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User capacities and operation forces

Requirements for body-powered upper-limb prostheses

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To my father

TABLE OF CONTENTS

1

INTRODUCTION

While reading these sentences, you are either holding a book or a digital reading device in your hands – or perhaps you are sitting behind a computer. Whatever medium you are using, you need your hands to hold, turn a page, swipe, scroll or click. Your hands are continuously active. Think about your day. Think about which routines you perform when getting up, getting dressed, preparing and having breakfast, going to the toilet, brushing your teeth, leaving the house (or not), and what kind of gestures you were making when talking to another person; maybe you were waving to your child to say goodbye or sending a kiss by hand to your loved one. Maybe you caressed him/her after saying something nice. Did you recognize that the water you were washing your hands with was warm or cold? Was it a soft or hard towel you used?

Imagine you would miss a hand, or even both. For many people, this is reality. Today, in the Netherlands approximately 3750 persons have an upper-limb deficiency, that means that they miss (a part of) their hand, (a part of) their lower arm, (a part of) their upper arm or even their entire arm including the shoulder joint. 1350 of them were born without a hand (congenital defect) and 2400 lost their hand later in life, for example due to disease (for instance dysvascular conditions or cancer), or traumatic causes (for instance physical and thermal injuries, infections after injury, explosives) (1). In the US 41,000 (1.4 per 10.000 inhabitants) persons are estimated to have major (i.e. excluding fingers) acquired upper-limb deficiencies (2), whereas various studies report congenital upper-limb deficiencies of 2.8-5.0 per 10,000 births (3).

Would it not be great if we could engineer a replacement hand, which gives back the full functionality of the unaffected hand? Despite a long history of upper-limb prosthesis development, current prostheses by far do not match the functions of a human hand: grasping, holding, and manipulating objects, as well as feeling, touching, and expressing yourself during communication.

1.1 HISTORIC OVERVIEW OF UPPER-LIMB PROSTHESES

The oldest prosthetic hand known was found on an Egyptian mummy from around 330 B.C., which was without moving parts and was probably targeting the restoration of the wearer's outward appearance (4), thus a cosmetic prosthesis. Other known examples of passive hand prostheses are the first hand of Florence from the second half of the 15th century (4), the hands from Götz von Berlichingen from the early 16th century (4,5) and Ambroise Paré from the second half of the 16th century (5). In the early 19th century the harnessing of gross movements of body-parts to actively operate a prosthesis started, thus the era of body-powered prostheses began. Around 1818 Ballif designed a below elbow prosthesis. In this prosthesis arm abduction controls the fingers, whereas the extension of the elbow controls the thumb (5). In 1844 an above elbow prosthesis was designed by Van Peetersen. The elbow flexion and extension of the prosthesis is controlled by shoulder anteflexion and finger flexion can be achieved by arm abduction and as a result of the prosthesis' elbow flexion (5). Another below elbow prosthesis was designed by the Count of Beaufort in 1860. The thumb of his wooden prosthetic hand is operated by arm abduction and anteflexion. The harness system seems quite comparable to the figure-of-nine harness, which is used nowadays (4). Charrière designed in 1860 an elbow disarticulation prosthesis for an opera singer. Again the harness looks similar to the figure-of-nine harness. Elbow flexion, wrist supination, wrist flexion and hand closing were coupled together and were activated by a single control cable (5). In 1911 Carnes proposed two prostheses, one below and one above elbow prosthesis. Both prostheses were two-way controlled. Bi-scapular abduction and/or shoulder flexion opened the hand, whereas shoulder shrug closed the hand (5). The Dorrance split hook, which still finds broad application in body-powered prostheses nowadays, was patented in 1912 (6).

As a comparison to the long lasting history of body-powered prostheses: in 1948 Reiter introduced the first myo-controlled hand prosthesis (7), which became commercially available in the late 60's and early 70's (8) and has been under great research and commercial attention ever since.

Nowadays a wide variety of prostheses is commercially available in all three categories: cosmetic, passive and active prostheses. Cosmetic prostheses aim to replace the missing hand such that it looks as naturally as possible without offering any grasping function to the user. The thumb and/or fingers of a passive prosthesis can be opened by the sound hand and by a closing-spring mechanism an object can be held, for instance for carrying purposes. Available active prostheses are myo-electric and body-powered prostheses. Both come either with a (multi-articulating) hand or a hook as prehensor. Myo-electric prostheses are activated externally by electric motors. They are controlled by electrical signals generated by the user's muscles. Body-powered prostheses are activated and controlled by body-movements and rely on mechanical principles only, which are discussed in detail in the next section.

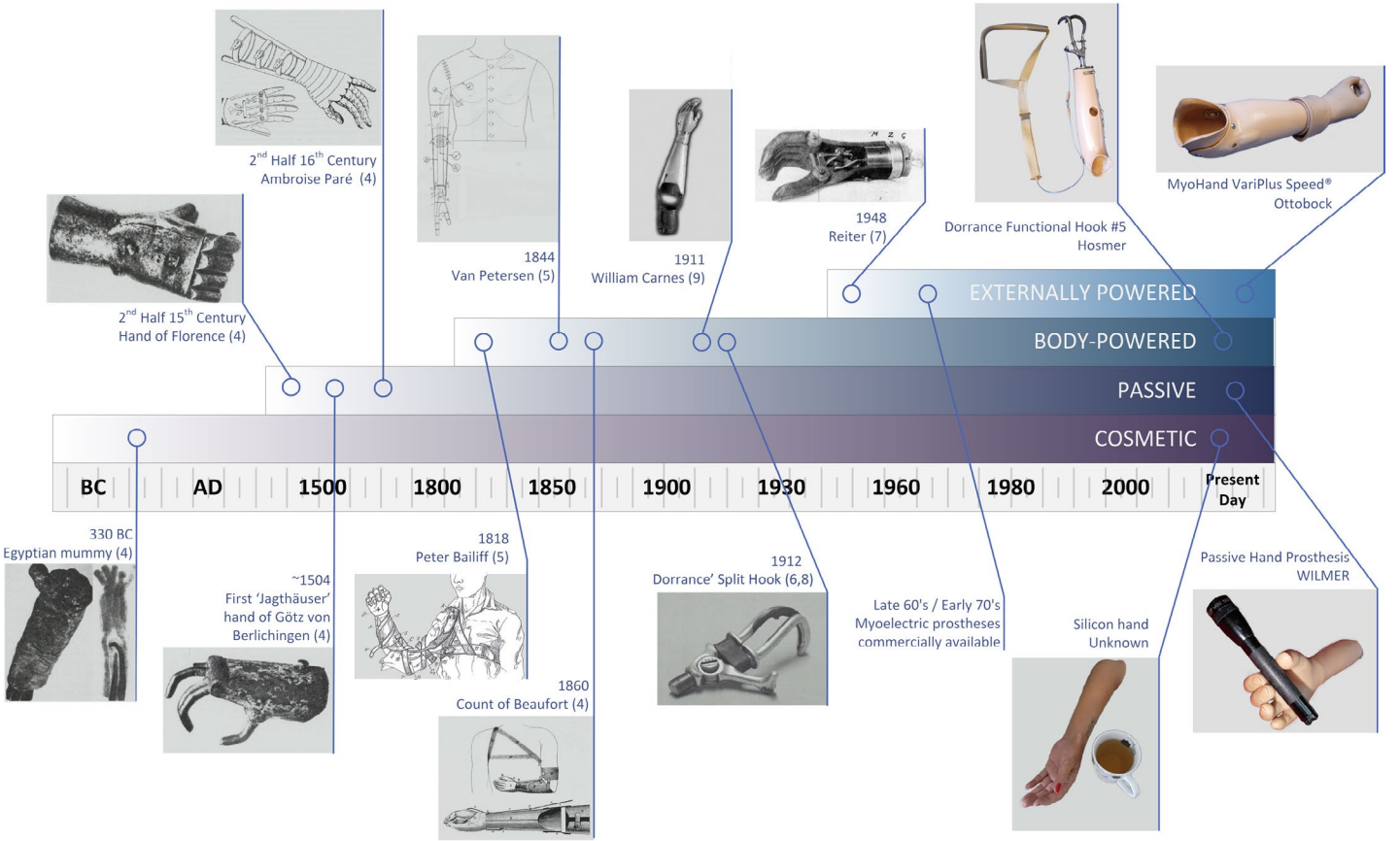

Figure 1–1. Timeline of prosthesis development 330 B.C. until the present day. Sources: (4-7,9)

A prosthesis is often used for fixating an object or for indirect grasping, in which the sound hand transfers an object to the prosthesis (for example during bi-manual tasks), and less frequently for direct grasping (10). Unilateral amputees execute tasks that require direct grasping mostly with their natural hand. Thus, the ability to grasp objects with a prosthesis is inferior to that of a natural hand. One explanation might be that the perceived information on what the object is like is limited (for example object deformability), if not absent (for example information on temperature or texture of the object). Furthermore, information on what the contact surface of the object and prehensor is like and the magnitude of the exerted force on the object, the pinch force, is of importance, but might not be available to the user. Perception and control of prosthesis activity remains a challenge due to the limited or low quality sensory feedback.

Promising developments on peripheral nerve interfaces, which connect the nerves in the arm to sensors in the prosthetic hand via electrodes, have shown that an amputee could identify stiffness and shape of different objects and effectively modulate the grasping force of his prosthesis without visual or auditory feedback (11). But so far this has only been implemented in a lab setting. Interface design problems such as varying fidelity of the resolution, relatively weak, noise-ridden electrical signals, inflammation (as a cause of signal instability over time but also a health risk on its own), as well as injury to nerve fibres and pain (12) have not been solved yet. Additionally, the required surgery with concomitant risks and costs complicate the practical application of peripheral nerve interfaces and delay the availability for prosthesis applications in clinical practise.

Furthermore, brain-computer interfaces show revolutionary developments. In February 2012, a woman, who lost the control and function of her limbs and torso due to tetraplegia, underwent brain surgery. Two microelectrodes were implanted in the motor cortex and neural signals were transmitted via the electrodes to a 7 degrees-of-freedom robotic arm. After 13 weeks of training she was able to control the robotic arm routinely and feed herself dark chocolate, which was her defined goal before she underwent surgery (13,14). As is the case for the peripheral nerve interfaces, up till now these brain-computer interfaces are only placed in lab settings for similar reasons.

Unlike peripheral nerve and brain computer interfaces, myo-electric and body-powered prostheses are available for the clinical practice. Both offer the user grasping function, which enables him to conduct daily activities. Interestingly, many potential prosthesis users decide not to wear it: 16-20% of potential users continue life without any prosthesis and rejection rates of upper-limb prostheses vary from 23 to 45% of the users for various reasons (15). This indicates that there is much room for improvement to satisfy the users' needs.

1.2 BODY-POWERED VERSUS MYO-ELECTRIC PROSTHESES

Advantages of body-powered prostheses range from lower mass, higher reliability, quiet operation, shorter training time, easier to clean, low costs, independence from external energy source, to proprioceptive force and position feedback. On the other hand, myo-electric prostheses offer increased pinch strength, advantages in appearance and increased comfort due to the absence of a harness (15).

The costs of a body-powered prosthesis are estimated between $4.000 – $10.000, whereas the costs for myo-electric prostheses range from $25.000 to $75.000 (16). Additionally, body-powered prostheses require less maintenance and a shorter training time (17). Thus, body-powered prostheses offer a low-cost and low-maintenance solution, which is beneficial for the public health sector. Furthermore, body-powered prostheses offer an attractive solution for developing countries.

Myo-electric prostheses rely mainly on visual feedback, although the user can also hear and feel the electric motors. Many approaches have been undertaken to engineer artificial feedback in myo-electric prostheses, such as vibro-tactile feedback (18,19), mechano-tactile feedback (pressure on skin) (20-22), electro-tactile feedback (electro-cutaneous stimulation) (23-25), skin stretch (26), and force feedback spanning the joint (27). Although some approaches showed significant improvement in feedback qualities, none of them have been implemented in clinical practice to date. Additionally, all above mentioned approaches target tactile feedback, which has an inferior role in dynamic force feedback tasks (28). The fastest, and preferred, form of feedback is proprioception with its fast spinal cord feedback loop. Haptic display mechanisms that feature force feedback might help improving myoelectric prosthesis control (29). Furthermore, myo-electric prosthesis users experience a delay between their muscle activation (control action), and the movement of the prosthetic digits. Through the direct connection of the user's movements to the prehensor, the body-powered prosthesis user 1) benefits from the fast proprioceptive feedback (29) and 2) experiences no delays in action and effect compared to a myo-electric prosthesis.

Unfortunately, body-powered prostheses do not exploit their advantages to their full potential since little effort is taken to improve body-powered prosthesis design. The Delft Institute of Prosthetics and Orthotics (DIPO) focuses on the enhancement of body-powered prostheses since the Thalidomide tragedy in the late 50's and early 60's of the twentieth century. Worldwide over 10.000 infants were born with malformation of the limbs due to the drug Thalidomide, which was used against nausea and to alleviate morning sickness in pregnant women (30). Suddenly the need for state-of-the-art prostheses exploded, which was the beginning of the upper-limb prosthesis research activities at the Delft University of Technology. The DIPO has worked ever since in close collaboration with rehabilitation centres in the Netherlands to ensure clinical interactions and applications. Despite the increased commercial and research focus on

myo-electric prostheses, DIPO's believe in the advantages of body-powered prostheses remains unchanged.

1.3 BODY-POWERED PROSTHESES

1.3.1 Working principles

A body-powered prosthetic system can be described by its individual elements:

Shoulder Harness ('SH' in Figure 1–2 and Figure 1–3). A body-powered prosthesis is operated by physical movements of the user, which are captured by a shoulder harness. Humeral abduction and anteflexion of the affected side and shoulder protraction of the contralateral side result in a change of distance between point A and B as indicated in Figure 1–2.

Transmission ('TM' in Figure 1–2 and Figure 1–3). Cable forces and excursions are transmitted via a Bowden cable at the user's back to the prosthetic prehensor. A well-established application of the Bowden cable is a bicycle hand brake. An inner cable guided through an outer cable housing transmits cable forces and excursions from handle bar to brake claws.

Prehensor ('PH' in Figure 1–2 and Figure 1–3). Hand mechanism and prosthetic digits together form the prehensor. The hand mechanism translates the cable movements into prosthetic digit movements. The prosthetic digits can resemble healthy human digits, or take the shape of a hook.

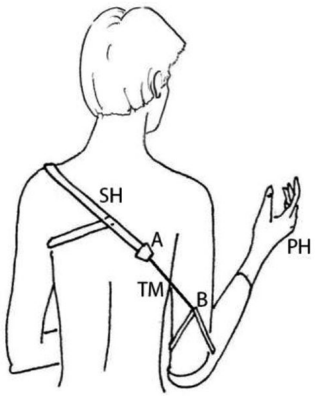

Figure 1–2. Shoulder controlled body-powered prosthesis. By increasing the distance between A and B, the control cable is pulled and the hand is actuated. (adapted from (8))

1.3.2 Voluntary closing and voluntary opening prehensors

In body-powered prostheses two types of prehensors are available: voluntary closing and voluntary opening, which closes or opens the prehensor respectively when the cable is pulled. A spring (or rubber bands) return the prehensor to its initial state. For a voluntary opening prosthesis this implies that the pinch force is dependent on the spring's properties (or the

number and properties of rubber bands on the hook), which can be chosen by the user in case of a hook prosthesis, dependent on the highest desirable pinch force. A cable force overcoming this spring force needs to be exerted each time the prehensor opens to grasp an object. The advantage that holding an object does not require any user effort comes together with the disadvantage that it "requires several times more mechanical work to operate" (31). Also, in case the spring force is too high to hold delicate objects, the user needs to counterbalance the pinch force by applying cable force.

The pinch forces of the voluntary closing prosthesis are directly related to the exerted cable forces by the user's movements. Thus, the control movements are directly related to the actions at the prehensor and consequentially the user gets feedback of prehensor and digit positioning as well as pinch forces due to Extended Physiological Proprioception (EPP) (17). However, this requires the user to hold on to the exerted cable force as long as he holds and manipulates the object, but enables him to adjust pinch forces in an intuitive, fast and easy way, which is desired in prosthesis control. Alternatively, the user can activate a locking mechanism, which is designed to maintain the hand opening and pinch force when releasing the tension from the cable.

1.3.3 Shortcomings of body-powered prostheses

Rejection rates of body-powered prostheses vary in different studies from 16 to 66%. They are reported to be mechanically inefficient (32-34), offer a limited pinch force and require at the same time high cable operation forces from the user (32,33). This might explain why users are complaining about exhaustion, (upper body) pain, sores, and skin irritation leading to discomfort and resulting in disuse of their prosthesis (35). Harness comfort is one of the consumer design priorities. The reported harness discomfort might partly be provoked by the high operation forces, but harness elimination and a greater choice in harnessing configurations is desired (35), probably also for cosmetic reasons. Clearly users are not satisfied with their prosthesis and desire better prosthesis design.

1.4 PROBLEM DEFINITION AND AIMS

Improved prosthesis design can contribute to higher prosthesis acceptance and can enhance the quality of life of upper-limb prosthesis users. However, quantified design requirements for body-powered prostheses are scarce.

The scheme in Figure 1–3 illustrates the human-prosthesis-object interaction. The user's central nervous system initiates muscle contractions resulting in muscle forces and body movements, which are fed back to the central nervous system by the proprioceptive feedback paths of muscle spindles and Golgi tendon organs. The body movements are captured by a shoulder harness and result in cable operation forces (F_{SH}) and excursions (x_{SH}) at the user's back (Point A and B in Figure 1–2). The resulting pressure on the skin at harness and socket ('SH' in

Figure 1–2 and Figure 1–3) is fed back to the central nervous system and serves as feedback loop of the produced cable forces and excursions (tactile feedback). These forces and excursions are transmitted via the Bowden cable ('TM' in Figure 1–2 and Figure 1–3) to the prosthetic prehensor ('PH' in Figure 1–2 and Figure 1–3). The prehensor's input cable forces and excursions result then in pinch forces (F_{PH}) exerted on an object and the prehensor's finger positioning (x_{PH}). The prehensor's digit positioning, object movement and an eventual deformation of the manipulated object result in visual information, which is fed back to the central nervous system. An eventual deformation of the manipulated object may also result in audible sound, which might serve as information of the applied pinch force on the object.

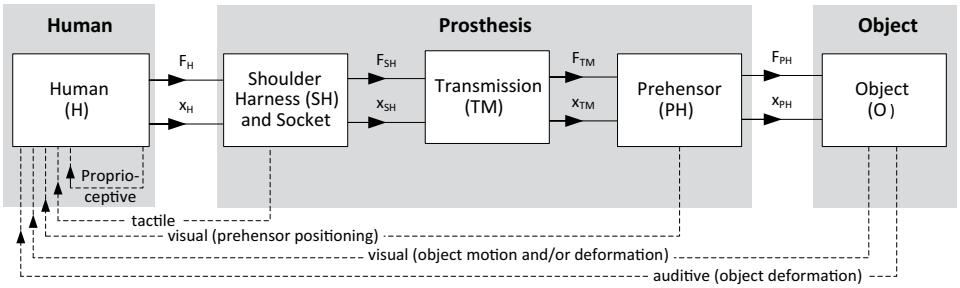

Figure 1–3. Scheme of the human-prosthesis-object interaction of a body-powered prostheses.

This scheme can serve as a design guideline for body-powered prostheses. Data on the Bowden cable and its efficiencies (36,37) as well as on different prehensors, their mechanical properties and available pinch forces (32,33) has been published. However, little is known on the prosthesis-input requirements, for instance the users' capabilities to exert, perceive and control cable operation forces and excursions and the resulting pinch forces and digit positioning.

The user demands an adequate pinch force of his prosthesis, which can be exerted onto an object, and high quality feedback of the prehensor-object interaction. To meet these demands the user's capacities need to be considered to realize a user-centred body-powered prosthesis design. As indicated in the section 1.3.3, the magnitude of cable forces appears to provoke problems. To date no information is available which operation force levels prosthesis users are capable to exert on the control cable; what constitutes a fatigue- and pain-free operation range feasible for daily activities; and what is the influence of the magnitude of cable forces and excursions on perception and control of cable forces and resulting pinch forces. Furthermore, a new harnessing configuration is desired for improved outer appearance and comfort, which should not deteriorate prosthesis control compared to the traditional harness.

This thesis aims to quantify user capacities to operate a body-powered prosthesis and establish a better understanding of the prosthesis-input requirements in order to frame quantified user-centred body-powered prostheses design requirements. Quantitative requirements facilitate improved prosthesis design, which enhances the quality of life of upper-limb prosthesis users and prevents (repetitive strain) injuries.

1.5 OUTLINE OF THIS THESIS

Chapter 2 quantifies the user's strength by identifying the maximum forces a prosthesis user is able to exert on the control cable to operate a body-powered prosthesis. This maximum force is not representative for daily activities, since the prehensor is activated many times during a day to grasp and manipulate objects. Exerting the maximum cable force for each prehensor activation would result in tiring and painful use. Therefore the maximum force is corrected for long-duration use. The maximum and corrected forces are used to evaluate current prosthesis.

The ability to control pinch forces is essential for adequate dexterity. The influence of high cable operation forces on the ability to control pinch forces is evaluated in *Chapter 3* by a pick-transfer-place task of a collapsible object with a prehensor. *Chapter 4* describes a force reproduction task, which was executed to identify the low cable operation force levels which can be perceived and controlled optimally by prosthesis users. Additionally, the influence of cable excursions on the control accuracy of cable forces is evaluated. To allow for a general advice on cable operation forces and excursions independent of the mechanical properties of one prehensor, the experiments in this chapter were performed without a prehensor, but included simulation of different prehensor properties.

Improved harness design is one of the consumer design priorities (35). The Ipsilateral Scapular Cutaneous Anchor System is a commercially available alternative to the traditional harness. *Chapter 5* compares the user's cable force control abilities of the Anchor system with the figure-of-nine harness at low operation forces utilizing a force reproduction task similar to that of *Chapter 4.*

2

FATIGUE-FREE OPERATION OF MOST BODY-POWERED PROSTHESES NOT FEASIBLE FOR MAJORITY OF USERS WITH TRANS-RADIAL DEFICIENCY

Mona Hichert, Alistair N. Vardy, Dick H. Plettenburg

Submitted

ABSTRACT

Background: Body-powered prostheses require cable operation forces between 33 and 131 N. The accepted upper limit for fatigue-free long-duration operation is 20% of a users' maximum cable operation force. However, no information is available on users' maximum force.

Objective: To quantify users' maximum cable operation force and to relate this to the fatigue-free force range for the use of body-powered prostheses.

Methods: 23 subjects with trans-radial deficiencies used a bypass-prosthesis to exert maximum cable force three times during three seconds and reported discomfort or pain on a Body-map. Additionally, subjects' anthropometric measures were taken to relate to maximum force.

Results: Subjects generated forces ranging from 87 to 538 N. Twelve of the 23 subjects generated insufficient maximum cable force to operate 8 of the 10 body-powered prostheses fatigue-free. Discomfort or pain did not correlate with the magnitude of maximum force achieved by the subjects. Nine subjects indicated discomfort or pain. No relationships between anthropometry and maximal forces were found except for maximum cable forces and the affected upper-arm circumference for females.

