and testing of the hand prototype. The hand was subjected to mechanical performance tests and an endurance test. Furthermore the hand was functionally tested by a group of able bodied subjects.

In Chapter 8 the results of the prototype testing are discussed and compared to state-of-the-art body powered and externally powered prostheses. The chapter concludes with the main conclusions of this thesis and with recommendations for future work.

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2

Efficiency of voluntary closing hand and hook prostheses

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Abstract

The Delft Institute of Prosthetics and Orthotics has started a research program to develop an improved voluntary closing, body-powered hand prosthesis. Five commercially available voluntary closing terminal devices were mechanically tested: three hands [Hosmer APRL VC hand, Hosmer Soft VC Male hand, Otto Bock 8K24] and two hooks [Hosmer APRL VC hook, TRS Grip 2SS]. The test results serve as a design guideline for future prostheses. A test bench was used to measure activation cable forces and displacements, and the produced pinch forces.

The measurements show that the hands require higher activation forces than the hooks and 1.5 to 8 times more mechanical work. The TRS hook requires the smallest activation force (33 N for a 15 N pinch force) and has the lowest energy dissipation (52 Nmm). The Hosmer Soft hand requires the largest activation force (131 N for a 15 N pinch force) and has the highest energy dissipation (1409 Nmm).

The main recommendations for future prostheses are the following: (1.) Required activation forces should be below the critical muscle force (~18% of maximum), to enable continuous activation without muscle fatigue. (2.) Hysteresis of mechanism and glove should be lowered, to increase efficiency and controllability.

2.1 Introduction

Many patients abandon their upper-limb prosthesis after some time. Studies show rejection-rates varying from 23 to 45% [1]. Patients are often not satisfied with their prosthesis because it does not fulfil their basic demands. These basic demands can be summarized by the words: cosmetics, comfort and control [2]. Prosthesis users have a large range of needs and priorities. They often want their prosthesis to be aesthetically pleasing, comfortable to wear all day, easy to don and doff, and intuitive to control without a high mental or physical load. Current prostheses do not fulfil these demands simultaneously.

This study focuses on the control issue. Currently two types of active prostheses are available: the electric prosthesis and the body-powered (BP) prosthesis [3]. The electric prosthesis most commonly uses surface electromyography (EMG) to control the terminal device, but it can also be controlled by using switches or other sensors (myoacoustic, FSR). The electric prosthesis provides visual feedback and incidental feedback (motor sound, vibration) [4]. It does not provide proprioceptive feedback to the user regarding the opening width of the terminal device, the applied pinch force or the external pinch force disturbances. The absence of proprioceptive feedback decreases the speed and accuracy of both fine [5] and gross [6] motor skills. It also reduces the ease of use of the prosthesis [7, 8]. The BP prosthesis is most commonly controlled by a Bowden cable anchored to a shoulder harness. Pulling the cable results in closing of the prosthesis in voluntary closing (VC) devices, or in opening in voluntary opening (VO) devices. Cable displacement and cable force provide proprioceptive feedback to the user regarding the opening width and the applied pinch force [9]. A major complaint about this type of control is the physical load imposed on the user. Often large activation forces are required. This results in muscle fatigue, discomfort and irritation, particularly in the axilla when using a shoulder harness [10, 11]. To solve one aspect of this problem most VC devices are provided with a locking mechanism. This prevents the user from fatiguing when holding an object for long durations. It also keeps the prosthesis closed while not being used.

The Delft Institute of Prosthetics and Orthotics (DIPO) has started a research program on the development of an improved VC BP hand prosthesis. This prosthesis should require significantly lower physical control effort than commercially available VC BP prostheses. In a first step to this development currently available VC devices for adults were analyzed on mechanical performance properties, as limited data is available on body powered prostheses. LeBlanc *et al*. performed mechanical tests on child size VO and VC devices [12]. Corin *et al*. tested adult size VO devices [13]. In both tests a materials testing machine was used. Various parameters were measured, for example maximum opening width, cable excursion, activation force and pinch force. The activation work was estimated by using the averaged slopes of the force-displacement diagrams, but no dissipated work was estimated. No tests were performed on adult size VC devices. Carlson and Long [14] tested one VO and one

VC hook. In this test the prostheses were measured as a complete system, worn by a user. The activation force and displacement were measured at the harness; therefore, the measured efficiency of the systems was also dependent of the efficiency of the Bowden cable transmission.

2.2 Goal

The goal of this study was to quantify and objectively compare the performance of several commercially available VC upper-limb prostheses. Hand prostheses, as well as hook prostheses were tested. Results of the tests give an impression of the state of the art in the performance of VC prostheses. The obtained values will serve as a guideline for the design of improved VC hand prostheses.

2.3 Methods

ISO 2253.2006 section D6.8 describes a test protocol for VC devices [15]. However the focus of ISO 2253 is primarily on prosthesis safety. Only the recording of the values of the activation force and displacement at which the pinch force reaches 20 N are prescribed. To quantify and compare the performance of the tested prostheses much more parameters were measured in this study:

- Mass of the prosthetic device
- Maximum opening width
- Excursion range of the activation cable
- Work needed for closing the device
- Hysteresis of one cycle (closing and reopening)
- Work needed for closing the device and pinch 15 N
- Activation cable force needed to generate a pinch force of 15 N
- Generated pinch force at an activation cable force increasing from 0 to 100 N
- Pinch force drop induced by the locking mechanism

A pinch force of 15 N was chosen to compare the required activation forces. This force is a bit larger than the 10 N pinch force which is considered to be sufficient for children to perform most tasks of daily living [16]. The amount of work needed to close the prosthesis can be calculated by integrating the required activation force over the path length (cable excursion) over which the force is acting (Equation 2.1). The amount of work can be graphically displayed as the area below the force-displacement-curve (Figure 2.1a).

$$W = \int_0^{\ell} F(x) \cdot dx \qquad (2.1)$$

in which:

W	= Work	[Nm]
ℓ	= Maximum cable excursion	[m]
F(x)	= Force as function of cable excursion	[N]
x	= Cable excursion	[m]

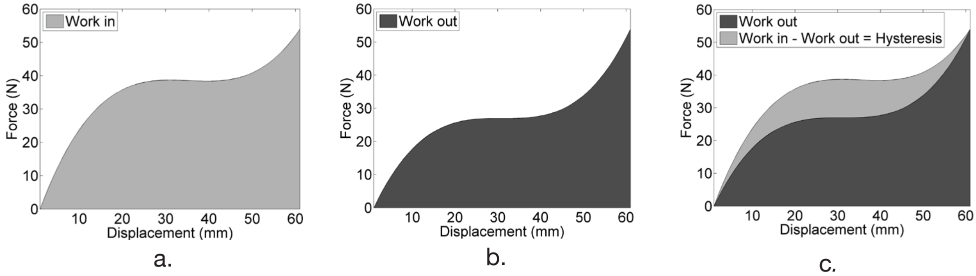


Figure 2.1 Work can be represented by the area below the force-path-curve. The hysteresis or dissipated energy (c) is the difference between the work done on the system (a) and the work returned by the system (b).

The amount of hysteresis, or dissipated energy, of one cycle is a measure of the (in)efficiency of the prosthesis. The difference between the amount of work required to close the prosthesis (Figure 2.1a) and the work returned by the prosthesis during reopening (Figure 2.1b), is defined as the hysteresis (Figure 2.1c, Equation 2.2). An efficient mechanism has a low hysteresis.

$$Hysteresis[Nm] = Work_{closing}[Nm] - Work_{opening}[Nm] \qquad (2.2)$$

Tested prostheses

All tested prostheses are commercially available VC prostheses (Figure 2.2). The oldest designs are the ARPL devices, which were developed in 1945 [17]. The newest design is the Hosmer Soft hand, which was introduced in 2002 [18]. The Lite Touch Adult hand of TRS was not tested. It resembles a hand shaped hook and is not provided with any glove. Therefore its efficiency is expected to be similar to the TRS hook.

1. Hosmer APRL VC Hand (L), Size 8	*2. Hosmer APRL VC Hook (R)*	*3. Hosmer Soft VC Male Hand (R), Size 7 3/4*	*4. Otto Bock VC Hand 8K24 (L), Size 7 3/4*	*5. TRS VC Hook Grip 2SS*

Figure 2.2 Overview of the tested voluntary closing prostheses: three hands and two hooks.

This study focuses on the efficiency of the mechanism, rather than on the characteristics of the cosmetic glove. Therefore, the hands were tested without a cosmetic glove. The Otto Bock hand and the Hosmer Soft VC hand were tested with their inner glove applied. The APRL hand has no inner glove. The tests with the Otto Bock hand were repeated with the cosmetic glove and the inner glove applied, to study the effect of the cosmetic glove. The tests were also repeated with the bare frame in order to study the effect of the inner glove. All tested devices were new and previously unused. No adjustments were made to the devices. With the exception for the TRS hook, all tested prostheses have an automatic locking mechanism.

Apparatus and procedure

A custom-build test bench was used to measure the tensile force and the displacement of the activation cable of the prosthesis (Figure 2.3 and 2.4). The bench was manually operated. The prostheses were controlled at a low opening and closing speed (fingertip speed about 3 mm/s), to reduce the viscous behaviour of the inner gloves and the cosmetic glove. The pinch force applied by the prosthesis was measured using a custom-build pinch force sensor. The sensors were connected to a laptop by a data acquisition interface. All components used are listed in Table 2.I.

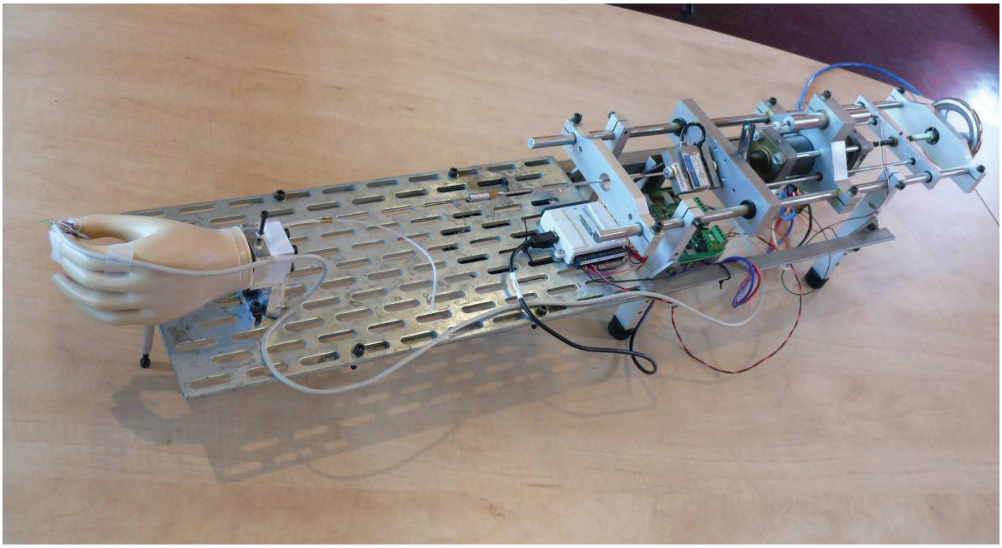


Figure 2.3. The Otto Bock hand mounted in the test bench. The bench was used to measure the cable force and the cable excursion together with the pinch force produced by the terminal device.

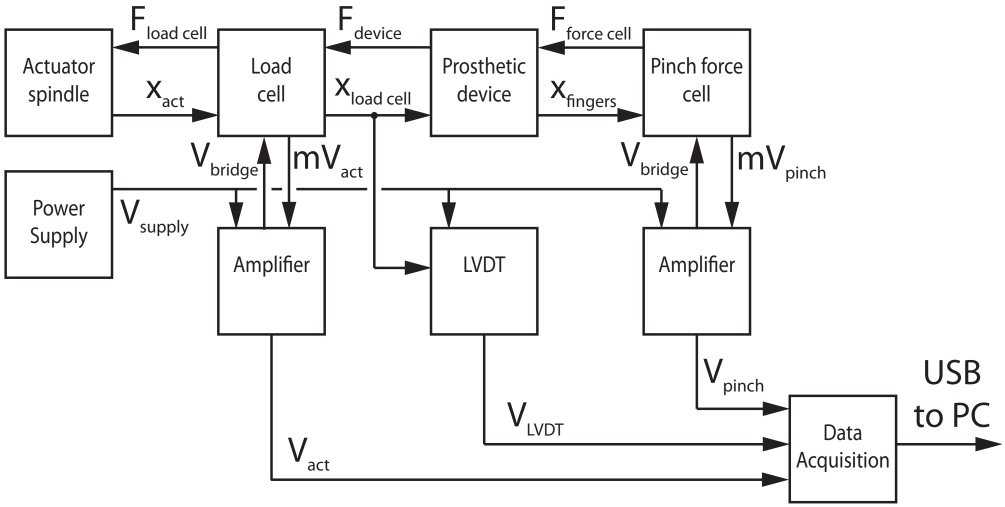


Figure 2.4 Schematic overview of the test bench. In which F = force, x = displacement, V = volt, mV = millivolt and LVDT = Linear Variable Differential Transducer.

Table 2.I Components used in the test bench

Component	Description
Force sensor	Zemic: FLB3G-C3-50kg-6B
Amplifier	Scaime: CPJ
Linear displacement sensor (LVDT)	Schaevitz: LCIT 2000
Power supply	EA: EA-PS 3065-05 B
Computer interface	National Instruments: NI USB-6008
Pinch force sensor	Double leave spring with strain gauges

All devices were subjected to three different tests.

- **Closing test.** A small steel plate (thickness = 1 mm) was placed in between the fully opened fingers (Figure 2.5.1). The cable was pulled until the prosthesis was closed. Thereafter the cable was released for the first time, thus activating the locking mechanism. The prosthesis was reopened by pulling and releasing the cable for the second time.
- **Pinch test.** The pinch force sensor (thickness = 10 mm) was placed in between the fully opened fingers (Figure 2.5.2a). The cable was pulled until a pinch force of 15 N was reached. Thereafter the cable was released for the first time, thus activating the locking mechanism. The prosthesis was reopened by pulling and releasing the cable for the second time.
- **Pull test.** The pinch force sensor (thickness = 10 mm) was placed in between the fully opened fingers (Figure 2.5.3). The cable was pulled until an activation force of 100 N was reached.

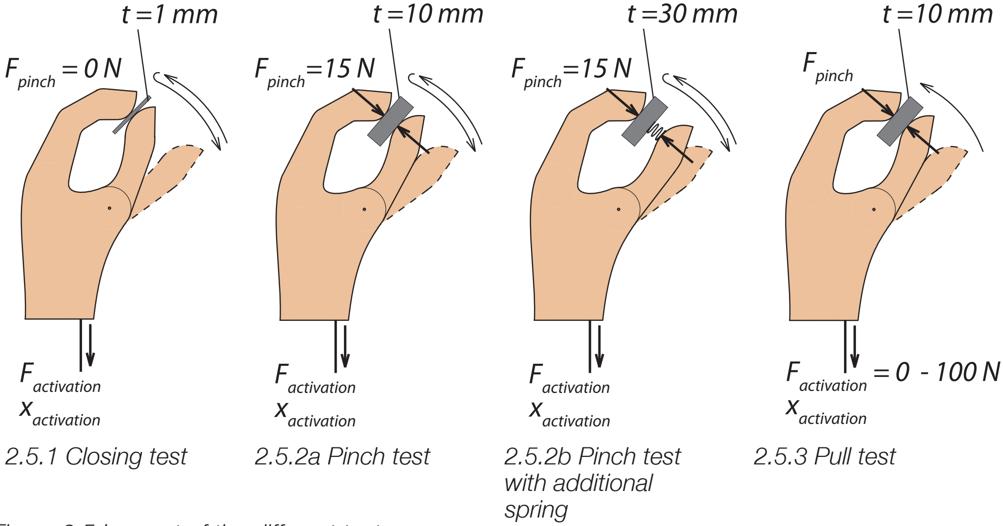


Figure 2.5 Lay-out of the different tests.

The closing and pinch tests were repeated four times for each device, to obtain an average value. The acquired data was processed in MATLAB [19]. Plots were made showing the ‘cable displacement vs. cable activation force’ and the ‘cable activation force vs. pinch force’. The work and hysteresis values were calculated for the last 35 mm of the cable excursion, which is within the range of all devices. This enabled comparison of the different prostheses. The pull test was performed once for each device. One combined plot was made showing the ‘cable activation force vs. pinch force’ of all pull tests.

Testing the locking mechanism

After activation of the locking mechanism, the pinch force drops somewhat. The magnitude of this drop was, where present, obtained from the data of the pinch test. It was used as a measure of effectiveness of the locking mechanism. A larger drop will result in a reduced grip, which means that the locking mechanism is less effective. One supplementary test was performed with the Otto Bock hand, because the results of its locking mechanism showed an unexpected behaviour. For this prosthesis, the pinch test was repeated with the inner glove and cosmetic glove applied, while a spring (length = 20 mm, stiffness k = 4 N/mm) was placed between one finger and the pinch force sensor (Figure 2.5.2b). This test was repeated twice.

2.4 Results

An overview of the geometrical properties and the test results for the prostheses is given in Table 2.II. Notice that the Hosmer APRL hand and hook have two opening spans. The hand has an adjustable thumb, which can be locked in two positions. The hook has a setting in which the maximum opening of the hook is limited.

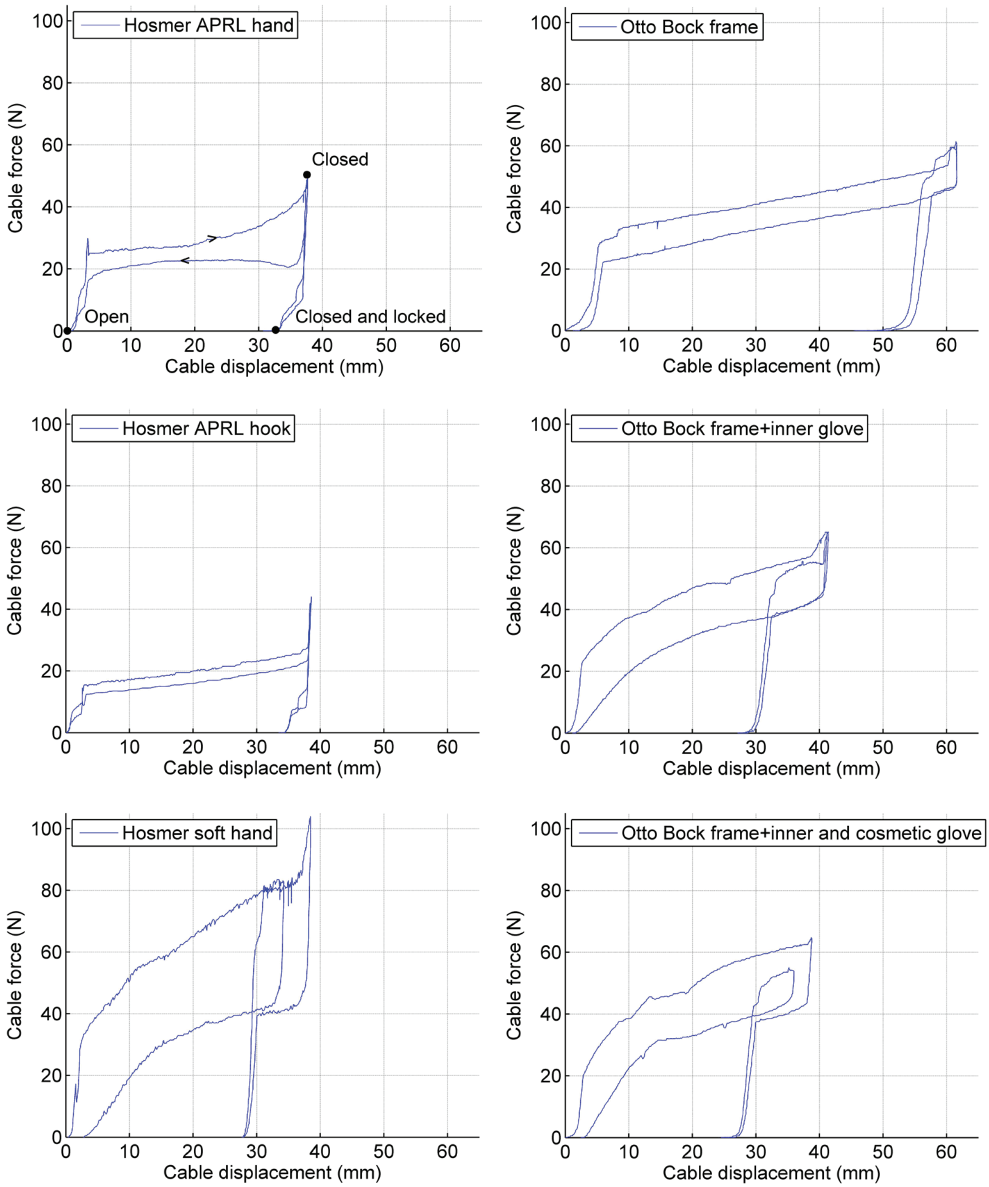
Table 2.II *Overview of the geometrical properties and the test results of the tested prostheses.*

	Prosthesis	Mass (gr)	Opening width (mm)	Maximum cable excursion (mm), $n=4$	Work closing (Nmm), $n=4$	Cycle hysteresis (Nmm), $n=4$	Work closing and pinching 15 N (Nmm), $n=4$	Required cable force for a 15 N pinch (N), $n=4$	Pinch force at a cable force of 100 N (N)	Pinch force drop at a 15 N pinch (N), $n=4$
1	Hosmer APRL hand, 52541 (L) size 8	347	44 (70*)	37 ± 0.1	1058 ± 4	298 ± 8	831 ± 1	61 ± 0.6	41	7.3 ± 0.4
2	Hosmer APRL hook, 52601 (R)	248	73 (33**)	38 ± 0.1	720 ± 6	138 ± 3	687 ± 2	62 ± 0.0	30	10 ± 1.5
3	Hosmer soft hand, 61794 (R) size $7\frac{3}{4}$	366	71	38 ± 0.3	2292 ± 12	1409 ± 37	2176 ± 16	131 ± 0.7	5	14 ± 1.7
4	Otto Bock, 8K24 (L) size $7\frac{3}{4}$, frame	220	100	60 ± 0.5	1624 ± 8	389 ± 19	1545 ± 1	78 ± 0.3	28	6.7 ± 0.5
5	Otto Bock, 8K24 (L) size $7\frac{3}{4}$, frame + inner glove	350	69	41 ± 0.2	1639 ± 24	672 ± 8	1694 ± 16	90 ± 0.9	19	5.9 ± 0.4
6	Otto Bock, 8K24 (L) size $7\frac{3}{4}$, frame + inner glove and cosmetic glove	423	57	38 ± 0.5	1710 ± 20	681 ± 23	1636 ± 29	98 ± 0.5	14	6.5 ± 0.3
7	TRS hook, Grip 2S	318	72	49 ± 0.1	284 ± 3	52 ± 1	243 ± 3	33 ± 0.2	58	–

*Thumb positioned in 'wide' position. **Hook adjusted to small range.

Closing test

The measured activation cable forces and displacements are plotted in Figure 2.6. The calculated amount of work needed for closing the hand or hook, and the calculated hysteresis of one cycle of closing and reopening are shown in Figure 2.7.



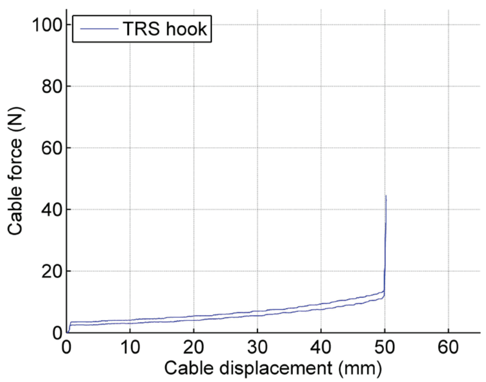


Figure 2.6 The measured forces at the activation cable as function of the cable displacement, during the closing test. The clockwise cycle starts and ends at 0 mm and 0 N, when the hand is fully open. At the maximum cable displacement and force, the hand is closed.

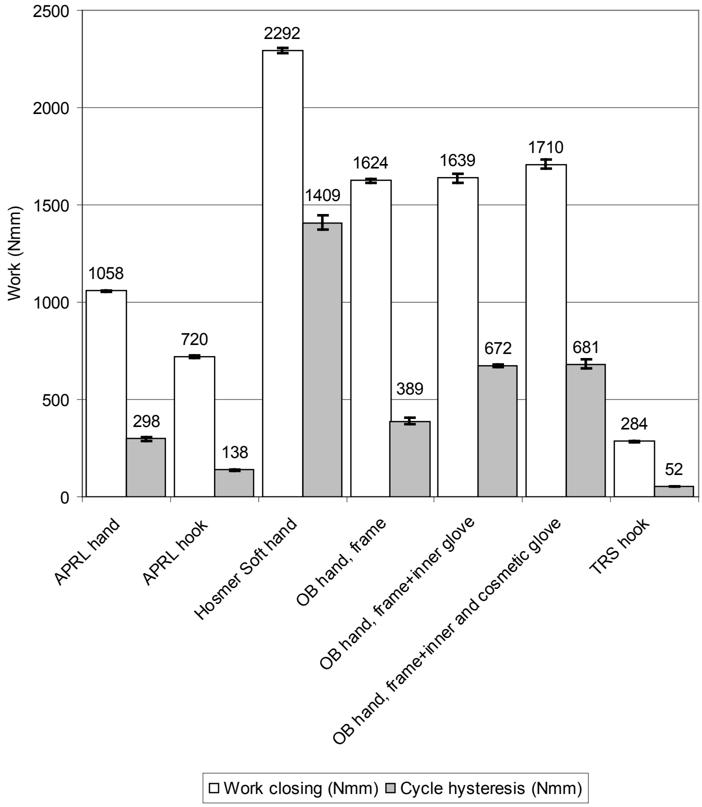


Figure 2.7 Results of the closing test: The work to close the device is displayed together with the amount of energy dissipated during one cycle of opening and closing.

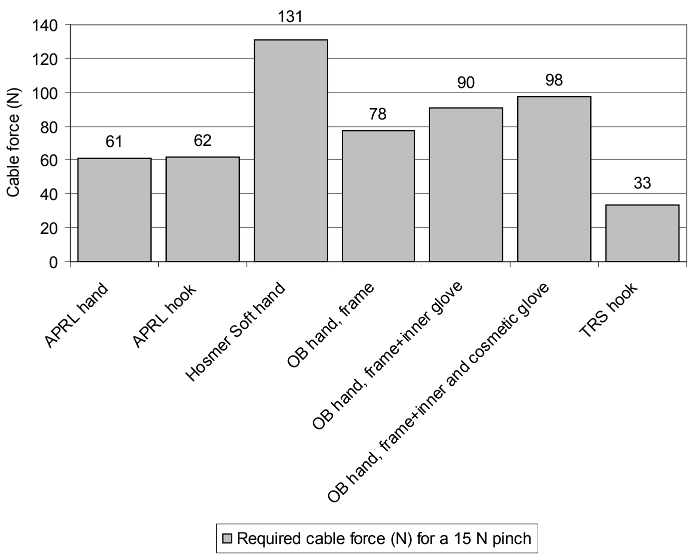


Figure 2.8. Required cable force (N) to produce a pinch force of 15 N.

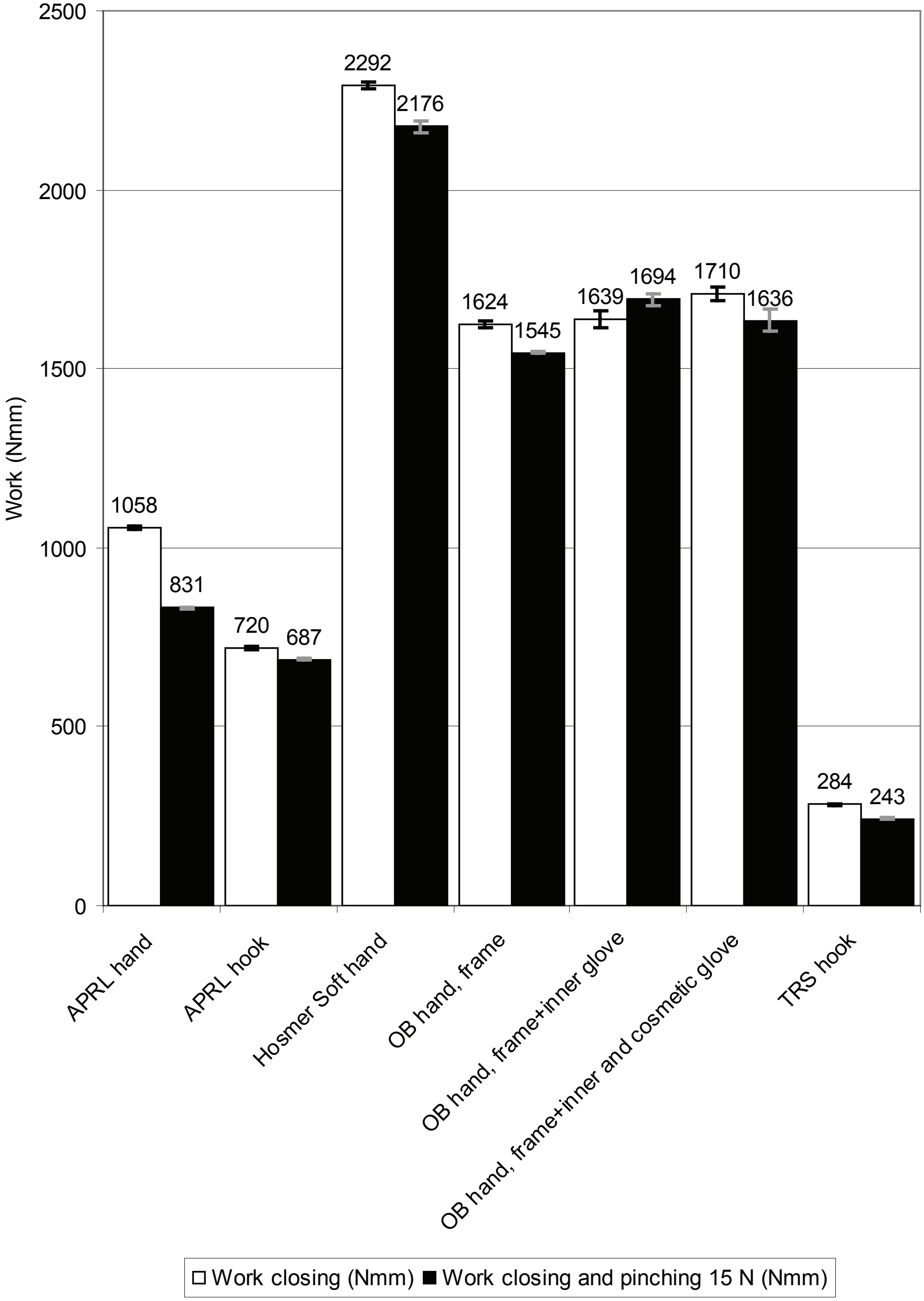


Figure 2.9 Calculated work required for the closing test and for the pinch test. Closing the prostheses entirely requires a different amount of work than clamping the pinch force sensor (thickness = 10 mm) to a pinch force of 15 N.

Pinch test

The activation cable force required to generate a pinch force of 15 N, varies from 33±0.2 to 131±0.7 N among the different devices (Figure 2.8). Closing the prostheses entirely, i.e. with no object present, requires a different amount of work than clamping the pinch force sensor (thickness = 10 mm) to a force of 15 N (Figure 2.9).

Pull test

Figure 2.10 shows the pinch forces as a function of the activation force for each device. All devices show a linear relation. The minimum required activation force to initiate pinching is different for each device and varies from 10 to 85 N.

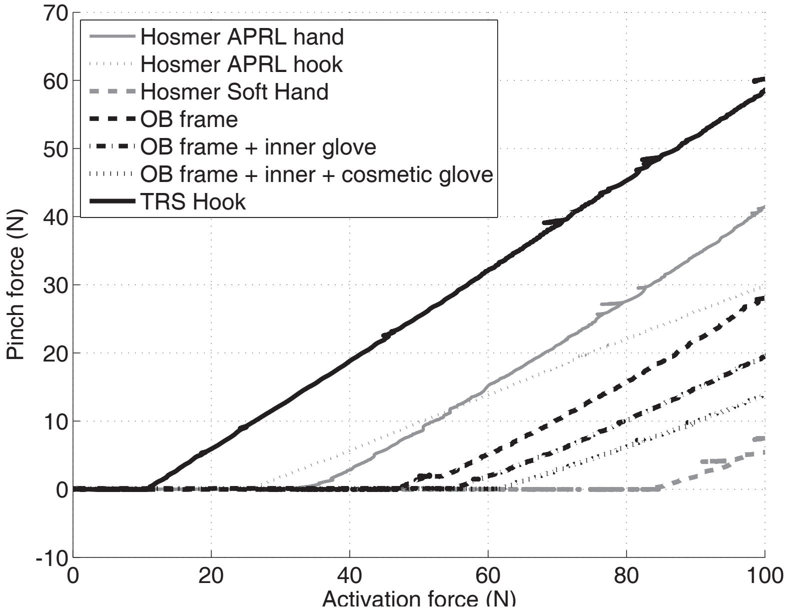


Figure 2.10 The 0 to 100 N pull test. The curve is initially horizontal as the activation force increases, while the pinch force remains at zero. When the moving finger touches the force sensor, the pinch force begins to increase along with the activation force.

Locking mechanism

In the devices provided with a locking mechanism, the pinch force drops after the mechanism is activated (Figure 2.11, 3rd arrow). This drop varies in magnitude from 50 to 90% of the initial pinch force (~ 15 N) for the different devices (Figure 2.12). During deactivation of the Otto Bock locking mechanism, before the cable is released for the second time, the hand opens slightly and the pinch force drops close to zero (Figure 2.11, 4th arrow). In the supplementary test, in which the Otto Bock hand pinches a spring, the pinch force drops 20% during activation of the locking mechanism. Again, during deactivation the pinch force drops further, close to zero, before the cable is released. No results were obtained from the TRS hook, as it is not provided with an internal automatic locking mechanism.