Conclusions: For a majority of subjects, the maximal cable force was lower than acceptable for fatigue-free prosthesis use. Discomfort or pain occurred in ~40% of subjects, suggesting a suboptimal force transmission mechanism.

2.1 BACKGROUND

Body-powered prostheses are rejected by 26-45% of the users (15). One of the reasons for rejection is the high operation force required for prosthesis activation (32,33), leading to pain or fatigue or, in the worst case, nerve and vessel damage (15,35). Required operation forces to pinch 15 N with a voluntary closing prosthesis vary between 33 and 131 N (32). For a 50 mm opening of voluntary opening prostheses, which are able to pinch at least 15 N, cable forces between 50 and 94 N are required (33). Using a prosthesis on a daily basis implies that the user should not feel tired after a number of manipulations and should also not experience any pain (e.g. sore muscles, pinching) during or after use. Humans can conduct isometric contractions without fatigue effects at a critical force level of 15-20% of their maximum voluntary contraction (38). Hence, taking the conservative value and maintaining 20% of users' maximum cable operation force as an upper boundary for daily use will enable users to operate their body-powered prosthesis fatigue-free.

However, the user's maximum cable forces have never been quantified. Current research is based on measurements on 50 'normal' subjects by Taylor in 1954 (39), who measured cable forces of 280±24 N (mean ± standard deviation) for arm flexion, 270±106 N for shrug and 251±29 N for arm extension. Unfortunately, the measurement procedure and the subject characteristics were not described. Moreover, the study reported forces and displacements from isolated movements instead of combinations of movements typically used for body-powered prosthesis operation. A recent unpublished pilot experiment on 10 male controls (28±2 years old), revealed average values of 475 N and a peak value of 970 N for one subject, which are significant higher than the reported values of Taylor (39).

Prosthesis user strength will probably show a large variety, resulting in a challenge for the clinical team to find the best individual suitable prosthesis. Predicting maximum cable operation forces by anthropometric measures might facilitate the prosthesis fitting procedure and prevent the need for costly measurement equipment. Furthermore, although discomfort has been reported for body-powered prosthesis operation (15,35), extent and locations of discomfort have never been related to the exerted cable forces.

This study aims to quantify users' maximum cable operation forces and to relate these to a fatigue-free force range for the use of body-powered prostheses. In addition we aim to identify extent and locations of discomfort provoked by the exertion of cable forces and to explore the predictability of maximum cable operation forces by the anthropometric measures of users.

2.2 METHOD

This study was approved by the medical ethical committee of University Medical Centre Groningen (UMCG). The subjects were recruited from University Medical Centre Groningen, Erasmus Medical Centre, Rotterdam, and the rehabilitation institute De Hoogstraat, Utrecht.

2.2.1 Subjects

Twenty-three adults (11 males, age: 49±13 years) all with an unilateral trans-radial deficiency participated (Table 2–1). All participants were free of neurological, muscle, joint, or motor control problems concerning the upper extremity or the torso (exclusion criteria). Nine participants had a right deficiency, 14 had a congenital defect, and 11 had experience with body-powered prostheses.

2.2.2 Materials

2.2.2.1 Maximum force measurements

A custom-made prosthesis simulator (Figure 2–1) was connected by the experimenter to the participant's prosthesis. For two participants, who did not own a prosthesis, the prosthesis simulator was placed on a temporary WILMER Open Socket (40). For two other participants the bypass-prosthesis was attached to the remnant arm since its length was sufficient for a firm connection. The prosthesis simulator consisted of an adjustable "figure-of-nine" harness linked to a standard 1/16'' (.159 cm) diameter stainless steel cable (C100, Hosmer Dorrance Corporation, Chattanooga, USA). Cable excursions were disabled in this setup. The Bowden cable was interrupted by a force sensor (S-Beam load cell ZFA 100kg, Scaime, Juvigny, France). The measured forces were amplified (CPJ, Scaime, Juvigny, France), sampled at 1 kHz (NI USB-6008, National Instruments, Austin, USA), and finally stored using a custom LabVIEW program (LabVIEW 2012, National Instruments, Austin, USA).

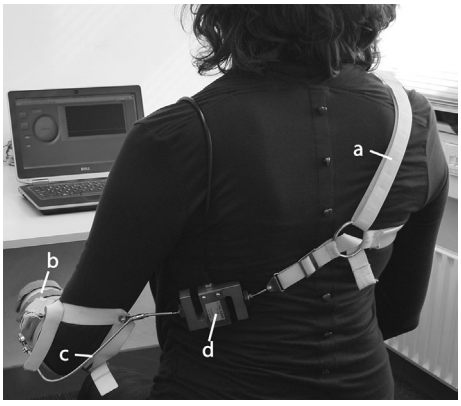

Figure 2–1. Measurement set-up for maximum force measurements: the "figure-of-nine" harness (**a**) and thermoplastic shell (**b**) are connected through a Bowden cable (**c**), which is interrupted by a force sensor (**d**). In this set-up cable excursions are disabled.

Table 2–1. Overview of the subject characteristics. Subjects are sorted by gender and the cause of their arm defect.

Subject no.	Gender	Age	Acquired/ congenital defect	Affected side	Dominant side	Currently used prosthesis	Experience in body-powered prosthesis use	Frequency of prosthesis use
9	female	54	acquired	left	right	myo	no	daily use
11		60		right	right	myo	no	daily use
14		69		left	right	myo & body-powered	yes	daily use
2		20	congenital	left	right	none	no	never
4		43		left	left	cosmetic	yes	use for specific tasks
8		46		left	right	myo	no	use for specific tasks
12		52		right	right	cosmetic	no	use for specific tasks
17		43		right	left	cosmetic	yes	use for specific tasks
18		34		left	left	none, only fork on socket	yes	use for specific tasks
19		58		left	right	none	no	got one, but never wore it
7	male	49	acquired	left	right	body-powered	yes	daily use
10		54		right	left	myo	yes	daily use
1		57	congenital	left	right	myo	no	daily use
3		68		right	right	myo & cosmetic	no	daily use
6		35		left	right	myo	no	daily use
16		50		left	left	myo	no	daily use
20		68		right	right	myo	yes	daily use
23		49		left	right	myo & body-powered	yes	daily use
13		47		right	right	none	no	got one, but never wore it
5		63		left	right	myo	no	daily use
15		25		right	right	myo	yes	daily use
21		37		right	right	myo	yes	daily use
22		55		left	left	myo	yes	daily use

2.2.2.2 Anthropometric data

The subjects shoulder width, upper-arm length and remaining lower-arm length was measured by the experimenter with an anthropometer (Model 101, GPM, Zurich, Switzerland). The upper-arm circumference was measured with a sewing tape.

2.2.3 Procedure

After signing an informed consent form the anthropometric data were taken following the instructions of the NASA Reference Publication 1024 (41); 103. Biacromial Breadth, 751. Shoulder-Elbow Length, 113. Biceps Circumference, Relaxed, 381. Forearm-Hand Length (the fingertips are represented by the far end of the subjects' stump). Then a prosthetic simulator was connected to the subjects' prosthesis. The subject was instructed to use their preferred combination of shoulder protraction of the sound side, humeral abduction and forward flexion on the affected side to create cable forces. Before starting the measurements, the subjects were allowed trial movements until they felt familiar with exerting forces on the control cable. Subjects were asked to deliver their maximal cable forces within three seconds. This was repeated three times. The three second time interval was chosen based on trial measurements, testing how much time is required to achieve the maximum force. The trial measurements were done with able-bodied subjects. Finally, subjects were requested to report locations of pain or discomfort on a Body-Map (Figure 2–2) after completing the experiment.

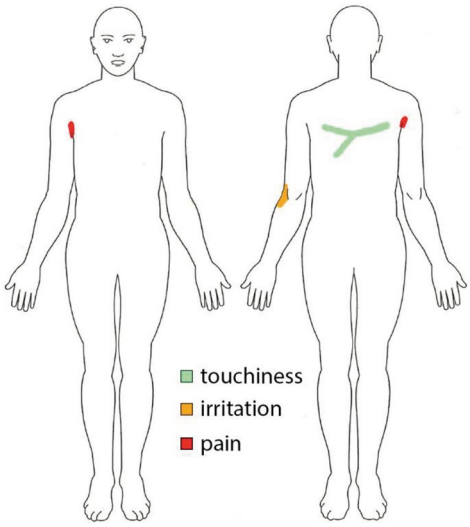

Figure 2–2. Body-Map coloured by one subject indicating pain (red) in the right arm pit, irritation (orange) at the back of the left elbow and touchiness (green) on a stripe of his back.

2.2.4 Data analysis

2.2.4.1 Maximum force measurements

The highest values of the three maximum force measurements were determined. Only trials where the maximum force was attained within the predetermined 3 seconds were included (56 of 69 trials).

The subjects' maximum cable forces and fatigue limits were compared to the required forces to 1) create a 15 N pinch force with a voluntary closing prostheses (32) and to 2) achieve a 50 mm prehensor opening with voluntary opening prostheses which is capable to pinch at least 15 N (33) (Table 2–2).

Table 2–2. Required cable forces to operate voluntary closing and opening prostheses.

Voluntary closing prostheses	Required cable force to create a 15 N pinch force (32)	Voluntary opening prostheses	Required cable force to achieve a 50 mm prehensor opening (33)
TRS Hook, Grip 2S	33 N	Hosmer Hook 5XA, 2 bands	50 N
Hosmer APRL Hand, 52541	61 N	Hosmer Sierra Hand, ungloved	70 N
Hosmer APRL Hook, 52601	62 N	Hosmer Hook 5XA, 3 bands	71 N
Otto Bock Hand, 8K24	98 N	Hosmer Sierra Hook, Setting 2	82 N
Hosmer Soft Hand, 61794	131 N	Otto Bock Hook, Setting 2	94 N

2.2.4.2 Body-Maps

The body-maps were inspected visually and were summarized in the highest discomfort-intensity and its affected body-part by the experimenter. The analysis procedure and results for different subjects were discussed with the other authors.

2.2.5 Statistics and prediction of maximum forces

For statistical analysis SPSS version 20 was used, and a significance level of α=0.05 was maintained. Gender effects for the force magnitude were analyzed with a paired sample t-test. A linear multiple regression analysis was conducted to predict the maximum forces body-powered prosthesis users can create on the control cable from 1) shoulder width, arm circumferences, upper-arm length of the affected arm and remnant length as well as 2) gender, experience in body-powered prosthesis use, and cause of defect.

2.3 RESULTS

The maximum cable operation force averaged over all subjects was 257±124 N. The individual maxima ranged from 87 to 360 N (188±87 N) for female subjects and from 199 to 538 N (332±117 N) for males (Figure 2–3), which is a significant difference ($t(22)=9.89$, $p<0.001$).

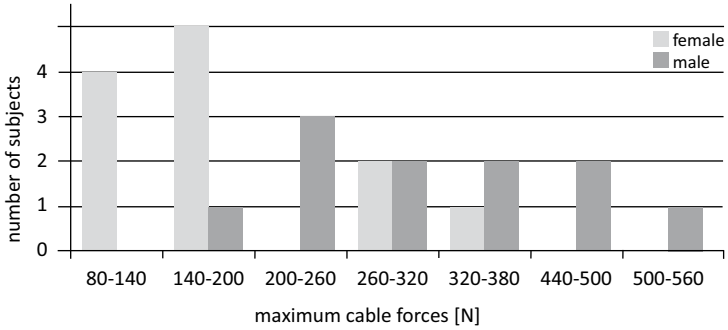

Figure 2–3. Male subjects attained significantly higher forces than female subjects ($t(22)=9.89$, $p<0.001$). Subjects created cable forces of 257±124 N (mean ± standard deviation). The maximum attained forces range from 87 to 360 N (188±87 N) and 199 to 538 N (332±117 N) for female and male subjects, respectively.

Assuming fatigue-free operation at 20% of the users' maximum cable forces (38), females can operate a body-powered prostheses fatigue-free up to 38±17 N, whereas males can handle forces up to 66±23 N.

The subjects' maximum cable forces and fatigue-limits were compared to the required operation forces of 1) voluntary closing prostheses creating a 15 N pinch force (32) (Figure 2–4) and 2) voluntary opening prostheses achieving a 50 mm prehensor opening with prostheses which can at least pinch 15 N (33) (Figure 2–5). The results indicate that three out of ten evaluated prostheses cannot be operated by all subjects even when exerting their maximum cable forces. More than 50% of the subjects will not operate eight of the 10 evaluated prostheses in daily live fatigue-free. One prosthesis included into the study, the Hosmer Soft Hand, cannot even be operated by a single user without exhaustion.

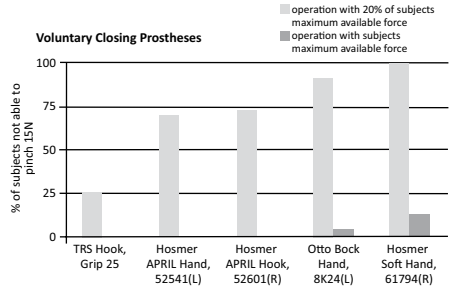

Figure 2–4. Pinching 15 N repetitively with five voluntary closing prostheses fatigue-free is impossible for 26 to 100% of prosthesis users. Fatigue-free operation is considered at 20% of users' maximum cable force (38) and is desired for ADL. Cable forces required to pinch 15 N with five voluntary closing prostheses vary between 33 and 131 N (32). The maximum strength of 13% of prosthesis users is insufficient to pinch 15 N with the Hosmer Soft Hand.

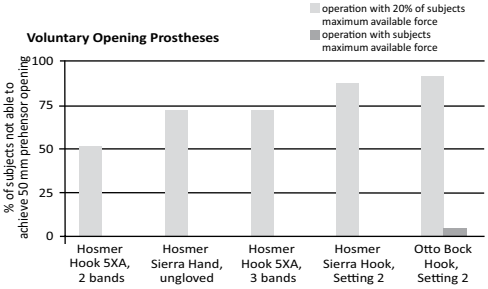

Figure 2–5. Achieving a 50 mm prehensor opening repetitively with five voluntary opening prostheses fatigue-free is impossible for 52 to 91% of prosthesis users. Fatigue-free operation is considered at 20% of users' maximum cable force (38) and is desired for ADL. Voluntary opening prostheses, which are able to pinch at least 15 N, require between 50 and 94 N cable force to achieve a 50 mm prehensor opening (33). The maximum strength of 4% of prosthesis users is insufficient to open the Otto Bock Hook.

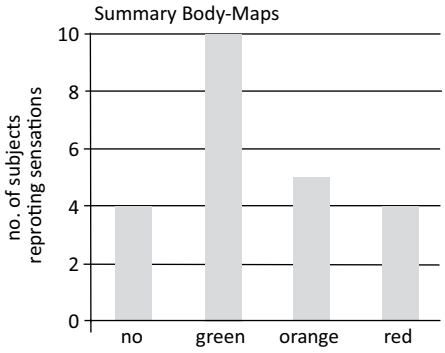

Figure 2–6. Number of reported sensations on the Body-Map after exerting maximum forces on the operation cable. Four subjects reported pain ('red'), five irritation ('orange') and ten a mild form of sensation ('green'). Four subjects did not report any sensation ('no').

Reported sensations after exerting the maximum cable forces are summarized in Figure 2–6 and Table 2–3. Sensations were mostly reported in armpit, neck/shoulders and upper back. Nine of the 23 subjects reported pain or discomfort, of which six reported the armpit as affected body-part. Detailed information on extend and locations of reported sensations can be found in Appendix A.

Reported sensations appear to vary randomly between the subjects, and are independent of the maximum force they could generate (as can been seen in Table 2–3). The Hosmer Hook 5XA with 3 bands requires a cable operation force of 71 N, which is the average operation force of all tested prostheses. With its individual maximum and fatigue-free cable operation forces, Table 2-3 indicates that all users can operate the hook, but only six of 23 subjects would be able to operate the prosthesis fatigue-free on daily basis.

Table 2–3. Reported sensations in the Body-maps were independent of the subjects' maximum cable forces. The Hosmer Hook 5XA with three bands requires 71 N cable operation force, the average operation forces over all prostheses. The individual maximum cable forces indicate that all users are capable to operate the hook, but only six of the 23 subjects could operate the hook fatigue-free on daily basis.

Subject no.	Maximum cable force [N]	Fatigue-free operation force [N]	Sufficient force to operate Hosmer Hook 5XA fatigue-free	Body-maps
18	86,6	17,3	No	irritation
19	100,4	20,1	No	none
11	117,9	23,6	No	pain
14	134,1	26,8	No	mild sensation
16	147,9	29,6	No	pain
12	164,1	32,8	No	none
4	165,4	33,1	No	mild sensation
8	181,6	36,3	No	pain
3	197,9	39,6	No	irritation
22	199,1	39,8	No	pain
9	212,9	42,6	No	none
23	229,2	45,8	No	mild sensation
21	259,2	51,8	No	mild sensation
5	272,9	54,6	No	mild sensation
1	277,9	55,6	No	mild sensation
15	290,4	58,1	No	mild sensation
17	307,9	61,6	No	mild sensation
2	360,4	72,1	Yes	none
13	360,4	72,1	Yes	mild sensation
7	369,2	73,8	Yes	mild sensation
6	441,7	88,3	Yes	irritation
10	490,4	98,1	Yes	irritation
20	537,9	107,6	Yes	irritation

The maximum cable force (MCF) can be predicted for the females by the following model:

MCF = -200.102+14.139 * *upper arm circumference affected arm*

The affected upper-arm circumference shows a Pearson correlation with the maximum cable force of 0.646 for female subjects (n=12, p =0.023). Other predictors and correlations were not significant. Anthropometric measures and subject characteristics are summarized in Appendix B.

2.4 DISCUSSION

This study aimed to quantify users' maximum cable operation forces and to relate these to a fatigue-free force range for the use of body-powered prostheses. In addition we aimed to identify extent and locations of discomfort provoked by the exertion of cable forces and to explore the predictability of maximum cable operation forces by the anthropometric measures of users. Subjects created maximum cable forces of 257 (124) N. The majority of subjects cannot use most body-powered prostheses fatigue-free on a daily basis. Nine subjects reported discomfort or pain after exerting maximum cable forces, of which six reported the armpit as affected body part. Pain and discomfort do not correlate with the maximum force a subject could generate. The affected upper-arm circumference can predict the maximum cable force exerted by females.

Comparing the attained maximum cable forces with the reported cable forces of Taylor (39) the maxima are comparable, although isolated movements of controls were measured. The results show that the required cable forces of available prostheses are generally speaking not befitted to the user's strength when corrected for fatigue-free operation. Accordingly, more than half of all users cannot operate eight out of the ten prostheses that were evaluated, which might explain the high rejection rates of body-powered prostheses (15). As a design recommendation for body-powered prostheses the fatigue-free operation force for the average female of 38 (17) N and for the average male of 66 (23) N should be considered. When for instance a higher pinch force with a voluntary closing prosthesis is needed than required for most daily activities, the fatigue-free boundary can be exceeded without further consequences. However, for repetitive daily tasks this fatigue-free boundary should not be exceeded. An alternative way to derive a design criterion could be to determine the cable force level that can be exerted, fatigue-free, by 90% of the users. Based on the results of this experiment, this would result in even lower allowed forces (<23 N), which is considered quite challenging for prosthesis design.

Discomfort and pain are reasons for prosthesis rejection (15,35,42) and occasional non-use in frequent wearers (43). In body-powered prostheses use the harness system can cause skin irritations and upper body pain (35). Supplementary to literature the results of this study show that the exertion of maximum forces provoke discomfort and pain for approximately 40% of the users. However, it is noted that daily activities may not require the user to exert maximum forces.

Reported locations of discomfort and pain are mainly the armpit, neck/shoulders and upper back, which is caused by the harness system. It was expected that anthropometric measures could predict user strength in terms of maximum cable forces. This might help clinicians to make a quick estimate whether a body-powered prosthesis is suitable for a patient. However, this study showed predictability of maximum forces only by the affected upper-arm circumference of females. Unfortunately, the strength of males is not predictable by anthropometric measures, since the predictions were not significant for this subject population.

2.4.1 Study limitations

The magnitude of maximum cable operation forces partly depends on the eagerness and motivation of subjects. The encouragement of the experimenter may contribute, but may not be sufficient to reach the maximum strength.

The results show a wide variability in achieved maximum forces over the subjects. Data of more participants might have allowed more (gender related) predictions of maximum cable forces by anthropometric data. However, the subject population was a representative group of (potential) prosthesis users with trans-radial defects, since the group covered of a wide variety of characteristics as indicated in Table 2–1 and Appendix B.

We concluded that pain and discomfort does not correlate with the magnitude of the maximum force achieved by the subjects. In other words, subjects who were able to attain higher forces were not experiencing more discomfort or pain than subjects who created significant lower maximum forces. This study did not investigate the subjects' individual discomfort thresholds of exerted cable forces. Also conclusions on the severity of the pain cannot be drawn by the subjective data of the Body-Maps. Furthermore, depending on his physical strength and desired daily activities, a user may not need the maximum force to operate his prosthesis. The recorded pain or discomfort associated with maximum forces therefore may or may not be experienced in daily life.

2.5 CONCLUSION

In conclusion, in many cases the user's strength is insufficient to operate body-powered prostheses fatigue-free on a daily basis. Exerting maximum cable forces provokes discomfort and pain, especially in the armpit. The fatigue-free operation forces for the average female of up to 38 N and for the average male user of up to 66 N should be considered as input design target of body-powered prostheses to conduct most daily activities. This implies that a significant number of users may not be able to achieve this group average in consideration of their personal fatigue-limit. The choice of a body-powered prosthesis should be based on the user's strength, which can be predicted for females only by the affected upper-arm circumference.

3

HIGH CABLE FORCES DETERIORATE PINCH FORCE CONTROL IN VOLUNTARY CLOSING BODY-POWERED PROSTHESES

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ABSTRACT

Background: It is generally asserted that reliable and intuitive control of upper-limb prostheses requires adequate feedback of prosthetic finger positions and pinch forces applied to objects. Body-powered prostheses (BPPs) provide the user with direct proprioceptive feedback. Currently available BPPs often require high cable operation forces, which complicates control of the forces at the terminal device. The aim of this study is to quantify the influence of high cable forces on object manipulation with voluntary-closing prostheses.

Method: Able-bodied male subjects were fitted with a bypass-prosthesis with low and high cable force settings for the prehensor. Subjects were requested to grasp and transfer a collapsible object as fast as they could without dropping or breaking it. The object had a low and a high breaking force setting.

Results: Subjects conducted significantly more successful manipulations with the low cable force setting, both for the low (33% more) and high (50%) object's breaking force. The time to complete the task was not different between settings during successful manipulation trials.

Conclusion: High cable forces lead to reduced pinch force control during object manipulation. This implies that low cable operation forces should be a key design requirement for voluntary-closing BPPs.

3.1 INTRODUCTION

3.1.1 Myo-electric prostheses

It is generally asserted that upper-limb prosthesis operation requires sufficient feedback to obtain adequate dexterous manipulation (44,45). Myo-electric prostheses require visual confirmation of movements of the terminal device as there is no other direct form of feedback about the action of the prehensor. Several approaches to pinch force feedback have been investigated in the last decades such as vibro-tactile feedback (18,19), mechano-tactile feedback (pressure on skin) (20-22), electro-tactile feedback (electro-cutaneous stimulation) (23-25), skin stretch (26), and force feedback spanning the joint (27). None of them have been implemented in commercial myo-electric prostheses and all except the latter target tactile feedback. However, in dynamic force feedback tasks, proprioception is the key player and tactile feedback has only an ancillary role (28).

Body-powered prostheses

The first body-powered prosthesis (BPP) was designed by Ballif in 1818 (5). Current BPPs still rely on the same principle: A shoulder harness captures the relative motion of shoulder and arm movements and transmits their action via a Bowden cable to operate a prosthetic prehensor. Two types of prehensors are used: Voluntary-Closing (VC) and Voluntary-Opening (VO) which open or close when the cable is pulled. The VC BPP provides the user with Extended Physiological Proprioception (EPP) (46). EPP extends the concept of proprioception to tools connected to the body, in this case a prosthesis. This has the inherent benefit of direct proprioceptive feedback about the prehensor's movement and forces through the movement and forces of the harness.

To date, body-powered hooks are equally preferred to myo-electric hands (35). Stated advantages of body-powered prostheses compared to myo-electric prostheses (15,47,48) include mass, robustness and cost-efficiency. However, BPPs are still far from optimal in spite of the advances since the patenting of the Dorrance split hook in 1912. Body-powered hands are less preferred than hooks (35). A user might prefer a prosthetic hand instead of a hook for cosmetic reasons, but then he needs to exert 1.5–8 times more mechanical work and will experience 2–27 times higher hysteresis or energy dissipation (32). Further advances in harness design (35), reduction of friction in the transmission (32,33), and weight reduction of the prosthesis (48) are possible. Fundamental improvement in BPP design could be realized by optimizing the relationship between the forces and displacements at the prehensor and those at the shoulder harness (49). Progress is currently impeded by the limited understanding of how cable forces influence grasping performance and comfort.

3.1.2 Cable forces in prosthesis operation

Current BPPs usually require high operating forces (49), which lead to pain and fatigue during or after operation (35) and may additionally disturb the feedback and control of pinch forces. Previous work in our group demonstrated that the control of operation forces decreases with higher cable forces (50).However, these experiments were done without prehensor and objects. This means the dynamic effects of prosthesis-object interaction and compensatory strategies of the user were not considered. Therefore, the effect of high cable operation forces for prosthesis-object-interaction remains unexplored.

This study aims to quantify the influence of high cable forces on the accuracy of pinch force control, when a VC BPP is used to grasp an object and transport it without exceeding pre-defined force boundaries. We hypothesize that high cable operation forces reduce the task performance in terms of the amount of successfully transported objects.

3.1.3 Approach

Able-bodied subjects were equipped with a by-pass socket and BPP. They were instructed to execute a repeatable abstract task of grasping an object and transferring it to another predefined position. The object transfer involves arm movements, which influence the pinch forces if the subject does not correct for this effect. Therefore the object transfer simulates the type of challenges that VC body-powered prosthetic users experience in daily activities. In order to inherently include interaction force limits in this manipulation task, a “mechanical egg” (20) was used which offers repeatable limits: at too little force subjects can’t lift it, and at an adjustable level it “breaks” mechanically. Abstract collapsible objects have been used in diverse studies investigating feedback and pinch force control (20,51), since they offer a natural challenging dynamic grasping task. As prosthetic users aim to execute grasping tasks as quickly as they would with an intact hand, time to execute the task was taken as an outcome measure. Subjects were asked to execute the task as quickly as they could without breaking the object. Breaking an object in daily life is inconvenient and is generally avoided.

3.2 MATERIALS AND METHODS

3.2.1 Subjects

Twelve able-bodied male subjects (11 right & 1 left handed, age: 30±8 (mean ± standard deviation) years old, height: 179±5 cm weight: 88±8 kg) participated in this study. The data of one of the subjects was excluded from data analysis because he was unable to successfully complete 80% of the trials. In addition, the force data of a second subject were not available for analysis. None of the subjects had experience operating BPPs. The Research Ethics Board of the University of New Brunswick, where the experiments were conducted, approved the experiments (REB #2014-064). All subjects signed an informed consent form prior to the experiments.

3.2.2 Apparatus

Subjects were fitted with a custom-made prosthesis consisting of a modified prehensor, which was attached to an adjustable bypass fitting, and linked to an adjustable "figure-of-nine" harness to provide the cable forces to close the prehensor (Figure 3–1). The equipment was manufactured and modified in the Atlantic Clinic for Upper Limb Prosthetics in Fredericton, Canada. The length of the socket was adapted to the subject's lower arm. Likewise, the harness could be modified and adapted to the subject's shoulder width and upper arm length. A standard 1/16'' (.159 cm) diameter stainless steel cable (C100) running through a cable housing for C-100HD cable (CH-100HD). To reduce friction in the cable a Teflon liner for heavy duty cable housing (CH100-HD) (all from Hosmer Dorrance Corporation, Chattanooga, USA) was placed in the inside of the cable housing. The coefficient of friction was reported to be 0.092 and assuming a maximum cable curvature of 90 degrees we would expect the static efficiency of force transmission of the Bowden cable to be more than 90% according to Carlson et al. (37).

Figure 3–1. Side-view (**a**) and back-view (**b**) of one subject wearing the custom-made bypass-prosthesis. The prehensor (**1**) is connected to the fitting. The prehensor (**1**) was connected via a Bowden cable (**3**) to the "figure-of-nine" harness (**5**). The Bowden cable forces were measured before and after the outer cable housing with force sensor 1 (**2**) and force sensor 2 (**4**).

3.2.2.1 Prehensor

The voluntary-closing Grip 3 prehensor (TRS Inc., Boulder, USA) was chosen because of its mechanical efficiency and linear relationships between cable operation forces and cable excursions as well as between cable operation forces and pinch forces (Figure 6 and Figure 10 of Smit and Plettenburg's study (32)). The relationship between the pinch force and the cable force of a non-deformable object was determined to be

$$\frac{F_{pinch}}{F_{cable}} = 0.64 \qquad (3\text{-}1)$$

The cable force required to start building up a pinch force is dependent on the prehensor's spring stiffness and the prehensor's opening. Thus, with small modifications, the prehensor could facilitate different cable force settings to generate the same pinch force. The original prehensor's torsion spring was replaced by interchangeable linear springs of different stiffness fixed at the prehensor's thumb lever (Figure 3–2). The settings consisted of either two parallel springs (0.11 N/mm each), or three parallel springs (1.7 N/mm each). These different settings then required either low or high cable forces to close the prehensor. The high force setting (~40-50 N) represents the required forces to operate a TRS hook, a Hosmer APRL hand or hook as shown in the study of Smit and Plettenburg (Figure 10 of Smit and Plettenburg's study (32)). The low force setting (~10-15 N) was chosen according to the preferred forces of prostheses users as shown in the results of a preliminary study of our group (50).

Figure 3–2. TRS hook with the internal torsion spring replaced by external linear springs in the high force setting (3 x 1.7 N/mm springs); 2 x 0.11 N/mm springs were used for the low force setting.

3.2.2.2 Test object: "mechanical egg"

Subjects needed to interact with a force-sensitive test object (Figure 3–3). The object was called a "mechanical egg" since it "breaks" when excessive pinch force is applied. This "mechanical egg" is the same device as designed and used in the study of Meek et al. (20). The original grasping surface of the egg was rounded to match the TRS finger's shape and covered with non-slip material (Dycem Ltd, Bristol, UK) at the finger and the thumb grasping surface, in order to enhance the grip quality. The weight of the object (and thereby the slipping force) remained constant during the experiments (385 g).

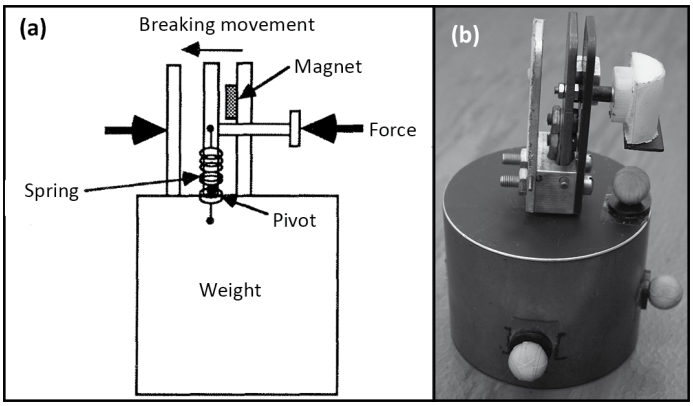

Figure 3–3. The "mechanical egg's" breaking mechanism (20) is shown in the left picture (**a**) and the experimental setup is shown to the right (**b**).

The object's breaking force was adjusted to a high and low breaking force setting, resulting in two pinch force margins at which the egg will not break or drop during manipulation. Table 3–1 contains the statically determined cable forces for both prehensor's spring stiffness settings at which the object slips of the prehensor (F_{slip}), thus the minimum required cable force to hold the test object, and the cable forces at which the object breaks for the low and high breaking forces (F_{break}). Figure 3–4 illustrates the relationship between cable and pinch forces. For training purposes, a third setting with an even higher breaking force was applied.

Table 3–1. The statically determined minimum required cable forces to hold the "mechanical egg" (F_{slip}) and its maximum allowed cable forces (F_{break}) for the two object's breaking force settings derived for the prehensor's two spring stiffnesses.

spring stiffness	**0.22 N/mm**	**0.22 N/mm**	**5.1 N/mm**	**5.1 N/mm**
breaking force	**high**	**low**	**high**	**low**
minimum required cable force (F_{slip}) [N]	5.3±0.3		28.8±0.3	
maximum allowed cable force (F_{break}) [N]	14.3±1.3	10.1±0.8	42.2±0.6	38.8±0.4

3.2.2.3 Measured signals

A custom-made timer was pressed by the subject to indicate the start and end of each trial. The subject reported the task completion time to the experimenter. Cable operation forces were measured at both the forearm and back of the subject. Forces were measured with two mini S beam 222N load cells (FUTEK Advanced Sensor Technology, Inc., Irvine, United States), amplified with a CPS amplifier (SCAIME S.A.S., Juvigny, France) and fed into the analogue input of a motion

capture system (Vicon Motion Systems Ltd., Oxford, United Kingdom) at 1000 Hz. The signals were recorded using Nexus 1.8.3 software (Vicon Motion Systems Ltd., Oxford, UK), and stored for off-line analysis after each trial. The recorded motion capture data were not used for the current study.

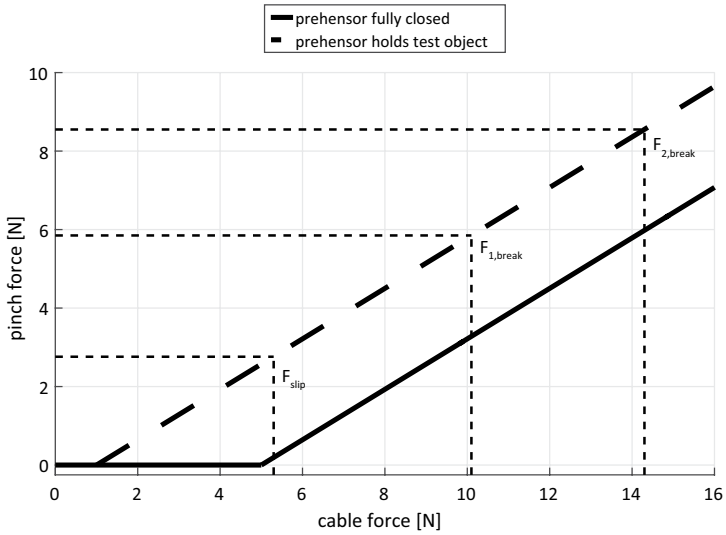

Figure 3–4. Cable to pinch force. The cable force to pinch force relationship is shown when the TRS hook is fully closed and when the test object is held utilizing the prehensor's low spring stiffness setting. The force at which the object slips out of the prehensor (F_{slip}), and the forces at which the "mechanical egg" breaks span the operating window in which the test object can be manipulated, for both the low ($F_{1,break}$) and high ($F_{2,break}$) breaking force settings. Note that the cable force at which the TRS hook starts to build up a pinch force on the test object is an estimation, since it was not experimentally determined. As a consequence the pinch force values are not representative.

3.2.3 Metrics

The number of failures and the time required completing the task served as the task metrics. Prosthetic users should be able to manipulate objects efficiently without breaking or dropping them.

3.2.4 Procedure

Each subject wore the bypass-prosthesis on the left arm (Figure 3-1) and was seated at a table (height: 73 cm). After adjusting the prosthesis and the seat to a height comfortable for each subject, the training session commenced. Subjects were instructed to operate the prosthesis using shoulder protraction of the right side, and humeral adduction and anteflexion of the

left side and had freedom of choice in their control movements. First, the subject familiarized themselves with the operation of the device by moving wooden blocks (2.5 x 2.5 x 2.5 cm) from the predefined low (1 cm above the table) to high position (16 cm above the table), start position B to target position C in Figure 3–5. Training continued with the "mechanical egg", starting with the stiffest setting, followed by the two test conditions, the high and low breaking force settings. Once the subject was familiar with the "mechanical egg's" function at the training setting, the timer (A in Figure 3–5) was introduced. For training purposes, each setting had to be conducted at least 10 times with 3 successful trials in a row before subjects moved on to a lower breaking force setting. Training ended when they could successfully execute the trial at the egg's low force setting.

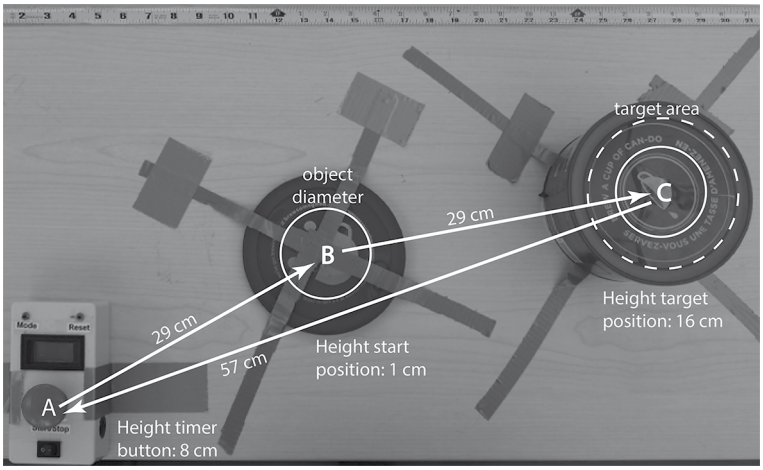

Figure 3–5. Visualization of one trial. The subject hits the self-timer button A to start the time measurement, moves 29 cm to grasp the object at the lower area B, then moves the object 29 cm to the higher target area C. After releasing the object, the subject needs to hit the self-timer to stop the time measurement.

The four experimental conditions were tested in a counterbalanced order, combinations of low and high cable forces and low and high breaking force setting. A trial consisted of starting the timer with the prosthesis, transferring the test object from the low to high position, and stopping the timer. The subjects were instructed to transfer the egg as quickly as possible without breaking or dropping it. Subsequently, the subject reported the time or a failure to the experimenter. Each of the four experimental conditions was tested 25 times, resulting in a total of 100 trials per subject. After the experiment was completed, the subjects were asked during a semi-structured interview which system they preferred, the low or the high cable force setting, and why they preferred that system.

3.2.5 Data analysis

For 11 subjects the number of failures and the average times over the 25 trials per condition were analyzed with a repeated measures ANOVA (IBM SPSS Statistics Version 20 – IBM Corporation, Armonk, United States).

The recorded Voltage of the force sensors was converted into Newton and filtered with a 3rd order filter (filtfilt function) at 10 Hz (Matlab Version 2013b – The MathWorks, Inc., Natick, United States) for 10 subjects. The peak forces (maxima) were determined for each successful trial and averaged per condition.

Friction losses were determined by comparing measured input and output forces of the Bowden cable.

3.3 RESULTS

The prehensor's high spring stiffness of 5.1 N/mm resulted in a 3.5 to 4 times higher cable operation force measured at the forearm than the low prehensor's spring stiffness (0.22 N/mm) as indicated in Table 3–2.

Table 3–2. The peak forces for successful trials measured at the forearm (2 in Figure 3–1) and at the back (4 in Figure 3–1) of the subject and averaged over all subjects per condition (values in mean ± standard deviation).

spring stiffness	**0.22 N/mm**	**0.22 N/mm**	**5.1 N/mm**	**5.1 N/mm**
breaking force	**high**	**low**	**high**	**low**
force@forearm (F1) [N]	12.6±0.9	10.7±0.9	43.5±2.1	42.0±2.5
force@back (F2) [N]	15.5±1.2	13.3±1.2	51.5±2.2	49.8±2.9
efficiencies Bowden cable	81%	80%	84%	84%

High cable operation forces resulted in significantly more unsuccessful trials ($F(10,1)=6.763$, $p=0.026$, Figure 3–6). The task completion time, however, was not significantly affected by the magnitude of the cable force ($F(10,1)=4.097$, $p=0.071$, Figure 3–7).

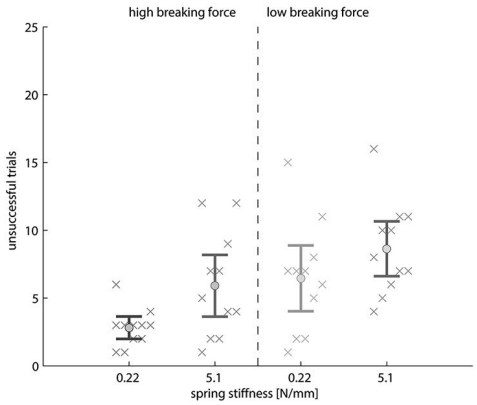

Figure 3–6. Number of unsuccessful trials. The number of unsuccessful trials out of 25 trials per condition are indicated by "x" per subject (N=11), averaged over all subjects ("o") with the 95% confidence intervals (whiskers). The results are compared for the high (left) versus the low (right) breaking force setting of the test object as well as the low (0.22 N/mm) versus high (5.1 N/mm) spring stiffness of the prehensor.

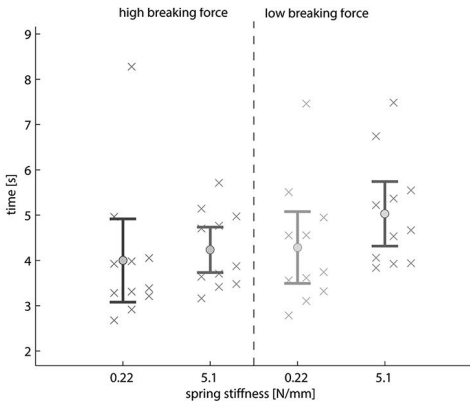

Figure 3–7. Task completion time. The time to complete the experimental task was determined by the average of all successful trials per condition per subject (N=11) indicated by "x". The error bars represent the average of all subjects ("o") with the 95% confidence intervals (whiskers). High (left) versus low (right) breaking force setting of the test object as well as the low (0.22 N/mm) versus high (5.1 N/mm) spring stiffness of the prehensor were compared.

Subjects exerted significantly less force on the control cable during the task execution at the object's low breaking versus the high breaking force condition (Table 3–2; force@forearm: $F(9,1)=114.608$, $p<0.001$; force@back: $F(9,1)=123.013$, $p<0.001$). The low object's breaking force resulted in significantly more unsuccessful trials than the high breaking force setting ($F(10,1)=25.817$, $p<0.001$, Figure 3–6). Subjects completed the experimental task significantly quicker at the high object's breaking force setting ($F(10,1)=25.346$, $p<0.001$, Figure 3–7).

The prehensor's spring stiffnesses and the object's breaking forces did not show interaction effects for the number of unsuccessful trials ($F(10,1)=0.225$, $p=0.646$) and the average task execution time ($F(10,1)=1.461$, $p=0.255$).

The outcome of the semi-structured interviews of the subjects showed that ten of the eleven subjects preferred the low spring-stiffness setting. The reported reasons for the low spring-stiffness system preference were the ease to control and distinguish pinch force and a higher long term comfort (less load on the axillar region, less tiring, less required effort).

3.4 DISCUSSION

This study aimed to quantify the influence of high cable forces on the accuracy of pinch force control, when a VC BPP is used to grasp an object and transport it, while not exceeding pre-defined force boundaries. To create two different cable force levels, springs with different endpoint stiffnesses were mounted on the prehensor. We hypothesized that high cable operation forces reduce the task performance. The results showed that higher cable operation forces resulted in more unsuccessful trials and thus an inferior task performance compared to lower cable forces, which is in line with our hypothesis. Subjective interview reports showed that subjects preferred the lower cable forces as well. Interestingly, high cable forces did not increase the task execution time. This finding might indicate that the subjects either did prioritize the task execution time over the successful task performance, or were not aware of their accuracy in controlling the decreased pinch force. A recent study has shown that high cable operation forces result in a larger deviation in the targeted cable forces (50). The TRS prehensor cable forces relate linearly to the pinch forces as shown in Figure 10 of Smit and Plettenburg's study (32). Consequently, a wider deviation of pinch forces could be expected at higher cable forces. This wider deviation seems to result in decreased pinch force control accuracy. Required cable forces for a 15 N pinch force range from 33 (TRS hook) to 131 N (Hosmer soft hand) for VC prehensors (32). The cable forces of this study ranged from 16 to 52 N (as measured at the subjects' back) for the two cable force conditions and show effects in pinch force control accuracy. This emphasizes the urgency of lowering the required cable operation forces for VC BPPs to achieve better pinch force control. Additionally, users of body-powered hands also complained about "slowness in movement, insufficient grip strength and high-energy expenditure" (15). These problems can be tackled with lower cable operation forces.

The difficulty of the task was manipulated by utilizing two breaking force settings of a "mechanical egg". The narrower the object's grasping force margin, the more critical the pinch forces on the object became. Thus the task gets more challenging, which is indicated by the higher number of failures and the longer task execution time in the low breaking force setting. Fragile objects were hypothesized to require more attention from the user and the manipulation to be more time-consuming than rigid objects. We did not find interaction effects between the magnitude of breaking and cable force. Thus irrespective of the task difficulty, we can conclude that higher cable operation forces deteriorate the pinch force control accuracy.

The differences of cable forces measured simultaneously at the forearm and at the back of each subject are striking (Table 3–2). These differences are mainly due to friction, a well-known disadvantage of Bowden cables, which increases with the curvature of the cable (37). In our experiment the friction losses result in efficiencies between 80 and 84% of the exerted forces, despite the Teflon liner in the outer cable housing to improve the efficiency of force transmission. According to Carlson, in a static set up an efficiency of 80% implies a cable curvature

of approximately 150 degrees (37). Since the angle was never more than 90 degrees in this set up, it would suggest that there is a different behavior of the Bowden cable during dynamic prosthesis operation. This corresponds to recent evidence presented at the ISPO Europe conference 2016: Preliminary results on the dynamic properties of different types of Bowden cables were discussed (52), that suggest decreasing efficiencies for increasing cable velocities. Unfortunately, in this experiment cable velocities were not measured, preventing further analysis on the impact of dynamic properties of the Bowden cable on the pinch force control accuracy and the human controller's abilities to anticipate for this apparently unknown behavior of the system. However, it is clear that the Bowden cable introduces additional inefficiencies above those of the prehensor (32,33). Interestingly, the measured cable forces during the object transfer task at the prehensor's 5.1 N/mm spring stiffness and object's low breaking force (Table 3–2) exceed the statically determined cable forces at which the "mechanical egg' breaks (Table 3–1). We speculate that dynamic effects might have allowed subjects to exceed the statically determined breaking forces.