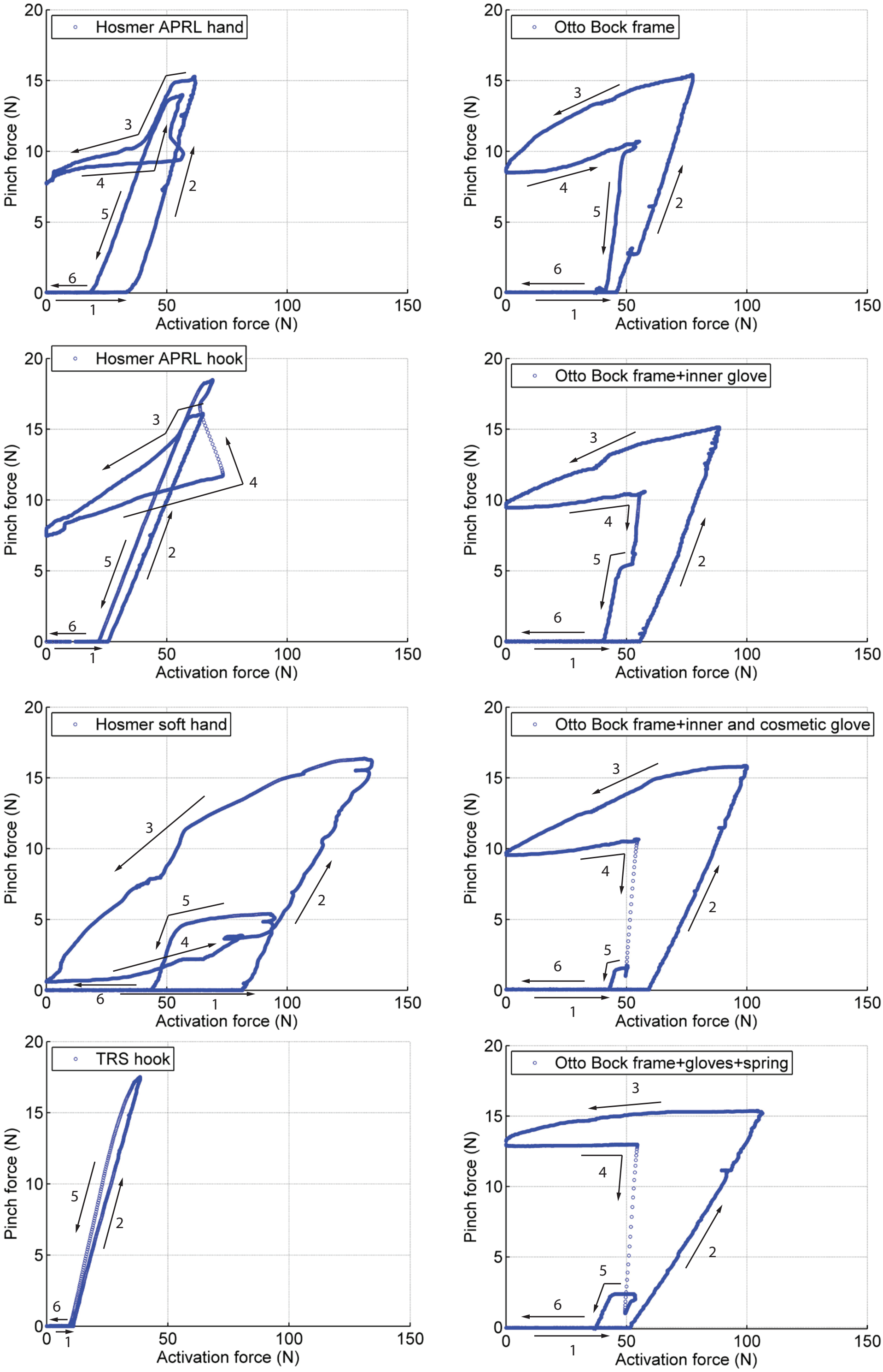


Figure 2.11 Activation force vs. pinch force-diagrams. Explanation on basis of the Otto Bock hand: 1 First cable pull, the fingers close. 2 Fingers touch the pinch load cell, the pinch load builds up. 3 First cable release, the lock is activated and the pinch force drops. 4 Second cable pull, lock unlocks. 5 Final cable release, pinch load decreases. 6 The fingers open.

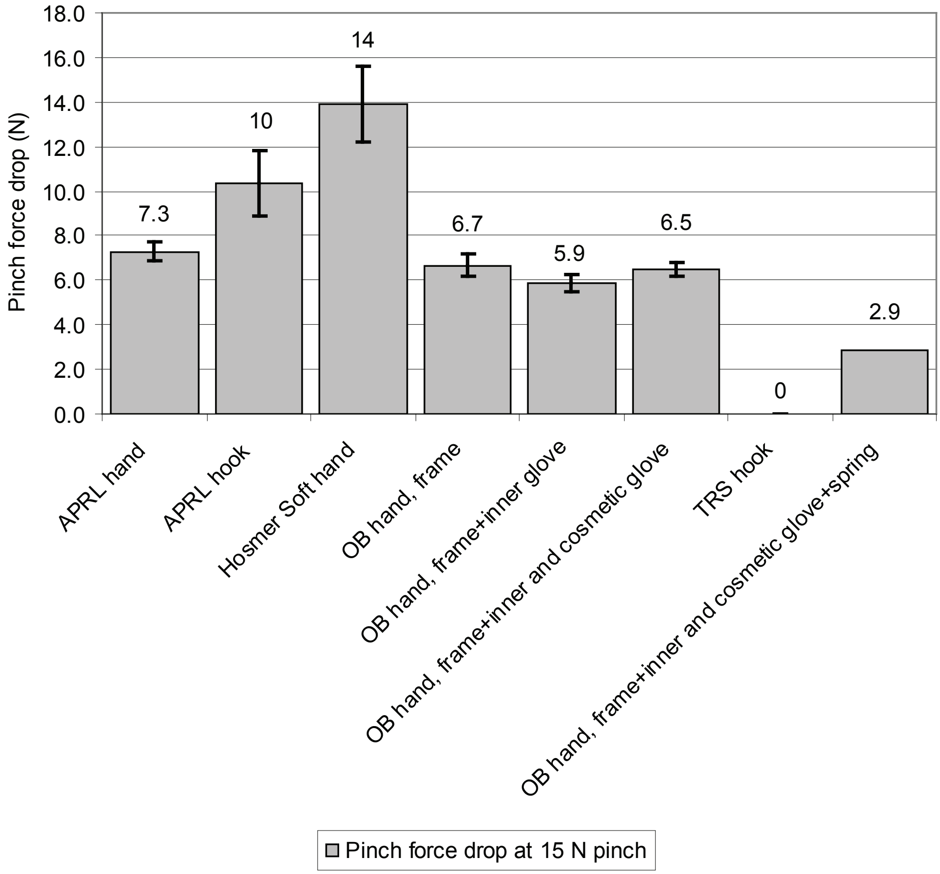


Figure 2.12. Drop in pinch force, after activation of the locking mechanism at an initial pinch force of approximately 15 N. The TRS hook is not tested, as it is not provided with an automatic locking mechanism.

2.5 Discussion

Closing test

The results show that VC mechanical hands require higher activation forces than VC hooks (Figure 2.8), and require 1.5 to 8 times more mechanical work (Figure 2.7). This is in line with previous studies performed on VO devices [12, 13]. The energy dissipation in hands is 2 to 27 times higher than in hooks. The Otto Bock device has a larger hysteresis when the inner glove is applied. Still, without the inner glove applied, the Otto Bock hand mechanism has a larger hysteresis and requires more work than the APRL hand, which has no inner glove. The inner glove also accounts for 30% of the total mass of the Otto Bock hand. In future designs it is recommended to decrease the mass and hysteresis of the inner glove, or abandon its use.

Applying the cosmetic glove on top of the inner glove at the Otto Bock hand gives a small increase of the required work. Remarkably, the amount of hysteresis does not increase significantly. Herder *et al*. [20] measured cosmetic glove hysteresis values between 30 and 90 Nmm, for a different glove size, by using a different set-up. Possibly the hysteresis of the cosmetic glove is compensated by the behaviour of the locking mechanism. The mechanism has a smaller hysteresis loop when the cosmetic glove is applied (Figure 2.6). The cause of this behaviour is unknown.

Pinch test

The difference in activation forces is the largest between the Hosmer Soft hand and the TRS hook (Figure 2.8). Even without a cosmetic glove, the Hosmer Soft hand requires almost four times more force than the TRS hook, to create a pinch force of 15 N. In most devices, closing the prosthesis entirely requires more work than clamping the pinch force sensor with a force of 15 N (Figure 2.7). Because of the sensor thickness the fingers do not fully close. Consequently, the considered closing trajectory shifts forward. As a result the opening spring is less loaded, so less work is required. The required extra work to build up the pinch force is relatively low, due to the stiff pinch force sensor. In this case [object size 10 mm; pinch force 15 N] the amount of work “gained” because of the trajectory shift, is larger than the required extra work to apply the pinch force.

Pull test

The minimal force necessary to close the fingers and to start building up a pinch force differs widely among the various devices (Figure 2.10). The Hosmer Soft hand requires the largest activation force to start pinching (83 N): 7.5 times more than the TRS hook, which requires the lowest force (11 N). The results are in accordance with the outcome of the closing tests and the pinch test (Figure 2.6, Figure 2.8). Carlson and Long [14] measured a 40% lower pinch force at an activation force of ~83 N for the TRS hook (29 N instead of 48 N in the current test). This can largely be explained by the inefficiency of the Bowden cable, which was included in their test.

Required activation force

The maximum force that can be generated using a shoulder harness is 280±24 N [9]. Although the measured maximum forces in the pinch test are within this range, some remarks have to be made:

- The maximum force, as mentioned in the literature, was obtained by measurements on non-amputees. A study showed that children with a congenital arm defect have much less strength in their arms than typical-bodied children [21]. It is expected that the same is true for adults.
- Exerting the maximum force for a longer time is impeded by discomfort, caused by the harness and fatigue of the muscles. A muscle can only be contracted continuously without fatigue when the muscle force is lower than the critical force, which is about 18% of the maximum muscle force [22]. Intermittent contractions, at a work-to-rest ratio of 0.5, can be performed without fatigue at about 38% of the maximum muscle force. Psychophysical aspects of body control, such as

maximum comfortable activation force and range, and control accuracy, will be part of future studies of DIPO.

- The pinch force of 15 N produced in the test is relatively low. For some activities of daily living a larger pinch force is required (e.g. prehension of a folded sock: 34 N) [23].
- The Hosmer Soft VC hand and the APRL hand were tested without a cosmetic glove. The required activation force with the cosmetic glove applied is expected to be somewhat larger.
- The harness activation force has to be 20 to 40% larger than the cable force measured in the test, due to the inefficiency of the Bowden cable transmission [14, 24, 25].

Taking these remarks into account, only the TRS hook can be used without fatigue. With this device it is also possible to produce the largest pinch forces for a given activation force. For the other devices there is a trade-off between the produced pinch force and the duration the force can be maintained. The less efficient the device is, the larger the required activation force must be, and the faster the user gets fatigue. In this respect, the usability of the Hosmer Soft hand will be very limited.

Cable excursion

The maximum cable excursion by shoulder control is 53±10 mm [9]. All measured cable excursions are within the average range (Table 2.2). Having a maximum cable excursion of 49±0.1 mm, the TRS hook is not within the average range minus the standard deviation. A part of the users will not be able to use the full opening range of the hook. The maximum cable excursion of the Otto Bock bare frame is also not within the maximum range. However this is not relevant, as it is never used without both gloves.

Locking mechanism

The measured pinch force drops (5.9±0.3 to 14±1.7 N) are relatively high, compared to the initial pinch force of approximately 15 N (Figure 2.12). To maintain a secure grip after the lock is activated, a larger initial pinch force is required. The maximum producible pinch force is limited by the object strength and by the capacity of the user. Therefore it will often not be possible to hold an object secure using one of the tested locking mechanisms. In future designs the locking mechanisms in all prostheses should be improved to maintain a better grip, or be abandoned to improve the efficiency of the device.

The pinch force sensor used in this study was stiff. Pinching a compliant spring reduces the pinch force drop in the Otto Bock hand from 43% to 19%, which is still quite large. The behaviour of the Otto Bock locking mechanism during unlocking is remarkable. When the cable is pulled for the second time, the fingers suddenly open a little and the pinch force instantaneously drops close to zero. During step 4 in Figure 2.11 it is not possible to control the decrease of pinch force in the Otto Bock mechanism. The pinched object is suddenly released. This behaviour was the same for the stiff and the compliant object.

Study limitations

One test was performed with a cosmetic glove applied, to compare the magnitude of the added work and hysteresis to that of the mechanism without one. Mechanic characteristics vary widely among gloves, even for gloves of the same brand and size [20]. To study the effect of a cosmetic glove on a mechanism, multiple tests with different gloves have to be performed. Therefore the effect of the cosmetic glove was left out of the scope of this study. It would be interesting to perform such a study in the future.

All devices were tested using factory settings. The pre-tension of the opening spring in the Hosmer Soft hand is adjustable by disassembling the hand. The spring in the Otto Bock hand can be adjusted by the Otto Bock Service centre. The pre-tension ensures full opening of the device. It has to overcome the hysteresis of the glove plus the friction in the Bowden cable. The pre-tension values in the hands are between 20 and 30 N. The values in the hooks are around 12 N for the APRL hook and around 3 N for the TRS hook. Reducing the pre-tension value of the APRL hook to that of the TRS hook, would reduce its amount of work by one third. The amount of hysteresis might slightly reduce due to the reduction of internal friction. Reducing the pre-tension in the hands might also be possible. However, this will also result in an undesirable reduction of the maximum opening width, because of the glove hysteresis. It would be interesting to study the effects of the spring pre-tension and stiffness. The spring stiffness can be changed by replacing the spring.

2.6 Conclusions

Five VC devices were tested: three hands and two hooks.

- Large differences were observed among the devices. Mechanical hands require 1.5 to 8 times more mechanical work than hooks. The hysteresis or energy dissipation in hands is 2 to 27 times higher than in hooks. The TRS hook requires the smallest activation force (33±0.2 N), the Hosmer Soft hand the largest (131±0.7 N). The results are in line with previous studies performed on VO devices.
- All measured activation forces are within the maximum range as determined by Taylor [9]. The activation force of the TRS hook is also within the critical force range and can therefore be maintained continually without fatigue. For the other devices the duration over which the pinch force can be maintained is limited by the magnitude of the required activation force, and is dependent on the desired pinch force and the efficiency of the prosthesis.
- All measured cable excursions are within the average of the maximum range determined by Taylor [9]. The range of the TRS hook is not within the average range minus the deviation. Therefore a part of the users will not be able to use the full opening range of the hook.

- The measured drops in pinch force, after activation of the locking mechanism, are relatively high compared to the initial pinch force (~ 40-90%). A larger initial pinch force is required to maintain a secure grip after the lock is activated. It will often not be possible to hold a stiff object secure, using one of the tested locking mechanisms. When pinching a compliant spring, the pinch force drop in the Otto Bock hand was reduced, but it remained quite large (19%). Remarkably the Otto Bock hand has a second pinch force drop, directly after unlocking. It is not possible to decrease the pinch force gradually.
- The following recommendations can be given for future designs:
- Activation forces should be lowered within the critical force range, to enable continuous activation without muscle fatigue.
- The cable activation range should be within the range of all users, or should be adjustable to each individual user.
- Hysteresis of the mechanism and the glove should be lowered, to increase the efficiency and controllability.
- The mass and hysteresis of the inner glove should be decreased, or its use should be abandoned.
- Locking mechanisms should either be improved or abandoned.

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3

Efficiency of voluntary opening hand and hook prosthetic devices, 24 years of development?

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Abstract

Quantitative data on the mechanical performance of upper limb prostheses are very important in prostheses development and selection. The primary goal of this study was to objectively evaluate the mechanical performance of adults' size voluntary opening prosthetic terminal devices, and to select the best tested device. A second goal was to see whether voluntary opening devices have improved the last two decades. Nine devices, four hooks and five hands, were quantitatively tested [Hosmer 5xAl Hook, Hosmer Sierra Hook, RSL Steeper Carbon Gripper, Otto Bock 10A60 Hook, Hosmer Becker Imperial Hand, Hosmer Sierra Hand, Hosmer SVO Hand, RSL Steeper VO Hand, Otto Bock VO Hand]. The pinch forces, the activation forces, cable displacements, mass and opening span were measured. The work and hysteresis were calculated. The results were compared to data from 1987. Hooks required lower activation forces and delivered higher pinch forces than hands. The activation forces of several devices were very high. The pinch forces of all tested hands were too low. The Hosmer 5xAl Hook with 3 bands was the best tested hook. The Hosmer Sierra Hand was the best tested hand. No improvements of voluntary opening devices were found, compared to the data from 1987.

3.1 Introduction

Despite the developments made in electrical prostheses, a significant number of adults and children wear body powered prostheses [1, 2]. These users often prefer the relative benefits of the body powered prostheses, like low weight, technical reliability, low cost, and proprioceptive feedback, over the benefits of the electrical prostheses, such as the grip strength and the fact that a harness is not necessary in most cases. Body powered prostheses however, also have a number of drawbacks. A major complaint is activation force, which is often quite large [3]. This is uncomfortable, and can lead to complaints and irritation of the shoulder and the axilla [4, 5]. A further problem is frequent failure of the activation cable [4]. Although being lighter than electrical devices, also body powered devices are often perceived to be heavy by their users [4]. The current paper determines the mechanical efficiency of currently available body powered voluntary opening (VO) hands and hooks. This helps patients in selecting an appropriate device and manufactures in improving their designs.

To efficiently and smoothly use a body-powered prosthesis it is necessary that the device is mechanically efficient and requires a low activation force. A previous study on voluntary closing (VC) devices showed that, except for the TRS hook, nearly all tested devices were inefficient and required high activation forces, that users could find uncomfortable. It is therefore interesting to study the performance of voluntary opening (VO) devices [6]. Therefore, the study described in this paper will objectively measure the mechanical performance of VO terminal devices for adults. On the basis of these measurements it is possible to compare the performance and the efficiency of the tested devices.

No recent data is available on the mechanical efficiency of VO devices. Corin *et al*. [7] published measurements on a broad range of adult- and child-size VO devices, in 1987. Around that period, Carlson and Long [8] also measured one VO and one VC hook and LeBlanc *et al*. [9] measured several VO and VC child-size prehensors. Our study presents new test data, which is obtained from experiments similar to that of Corin. Comparing the current findings with those of Corin *et al*. makes it possible to give an objective view of how much VO terminal devices have been improved during the past decades. In the clinic the results can be used to select an appropriate prosthetic terminal device for a patient. Manufacturers can also use the results to improve their prosthetic components.

3.2 Goal

The primary goal of this study was to objectively compare different voluntary opening terminal devices for adults, by quantitatively measuring the mechanical work, energy dissipation, maximum cable force and excursion, pinch force, opening span and device mass. This comparison was used to select the most suitable hand and hook prosthesis based on the measured mechanical characteristics. The second goal was to see whether voluntary opening devices have improved over the last two decades. Therefore, the results will be compared to results from a study in 1987.

3.3 Methods

Tested materials

In this study nine VO terminal devices were tested, four hooks (Figure 3.1) and five hands (Figure 3.2). All devices were of comparable size (around size 7 ¾), which corresponds to a small adult male hand, or a large adult female hand. All devices were for left side use. All devices were brand new and not used before.

Hosmer Hook 5xAl 1-3 bands	*Hosmer Sierra Hook 2 settings*	*RSL Steeper Car- bon Gripper 2 settings*	*Otto Bock Hook 10A60 2 settings*

Figure 3.1 Overview of the tested hooks. The pictured devices are not the actual measured devices.

Hosmer Becker Imperial hand size 8	*Hosmer Sierra Hand size 8 2 position thumb*	*Hosmer Soft VO Hand size 7 3/4*	*RSL Steeper VO Hand size 7 3/4*	*Otto Bock VO Hand size 7 3/4*

Figure 3.2 Overview of the tested hands without a cosmetic glove applied. The three hands on the right have an inner glove, which protects the cosmetic glove. The hands on the left side do not have an inner glove. The pictured devices are not the actual measured devices. (VO = Voluntary Opening)

All the hooks have adjustable settings of the spring force. The Sierra Hand has an adjustable thumb, which has two different opening positions. These settings can be easily adjusted by the user. The devices were tested for each individual setting. Some devices have a spring that is adjustable by the prosthetist or manufacturer. These devices were tested with the manufacture’s settings.

Test equipment

A simple test bench was used to measure the force and displacement of the activation cable (Figure 3.3). The test bench was custom build, and consisted of standard components. An LVDT (Positek: LPIS P101) measured the displacements, and a load cell (Zemic: FLB3G-C3-50kg-6B) measured the cable activation force. The pinch forces were measured with a custom build double leave strain gauge load cell. The voltage of each load cell was amplified with an amplifier (Scaime: CPJ). All data were recorded using a data acquisition system (National Instruments: NI USB-6008).

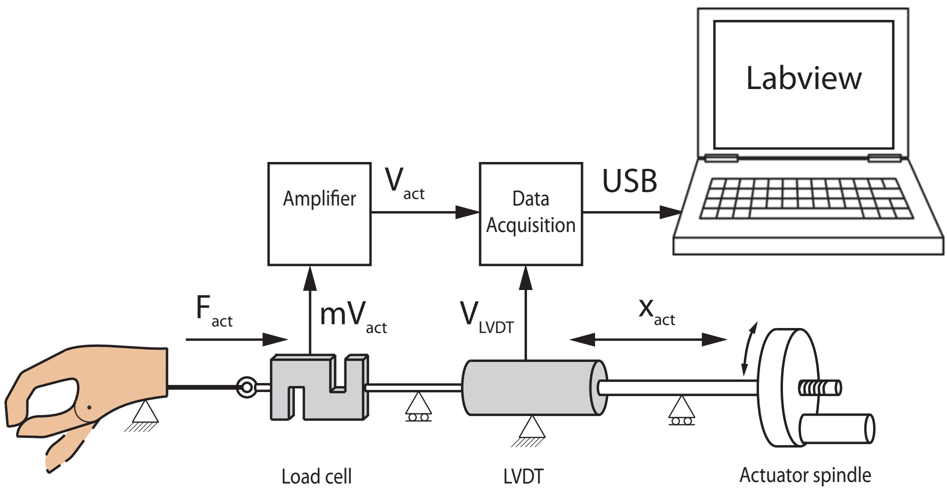


Figure 3.3 Schematic overview of the test set up during the first two tests. Cable activation force and displacement are measured.

Test protocol

For each device the mass and the maximum opening span were measured. Subsequently each device was subjected to three different tests:

1. Open and close test. The cable was slowly pulled (~2 mm/s) till the device was fully opened. Then the cable was released at about the same speed, till the device was closed again. During this test the cable force and displacement were measured.
2. Open and close test (50 mm). Similar as test 1. Only in this test the device was opened till the minimum opening between the fingers was 50 mm.
3. Pinch force test. The pinch force sensor was placed between the fingertips of the device, and the pinch force was measured. The force was measured for an opening width of 10, 20 and 30 mm.

Both, the hands and the hooks, were tested according to the same protocol. The hands were tested with and without a cosmetic glove. During the gloved tests, each hand was gloved with the standard PVC glove recommended by the manufacturer of that hand. To normalize the mass data, the long sleeved Steeper cosmetic glove was shortened to a length similar to that of the other cosmetic gloves (around 10 cm below the wrist plane). During the ungloved tests, the hands which had an inner glove, were tested with the inner glove still applied. The Becker hand was not tested with a cosmetic glove, as there was no matching glove available at the moment of testing. Each test was preceded by two initial runs, to prevent for transient behaviour. Test 1 and 2 were repeated four times, to obtain an average value. Test 2 was performed to enable a comparison among the different devices, since they all have different opening spans. Test 3 is a static test, and was therefore not repeated. The pinch force depends on the opening width of the fingers, not on their motion. The opening width before closing and pinching might have a small influence on the pinch force. To minimize this influence the devices were opened only a few millimetres more than the thickness of the pinched object.

Processing the data

From the data the maximum excursion and activation force were determined. To calculate the amount of energy (or work) to open the device, the measured forces were integrated along the displacement path. Also the amount of energy that was 'returned' during opening was calculated. The difference between the 'input' and the 'output' energy, is the dissipated energy, or hysteresis. This is an indicator of the efficiency of the mechanism. The higher the efficiency, the lower the hysteresis. Before processing the data, the start and end data with a cable force below 1 N were cut off, as these data are not of interest.

Comparison to the data from 1987

The acquired data was compared to the data of the adult-size devices, tested by Corin *et al*. in 1987 in a similar test. This was done for the required work and activation force to fully open the device (test 1) and for the pinch force for small objects (≤ 10 mm, test 3). The mean values of our data were calculated for the hooks and gloved hands in all settings. The mean values of Corin's data were calculated for the following adult-size hook devices: Hosmer SSS-555 (1-3 band), Hosmer 88x (1-3 bands), Hosmer 99x (1-3 bands), and the gloved hand devices: UNB Steeper 2.50" and 2.75", Hosmer Sierra hand, Hosmer Robins Aids, Hosmer Becker, Hosmer #201, #301 and #401, Otto Bock 6.75" and 7.75". An unpaired t-test was used to test whether significant ($a < 0.05$) differences could be found, between the data of both studies.

Best tested prosthetic devices

Finally it is interesting to see which hand and hook were most suitable for daily use, according to the demands of the user. To enable a number of activities of daily living, the devices should have a pinch force above 20 N [10, 11]. Their activation force should be as low as possible. The cable excursion should be within the acceptable range (< 53 mm [12]). The mass should be as low as possible. The results will be compared to these user demands.

Table 3.I Overview of the test results of the VO hand and hook tests

Voluntary opening terminal device	Mass	Max. open.	Max. cable excursion (n=4)				Max. cable force (n=4)				Open and close test Test 1, full (n=4)				Open and close test Test 2, 50 mm (n=4)				Pinch force Test 3		
			Test 1 full		Test 2 50 mm		Test 1 full		Test 2 50 mm		Work		Hysteresis		Work		Hysteresis		10 mm	20 mm	30 mm
	(g)	(mm)	(mm)		(mm)		(N)		(N)		(Nmm)		(Nmm)		(Nmm)		(Nmm)		(N)	(N)	(N)
			AVG	SD	AVG	SD	AVG	SD	AVG	SD	AVG	SD	AVG	SD	AVG	SD	AVG	SD			
Hosmer Hook 5XAI 1 band	87	88	45	± 0.2	24	± 0.1	48	± 12.0	25	± 0.3	1128	± 14	290	± 3	574	± 3	120	± 4	9	9	9
Hosmer Hook 5XAI 2 bands	90	88	46	± 0.1	25	± 0.1	72	± 3.5	50	± 0.2	2248	± 10	394	± 6	1173	± 6	154	± 3	14	19	20
Hosmer Hook 5XAI 3 bands	92	88	46	± 0.1	25	± 0.0	95	± 4.2	71	± 0.2	3206	± 18	458	± 4	1684	± 4	186	± 4	24	29	33
Hosmer Sierra Hook Setting 1	242	66	34	± 0.1	26	± 0.0	67	± 7.9	40	± 0.3	1243	± 11	379	± 1	868	± 1	245	± 3	9	11	11
Hosmer Sierra Hook Setting 2	242	66	35	± 0.0	26	± 0.0	117	± 6.4	82	± 0.1	2642	± 14	571	± 2	1820	± 2	337	± 2	24	27	29
RSL Steeper Carbon Gripper Setting 1	171	97	43	± 0.2	28	± 0.1	70	± 0.4	43	± 0.3	1619	± 2	487	± 4	846	± 4	267	± 4	11	11	11
RSL Steeper Carbon Gripper Setting 2	171	97	43	± 0.1	28	± 0.1	75	± 0.2	48	± 0.1	1848	± 7	510	± 2	992	± 2	272	± 6	14	13	14
Otto Bock hook Setting 1 (2x2 springs)	223	67	35	± 0.1	27	± 0.0	36	± 0.0	32	± 0.5	1002	± 3	482	± 5	775	± 5	353	± 6	11	11	11
Otto Bock hook Setting 2 (2x2 springs)	223	67	35	± 0.1	27	± 0.2	101	± 0.5	94	± 0.3	2752	± 6	555	± 15	2033	± 15	421	± 16	24	31	37
Hosmer Becker Imperial Hand (ungloved)[1]	367	75	49	± 0.3	30	± 0.0	65	± 0.3	63	± 0.4	2748	± 17	1710	± 9	1613	± 9	1031	± 6	10	11	10
Hosmer Sierra Hand (ungloved)	356	41 / 67[2]	25	± 0.1	16	± 0.1	75	± 1.7	52	± 0.7	1152	± 8	637	± 6	581	± 6	313	± 1	15	12	10
Hosmer SVO hand (ungloved)	310	74	38	± 0.1	30	± 0.2	121	± 0.9	81	± 0.9	2170	± 16	858	± 6	1406	± 6	536	± 10	15	12	12
RSL Steeper hand (ungloved)	328	81	46	± 0.4	34	± 0.1	120	± 1.3	48	± 0.5	1758	± 27	855	± 6	804	± 6	338	± 4	3	6	7
Otto Bock VO hand (ungloved)	326	85	56	± 0.2	41	± 0.2	105	± 0.4	61	± 0.4	2545	± 11	917	± 5	1291	± 5	451	± 4	10	10	10
Hosmer Sierra Hand (gloved)	447	38 / 62[2]	22	± 0.3	16	± 0.3	84	± 0.8	70	± 0.6	1017	± 24	583	± 11	639	± 11	358	± 9	7	11	18
Hosmer SVO hand (gloved)	400	70	36	± 0.3	32	± 0.1	138	± 4.5	104	± 0.9	2266	± 29	1082	± 9	1690	± 9	806	± 19	12	12	14
RSL Steeper hand (gloved)	444 (670)[3]	75	47	± 0.1	37	± 0.2	174	± 1.5	81	± 0.7	2271	± 25	1167	± 14	1100	± 14	512	± 9	9	7	8
Otto Bock VO hand (gloved)	395	80	53	± 0.3	40	± 0.2	146	± 1.7	79	± 0.5	2927	± 68	1240	± 9	1472	± 9	637	± 4	9	10	12

1. The Hosmer Becker Imperial Hand was only tested without a cosmetic glove. During the tests there was no cosmetic glove available. 2. Two thumb settings: small and wide.
3. The sleeve of the RSL cosmetic gloved was shortened to a length similar to that of the other cosmetic gloves. The mass of the hand with uncut glove applied, is given in parentheses.

3.4 Results

Table 3.I presents characteristics of the hands as well as an overview of all test data.

Mass

The measured hands had a mass 1.6 to 5.1 times higher than that of the hooks. The device with the lowest mass was the Hosmer Hook 5xAl with one rubber band (87 grams). The gloved Sierra hand had the highest mass (447 grams). The Becker hand was the heaviest ungloved hand device (367 grams).

Maximum opening span

The opening span of the hooks had a range from 67 mm (Otto Bock Hook) to 97 mm (RSL Steeper Carbon Gripper). The span of the gloved hands ranged from 62 mm (Hosmer Sierra Hand) to 80 mm (Otto Bock Hand). The use of a cosmetic glove reduced the opening span of the hands a few millimetres.

Maximum cable excursion

The maximum cable excursion ranged from 34 mm (Hosmer Sierra Hook) to 46 mm (Hosmer 5xAl hook) for the hooks, and from 22 mm (Hosmer Sierra Hand) to 53 mm (Otto Bock VO Hand) for the gloved hands.

Activation force

The maximum required activation forces during the 50 mm open and close test, are shown in Figure 3.4. Applying a cosmetic glove increased the activation force by 22 to 41% in this test.

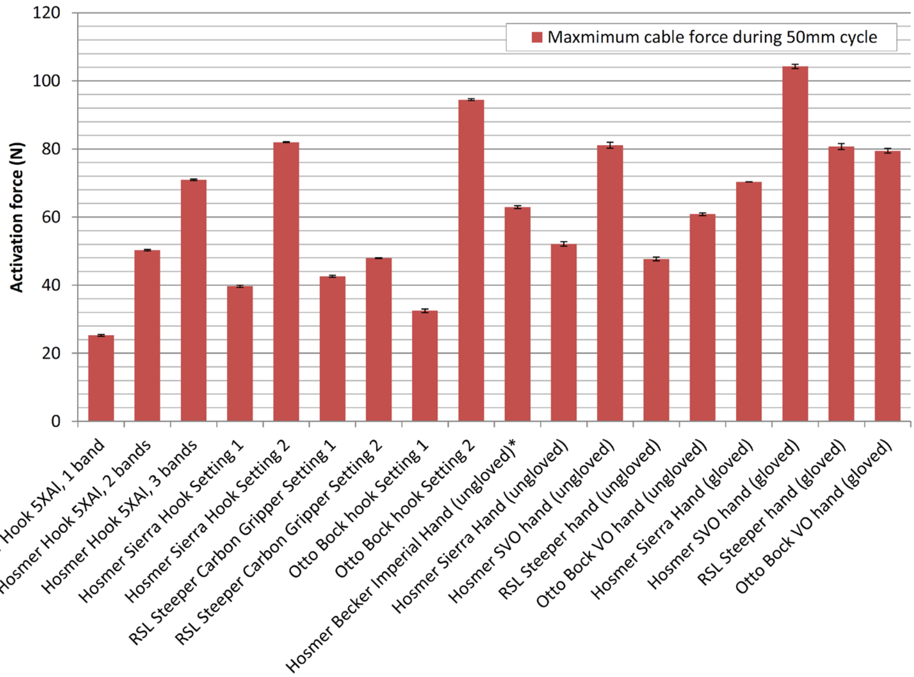


*Figure 3.4 The maximum cable activation force to open the devices 50 mm, during the second test. *The Becker hand was only tested without a cosmetic glove.*

The force displacement graphs of the fully opening and closing test are depicted in Figure 3.5 and Figure 3.6. The graphs show the different slopes of the devices, caused by the differences in stiffness. The graphs also show that the return trajectory is lower than the opening trajectory. This is caused by the mechanical friction in the mechanism, and by the internal friction in cosmetic glove and in the stiff inner glove. The Otto Bock hook (setting 1) is the only device that has a decreasing activation force characteristic (Figure 3.5). Another remarkable phenomena can be seen in the plot of the Hosmer SVO hand (Figure 3.6). The force displacement graph shows a saw tooth pattern.