3.4.1 Study limitations

Instead of amputees, the experiment was performed by able-bodied subjects without experience in prosthesis operation. The task performance might differ between subjects with and without arm deficiencies due to different anatomy and the lack of experience in prosthesis operation. Experienced users might have developed strategies to grasp several objects efficiently in activities of daily living (ADL). However, a predefined, non-varying grasping surface was utilized in this experiment, removing the need to develop different grasping strategies. The subjects learned quickly how to grasp the test object due to the intuitive operation of a voluntary closing prehensor and were already able to distinguish grasping forces during the training session. In a previous study (50), no differences in the deviations of the controlled forces were found between subjects with and without arm deficiencies. As a consequence, performance differences between prosthetic users and our healthy control subjects are not expected for these experimental conditions.

The prosthesis simulator was placed at the left arm, which was the non-dominant arm of 10 out of the 11 subjects. This is presumed to best reflect the actual situation of prosthesis users, who usually prefer to manipulate objects with their natural hand, making the affected side the non-dominant side. The effect of operating the prosthesis simulator with the dominant versus the non-dominant side was not investigated.

The task instructions of transferring the "mechanical egg" as quickly as possible from and to a predefined position without breaking the "egg" might have been interpreted in different ways. Subjects might have prioritized the task execution time over the number of failures or vice versa. If a penalty for failure would have been applied, the subjects would probably all have prioritized to execute the task successfully rather than completing the task as quickly as possible. This might also explain why we did not find significant effects in the task execution time for the two cable force settings.

3.4.2 Implications and future research

Design requirements for body-powered prostheses lack quantitative values especially when considering the user's capacities and demands. This study clearly shows that pinch force control should be improved when utilizing low cable operation forces in the voluntary closing prosthetic design. However, there might be a disadvantage to reducing cable operation forces in terms of perception. A preliminary study showed that cable operation forces between 20-30 N are the preferred forces with fixed cable excursions (50). Plettenburg et al. suggest a relationship between the strength of the user and the preferred operation forces. This relationship as well as the influence of cable excursions on the preferred forces need to be investigated. Additionally, the optimal ratio between cable operation forces and pinch forces is yet to be determined. Future research should prioritize achieving a satisfactory grip strength at the best possible sensory feedback as a design criterion (35).

A second crucial factor for the body-powered prostheses efficiency is the reduction of friction. The unpredictable behavior and the inefficiencies of the Bowden cable during dynamical task execution suggest a need for better solutions in body-powered prosthesis design. The effects of the Bowden cable on pinch force control accuracy are unclear. This impedes the development of better prostheses.

Another crucial factor for body-powered prostheses is the harness design: The primary concern for users is skin irritation, pain and exhaustion during or after operation of a body-powered prosthesis (15,35). The results of the semi-structured interviews imply a higher long term comfort (less load on the axillar region, less tiring, less required effort) with lower cable operation forces. However, users feel restricted in their movements due to the harness. New harness designs are required. One commercially available alternative to the harness is the Ipsilateral Scapular Cutaneous Anchor (53); a patch is glued to the back of the user and is connected to the Bowden cable. The ability to distinguish operation forces compared to the traditional harness and the range of possible operation forces is unclear. Additionally, with a new harness design the appearance as well as the ease of donning and doffing the prostheses could be improved, which are two additional user design preferences (15).

The study of Lum et al. shows that fragile object manipulation is inferior with a prosthesis than with the intact biological hand (51). However, the performance of the voluntary closing body-powered prosthesis user was exceptional compared to the other prosthetic users. This single user performed the task without breaking a fragile object as successful as the able-bodied controls. Although it was a single user, it emphasizes the high potential benefits users may gain with improved body-powered prostheses design. This study indicated the benefits of one of the body-powered prosthesis design criteria: a decreased cable operation force. More body-powered prosthesis design criteria should be quantified, like the required pinch forces to manipulate objects and the resulting transmission ratio between cable forces and pinch forces for optimal voluntary closing body-powered prosthesis operation.

3.5 CONCLUSION

The goal of this research was to quantify the influence of high cable forces on object manipulation with a voluntary-closing, body-powered prosthesis. Lower cable operation forces lead to better control as shown by fewer unsuccessful trials, even though lower cable forces had no effect on task execution time. For the experimental conditions studied, we conclude that a lower cable force leads to improved performance during object manipulation. Therefore, we argue that low cable operation forces should be a key design requirement for voluntary-closing body-powered prostheses.

4

PERCEPTION AND CONTROL OF LOW CABLE OPERATION FORCES IN VOLUNTARY CLOSING BODY-POWERED UPPER-LIMB PROSTHESES

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Submitted

ABSTRACT

Background: Operating a body-powered prosthesis can be painful and tiring due to high cable operation forces, illustrating that low cable operation forces are a desirable design property for body-powered prostheses. Perception and controllability of low cable operation forces have never been investigated, and can be quantified by force reproduction experiments. This study aims to quantify the accuracy of cable force perception and control for body-powered prostheses use in a low cable operation force range by utilizing isometric and dynamic force reproduction experiments.

Method: Twenty-five subjects with trans-radial defect conducted two force reproduction tasks; first an isometric task of reproducing 10, 15, 20, 25, 30 or 40 N and second a force reproduction task of 10 and 20 N, for cable excursions of 10, 20, 40, 60 and 80 mm. Task performance was quantified by the force reproduction error and the variability in the generated force.

Results: The results of the isometric experiment demonstrated that increasing force levels enlarge the force variability, but do not influence the force reproduction error for the tested force range. The second experiment showed that increased cable excursions resulted in a decreased force reproduction error, for both tested force levels, whereas the force variability remained unchanged.

Conclusions: In conclusion, the design recommendations for voluntary closing body-powered prostheses suggested by this study are to minimize cable operation forces: this does not affect force reproduction error but does reduce force variability. Furthermore, increased cable excursions facilitate users with additional information to meet a target force more accurately.

4.1 INTRODUCTION

Body-powered prostheses are operated by body movements of the user. In case of a voluntary closing body-powered prosthesis movements and forces are typically transferred by a shoulder harness to close the prosthetic fingers and generate grasping forces. Grasping forces of a voluntary opening body-powered prosthesis are generated by a spring. The prehensor is opened by the movements of the user, which is the opposite mode of operation compared to a voluntary closing device. Grasping objects with a body-powered prosthesis generates proprioceptive feedback of finger positions and grasping forces due to Extended Physiological Proprioception as described by Simpson (46). This is different compared to a myo-electric prosthesis, which relies on visual, auditory and tactile feedback from motor vibrations. In any case, in dynamic force feedback tasks proprioception remains superior to visual and tactile feedback (28). Therefore body-powered prostheses provide the user with superior feedback in daily activities, which theoretically results in substantial control of pinch forces during object manipulation.

For object manipulation, the prosthesis user estimates the pinch force required to lift the object and intends to produce the estimated force. If the exerted pinch force on the object is too high, the object might break, whereas if the pinch force is too low, the object might fall. Both scenarios lead to unsuccessful object manipulation, which discourages the user to use his prosthesis. The difference of the estimated and the produced force is a measure for how well the user is capable to perceive and control the exerted forces of his prosthesis on the object. Substantial perception and control of pinch forces results in successful object manipulation, which makes prosthesis use appealing. Users of body-powered prostheses desire appropriate grip control and strength (35). In voluntary closing body-powered prosthesis the pinch force is directly related to cable operation forces (32). However, current body-powered prostheses require high operation forces from users to generate appropriate pinch forces for daily activities (32,33). High cable operation forces do not only lead to fatigue and possibly painful use (35), but also deteriorate pinch force control accuracy in voluntary closing prostheses (54). Some (potential) users are even not capable to exert the cable forces required to operate low-efficiency body-powered prostheses (49). Low cable operation forces are therefore desirable for future prostheses design.

However, the user's perception and control at low cable operation forces might differ from that at high force levels. Static force perception experiments (55-58) investigated the systematic force reproduction error, i.e., the difference of target force, perceived in a first trial, and reproduced force, generated in the following trial. Motor control literature shows that in isometric force reproduction tasks the target forces are overestimated for low and underestimated for high force levels and that the crossover point from over- to underestimation in percentage of maximum voluntary contraction of the target force is dependent on the measured muscle group (55-58). Furthermore, motor noise, more specifically force variability, increases with increasing target

force levels (59,60) and for stronger muscle groups the motor noise is smaller than for weaker muscle groups (61).

Since a body-powered prosthesis is operated by a combination of shoulder protraction, humeral abduction and anteflexion, weak and strong muscle groups with their different characteristics are interacting when powering the prosthesis. Motor control literature does not provide quantitative data of force variability or noise for the combination of muscle groups used during body-powered prosthesis operation. Therefore the effect of cable operation force magnitude on force reproduction error and force variability remain unknown. In addition, a mechanical linkage (shoulder harness, Bowden cable, and hand mechanism) with its inefficiencies transfers body movements and forces into finger positions and pinch forces, respectively. Its influence on perception and control of cable operation force remains unknown.

Preliminary experiments (50) examined static force reproduction tasks with a body-powered prosthesis. The results indicated a crossover point from over- to underestimation between 20-30 N for controls (n=13) and between 10 and 20 N for prosthesis users (n=7, only three participants completed all examined conditions). The force variability increased with increasing target forces. Shortcomings of these preliminary experiments are that cable operation forces were investigated isolated from cable excursions, which only simulates holding a rigid object with a voluntary closing prehensor. Approaching the desired pinch force to hold an object when closing the voluntary closing prehensor was not investigated. Motor control literature suggests that additional position information next to the perceived and controlled forces might decrease the force reproduction error (62). Additionally, in the preliminary study (50) the group size of prosthesis users was insufficient to generalize the results.

The aim of this study is to quantify the accuracy of cable force perception and control for body-powered prosthesis use in a low cable operation force range by utilizing isometric and dynamic force reproduction experiments. We hypothesize that cable forces between 10-20 N will result in the smallest force reproduction error, based on preliminary experiments (50). Furthermore, we hypothesize that introducing cable excursion during the force reproduction task will provide subjects with additional information to complete the task successfully and we therefore expect a decreasing force reproduction error with increasing cable excursion at cable operation forces of 10 and 20 N, which were the examined forces. Furthermore, we hypothesize that the force variability will increase with increasing target force levels.

4.2 METHODS

4.2.1 Approach

Force reproduction experiments either request subjects to reproduce a force generated on the participant (56,57), or reproduce a self-generated force (55,58). We choose the second, to let the subject first reproduce a target force which is illustrated visually on a screen (visual bock), and consequentially receiving proprioceptive feedback of his body movements and tactile feedback of the exerted forces on the skin by prosthesis parts (harness and socket). Based on the perceived forces the subject reproduces the same force again without visual information (blind block). It is mainly the proprioceptive feedback perceived during the visual block, which enables the user to reproduce the same force during the blind block (62). This simulates prosthesis use: the user estimates a force required to manipulate an object (experimental: target force) and based on his experience of former perceived forces (experimental: visual blocks), he applies the required force (experimental: blind block).

The experimental setup should be unaffected by (mechanical) properties of available prehensors and therefore either 1) a threaded rod or 2) springs of different stiffness were mounted on the end of the control cable instead of a voluntary closing prehensor. The threaded rod setting simulates holding a rigid object with a voluntary closing prehensor at a constant cable excursion. The “variable-spring-stiffness” setting simulates the approach of a desired pinch force to hold an object with a voluntary closing prehensor.

Cable forces of interest were based on the examined cable force levels on TRS hook data of Smit and Plettenburg’s study (32), since the TRS hook requires the lowest cable force of all tested devices. At 10 N the TRS hook starts building up a pinch force. At 40 N the TRS hook pinches approximately 20 N. A pinch force of 20 N is reported to be sufficient to complete most daily activities with an upper-limb prosthesis (63,64). Additionally, the critical force, which is the force that humans can conduct without fatigue effects during continuous isometric contractions, should be considered as upper force boundary for prosthesis use. Monod determined the critical force at 15 and 20% of the maximum voluntary contraction (38). Considering maximum cable operation forces reported by Taylor (arm flexion: 280±24 N; shrug: 270±106 N; arm extension: 251±29 N (mean ± standard deviation)) (39) and Hichert et al. (combination of shoulder protraction, humeral abduction and anteflexion: 267±123 N) (49) as maximum voluntary contraction, the target forces should not exceed 40 N (251 N x 15%) to enable participants to complete all trials. The upper boundary for operation forces of 40 N is also supported by the results of Chapter 2, which suggest fatigue-free operation up to 38±17 N for female subjects. Based on this, we decided to examine six force levels (10, 15, 20, 25, 30 and 40 N) for the threaded rod setting.

In contrast to the reported magnitude of maximum cable excursion of 58±1.7 mm for arm extension (39) by Taylor, we measured maximum cable excursions of 160 to 260 mm in preliminary experiments. These experiments also showed that up to 50% of the maximum cable excursion

the subjects' operation force levels were unchanged. The cable excursion should therefore not exceed 80 mm (160 mm x 50%). Based on this, we decided to examine five cable excursions (10, 20, 40, 60 and 80 mm) for the "variable-spring-stiffness" setting. The five excursions were tested at two force levels, 10 and 20 N, at which the crossover point from overestimation to underestimation of target forces for prosthesis users were found in preliminary experiments (50). This results in ten force-excursion conditions.

4.2.2 Participants

Twenty-four adults (12 females, age: 49±13 years, height: 175±8 cm, weight: 75±14 kg) with congenital and acquired unilateral trans-radial defects participated. All participants were free of neurological, muscle, joint or motor control problems concerning the upper extremity or the torso (exclusion criteria). A total of 16 participants had a left deficiency, 15 had a congenital defect, 13 had experience with body-powered prostheses and five are current body-powered prosthesis users.

This study was approved by the medical ethical committee of University Medical Centre Groningen (UMCG) (NL41112.042.12). The participants were recruited from University Medical Center Groningen, Erasmus Medical Center, Rotterdam, and the rehabilitation institute De Hoogstraat, Utrecht.

4.2.3 Materials

A custom-made prosthesis simulator (Figure 4–1) was connected to the participant's prosthesis. For two participants, who did not own a prosthesis, the bypass-prosthesis was placed on a temporary WILMER Open Fitting socket [7]. For two other participants the bypass-prosthesis was attached to the remnant arm since its length was sufficient for a firm connection. The prosthesis simulator consisted of an adjustable "figure-of-nine" harness linked to a standard 1/16'' (.159 cm) diameter stainless steel cable (C100, Hosmer Dorrance Corporation, Chattanooga, USA). The end of the control cable, which was positioned in a U-profile, was attached to either 1) a threaded rod or 2) springs of different stiffness. The steel cable was interrupted by two force sensors (FLLSB200 222 N, FUTEK, Irvine, USA), one before and one after the stainless steel cable housing for C-100HD cable (CH-100HD, Hosmer Dorrance Corporation, Chattanooga, USA). To decrease friction in the cable a Teflon liner for heavy duty cable housing (CH100-HD, Hosmer Dorrance Corporation, Chattanooga, USA) was placed in the inside of the cable housing. The U-profile was fixated to the thermoplastic shell with 3 mm Neoprene at the inside. In the U-profile one of the two force sensors was placed and one displacement sensor (13FLP100 A, Sakae, Zhejiang, China). The two force sensors were amplified (CPJ, Scaime, Juvigny, France) and sampled together with the displacement sensor at 50 Hz (NI USB-6008, National Instruments, Austin, USA), and finally stored using a custom LabVIEW program (LabVIEW 2012, National Instruments, Austin, USA).

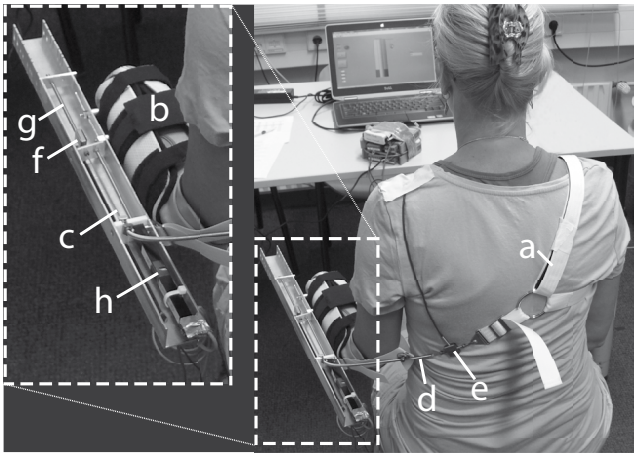

Figure 4–1. The measurement set-up consisted of a "figure-of-nine" harness (**a**) and thermoplastic shell (**b**) which are connected through a Bowden cable (**c**) running through a cable housing (**d**). The cable is interrupted by two force sensors (**e** & **f**), which measure the cable forces before (F_{back}) and after (F_{arm}) the cable housing respectively. In this figure a thread-rod (**g**) is illustrated leading to disabled cable excursions. The thread-rod is interchangeable with springs of different stiffness, which resulted in different cable force-excursion characteristics. A displacement sensor is recording cable excursions (**h**).

To investigate ten different force-excursion conditions, ten interchangeable springs with varying spring stiffness and pretensions were utilized as shown in Table 4–1.

Table 4–1. Stiffness and pretension of the utilized springs in each condition (values in mean ± standard deviation).

Condition	Spring stiffness	Spring pretension
10 N – 10 mm	0,44 ± 0,06 N/mm	5,5 ± 0,6 N
10 N – 20 mm	0,19 ± 0,04 N/mm	6,3 ± 0,8 N
10 N – 40 mm	0,20 ± 0,01 N/mm	2,0 ± 0,3 N
10 N – 60 mm	0,08 ± 0,00 N/mm	5,6 ± 0,1 N
10 N – 80 mm	0,44 ± 0,06 N/mm	4,0 ± 0,1 N
20 N – 10 mm	1,50 ± 0,18 N/mm	5,3 ± 1,6 N
20 N – 20 mm	0,57 ± 0,01 N/mm	8,9 ± 0,2 N
20 N – 40 mm	0,26 ± 0,01 N/mm	10,0 ± 0,1 N
20 N – 60 mm	0,22 ± 0,00 N/mm	7,2 ± 0,1 N
20 N – 80 mm	0,21 ± 0,04 N/mm	5,2 ± 1,1 N

4.2.3.1 Maximum force measurements

Another similar custom-made prosthesis simulator (Figure 4–2) was utilized to measure the participants' pre and post experimental maximum forces. Cable excursions were disabled in this setup. The Bowden cable was interrupted by a force sensor (S-Beam load cell ZFA 100kg, Scaime, Juvigny, France). The measured forces were amplified (CPJ, Scaime, Juvigny, France), sampled at 1 kHz (NI USB-6008, National Instruments, Austin, USA), and finally stored using a custom LabVIEW program (LabVIEW 2012, National Instruments, Austin, USA).

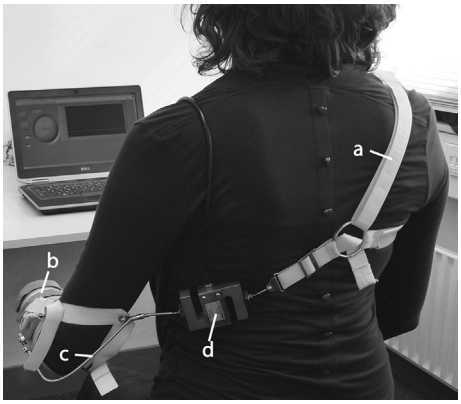

Figure 4–2. Measurement set-up for maximum force measurements: the "figure-of-nine" harness (**a**) and thermoplastic shell (**b**) are connected through a Bowden cable (**c**), which is interrupted by a force sensor (**d**). In this set-up cable excursions are disabled.

4.2.3.2 Questionnaires

To analyze the given task and the used system with its force -excursion combinations and the differences between the different conditions, subjective data of perceived workload were gathered via the Nasa Task Load Index (NASA-TLX) questionnaire (Desktop Version 2.1.2, developed by David Sharek, NASA Ames Research Center, Moffett Field, USA). A Dutch translation of the questionnaire was provided. The questionnaire assesses the total workload divided into six subscales: Mental Demand, Physical Demand, Temporal Demand, Performance, Effort, and Frustration.

Furthermore, subjects were requested to indicate regions of no, mild or severe discomfort on a map of the body (Body-Map) by coloring the respective body parts green (touchiness), orange (irritation), or red (pain) (Figure 4–3).

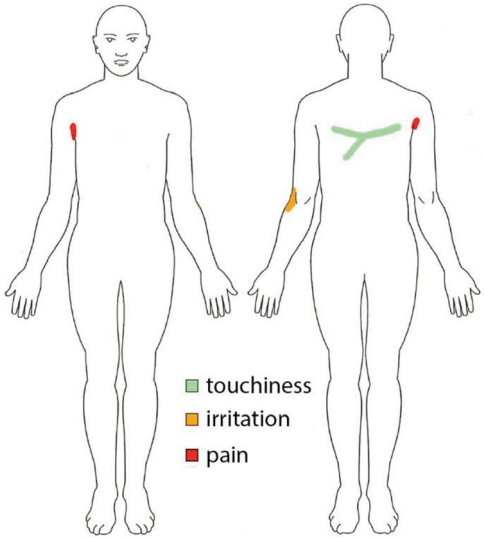

Figure 4–3. Body-Map coloured by one subject indicating pain (red) in the right arm pit, irritation (orange) at the back of the left elbow and touchiness (green) on a stripe of his back.

To monitor post experimental pain and fatigue effects, a few days after the experiment each participant was asked in an email whether he/she had experienced any post-experimental pain the day of the measurement or the following days, and if so in which part of the body.

4.2.4 Procedure

The chronological experimental procedure is shown in Figure 4–4. First, subjects were requested to exert their maximum force on the cable utilizing the equipment shown in Figure 4–2. Three measurements were taken with a duration of 3 seconds each. This procedure was repeated at the end of the experimental procedure to monitor physical fatigue caused by the experiment. Then the subjects conducted the force reproduction experiments, consisting of two parts, six trials with cable excursion disabled, followed by ten trials with cable excursion. After completing each of these 16 trials the subject was requested to fill in a Nasa-TLX questionnaire. The individual relevance of each of the six subscales to the total workload was supplemented by a paired comparison of the six subscales, ascertained during the first and last questionnaire. The Body-Map questionnaire was provided four times: after the pre and post maximum force measurements as well as after the force reproduction experiments without and with cable excursion.

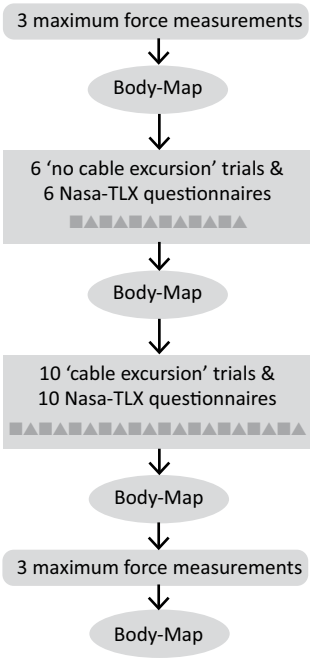

Figure 4–4. The flowchart of the experimental procedure illustrates the chronologic order of maximum force measurements (rounded rectangles), body-map questionnaires (ellipses), force reproduction experiments with and without cable excursion (rectangles) with alternating force reproduction trials (squares) and Nasa-TLX questionnaires (triangles).

For the force reproduction trials the measurement set-up of Figure 4–1 was fitted to the subject. During the 'no cable excursion' trials a threaded rod was placed in the U-profile disabling cable excursion. For the 'cable excursion' trials, the threaded rod was replaced by linear springs of different stiffness. Six force levels (10, 15, 20, 25, 30 and 40 N) for the 'no cable excursion' trials and ten force-excursion combinations (10, 20, 40, 60 and 80 mm each at 10 and 20 N) for the 'cable excursion' trials were examined resulting in 16 test conditions. Before each trial, the subject was allowed one training run at 22 N to familiarize himself with the task. Figure 4–5 shows the experimental procedure of the six 'no cable excursion' trials. The order of the force levels (part 1) and force-excursion conditions (part 2) were counterbalanced over participants. One trial consisted of eleven alternating visual and blind blocks. One block lasted 5 seconds followed by a 2 second break, resulting in a duration of 152 seconds per trial. During a visual block the reference and produced force measured on the arm of the subject (F_{arm}) was shown on the laptop screen, whereas during a blind block only the target force was displayed. In other words, during the visual blocks subjects reproduced the target force based on the visual information on the screen, whereas during the blind blocks subjects based the magnitude of the reproduced force on the perceived force during a visual block. Participants were instructed to produce the force as stable as possible. During the 'cable excursion' trials visual feedback to the subjects' arm was disabled with a hairdressers cloth tightened to the walls, as the arm position would have given information about the cable excursion. Subjects had the opportunity to practice the given task for 120 seconds. For the 'cable excursion' trials subjects were given 60 seconds to become

accustomed to the new condition. In the event that a subject experienced (concentration) difficulties in one block, another visual and blind block was added to the condition to complete the measurement.

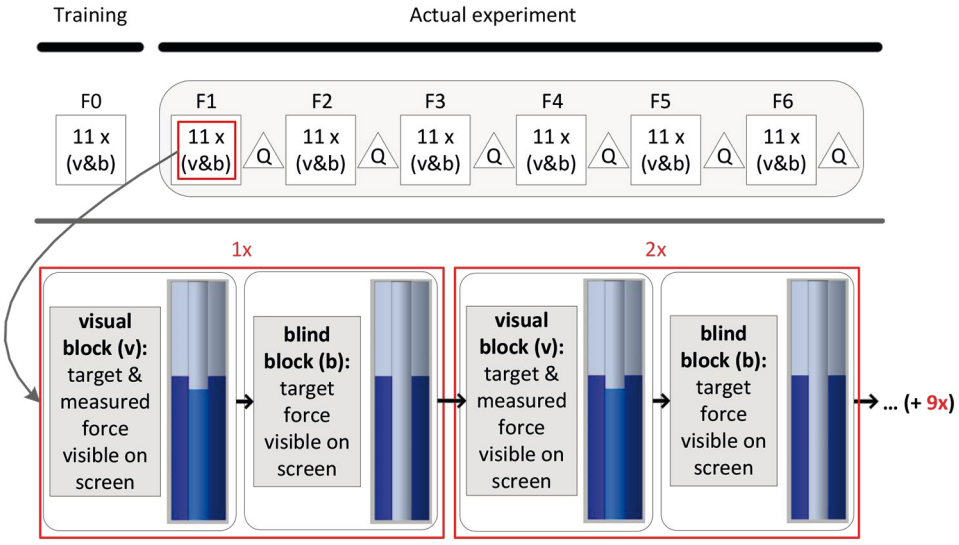

Figure 4–5. Flowchart illustrating the experimental procedure of the 6 'no cable excursion' trials as shown in Figure 4–4. After practicing the force reproduction task at 22 N (F0), six force levels (10, 15, 20, 25, 30 and 40 N) were examined during 11 alternating visual and blind blocks. The force reproduction task at each force level (squares) was followed by a Nasa-TLX questionnaire (triangle). The order of force levels (F1 to F6) was counterbalanced over the subjects. The outer (purple) bars indicate the target force; the inner (blue) bar indicates the measured force.

4.2.5 Data analysis

4.2.5.1 Metrics

Participants' performance was assessed by the force reproduction error, which is the difference between target and reproduced force, and the force variability, which is the noise of the reproduced force. These metrics were determined from the cable forces measured at the back of the subject.

The last 2.5 seconds of measured force were analyzsed by calculating the mean and standard deviation (Figure 4–6). Because the perceived force during the visual block must be reproduced during the blind block, the force reproduction error (FRE) per block was calculated as the average force of a blind block minus the average force of the foregoing visual block (Equation 4–1). The results per block were then averaged over all blocks of the trial to obtain the overall force reproduction error (per subject, per force level) (Equation 4–2). The first visual and blind blocks of each trial were eliminated from data analysis.

$$FRE_{block_i} = mean(F_{blind,block_i}) - mean(F_{visual,block_i}) \quad (4–1)$$

$$FRE = \frac{1}{10}\sum_{i=2}^{11} FRE_{block_i} \quad (4–2)$$

The force variability (FV) results from the standard deviation of the blind blocks (Equation 4–3) averaged over all analyzed blocks (Equation 4-4).

$$FV_{block_i} = std(F_{blind,block_i}) \quad (4–3)$$

$$FV = \frac{1}{10}\sum_{i=2}^{11} FV_{block_i} \quad (4–4)$$

The force reproduction error and force variability were determined for each condition (six force levels for 'no cable excursion' and ten force-excursion combinations for 'cable excursion' trials).

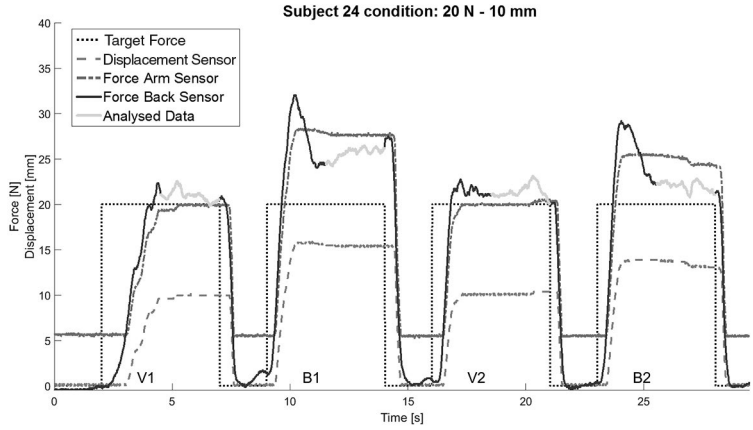

Figure 4–6. The raw data of the first 30 seconds of a typical trial, condition 20 N – 10 mm, represents the target force of 20 N, the approximate 10 mm cable excursion measured by the displacement sensor and the two cable forces measured at the arm (F_{arm}) and the back (F_{back}) of the subject. Visual blocks (V1, V2) are alternating with blind blocks (B1, B2). The last 2,5 seconds of each block were used for analysis.

4.2.5.2 Maximum force measurements

The highest values of the three pre and three post maximum force measurements were determined. Only trials where the maximum force was attained within the predetermined 3 seconds were included (114 of 150 trials). The maxima of the three pre and post measurements were taken to analyze for fatigue effects.

4.2.5.3 Statistics

For statistical analysis SPSS version 20 was used. Pre and post experiment maximal force levels were compared using a paired Student t-test. Repeated measures ANOVAs were used to determine the experimental effects ('no cable excursion' trials: target force; 'cable excursion' trials: target force × excursion) for force reproduction error and force variability. A significance level of α=0.05 was maintained.

4.3 RESULTS

The force reproduction error for 'no cable excursion' trials showed no difference for the measured target forces between 10 and 40 N ($F(5,19)=0.936$, $p=0.48$, Figure 4–7). The target force was overestimated for all force levels, and consequentially we did not find a crossover point from over- to underestimation. With increasing target force the force variability was increasing ($F(5,19)=23.767$, $p<0.001$, Figure 4–8).

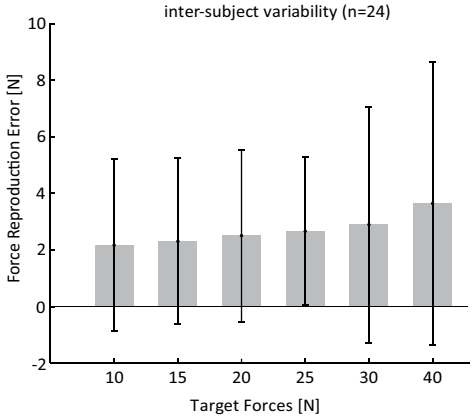

Figure 4–7. The force reproduction error for the 'no cable excursion' trials shows no significant differences between the tested conditions of target forces between 10 and 40 N. The bars indicate the group's average and the whiskers the standard deviation.

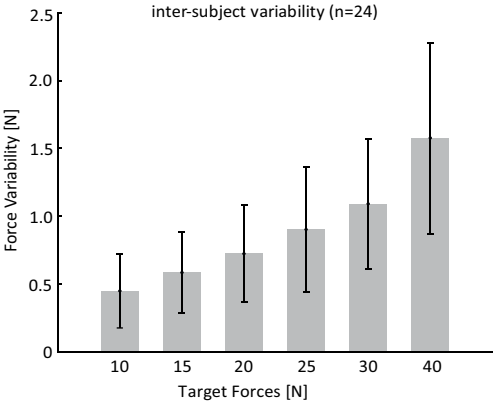

Figure 4–8. The force variability increases with increasing target force for the 'no cable excursion' trials. The bars indicate the group's average and the whiskers the standard deviation.

In the 'cable excursion' trials the force reproduction error was decreasing with increasing cable excursions for both target forces 10 and 20 N ($F(4,20)=8.865$, $p<0.001$, Figure 4–9), whereas no difference in force variability was found for increasing cable excursions at both target forces ($F(4,20)=1.878$, $p=0.154$), Figure 4–10). No difference in force reproduction error between

10 and 20 N target forces was found for the 'cable excursion' trials ($F(1,23)=2.554$, $p=0.124$, Figure 4–9). The force variability increases for increasing target forces for the 'cable excursion' trials ($F(1,23)=9.576$, $p=0.05$, Figure 4–10). The target force was overestimated for all conditions, except the 10 N – 80 mm condition. As a result, we found a crossover point from over- to underestimation for a target force of 10 N between 60 and 80 mm cable excursion, whereas we did not find a crossover point for a target force of 20 N.

The pre and post maximum force measurements did not differ ($T(24)=-0.50876$, $p=0.61557$).

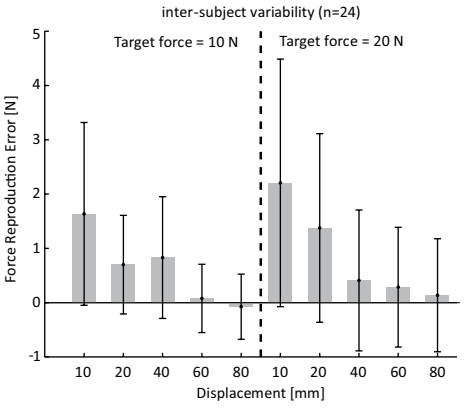

Figure 4–9. The force reproduction error decreases with increasing cable excursion for the 'cable excursion' trials for both target forces of 10 and 20 N. The force reproduction error does not differ between force levels. The zero line indicates when the target force is met. A negative force reproduction error indicates a lower reproduced force than target force. The bars indicate the group's average and the whiskers the standard deviation.

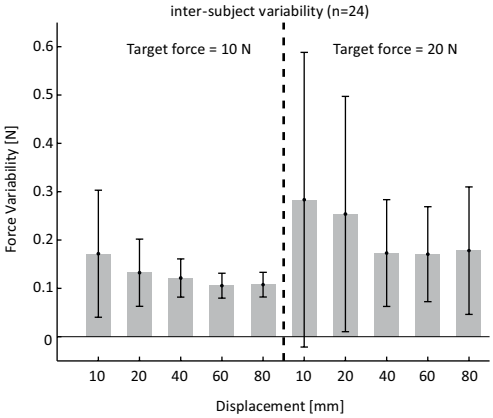

Figure 4–10. The force variability remains constant with increasing cable excursion for 'cable excursion' trials. The force variability is lower for 10 N target force than for 20 N. The bars indicate the group's average and the whiskers the standard deviation.

4.3.1 Subjective data

The NASA-TLX questionnaires did not show any differences between the tested conditions for the measured indexes (mental, physical and temporal demand, overall performance, frustration level and effort).

The Body-Maps indicated that not only the magnitude of forces but also the duration of the experiment seems to provoke discomfort and pain, since multiple subjects indicated discomfort and pain after the 'cable excursion' trials, where the examined forces did not exceed 20 N. After the maximum force measurements (pre and post) four subjects of the twenty-four subjects

reported pain in neck, upper back, shoulder or axilla. One subject reported pain in the upper back after completing the 'no cable excursion' trials, and four subjects had pain in neck, upper back, shoulder or their remnant arm (two subjects) after completing the 'cable excursion' trials. One subject reported pain at all four monitored moments and another subject indicated pain at three of the four monitored moments. Both subjects wanted to continue the experiment despite the experienced pain.

Fourteen subjects had no post-experimental pain the next day(s), eight reported that they had aching arms, shoulder or necks varying from light to heavy, one was not able to sit comfortably during the experiment with resulting muscle pain in his right leg, two were mentally tired after the experiments and two did not reply.

4.4 DISCUSSION

Contrary to our expectations, we did not find a difference in force reproduction error for measured target forces between 10 and 40 N during the 'no cable excursion' trials, whereas the force variability increased with increasing target forces, as hypothesized. The 'cable excursion' trials showed, as hypothesized, a decreasing force reproduction error with increasing cable excursions, for both target forces 10 and 20 N, whereas we did not find any difference in force variability at increasing excursions. The target forces were overestimated for all conditions, except for the 10 N -80 mm condition.

The *'no cable excursion' trials* were designed to simulate grasping a rigid object with a voluntary closing body-powered prosthesis. When gasping a rigid object with a voluntary closing body-powered prosthesis, the pinch force increases proportionally with the cable force. This occurs without a change in cable excursion and thus a change in prehensor opening. The increasing force variability for increasing target forces indicates a higher deviation of cable forces for higher forces. As a result, the deviation of pinch forces on an object increases with increasing required pinch forces. The controllability of pinch forces therefore decreases for increasing force levels. With current voluntary closing prostheses this implies less control on for example heavy objects where higher pinch forces are required. When a pinch force exerted on an object is too small the object might slip off the prehensor and fall, when the pinch force is too high the object might break. Both, results in unsuccessful object manipulation, which discourages the user to manipulate objects with his prosthesis. We expected a crossover point from over- to underestimation around 10 to 20 N, as in preliminary experiments obtained for prosthesis users (50), but surprisingly the overestimation did not decrease with increasing force levels. This might be explained by a shorter force reproduction duration and less repetitions per force level. The overestimation of target forces indicates that the exerted pinch force on a rigid object would be higher than intended by the body-powered prosthesis user at low cable operation forces.

However, for the tested force levels the offset of estimated and produced force on an object is expected to remain constant, based on the unchanged force reproduction error. The relationship between cable operation and pinch force can be described by $F_{pinch} = k*F_{cable}$ for voluntary closing body-powered prosthesis. Since the proportionality constant k is smaller than 1 for current voluntary closing prostheses (32), the effect of the overestimated cable force is smaller for the pinch force. For example: The proportionality constant k for the TRS hook is 2/3 (32). A deviation of ±3 N in cable force results in a deviation of ±2 N in pinch force. Overall the force reproduction error has a larger impact than the force variability for the measured forces up to 40 N, ±3 N versus ±1.5 N. In other words, for the manipulation of light objects has the difference between estimated and produced pinch force a larger impact than the ability to hold a pinch force at a constant level.

The *'cable excursion' trials* were designed to simulate approaching a desired pinch force to hold an object with a voluntary closing body-powered prosthesis. Building up the cable force and increasing the cable excursion closes the voluntary closing prehensor. When the voluntary closing prehensor is fully closed or touches an object a pinch force is created. From the experimental results of the 'cable excursion' trials we learned that increasing cable excursions may help to estimate and approach the desired pinch force more accurately. Increasing cable excursions do not affect the deviation of produced pinch forces. This implies better control of pinch forces when the voluntary closing prosthesis requires a long stroke to close the device. Or, since smaller objects require a longer closing stroke than larger objects, the pinch force on small objects can be controlled better than on large objects with a voluntary closing prosthesis. This counts for a voluntary opening prosthesis vice versa: the pinch force on large objects can be controlled better than on small objects. This might be taken into account when operating the voluntary opening and voluntary closing prosthetic terminal device developed by Berning et al. (65,66), a body-powered prosthesis which allows to switch between voluntary opening and voluntary closing operation mode. Although cable excursions of 80 mm show the lowest force reproduction error it is questionable whether large cable excursion is feasible for practical use. Utilizing this amount of cable excursion during grasping tasks implies that the prosthesis has to be held far away from the body, which makes object manipulation impractical, especially during feeding tasks. Current voluntary closing and voluntary opening body-powered prostheses demand cable excursions of up to 53 mm to fully close or open the prehensor respectively (32,33), and therefore we consider cable excursions of 53 mm as clinically approved. Practicality of cable excursions higher than 53 mm need to be examined in daily activities before recommending it for body-powered prosthesis design. Cable routing across the laterlae epicondyle enables users to operate a voluntary closing prosthesis with elbow flexion and might increase the functional cable excursion.

Interestingly, differences in the measured forces at the back and at the forearm of participants typically ranged between 2 to 3 N, but incidentally even up to 9 N. Such differences occur due to

friction losses of the Bowden cable. However, irrespective of the magnitude of friction losses we found significant differences in the force reproduction error between conditions for the 'cable excursion' trials. Hence, the magnitude of friction should not have influenced the outcome of these experiments.

Note that in the present study we provided visual feedback of the force measured at *the forearm*; whereas subjects received proprioceptive feedback from back muscles. We chose to provide feedback of the force measured at the forearm of the subject to the screen, since this cable force is directly related to the created pinch force of the voluntary closing body-powered prosthesis (32). After all, the user gets visual information of the created pinch forces when manipulating deformable objects with his prosthesis.

The experiments imitate intensive prosthesis use. All subjects could complete the full experiment, which suggests that the tested range of 10 to 40 N cable operation force is feasible for daily prosthesis operation. Interestingly, although pre and post experimental maximum force measurements did not show differences, eight subjects reported post experimental pain the next day(s). Furthermore, not only the magnitude of applied forces (maximum force measurements), but also the duration of the experiment seemed to provoke discomfort and pain as indicated by the results of the Body-Maps. Unfortunately, the subjective data of the Body-Maps does not include the severity of the pain, which makes interpretation of this subjective data difficult.

The perceived workload reported in the NASA TLX questionnaires did not differ between conditions. This implies that subjects do not show any preference for one of the tested force or force-excursion combinations.

4.4.1 Study limitations

We only tested for force levels between 10 and 40 N. In preliminary experiments we observed inferior control and perception of cable forces lower than 10 N. The observed friction losses in the Bowden cable probably also complicate control and perception of forces below 10 N proportionally more than for higher force levels. Force levels higher than 40 N would probably lead to fatigue during long-term operation. Of course in clinical practice the individual fatigue force level should be considered. Furthermore, cable excursions are only investigated for two low force levels of 10 and 20 N. Between these two force levels we expected the crossover point from over- to underestimation based on preliminary experiments (50), but we found over-estimation for all tested force levels.

A second limitation in our study was the rather abstract task, focused on obtaining results that could generalize over different prehensor types. Therefore, we chose to simulate prosthesis behaviour by utilizing springs of different stiffness and disabling cable excursion by a thread-rod. Of course, this experimental set-up is different than manipulating objects with a body-powered prosthesis, but gave us the opportunity to test different prehensor settings to make an informed choice of voluntary closing body-powered prosthesis design parameters. Also the duration and

intensity of the experiment were considerable. Participants were requested to reproduce a force at one force level with many repetitions in short time. Since the pre and post maximum force measurements did not show a significant difference, we conclude that the data was not influenced by physical fatigue effects, which is in line with the answers given in the NASA-TLX questionnaire. Also the long duration of the experiments (±2 hours) might have influenced the participants' performance due to mental fatigue, although the NASA-TLX questionnaire did not indicate mental fatigue.

4.4.2 Further research and implications

Users of upper limb prostheses have shown a preference for electric hands and body-powered hooks (n=242) (43). Body-powered hands show, compared to body-powered hooks, low mechanical efficiency (32,33) and are probably therefore often rejected. The maximum attainable efficiency of body-powered hands is insufficient to operate them with low cable forces as those measured. Cosmetically, however, prosthesis hands seem to be more appealing than hooks. A possible solution might be found in introducing power assistance systems to body-powered prosthesis hands. The results of this study could serve as input design requirements. Output requirements of such a system in terms of desired pinch forces for daily activities remain unknown.

4.5 CONCLUSION

The aim of this study was to quantify the accuracy of cable force perception and control for body-powered prosthesis use in a low cable operation force range by utilizing isometric and dynamic force reproduction experiments. For the experimental conditions studied, the following can be concluded:

- Contrary to our hypotheses, force reproduction accuracy did not depend on the tested force levels (10 – 40 N): the target force was constantly overestimated during the force reproduction task.
- As hypothesized, motor noise significantly increased with increasing force levels.
- As hypothesized, the presence of cable excursions contribute to a higher force accuracy, as compared to isometric for reproduction.

When translating cable forces proportionally to pinch forces of a voluntary closing body-powered prosthesis the results imply a higher deviation of pinch forces at higher force levels due to motor noise. The estimation error of created pinch forces on rigid objects does not vary for the examined low force levels, but the created pinch force is constantly higher than intended. A long closing stroke for voluntary closing body-powered prosthesis accommodates the right estimation of pinch forces on objects.

5

IPSILATERAL SCAPULAR CUTANEOUS ANCHOR SYSTEM: AN ALTERNATIVE FOR THE HARNESS IN BODY-POWERED UPPER-LIMB PROSTHESES

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ABSTRACT

Background: Body-powered prosthesis users frequently complain about the poor cosmesis and comfort of the traditional shoulder harness. The Ipsilateral Scapular Cutaneous Anchor System offers an alternative, but it remains unclear to what extent it affects the perception and control of cable operation forces compared to the traditional shoulder harness.

Objective: To compare cable force perception and control with the figure-of-nine harness versus the Ipsilateral Scapular Cutaneous Anchor System and to investigate force perception and control at different force levels.

Method: Ten male able-bodied subjects completed a cable force reproduction task at four force levels in the range of 10 – 40 N, using the figure-of-nine harness and the Anchor System. Perception and control of cable operation forces were quantified by the force reproduction error, and the force variability.

Results: In terms of force reproduction error and force variability, the subjects did not behave differently when using the two systems. The smallest force reproduction error and force variability were found at the smallest target force level of 10 N.

Conclusions: The Anchor System performs no differently than the traditional figure-of-nine harness in terms of force perception and control, making it a viable alternative. Furthermore, users perceive and control low operation forces better than high forces.