Work and hysteresis

The bar chart (Figure 3.7) enables easy comparison of the work and hysteresis measured for the devices in the 50 mm open and close test. In this test the hooks dissipated 11% (Hosmer Hook 5xAl, 3 bands) to 46% (Otto Bock, setting 1) of the input energy. The ungloved hands dissipated 35% (Otto Bock VO Hand) to 64% (Hosmer Becker Imperial Hand), and the gloved hands dissipated 43% (Otto Bock VO Hand) to 56% (Hosmer Sierra Hand) of their input energy.

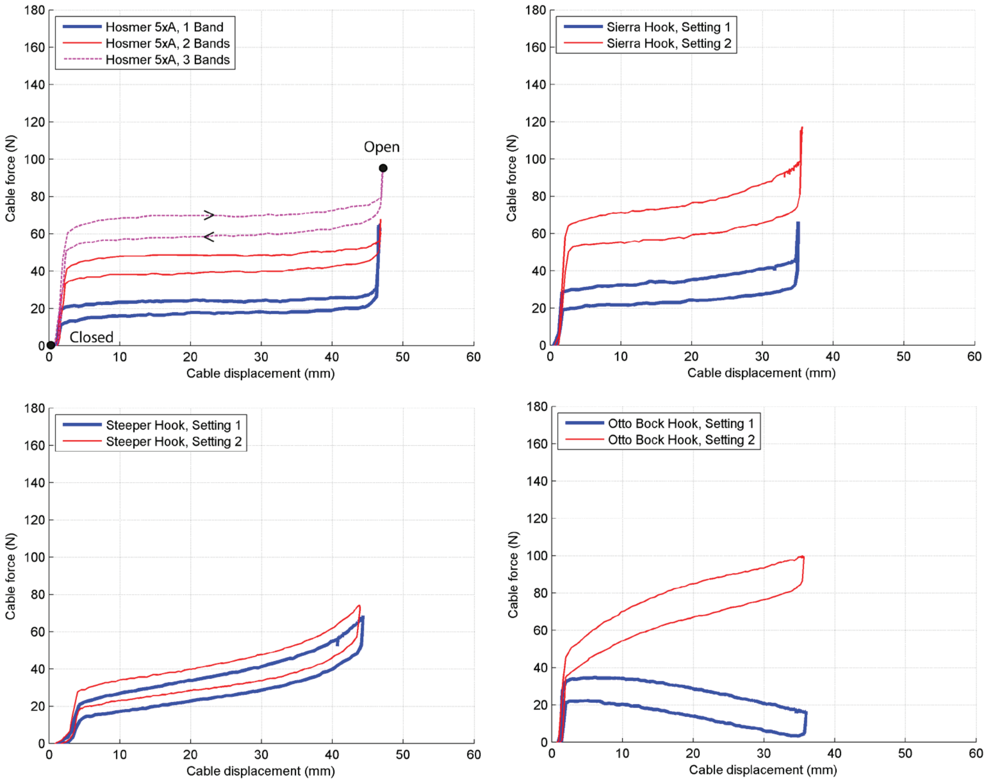


Figure 3.5 Force displacement graphs of the hook devices in their different settings, during the full opening and closing test.

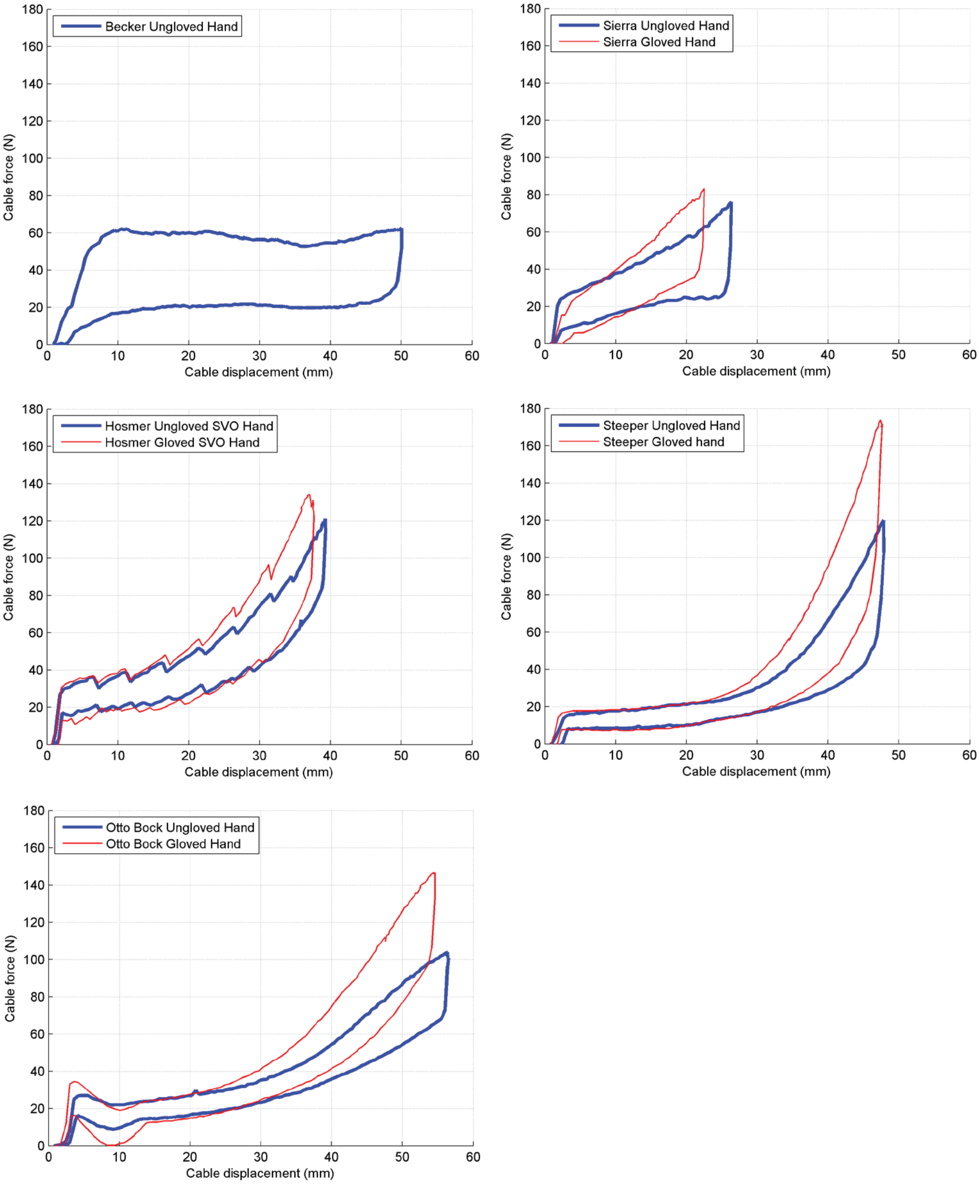


Figure 3.6 Force displacement graphs of the hand devices, with and without cosmetic glove applied, during the full opening and closing test. The Becker Imperial hand was only tested without a glove.

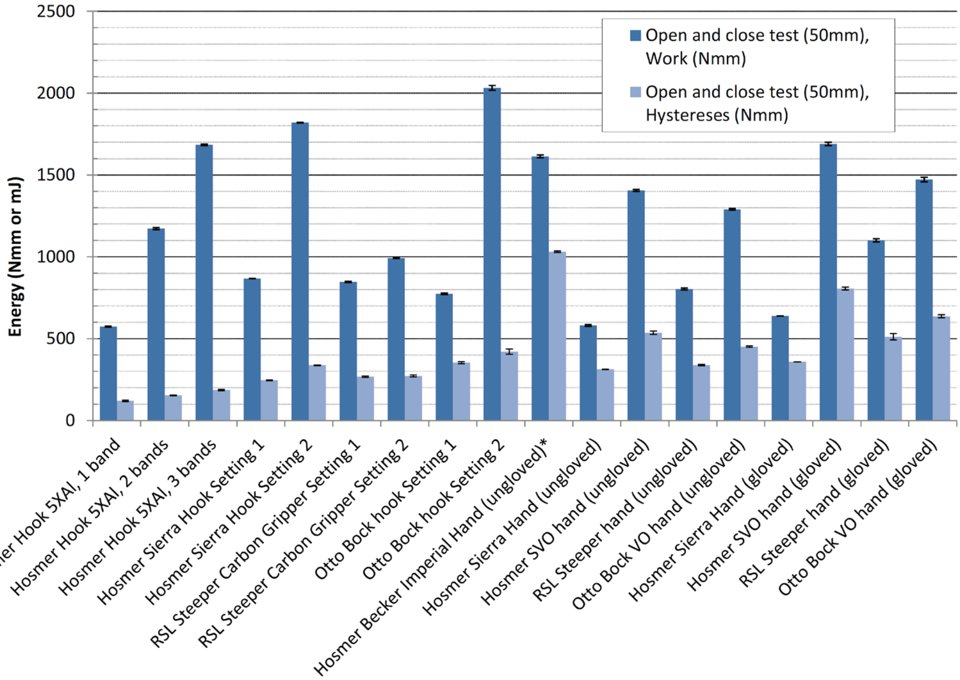


Figure 3.7 The required work to open the devices 50 mm, and the energy dissipated during one cycle of opening and closing 50 mm, during the second test.
**The Becker hand was only tested without a cosmetic glove.*

Pinch forces

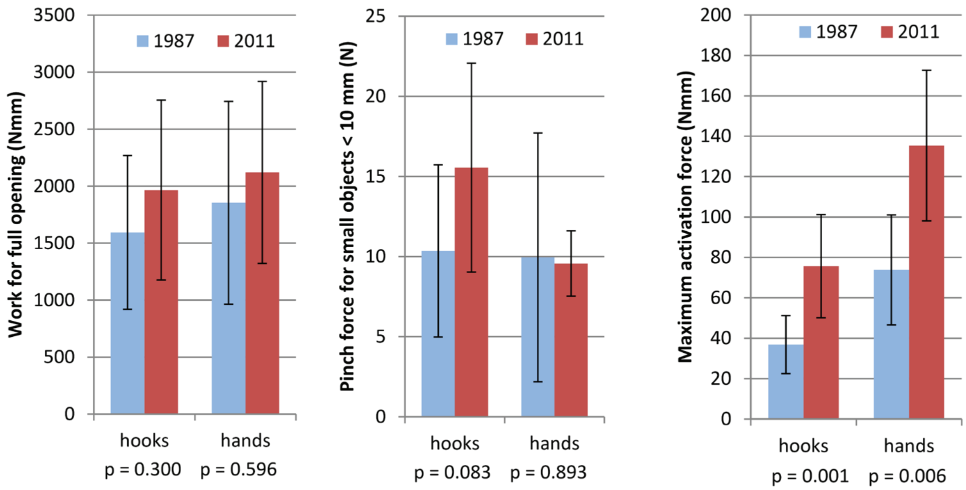
The measured pinch forces of the hooks ranged from 9 N (Hosmer 5xAl, 1 band) to 37 N (Otto Bock Hook, setting 2). The pinch forces of the gloved hands ranged from 7 N (Hosmer Sierra hand, 10 mm object, and RSL Steeper Hand, 20 mm object) to 18 N (Hosmer Sierra Hand, 30 mm object).

Best tested prosthetic devices

Of the hooks which could pinch over 20 N, the Hosmer 5xAl hook with 3 bands (24-33 N) required the lowest activation force (95 ± 4.2 N). Its cable excursion (46 ± 0.1 mm) was within the acceptable range (< 53 mm), and it had the lowest mass (92 grams). None of the hands complied with the demand of having a pinch force above 20 N. The hands which had the largest pinch force with the cosmetic glove applied, were the Hosmer Sierra Hand (7-18 N) and the Hosmer SVO Hand (12-14 N). Of all hands the Hosmer Sierra Hand required the lowest activation force (84 ± 0.8 N).

Comparison to the data from 1987

Figure 3.8 shows the mean values and the standard deviation of the devices tested in this study and of the adult devices tested by Corin *et al*. The p-values of the t-test are presented below each plot. There is only a significant difference for the activation force of the hooks and hands. All other parameters did not differ significantly ($p > 0.05$).



a. *b.* *c.*

Figure 3.8 Comparison between the values of the work (a), pinch force for small objects (b) and activation force (c) of the hooks and hands tested in 1987 and in 2011.

3.5 Discussion

The goal of this study was to objectively compare current voluntary opening prosthetic terminal devices amongst each other, by using quantitative measurements. The results were used to select the best tested hand and hook and to see whether voluntary opening devices have improved over the last decades.

Mass

The human hand has a mass of 400±90 grams [13]. The mass of a prosthetic device should be considerably lower, to enable comfortable wearing. All measured hooks were significantly lighter (87-242 grams) than a human hand. The gloved hands were of the same weight as the human hand (395-447 grams). This might be one reason that a high weight is also an issue in body powered devices, [4] and why hooks are preferred over hands. The results showed that the cosmetic glove significantly contributes to the mass of the hand. The Becker hand was the heaviest hand without cosmetic glove, and it may have also the highest mass with cosmetic glove applied.

Cable excursion

For a harness using ‘arm-flexion control’, Taylor [12] measured a maximum cable excursion of 53±10 mm. All hooks and gloved hands were within this range. The Hosmer Hook 5xAl, the gloved Otto Bock and RSL Steeper hand, and the ungloved RSL Steeper and Hosmer Becker hand, were not within the range minus one standard deviation (43 mm). Only the Hosmer Sierra hand was within the range minus two times the standard deviation (33 mm). This implies that not all users can fully open all devices.

Level of activation force

The maximum activation forces of all devices were well within the maximum activation force of shoulder control (280±24 N), measured by Taylor [12]. However, his data was measured on able-bodied subjects. A study by Shaperman *et al*. [14] showed that children with an congenital upper limb deficiency had lower strength, in both their deficient and their sound arm, than able-bodied children (~1.5-2 times lower). Furthermore, Taylor only measured maximum forces. However, the critical force is not known. The critical force is a percentage of the maximum force, which a person can exert over a certain period of time, without getting fatigued [15]. Because VO devices are usually only opened for short periods of time, it is difficult to determine the critical force. From literature, it does not become clear whether the required forces to operate a prosthetic hand during daily life activities are a problem for the user. Therefore further research on this topic is needed.

The decreasing force characteristic (Figure 3.5) of the Otto Bock hook (setting 1) is caused by the configuration of the spring in this setting. During opening the distance between the spring and the joint decreases close to zero. As a result the hook almost acts like a bi-stable mechanism, which is stable when it is closed or fully opened, but which is unstable in positions in between. Although this makes it easier to keep the hook fully opened, it may be more difficult to accurately control the opening width of the hook in between these extreme positions. The jerky behaviour of the Hosmer SVO hand (Figure 3.6) is probably caused by stick-slip behaviour of the mechanism. This behaviour could make it difficult to accurately control the hand.

Level of pinch force

A pinch force of 10 N is considered to be sufficient for most activities of children [16]. For adults the desired pinch is about two times higher and occasionally more. A study by van der Niet *et al*. [11] showed that an iLimb, which had a maximum pinch force of 15-20 N, did not exert enough force to complete all tasks. A study of Keller *et al*. [10] showed that the required pinch force is even higher for several activities (e.g. holding a tea cup: 28 N, pulling on a sock: 34 N). Therefore it is assumed that the VO devices should be able to pinch over 20 N. In their highest setting all hooks could pinch well over 20 N, except for the RSL Steeper Carbon gripper, which had a maximum pinch force of 14 N. None of the tested hands could pinch over 20 N. The maximum measured pinch force was 18 N, in the gloved Sierra Hand. All other measured hand pinch forces did not exceed 15 N. These results indicate that the produced pinch forces of the VO hands are not large enough to complete all tasks of daily living. The low pinch forces measured for the hands, and the relative high pinch forces measured for the hooks, might be an important reason of the high rejection rates of body powered hands, and the relative good acceptance for hooks.

Note that the method of the pinch force test we employed, differed from that of Corin et al. They attached the fingers of the fully opened hand to the test bench, using two vertical spanned cables. The pinch force was then measured, while the cables were slowly released. In this method the hand is only supported by the cables. Initial trails in our study showed that the measured pinch force was influenced by the mass of the device, which was not in line with the vertical cables. The outcomes were also influenced when the cables were spanned horizontally, especially for small finger openings. When the method of Corin *et al*. is used, the measured pinch forces can be higher than the actual pinch force. This effect is larger for hands than for hooks, because of their larger mass. Although these deviations are relatively small, it is more accurate to measure the pinch force directly. Therefore in our study we used a pinch force sensor, to measure the pinch force for three opening spans. Because the overestimation of the pinch force in the data of Corin *et al*. would be small, and because the standard deviations are large (Figure 3.8b), it is unlikely that this has affected the outcome of the pinch force comparison significantly.

Best performing devices

It was easy to select the best performing prosthetic hook. In general the hooks performed much better than the hands. Selecting a suitable hand was difficult, as none of the hands produced a sufficient pinch force. The use of the hands for activities of daily living is therefore expected to be limited.

Past, present, future…

It is interesting to compare the results to the data of the test performed by Corin *et al*. in 1987 [7]. In the past two decades many new materials have become available, which can be used for mechanism, bearings, gloves etc. Meanwhile various user studies clearly mapped the needs of the prosthesis user [4, 17]. However, comparisons between our results and Corin's results showed no significant difference for the required work or pinch forces for the hooks nor for the hands (Figure 3.8a, b). However, the activation force for the hands and the hooks was significantly higher in this study (Figure 3.8c), which means that they performed worse. In general it can therefore be concluded that, despite all technologic advantages in other fields, VO prosthetic devices did not improve since 1987.

This study also shows that some newer devices had a poorer performance than devices tested in 1987. For example, the RSL Steeper Carbon Gripper required a maximum activation force of 75 ± 0.2 N, and it delivered a pinch force of 14 N. The Hosmer SSS-555 hook, and the Hosmer 10P hook, tested by Corin *et al*. in 1987, required a lower activation force of 49 and 65 N respectively. They also delivered a higher pinch force of 15 and 26 N respectively. Another example is the current Otto Bock VO Hand, which required an activation force of 146 ± 1.7 N, and it delivered a pinch force of 9 N for objects of 10 mm. Its predecessor, the Otto Bock Hand tested in 1987, required an activation force of 62 N, and delivered a pinch force of 14 N for small objects.

The outcome of this comparison with the data from 1987 raises a number of questions:

- "Why have VO devices not been improved over the past decades?" This is especially interesting to know for hands, which still have an insufficient pinch force, and a high activation force that may be uncomfortable.
- "How is it possible that new devices have become available that perform worse than their predecessors?" Why do prosthesis users not benefit from apparent innovations?

Perhaps the most interesting question is:
- "What would be the real future potential of body powered prostheses, if we would invest the same effort and resources into the improvement of body powered prostheses, as is currently invested in electric devices?"

For now these questions remain unanswered. However they deserve studying. There are many opportunities for improvements in body powered prosthetics, especially for prosthetic hands:
- The mass of the hands should be reduced, to enable comfortable wearing.
- The activation force should be reduced to a comfortable level. A study should reveal the comfortable force level. Moreover, it needs to be examined what is the optimal shoulder movement to produce the most efficient force.
- The pinch force should be increased to an acceptable level (> 30 N). To enable this the mechanism should be more efficient. This could be achieved by redesign of the mechanism, and by improving the bearings of the mechanism.
- A more flexible cosmetic glove should be developed.
- The inner glove should be more flexible, or its use should be avoided.

There are indications in literature that overload of the contralateral shoulder might lead to symptoms of overuse [18]. Therefore improvement of body powered prostheses is not only desirable, it is a necessity. Prosthesis users should be offered a prosthesis that is optimized to their needs and demands.

Study limitations

Of each prosthetic device only one specimen was tested. Also for each cosmetic glove type only one glove was used. Variations in individual gloves, prostheses, and factory spring settings, might result in deviations from the measured data. These effects are expected to be the smallest in the hooks, and the largest in the hands with adjustable spring settings and a cosmetic glove. From cosmetic gloves it is known that their properties can vary, due to variations in thickness [19]. Although some variables might cause variations in the results for the gloved hands, they are expected to give only minor variations for the hooks and the ungloved hands. Testing of multiple devices of individual types, with multiple gloves might give a better insight of this variation. A second limitation in this study was the low speed in which the devices were activated. A higher activation speed might slightly increase the activation forces and hysteresis, due to viscous behaviour.

3.6 Conclusions

Nine voluntary opening prosthetic devices, four hooks and five hands, were quantitatively tested. All hooks weighed less than a human hand. The masses of the gloved hands were similar to that of the human hand. All cable excursions were within the average movement range for shoulder activation. The hooks required a lower activation force and work than the hands. Nearly all hooks could pinch over

20 N. The hands required a high activation force, and could not pinch over 20 N. Their use in daily life is expected to be limited. The Hosmer 5xAl Hook with 3 bands was the best tested hook. The Hosmer Sierra Hand was the best tested hand.
Comparison to data of Corin *et al*. showed that VO prosthetic devices have not improved since 1987. Some newer devices even performed worse than the devices tested in 1987. The results of this study are helpful in selecting the right prosthetic device for a patient, and in improving current devices. Future research should focus on: reduction of the mass of cosmetic glove and hand mechanism, determining the comfortable activation force level for shoulder activation, decreasing the required activation force level, increasing the pinch force of the hands.

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4

Design and Evaluation of Two Different Finger Concepts for a Body Powered Prosthetic hand

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Abstract

The goal of this study was to find an efficient way of energy transmission for application in an anthropomorphic underactuated body powered prosthetic hand. A pulley-cable driven finger and a hydraulic cylinder driven finger were designed and tested to compare the pulley-cable transmission principle to the hydraulic cylinder transmission. Both fingers had identical dimensions and had a low mass. The only thing that differed between the fingers was the transmission principle. The input energy was measured for a number of tasks. The pulley finger required more input energy than the cylinder finger to perform the tasks. This was especially the case in tasks which required high pinch forces. The hydraulic transmission is therefore the more efficient transmission for application in body powered prosthetic fingers.

4.1 Introduction

For many applications of artificial hands in the field of robotics and prosthetics, it is desirable to have a low hand mass. In the field of prosthetics a high hand mass is a major cause of the rejection of a prosthetic hand by the user [1]. Artificial hands are often heavy, due to the fact that they have multiple motors placed inside the hand [2, 3]. Commercial available articulating prosthetic hands, like the iLimb [4] and the bebionic [5], use one electric motor for each finger. The number of actuators can be reduced by using the principle of underactuation. An underactuated mechanism has by definition more degrees of freedom than actuators [6, 7]. The configuration of such a mechanism depends not only on the actuator force, but also on the external forces acting on the mechanism, e.g. the force acting on the fingers. As an amputee usually has only one control signal available, just one actuator would be enough to control all finger joints of the entire hand. Using only one actuator could drastically reduce the hand mass. Instead of using an actuator it is also possible to have the amputee mechanically controlling the hand, e.g. by means of a shoulder harness. In such a body powered prosthesis no electric motor is needed, reducing the mass even further.

4.2 Problem

The problem with current body powered hands is that they require a large amount of input energy by the prosthesis user in order to produce a limited pinch force at the finger tip. A body powered hand requires up to 2292 Nmm of energy to pinch 15 N, at an user effort that is uncomfortably high [8]. User needs of hand amputees include: a higher pinch force [9], a lower activation effort [8] and a lower hand mass [10]. To achieve these goals an efficient energy transmission is required. The input energy can be transmitted to the fingertip through e.g. a cable or a hydraulic transmission. The input energy is required through the compliancy of the components (segments, transmission) and by the friction in the joints, resulting in actuator displacement and force while the contact point of the fingertip itself does not move. The low ratio of the actuator lever over the pinch force lever (the total finger length) results in very high actuator forces. This unfavourable lever ratio directly results from the anthropomorphic dimensions. The problem can be illustrated by an example . To produce a pinch force of 30 N at a total finger length of 67 mm (l_{total}), a joint torque of 2010 Nmm (M_{MCP}) is required in the MCP-joint (Figure 4.1). When the activation lever in the MCP-joint has a length of 5 mm, a very high activation force of 402 N is required.

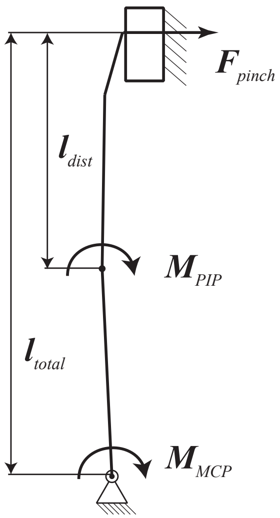


Figure 4.1 High activation joint torques are required to produce a pinch force of 30 N. When the total finger length (l_{total}) is 67 mm and the distal finger length (ldist) is 37 mm, the joint torque has to be 2010 Nmm in the MCP-joint (M_{MCP}), and 1110 Nmm in the PIP-joint (M_{PIP}).

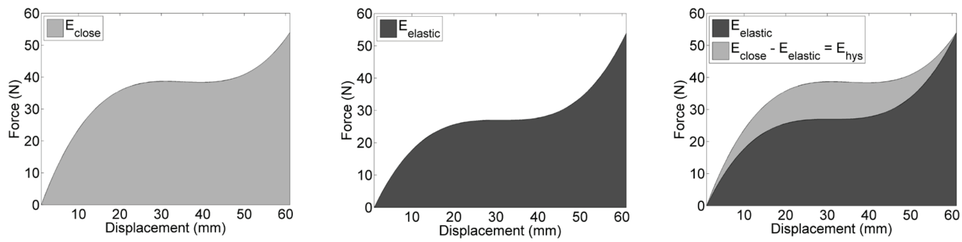
Ideally the pinched object and the finger parts would be totally rigid. The displacement of the actuator during pinching would then be zero. Hence, the input energy or work (force times displacement, eq. 4.1) would then be zero, so the actuator does not have to produce input energy (~work). However, in practice the pinched object and the finger parts will be compliant. As a result of the high input force, the object and the finger parts will start to elastically deform. To build up a pinch force the actuator has to produce an input displacement, even when the contact point is not moving. This will result in a considerable work by the prosthesis user:

$$W = \int_0^{\ell} F(x) \cdot dx \qquad (4.1)$$

Because of the high activation force (F(x)), even a small input displacement (dx) will require a large amount of energy (W). The finger parts act like springs, which store large amounts of input energy. The actuator has to deliver this input energy. Although the stored 'elastic energy' is returned during re-opening ($E_{elastic}$ in Figure 4.2), it is not useful to the user of the body-powered prosthesis. This is the first cause of energy loss. A second cause is located in friction in the finger joints. Due to the deflection of the finger parts, there will be small joint rotations. The high activation force causes high joint loads. The rotational friction in the joint will therefore also be high, which results in a considerable energy dissipation even at a small joint rotations (E_{hys} in Figure 4.2). This energy dissipation due to friction occurs during the closing motion and during the opening motion. The actuator also has to deliver this dissipated energy. Therefore to design an efficient mechanism that requires a low amount of input energy (E_{close} in Figure 4.2), such a mechanism should have a high stiffness (for a low $E_{elastic}$, eq. 4.2) and should have a low energy dissipation (for a low E_{hys}).

$$E_{close} = E_{elastic} + E_{hys} \qquad (4.2)$$

Two recent studies showed that all seven tested body powered prosthetic hands, with one degree of freedom, require much input energy which is already problematic [8, 11]. Adding more degrees of freedom will increase the number of joints, which will reduce the efficiency even further. When we want a to achieve a firm grip, with an articulating underactuated hand, using only a small amount of input energy, we need an efficient energy transmission between actuator and fingertip.



a. b. c.

Figure 4.2 Work can be represented by the area below the force-path-curve. The hysteresis or energy dissipated by the finger (E_{hys} , figure c) is the difference between the work done on the finger during closing (E_{close}, figure a) and the elastic energy returned by the finger during re-opening ($E_{elastic}$, figure b), adapted from [8].

4.3 Goal

The goal of this study is to find an efficient way of energy transmission, to enable underactuated articulating finger movement in an anthropomorphic body powered prosthetic hand. The mechanism should have a low mass and should be able to deliver a requested pinch force, with only a small amount of input energy. The pulley cable transmission principle will be compared to the hydraulic cylinder transmission principle, in order to select the most efficient principle.

4.4 Methods

In this study two different ways of energy transmission will be compared: the pulley cable transmission, and the hydraulic cylinder transmission. Both transmission principles will be briefly explained, as well as their advantages and disadvantages.

Pulley-cable transmission

A common way of energy transmission is the use of pulleys and cables. This principle has been used in various hand prototypes to achieve underactuation [12, 13]. Figure 4.3 (left) shows the schematic overview of a pulley cable transmission in a finger with two degrees of freedom, that will be used in this study. The pulley at the MCP-joint can move independently from the proximal phalanx. The pulley at the PIP-joint is attached to the distal phalanx. When the cable is pulled, a torque will be applied to the MCP- and the PIP-joint. The torques are independent of the configuration of the finger. The magnitude of both torques depends on the ratio of the PIP pulley diameter over the MCP pulley diameter. The pulley cable transmission has several possible advantages and disadvantages compared to a hydraulic cylinder transmission.

Possible advantages:
Very lightweight
No strict dimension tolerances

Possible disadvantages:
Cable wear and tear
Cable elasticity
Cable run off

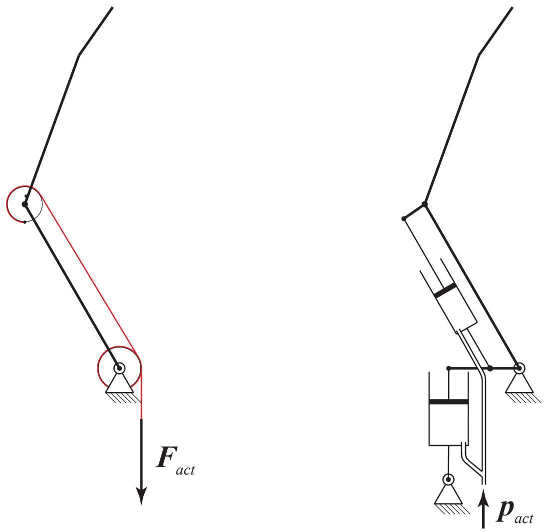


Figure 4.3 Schematic drawings of both transmission principles: the pulley finger (left) and the cylinder finger (right). The MCP-pulley can move independently, the PIP-pulley is attached to the distal phalanx. The distal cylinder of the cylinder finger is located inside the finger. The proximal cylinder is located in the palm of the hand.

Hydraulic cylinder transmission

Another common way of energy transmission is hydraulics. Although this principle has been used for decades in many fields, there are only a few examples in the field of hand prosthetics [14-18]. There are various actuators which can be used for hydraulic transmission, e.g. metal bellows, cylinders with o-ring sealing, rolling diaphragms, or McKibben muscles. In this study hydraulic cylinders with o-ring sealing were used. Cylinders can withstand higher pressures (>10 MPa) [19] than other hydraulic actuators (varying from 0.8 MPa for McKibben muscles [20] to 6.9 MPa for rolling diaphragms [21]). As a result cylinders offer the highest force, given a limited cross sectional area. Figure 4.3 (right) shows the schematic overview of the hydraulic cylinder transmission in a finger with two degrees of freedom, that will be used in this study. A master cylinder, or a hydraulic pump, can pump fluid into the inlet tube. This will increase the activation pressure (pact). Due to the increasing activation pressure, the cylinders will start to apply a torque around the MCP and PIP joint. The magnitude of both torques is dependent on the effective cross sectional area of the cylinders, the lever length, and the lever orientation.

Possible advantages:	**Possible disadvantages:**
Efficiency independent of hose curvature	Risk of leakage
Flexible to install	Bulkier than pulley cable transmission
High system stiffness	Strict dimension tolerances
	Sealing friction
	Large hoses
	Stiffening up of hoses at high pressures

Tested fingers

Two fingers were designed to compare the transmission principle of a pulley-cable transmission (Figure 4.3, left) and of the hydraulic cylinder transmission (Figure 4.3, right). Except for the transmission principle, all parameters of both fingers were identical (e.g. dimensions, axis diameters, bearings). The fingers had to comply with the following demands:

- Be able to pinch 30 N, to handle a broad range of objects [22].
- Torque ratio between MCP and PIP joint should be around 0.5, to enable a stable pinch grip [23].
- Anthropomorphic dimensions (fit inside a finger of a cosmetic glove, size 7¾)
- Maximum mass 25 gram (so four max. mass of four fingers is 100 gram)
- MCP-joint range of 0-90° (natural range of motion)
- PIP-joint range of 0-90° (natural range of motion)
- DIP-joint fixed at 15° (this angle is used in arthrodesis of the DIP-joint)[24]

Designed fingers

Two fingers were designed, a pulley cable finger (Figure 4.4, left) and a hydraulic cylinder finger (Figure 4.4, right). The fingers were identical, except for the way of transmission (Table 4.I). Both fingers had the same dimensions. Each transmission principle was optimized for the test, in such a way that its required input energy, energy dissipation and mass were all as low as possible. The fingers had to produce high joint torques, e.g. pinching 30 N with a stretched finger (Figure 4.1) requires a joint torque of 2010 Nmm around the MCP-joint and a torque of 1110 Nmm around the PIP-joint. This imposed a challenge to the design of the fingers.