5.1 BACKGROUND

When users reject body-powered prostheses, they frequently describe the poor comfort and cosmetic properties (15,35) associated with the traditional figure-of-eight and figure-of-nine harnesses. The traditional harness design is essentially the same as the design made by the Count of Beaufort in 1860 (4) although improved harness comfort and appearance under clothing has been the main design priority of users since then (35). Attempts to achieve improved harness comfort or appearance include the introduction of the axillar bypass ring (67), the T-shirt system (68), and the Ipsilateral Scapular Cutaneous Anchor System (53). The Anchor System, patented in 2006 (69), is the only one of these alternative systems that is available commercially. It consists of a flat plastic patch that is adhered directly onto the skin at the scapula and contains a button that connects the body to the prosthesis' Bowden cable. The Anchor System returns the unimpeded use of the unaffected side, and reduces the strain on the armpit by eliminating the need for straps altogether; thereby resulting in both increased cosmetic value and comfort (53).

However, body-powered prostheses – as a natural extension of the body – should provide the user with proprioceptive feedback and control of operation forces. An alternative harness design might alter the extended physiological proprioception (46), one of the main advantage of body-powered prostheses compared to current myo-electric prostheses (51,70,71). On the one hand, the Anchor System is adhered directly onto the skin, which may result in a more direct force transmission and tactile feedback of high resolution. On the other hand, perception and control might be reduced because the Anchor System eliminates shoulder movements of the contralateral side and the resulting proprioceptive information of these movements. The user effectively has one less degree of freedom to operate the prosthesis. For this reason it is expected that the traditional figure of nine harness would offer superior perception and control of operation forces compared to the Anchor System. However, this has never been investigated.

In motor control literature force reproduction tasks are used to quantify force perception and control (55-58). Recently these have also been implemented to quantify perception and control of low cable operation forces in voluntary closing body-powered upper-limb prostheses (72), by investigating force reproduction error and force variability. The force reproduction error, which is the difference between the reproduced and target force, indicates the difference between the intended and exerted grasping force in clinical practice. The force variability implies the deviation in grasping force. A small force reproduction error and a small force variability are desired because this indicates that the user is in control of the forces he exerts on an object.

This study aims to compare force perception and control with the figure-of-nine harness versus the Ipsilateral Scapular Cutaneous Anchor System for a range of relevant (daily use) force levels. A second objective of this study is to investigate differences in force perception and control these force levels. This is done by comparing two metrics of a force reproduction task,

force reproduction error and force variability, which are attained for both systems at four target force levels. We hypothesize that the force reproduction error, as well as the force variability of the Anchor System, is higher compared to the figure-of-nine harness. Furthermore, based on the results of a another study (72), we hypothesize no differences in force reproduction error and an increasing force variability with increasing target force levels.

5.2 METHODS

5.2.1 Subjects

Ten right-handed male able-bodied subjects (age: 28±2 years old (mean ± standard deviation)) participated in our research. This study was approved by the Ethics board of Delft University of Technology (ID number 1481).

5.2.2 Materials

A custom-made prosthesis simulator (Figure 5–1) consisting of a thermoplastic shell with 3 mm Neoprene on the inside connected via a standard 1/16" (.159 cm) diameter stainless steel cable (C100, Hosmer Dorrance Corporation, Chattanooga, TN, USA) running inside a stainless steel cable housing for C-100HD cable (CH-100HD, Hosmer Dorrance Corporation, Chattanooga, TN, USA) to either an adjustable "figure-of-nine" harness or the Anchor System (69) (Cutaneous Anchor Technology, Single-Handed Solutions, LLC, Springfield, MA, USA distributed by TRS Prosthetics, Boulder, CO, USA). The "figure-of-nine" harness and Anchor System were interchangeable. The cable excursion was disabled in this setup, and no prehensor was used in order to eliminate any influence from its mechanical properties. Operation cable forces were proportional to pinch forces exerted on objects in voluntary closing prostheses and cable excursions remained constant when building up pinch forces on rigid objects (32). The thermoplastic shell was attached to the participant's lower left arm. The steel cable was interrupted by one force sensor (FLLSB200 222 N, FUTEK, Irvine, CA, USA). The measured forces were amplified (CPJ, Scaime, Juvigny, France) and sampled together with the displacement sensor at 50 Hz (NI USB-6008, National Instruments, Austin, TX, USA), and finally stored using a custom LabVIEW program (LabVIEW 2012, National Instruments, Austin, TX, USA).

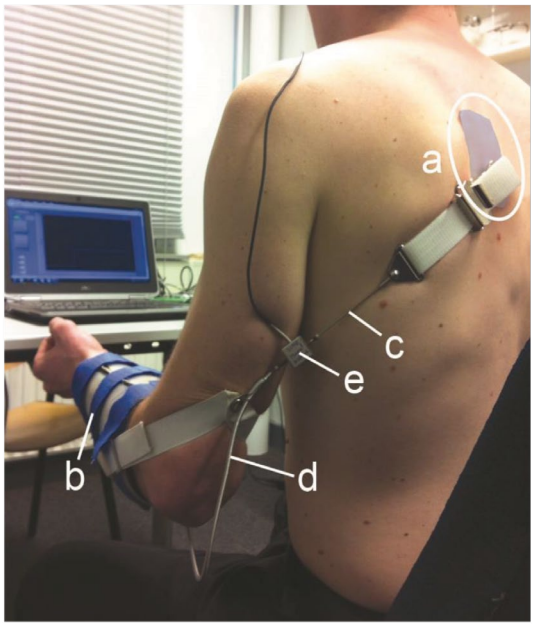


Figure 5–1. The Anchor System (**a**) is connected to a thermoplastic shell (**b**) via a Bowden cable (**c**) running inside a cable housing (**d**). There is a force sensor (**e**) in the middle of the Bowden cable to measure the cable operation forces. The original color (transparent) of the Anchor System was edited to clearly show its dimensions in the figure.

5.2.3 Procedure

The experimental procedure is illustrated in Figure 5–2. After the first system, the Anchor System or the harness was fitted to the subject. Next the control movements to operate a body-powered prosthesis and the experimental task were explained and the subject practiced these during a training session. The subjects were requested to produce a target force as shown on a screen in front of them, using humeral anteflexion and abduction of the affected side, together with shoulder protraction of the contralateral side in the case when the harness was used. During a visual block, the target and measured force were shown on the laptop screen, whereas during a blind block only the target force was displayed for the duration of the force reproduction. In other words, during visual blocks, subjects reproduced the target force based on the visual information on the screen, whereas, during the blind blocks, they based the magnitude of the reproduced force on the perceived force during a visual block. Participants were instructed to produce the force as stable as possible. The training was completed once the subject was familiar with the prosthesis operation and the experimental task. After the training session at 15 N, subjects conducted the actual force reproduction experiments at four force levels (10, 20, 30 and 40 N). One trial of the actual experiments consisted of ten visual and ten blind alternating blocks. One block lasted seven seconds followed by a three second break, resulting in a duration of 200 seconds per trial. Then the second system was fitted and a second training session started at 15 N in order to allow the subjects to familiarize themselves with the other system. This was followed by the actual force reproduction experiments at the four force levels. The order of

tested system as well as the force levels, were counterbalanced over the subjects. The Anchor System was fitted in accordance with the TRS instruction video (73).

The force levels examined were limited to prevent discomfort or, worst-case, skin damage, but were still representative of daily activities. At 10 N the TRS hook started pinching and with 40 N cable force the hook pinched at approximately 20 N (32). A pinch force of 20 N is reported to be sufficient to complete most daily activities with an upper-limb prosthesis (63,64).

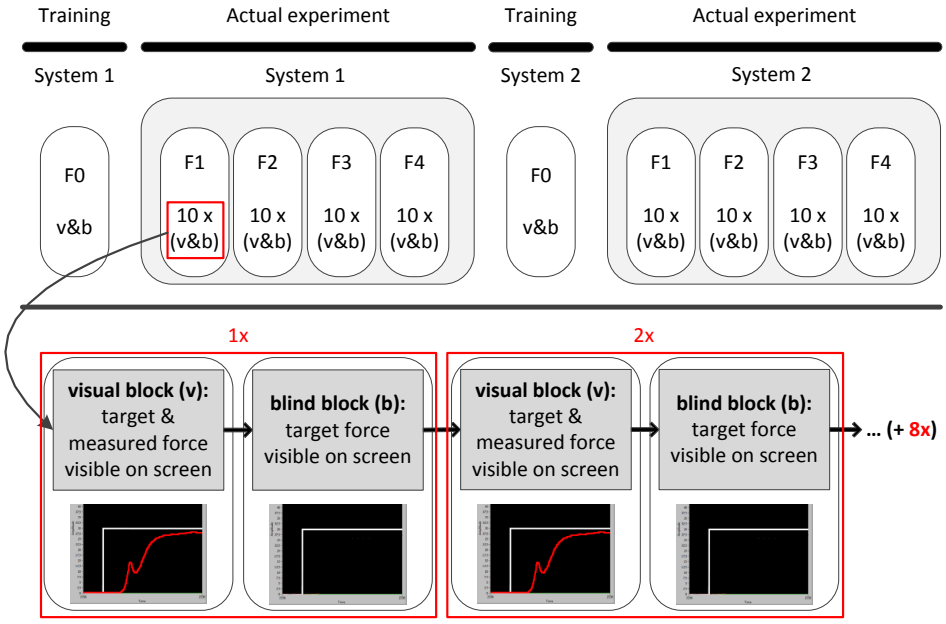


Figure 5–2. Flowchart illustrating the experimental procedure. Subjects performed the experiments with the two systems, Anchor System and figure-of-nine harness, with the order counterbalanced over subjects. After practicing at 15 N (F0), each system was examined at four force levels (10, 20, 30 and 40 N) during 10 visual and 10 blind blocks in alternating order. The force levels (F1 to F4) were counterbalanced over the subjects. The order of force levels differed per subject.

5.2.4 Data analysis

5.2.4.1 Metrics

Participants' performance was assessed by the force reproduction error, which is the difference between the target and reproduced force, and by the force variability, which is the noise of the reproduced force. The last four seconds of measured force were analyzed by calculating the mean and standard deviation (Figure 5–3). Because the perceived force during the visual block must be reproduced during the blind block, the force reproduction error (FRE) per block was

calculated as the average force of a blind block minus the average force of the foregoing visual block (Equation 5–1). The results per block were then averaged over all blocks of the trial to obtain the overall force reproduction error (per subject, per force level) (Equation 5–2).

$$FRE_{block_i} = mean(F_{blind,block_i}) - mean(F_{visual,block_i}) \quad (5–1)$$

$$FRE = \frac{1}{10}\sum_{i=1}^{10} FRE_{block_i} \quad (5–2)$$

The force variability (FV) results from the standard deviation of the blind blocks (Equation 5–3) averaged over all analyzed blocks (Equation 5–4).

$$FV_{block_i} = std(F_{blind,block_i}) \quad (5–3)$$

$$FV = \frac{1}{10}\sum_{i=1}^{10} FV_{block_i} \quad (5–4)$$

The force reproduction error and force variability were determined for each of the four target force levels for both the Anchor System and harness per subject. The mean force reproduction error and mean force variability represent the average values of the group.

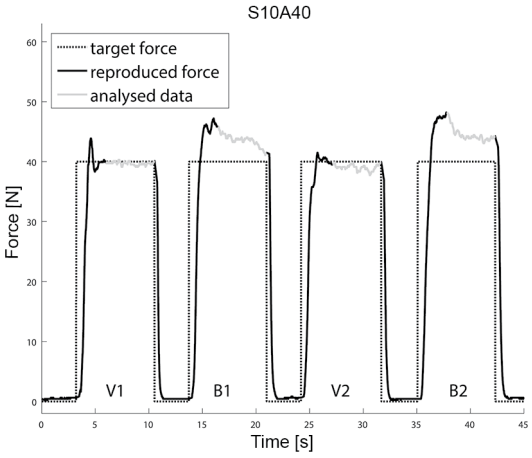


Figure 5–3. The raw data of the first 45 seconds of a typical trial (subject 10 – Anchor System @ 40 N) shows the target force (dotted line) of 40 N and the measured cable force (black solid line) at the back of the subject. One full trial consisted of ten visual blocks alternating with ten blind blocks (only first four blocks shown). Each block lasts 7 seconds, followed by a 3 second break. The last 4 seconds of each block are used for data analysis.

5.2.4.2 Statistics

For statistical analysis SPSS version 20 was used. A two-way repeated measures ANOVA with two levels for harness system and four levels for target force levels was conducted to determine the experimental effects for the two outcome measures, force reproduction error and force variability. A significance level of α=0.05 was maintained.

5.3 RESULTS

No significant differences were found between the Anchor system and harness, for both force reproduction error (F(1,9)=3.134, p =0.11) and force variability (F(1,9)=1.002, p =0.343), as shown in Figure 5–4 and Figure 5–5.

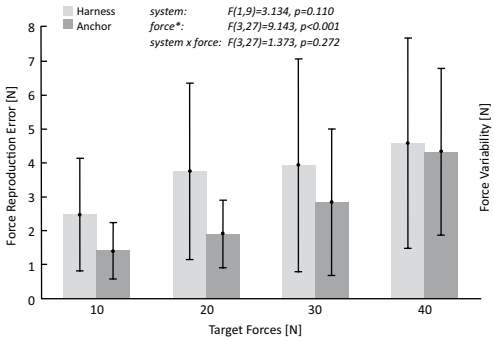


Figure 5–4. The force reproduction error (y-axis) is defined as the difference between target and reproduced force. It is presented for the harness and the Anchor System (see legend) at the four examined target force levels, 10 N, 20 N, 30 N and 40 N (x-axis). The bars indicate the group averages of the force reproduction error, whereas the whiskers show the standard deviations over the subject group (one standard deviation above and one below the group average). Between the two systems no significant differences were found, whereas the differences between force levels were significant (*). The interaction (system x force) did not have a significant effect.

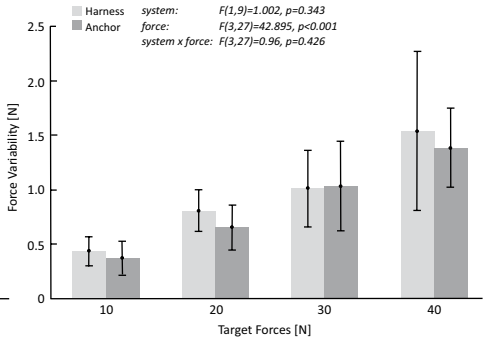


Figure 5–5. The force variability (y-axis) is defined as the deviation of the reproduced force (y-axis). It is presented for the harness and the Anchor System (see legend) at the four examined target force levels, 10 N, 20 N, 30 N and 40 N (x-axis). The bars indicate the group averages of the force variability, whereas the whiskers show the standard deviations over the subject group (one standard deviation above and one below the group average). Between the two systems no significant differences were found, whereas the differences between force levels were significant (*). The interaction (system x force) did not have a significant effect.

However, significant differences between force levels were found for force reproduction error and force variability (force reproduction error: F(3,27)= 9.143, p <0.001; force variability: F(3,27)= 42.895, p <0.001). Both metrics increased as the force level increased. Target forces were overestimated for both systems at all target force levels, which is indicated by the positive mean force reproduction error.

The interaction (system x force) did not have a significant effect (force reproduction error: F(3,27)=1.373, p =0.272, force variability: F(3,27)=0.96, p =0.426).

5.4 DISCUSSION

In contrast to what was hypothesized, we did not find a difference in force reproduction error or force variability between the Anchor System and the figure-of-nine harness. Both the mean force reproduction error and mean force variability increased significantly as the target force levels increased. This is in accordance with our hypothesis regarding force variability, but not force reproduction error.

5.4.1 Systems

Since there were no differences in force reproduction error or force variability between the two systems, subjects had no preference for either system in terms of perception and control of operation forces. This suggests that the disadvantage of less proprioceptive information in the Anchor System might be counterbalanced by the advantage of more direct force transmission and superior tactile feedback. Alternatively, the effects of each aspect, less proprioceptive information and more direct force transmission and superior tactile feedback, may be negligible.

Although not statistically significant, the mean and standard deviation of the force reproduction error across the group of subjects appear lower for the Anchor System than for the harness at all force levels (Figure 5–4). The larger mean force reproduction error and variability across the group with the harness might result from two outlier subjects, whose force reproduction error was much larger than the other subjects. This might indicate individual preferences for one system over the other, but since the observed differences are not statistically significant, this does not justify a generalization of this preference for all users in terms of the accuracy to meet an (estimated) target force.

5.4.2 Force levels

In contrast to what was hypothesized, the *force reproduction error* showed a significant difference between force levels. Post-hoc analysis showed that the force reproduction error was significant different between 10 N and 40 N, as well as 20 N and 40 N.

The increasing force reproduction error with increasing target forces, implies for prosthesis operation that users can exert the intended grasping force more accurately at low force levels.

The difference in hypothesized and determined results of the force reproduction error might be explained by different subject populations of the current study compared to the study on which we based our hypothesis. The current study used a relatively homogeneous group of ten right handed male controls. The previous study used a heterogeneous group of 24 subjects with unilateral trans-radial deficiencies (left or right side affected) of both genders (72). In addition, there are some differences between the two measurement protocols, but these are not expected to have a significant influence.

As hypothesized, the force variability was significant different between force levels. Here, the post-hoc analysis showed significant differences for all combinations of force levels. The increasing force variability with increasing target forces, implies that users can stabilize a pinch force exerted on an object better at low operation forces.

5.4.3 Anchor system

The overall force perception and control of both systems are comparable, making the Anchor System a possible alternative to the traditional harness. Still, some practical questions remain. One subject remarked that attaching the Anchor System might prove difficult on your own. The inventor, Debra Latour, explained in an email conversation (20 October 2016), that an assistive mounting device can be used to place the Anchor System on one's back if it is not within the individual's normal range of motion.

Additionally, while the direct skin contact was overall thought to be beneficial for transmitting force information at low force levels, one subject expressed the concern that the Anchor System might feel really uncomfortable at higher force levels. Regardless of whether the Anchor or harness system is used, we believe excessively high operation forces should be avoided not only to decrease force reproduction error and force variability, but also to minimize fatigue and discomfort caused by repetitively exerted high operation forces (74). Furthermore, the Anchor System is not feasible for users who are allergic to adhesive substances. To minimize this concern, medical-grade hypo-allergenic tape is used to connect the Anchor to the skin. Alternatively, other latex-free products could be used instead of the current adhesive, according to Latour.

5.4.4 Study limitations

Due to the limited availability of prosthesis users, the subject population of this study consisted only of able-bodied individuals. However, the magnitude of force reproduction error and force variability is consistent with the values found for subjects with trans-radial deficiency (72).

The examined force levels of this study were limited to 40 N. Hence, the perception and control differences between the two systems at higher operation forces cannot be concluded based on this study.

The force reproduction experiments aim to simulate short and intensive prosthesis use, but remain different from daily prosthesis operation. The attained freedom of the contralateral side with the Anchor System and reduced discomfort of the armpit through elimination of the straps altogether might be beneficial during daily activities. The resulting advantages or disadvantages in terms of cosmesis, comfort or control have not been quantified here and would require further attention.

5.5 CONCLUSION

This study aimed to compare force perception and control with the figure-of-nine harness versus the Ipsilateral Scapular Cutaneous Anchor System. A force reproduction task was used to investigate force perception and control for various relevant (daily use) force levels. The metrics of force reproduction error and force variability revealed no differences between the two systems. Furthermore, force perception and control abilities improved with decreasing force levels. Our advice is to consider the Anchor System for body-powered prosthesis operation, particularly at low operation forces, since its performance is comparable to the harness and it offers increased cosmetic value and comfort.

6

DISCUSSION

The studies performed in this thesis aimed to quantify user capacities to operate a body-powered prosthesis and establish a better understanding of the prosthesis-input requirements in order to frame quantified user-centered body-powered prosthesis design requirements.

This final chapter will combine the results from the different studies and discusses the various steps towards a well-designed body-powered prosthesis, starting with the implications for the prosthesis input requirements (the human-prosthesis interface), followed by the prosthesis-output requirements (the prosthesis-object interface) and the prosthesis components (shoulder harness ('SH'), transmission ('TM') and prehensor('PH')). Thus the results will be discussed based on the scheme of the human-prosthesis-object interaction of a body-powered prosthesis discussed in the introduction (Figure 1–3) and depicted here again in Figure 6–1 for the readers ease.

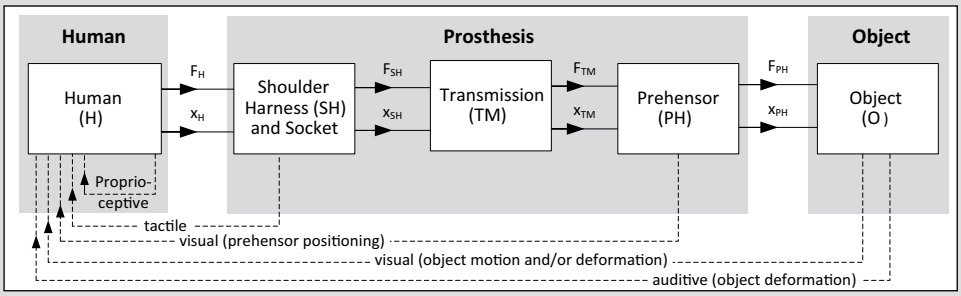


Figure 6–1. Scheme of the human-prosthesis-object interaction of a body-powered prosthesis.

6.1 PROSTHESIS-INPUT REQUIREMENTS OR THE USERS' CAPACITIES

Investigating the prosthesis-input requirements started with the quantification of users' strength in terms of maximum cable operation forces (Chapter 2). The force a user can create on the control cable was measured 'F_{SH}' in Figure 6–1. This force is transferred via the Bowden cable to the prehensor and thus serves as input to the prehensor 'F_{TM}' in Figure 6–1. The attained maximum forces ranged from 87 to 360 N for females and from 199 to 538 N for males, which implies that not only differences between genders were found, but also shows a wide variety of prosthesis users' strength. It is important for prosthesis designers and clinicians alike to take into account that users differ a lot in terms of strength. The comparison of the measured maximum cable forces with the required operation forces of available prehensors indicated that three out of ten body-powered prostheses cannot be operated by all users. Tiring and painful operation is one of the main body-powered prosthesis user complaints (35). Therefore forces up to a 20% limit of the maximum forces were considered as a fatigue-free operation range. This limit was chosen based on literature on isometric contractions (38). The results imply that males could operate a prosthesis fatigue-free up to 66±23 N (mean ± standard deviation), and females up to 38±17 N. Furthermore, nine out of ten body-powered prostheses cannot be operated by more than 50% of the users when comparing the required operation forces with users' fatigue-corrected forces. The best performing prosthesis was the TRS hook. Still 25% of the users were not able to operate the TRS hook fatigue-free, which suggests that the current state-of-art of body-powered prostheses is not suitable for prosthesis users.