For the pulley finger a 1 mm thick cable made of steel was selected. The steel cable has a high stiffness, as steel has a high elastic modulus (~200 GPa). The stiffness of a cable is dependent on the material stiffness and on the constructional stiffness, which is dependent on the plait or braid of the cable [25]. Other commonly used high strength cable materials all have a 2 to 4 times lower modulus of elasticity than steel, e.g. Vectran (52-103 GPa), Aramid (70-110 GPa) and Spectra/Dyneema (120 GPa). The tension force in the cable of the pulley finger had to be minimized, to reduce elastic behaviour of the cable. A lower cable force will also reduce the bearing load

and thus the bearing friction. To minimize the cable force, the diameters of the pulleys were maximized. The diameter of the proximal pulley was set at 10 mm, which was the maximal diameter that would fit inside an anthropomorphic finger together with the finger frame. To match the transmission ratio of the cylinder finger, the distal pulley diameter had to be 0.55 times the proximal pulley. Given the proximal pulley diameter (d_2) and a cable thickness of 1 mm (t_c), the distal pulley diameter (d_1) was set at 5 mm $((d_1+t_c)/(d_2+t_c)=(2.5+0.5)/(5+0.5)=0.55)$. The finger had a total mass of only 15 grams.
The piston diameters of the hydraulic finger were maximized, to enable the high activation forces. First the distal cylinder was maximized, as there was only limited space to fit the distal cylinder inside the finger. Its diameter was set at 7 mm, which was the largest cylinder diameter that would fit inside the anthropomorphic finger alongside the finger frame. The distal cylinder had a moment arm of 5 mm. There was more space for the proximal cylinder, as it was placed inside the palm of the hand. Its diameter was set at 8 mm. Together with a moment arm of 7 mm, this yields a transmission ratio of 0.55, compared to the distal cylinder. The finger had a mass of 25 grams. This is 10 grams more than the mass of pulley finger. The difference in mass is only 2% of the total mass of an average prosthetic hand (which has a mass of 450 grams). Therefore both fingers can be considered as very lightweight fingers. Flexible nylon hoses (3x1.8 mm) were used, which can stand a pressure of up to 6 MPa. As a hydraulic fluid water was used instead of oil, to limit the consequences in case of a system failure.
Both fingers were unidirectionally activated. After activation they were returned to their initial position by helical springs. The same springs in the same configuration were used for both fingers. Plain bearings made of PCTFE were used. The diameter of the axis in the fingers were minimized to 1.5 mm, to reduce the joint friction inside the bearings.

Table 4.I Specifications of both finger prototypes

	Pulley finger	Cylinder finger
Max. joint angles MCP	90°	90°
Max. joint angles PIP	90°	90°
PIP/MCP-ratio	0.55	0.55
Total finger length (L)	67 mm	67 mm
Length Proximal Phalanx (L_1)	30 mm	30 mm
Total length Middle and Distal Phalanx (L_2)	37 mm	37 mm
Diameter axes	1.5 mm	1.5 mm
Total finger mass	15 gram	25 gram

Figure 4.4 Two fingers with identical parameters were made: the pulley finger (left) and the cylinder finger (right). The only parameter that differed was the transmission principle. The same springs were used in an identical configuration in each finger, to extend the finger.

Test protocol

The fingers were compared using an energy based approach. The required activation energy and the energy dissipation were measured while the fingers had to perform the following tasks:

1. Pinch 30 N with a stretched finger (Figure 4.5, left).
2. Close the finger 90°, pinch 0 N, and reopen the finger (Figure 4.5, middle).
3. Close the finger 90°, pinch 30 N, and reopen the finger (Figure 4.5, middle).
4. Close the finger 180°, pinch 0 N, and reopen the finger (Figure 4.5, right).
5. Close the finger 180°, pinch 30 N, and reopen the finger (Figure 4.5, right).

To simulate the effect of a cosmetic glove on a prosthetic finger, the tests were repeated with a PVC cosmetic glove placed over the finger. All tests were performed 4 times for each finger type, to obtain average values. The energy transmission which requires the lowest amount of energy, was selected as the most efficient transmission.

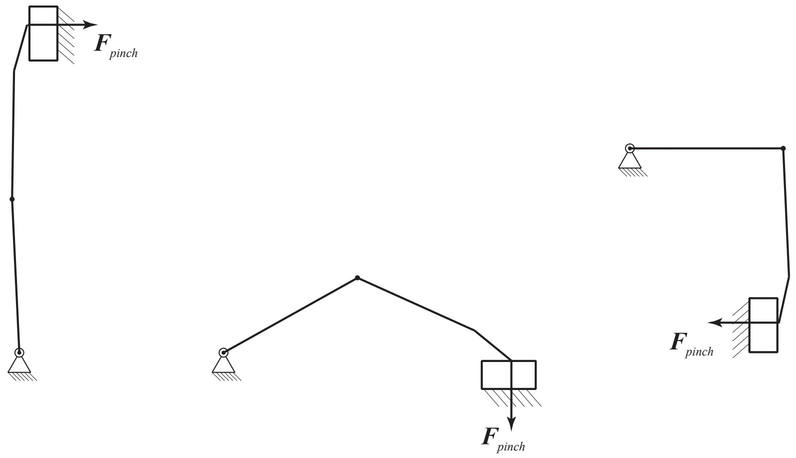


Figure 4.5 Schematic overview of the tests. Pinch 30 N in stretched position (left), close 90° and pinch 0 or 30 N (middle), close 180° and pinch 0 or 30 N (right).

A manual operated test bench (Figure 4.6) was used to actuate the fingers. The pulley cable finger was actuated by pulling the actuation cable. The hydraulic cylinder finger was actuated by pulling a master cylinder (d_{piston}=10 mm). A load cell (Zemic, B3G-C3-50kg-6B) measured the force acting on the cable or the master cylinder piston. A LVDT (Positek P101.200CL100) measured the displacement of the actuation cable or the piston of the master cylinder. The pinch force was measured using a custom build pinch strain gauge load cell. The required work and hysteresis were obtained from the measured actuation forces and displacements.

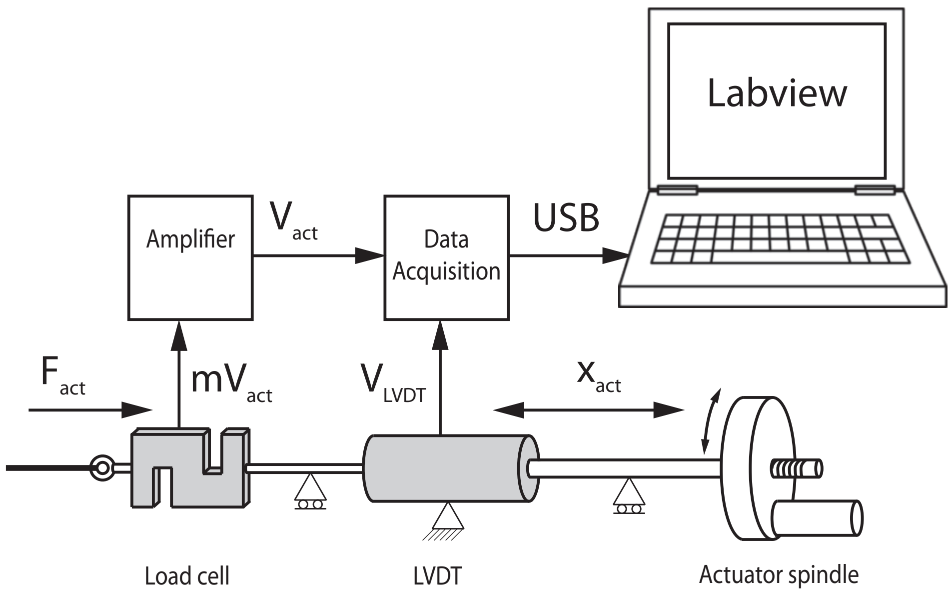


Figure 4.6 Schematic overview of the manual test bench, which was used to apply the load to the cable and the master cylinder. A load can be applied by turning the actuator spindle. The load cell measures the actuation force, the LVDT measures the displacement (adapted from [11]).

4.5 Results

Figure 4.7 shows an example of the raw data of a measurement on the pulley finger without a glove, bending 90° and pinching 30 N. The arrows mark the subsequential steps: closing 90° (1), increase pinch force up to 30 N (2), unload the finger (3), stretch the finger (4). The input energy (E_{close}) is the area below line 1 and 2. The returned elastic energy ($E_{elastic}$) is the area below 3 an 4. The dissipated energy (E_{hys}) is the area enclosed by 1, 2, 3 and 4. Table 4.II presents the measured work and hysteresis values of all tests.

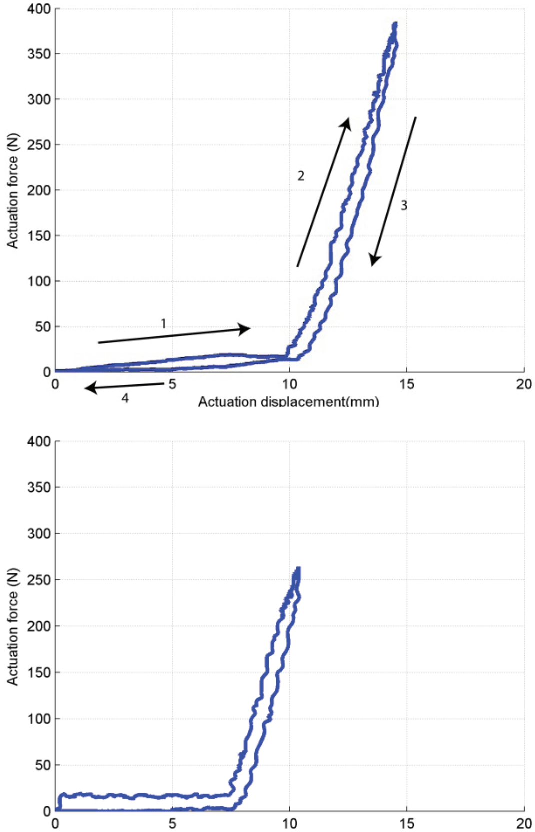


Figure 4.7 Example of the raw data of the pulley finger (left) and the cylinder finger (right), 90° finger flexion, pinching 30 N, without glove. The steps are marked by arrows: closing 90° (1), increase pinch force up to 30 N (2), unload the finger (3), stretch the finger (4). In this test the pulley finger requires both a larger actuation displacement as well as a higher actuation force.

Table 4.II Required work and dissipated energy for the different tasks.

Finger	Angle (°)	Pinch force (N)	Glove	Work (Nmm)		Hysteresis (Nmm)	
				AVE	STD	AVE	STD
Pulley	0	30	frame	858	15	214	15
Pulley	0	30	frame + glove	1148	95	431	70
Pulley	90	0	frame	111	1	58	1
Pulley	90	0	frame + glove	127	10	47	9
Pulley	90	30	frame	954	15	243	8
Pulley	90	30	frame + glove	1009	26	253	13
Pulley	180	0	frame	245	3	150	4
Pulley	180	0	frame + glove	486	45	288	39
Pulley	180	30	frame	1110	14	303	3
Pulley	180	30	frame + glove	1214	57	359	48
Cylinder	0	30	frame	708	8	211	9
Cylinder	0	30	frame + glove	761	7	219	5
Cylinder	90	0	frame	113	2	106	3
Cylinder	90	0	frame + glove	125	4	115	4
Cylinder	90	30	frame	549	14	251	7
Cylinder	90	30	frame + glove	782	20	323	23
Cylinder	180	0	frame	218	2	188	2
Cylinder	180	0	frame + glove	360	19	284	20
Cylinder	180	30	frame	990	19	439	16
Cylinder	180	30	frame + glove	1111	35	513	28

Work

Figure 4.8 shows the input energy or total work (E_{close}) for both fingers for every performed test. The required amount of work ranged from 111 to 1214 Nmm for the pulley finger, and from 113 to 1111 Nmm for the cylinder finger. When the cosmetic glove was applied, both fingers required more input energy than without glove. For tasks which involved only moving, the pulley finger required up to 35% more input energy than the cylinder finger. For pinching tasks it required up to 74% more energy than the cylinder finger.

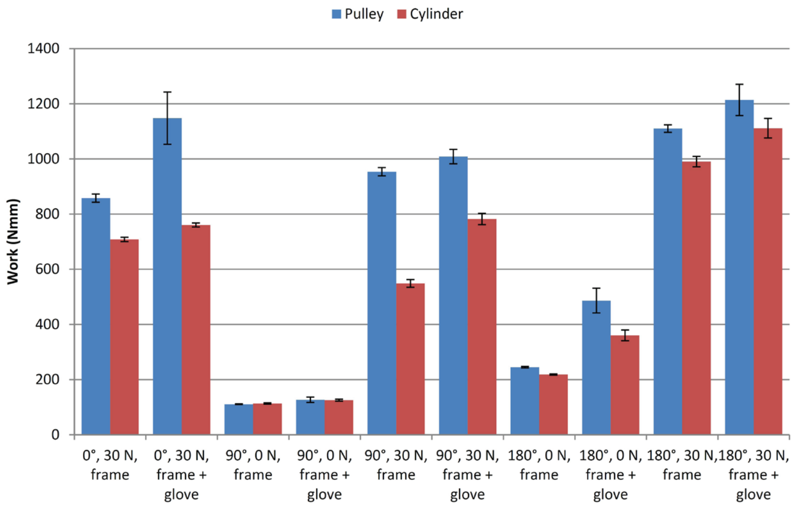


Figure 4.8 The work, or required input energy (E_{close}), to operate the fingers during the different tasks. The pulley finger required more input energy (or work) than the cylinder finger to perform the same tasks.

Hysteresis

Figure 4.9 shows the dissipated energy or hysteresis (E_{hys}) for both fingers for every performed test. The pulley finger dissipated 47 to 431 Nmm. The cylinder finger dissipated 106 to 513 Nmm. When the cosmetic glove was applied, both fingers dissipated more input energy than without glove. The cylinder finger dissipated up to 51% more energy when the glove was applied. During pinching up to 5.4 times more energy was dissipated in the pulley finger and up to 2.8 times in the cylinder finger, compared to tasks without pinching.

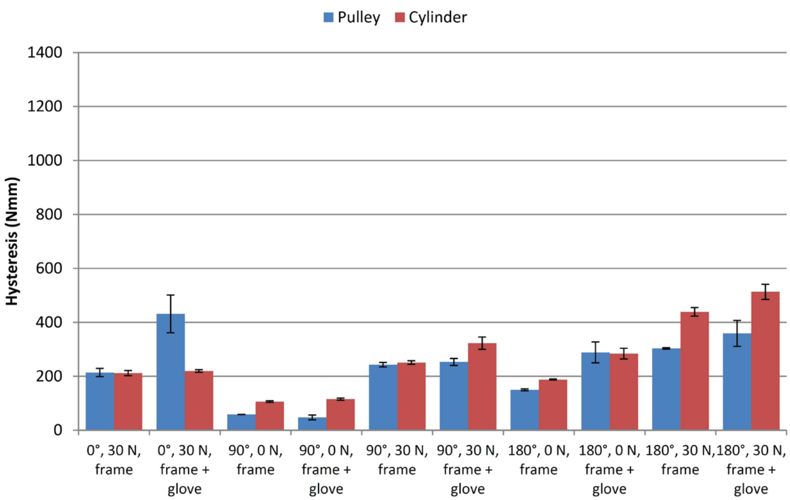


Figure 4.9 The hysteresis, or dissipated energy (E_{hys}), to operate the fingers during the different tasks.

4.6 Discussion

Based on their dimensions and mass, both fabricated fingers are suitable for application in a anthropomorphic hand prosthesis. The fingers have an anthropomorphic range of motion and they are capable of pinching 30 N, which enables a sufficient range of activities and tasks. The mass of each finger (25 gram or less) is only 3-6% of the mass of a current prosthetic hand, which enables the design of a lightweight body powered hand.

Friction and hysteresis

The energy that was dissipated during the tests, is dissipated by multiple components:

- In the first place there is the bearing friction, which increases when the bearing loads increase. To get an idea of the amount of energy dissipation by one bearing, we can estimate the friction in the bearing of the pulley of the MCP joint during 90° finger flexion and pinching of 30 N without glove (Figure 4.7). The joint friction is expected to be the highest when the finger starts pinching. During pinching the cable force increases linearly (from ~20 to ~380 N), so the average cable force is about 200 N. When we assume that the normal force in the bearing of the MCP pulley is equal to the cable force and the friction coefficient of PCTFE is 0.35 [26], the tangential friction force in the bearing (d=1.5 mm) will be 70 N. The tangential force at the pulley diameter will be 0.75 mm/5.5 mm*70 N=9.55 N. As the measured cable displacement during pinching is 5 mm, the bearing will dissipate 5 mm*9.55 N=48 Nmm during closing and the same amount during opening, which yields a total energy dissipation of 95 Nmm per cycle for the bearing of the MCP pulley. Beside the friction in the MCP pulley there will also be friction in the MCP- and PIP-joint. Doubling the pulley diameter would halve the bearing friction. It is however not possible to increase the pulley diameter, as then the pulley finger would not fit inside the cosmetic glove.
- In the second place there is energy dissipation in all components that have viscous behaviour, like the cosmetic glove [27]. Therefore the dissipated energy of both fingers is expected to be lower during movements without pinching or without a counteracting cosmetic glove. This was confirmed by the results (Figure 4.9). When the glove was applied the fingers required more input energy. In the cylinder finger an extra cause of extra energy dissipation is the sealing friction, which is caused by the o-ring sliding along the cylinder wall [19]. This friction is always present, even when the piston is moving without external loads acting on the finger. This is a disadvantage compared to the pulley finger. The effect could be clearly seen when the fingers were bend 90° without pinching (Figure 4.9). In this unloaded test the cylinder finger dissipated significant more energy than the pulley finger, due to the o-ring friction. After the cosmetic glove is applied the pulley finger shows a remarkable increase of the hysteresis in the stretched configuration. This can possibly be explained by contact friction between the glove and the mechanism in this configuration.

System stiffness and required work

For unloaded movements the required work is mainly dependent on the helical springs, the elastic deformation of the glove, if present, and the energy dissipation as described above. As soon as the finger has to deliver a pinch force, the forces acting on the finger will increase and various components of the finger start to elastically deform and start acting like springs. This is clearly shown in the test in which the stretched fingers pinch 30 N. Although the helical springs are not extended during this test, the fingers still store elastic energy. The amount of deformation of components like frame, cable or hose, are dependent on their stiffness. The individual stiffnesses of all the components together, determine the total stiffness of the system. The stiffer the system is, the less extra energy is required to deliver the demanded pinch force. Although the components return their elastic energy when turning into their original shape, the returned energy is not useful anymore as it is not possible to return the energy back to the user in a useful way. A system with a low stiffness will therefore have a higher energy demand, which will result in extra physical effort for users of body powered prosthetic hands. Elastic deformation is therefore undesirable. It should be avoided by making the system stiffness as high as possible.

The results show that the pulley finger requires more energy in tasks which involve pinching, even in tasks in which its hysteresis is lower. The pulley cable finger stores more elastic energy during one cycle, which means that it has a lower stiffness than the cylinder finger. A cause of its lower stiffness lies in the elastic behaviour of the cable, which elastically elongates when the activation force increases. This effect could be reduced by using a thicker cable, or two cables, instead of one. However this is not possible in the designed finger, as the limits were already reached during the optimisation. Increasing the cable diameter, at a constant pulley radius, will increase the stress in the outer cable filaments, which will result in cable failure. Also the pulley width is too small for a thicker cable, or a double cable. It is also not possible to use a stiffer cable of a material with a much higher Young's Modulus than the current steel cable, as alternative cable materials, like Vectran, Aramid and Spectra/Dyneema all have a 2 to 4 times lower elastic modulus. Therefore cables made of materials other than steel, will be less stiff and will require even more energy input. A second cause for the larger elastic deformation of the pulley finger is the high cable force that exerts an external force to the finger frame, causing small elastic deformations of the frame. The hydraulic hose of the cylinder finger does not impose an external force to the finger frame of the hydraulic finger. Less elastic deformation takes place in the frame of the cylinder finger, as there only act internal forces on the frame of the cylinder finger.

Future clinical implications

The fingers in this study were both able to produce a pinch force of 30 N. This is a high pinch force for an articulating underactuated finger and it is a 1.7 to 4.3 times higher pinch force than current body powered hands [8, 11]. With two fingers this would enable a tripod grip of even 60 N. The hydraulic hose of the hydraulic finger allows for an increase in pressure and in pinch force. The pinch force of the pulley finger cannot be increased, due to the limited strength of the actuation cable. Both fingers are lightweight, having a mass of only 3-6% of the total mass of an average prosthetic hand. When applied in a body powered prosthetic hand, no electric actuators are needed, which allows for a low total hand mass. The presented finger concepts enable the construction of an articulating hand that is lighter, can pinch harder and have a higher energy efficiency than current BP hands, thereby meeting the most important user demands. This enables a breakthrough in the development of BP powered hands, in a field that has not changed significantly in the past decades [11]. Of the two tested principles the finger with the cylinder transmission required the least input energy, and was therefore selected to be the most efficient transmission. An extra benefit of the cylinder transmission is that it can also replace the Bowden cable transmission between the hand and the user [28].

Study strengths and limitations

To the authors' knowledge this study is the first one which makes a quantitative comparison between a pulley transmission and a hydraulic transmission in anthropomorphic finger design. The focus on energy is important to enable the development of efficient and lightweight prosthetic body powered hand. Because the dimensions are the same for both fingers, and because the fingers were optimized for the tasks, the study shows a fair comparison between both transmission principles. A limitation of this study is that it focuses on just one finger, while most hands have multiple fingers. Although this is a limitation, it is unlikely that the difference in energy requirement will decrease between both principles when multiple fingers are added. The difference might even increase when extra pulleys and cables are introduced, to enable underactuation among the pulley fingers. To enable underactuation among cylinder fingers, a simple manifold can be used, which will not introduce major energy losses. The designed fingers can also be used in robotic hands, as the constrains to a robotic hand are usually less tight than the constraints in for prosthetic application.

4.7 Conclusions

Two finger prototypes were designed and constructed, to quantitatively compare a pulley cable transmission to a hydraulic transmission. The fingers were optimized for application in a finger of a cosmetic glove of a prosthetic hand. The fingers have identical dimensions and their mass is only 3-6% of the total mass of a current prosthetic hand. The pulley finger required up to 35% more energy than the cylinder finger for tasks which required only joint movement without pinching. Also it required up to 74% more energy for moving and pinching 30 N for various configurations. The test showed that the cylinder finger required the lowest amount of input energy to perform identical tasks, as it had the highest system stiffness. Both fingers enable the construction of an articulating BP hand that is lighter, can pinch harder and has a higher energy efficiency than current BP hands. Of both concepts the cylinder finger is the most suitable for application in a prosthetic hand, as it has a higher energy efficiency than the pulley finger.

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5

Comparison of mechanical properties of silicone and PVC cosmetic gloves for articulating hand prostheses

In press:

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Abstract

Current articulating electric hands and body powered hands have a low pinch force (15-34 N), when compared to electric hands with stiff fingers (55-100 N). The cosmetic glove, which covers a hand prosthesis, has a negative effect on the mechanical efficiency of a prosthesis. The goal of this study is to mechanically compare a PVC and a silicone cosmetic glove, and to quantify, during joint articulation, the stiffness of the finger joints, the required actuation energy and the energy dissipation. Three pair of cosmetic gloves, identical of size but made from different materials, were mechanically tested: three PVC gloves and three silicone gloves.

The silicone gloves required less work and dissipated less energy during flexing. They also had a lower joint stiffness and required a lower maximum joint torque. Based on energy requirements, joint stiffness, and required joint torque, the tested silicone glove is most suitable for application on an articulating hand prosthesis.

5.1 Introduction

Problems of current hand prostheses

Many users of hand prostheses are dissatisfied with various aspects of their prosthesis [1, 2]. Rejection rates of hand prostheses are high (20-40%) [3]. Prostheses should meet the basic user demands, which can be summarized by the words cosmesis, comfort, and control [4]. In practice however, hand prostheses do not meet all demands simultaneously. Body powered hand prostheses require the user to deliver an uncomfortable high activation force (60-130 N), for producing only a small pinch force of 15 N [5, 6]. Also electric hand prostheses with articulating fingers produce a relatively low pinch force of 15-34 N [7-9]. The cosmetic glove counteracts closing of the hand. Therefore it reduces the pinch force in articulating electric hands, and in voluntary closing body powered hands [5, 10].

Cosmetic gloves

The main function of the cosmetic glove is to cover the hand mechanism and give the hand prosthesis a natural and cosmetically pleasing appearance. As an additional benefit the glove protects the mechanism against moisture and dirt. Currently two types of cosmetic gloves are available: the PVC (polyvinylchloride) glove, and the silicone glove [11].

The PVC glove is relatively durable. It has a higher resistance to mechanical damage (e.g. puncture, tearing, abrasion) than the silicone glove. However it is also relatively stiff and it gets easily stained [11, 12]. The sensitivity to staining can be reduced by treating the PVC glove with a special surface coating. Plasticisers inside the PVC keep the glove flexible. However in the longer term (e.g. long storage) the plasticisers migrate out of the material. The PVC will degrade, and will become stiffer and brittle.

The silicone glove is more flexible. It is less susceptible to light radiation (visible and UV) and heat. However it gets easily mechanically damaged, and is less durable than the PVC glove [11, 13]. Therefore it needs to be replaced more often. Because the material and production costs of a silicone glove are also higher, the replacement costs are higher than that of a PVC glove. Silicone gloves have a higher surface friction [11], which is a benefit when holding objects, but it is a drawback during dressing when the silicone sticks inside the sleeves.

From a mechanical point of view a cosmetic glove has undesirable properties. The glove imposes parasitic forces to the mechanism, due to the stiffness of the glove material. The glove also dissipates energy, due to the internal hysteresis of the glove material [10]. As a result the user of a body powered prosthesis has to deliver more energy. Also electric prostheses require batteries with a larger capacity and motors which are more powerful, which may contribute to a higher device mass.

Articulating fingers

Recently electric hands with articulating fingers have become commercially available [7, 8]. For body powered prostheses only one articulating hand is available, the Becker Imperial Hand [14]. Hands with articulating fingers have mechanisms with multiple joints. Therefore they have a higher energy dissipation than hands with stiff fingers.

This results in either a lower pinch force, or an increase in the required operating energy. The pinch force of the current articulating electric hands (15-35 N) [7, 8] is lower than the pinch force of the stiff fingered electric prostheses (~100 N) [15]. In order to increase the pinch force and to reduce the energy demand, of both body powered and electrical powered hands, it is desirable to use a glove with a low stiffness and hysteresis. Although PVC and silicone gloves are used for decades now [11], there is very limited quantitative data available about the stiffness and hysteresis of cosmetic gloves. Special gloves were designed for the i-limb and the bebionic. However no data was published on the mechanical properties of these gloves. Herder *et al.* [10] measured the stiffness and hysteresis of a PVC cosmetic glove for movements of the thumb. Currently no data is available of the effect of cosmetic gloves on other joints. Also no data is available on the stiffness and hysteresis of silicone cosmetic gloves. The Delft Institute of Prosthetics and Orthotics (DIPO) currently develops a new prototype of a body powered articulating hand. It would be desirable to use a standard cosmetic glove for this prototype, instead of a special glove designed for the i-limb or the bebionic, as the purchase prices of these special gloves are considerably higher (3.5 to 6.5 times higher for the bebionic and 5 to 10 times for the i-limb glove than standard gloves). Furthermore standard sized cosmetic gloves are available through multiple manufacturers. To select the right cosmetic glove, it is necessary to know its mechanical properties.

5.2 Goal

The goal of this study is to determine the contribution of a standard PVC and silicone cosmetic glove to the stiffness of the finger joints, and to quantify the energy dissipation during articulation of the finger joints. This data enables the selection of the most efficient glove for a new prototype of an articulating hand. Furthermore the data can be used in future development of prosthetic hands and cosmetic gloves.

5.3 Methods

Tested gloves

Three pair of midsize cosmetic male gloves were tested, size 7 ¾ (Figure 5.1). This corresponds to the size of a small adult male hand, or a large female hand. Three gloves were made of PVC, the other three of silicone. The gloves were manufactured by RSL Steeper (Table 5.I). The gloves were slush moulded gloves from the same hand model, and were therefore almost identical.

Figure 5.1 Two of tested cosmetic gloves: The PVC glove (left) and the silicone glove (right). The gloves were identical of size, shape and texture. Only the material differed. The fingers are slightly flexed in their neutral position.

Table 5.I Tested gloves

Glove number	*Material*	*Side*	*Size*	*Mass (gram)*	*Brand*
CG302/E4	PVC	L	7 ¾	338±23	RSL Steeper
SG302/E4	Silicone	L	7 ¾	247±6	RSL Steeper

Test set up

To test the fingers an articulating finger frame was designed and built. The finger frame has one joint, which has two low friction roller bearings. The finger frame can be flexed by pulling the flexor cable (d=0.8 mm) attached to a pulley (r=4 mm). A dead mass of 1.0 kg, attached to the extensor cable, exerts an extension torque of 43.2 Nmm to the finger joint. The mass exerts a constant extension torque to the joint. It does not add stiffness to the joint. The flexor cable is attached to a test bench, which measures and records the cable force and displacement. A schematic overview of the set-up is given in Figure 5.2. The thumb of the glove was moved away, to avoid collision with the fingers.

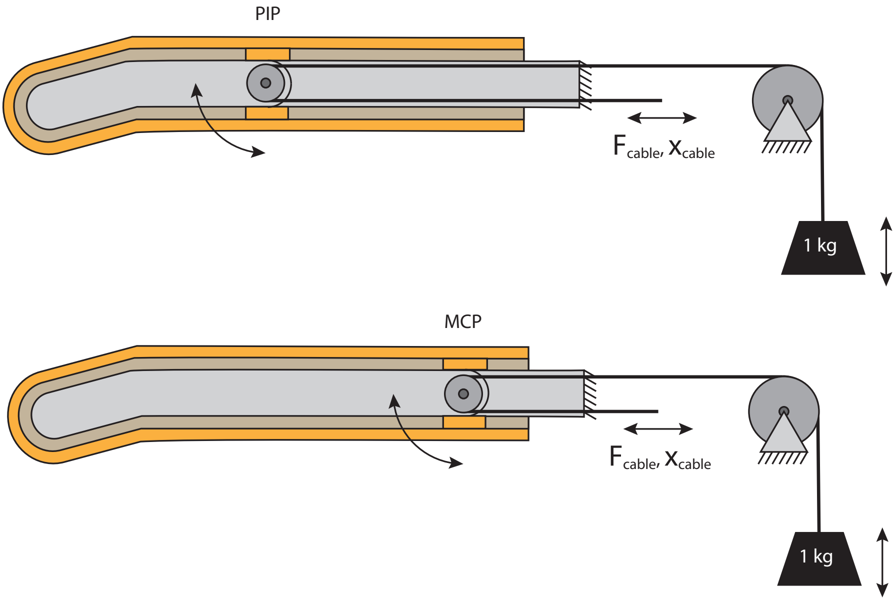

Figure 5.2 Schematic overview of the test set-up configurations of the PIP and MCP test. The tested cosmetic gloved is fitted over the finger frame. The joint of the finger frame is aligned with the joint location of the glove that will be tested. The spaces between glove and frame are filled with soft foam (dark grey).

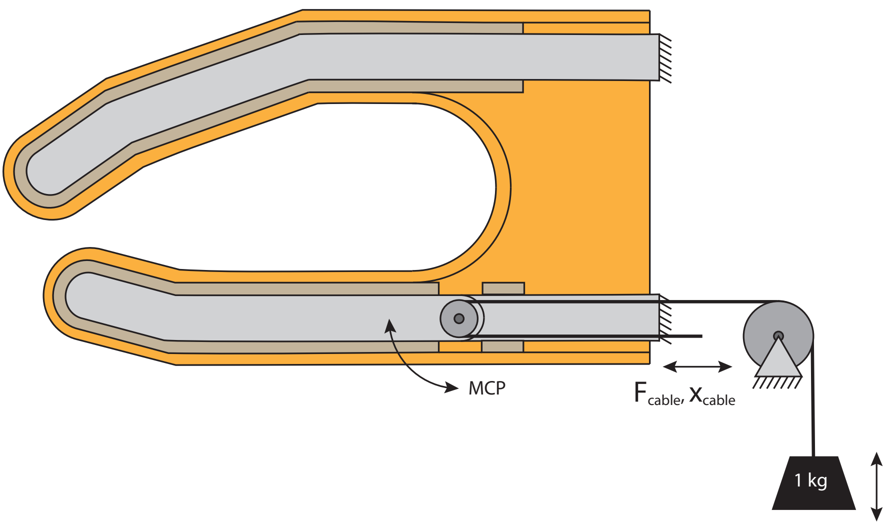

Figure 5.3 Schematic overview of the test set-up configurations of the MCP joint of the thumb and the web space between thumb and fingers. Activation of the cable opens the thumb. The counterweight closes the thumb when the cable is released. The spaces between glove and frame are filled with soft foam (dark grey).

The MCP joint of the thumb and the web space between the thumb and the finger were tested in a similar way. The thumb was pulled open from its neutral position. The thumb was closed again by the 1 kg counterweight (Figure 5.3).

Test protocol

The thumb, the index finger, the middle finger and the ring finger of the gloves were tested. For each finger the MCP-joint and the PIP-joint were tested. For the thumb the MCP-joint was tested. In each test the finger frame was placed inside the tested finger of the cosmetic glove. The joint of the frame was aligned with the tested joint location (MCP or PIP) of the cosmetic glove (Figure 5.2). Each finger joint was flexed 0.5 π·rad (90°) and extended again. The thumb was extended π/3·rad (60°) from its neutral position and flexed again. Cable force and displacement were measured. The cable translations were used to calculate the joint angles, the cable forces were used to calculate the joint torques. All tests were repeated three times, to obtain average values. The tests were preceded by three unrecorded trials, to avoid transient effects. The following parameters were measured or calculated from the measured data:

Work and hysteresis

For each finger joint the amount of work, that was required to flex the joint from 0.1 to 0.5 π·rad, was measured. The amount of dissipated energy, or hysteresis, was measured for the same interval. Instead of an interval of 0 to 0.5 π·rad, the interval of 0.1 to 0.5 π·rad was considered. A counterweight of 1.0 kg was not heavy enough to fully extend the gloved PVC fingers, as the fingers are slightly flexed in their neutral position in which they are moulded. For the thumb an interval of 0 to π/3·rad was considered.

Four empty runs were performed, to determine the required work and hysteresis of the test set-up. The setup required 18.9±0.04 Nmm work, and had a hysteresis of 1.2±0.02 Nmm for a joint rotation of 0.5 π·rad. The system work and hysteresis were substracted from the measured joint work and hysteresis.

Maximum joint torque

The maximum required joint torque (M_{max}) was recorded for each joint.

Average joint stiffness

The average joint stiffness was calculated for each joint, by making a linear least squares fit to the data at the interval of 0.1 to 0.5 π·rad for the fingers, and at an interval of 0 to π/3·rad for the thumb.

Glove or skin thickness

As a result of the slush moulding process there is a variation in glove thickness within a single glove and between different gloves. The glove thickness was measured for each tested joint for all gloves, at the MCP and at the PIP joint. First the finger of the glove was squeezed, then the total thickness was measured using a micrometre calliper. The ratchet knob at the micrometre guarantees a constant calliper tip pressure for each measurement. Each joint was measured four times at different positions. The average of the measured thickness was divided by two, to obtain the thickness of one glove layer.

5.4 Results

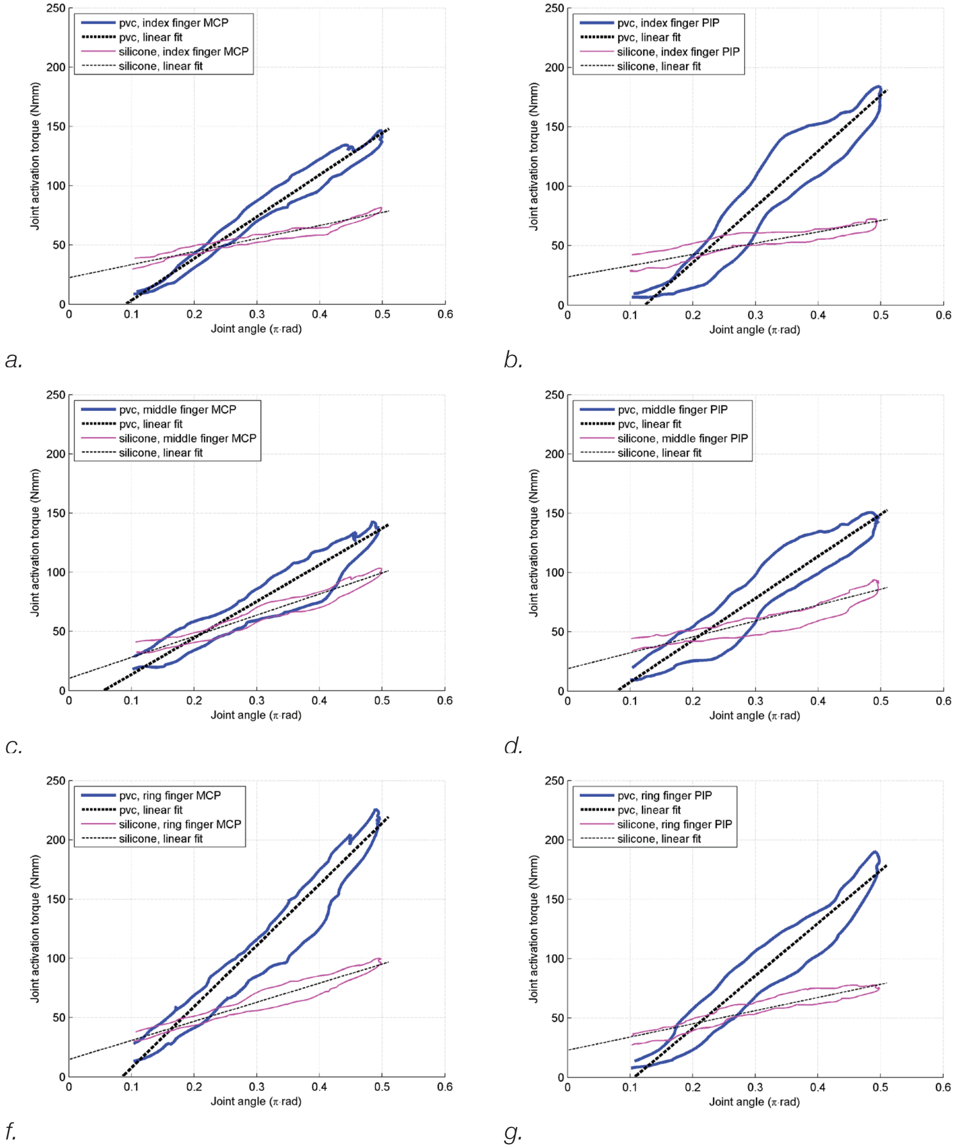

Figure 5.4 Angle-torque-diagrams of the MCP (on the left) and PIP-joints (on the right) of each finger (index finger: a, b; middle finger: c, d; ring finger: f, g). The thick line represents the PVC-glove, the thin line the silicone glove. For reasons of clarity the data of one trial of one PVC and one silicone glove is shown. The measured torque values include the constant torque (43.2 Nmm) produced by the counter mass.

Angle-torque-diagrams

The angle-torque-diagrams of the finger joints (Figure 5.4) show the measured torque for each joint angle between 0.1 and 0.5 π·rad, for one cycle. For reasons of clarity the data of one trial of one PVC and one silicone glove is shown. The data were representative for the other trials and gloves. Joint angles smaller than 0.1 π·rad were disregarded, and are therefore not displayed. The diagrams show higher maximum joint torques for the PVC glove, for each joint. The diagrams of the PVC glove are running steeper, indicating a higher joint stiffness. They also enclose a larger area, which indicates a larger hysteresis for the PVC glove. This is similar for the thumb joint (Figure 5.5).

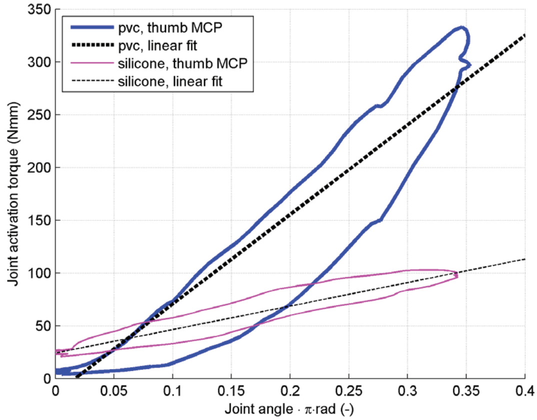

Figure 5.5 Angle-torque-diagrams of the MCP thumb joint and thumb web space. Note that the scale for the torque differs from those of the fingers. For reasons of clarity the data of one trial of one PVC and one silicone glove is shown. The measured torque values include the constant torque (43.2 Nmm) produced by the counter mass.

Work and hysteresis

The amounts of work and hysteresis that were measured for each joint, are given in Table 5.II. The work and hysteresis of the test set-up are already subtracted from these values. The area between the upper line of a diagram and the x-axis, represents the amount of work (Nmm). The area enclosed by a diagram represents the hysteresis (Nmm). The PVC glove required 1.8 to 3.8 times more work to flex the individual joints of the fingers. The PVC glove also dissipated 1.7 to 3.4 times more energy.

Maximum joint torque

The maximum joint torque (M_{max}), required to fully flex a joint, was 2.2 to 4.2 times higher in the PVC-gloves (Figure 5.6).

Stiffness and glove thickness

The stiffness of the PVC gloves was 2.5 to 4.5 times higher than that of the silicone gloves (Figure 5.7). The measured finger thickness is also given in Table II. The fingers of the PVC gloves were 1.5-1.7 times thicker.

Table 5.II This table presents the work and hysteresis that was measured for every joint, during a 90° flexion and extension of the finger joints and 60° of the thumb joint and web space. M_{max} is the maximum measured joint torque, minus the torque produced by the counterweight (43.2 Nmm). The joint stiffness was calculated from the measured data. The glove thickness was measured for each joint.

Finger, Joint	**Material**	**Work (Nmm)**		**Hysteresis (Nmm)**		**Mmax (Nmm)**		**Stiffness (Nmm/rad)**		**Thickness (mm)**	
		AVE	**STD**	**AVE**	**STD**	**AVE**	**STD**	**AVE**	**STD**	**AVE**	**STD**
Thumb, MCP	PVC	27,5	7,7	15,0	4,7	218	42,4	698	115	1,6	0,2
Thumb, MCP	Silicone	13,9	0,5	6,3	0,3	69	6,5	241	20	0,9	0,1
Index, MCP	PVC	17,8	2,7	7,7	2,1	124	16,1	365	71	1,3	0,2
Index, MCP	Silicone	6,4	1,7	3,1	1,8	43	3,5	108	11	0,9	0,1
Index, PIP	PVC	19,7	1,7	12,5	2,1	156	18,0	505	74	1,5	0,2
Index, PIP	Silicone	5,2	0,9	4,0	1,0	37	8,6	115	34	0,9	0,1
Middle, MCP	PVC	19,1	2,6	11,1	1,4	134	24,9	365	111	1,3	0,1
Middle, MCP	Silicone	10,7	2,7	6,5	3,0	61	10,2	145	33	0,9	0,1
Middle, PIP	PVC	17,3	2,1	11,8	1,6	133	23,4	429	91	1,5	0,2
Middle, PIP	Silicone	5,4	0,6	4,3	0,9	43	5,9	130	20	0,9	0,1
Ring, MCP	PVC	24,9	2,1	11,0	1,6	153	38,5	413	146	1,4	0,2
Ring, MCP	Silicone	9,9	1,4	4,6	1,1	58	3,8	148	10	0,9	0,1
Ring, PIP	PVC	20,1	1,5	12,4	1,4	165	21,5	508	70	1,5	0,2
Ring, PIP	Silicone	6,3	0,7	3,7	0,7	41	10,3	113	23	0,9	0,1
Average	PVC	19,8	2,5	11,1	1,6	144	15	431	94	1,4	0,2
Average	Silicone	7,3	2,2	4,4	1,1	47	9	126	22	0,9	0,1

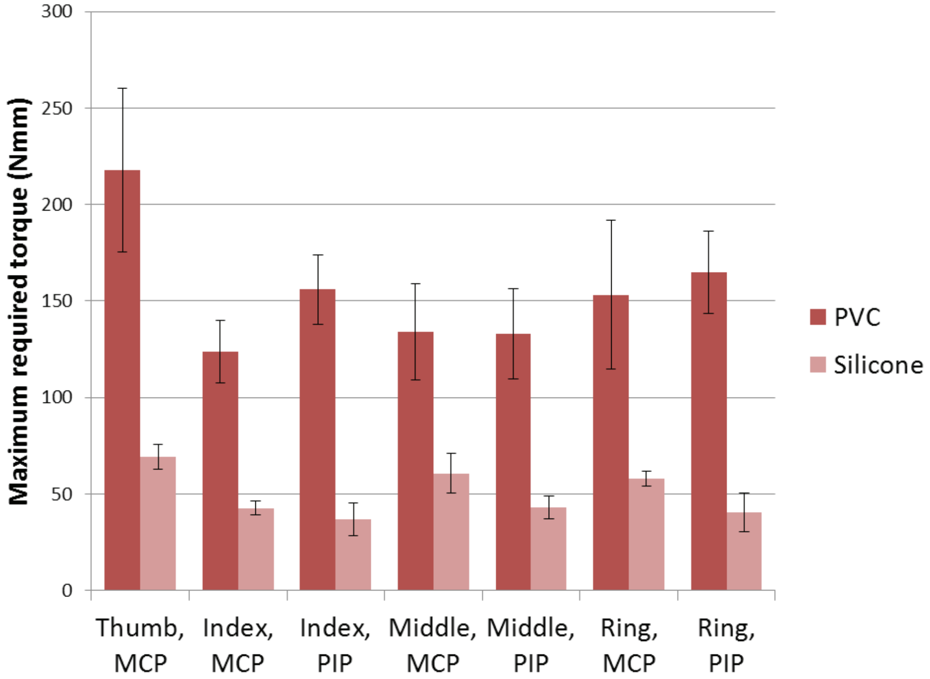

Figure 5.6 The maximum joint torque to flex the joint 0.5 π·rad (90°). The maximum joint torque was up to 4.2 times higher in the PVC-glove

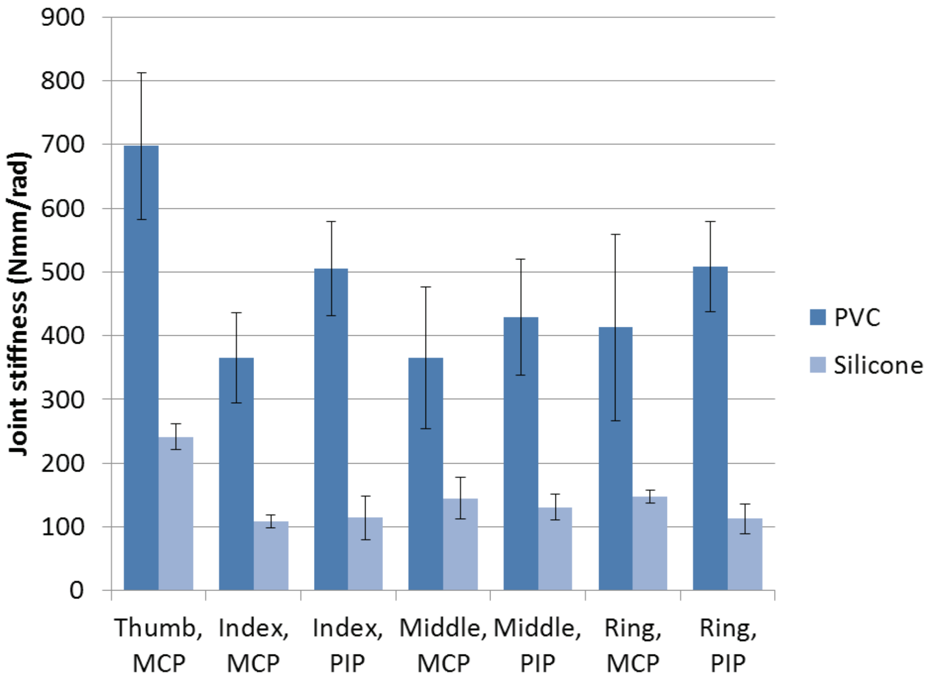

Figure 5.7 The stiffness of the different joints, for the PVC and the silicone cosmetic gloves. The stiffness of the PVC glove was up to 4.5 times higher than that of the silicone glove.

5.5 Discussion

Angle-torque-diagrams

The angle-torque-diagrams of the fingers are only shown for 0.1 to 0.5 π·rad joint flexion. The finger joint did not fully extend when a PVC glove was applied. This is because the fingers of both gloves are already a little flexed in their neutral position in which they are moulded, like in the real human hand. As a result it required a negative torque to fully extend the finger. It turned out that the counterweight of 1 kg was not heavy enough to fully extend the joint of a PVC glove. This was not a problem when a silicone glove was applied, as it had a lower stiffness. It was undesirable to further increase the counter mass, as this would result in a high load and friction inside the roller bearings. To make a fair comparison between both cosmetic gloves, it was decided to disregard the data between 0 to 0.1 π·rad for the finger joints.

Work and hysteresis

The PVC gloves required considerably more work. They also dissipated considerably more energy. The sum of the required work, of the MCP- and PIP-joints of the three tested fingers of the PVC glove, was 119 Nmm. This seems relatively low, when compared to closing a voluntary closing Otto Bock hand (1710 Nmm) or a Hosmer APRL hand (1058 Nmm) [5]. However, these hands all require very high activation forces. Non-hand-like gloveless prehensors, e.g. the TRS-GRIP (284 Nmm), require a much lower activation force, because they have a relatively simple mechanism.
These devices are not meant to be covered. When we would add a cosmetic covering to such a device, this would considerably increase the total required work of the prehensor. Devices which have a more complex mechanism, with more joints and transmissions, require much more work than relatively simple mechanisms. Therefore for the more complex devices the work that is added by a cosmetic glove is relatively smaller. However also for these devices the work added by the cover should be as low as possible, because the more complex mechanism itself is requiring so much more work from the user.

Stiffness and glove thickness

All the joints had a positive stiffness, which could be approximated well by a linear fit. The PVC gloves were much stiffer (2.5 to 4.5) than the silicone gloves. This can only partially be explained by the larger thickness of the PVC gloves, as the PVC gloves were only 1.5 to 1.7 times thicker. The largest difference in stiffness (350%) was measured for the PIP joint of the index finger, for which the thickness between the gloves only differed 67%. Therefore it can be concluded that the main cause of the difference in glove stiffness is the difference in stiffness properties of the PVC and silicone glove material. For a minimal required input energy, and a maximal pinch force, the stiffness should be as low as possible. To further reduce the drawbacks of the glove stiffness, the glove stiffness can be compensated. This can be achieved by using a stiffness compensation mechanism [16]. As the glove has a linear positive stiffness, the compensation mechanism should have a linear negative stiffness.

Study limitations and strengths

In this study three pair of gloves of one manufacturer were tested. The outcomes may vary among other brands and types. However, it is unlikely that such variations will give a total different outcome. The variations in the results among the joints and the fingers were small, whereas the variations between the gloves were large. Both gloves were identical of size, shape and texture. The only parameter that differed between the gloves was the glove material. This study clearly shows that the glove material has a considerable effect on the mechanical performance of a cosmetic glove and on the energy requirement of a prosthetic hand. The results are in line with the design choice of the designers of the i-limb and of the bebionic. They used silicone when they designed special multi-layered reinforced gloves for these new hands.

Implications for non-articulating hands

For the non-articulating, or 'stiff fingered' hands, the largest glove deformation takes place in the web space between the thumb and the fingers. The properties of this part of the glove are represented by the measurement of the thumb MCP-joint and the web space. These results show that for the thumb joint the stiffness is 2.9 times higher. The required work is 2.0 times higher. So also for non-articulating hands the tested silicone cosmetic gloves will require less energy than the PVC gloves.

Clinical significance

The outcomes of this study can help the clinician to select the most suitable cosmetic glove for a patient. Based on the mechanical properties, the silicone gloves outperformed the PVC gloves. Using a silicone glove, will increase the battery life for an externally powered prosthesis. For a body powered prosthesis, it will result in a lower user effort and an increased user comfort. However, when selecting a cosmetic glove, also other properties should be taken into account (e.g. durability, cosmetic appearance, cost and resistance to staining). The results of this study also give directions for manufacturers and researchers in order to develop improved cosmetic gloves and prosthetic hands.

5.6 Conclusion

Three pair of identical standard cosmetic gloves of different materials were mechanically tested, three PVC gloves and three silicone gloves. The most efficient glove was selected to be used for a new developed articulating hand prototype. Both types of glove showed a linear joint stiffness characteristic. The silicone gloves had the lowest joint stiffness for all joints. As a result they required less energy for flexing the joint and dissipated less energy. Also the joint torque to fully flex the joints was considerably lower. Based on energy requirements, joint stiffness, and required joint torque, the silicone gloves had a higher mechanical efficiency than the PVC gloves. They dissipated less energy and they required a lower activation force. This will result in an increased battery life, or user comfort, depending on the type of prosthesis that is used.

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6

A mechanism to compensate undesired stiffness in joints of prosthetic hands

In press:

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Abstract

Background: Cosmetic gloves that cover a prosthetic hand, have a parasitic positive stiffness that counteracts the flexion of a finger joint. *Objectives:* Reducing the required input torque to move a finger of a prosthetic hand, by compensating the parasitic stiffness of the cosmetic glove. *Study Design:* Experimental, test bench. *Methods:* The parasitic positive stiffness and the required input torques of a PVC and a Silicone glove were measured when flexing a MCP finger joint 90°. To compensate this positive stiffness, an adjustable compensation mechanism with a negative stiffness was designed and built. A Matlab model was created to predict the optimal settings of the mechanism, based on the measured stiffness, in order to minimize the required input torque of the total system. The mechanism was tested in its optimal setting with an applied glove. *Results:* The mechanism reduced the required input torque with 58 % for the PVC glove and with 52 % for the silicone glove. The total energy dissipation of the joint did not change significantly. *Conclusions:* This study shows that the undesired positive stiffness in the joint, can be compensated with a relatively simple negative stiffness mechanism, which fits inside a finger of a standard cosmetic glove.

6.1 Background

The artificial hands used in the prosthetic field are usually covered by a cosmetic glove, made out of silicone or PVC [1]. The cosmetic glove gives the hand a more lifelike appearance and it protects the hand mechanism against water and dirt. Unfortunately the cosmetic glove has also some negative properties, especially from a mechanic point of view. The stiffness of the glove material counteracts the movement of the hand mechanism [2] and of the finger joints [3]. As a result the operator has to deliver extra input energy to operate the prosthesis. This results in an increased user effort in body-powered prosthesis, which are directly operated by the user [4, 5]. It also results in a decreased battery life in electric-powered prostheses. These undesired effects can be reduced by lowering the glove stiffness [2]. Another solution is to counteract the parasitic glove stiffness, by using a stiffness compensation mechanism. Such a mechanism has a stiffness which is opposite to the parasitic stiffness of the glove [6, 7]. Although such a mechanism has been described in some papers [2, 6, 7], it has not been built and evaluated. The myoelectric System Hands by Otto Bock do have some glove compensation, which is provided by a wire spring inside the palm of the hand. However this spring is only applied in the electric hands, not in the body-powered hands, and it compensates only one degree of freedom. It would be desirable to be able to compensate more degrees of freedom, in multiple joints.

6.2 Problem definition

The stiffness of the cosmetic glove causes an undesired joint torque in the finger joint of a prosthetic hand. As a result an extra input torque is required to move a finger joint.

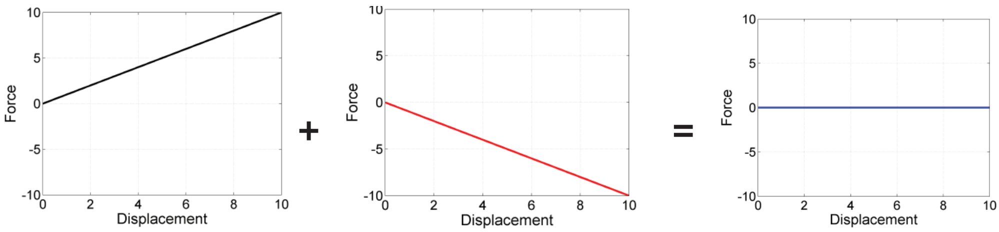
6.3 Goal

The goal of this study was to design and evaluate a mechanism that compensates the undesired stiffness in finger joints of prosthetic hands, which is induced by the cosmetic glove. In this way the compensation mechanism should reduce the input torque that is required to move a finger joint.

6.4 Methods

The positive parasitic stiffness of the cosmetic glove can be compensated by a spring with a negative stiffness (Figure 6.1). Figure 6.1a shows the force-displacement curve of a typical positive stiffness linear spring. When such a spring is elongated (displacement on the x-axis increases) the spring pull force increases linearly (the force on the y-axis increases). The curve has a positive slope. In order to enable a forceless motion, the positive stiffness spring needs to be compensated. This can be achieved by using a spring which has exactly the same behaviour in the opposite direction. Instead of an increasing pull force, such an ‘imaginary spring’ would need to deliver a linearly increasing push force when the spring is elongated.

As this force acts in the opposite direction, the force of this spring is considered to be negative. The force-displacement curve of such a negative linear stiffness spring is shown in Figure 6.1b. This curve is the exact reflection of the positive stiffness curve (Figure 6.1a), with respect to the x-axis. The curve has a negative slope, which means that the stiffness of the spring is negative. When both graphs are added, (i.e. the ends of both springs are attached together) the resultant curve is a line which runs along the x-axis, with a value of 0 (Figure 6.1c). In this way the sum of the spring forces of both springs is zero for every position. Therefore the moving point can be moved to every position, without any force. The slope of the curve is zero, which means that such a mechanism has a stiffness which is zero.



a. *b.* *c.*

Figure 6.1 An undesired positive stiffness of a component (left) can be compensated by using a mechanism with a negative stiffness (middle). The result is a mechanism which can move with a zero stiffness.

Although there exists no spring with a negative stiffness, it is possible to create a negative stiffness by using a tension spring with a positive stiffness, together with two bars and three joints (Figure 6.2). When angle α increases the spring moves away from the joint axis and the moment arm increases. At the same time the spring shortens, which results in a decreasing force in the tension spring. The moment in the joint (M_{spring}) is the product of the moment arm and the spring force. It is possible to choose the parameters, e.g. spring stiffness, spring dimensions, and spring configuration, in such a way that the joint moment decreases when the angle displacement increases. In this way the slope of the resulting curve will be negative, which means that such a mechanism has a negative stiffness.

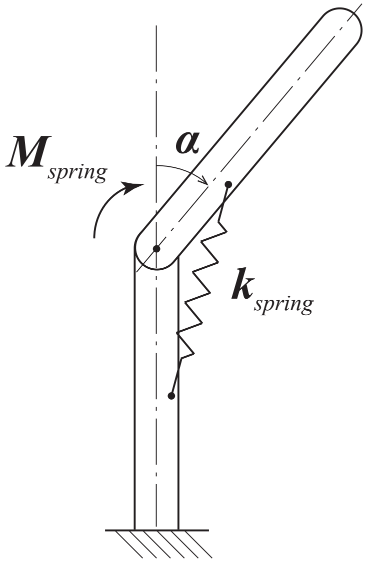


Figure 6.2 Schematic example of a negative stiffness spring mechanism. A tension spring with stiffness k_{spring} exerts a torque M_{spring} to the upper bar. When the right spring parameters have been selected, the joint torque decreases, as the joint angle α increases.

Protocol

To compensate the parasitic glove stiffness in a finger's joint, the following steps were followed:

- The parasitic glove stiffness in the finger joint was measured, together with the required input torque and the energy dissipated by the glove. The glove was placed on the set-up and the finger was flexed from 0° to 90° and extended again.
- The measured force-displacement curve was fit with a linear approximation to approach the glove stiffness.
- The different feasible configurations of the compensation mechanism were modelled.
- The optimal solution was selected.
- The selected solution was implemented.
- The resultant stiffness in the finger joint was measured.

Test set up

A finger frame was placed inside the finger joint, to measure the joint stiffness (Figure 6.3). In this study the MCP-joint of the index finger was used. The finger frame consisted of two bars, connected with a joint. The joint had ball bearings, to minimize the joint friction in the finger frame. A cable running over a pulley controlled the joint angle. One end of the cable was attached to the measurement set up. The measurement set up activated the cable and measured the cable force and displacement. The other end of the cable was attached to a counter mass, which returned the finger joint to its initial position. The thumb of the glove was moved sideways out of the trajectory of the index finger, to avoid collision. The extension force applied by the counter mass was subtracted from the cable force. The resulting force, or activation force, was multiplied by the pulley radius to obtain the activation torque, which was recorded for every joint angle.

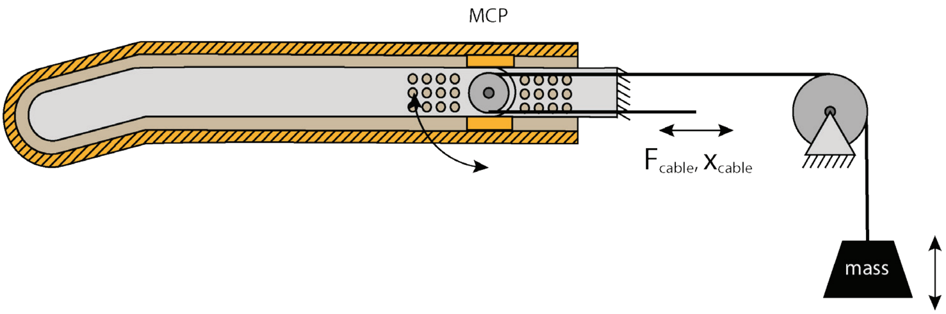


Figure 6.3 Test set-up to measure the joint stiffness. The finger was flexed by pulling the cable. The cable force and displacement were recorded. The counter mass extended the finger when the cable was released. The extension force applied by the counter mass was subtracted from the cable force. The mechanism was also used as a stiffness compensation mechanism, by placing two pins in the holes at the left and right side of the joint and by attaching a spring to those pins (see Figure 6.5).

Figure 6.4 The RSLSteeper cosmetic gloves that were used in this study, the PVC glove (left) and the silicone glove (right).

Gloves

For the test two different cosmetic gloves were used, a silicone cosmetic glove (type: SG302/E4) and a PVC cosmetic glove (type: CG302/E4). Both gloves were manufactured by RSLSteeper (Figure 6.4) and had identical shape and size.

Compensation mechanism

The finger frame of the test set-up was used as a compensation mechanism by attaching a helical tension spring to the frame. Above the joint and below the joint there were twelve spring attachment points, in a grid of 3 by 4. This enabled 12x12=144 different configurations. The distance between the centres of the holes was 3 mm. The distance of the first row to the joint centre was 6 mm. A pin could be placed in one of the holes to serve as a spring attachment point (Figure 6.5). The forces acting on the distal phalanx, when a spring was applied, are shown in Figure 6.6. Spring manufacturers offer many different springs. Important spring parameters are: spring stiffness, rest length, maximum elongation, spring diameter and wire diameter. As the spring had to fit inside a finger, along the finger frame, the maximum spring diameter was set at 5 mm. Because a prosthetic hand has to function in a moist environment (e.g. ambient humidity or seepage from immersion of the hand) the springs had to be made of stainless spring steel. From the product range of standard helical springs of a large spring manufacturer (www.alcomex.nl) a total of 52 springs met both criteria.

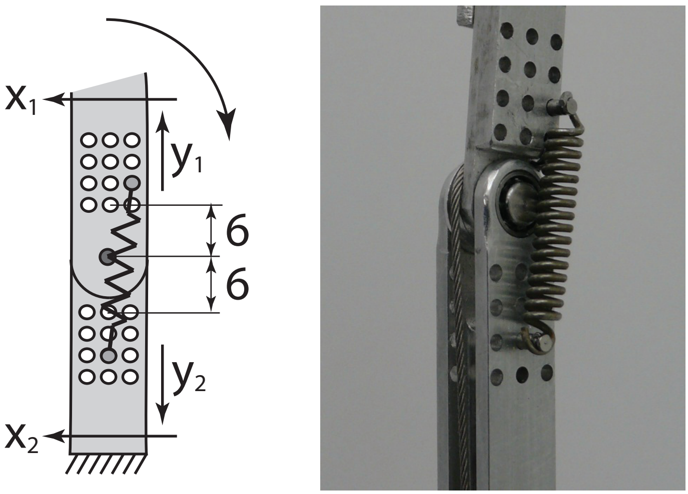


Figure 6.5 There were 3x4x3x4=144 different configurations to attach the spring to the mechanism. The distance between the centres of the joint was 3 mm. The curved arrow indicates the direction of the joint flexion. The figure shows configuration x_1=1, y_1=2, x_2=2, y_2=3, both schematically (left) and on the real mechanism (right).

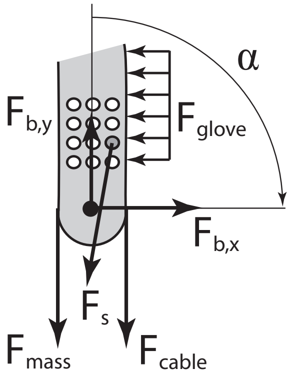


Figure 6.6 Forces acting on the distal phalanx, inducing torques around the joint axis. The forces of the cosmetic glove (Fglove) and the extension force of the counter mass (Fmass) induce a counter clockwise or positive torque. The cable force (Fcable) and the force of the compensation spring (Fs) induce a clockwise or negative torque. Fb,x and Fb,y represent the reaction forces in the joint, which do not contribute to the torque. The range of motion of the phalanx is indicated by arrow α.

Matlab model

A model was created in Matlab to select the optimal spring together with the optimal spring configuration. The parameters of the 52 selected springs were included in the model. The model calculated the force-displacement curve for every spring, in every configuration. The force-displacement curve of the compensation mechanism was added to the curve of the glove, to calculate the resultant curve (Figure 6.1). In the final step the algorithm selected the resultant curve that had the smallest deviation from the horizontal axis. This is the combination of spring and attachment configuration that requires the lowest input torque.

6.5 Results

Figure 6.7 shows the measured joint torque as a function of the angle, for one cycle of 90° flexion and extension of both gloves. The torque increased almost linear with the joint angle. The area enclosed by the entire curve represents the amount of energy dissipated by the glove during one cycle. When the PVC glove was applied the finger required a torque range of 193 Nmm (from a minimum torque of -77 Nmm to a maximum torque of 116 Nmm). During one cycle 69 Nmm or mJ of energy was dissipated. The silicone glove required a torque range of 73 Nmm (from -33 Nmm to 40 Nmm). During one cycle 27 Nmm or mJ of energy was dissipated. Figure 6.7 shows the linear approximation of the joint stiffnesses of both gloves. The linear approximation of the joint torque and the joint angle θ (in degrees) could be described by $M_{pvc}=-72+2.07\cdot\theta$ for the PVC glove and by $M_{sil}=-29+0.73\cdot\theta$ for the silicone glove.