In Chapter 2 the cable forces required to pinch 15 N were evaluated. Pulling on a sock requires a pinch force of 34 N (75), which results in a required cable force of ±60 N for the TRS hook instead of the 33 N cable force required for a 15 N pinch as illustrated in Figure 10 of Smit and Plettenburg's study (32). Incidentally applying higher forces than the fatigue limit would not lead to fatigue, but will probably lead to reduced control of the applied pinch forces 'F_{PH}' in Figure 6–1 (Chapter 3). In the study of Chapter 3 the input force of the prehensor 'F_{TM}' in Figure 6–1, was manipulated to see its effect on the prehensor's pinch force control on objects 'F_{PH}' in Figure 6–1. The two chosen cable operation force levels of ±16 N and ±52 N 'F_{SH}' in Figure 6–1 were below the average fatigue limit for males of 66±23 N and only male controls were measured (Chapter 2). Additionally, male controls are considered to be stronger than prosthesis users (Chapter 2) and thus have a higher fatigue limit. This suggests that fatigue effects are negligible in this study and did not influence the users' task performance. In summary, high cable operation forces lead to inferior prehensor control.

The dependency of prehensor control on the operation force level was investigated for two force levels (±16 N and ±52 N; 'F_{SH}' in Figure 6–1) in Chapter 3. In Chapter 4, experiments were done without prehensor to investigate perception and control differences of cable forces at levels ranging from 10 to 40 N ('F_{TM}' in Figure 6–1), by a force reproduction task assigned to prosthetic

users. Cable forces exerted on the prehensor ('F_{TM}' in Figure 6–1) are directly related to the pinch forces produced by the voluntary closing prehensor 'F_{PH}' in Figure 6–1 (32). To allow a general advice for all prostheses, the results should be independent of the mechanical properties of one prehensor. Therefore in this study the end of the control cable was fixated either on a threaded rod or springs of different stiffness instead of the prehensor. The threaded rod setting disabled cable excursions and therefore simulated holding a rigid object. The different spring settings resulted in predefined cable force-excursion relationships simulating different prehensor properties. So the control accuracy of the prehensor input ('F_{TM}' and 'x_{TM}' in Figure 6–1) was investigated aiming to indicate a force level ('F_{TM}') which can be optimally perceived and controlled. Two metrics were investigated, the force reproduction error and the force variability. The force reproduction error indicates the difference between intended and applied force, whereas the force variability identifies the users' ability to hold the applied force constant. When the applied force on an object is too small the object might slip out of the prehensor and fall. Is the force too high the object might break. Both result in unsuccessful object manipulation, which discourages the user to manipulate objects with his prosthesis.

Based on a preliminary study (50) we hypothesized a minimum force reproduction error between 10 and 20 N operation force. However, the results showed no differences of the force reproduction error at different force levels. The force variability was increasing, as hypothesized. Thus based on the results of Chapter 4 we cannot recommend an optimum operation force range, but point out that the user is able to hold the operation forces ('F_{TM}' in Figure 6–1) and thus the resulting pinch forces ('F_{PH}' in Figure 6–1) more stable at lower force levels, which is in line with the results of Chapter 3. Interestingly, the results of Chapter 5 showed for increasing force levels significantly increasing force reproduction error (and force variability). The variability of the force reproduction error amongst users (Chapter 4) is higher than amongst controls for both systems, harness and anchor system (Chapter 5) as indicated in Figure 6–2. This high variability amongst users might explain why no differences between force levels were found in Chapter 4. The difference in force reproduction error between the two groups, users versus controls, might be explained by different subject populations or the experimental set up. The user population consisted of 24 male and female subjects of 49±13 years of age (Chapter 4), whereas the control group (n=10) was more homogenous in age (28±2 years old) and gender (males) (Chapter 5). In the experimental setup described in Chapter 4 the forces measured at the arm ('F_{TM}' in Figure 6–1) were shown on the monitor alongside the target force, whereas in the study of Chapter 5 the forces measured at the subjects' back ('F_{SH}' in Figure 6–1) were presented on the monitor (alongside the target force). Additionally, differences between force reproduction time and the duration of breaks between the force reproduction and the number of examined target forces, six versus four, might yield different results. In both studies (Chapter 4 and Chapter 5) the target forces were constantly overestimated, which is a known phenomenon when a low externally generated target force has to be matched (57). Chapter 4 and Chapter 5

indicate that the Bowden cable introduces friction losses, which results in efficiencies of 80 to 84% (Chapter 3) or force differences measured before and after the Bowden cable of 2 to 3 N ('F_{SH}' and 'F_{TM}' in Figure 6–1), incidentally up to 9 N (Chapter 4). In summary, the general advice for body-powered prosthesis design is to keep the operation forces as low as possible to increase the control accuracy of operation forces and resulting pinch forces. However, the perception and control accuracy of cable operation forces lower than 10 N might be significantly influenced by the high friction losses of the Bowden cable and are therefore not advisable.

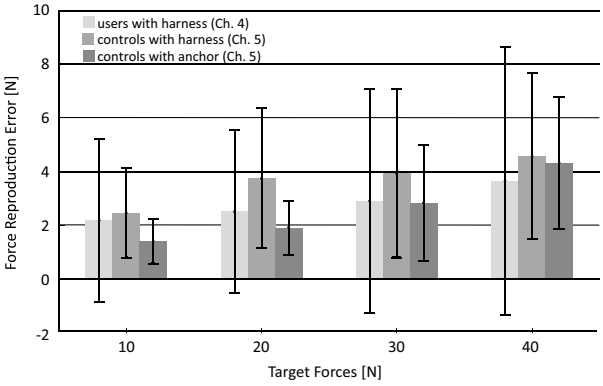


Figure 6–2. Force reproduction error of prosthesis users (n=24) utilizing the traditional harness (Chapter 4), compared to that of controls (n=10) utilizing either the harness or the anchor system (Chapter 5). The variability of the force reproduction error over the user group is larger than over the control group, which may contribute to the (statistically significant) differences *between force levels* that were present with controls, but not with users.

Furthermore, results in Chapter 4 indicated that cable excursions ('x_{TM}' in Figure 6–1) contribute to a more accurate force reproduction, whereas the ability to hold the force stable was independent of the cable excursion. Thus, the combination of low operation forces with high cable excursions contributes to an increased perception and control of cable forces. For prosthesis design, however, long strokes required to achieve large cable excursions might prove to be impractical when operating the prehensor close to the body. As a comparison, available body-powered prehensors require cable excursions up to 53 mm [32,33] ('x_{TM}' in Figure 6–1). In Chapter 4 cable excursions of up to 80 mm were examined. However, the practicality of larger cable excursions needs to be investigated in daily use.

The discomfort during or after the experiment was monitored in Chapters 2 to 4. Subjective discomfort perceptions were indicated on a body-map and showed that creating maximum cable forces as well as a long experimental duration provoked discomfort (Chapter 2 and Chapter 4). Furthermore, the results of a semi-structured interview suggest a lower long term discomfort (lower load on the axillar region, less tiring, less required effort) with lower cable operation forces (Chapter 3).

The acquired knowledge on the prosthesis-input requirements and its resultant design requirements for body-powered prostheses to reduce the operation forces, address the user complaints of exhaustion, (upper body) pain, sores, and skin irritation (35). The user's wish of eliminating the harness and providing a greater choice in harnessing configurations (35), led to Chapter 5 in which an alternative harnessing system, the Ipsilateral Scapular Cutaneous Anchor System, was proven to be a good alternative for the traditional harness in terms of force control and is deemed to be more comfortable than the harness. The system aims to improve the comfort and cosmesis of the traditional harness and at the same time provides the user with comparable perception and control at low operation forces.

In summary, based on this thesis the following design guidelines for body-powered prostheses are given:

1. Keep the required operation forces as low as possible, but no lower than 10 N.
2. A maximum operation force of 38 N should be maintained to enable the average female (66 N for the average male) to operate a prosthesis fatigue-free during the day every day. Higher forces are possible for non-repetitive tasks that are unlikely to cause fatigue. Note when applying this upper boundary that this is a group average, meaning that there will be users, who are not able to operate a prosthesis at this low force boundary fatigue-free.
3. Prosthesis designers and clinicians need to realize that users differ a lot in terms of strength.
4. Larger cable excursions ('x_{TM}' in Figure 6–1) facilitate better perception and control of operation forces, but keep in mind that long strokes required to achieve these cable excursions might not be practical when operating the prehensor close to the body. The practical feasibility of employing cable excursions up to 80 mm was not evaluated in this thesis.
5. The Ipsilateral Scapular Cutaneous Anchor System is a viable alternative to the traditional shoulder harness. It offers the same operating performance, and is rated as more comfortable.

6.1.1 Future research on the prosthesis-input requirements

This thesis did not specifically address the minimum operation forces for users or the relationship between user strength and preferred operation forces ('F_{SH}' in Figure 6–1). It is worthwhile to verify the presented results in clinical practice by implementing a low operation force prehensor and evaluate its effect on fatigue, pain, and perception as well as its documented use. Unfortunately, the optimal prehensor remains unavailable, since even the best performing body-powered prosthesis, the TRS hook, requires too high operation forces to enable all users to use the device. Furthermore, since large cable excursions ('x_{TM}' in Figure 6–1) help prosthesis control it is recommended to quantify the range of movements and strokes that body-powered prosthesis users can achieve in daily activities.

6.2 PROSTHESIS-OUTPUT REQUIREMENTS OR THE USERS' DEMANDS

When operating a prosthesis the user demands adequate pinch force of his prosthesis ('F_{PH}' in Figure 6–1), which can be exerted on an object, and high quality feedback of the prehensor-object interaction. This thesis shows that the feedback of finger positioning ('x_{PH}') and pinch force ('F_{PH}' in Figure 6–1) improves with lower operation forces ('F_{SH}' in Figure 6–1). However, required pinch forces for daily activities are expected to vary widely depending on the activities and nature of the manipulated object. Therefore, quantification of the range of pinch forces required for all daily activities is complicated. The Split-Hook grasp taxonomy shown in Figure 6–3 illustrates the variety of prehensor-object interactions possible with a hook prosthesis. The taxonomy distinguishes between non-prehensile and prehensile object manipulations. Prehensile manipulations are subdivided in not within fingers and within finger manipulations. The latter distinguishes between the number of contact points of the prehensor with the object.

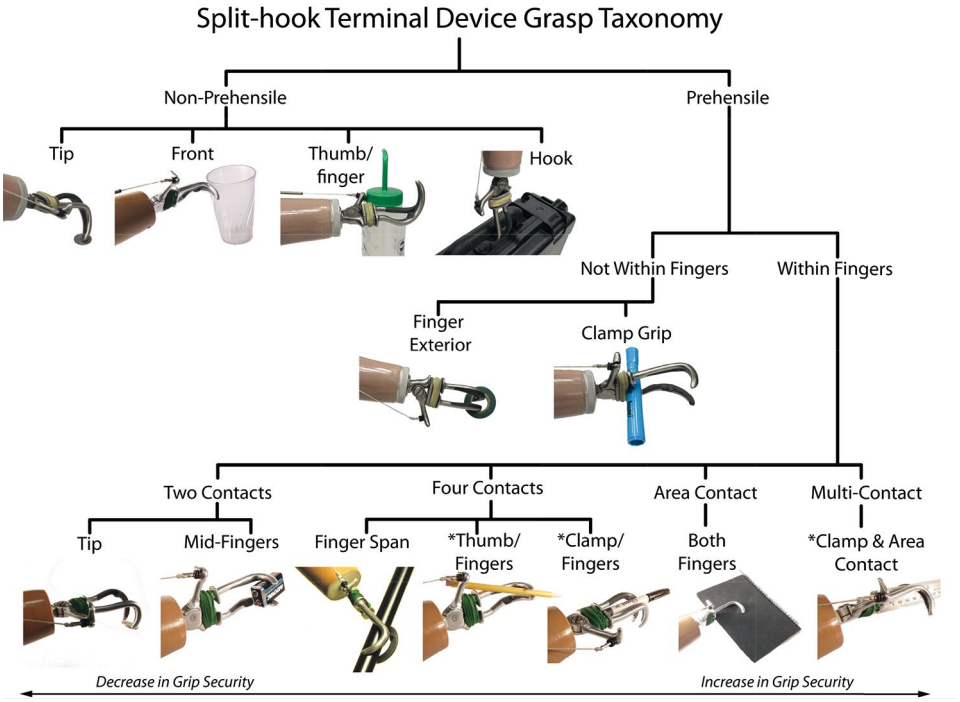


Figure 6–3. "The Split-Hook grasp taxonomy shows all observed uses for the voluntary-opening split-hook as controlled by a body-powered control cable and harness. *Grasps only observed by placing the object within the hook using the able hand." (77)

For a prehensile within fingers object manipulation the required pinch forces are dependent on the prehensor's and object's properties, the type of grasp and the resulting prehensor-object contact area. For instance, holding an object with a hook prosthesis results in two relatively small contact areas. The free-body diagram in Figure 6–4 shows that the mass of the object needs to be counterbalanced by the pinch force orientated in a single axis. The required pinch force is dependent on the mass of the object and the coefficient of friction of the contact area. When holding the same object with a prosthesis with adaptive fingers (Figure 6–5), the fingers are folding around the object, thus the little finger can slide under the object and balance the object's mass. The coefficient of friction of the contact area is dependent on the object's material and the prosthetic hand's glove material, PVC or silicone, or the metal of which the hook is made. In some hooks the grasping surface is coated with rubber or similar materials.

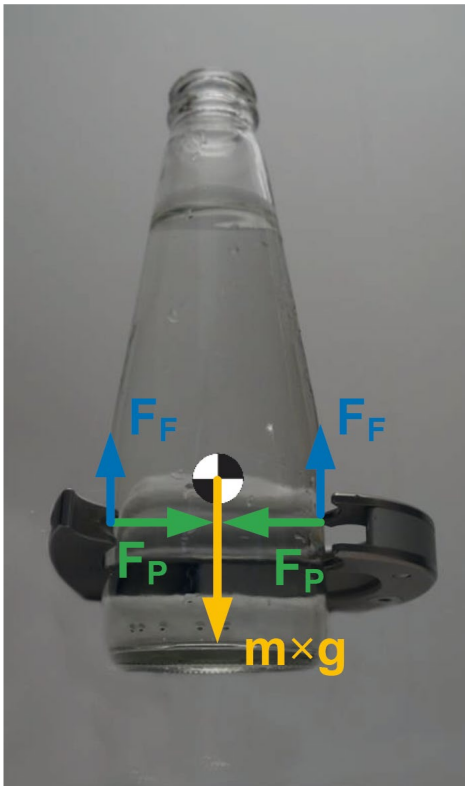


Figure 6–4. Free-body diagram of an object grasped with a hook prosthesis.

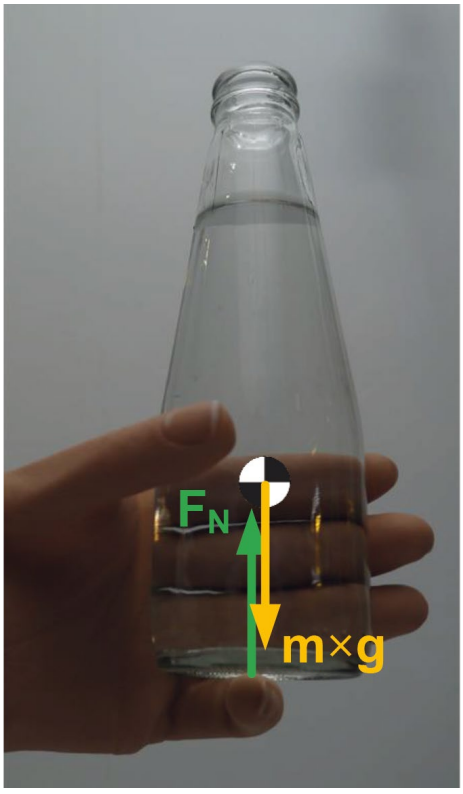


Figure 6–5. Free-body diagram of an object grasped with an adaptive hand prosthesis.

When searching for prosthesis-output design requirements, one might choose to take the literature on the required pinch forces for daily activities with the natural hand into account. However, the resulting coefficient of friction from natural hand-object contact is not comparable to the materials used in prostheses and not easy to determine. Additionally, the ability of the natural hand to fold around an object cannot be matched by the most advanced adaptive hand prosthesis.

Quantifying the number of prehensor activations and the duration a user is holding onto an object is not only dependent on a user's activity level, preferences and attitude, but also on how well the prosthesis accompanies his capacities and demands. For instance, if a prosthesis requires too high operation forces ('F_{SH}' in Figure 6–1) to hold onto an object for a long period of time and the user is getting tired, he will search for an alternative to complete his task. However, limited information is available on the number of prehensor activations, the pinch force magnitude and duration of maintaining the pinch force. A portable or mobile app-based data acquisition system, which can be connected to the prosthesis and monitor daily activities outside the clinic, is probably a very helpful tool to attain this valuable piece of information.

Alternatively, the range of pinch force magnitudes ('F_{PH}' in Figure 6–1) required for daily activities can be quantified experimentally. Therefore, a set of (bimanual) tasks should be chosen to elaborate the whole range of pinch forces, from low to high. Since the relationship between cable activation forces and pinch forces ('F_{TM}' and 'F_{PH}' in Figure 6–1) of the TRS hook is well known, a simple set-up adapted from the experiments in Chapter 3 and 4 can be utilized. The appurtenant benefit of this set-up is that no force sensors, force sensing resistors (FSR) or strain gauges (resistance changes with elastic elongation of the strain gauge) need to be placed and the contact surface between object and prehensor remains unaffected. Measuring the pinch forces required with the TRS hook will probably result in higher pinch forces than would be measured with an (adaptive) hand prostheses, due to the poor prehensor-object contact surface and force distribution on a single axis as described earlier. Although the relevance of the results will strongly depend on the careful choice of examined tasks, such an experimental approximation might serve as a good starting point for prosthesis design requirements. Next to body-powered prostheses, also myo-electric prosthesis design might benefit from such a study.

6.2.1 Future research on the prosthesis-output requirements

In summary, required pinch forces to manipulate objects need to be quantified ('F_{PH}' in Figure 6–1). Furthermore, identification of the number of prehensor activations and the object manipulation durations would contribute to a better understanding of prosthesis requirements to effectively execute daily activities. The information on the prosthesis-input and -output requirements can be combined in an optimal ratio between cable operation forces and pinch forces aiming for satisfactory grip strength at the best possible sensory feedback.

6.3 PROSTHESIS COMPONENTS

This section discusses the state of the art and future developments of the different prosthesis components, shoulder harness ('SH'), transmission ('TM') and prehensor ('PH') as indicated in Figure 6–1 and links to the design guidelines as summarized at the end of section 6.1.

6.3.1 Shoulder harness

An extensive overview of different harness patterns for upper-extremity prostheses was described by Pursley in 1955 (77), which are used up until now ('SH' in Figure 6–1). Users desire elimination of the harness and a greater choice in harnessing configurations (35). The anchor system evaluated in Chapter 5 was shown to be an alternative. The axillar bypass ring (67) and the T-shirt system (68) are proposed as other alternatives. Several attempts have been made by the DIPO to implement the harness in a bra or as a 'Chest Strap' (78). However, none of them are commercially available and further improvement of these alternatives is needed. The proposed low operation forces a body-powered prosthesis should be operated with might make some of these attempts practically more feasible and may inspire designers to come up with more alternatives. Improvements on the user design preferences and points of concern of the current design should be centralized in the designing process, such as skin irritation, pain and exhaustion during or after operation, restrictions in movements, appearance as well as the ease of donning and doffing the prostheses (15,35). Some individuals might prefer the Anchor System over the traditional harness for cosmetic reasons, experience wearing the Anchor system as more comfortable, or may have better perception and control of operation forces with the Anchor System compared to the harness. Individual preferences play an important role when choosing the harness system. Anchor System and harness are also easily interchangeable and could be chosen dependent on the planned activities of the user. However, the awareness that the Anchor System is a possible alternative for the harness needs to increase in rehabilitation centers and more alternatives to the harness are desired.

6.3.2 Transmission

The Bowden cable is transmitting forces and displacements exerted by the prosthesis user to the prehensor ('TM' in Figure 6–1). Unfortunately, the Bowden cable introduces friction losses. Static friction losses dependent on the Bowden cable's materials and its curvature have been investigated (36,37), also in comparison with a hydraulic transmission system (79). However, the Bowden cable force transmission efficiencies measured during the experiments described in Chapter 3 were between 80 and 84%, which corresponds to a wrapping angle of 270 degrees according to Carlson (37). Reported deviations of the measured forces at the users arm and back, thus before and after the Bowden cable ('F_{SH}' and 'F_{TM}' in Figure 6–1), were typically 2 to 3 N and incidentally went up to 9 N in the study described in Chapter 4. Since the wrapping angle

never exceeded 90 degrees, this suggests that the friction losses of a Bowden Cable during dynamic operation play a prominent role in body-powered prosthesis operation. The Bowden cable introduces unknown behaviors to the operation system and might act as a filter with unknown characteristics. Its influence of the control accuracy of pinch forces and to what extent the human controller can anticipate this apparently unknown behavior of the system needs to be investigated. During the ISPO Europe conference 2016 in Rotterdam, the results of an experiment on the dynamic efficiency of the Bowden cables were presented (52). This preliminary study showed that the Bowden cable with Teflon liner, which is used in body-powered prosthesis and the experiments of Chapter 3 and 4, was less efficient than the Igus Robolink™, a high performance bicycle Bowden cable. Thus, the Igus Robolink™ could be considered for body-powered prosthesis application to archive higher system efficiencies. Overall, more efficient force transmission with known static and dynamic characteristics is desired to contribute to a low operation force prosthesis design that offers better perception and control to the user.

6.3.3 Prehensor

Available body-powered prehensors ('PH' in Figure 6–1) showed relatively advanced mechanical properties of prosthetic hooks compared to prosthetic hands (32,33). The concluding advice for body-powered prostheses of both studies targets a lower mass, higher efficiency, higher pinch force, and lower actuation force without the provision of specific values. The results of this thesis suggest that body-powered prosthesis operation forces ('F_{SH}' in Figure 6–1) should not increase beyond 38 N for the average female and 66 N for the average male user to facilitate fatigue-free prosthesis operation with adequate perception and control of cable and pinch forces ('F_{PH}' in Figure 6–1). Assuming that pinch forces of 15 N are sufficient to conduct daily activities, and taking this as requirement for the prostheses tested by Smit et al. (32,33), the TRS hook is the only prosthesis available, which can be operated fatigue-free by the *average* female user. A comparison of different types of upper-limb prostheses showed that the TRS hook, as the only tested voluntary closing body-powered prosthesis, performed superior to other myo-electric and voluntary opening body-powered prosthesis when grasping and lifting a fragile object (51). This finding supports the perception and control theories of this thesis.

Nowadays voluntary opening prostheses are more frequently used than voluntary closing prostheses. Users might not like to continuously apply the required operation forces when holding onto an object with their voluntary closing prosthesis. The locking mechanisms of voluntary closing prostheses cannot maintain the pinch force at which the user intended to lock his prehensor (32). Consequentially, the pinch force exerted on the object might not be sufficient and the object might fall, which is frustrating for the user. The TRS SURE-LOK is placed outside the prehensor on the prosthetic socket and locks the cable in each chosen pinch force configuration (80). Its mechanical properties were not evaluated in (32), but is seemingly a simple and robust solution.

A Voluntary Opening Voluntary Closing (VOVC) device might facilitate voluntary opening supporters with a voluntary closing mode utilizing the same device. The Hand of Dalisch (4), LeBlanc's prehensor (81,82), Nelson's prehensor (81,82), LESA prehensor (83), Kuniholm's prehensors (84), Sullivan's prehensor (85) and Sensinger's prehensor (66) are proposed VOVC devices, which can switch between the modes by manually reversing the spring action. However, none of them is commercially available.