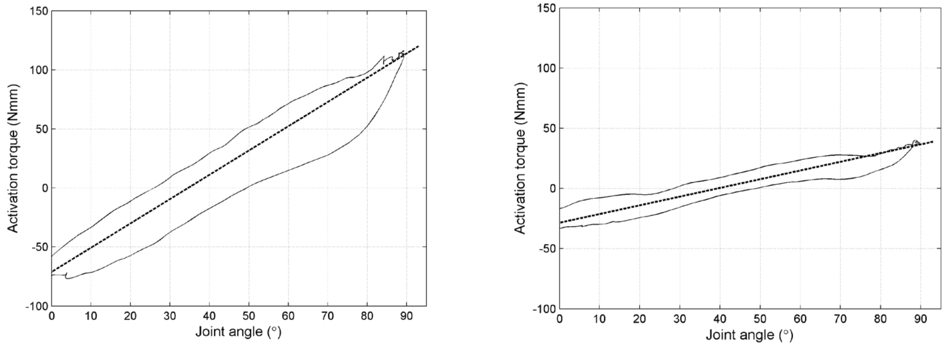


Figure 6.7 Measured joint stiffness (black line), and linear approached stiffness (dashed linear line), of the PVC glove (left) and the silicone glove (right). The diagram represents one cycle of 90° flexion and extension. The difference between the minimum and the maximum measured torque was 193 Nmm for the PVC glove and 73 Nmm for the silicone glove.

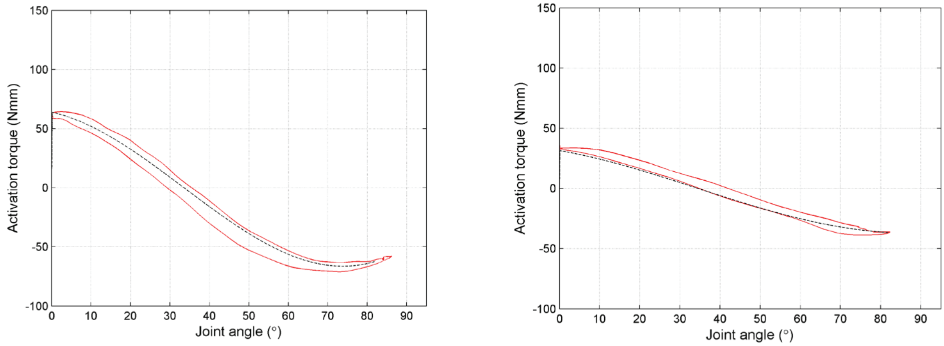


Figure 6.8 The predicted (black) and the measured (red) characteristic of the suggested spring, installed on the compensation mechanism in its optimal configuration. The TR 480 spring for compensation of the PVC glove (left). The TR390 spring for compensation of the silicone glove (right). The diagram represents one cycle of 90° flexion and extension.

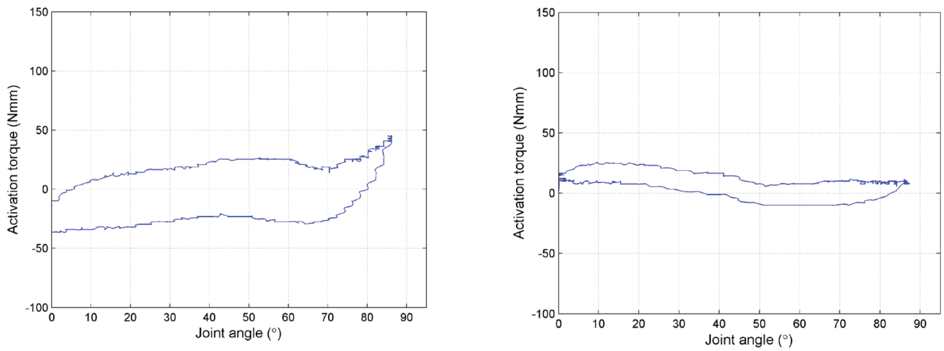


Figure 6.9 The resultant torque characteristic after the glove was placed over the finger. The difference between the minimum and the maximum measured torque was 81 Nmm for the PVC glove and 35 Nmm for the silicone glove. This is a reduction by 58 % and 52 % respectively, compared to the uncompensated glove.

To compensate the stiffness of the PVC glove, the MatLab model predicted the optimal result for spring TR480 (Alcomex.nl), in configuration x_1=1, y_1=2, x_2=1, y_2=2. In this configuration the system dissipated 37 Nmm or mJ of energy. For the silicone glove the optimal compensation was predicted for the TR390 spring, in configuration x_1=1, y_1=1, x_2=1, y_2=4. Figure 6.8 shows the predicted characteristic of the spring, together with the measured characteristic. In this configuration the system dissipated 21 Nmm or mJ of energy.

Figure 6.9 shows the resultant characteristics of the joint of both fingers, with the spring and the glove applied. The PVC glove in combination with the compensation mechanism, required a torque range of 81 Nmm (from a minimum torque of -36 Nmm to a maximum torque of 45 Nmm). This is 42 % of the required torque range of the uncompensated glove. During one cycle 67 Nmm or mJ of energy was dissipated. The silicone glove in combination with the compensation mechanism, required a torque range of 35 Nmm (from -10 Nmm to 25 Nmm), which is 48 % of the required torque range of the uncompensated glove. During one cycle 28 Nmm or mJ of energy was dissipated.

6.6 Discussion

This study showed that it was possible to compensate the undesired joint stiffness caused by the cosmetic glove. The compensation mechanism reduced the required joint torque range of the PVC glove by 58 % and of the silicone glove by 52 %. It was however not possible to achieve a perfect compensation, as in Figure 6.1c. In the first place the glove stiffness was linearly approached. The real glove characteristic was however not entirely linear and had also a considerable amount of hysteresis (Figure 6.7). In the second place the real spring compensation characteristic might also have deviated a little from the measured characteristic (Figure 6.8, right), as the spring properties usually slightly differ from the properties specified by the manufacturer. Also the spring mechanism added a small amount of hysteresis. In the third place the optimal calculated resultant characteristic (Figure 6.9) was not exactly zero for every joint angle, due to the limited amount of springs and spring attachment locations. Despite these inaccuracies and hysteresis, the input torque was considerably reduced.

Implications for prosthetic hands

The mechanism can be applied to both body-powered and electric hands. In electric hands the compensation mechanism will help to reduce power consumption. Furthermore it might be possible to select other electric motors, which have a lower mass. Excessive prosthesis mass is currently one of the most important factors in prosthesis rejection [8]. Finger movement requires a fast movement at a low torque, pinching requires a small movement at a high torque. Bi-phasic systems [9, 10] and systems with ‘synergetic prehension’ [11, 12] use this property to reduce the motor mass and required energy. When the torque is reduced by the glove compensation mechanism, these systems could be optimized to even further reduce energy consumption and to obtain a small reduction in motor mass. However the real benefit of the compensation mechanism will be for body-powered prostheses. As in these hands the user of the hand has to deliver the input energy himself. In many current hands this is a problem, as the forces the user has to deliver are too high [4]. Applying the compensation mechanism to a prosthetic hand will help to reduce the force that the user has to deliver.

The study showed that the undesired stiffness of the cosmetic glove can be compensated by a relatively simple mechanism. All that is needed are a spring, the attachment points, and a small build-in space. The mechanism works for individual fingers, every single finger can be compensated. When applying the mechanism to a prosthetic hand, the biggest challenge will be the limited build-in space inside the finger. Other aspects, e.g. costs and mass will be of minor influence. Because of the small number of extra required parts and the low mass of the parts (less than 1 gram per joint), the glove compensation will only add little extra mass, relative to the mass of a standard prosthetic hand (~500 gram).

Inner glove compensation?

Beside a cosmetic glove, many prosthetic hands also have an inner glove. The inner glove gives the hand its shape and it provides an extra protection of the mechanism against the environment. From a mechanical point of view the inner glove has the same undesired properties as the cosmetic glove. The inner glove is even thicker and stiffer than the cosmetic glove and therefore it dissipates more energy than the cosmetic glove and it requires a higher input force [4]. Although it might also be possible to compensate the stiffness of the inner glove, this will be harder to achieve due to its higher stiffness. However, the use of an inner glove is not a necessity. Hands like the Becker Imperial hand or the Hosmer APRL hand do not use an inner glove. Also new hands like the i-limb and the bebionic do not use inner gloves. Because of its negative mechanical properties the use of an inner glove should be avoided in future designs [4]. Omission of the inner glove will also leave more build-in space for the compensation mechanism.

Other applications

This study was performed to solve the problem of the undesired stiffness of finger joints of prosthetic hands, caused by the cosmetic glove. The presented solution can however be used for every joint which has an undesired joint stiffness, e.g. a bundle of cables or a hydraulic hose attached to a joint of a robot arm.

Strengths and limitations

Although the glove compensation principle was already described in earlier studies [2, 6, 7], it was still unknown whether it would be feasible to design a mechanism that would fit inside a prosthetic hand. This study showed that it was even possible to fit such a mechanism inside a finger. The mechanism used standard springs and was adjustable to different gloves, even when the gloves were made of different materials and had different stiffness characteristics. This enabled the use of different gloves, which is important as there are various different glove manufacturers who produce a broad range of PVC and silicone cosmetic gloves of different sizes and thicknesses. The most important limitation of the compensation mechanism was the hysteresis band, which was largely caused by the hysteresis in the glove. The compensation mechanism added a small amount of extra friction. As a results of the hysteresis band, the system always needed an input torque and always dissipated energy during motion. A strength of the presented compensation mechanism is that it only required two extra joints (the attachment points of the spring), so the added amount of joint friction was small.

6.7 Conclusions

This study showed that the undesired positive stiffness in the joint of the cosmetic glove, can be compensated with a relatively simple negative stiffness mechanism, by using a standard helical tension spring. The mechanism was designed, built and tested. The entire mechanism fit inside a finger of a standard cosmetic glove. The mechanism could compensate the positive stiffness of a silicone cosmetic glove, as well as of a much stiffer PVC glove. The compensation mechanism reduced the required input torque range by 58 % for the PVC glove and by 52 % for the silicone glove. The addition of the compensation mechanism did not significantly change the total energy dissipation of the finger joint.

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7

The lightweight Delft Cylinder Hand, the first multi-articulating hand that meets the basic user requirements

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Abstract

Problem: Although body-powered (BP) hooks are used by many amputees, BP hands show the greatest user rejection rate of all available upper limb prostheses. BP hands require a high user operation force, produce a low pinch force and have rigid fingers. Furthermore they are heavy, which users indicate as being one of the first priorities in prosthesis improvement. There have not been any improvements reported for many years. *Goal:* To develop and evaluate a lightweight prosthetic hand with articulating fingers and body-powered control, that meets the basic user demands. The hand should require less actuation energy than current body-powered hands (1058-2292 Nmm) and it should produce sufficient pinch force (>30 N) for a broad functional range. *Results:* This study presents the Delft Cylinder Hand, an underactuated body-powered hand with articulating fingers which adapt to the grasped object. Through the application of a hydraulic cylinder transmission, the hand requires 49-162% less energy from the user than commercially available body-powered hands and it has a higher maximum pinch force (30-60 N). In functional tests (Box and Block Test and Nine Hole Peg Test) the hand scored similar to current myoelectric hands. Yet its mass (152 gram without a glove; 217 gram with a glove) is 68% lower than the lightest available articulating myoelectric hand and 55% less than the lightest BP hand of similar size. *Conclusion:* The Delft Cylinder Hand provides the user with an adaptive grip, a lower operating effort and a higher pinch force than current body-powered hands, which is a significant improvement in body-powered prosthetics. Its very low mass, its anthropomorphic shape and kinematics, high functional scores, proprioceptive feedback and fast control make this hand prototype the first prosthetic hand that meets the basic user requirements in one device.

7.1 Introduction

Replacement of a missing hand by an artificial alternative remains one of the biggest challenges in prosthetic rehabilitation [1]. Although many different prosthetic terminal upper limb devices are available [2-4], around 27% of the amputees does not actively use its device and another 20% stops wearing it at all [5]. There are various reasons for prosthesis abandonment; wearing discomfort (too heavy, too hot), too little added functionality, difficult or tiring to use, or the lack of sensory feedback [5]. The most important design priority is the device mass, which should be significantly reduced [6]. A hand prosthesis should at least meet the basic user demands, summarized by the words comfort, cosmetics and control; 'a patient wants and expects a prosthesis that looks naturally beautiful, that is comfortable to wear and that is easy to use' [7].

Currently two types of active prostheses are available; myoelectric and body-powered (BP) prostheses. In recent years new myoelectric hands have become available: the i-limb (www.touchbionics.com), the bebionic (www. bebionic.com), and the Michelangelo hand (www.ottobock.com). These hands provide a set of different grip patterns. Unlike other prosthetic hands, the fingers of the i-limb and the bebionic can articulate to provide an adaptive grip. Although these new hands provide more grip patterns, their additional functional value has not been proven yet [8], and the hands still remain too heavy (an average articulating hand weights ~550 gram, without battery and glove). A device mass of ~350 gram is already too heavy for amputees with a short residual arm [9].

The alternative to the myoelectric device is the BP hand or hook. The reported rejection rates of BP hands (65%-80%) are significantly higher than that of BP hooks (32-51%), or electric hands (17-41%) [6, 10]. A possible explanation of this low acceptance can be found in the limitations of the currently available BP hands. Although they have a lower mass (~ 350 gram) than electric hand they are still heavy. They produce a low pinch force (~15 N) and require a high actuation force (61-131 N), due to their low mechanical efficiency [11, 12]. Furthermore their shape is not closely anthropomorphic and they have rigid fingers, which do not adapt to the shape of the grasped object. This reduces the contact area between hand and object and limits the grasping functionality [13]. Despite these drawbacks, no improvements have been made to BP hands for decades [11, 12]. If these matters could be resolved, a BP hand could offer several benefits over a myoelectric hand; the harness system of a BP hand offers proprioceptive force and position feedback to the user. Also a BP hand could be considerably lighter than a myoelectric hand, as no motors and batteries are required.

To increase the mechanical efficiency of a mechanism, mechanical linkages can be replaced by hydraulics. Hydraulics potentially offers an efficient way of energy transmission, as the friction losses in hydraulic systems are typically low. There have been various attempts to use hydraulics in upper limb prosthetics, in an effort to develop an efficient externally powered or body-powered terminal device. Several prototypes have used hydraulics to operate an externally powered prosthetic hand, e.g. Janovsky *et al.* [14], Broome *et al.* [15], Witte [16], Tobergte [17], Kato [18], Kargov *et al.* [19]. In other prototypes Goller *et al.* [20] and LeBlanc *et al.* [21] used a hydraulic transmission to operate a BP hook. There are no examples of the application of a hydraulic transmission in a BP hand in literature, nor in any commercially available BP or electric hand.

7.2 Problem

Body-powered hands have the highest user rejection rate of all available terminal devices. They are heavy, require a high operation force from the user and then produce a low pinch force, due to mechanical inefficiencies. Furthermore their shape is not closely anthropomorphic and they have rigid fingers, which limits their grasping functionality.

7.3 Goal

To develop and evaluate a lightweight prosthetic hand with articulating fingers and body-powered control, that meets the basic user demands. The hand should require less actuation energy than current body-powered hands (1058-2292 Nmm) and it should produce sufficient pinch force (>30 N) for a broad functional range. The hand will be evaluated by quantifying its mechanical performance by mechanical tests and its functional performance by functional user tests. The results will be compared to other prostheses described in literature.

7.4 Methods

Requirements and design principles

A new lightweight articulating hand was designed, which has voluntary closing body-powered control. The requirements to which the hand had to comply are listed below. Various design principles were used to meet these requirements:

Cosmetic appearance

The hand mechanism should fit inside a standard 7 ½” size anthropomorphic silicone cosmetic glove, e.g. the SGL5/E4 glove (RSL Steeper) or the 101L M2 Male glove (Regal Prosthesis Ltd.). Preferable a silicone glove should be used, as they are more compliant and have a lower hysteresis than PVC gloves [22]. In order to give the hand an anthropomorphic appearance, the shape of the mechanism should be adapted to the shape of the glove, instead of the other way around.

Mass

In order to increase wearing comfort, the mass of the hand should be lowered [6]. Mechanism and glove should weigh less than a human hand (426 ± 63 g [23]), as the hand is not directly attached to the musculoskeletal system of the user and is therefore perceived as an external load. As a cosmetic glove weighs around 90 g [12] and one standard deviation is 63 g below the average mass of a human hand, the mass of the hand mechanism should be lower than 426 g - (63 +90)g = 273 g.

Actuation energy and pinch force

The energy dissipation by the transmission should be minimized, to reduce the required input effort by the user and to produce a high pinch force (> 30 N) to enable a broad range of activities [24]. A hydraulic system can transmit and distribute the energy and forces in an efficient way to multiple actuators, without the need of a whiffletree like mechanism with extra joints, with pulley cables or a linkage system. In this way extra friction can be avoided. Although the hand will have more DoF's than current BP hands, it should require less input energy than current hands (range of current hands: 1058-2292 Nmm [11]), to enable easy and comfortable operation.

Control of the hand

Control of the prosthetic hand should be fast, easy and intuitive. This can potentially be achieved by body-powered (BP) control. BP-control provides the user with direct proprioceptive force and position feedback, which enables precise and fast control of force as well as hand opening. This is a benefit over myoelectric control, which does not provide proprioceptive feedback. Furthermore BP- control requires no batteries and motors, which makes BP controlled prostheses potentially lighter. There are two modes of body-powered control, voluntary closing and voluntary opening control. For the new hand voluntary closing is preferred, as this provides the user with a direct and intuitive relation between actuation force and pinch force [25-27].

Articulating fingers

To enable grasping and holding of a large variety of objects, the hand should be able to perform the two basic grip patterns: the precision or pinch grip, and the cylinder or power grip [28]. The pinch grip can be used for picking up small objects by means of force closure. The cylinder grip can be used to pick up larger objects, by means of force and form closure (the fingers wrap around the object). A pinch grip can be achieved with rigid fingers with one degree of freedom (DoF). To enable a power grip the fingers should have at least two DoF's, one in the MCP and one in the PIP joint. It is not necessary to have an extra DoF in the DIP joint, this joint can be fixed at an angle of 15°. This is the same angle that is used in case of arthrodesis surgery of the DIP joint [29]. As there is only one control signal available, to control the multiple DoF's of the hand, the hand is by definition underactuated [30, 31]. To obtain stable grasping in both pinch and power grasp tasks, without ejecting the grasped object, the torque ratio between MCP and PIP joint should be around 0.5 [32]. The thumb of the hand should be opposable to the fingers, to enable a pinch grip with one or two fingers. This could be done passively. The opening of the hand should be at least 70 mm, to enable grasping of a broad range of objects.

Environmental influences

The hand should be able to function in wet and dirty conditions, to enable a broad application and to guarantee a high reliability. Therefore no electronic components should be used, and the selected materials should be corrosion resistant.

Modular system

The hand should be modular to enable easy maintenance and replacements of parts or modules by the prosthetist (e.g. one finger or one actuator). Modularity also provides the prosthetist with more freedom to customize the hand to an individual patient. The number of parts that can only be used for either a left or a right hand should be minimized [33].

Hand prototype evaluation

The mechanical performance of the constructed hand was evaluated in mechanical tests. The functional performance was assessed by functional user tests. Both tests used quantitative outcomes, to enable comparison to current available prosthetic devices and to past and future prototypes.

Mechanical evaluation

A mechanical test bench measured the energy required to operate the hand and the energy that was dissipated by the hand. A load cell (Zemic: FLB3G-C3-50kg-6B) measured the force that was required to operate the master cylinder. A displacement sensor (Schaevitz: LCIT 2000) measured the actuation displacements of the master cylinder. The required energy and the dissipated energy could be derived by integrating the measured forces along the measured displacements. A custom built pinch force load cell (thickness=10 mm) was used to measure the pinch force.

Protocol for mechanical evaluation

In the mechanical evaluation the hand was subjected to four different test protocols:

1. A full closing and opening cycle, without pinching (measured four times)
2. A full cycle of closing, opening and pinching an object (thickness=10 mm) with a pinch force of 15 N (measured four times).
3. Closing and pinching the pinch load cell (thickness=10 mm), until an actuation force of 100 N was reached.
4. Endurance was tested by closing, pinching 15 N and re-opening of the hand 100,000 times. Two sensors detected whether the index finger of the hand had reached sufficient pinch force and had fully opened during each cycle. When these criteria were not reached, the cycle discontinued and the setup was checked prior to continuation.

During test 1 and 2, the required input energy (work) and the energy dissipation (system hysteresis), which is a measure of the energy efficiency of the hand, were recorded and were compared to that of current hands. In test 3 the pinch force is measured as function of the actuation force. The characteristic was compared to that of other hands. The actuation force at which the pinch force starts building up should be as low as possible. Finally the hand prototype should pass test 4, the endurance test, without the need of major repair.

User evaluation

The hand was tested by able-bodied right-handed male subjects (n=13, average age 27±1 years). Able-bodied subjects were chosen, instead of amputees, as using amputee subjects would impose an unnecessary burden to the small group of available subjects. Ethical approval for this study was obtained from the institutional ethical committee (file number: # 1-4-2012). The Delft Cylinder Hand was attached at the palmar side of the left hand (Figure 7.1), using a Pro Cuff (TRS inc.). A second Pro Cuff was used to create an attachment point for the shoulder harness. To evaluate the functionality of the hand, two tests were selected, the Box and Block Test (BBT) [34, 35] and the Nine Hole Peg Test (NHPT) [36]. Both tests are commonly used to assess hand function, e.g. in patients with stroke [37, 38] or Parkinson disease [39, 40]. There are several studies in the field of upper limb prosthetics which used the BBT to measure functional outcomes [41-44]. Miller *et al.* identified the BBT as a promising functional evaluation test for upper limb prosthetics [45]. Until now the application of the NHPT for the evaluation of terminal devices is rare [46]. This test was included as it requires more fine motor skills than the BBT.

Both tests can be performed within a few minutes, so multiple trials can be performed in one session. This is a benefit over other test instruments, like the Southampton Hand Assessment Procedure (SHAP) [47, 48], the Jebsen Hand Function Test (JHFT) [49] and the Sollerman Hand Function Test (SHFT) [50]. As these tests take a much longer time to administer (around 15-30 minutes), they are less suitable for many repetitions in a cohort study [51]. The outcomes of the BBT were compared to the outcomes published by Farrell and Weir [43] and the outcomes found by Resnik and Borgia [42]. The results of the NHPT were compared to outcomes found by Schabowsky *et al.* [46]. A commercially available BBT was used and a home-made NHPT (holes: Ø10 x 15 mm with a 32 mm centre distance, pegs: Ø9 x 32 mm).

Protocol user evaluation

The subjects performed the tests in three sessions at three different days. Each session consisted of five trials of the BBT, followed by five trials of the NHPT. As the BBT is a simpler task and it takes less time and effort from the user than the NHPT, the BBT was performed first. In this way the mutual influence of the tests was minimized. It was not necessary to randomize the order of the tests, as the tests were not mutually compared. Both tests were executed according to the test instructions of the BBT [34] and the NHPT [36]. Before the test started the test procedure was explained to the subject and an informed consent from the subject was acquired. A basic instruction on how to operate the hand was given to the subject. The subjects were given a minute to get used to operating the hand before the session started.

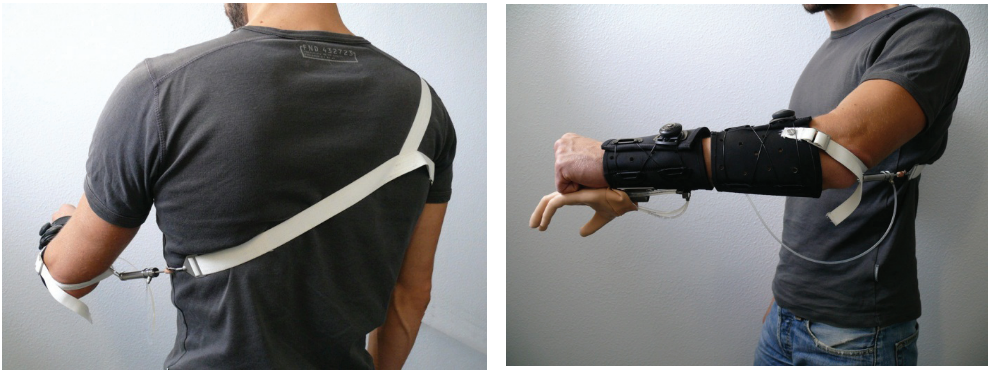

Figure 7.1 Two pro-cuffs (TRS inc.) were used to attach the Delft Cylinder Hand to a healthy test subject. The hand can be closed by pulling the white shoulder strap, which is attached to the master cylinder.

Amputee opinions

As a final part of the user evaluation three transradial amputee subjects were allowed to use the Delft Cylinder Hand for several minutes. They were asked to give their comments on the hand. The first subject was a male who was a regular user of a myoelectric prosthesis, the second was a male who was a regular user of both myoelectric and body-powered hands, the third was a female subject who abandoned the use of a prosthesis, as she considered it to be to heavy and of little use.

7.5 Results

Hand prototype

The fingers of the hand use 7 active DoF's that are actuated by miniature hydraulic slave cylinders (S1-7, Figure 7.2, Figure 7.3).The cylinders were specially designed for the hand, to enable a lightweight and compact design. Three cylinders (D=8 mm) are located inside the index, middle and ring finger. These cylinders control the PIP-joints. Three other cylinders (D=7 mm) control the MCP-joints. These cylinders are located inside the palm of the hand. A seventh cylinder (D=7 mm) controls the MCP- and PIP-joint of the little finger by a four bar mechanism. This cylinder is also located in the hand palm. The user controls the slave cylinders by pulling a shoulder strap (Figure 7.1), which is attached to a master cylinder (D=10 mm, M1 in Figure 7.3). The working principle of the hydraulic system (Figure 7.3) is analogous to the braking system of a car, in which the master cylinder of the brake pedal actuates the slave cylinders at the four wheels. When the user of the prosthesis pulls the shoulder strap, the master cylinder extends and actuates the slave cylinders at the fingers. The fingers of the hand then close. The cylinders of the fingers act like communicating vessels. Therefore the fluid pressure in the master cylinder and in each slave cylinder, are equal to each other at every moment. The pressure in the master cylinder also results in a reaction force at the shoulder harness. This provides the user with proprioceptive force feedback on the grip strength and it provides an indication of the object stiffness. No additional sensors are needed to control the grip force or the finger motion of this underactuated hand. During closing of the hand if one finger is blocked, the other fingers will continue closing. When there is an object present within the hand, the fingers will adapt to its shape. Depending on the shape of the object, the hand will adapt to a pinch grip or a power grip (Figure 7.4). The users can control the grip force and the degree of hand opening, by exerting an actuation force to the shoulder harness. The hand has a maximum opening span of 75 mm. Recordings with a high-speed camera showed that the hand closed within 250 ms. The hydraulic system could stand a system pressure up to 6 MPa (60 bar). This enables forces up to 300 N per cylinder and a pinch force exceeding 30 N per finger, or 60 N for a tripod grip.

In addition to the 7 active degrees of freedom, the hand has 7 passive degrees of freedom. The hand has a passively opposable thumb (1 DoF), which can be adjusted by using the sound hand. The thumb has no flexing joints. This makes it easier to align the prosthesis with an object, as the tip of the thumb will not move during hand closing. The fingers allow for +/-8° passive ad- and abduction (4 DoF's) at the Carpometacarpal (CMC) joints. This enables placement of objects between the fingers and it protects the fingers against side impact. The wrist has +/-10° passive flexion/extension (1 DoF), to reduce compensatory motion and to protect the fingers from palmar or dorsal impact. The hand can be connected to standard wrists, to allow for passive wrist pro- supination (1 DoF).

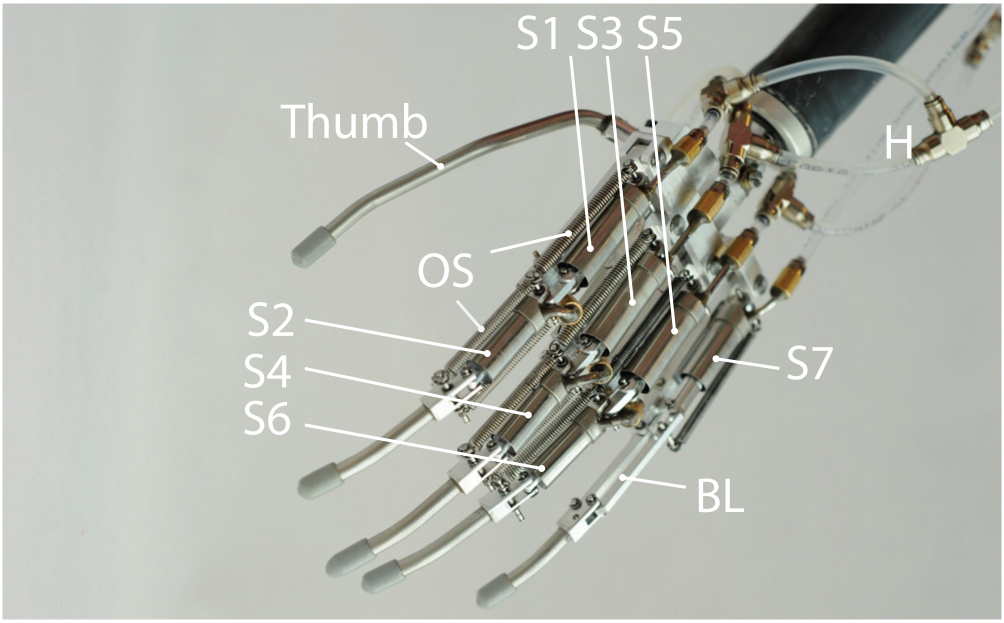

Figure 7.2 Hand frame of the Delft Cylinder Hand. The index, middle and ring finger have two slave cylinders each (S1-S6), the little finger is actuated by one slave cylinder (S7) combined with a four bar linkage (BL). The hoses (H) connect the slave cylinders to each other and to the master cylinder (see Figure 7.1), and distribute the hydraulic fluid among the slave cylinders. The fingers flex when the cylinders are actuated. After actuation the fingers are extend by the opening springs at the side of each finger (OS). The thumb is passively opposable.

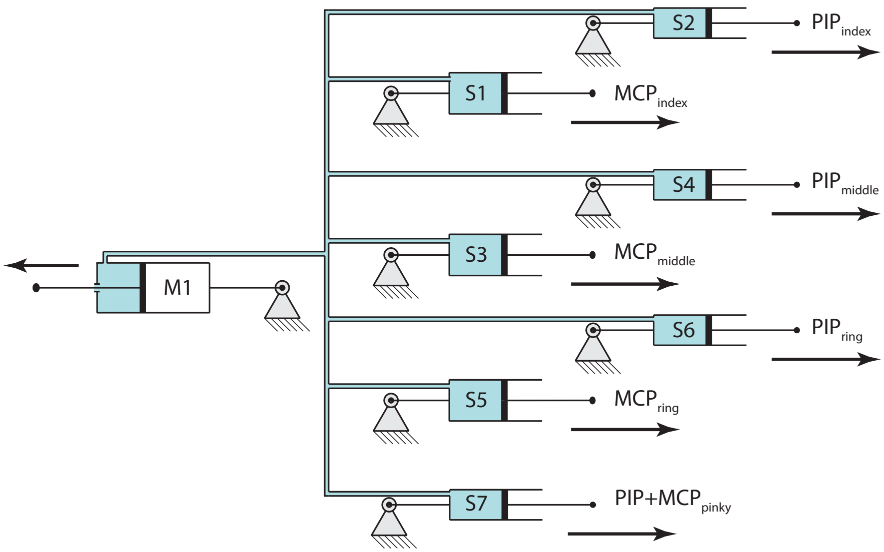

Figure 7.3 Schematic representation of the hydraulic system. When the master cylinder (M1) is pulled, the slave cylinders (S1-S7) will exert a force at the finger joints. The slave cylinders act like communicating vessels.

The mechanical as well as the hydraulic elements of the hand are fully modular. If instead of one input signal more signals would become available, the hydraulic system can be easily rearranged for multiple DoF control. Finger parts or complete fingers can be easily replaced or reconfigured. Reconfiguring the entire hand from a right-hand to a left-hand version, requires replacement of only two parts from the thumb.

Hand materials

The hand and finger frames were made out of aluminium Al7075 T6. The cylinders and axis were made out of stainless steel. After the aluminium was anodized the hand is completely water and dirt resistant. The hand can be covered with a standard sized 7½" cosmetic glove. The hand frame was covered with foam parts, to protect the cosmetic glove. The hand frame, including the protection foam, has a mass of 152 gram. The total hand including silicone glove has a mass of 217 gram. The master cylinder has a mass of 54 gram. Table 7.I shows an overview of the specifications of the Delft Cylinder Hand, together with the specifications of current available BP hands.

Figure 7.4 The underactuated hand adapts to the shape of the grasped object, without using any sensor and by using only one control signal. The hand can perform the two basics grasp patterns: precision grip and power grip. The hand can pick up small objects with the precision grip (top left and middle). The feedback allows precision tasks, like handling a pair of tweezers (top right). The hand can hold cylindrical objects with the power grip (below left). The little finger can be used to support the object, e.g. a cup (below middle). The power grip can also be used to carry a load in a horizontal position, e.g. the handlebar of a suitcase (below right).

Table 7.I Specifications of the Delft Cylinder Hand, compared to current available BP hands.

	Current BP hands	**Delft Cylinder Hand**
Cosmetic appearance	Cosmetic glove adapted to the mechanism	Mechanism adapted to an anthropomorphic cosmetic glove (size: 7 ½”)
Mass entire hand	~ 350 g, excl. cosmetic glove ~ 423 g, incl. cosmetic glove	152 g, excl. cosmetic glove 217 g, incl. cosmetic glove
Energy transmission	Bowden cable and mechanic linkages	Hydraulic
Actuation energy	~1600-1700 Nmm	828 Nmm
Maximum pinch force	~ 15 N precision grip (or three-jaw chuck)	30 N one finger 60 N precision grip (or three-jaw chuck)
Control	Shoulder control	Shoulder control
DoF’s	1 in the entire hand	2 active per finger 1 active in the little finger 1 passive per finger 1 passive for the thumb
DoF control in hand	1 signal -> 1 DoF (fully actuated)	1 signal -> 7 DoF’s (underactuated)
Basic grasp patterns	Pinch/Tip grip	Pinch/Tip grip Power grip
Other grasp patterns	Tripod grip	Hook grip Spherical grip Extension/Palmar grip Tripod grip
Maximum hand opening	~70 mm	75 mm

Hand evaluation

Mechanical evaluation

The mechanical performance of the hand was quantified by four different test. The results of the tests were compared to data of current BP hands.

1. In the first test the actuation force and displacements were measured, while the hand opened and closed without pinching. Figure 7.5 shows an example of a force displacement curve. One cycle is shown, as the deviations between the four cycles were small. For one cycle the hand (with glove and master cylinder) required 885±3 Nmm (or mJ) of energy. This energy is represented by the surface between the upper line of the graph and the x-axis, see Figure 7.5. The hand dissipated 642±3 Nmm (or mJ) of energy, which is represented by the surface enclosed by the curve.

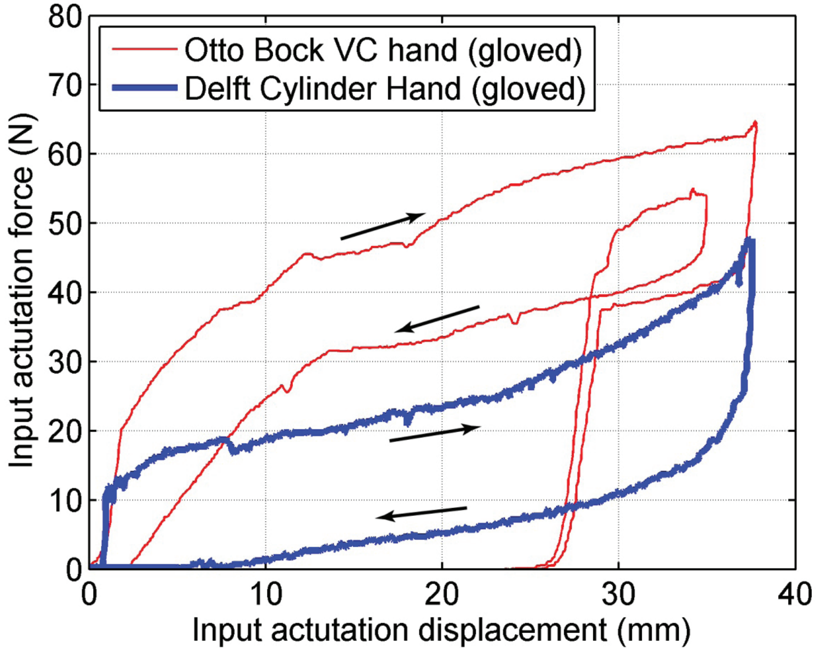

Figure 7.5 Force displacement curve of the closing test, representing a full cycle of closing and re-opening of the Delft Cylinder Hand (thick blue line). For clarity only one cycle is shown, as the four cycles were very similar. As a reference the curve of the Otto Bock VC hand (thin red line) is shown, adapted from [11]. Note the difference in force level between both hands, and note the small extra loop of the Otto Bock hand, due to its automatic locking mechanism.

2. In the second test the actuation force and displacement were measured, during a full cycle of closing, opening and pinching with a force of 15 N. During one cycle the hand required 828±3 Nmm (or mJ) of energy. Of this energy 546±3 Nmm was dissipated by the mechanism and the cosmetic glove.
3. During the third test the pinch force and the actuation force were measured. Figure 7.6 shows the pinch force a as a function of an actuation force increasing from 0 to 100 N. At an actuation force of 20 N the fingers touched the 10 mm thick load cell and commenced pinching. When the actuation force increased further the pinch force increased linearly. The slope of the linear line is determined by the transmission ratio of the system.

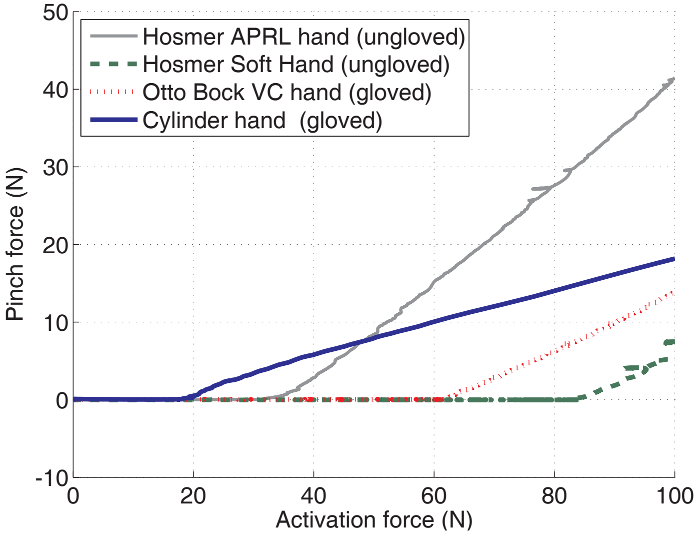

Figure 7.6 The force displacement curve of the 100 N actuation test. The graph displays the characteristic of the Delft Cylinder Hand together with the diagrams of the ungloved APRL and Hosmer Soft hand and a gloved Otto Bock hand. (Adapted from [11]). The Delft Cylinder hand requires a lower actuation force (20 N) to start building up pinch force, than the other hands. Its slope can be changed by using another master cylinder, whereas the other hands have a fixed slope.

4. In the endurance test the hand had to close, pinch 15 N and re-open 100,000 times. During the test, the apparatus paused several times, each time because one of the endpoint conditions was not met. After 33,493 cycles the MCP-springs had to be replaced, when the end loops of the MCP-springs were broken. After 54,696 cycles two MCP-springs had a broken end loop. The MCP-springs were therefore replaced by springs with a thicker wire diameter. After 76,695 cycles the bearings and the O-rings had to be re-lubricated. After 80,330 cycles the hydraulic system had to be refilled. At the end of the test the silicone cosmetic glove was worn out. The test was finished after 100,000 cycles without the failure of parts of the hand frame and without noticeable fluid leakage. No major repairs were required.

User evaluation

Box and block test

The number of blocks transferred increased over the length of the trials (Figure 7.7), from 17±6 blocks in the first trial to 26±8 blocks in the 15th trial.

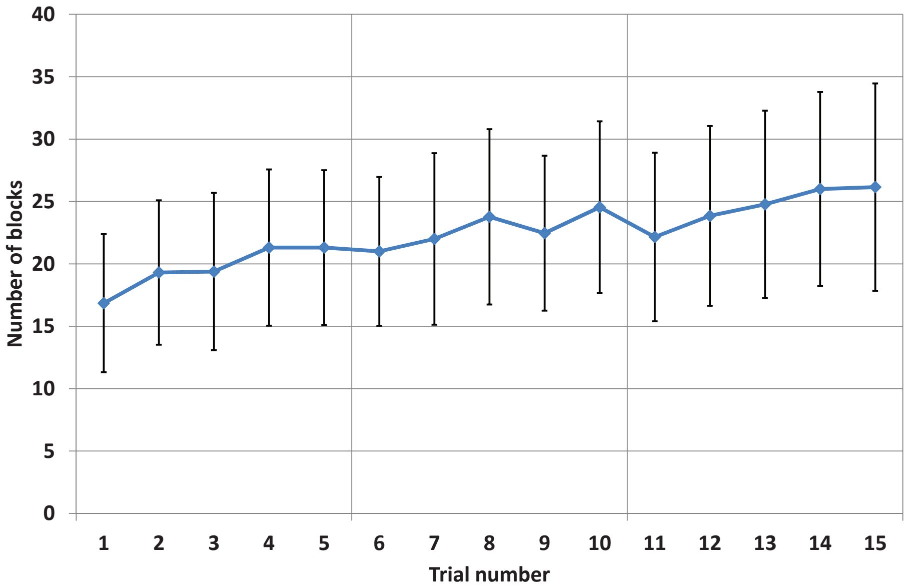

Figure 7.7 Scores of the Box and Block Test (BBT). The test was performed in three sessions of five trials at three different days (N=13, SD over 13 subjects).

NHPT

The required time to complete the NHPT decreased over the trials (Figure 7.8), from 117±41 s in the first trial to 62±11 s in the 15th trial.

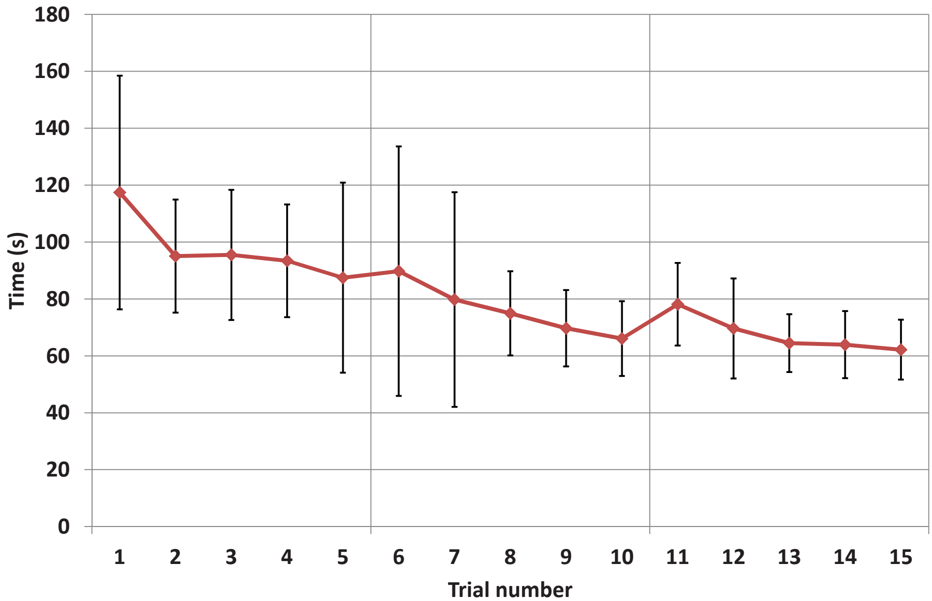

Figure 7.8 Scores of the Nine Hole Peg Test (NHPT). The test was performed in three sessions of five trials at three different days (N=13, SD over 13 subjects).

Amputee opinions

The amputee subjects that tried the prosthesis were all enthusiast about the hand and commented that it was easy to operate and that is was remarkably light, in comparison to their current and past prostheses.

7.6 Discussion

Mechanical evaluation

Actuation energy

Although the Delft Cylinder Hand has more joints and a larger range of motion, it required less input energy for closing (885±3 Nmm) than other body-powered VC hands, e.g. the Otto Bock VC hand (1639±24 Nmm), the APRL hand (1058±4 Nmm) and the Hosmer Soft VC hand (2292±12 Nmm) [11]. This means that the user has to deliver less input energy to operate the hand. Grasping and pinching an object with 15 N required less energy (828±3 Nmm) than fully closing all fingers, because pinching an object requires a smaller joint motion of the index and the middle finger. The required energy for closing and pinching was less than that of the Otto Bock VC hand (1694±16 Nmm), and the Hosmer Soft VC hand (2176±16 Nmm) and it was similar to the APRL hand (831±1 Nmm) [11]. It should be noted that the APRL hand, and the other hands, were tested without a Bowden cable transmission and without a cosmetic glove, which both will require considerable (10-40%) extra input energy. All current BP prosthesis use a Bowden-cable transmission, to transmit the force form the harness to the hand. The Delft Cylinder Hand uses a hydraulic transmission, instead of a Bowden-cable. The hand was tested with a hydraulic transmission and with a cosmetic glove applied.

Actuation force

The Delft Cylinder Hand required an actuation force of only 20 N to grasp a 10 mm thick object and start pinching. This actuation force is 15-65 N lower than that of other body-powered VC hands (Figure 7.6). The Delft Cylinder Hand has a flatter slope than the other hands, because of its lower transmission ratio. This ratio can however easily be changed, by using a master cylinder with a smaller or larger diameter. A cylinder with a smaller diameter will increase the slope. In this way the transmission ratio can be adapted to individual preferences. Another option is to add a dual phase mechanism which can switch between two transmission ratios [52, 53]. This principle can relatively easy be implemented in hydraulics [54], to increase the pinch force at a low actuation force. Its low required input energy and its low required actuation force are a major benefit over current hands, and make the Delft Cylinder Hand also suitable for people with less body strength.

Endurance

The hand passed the endurance test as it did not require major repair and showed no hydraulic leakage. A user who uses his/her hand prosthesis on average 274 times a day, would have to use the prosthesis for one year to reach 100,000 cycles. The hand required only small maintenance during the test. At 75% of the test the hand had to be re-lubricated and at 80% of the test the hydraulics had to be refilled. Although there was no fluid leakage, the fluid could slowly disappear from the system, as during every stroke some fluid could evaporate from the wetted cylinder surface. When 100,000 cycles equal one year, 75,000-80,000 cycles would correspond to 8-9 months. This is an acceptable interval for small maintenance, as the small maintenance can easily be done by the prosthetist when the cosmetic glove is replaced (typically every 3-6 months). Furthermore the test revealed the sensitivity to fatigue of the end loops of the MCP-springs. This can be solved by replacing the springs by springs with a larger wire diameter, as was done during the test, or by using a helical spring with an alternative and stronger end loop.

User evaluation

In the Box and Block Test as well as in the Nine Hole Peg Test the users were able to use their prosthesis directly after donning and they all were able to complete the tests on their first trial. The subjects had been given only a basic instruction and a minute to get used to the prosthetic hand. This illustrates that the control is intuitive and easy to learn. A clear trend can be observed in both tests (Figure 7.7, Figure 7.8). Over the trials the performance of the subjects increased. At the start of each new session the users performed less well than at the end of the previous session. During each session the performance increased gradually.

The results of the functional tests were compared to BBT-scores found in literature. Farrell and Weir [43] found BBT-scores of 25±3.5 blocks for slow devices and 28±4 blocks for fast devices, in a study with able-bodied subjects (n=20) using myoelectric terminal devices. Resnik and Borgia [42] found BBT-scores of 19.9±10.0 blocks for transradial amputees (n=26), who were full-time or part-time users of myoelectric or body-powered terminal devices. The scores of these studies are similar to the BBT-scores found in our study (26±8 blocks, during the 15th trial). It should however be noted that the scores of the Delft Cylinder Hand might still improve after the 15th trial. Furthermore the tests were performed by right-handed subjects, using a left-hand prosthesis. The tests results might improve when the tests would be performed by left-handed subjects, or by using a right-hand version for right-handed subjects.

The results of the NHPT illustrate the capability to perform fine motor tasks using the Delft Cylinder Hand. The values can serve as a quantitative measure for comparison to other prosthetic hands. In a study by Schabowsky *et al.* [46] skilled trans-radial amputees (n=6, mean age: 50.2±14 years) finished the NHPT in 76±29 s, by using their body-powered hook. This reference data suggests that the functional performance of Delft Cylinder Hand in the NHPT (62±11 s) was at least similar to the score of a BP-hook. This is a very interesting result, given the fact that the BP-hook is considered to be still one of the most functional terminal devices. More reference data is needed to support these findings, as Schabowsky *et al.* only measured six subjects.

Evaluation requirements

Cosmetic appearance

The hand mechanism fits inside a standard 7 ½" size anthropomorphic cosmetic glove, made of Silicone or PVC. These gloves, which are normally used for passive cosmetic hands, are moulded copies of a human hand. As the shape of the mechanism was adapted to the shape of the glove, the hand has the exact anthropomorphic dimensions of a human hand.

Mass

According to the study by Biddiss *et al.* the development of a more light-weight comfortable prosthesis is the most important design priority [6]. Other studies also report the importance of weight reduction in hand prostheses [55-57]. Fishman and Berger [9] already reported in 1955 that the APRL hand (347 gram, without a glove) was too heavy to be fitted to amputees with short or weak residuals. Despite these reports, the most recently developed hands are considerably heavier than the APRL hand; i-limb ultra (469 gram, without a glove), bebionic v3 (550 gram, without a glove) and Michelangelo (600 gram, without a glove). Figure 7.9 shows the mass of current commercially available prosthetic hands (347-600 gram, without a glove), compared to the mass of the Delft Cylinder Hand (152 gram, without glove). The Delft Cylinder Hand is 55% lighter than the lightest current VC hand and 68% lighter than the lightest current articulating hand. The low mass of the Delft Cylinder hand has the potential to improve wearing comfort and will enable more amputees to wear a prosthetic hand, e.g. amputees with a short or weak residual arm.

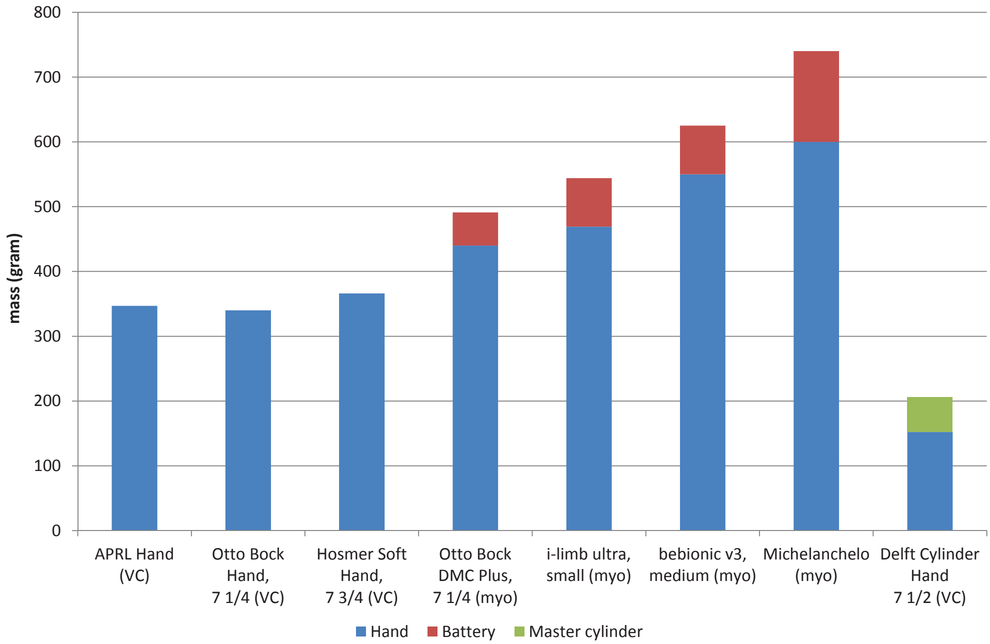

Figure 7.9 The mass of the Delft Cylinder Hand (right) compared to current state-of-the-art commercially available body-powered voluntary closing hands (VC) and myoelectric hands (myo). (All hand masses are without the cosmetic glove). The figure also shows the battery masses of the myoelectric hands.

Actuation energy and pinch force

The application of a hydraulic transmission enabled an efficient transmission and distribution of the actuation energy to the multiple finger joints, without additional friction forces. In this way the force feedback information is optimized and the input force is reduced. This reduces harness discomfort, which is one of the main problems associated with body-powered operation [6]. The elastic part of the required input energy was minimized, by making the system stiff, by choosing springs that were not stronger than necessary and by using a silicone glove, instead one made of PVC [22]. The resulting part of the required input energy consists mainly of hysteresis. The hysteresis was reduced by minimizing the bearing and the sealing friction. The Delft Cylinder Hand, which has 7 DoF's, required less energy than current body powered hands which have only one DoF. The hand could deliver a pinch force of over 30 N per finger, and 60 N for a tripod grip, which exceeds the requirements.

Control of the hand

The application of body-powered control in the Delft Cylinder hand, offers the user a prosthesis that has a low mass, provides force and position feedback, and that can be operated without unwanted hand movements, at a high speed and without additional noise. This is an advantage over myoelectric prostheses. Beside a high mass, users of electric devices report a lack of feedback [5, 56, 57], unwanted hand movement [56], a slow closing speed [55-57] and a disturbing motor noise [56, 57]. By using body-powered actuation these problems were avoided. The shoulder strap of the Delft Cylinder Hand offers intuitive force and position feedback to the user, based on the principle of Extended Physiological Proprioception (EPP)[58]. It enables easy and accurate control of the hand and avoids undesired opening or closing of the hand. The entire hand can be fully closed within <250 ms, which approaches the speed of a human hand (~100 ms). Electric hands require a minimum closing time ranging from 333 ms (Vari Plus Speed [59]) to 1200 ms (i-limb ultra [60]) and they require an additional delay ranging from 50 ms to over 300 ms [43], to process the myoelectric signal in the microcontroller. Unlike electric hands the Delft Cylinder Hand is completely silent, which avoids undesired attention.

Articulating fingers

The hand has articulating fingers, which can perform both pinch and power grasp. The underactuated fingers adapt to the shape of the grasped object. This enables stable grasping of a broad range of objects, without the need of extra sensors.

Environmental influences

The hand is corrosion resistant and does not use electronic parts. The hand can function in wet and dirty environment, which enables a broad application of the hand.

Modular system

The hand is fully modular. The finger parts or complete fingers can be easily replaced or reconfigured. All parts fit inside the hand, making the Delft Cylinder Hand also applicable for amputees with very long arm residuals or a transcarpal hand amputation. The transmission-ratio can be adapted to the individual patient by changing the master cylinder. These factors give the prosthetist the ability to adapt the hand to many different individual users with totally different needs. For example: people with very high amputations or very long residua, or people who have a low strength (including elderly users).

Clinical implications

The current prosthesis options do not meet all basic user demands, cosmetics, control and comfort, simultaneously. BP hooks are fast, accurate, and they have a low mass. However they have a poor cosmetic appearance. BP hands have a reasonable appearance, however they are uncomfortable to operate and have a poor function. The myoelectric hands combine function and a reasonable appearance. However they are slow, noisy, they lack force feedback and according to many users they are too heavy to be worn comfortably. The Delft Cylinder Hands is the first hand that combines the basic user demands in one device. The hand has a cosmetic anthropomorphic appearance, static as well as kinematic. It could be easily controlled with little familiarity and it performed at least similar to electric devices and BP-hooks. The hand is fast and accurate, due to the proprioceptive feedback. It is comfortable to operate and to wear, due to the low actuation force and its very low mass. The adaptive fingers enable the hand to perform the basic grasps, enabling the user to use the hand in many different activities of daily living. Combining the basic user demands in one device, makes the Delft Cylinder Hand a promising alternative to current available devices.

Study strengths and limitations

A strength of this study is the quantitative functional evaluation of the prototype. Most studies that describe prosthesis prototypes do not provide quantitative outcomes, which makes it impossible to objectively compare them to other devices. A limitation of this study was that right-handed subjects were tested, using a left-hand device. This could have influenced the results negatively. Also the prosthesis was only tested inside the laboratory. Extensive clinical testing is required, to see how the hand performs in daily living situations outside the laboratory. Therefore as a next step the prosthesis will be provided to a group of amputees, who will be able to use the prosthesis at home. This test is planned to start within a year. The home trial might also find an answer to the practical questions, e.g. whether it is desirable to add a locking mechanism to the hand, which can be used for holding objects for a longer time.

7.7 Conclusions

This study presents the Delft Cylinder Hand, a prosthetic hand prototype that is anthropomorphic, slender, fast and silent. The hand meets one of the most important user design criteria, which is a low hand mass. Its mass (152 gram without glove; 217 gram with glove) is 68% lower than the lightest available articulating myoelectric hand and 55% lighter than the lightest BP hand of similar size. The hand has articulating fingers which fully adapt to the grasped object, by using the principle of underactuation. Its body-powered actuation provides the user with proprioceptive force and position feedback, enabling accurate and fast control, without the need of additional sensors. In functional tests (Box and Block Test and Nine Hole Peg Test) the hand showed scores that were at least similar to current body-powered hooks and myoelectric hands. Through the application of a hydraulic cylinder transmission, the hand required 49-162% less energy from the user than current body-powered hands and it had a higher maximum pinch force (30-60 N). Its very low mass, its anthropomorphic shape and kinematics, and high functional scores, make this hand prototype the first prosthetic hand that meets the basic user requirements.

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8

Discussion, Recommendations and Conclusions

8.1 Discussion

The goal of this thesis was to design and evaluate a new lightweight and efficient body powered hand prosthesis.

A new body powered hand prototype was developed, the 'Delft Cylinder Hand' (Chapter 7). Main features of the hand:

- Its articulating fingers, which enable different grasp patterns. Each finger of the hand has two articulating joints, enabling the fingers to adapt to the shape of the grasped object. The hand can perform the pinch or precision grip, to pick up small objects. The hand can perform the cylinder or power grip, to firmly grip large objects. These two basic grip patterns enable grasping and holding of a variety of grasped objects with different shapes.
- The use of the principle of underactuation enables the adaption of the fingers to the shape of the object and it enables distribution of the force over the fingers, without the need of any sensor.
- The use of voluntary closing body powered (BP) control. The voluntary closing control enables intuitive and accurate proportional control of the pinch force and hand opening. The BP control provides proprioceptive force and position feedback to the user, which enables fast and accurate force and position control. This is an important advantage over myoelectric control, which lacks any proprioceptive feedback. Clinical tests (Chapter 7) showed that the hand can be operated directly, without any training.
- Its very low mass. A second benefit of the use of BP control is that no heavy motors and batteries are needed. Therefore the hand mass is 68% lower than that of lightest myoelectric articulating hand and 55% lower than that of current BP hands (Chapter 2).
- Its high mechanical efficiency. In order to maximize the mechanical efficiency The Delft Cylinder Hand uses a hydraulic transmission (Chapter 4) in combination with a silicone cosmetic glove (Chapter 5). The hand requires 49-162% less energy than similar BP hands for similar tasks (Chapter 7). The hydraulic system replaces the Bowden cable of the current BP system, yielding an additional improvement in efficiency of 10-40% compared to the current BP systems which use a Bowden cable. The efficient transmission enables the user to control the hand at a lower and more comfortable actuation force level, and enables the user to exert a higher pinch force. The hand can pinch more than 30 N per finger, which is more than twice the pinch force of current BP hands.

The 'Delft Cylinder hand' provides the amputee a level of function that is comparable to current contemporary hands, at a cost (mass and actuation effort) which is much lower than that of current hands. This makes the hand an attractive alternative to current available hands. In the following paragraphs we will discuss the different steps which led to the development of the new prototype.

State of the art in Body Powered prostheses

As a first step the state of the art of current body powered devices was evaluated, in order to establish a start off for the new designed hand. User studies report that body powered hands are associated with high rejection rates [1, 2]. Although clear evidence was lacking , there were also indications that there has been very little development in the field of body powered prosthetics during the past decades [3]. There is very little quantitative data available on the performance on prosthetic devices, functional as well as mechanical. The most recent studies date from two decades ago [4, 5]. These studies reported the need for improvement of the body powered devices that were available at that time. Although the devices that are currently available seem very similar to those that were tested at that time, their mechanical performance could possibly have been improved, due to refinement of the internal mechanisms or by the use of improved materials. To test this currently available voluntary closing (VC) terminal devices were tested in the study described in Chapter 2. The results of this study showed that the tested VC hands had a very low mechanical performance. They required a high actuation force to produce only a low pinch force. Based on these findings the VC hands are expected to have a very limited functional value for their user. Because voluntary opening (VO) devices are perhaps even more commonly used than VC devices, a second study was performed to evaluate the mechanical performance of VO devices (see Chapter 3). There could be a relation between the mechanical performance of the VO devices and the fact that they are used by so many people. Therefore results of the study were compared to data of a study performed in 1987 [4]. The results of the study showed that the performance of the VO hands was poor and that their functional value to the user was expected to be limited. There was no significant improvement of the mechanical performance of the tested hands, compared to the data of the data of 1987. Main conclusions from these studies were that, in order to meet the user demands, future body powered hand designs should have a much lower mass than current hands and should be capable of producing a higher pinch force than current hands (>15 N) at a lower actuation force and energy.

An energy efficient transmission

To reduce the required actuation force and to increase the pinch force that can be generated, it is necessary to increase the efficiency of a prosthetic hand. The amount of actuation energy that a hand requires, is a measure of the efficiency or inefficiency of a hand. The more inefficient the hand is, the higher the required actuation energy. The two most important parts of the hand that contribute to the inefficiency of the hand, are the hand mechanism and the cosmetic glove which covers the mechanism. Each of the two parts acts like a combination of a spring which stores energy and a damper which dissipates energy. This behaviour is undesired, as both spring and damper require extra input effort from the user. In the first place a more efficient hand mechanism should be designed. The mechanism inside the hand transmits the input force from the user to the fingertips. The mechanism should be stiff and it should transmit the energy without dissipating much energy itself. An efficient energy transmission becomes even more important when the complexity of the mechanism increases, due to the addition of multiple finger joints. In Chapter 4 two different finger concepts were designed and evaluated. The first concept had a pulley cable transmission, the second had a hydraulic cylinder transmission. The test results showed that the hydraulic transmission required 35-74% less energy than the pulley transmission in various tests. Therefore the hydraulic transmission was considered to be the most efficient and the most suitable for application in the hand. Beside application in the hand, the hydraulic transmission can also be used to replace the Bowden-cable for the shoulder control. This will provide an extra increase in efficiency of the entire system of 10-40%, compared to current BP systems [6]. The test results showed that a hydraulic transmission can significantly increase the efficiency of a hand system, therefore the hydraulic transmission principle was chosen to be used in the new hand design.

An energy efficient glove

The second part that contributes to the inefficiency of a prosthetic hand is the cosmetic glove [7]. Due to their stiffness cosmetic gloves impose undesired forces to the mechanism. The deformation of the cosmetic glove material also dissipates energy. It would be ideal to develop a new glove, which has improved mechanical properties. However this did not fit inside the scope and timeframe of this study. An alternative to designing a new glove, would be selecting an existing glove with good mechanical properties. In literature however, data on cosmetic gloves is scarce. In the study described in Chapter 5 the stiffness and energy dissipation of two sets of gloves were measured. One set was made out of Silicone the other was made out of PVC. The gloves were made by the same manufacturer and had identical size and shape. The test results showed that the Silicone glove had a 2.5-4.5 lower stiffness than the PVC glove. It also required 1.8 to 3.8 times less actuation energy. Therefore it was decided to use a Silicone glove to cover the new hand prototype. The results also showed that the gloves had a linear positive stiffness, which makes it possible to compensate this undesired stiffness by a linear negative stiffness.

Chapter 6 describes the design and evaluation of a mechanism that has a linear negative stiffness. The mechanism was optimized to compensate the positive stiffness of a PVC and Silicone glove. The system evaluation showed that the compensation mechanism could reduce the required input force and torque for over 50%, during finger movement without pinching. When applied to the hand prototype, this mechanism can significantly reduce the actuation effort for the amputee, to close the hand and when applying low pinch forces (~15 N). For larger pinch forces the reduction of the actuation force will be relatively small, as in that situation the negative force exerted by the glove is relatively small compared to the actuation force required to deliver the high pinch force. A negative stiffness mechanism was applied in the hand. The spring configurations in the current hand prototype were not optimized to the current glove, as the design of the hand was finished before the study of the mechanism optimisation was finished. To maximize the energy efficiency, the hand uses a Silicone glove, instead of a PVC glove.

Hand prototype

After the boundary conditions and possible solutions were identified, a new hand prototype was designed and built, the Delft Cylinder Hand (Chapter 7). The specification of the hand are shown in Table 8.I.

Table 8.I Specifications of the Delft Cylinder hand prototype, compared to that of current BP hands and compared to the human hand.

	Current BP hands	Hand prototype	Human hand
Cosmetic appearance	Cosmetic glove adapted to the mechanism	Mechanism adapted to an anthropomorphic cosmetic glove (size: 7 ½”)	Anthropomorphic
Mass entire hand	~ 350 g, excl. cosmetic glove ~ 423 g, incl. cosmetic glove	152 g, excl. cosmetic glove 217 g, incl. cosmetic glove	426 ± 63 g [8]
Energy transmission	Bowden cable and mechanic linkages	Hydraulic	Tendons
Actuation energy	~1600-1700 Nmm	828 Nmm	n/a
Maximum pinch force	~ 15 N precision grip (or three-jaw chuck)	30 N one finger 60 N precision grip (or three-jaw chuck)	69 ± 16 N one finger [9] 92 ± 23 N precision grip (or three-jaw chuck)
Control	Shoulder control	Shoulder control	Direct muscle control
DoF’s	1 in the entire hand	2 active per finger 1 active in the little finger 1 passive per finger 1 passive for the thumb	4 active per finger 4 active for the thumb
DoF control in hand	1 signal -> 1 DoF (fully actuated)	1 signal -> 8 DoF’s (underactuated)	Multiple signals -> Multiple DoF’s (nearly fully actuated)
Basic grasp patterns	Pinch/Tip grip	Pinch/Tip grip Power grip	Pinch/Tip grip [10] Power grip
Other grasp patterns	Tripod grip	Hook grip Spherical grip Extension/Palmar grip Tripod grip	Hook grip [11] Spherical grip Extension/Palmar grip Tripod grip Lateral grip/Key grip

The designed hand prototype is a lightweight hand, which has articulating fingers and is capable of performing both power and pinch grip. The hand has an energy efficient hydraulic cylinder transmission, which requires a low energy input and energy dissipation. As a result the required actuation force is lower than in current body powered hands and it can produce a higher pinch force. The hand is controlled by voluntary closing body powered shoulder control. The hand can be connected to the shaft of any prosthesis, with little or no modifications required. It is fully modular and can also be used for partial amputations. Unlike most current prosthetic hands the Delft Cylinder Hand is not bigger than a human hand, instead it has a fully anthropomorphic shape. The specifications of the prototype (Table 8.I) show large improvements compared to current BP hands.

Comparison to other hands

In the past decades many different prosthetic hands have been developed. Some of them did become commercially available, most did not. Various other prototypes are still under development. It would be interesting to see to what extend the new hand prototype is an improvement, compared to what is already commercially available and compared to what has been developed up till now. A first objective comparison can be made by comparing the different parameters of the hand prototype, to that of other prosthetic hands. Important hand parameters include: hand mass, adaptivity of the fingers, energy efficiency, pinch force level, functional scores and finger speed. To enable a more systematic comparison the prototype will be compared to three different groups of prostheses: BP hands, hydraulic prostheses (BP and externally powered), and electric prostheses. The prototype will be compared to commercially available hands and to other prototypes described in literature.

Body Powered hands

Compared to the commercially available current BP hands, which are described in Chapter 2 and 3, the new hand prototype offers several major advantages. First of all the hand has a 55 % lower mass than the lightest BP hand of similar size. This is a large benefit, as mass reduction has been identified to be one of the most important design demands in upper limb prosthetics [12]. Furthermore the hand prototype has adaptive gripping and the hand can perform both power and pinch grasp. Current BP hands have stiff fingers and are designed for the pinch grasp. The only hand which has fingers with some adaptivity is the Becker hand [13]. The fingers of this hand can however not move independently and their range of motion comes mainly from the PIP joints. The MCP and PIP-joints of the fingers of the new hand prototype have a range of motion of 90° and their motion is uncoupled. This enables secure adaptive grasping and holding of a broad range of objects. Unlike the current BP hands, the new hand prototype has a slender anthropomorphic design. Current hands are voluminous, due to their internal mechanism. The most important advantage of the new hand, compared to current BP hands, is its high mechanical performance. The hand prototype requires less mechanical work to operate the hand and it dissipates less energy than current hands. The hand can already be closed at an actuation force of 20 N. This makes the hand useful to much more amputees than current hands, which require 35 to 85 N to close the hand. Furthermore the hand is capable of pinching 30 N per finger and 60 N with a precision grip. This is an important improvement compared to current hands, which have very limited function as they have difficulties to pinch 15 N with a precision grip. For the first time a BP hand prototype has been built and tested that is capable of pinching such a high pinch force and requires such a low effort from its user. The increase in mechanical performance together with the large mass reduction, make the Delft Cylinder Hand a very promising alternative to current available BP hands, and might help to significantly increase both the user satisfaction and acceptance of BP hands.

In literature there are only a few recent studies on the development of BP hands. One of the studies that can be found describes the development of an hand prototype with adaptive fingers, the IOWA hand [14]. This hand has helical springs inside the fingers and it was covered by a cosmetic glove. The hand was not fitted to a test subject. The authors suggested that future development of the hand would focus on making the hand externally powered. Two other studies describe the development of an articulating hand that can be controlled by shoulder actuation. Laliberté *et al*. [15] describe the development of an underactuated articulating hand that uses pulleys and tendons, in combination with a seesaw mechanism. The hand was fitted to one test subject, who could grasp and hold various objects. Tang *et al*. [16] describe an articulating hand prototype that is based on the TRS GRIP prehensor. The hand can be operated by using a shoulder harness. The authors did not describe if the hand was really fitted to a test subject. Although both studies presented a working prototype, no functional outcome measures have been presented yet. Both hand prototypes could not be covered by a cosmetic glove. Another hand prototype worth mentioning is the Stark Hand [17], developed by Mark Stark. This articulating BP hand is fast and lightweight, and is used on a daily basis by a friend of Stark. It can however not be covered by a cosmetic glove. We can therefore conclude that up to now no suitable alternative prosthetic hands have been presented, to replace current BP hands or to improve them.

Hydraulic Body Powered prostheses

Although there have been a number of projects focusing on the application of hydraulics in upper limb prosthetics, no hydraulic hand prostheses are commercially available. Most studies focused on the application of hydraulics in externally powered prostheses. There are a few studies available on the application of hydraulics in body powered prostheses. Studies of Goller *et al*. [18] and LeBlanc [19] proofed the feasibility of using a hydraulic transmission in BP prostheses. Both studies used a hook, instead of a hand. The prostheses were fitted to test subjects, however no quantitative functional scores were published.

Hydraulic externally powered prostheses

Of the studies on the application of hydraulics in externally powered prostheses, some focused only on the hydraulic transmission and did not include the terminal device, e.g. Janovsky and Merten [20], Witte [21]. Several other studies did include the terminal device e.g. the NEC hand [22], the Waseda Hand-9H3 [23], a hydraulic Edingbrugh arm by Marsh [24], the hydraulic hand by Lin and Lin [25, 26] the hydraulic hand model by Stanciu and Stanciu [27], and the Mesofluidic finger [28]. The prototypes in these studies reached various stages of development. Only the Waseda hand was fitted to a test subject [23]. None of the studies reported quantitative functional data. Some data on hand parameters, like mass and pinch force were reported. The hands were either heavy or had a low pinch force. Except for the NEC hand all prototypes could not be covered with a cosmetic glove.

A recent study in which a hydraulic prosthesis was developed which was covered by a cosmetic glove, is the development of the Fluid Hand III [29]. This externally powered adaptive hand uses hydraulic bellows, to actuate the joints. The hand has a lower mass (400 gram) than other externally powered hands and has adaptive gripping. The Fluid Hand III prototype was tested by amputees in clinical trials. However as no quantitative measurements have been published, it is not possible to make a functional comparison to our prototype. Disadvantages of the Fluid Hand III compared to our prototype are drawbacks associated with the use of external power, e.g. slower than body powered control, absence of proprioceptive feedback, relatively high mass. The maximum fingertip force of the Fluid Hand III is 45 N [29], which is a relatively high pinch force for a articulating hand. Its maximum pinch force is limited by the maximum system pressure, which was 9.8 bar in the Fluid Hand III. One of the limiting factors in increasing the maximum pressure is the use of bellow actuators, which have a lower strength than the steel cylinders used in our prototype. The hand prototype in our study has a maximum system pressure of 60 bar, which is currently only limited by the hoses. This allows for a pinch force of 30 N per finger and 60 N precision grip force. However at the current transmission ratio a precision grip force above 30 N requires an uncomfortably high actuation force of the user. To enable higher pinch forces at a comfortable actuation force level, the pinch force/actuation force ratio should be increased. This can be achieved by using a master cylinder with a smaller diameter, or by using a bi-phasic hydraulic mechanism. By replacing the current hoses by stronger ones the maximum system pressure can potentially be increased up to 150 bar, without requiring redesign of other components. This will enable a more than double increase of the maximum pinch force.

Externally powered prostheses

Recently new externally powered hands have become commercially available which have articulating fingers: the i-limb by Touch Bionics [30], the bebionic by RSL Steeper [31] and the Vincent hand [32]. Another new available hand is the Michelangelo by Otto Bock [33], which has a powerful pinch force but has no articulating finger joints. These hands are considered to be the current state of the art in upper limb prosthetics. The hands can be controlled by using myoelectric skin-electrodes. By producing Morse code like signals, the amputee can select different grasp modes. Although these hands provide various grasp modes and adaptive gripping, they still have the same drawbacks as conventional myoelectric hands. They are relatively slow, they do not provide proprioceptive feedback, and they are heavy. Most newer hands are heavier than their stiff-fingered predecessors. As our hand prototype is body powered it does not suffer from the problems associated with externally powered hands. It is fast and it enables accurate force and position control, due to the proprioceptive feedback. Most important, the hand prototype has a 68 % lower mass (152 gram, ex. glove) than the lightest articulating commercially available prosthetic hand of similar size (i-limb ultra, small, 469 gram, ex. glove and battery).

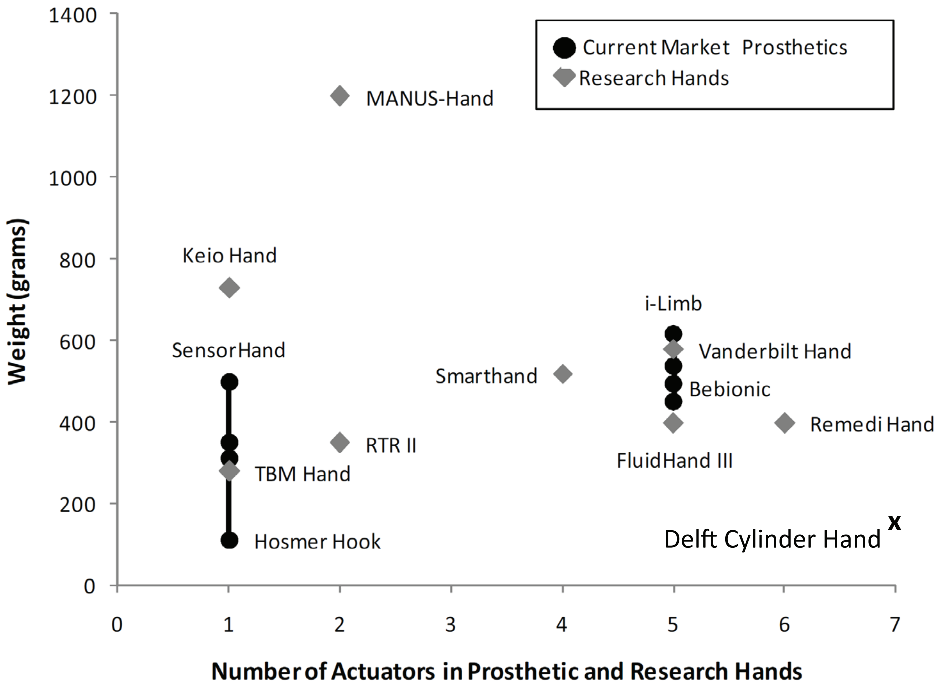

Figure 8.1 Distribution of hand weight compared to the number of actuators or actuated joints in the hand. The X marks the Delft Cylinder Hand prototype (adapted from [34]).

In literature various articulating hands can be found, which have been developed or are still under development, e.g. the Smarthand [35], UB hand 3 [36], UNB hand [37], MANUS hand [38]. An extensive overview has been given by Belter [34]. Figure 8.1, which is adapted from [34], shows that all current articulating hands have a significant higher mass than our prototype, despite the fact that they have less actuators. Most of the hands also have lower maximum pinch forces (<20 N). The MANUS hand has a high pinch force of 60 N, however this hand also has a very high mass (1200 gram) [38] as can be seen in Figure 8.2. This figure shows the hand mass as a function of the pinch or precision grip force. The green area seen in Figure 8.2 indicates the design space in which a prosthetic hand should be designed. It should have a mass of less than 273 gram (see Chapter 7), to be comfortable to wear and it should be able to pinch over 30 N (see Chapter 7), to enable grasping of a broad range of objects. Currently the Delft Cylinder Hand is the only hand which is within these boundary conditions. To make the Delft Cylinder Hand operable a master cylinder is required, with an additional mass of 54 gram. The other hands need batteries and electrodes, with an additional mass of 50-140 gram. The Delft Cylinder hand has a much lower mass than other articulating hands, because it uses BP actuation (similar to the Hosmer Hook in Figure 8.1). Furthermore the maximum pinch force of The Delft Cylinder Hand is higher than that of most other articulating hands.

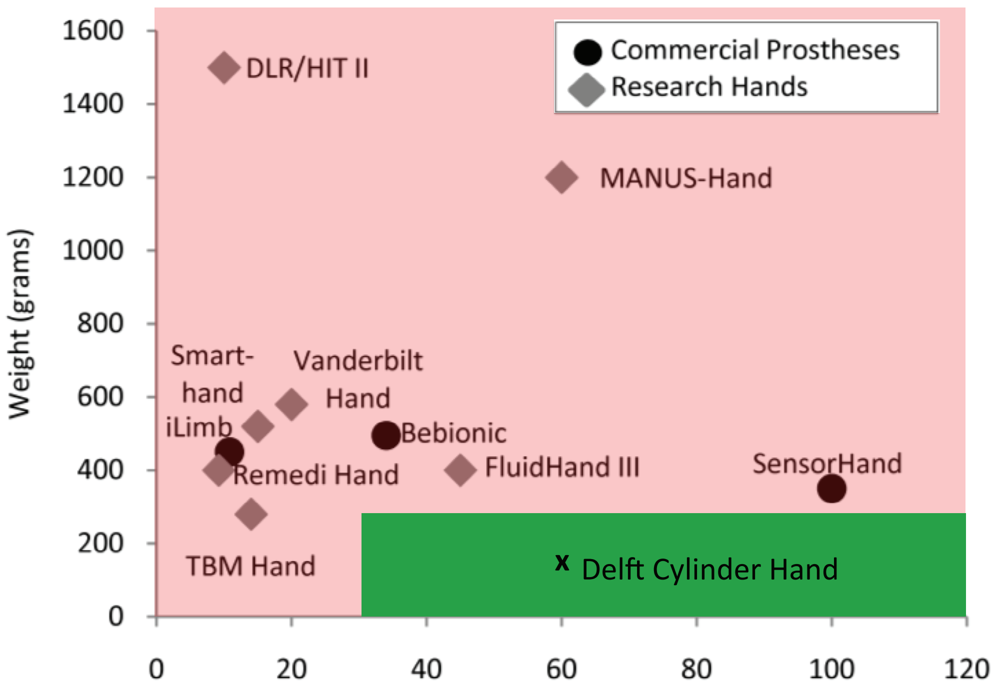

Figure 8.2 Hand mass as a function of the pinch or precision grip force (adapted from Belter [34]). The green area indicates the design space in which a prosthetic hand should be designed. A prosthetic hand should have a mass of less than 273 gram, to be comfortable to wear and it should be able to pinch over 30 N, to enable grasping of a broad range of objects. Currently the Delft Cylinder Hand (marked with an X) is the only hand which is within these boundary conditions.

Finally it is remarkable that none of the studies on articulating hands reported quantitative functional outcomes. It is not clear why no quantitative outcomes have been published. This can however be considered as a weakness of these studies, as it does not become clear whether a new prosthetic hand meets its demands, and whether it is better than other hands.

Functional evaluation

To enable objective functional comparison to current and newer prostheses, the performance of a new hand prototype should be subjected to objective testing. The evaluation of the hand prototype is described in the second part of Chapter 7. Objective data on the evaluation of prosthetic hands is scarce in literature. Instead most studies use questionnaires to evaluate upper limb prostheses, e.g. TAPES [39], PUFI [40] and OPUS [41]. Using questionnaires has two significant drawbacks. In the first place questionnaires are subjective. Their outcomes might therefore be biased by the subject [42]. Secondly there are many different questionnaires available in the field of upper limb prosthetics, which makes objective comparison complicated or even impossible. There are however several functional tests available, which enable a more objective comparison. Currently the biggest drawback of the functional tests is the limited amount of published test data. Also not every functional test can be used, e.g. the ACMC [43] test is specially designed for myoelectric prostheses and it is not suitable for the evaluation of BP prostheses. Other tests require considerable time to administer, e.g. the SHAP [44, 45]. This makes it difficult to do many repeated sessions with the same subjects. For the evaluation of the prototype the Box and Blocks Test (BBT) and the Nine Hole Peg Test (NHPT) were used. Both tests take a short time to administer, which enabled multiple repeated sessions with multiple subjects. Our prototype was tested by 13 healthy subjects. The hand could easily be controlled without any training, although a little practicing improved the performance of the controller. During the tests period (~30 min.) the functional performance improved to a level that was similar to the performance of the myoelectric hands tested by Farrel *et al*. [46] and of the BP-hooks tested by Schabowsky *et al*. [47]. The outcomes enable objective comparison to other hands. The functional evaluation of the Delft Cylinder Hand showed that scores of functionality similar to current myoelectric hand can be reached, with an anthropomorphic looking hand that has a 55-68% lower mass, without the need of any motor or battery. This is a very positive result, as it addresses one of the amputees' first design priorities in the design of prosthetic hands, being reduction of excessive mass.

8.2 Recommendations

A prototype has been developed and tested that combines the advantages of BP control, hydraulic transmission and adaptive gripping. The prototype proofed the technical feasibility and demonstrated it functional performance. However before the hand prototype can become widely available as a prosthetic device, further research and development is necessary.

Clinical testing

The results of the prototype testing in this study, described in Chapter 7, showed very promising results. The tests showed that the prototype is efficient, durable and has a good functional score. Now the prototype has proofed to function well in a controlled environment, the way is open for testing the prototype in an uncontrolled environment. Therefore the next step would be having the hand prototype tested by a group of patients in a daily situation, during a number of days or weeks. Such a clinical trial will provide more insight in the functionality of the hand prototype during daily activities. This will also reveal possible weaknesses of the prototype, as the hand will be subjected to unexpected loads. Ideally the hand would be instrumented during these trials. By instrumenting the hand the usage of the hand can be recorded. Measuring the loads that act on the prosthesis might help to identify possible causes of failure.

Locking mechanism

In the future it might be desirable to add a locking mechanism to the hand. The hand prototype that was developed in this study is an voluntary closing device. VC controlled devices have two drawbacks:

- The device is open, when not activated.
- Holding objects over a longer period of time can be tiring, as this requires constant actuation.

As we have seen in Chapter 2 most VC devices have an automatic locking mechanism. However the tests in this chapter also revealed that the automatic locking mechanisms did not function properly. After activation of the locking mechanism, the pinch forces dropped significantly. This undesired behaviour might be caused by backlash in the mechanisms. (This problem was already reported in 1955 [48]). The functional value of the locking mechanisms currently available is therefore limited. The only device that has no automatic locking mechanism is the TRS GRIP. When desired by the amputee the GRIP can be equipped with a manual controllable locking mechanism, but it is also used without a locking mechanism.
The new VC hand prototype that was developed in this study does not have a locking mechanism. Adding a locking mechanism might reduce the mechanical efficiency of the system. Especially when an automatic locking mechanism is used. Because the necessity of using a locking mechanism is still not clearly studied, it was decided not to include a locking mechanism in the prototype. Future clinical studies might however reveal that the addition of a locking mechanism is desired or deemed necessary by the test subjects. It is relatively easy to add a locking mechanism to the hydraulic system of the hand prototype, as the hydraulic system of the hand is fully modular. Such a locking mechanism can be a simple manual locking mechanism, or a more complex automatic locking mechanism. A manual locking mechanism could be a simple manual operated hydraulic valve, which can block the return flow, thus keeping the hand closed. A more complex hydraulic valve could engage and disengage automatically, at predefined conditions. Because of the modularity of the hydraulic system, different locking mechanism can easily be added or removed. In this way the amputee can try out different systems and choose the option that is most functional for his individual situation. A locking mechanism might help the amputee in specific tasks, which require holding an object over a longer period of time.

Increase pinch force

Another future direction in the development of the hand might include further increasing the pinch force, and or reducing the actuation force. The current hand prototype is capable of producing a pinch force of 30 N per finger. Most activities require a much lower pinch force. Therefore the required actuation force is in general quite low, as the hand already closes at an actuation force 20 N. Also for many tasks the adaptive cylinder grip can be used, instead of the pinch grip. Because of the adaptivity of the hand, the contact surface between the hand and the object is large and the contact forces can be low [49]. The available literature and the performed test trials did not reveal problems for the current levels of pinch force and actuation force. Future clinical trials might however indicate that a further reduction of the actuation force, and increase of the pinch force is necessary or desirable. Recent measurements on amputees [50] indicated that the actuation force that amputees can produce might be significantly lower than what we might expect based on available literature [11]. New measurements have to reveal the optimal control force, to operate a body powered device.

Even when the current levels of actuation force are acceptable, it might be desirable to reduce the required actuation force. Reduction of the actuation force will open up the use of the body powered hand to people that have less muscle force (e.g. children, elderly people, high level amputees). It will also make the control of the hand more comfortable and less physically demanding. Increasing the pinch force level will enable the amputee to perform even more tasks.

To obtain a higher pinch force at a lower actuation force level, requires an increase of the 'pinch force/actuation force ratio' of the hand. The increase in this force ratio can be achieved by two strategies:

In the first place a bi-phasic mechanism could be used [51]. Such a mechanism is based on the two grip phases. The first phase is the 'sizing phase'. In this phase there is only displacement of the fingers. The pinch force and actuation force are low. Ideally the transmission ratio should be small in this phase. The second phase is the "pinch phase". In this phase the fingers pinch the object, which requires a considerable pinch force. The displacements are small. Ideally the transmission ratio should be large in this phase. Current devices have a fixed ratio, which is a compromise between both phases. Also the new prototype has a fixed transmission ratio. A two phase mechanism has two transmission ratios. When the fingers touch the object, the bi-phasic mechanism switches from the sizing phase ratio, to the pinch phase ratio. This enables a higher pinch force at a lower actuation force. This principle is also used in some current electric devices, like the Otto Bock System hand and the Motion Control hand [52]. This first strategy can be implemented without making changes to the hand. A hydraulic bi-phasic mechanism can be added to the modular hydraulic system of the hand. Such a mechanism should specially be designed for the application in the hydraulic system of the hand. In the past a bi-phasic mechanism has already successfully been used in a hydraulic prosthetic arm [53].

A second strategy to obtain a higher pinch force at a lower load, is by using the principle of synergetic prehension [54]. Synergetic prehension is in fact a special type of bi-phasic mechanism, in which the finger and the thumb each have their own actuator, with each a different transmission ratio. During the "sizing phase" the finger with the low force transmission ratio moves and the thumb is locked in position. During the "pinching phase" the finger is locked in position and the thumb with the high transmission ratio builds up a high pinch force. This principle was used in the synergetic hook [54] and can also be used in hands [55, 56]. This principle could also be used in the hydraulic hand. This requires the addition of an actuator to the thumb, which is in the current prototype entirely passive. The thumb cylinder should have a larger transmission ratio than that of the finger(s). This can be realised by using a larger cylinder diameter, or a longer actuation lever. Also a hydraulic mechanism

should be added, which switches between the two phases.
Enabling the user to produce a larger pinch force at a lower actuation force, is expected to improve user comfort and prosthesis functionality. Also it will enable the use of a body powered system by amputees who are not able to generate sufficient actuation force for the current device.

8.3 Conclusion

This thesis described the development of a lightweight BP prosthesis with articulating fingers, the Delft Cylinder Hand. From the studies performed towards the development of this hand, the following conclusions can be drawn:

- The current status of body powered prosthetic hands, both VO and VC, has not improved during the past decades. In general the tested BP hands had a low efficiency, a low pinch force (< 15 N) and they required a high actuation force and actuation energy (1600-1700 Nmm) and they are heavy (~423 gram). The hands should be improved in order to meet the user demands.
- The principle of hydraulic cylinder transmission offers an efficient and powerful energy transmission, which is suitable for application in a prosthetic hand. Compared to a pulley transmission, the hydraulic system required 35-74% less energy. When the Bowden-cable is also replaced by hydraulics, the hydraulic system offers an additional improvement in efficiency of 10-40% of the entire system, compared to current BP systems. Therefore the hydraulic transmission was chosen to be used in the new hand prototype.
- The choice of material of the cosmetic glove can have a significant influence on the mechanical performance of a prosthetic hand. The tested Silicone gloves had a 2.5-4.5 lower stiffness than the PVC glove and required 1.8 to 3.8 times less actuation energy. Therefore for the new developed prototype a Silicone glove was used.
- The undesired positive stiffness induced by a Silicone or PVC glove, can be compensated by a negative stiffness compensation mechanism. The compensation mechanism could reduce the required input force and torque for over 50%, during finger movements without pinching. A negative stiffness mechanism was applied to the current prototype, however it has not yet been optimized to the current used cosmetic glove.
- A prototype of a body powered articulating hand was designed, built and evaluated. The hand is named the Delft Cylinder Hand. The hand prototype is voluntary closing and has a hydraulic cylinder transmission. It requires a low input energy (828 Nmm) and actuation force and has a 55-68% lower mass (217 gram, including cosmetic glove) than current available prosthetic hands. Unlike most current prosthetic hands the prototype is not bigger than a human hand. The hand is fully modular and can also be used for partial amputations.
- The hand prototype can easily be controlled without any training. Functional tests showed that The ‘Delft Cylinder hand’ provides the amputee a level of function that is comparably to current contemporary hands, at a cost (mass and actuation effort) which is much lower than that of all current available hands.

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Samenvatting

Natuurlijk Grijpen:
Ontwerp en evaluatie van een actief sluitende adaptieve hand prothese

Het vervangen van een ontbrekende menselijke hand, door een kunsthand is een van de grootste uitdagingen op het gebied van de revalidatie. Hoewel er veel verschillende handprothesen beschikbaar zijn, gebruikt 27% van de handprothesedragers de prothese niet actief en draagt 20% helemaal geen prothese. Er zijn verschillende redenen waarom mensen stoppen met het dragen van de prothese, bijv. draagcomfort (te zwaar, te warm), te weinig functioneel voordeel, moeilijk of vermoeiend in het gebruik, gebrek aan sensorische feedback. Gebruikersstudies laten zien dat verschillende aspecten van de prothesen verbeterd dienen te worden, om te voldoen aan de eisen van de gebruiker. Vermindering van de massa van de prothese heeft hierbij de hoogste prioriteit. De gebruikerseisen kunnen kortweg samengevat worden door de drie C's: Cosmetiek, Comfort en Controle. De prothese moet mooi zijn om te zien, comfortabel om te dragen en moet makkelijk te bedienen zijn.

Het doel van deze studie was het ontwerpen en het testen van een lichtgewicht en mechanisch efficiënte lichaamsbekrachtigde handprothese met articulerende ofwel scharnierende vingers. Een lage prothesemassa zal het draagcomfort verbeteren. Mechanische efficiëntie zal de bedieningskracht verlagen en daarmee het bedieningscomfort verhogen. Ook zal de hand hierdoor harder kunnen knijpen, wat zal resulteren in een verbeterde functionaliteit van de hand. De articulerende vingers maken het mogelijk om zowel de pincetgreep als de cilindergreep te vormen. Hierdoor kan een breed scala aan verschillende objecten worden vastgehouden. Bovendien verbetert dit de natuurlijke cosmetiek.

De eerste stap in de studie was om de state-of-the-art van de huidige lichaamsbekrachtigde prothesen te bepalen. Hoofdstuk 2 beschrijft het testen van actief sluitende prothesen en hoofdstuk 3 beschrijft het testen van actief openende prothesen. De mechanische efficiëntie van de geteste haken was beter dan die van de handen. Voor het bedienen van de handen was een hoge bedieningsinspanning en bedieningsenergie (1058-2292 Nmm) nodig. Ook dissipeerden de handen een groot gedeelte van de bedieningsenergie en leverden ze slechts een lage knijpkracht (~15 N). De massa van de handen was hoog (~423 gram). Een vergelijking met de resultaten van een studie uit 1987 liet zien dat de prothesen de afgelopen decennia niet verbeterd waren. Om te voldoen aan de gebruikerseisen moet een nieuw handontwerp een lagere massa hebben, minder bedieningsenergie vragen, minder bedieningsenergie dissiperen en de hand dient een hogere knijpkracht te leveren.

Hoofdstuk 4 beschrijft het ontwerp en het testen van twee ondergeactueerde vingerprototypen. Het ene vingerprototype was voorzien van een kabel-katrol transmissie, het andere van een hydraulisch transmissie. Beide vingers waren geoptimaliseerd voor toepassing in een cosmetische handschoen van een handprothese. De vingers hadden identieke afmetingen en hadden een erg lage massa. Er zijn kwantitatieve testen uitgevoerd om te bepalen welke transmissie het efficiëntst was. De vinger met de kabel-katrol transmissie had 35-74% meer energie nodig voor verschillende taken dan de hydraulische vinger. Vanwege de hogere mechanische efficiëntie van de hydraulische vinger, is deze vinger geselecteerd als meest geschikte vinger voor toepassing in een handprothese. Bovendien biedt een hydraulische transmissie een extra efficiëntieverbetering van 10-40%, wanneer de hydraulische transmissie gebruikt wordt om de Bowdenkabel te vervangen. Daarom is besloten om in het nieuwe handprototype de hydraulische transmissie toe te passen.

Hoofdstuk 5 beschrijft de vergelijking van mechanische eigenschappen van cosmetische handschoenen van siliconen en PVC. Beiden typen handschoen kunnen gebruikt worden op een handprothese. De testen zijn uitgevoerd om de handschoen te selecteren met de hoogste energie-efficiëntie. De stijfheid van de gemeten siliconen handschoenen was 2.5-4.5 keer lager dan die van de PVC handschoenen. De siliconen handschoen hadden een 1.8 tot 3.8 keer lagere actuatie-energie nodig en dissipeerden 1.7 tot 3.4 keer minder energie. Er is daarom gekozen om voor het nieuwe hand prototype een siliconen handschoen te gebruiken.

Hoofdstuk 6 beschrijft het ontwerp en het testen van een handschoencompensatie-mechanisme. Dit mechanisme, dat binnenin een vinger past, heeft een negatieve stijfheid die de ongewenste positieve stijfheid van de handschoen compenseert. De negatieve stijfheid van het mechanisme verminderde het benodigde actuatiemoment met 58% voor de PVC handschoen en met 52% voor de siliconen handschoen. In het nieuwe handprototype is ook een mechanisme met een negatieve stijfheid toegepast, om de vereiste bedieningsinspanning van de gebruiker te verlagen.

De laatste stap van de studie was het ontwerpen en het testen van een nieuwe handprototype, beschreven in hoofdstuk 7. Het nieuwe handprototype, de Delft Cylinder Hand, heeft ondergeactueerde articulerende vingers die zich aanpassen aan de vorm van het object. De hand wordt bediend met een lichaamsbekrachtigde actief-sluitende bediening. De transmissie van de hand is hydraulisch. De hand is onderworpen aan verschillende mechanische en functionele testen. Hoofdstuk 8 beschrijft de vergelijking van de prestaties van de hand met die van de huidige prothesehanden. Door de toepassing van een hydraulische transmissie hoeft de gebruiker 49-162% minder energie te leveren om de hand te bedienen dan voor de huidige lichaamsbekrachtigde handen nodig is. De hand kan een hogere knijpkracht leveren (30-60 N). In de functionele testen behaalde de hand vergelijkbare scores als die van myo-elektrische handen. De massa van de hand (152 gram zonder handschoen; 217 gram met handschoen) is 68% lager dan die van de lichtste articulerende myo-elektrische hand en 55% lager dan die van de lichtste lichaamsbekrachtigde hand van vergelijkbaar formaat. De functionele testen lieten zien dat de 'Delft Cylinder

Hand’ de prothesegebruiker een functionaliteit biedt die tenminste vergelijkbaar is met die van de huidige prothesehanden, tegen een belasting (handmassa en bedieningsinspanning) die veel kleiner is dan die van alle huidig beschikbare handen.

De Delft Cylinder Hand heeft articulerende vingers en is antropomorfisch, rank, snel, efficiënt en stil. De handmassa is veel lager dan die van de lichtste commercieel verkrijgbare hand. De hand beantwoord daarmee aan een van de belangrijkste gebruikerseisen, namelijk die van een lage handmassa. De hand kan harder knijpen (>30 N) met een lagere bedieningsinspanning.

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Gerwin Smit
Den Haag, 2013

Curriculum Vitae

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2005
Researcher at Philips Advanced Technology Centre (ATC), Drachten. Performing analyses and tests in the SEM electron microscope, using a new technique which was developed during the graduation internship.

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MSc-study Mechanical Engineering at University of Twente, Enschede, The Netherlands. Graduation internship at TNO, Defence and Security, Rijswijk. Design and validate an injury model, to prevent injuries from explosions. Title of the Master thesis: 'Injury assessment for blast induced whole body displacement.'

2008-2013
PhD-study at Delft University of Technology, Delft, The Netherlands. On the design and evaluation of a voluntary closing adaptive hand prosthesis.

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Co-founder of Delft Prosthetics BV. Manufacturer of upper limb prostheses and sockets.

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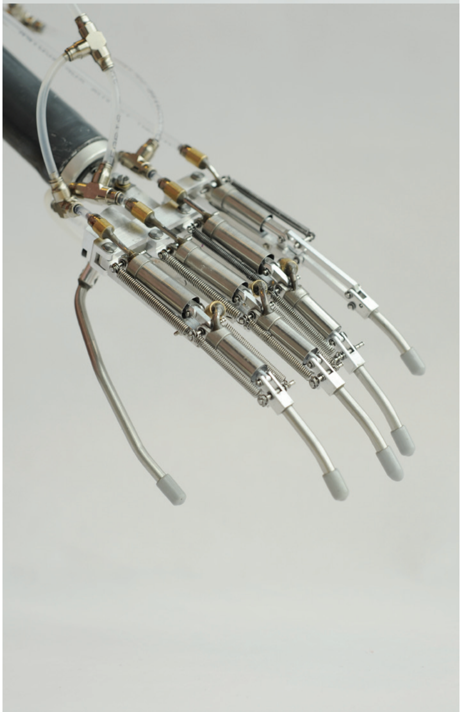
Co-founder of de 'Handenstichting' or 'Hand Foundation', a foundation which raises funds to support applied scientific research on upper limb prosthetics.



This thesis describes the design and functional evaluation of a new hand prosthesis prototype: the Delft Cylinder Hand. This hand prosthesis has articulating fingers which fully adapt to the shape of the object. The fingers are actuated by small hydraulic cylinders, inside the fingers of the hand. The user can easily control the hand by using a shoulder harness. The hand is lighter than all current hands. Therefore it can be comfortably worn. It can be fast and accurately controlled. Furthermore the hand has a anthropomorphic size and shape. This thesis presents the first prosthetic hand that meets all the basic user demands.

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