In voluntary opening and voluntary closing body-powered prostheses, springs are returning the prehensor to its initial state. The spring forces must be overcome for each activation. The efficiency of the prehensors could be improved by eliminating the springs and initiating the opening and closing by the user's movement: a two-way controlled prosthesis. Herewith the prehensor opening might appear also more natural, since the prehensor is not fully opened or closed in rest position, but something in between. Feasible movements to control such a prosthesis were evaluated by quantifying maximum cable excursions ('x_{SH}') and excursion dependent maximum forces initiated by these movements ('F_{SH}' in Figure 6–1). An overview table can be found in Appendix C. This information can also be applied to increase the control signals in a body-powered prosthesis, for instance for elbow movement for trans-humeral prostheses or to adapt grasp patterns of prehensors with multiarticulating fingers. Combining many movements implies "learning to activate the appropriate set of synergies and tailor their activation patterns to the task at hand" (71).

Body-powered hand prostheses show poor mechanical functioning (32,33) and the clumsiness of the hand does not only look unnatural, but also limits the view on the object to be grasped. A slender anthropometric hand prosthesis with adaptive fingers facilitates next to the grasping function also a more natural look of the prosthesis and is therefore a desired step towards improved body-powered prostheses. Prototypes of two body-powered prosthesis anthropometric hands were introduced; the Delft Cylinder Hand (86) and the Yale Body-Powered Anthropomorphic Prosthetic Hand (87,88). The challenge to design such a hand lies in the amount of parts the prosthesis consists of, introducing friction and energy losses. The Yale Hand pinches 23 N ('F_{PH}' in Figure 6–1) when exerting 100 N on the control cable ('F_{TM}' in Figure 6–1) (88). The first prototype of the Delft Cylinder Hand requires more than 80 N operation force ('F_{TM}' in Figure 6–1) to archive a 15 N pinch force ('F_{PH}' in Figure 6–1) (86). Both hands require too high input forces ('F_{SH}' in Figure 6–1) to guarantee fatigue-free long-duration operation. To make these new promising body-powered prostheses a successful user-centered design, the design requirements provided by this thesis should not be overlooked.

Transmission systems or servo mechanisms might bridge the gap between the low operation force requirement ('F_{SH}' in Figure 6–1) and the required pinch forces ('FPH' in Figure 6–1) to conduct daily activities with (adaptive) hand prostheses. Bi-phasic mechanisms have been proposed for a voluntary opening elbow controlled prosthesis for children (89,90) and for voluntary closing prostheses utilizing a variable mechanical advantage mechanism (91) or a force demand

valve (92). Like the rider of an e-bike receives pedal assistance, the user of a body-powered prosthesis can receive grip assistance. A power-assisted voluntary closing prehensors have been proposed in the past (93,94), but were not implemented in available prosthesis. The disadvantage of introducing such systems is that it devalues the beauty of body-powered prostheses in terms of independency of external energy, resistance against dirt and water, reliability, and the low mass. However, when external energy resources are required to match input and output forces of the prehensor, haptic display mechanisms should be considered to facilitate adequate feedback of the applied pinch forces and the prehensor's digit positioning ('F_{PH}' and 'x_{PH}' in Figure 6–1). When amplifying the user's control signal the quantification of forces and cable excursions which can be distinguished by the user, such as the just noticeable difference, becomes a necessity. Furthermore, implementing filters might become necessary since the motor control noise will be amplified along with the control signal and the offset of intended and applied forces. "

6.3.4 Future research on prosthesis components

In summary, further developments of alternative harness design ('SH' in Figure 6–1) centralizing user design priorities are strongly encouraged. Additionally, the application of more efficient force transmission systems ('TM' in Figure 6–1) will lower required operation forces and will probably improve the controllability of the prosthesis. Prehensor improvements ('PH' in Figure 6–1) should target a lower mass, higher efficiency, higher pinch force, and lower actuation force according to Smit et al. (32,33), which is supported by the findings of this thesis. VOVC devices and two way controlled prostheses would facilitate the user with the advantages of both control modes, voluntary opening and voluntary closing, in one device and a two way controlled prosthesis is an even more efficient prehensor than current ones due to the elimination of springs. However, neither VOVC nor two way controlled prostheses are available and need to be (further) developed. Two way control movements and its alternative applications, for instance controlling the elbow of a trans-humeral prosthesis or switching between grasping modes of hand prostheses with adaptive fingers, might be worthwhile further investigations. No body-powered anthropometric hand prostheses is currently on the market. Current prototypes consist of many parts which introduces friction and energy losses that result in too high operation forces. Further development in this area is needed. Transmission and power assistance devices might facilitate solutions to lower the high required operation forces, and haptic display mechanisms might become necessary to facilitate adequate feedback. Its application in body-powered prosthesis need to be investigated.

6.4 LIMITATIONS

6.4.1 Focus on user population

This thesis focused on below-elbow deficit cases, operating a shoulder controlled body-powered prosthesis. Conclusions for users with above-elbow deficiencies might be derived from the presented results, but a sharp eye should be kept open especially when estimating the user strength. The biomechanics differ between both cases, below and above elbow deficiencies. Additionally, since the subject population of all studies consisted of only adults no conclusions can be drawn for a pediatric population. Furthermore, the studies conducted were mainly orientated to voluntary closing prehensors, although general conclusions on voluntary opening prehensors can be drawn as well.

6.4.2 Prosthesis users versus controls

Upper-limb prosthesis research is often conducted with controls instead of prosthesis users, often because of the difficulty in obtaining prosthesis users as subjects. It is an ongoing discussion of professionals whether the group of controls is representative for prosthesis users. For the experimental conditions studied, this thesis provides evidence that body-powered prostheses users and controls have equal abilities to perceive and control cable operation forces (Chapter 4 and Chapter 5): the observed magnitudes of force reproduction error and force variability do not differ. However, the strength of prosthesis users is inferior compared to that of controls (Chapter 2) and measured group of users show a large variability in attained maximum forces and in the force reproduction error, whereas the deviation in force variability amongst users was more comparable to that of controls.

The measured control groups in the studies of Chapter 3 and Chapter 5 were only males of 30±8 years and 28±2 years old, whereas the user group was of both genders and 49±13 years of age. Thus, differences can result from these subject characteristics on its own, independent of user or control subject. Hence, drawing conclusions for prosthetic uses from control data is dependent on what and who is measured. It might be advisable to consider another subject population than young healthy male subjects to extend the scope of subject characteristics for enhanced generalization of results.

6.4.3 Prosthesis versus prosthesis simulator

Conducting experiments with controls implies the use of a prosthesis simulator. In this case the prehensor is not placed at the location of the missing natural hand, but distal or lateral to the natural hand. As a consequence the location of the prehensor might feel unnatural and approaching an object with the prehensor might require more visual attention. Additionally, the force distribution might differ from a body-powered prosthesis placed on a user's socket. However, the main conclusions of for instance Chapter 3 will remain unchanged since the low

and high cable force settings were evaluated with the same prosthesis simulator operated by the same subject. Well-trained body-powered prosthesis users might achieve more successful trials in a shorter task completion time than untrained controls.

6.4.4 Fatigue-limit

The fatigue-free operation range in Chapter 2 was determined by multiplying measured maximum forces with a derived value from the literature for isometric contractions (38). Ideally, fatigue-limit experiments should be conducted to relate the percentage of the maximum force of a subject to the duration for which the force can be held onto. The duration of force production should then be compared to the duration a pinch force is exerted on an object during daily activities to derive a more practical related fatigue limit.

6.4.5 Fundamental experiments as a simulation of daily activities

The fundamental experiments conducted in Chapter 4 and 5 simulated the operation of a body-powered prosthesis. A simulation corresponds to daily activities, but is of course not the same as conducting daily tasks with a body-powered prosthesis. The force reproduction task was the most obvious methodology to be able to draw prehensor independent conclusions on preferred operation forces. As already described in the section 'Prosthesis-input requirements or the users' capacities', the set of subjects, experimental set-up and/or procedure of the force reproduction task seemingly resulted in a significant difference of force reproduction errors between force levels in one study (Chapter 5), whereas in the other study no differences in force reproduction error were found between force levels. However, the magnitudes of the force reproduction error of both studies were comparable and the examined forces were overestimated in both studies. Also the findings on force variability were comparable with significant differences between force levels in both studies.

6.4.6 Minimum operation force

The advice on the minimum operation force of 10 N was mainly based on the poor force transmission performance of the Bowden cable. Which minimum forces can be perceived and controlled by a human when operating a body-powered prosthesis has not been investigated in this thesis in detail. The work was mainly focused on the determination of the upper operation force boundary.

6.4.7 Visual feedback

The evaluation of the prosthesis-input requirements were limited to the investigation of perception and control of cable forces based on the proprioceptive & tactile feedback loops (Figure 6–1). The visual feedback loop was only used for locating the prehensor on the object (Chapter 3) and not for achieving the desired pinch force accuracy, since the test object was non-

deformable and thus gives no visual information on the applied pinch force. Consequentially, the visual feedback loop did not contribute to the outcome of the study and was therefore not evaluated. In this thesis no priority was given to the investigation of the visual feedback loop since it is slow and demanding compared to the proprioceptive cues and plays therefore only an ancillary role in achieving high pinch force accuracy.

6.4.8 Discomfort evaluation

The methodology for measuring discomfort in Chapter 2 and 4 does not allow any conclusions on the severity of the experienced discomfort and pain, since the definition of the utilized terms on the body-maps ('feelings' (green), 'discomfort' (orange) and 'pain' (red)) might have been interpreted differently by subjects. Furthermore, the results could only be evaluated per individual and did not allow an inter-subject comparison in terms of locations of experienced discomfort, since no boundaries of body parts were defined on the body-map and thus the subjects were not forced to specify predefined locations.

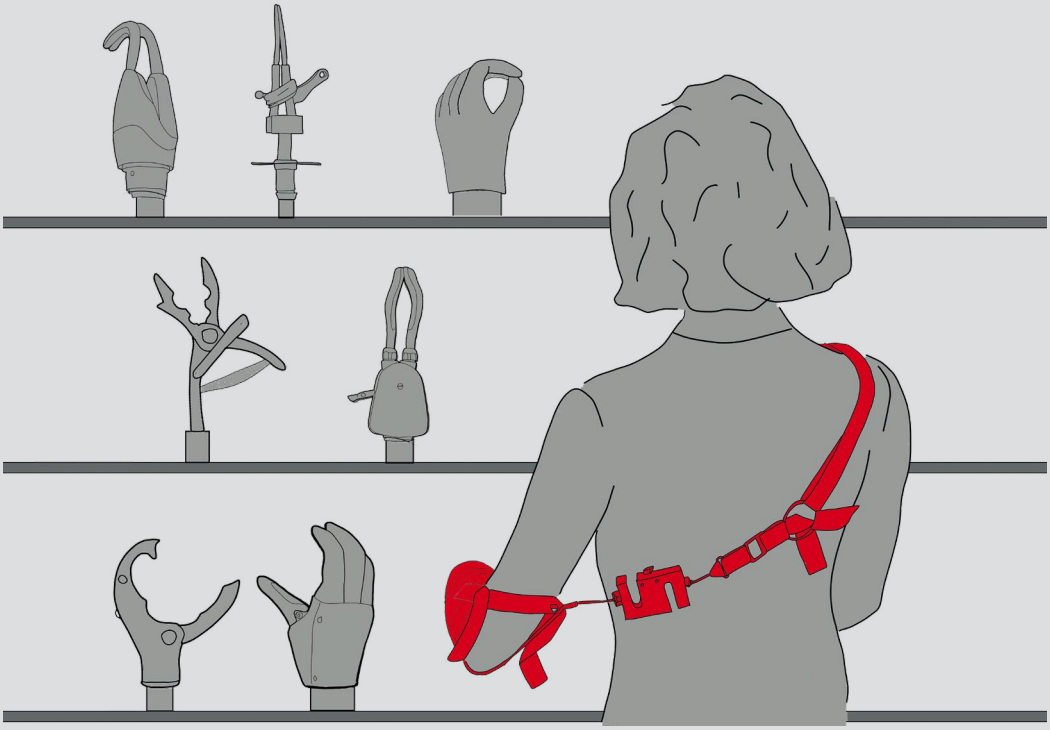
6.5 CLINICAL IMPLICATIONS

This thesis shows that clinical staff should carefully consider user strength when prescribing body-powered prostheses. Prosthesis users have different strength, which might in some cases be insufficient to operate a body-powered prosthesis, or a certain type of prehensor. Anthropometric data might help to estimate the available user strength for females (Chapter 2). Choosing a prehensor that requires low operation forces, for instance the TRS hook, facilitates the user with a better perception and control of his prehensor and the pinch forces he is applying to objects than with a prehensor that requires high operation forces.

The presence of proprioceptive feedback and the intuitiveness of body-powered prosthesis operation result in a shorter training period for new users compared to myo-electric prostheses (17). Increased use of body-powered prostheses therefore could potentially result in lower costs for the public health sector, but would be especially beneficial for the user, because he can conduct daily-life tasks with little training. This was confirmed in the experiments, where both prosthesis users (with no experience in body-powered prosthesis operation) and controls required no more than 5-10 minutes of training before starting the experimental task.

7

CONCLUSIONS



This thesis aimed to quantify user capacities to operate a body-powered prosthesis and establish a better understanding of the human-prostheses interface in order to frame quantified user-centered body-powered prostheses design requirements.

Based on the studies performed in this thesis the following conclusions can be drawn:

- A representative group of prosthesis users with trans-radial defects were able to generate maximum forces on the control cable ranging from 87 to 538 N (mean ± standard deviation: 257±124 N). Male users attained significant higher forces than females, 332±117 N versus 188±87 N (Chapter 2). The wide range of measured forces indicates that clinicians need to evaluate each individual's ability to operate an available body-powered prosthesis based on his strength.
- Applying the assumption that fatigue-free operation requires 20% of maximum forces, cable forces should be kept below 38 N for the average female user and below 66 N for the average male. With these operation forces, most available prostheses cannot create sufficient pinch force to conduct daily activities (Chapter 2).
- The affected upper-arm circumference can predict the maximum cable force exerted by females, which enables clinicians to make a quick estimate whether a body-powered prosthesis is suitable for a patient (Chapter 2).
- Superior pinch force control at low cable forces has been illustrated by the higher number of successful 'mechanical egg' transfers (Chapter 3) and indicated by the increasing force variabilities with increasing target cable forces (Chapter 4 and 5). Large cable excursions support the correct estimation of pinch forces on objects as shown by the decreased force reproduction error with increasing cable excursions (Chapter 4).
- In the force reproduction task described in Chapter 4 the target forces were constantly overestimated. For daily use of a voluntary closing prosthesis this means that the exerted pinch forces are higher than intended at low operation force levels.
- Exerting maximum cable forces provokes discomfort or pain for ~40% of prosthesis users of which two-thirds reported the armpit as affected body part (Chapter 2). Furthermore, the duration of the experiment seemed to provoke discomfort and pain (Chapter 4), which indicates intensive long duration use as a discomfort risk factor even at operation force levels of 40 N and lower with the traditional harness. However, results of the semi-structured interviews conducted in the study described in Chapter 3 suggest a lower long term discomfort (less load on the axillar region, less tiring, less required effort) with lower cable operation forces.
- The Bowden cable introduces inefficiencies and unpredictable behaviour of the exerted forces during dynamical tasks as indicated in the studies of Chapter 3 ('efficiencies between 80 and 84% of the exerted forces') and Chapter 4 ('differences in the measured forces at the back and at the forearm of participants typically ranged between 2 to 3 N, but incidentally even up to 9 N').

- Overall, low cable forces are desired for better prosthesis control. However, friction losses or breathing will influence controllability at very low force levels. Based on the incidentally measured friction losses of 9 N in the study described in Chapter 4, operation forces should not be lower than 10 N.
- The Anchor system is a viable alternative to the traditional harness, since the comfort of the anchor system is reportedly better and force perception and control of both systems were shown to be comparable at low operation forces (Chapter 5).

Summarized, the advice for body-powered prosthesis designers is to keep required cable operation forces as low as possible and employ the fatigue-free force limit as the highest required cable force to conduct daily activities with the improved prosthesis. This is expected to significantly decrease the experienced discomfort during or after prosthesis operation.

However, cable forces below 10 N are likely to complicate the force control for instance due to inefficiencies in the force transmission by the Bowden cable. Therefore, more efficient force transmission systems should be investigated. Additionally, further improved user comfort and force transmission through developments of the harness system are strongly encouraged.

REFERENCES

1. Dijk AJ van. Terminology, classification, registration and epidemiology of defects of the upper extremity [Terminologie, classificatie, registratie en epidemiologie van defecten aan de bovenste extremiteit]. In: Brouwers MAH, Mulders A, editors. Amputatie en Prothesiologie van de bovenste extremiteit. Nijmegen: Bureau PAOG-Heyendael; 2007. p. 10–35.
2. Ziegler-Graham K, MacKenzie EJ, Ephraim PL, Travison TG, Brookmeyer R. Estimating the prevalence of limb loss in the United States: 2005 to 2050. Archives of Physical Medicine and Rehabilitation. 2008;89(3):422–9.
3. Ephraim PL, Dillingham TR, Sector M, Pezzin LE, MacKenzie EJ. Epidemiology of limb loss and congenital limb deficiency: a review of the literature. Archives of Physical Medicine and Rehabilitation. 2003;84(5):747–61.
4. Löffler L. Substitutes for the upper extremity [in German: Der Erzatz für die obere Extremität]. Stuttgart: Enke Verlag; 1984.
5. Borchard M, Hartmann K, Leymann, Schlesinger G. Limb substitudes and labour assistive devices for war casualties and traumatic injured persons [in German: Ersatzglieder und Arbeitshilfen für Kriegsgeschädigte und Unfallverletzte]. Berlin: Verlag von Julius Springer; 1919.
6. Dorrance DW. Artificial hand. Vie medicale Paris France 1920. 1912. p. 32–4.
7. Reiter R. A new electric hand prosthesis [In German: Eine neue Elektrokunsthand]. Grenzgebiete der Medizin. 1948;1(4):133–5.
8. Plettenburg DH. Upper extremity prosthetics - current status and evaluation. 1st ed. Delft: VSSD; 2006.
9. Arm of William Carnes [Internet]. Smithsonian National Museum of American History - The Price of Freedom: Americans at War. 1915 [cited 2016 Nov 11]. Available from: http://amhistory.si.edu/militaryhistory/collection/object.asp?ID=609
10. Weeks DL, Anderson DI, Wallace S a. The role of variability in practice structure when learning to use an upper-extremity prosthesis. Prosthetic and Orthotic Science. 2003;15(3):84–92.
11. Raspopovic S, Capogrosso M, Petrini FM, Bonizzato M, Rigosa J, Di Pino G, et al. Restoring natural sensory feedback in real-time bidirectional hand prostheses. Science Translational Medicine. 2014;6(222).
12. Peripheral nerve interface - Problems and limitations [Internet]. Wikipedia: The Free Encyclopedia. Wikimedia Foundation, Inc. [cited 2016 Aug 28]. Available from: https://en.wikipedia.org/wiki/Peripheral_nerve_interface#Problems_and_limitations
13. Collinger JL, Wodlinger B, Schwartz AB. 7 degree-of-freedom neuroprosthetic control by an individual with tetraplegia. Lancet. 2012;381(9866):557–64.
14. University of Pittsburgh Medical Center. One giant bite: Woman with quadriplegia feeds herself chocolate using mind-controlled robot arm [Internet]. Available from: https://www.youtube.com/watch?v=76lIQtE8oDY
15. Biddiss EA, Chau TT. Upper limb prosthesis use and abandonment: a survey of the last 25 years. Prosthetics and Orthotics International. 2007;31(3):236–57.
16. Resnik L, Meucci MR, Lieberman-Klinger S, Fantini C, Kelty DL, Disla R, et al. Advanced upper limb prosthetic devices: Implications for upper limb prosthetic rehabilitation. Archives of Physical Medicine and Rehabilitation. Elsevier Inc.; 2012;93(4):710–7.
17. Carey SL, Lura DJ, Highsmith MJ. Differences in myoelectric and body-powered upper-limb prostheses: Systematic literature review. Journal of Research Rehabilitation and Development. 2015;52(3):247–62.
18. D'Alonzo M, Cipriani C, Carrozza MC. Vibrotactile sensory substitution in multi-fingered hand prostheses: Evaluation studies. In: Proceedings of the IEEE International Conference on Rehabilitation Robotics (ICORR); 2011 June 29 -July 1. Zurich, Switzerland: IEEE; 2011. p. 928–33.
19. Witteveen HJ, Rietman HS, Veltink PH. Vibrotactile grasping force and hand aperture feedback for myoelectric forearm prosthesis users. Prosthetics and orthotics international. 2014;39(3):204–12.
20. Meek SG, Jacobsen SC, Goulding PP. Extended physiologic taction: Design and evaluation of a proportional force feedback system. Journal of Rehabilitation Research and Development. 1989;26(3):53–62.
21. Patterson PE, Katz JA. Design and evaluation of a sensory feedback system that provides grasping pressure in a myoelectric hand. Journal of rehabilitation research and development. 1992;29(1):1–8.

22. Patterson PE, Katz JA. A sensory feedback system for grasping pressure in a myoelectric hand. In: IEEE Engineering in Medicine & Biology Society 10th annual internationsal Conference. 1988. p. 1564–5.
23. Shannon GF. A myoelectrically-controlled prosthesis with sensory feedback. Medical & biological engineering & computing. 1979;17(1):73–80.
24. Scott RN, Brittain RH, Caldwell RR, Cameron AB, Dunfield VA. Sensory-feedback system compatible with myoelectric control. Medical and Biological Engineering and Computing. 1980;18(1):65–9.
25. Prior RE, Lyman J. Electrocutaneous feedback for artificial limbs. Summary progress report. February 1, 1974, through July 31, 1975. Bulletin of prosthetics research. 1975.
26. Wheeler J, Bark K, Savall J, Cutkosky M. Investigation of rotational skin stretch for proprioceptive feedback with application to myoelectric systems. IEEE Transactions on Neural Systems and Rehabilitation Engineering. 2010;18(1):58–66.
27. Gillespie RB, Contreras-Vidal JL, Shewokis PA, O'Malley MK, Brown JD, Agashe H, et al. Toward improved sensorimotor integration and learning using upper-limb prosthetic devices. In: Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC), 2010 Aug 31-Sept 4. 2010. p. 5077–80.
28. Mugge W, Abbink DA, Schouten AC, Van Der Helm FCT, Arendzen JH, Meskers CGM. Force control in the absence of visual and tactile feedback. Experimental Brain Research. 2013;224(4):635–45.
29. Brown JD, Kunz T, Gardner D, Shelley MK, Davis AJ, Gillespie RB. An empirical evaluation of force feedback in body-powered prostheses. IEEE Transactions on Neural Systems and Rehabilitation Engineering. 2016;PP(99):1–1.
30. Vargesson N. Thalidomide-induced limb defects: Resolving a 50-year-old puzzle. BioEssays. 2009;31(12):1327–36.
31. Carlson LE, Long M. Quantitative evaluation of body-powered prostheses. Chicago, IL, USA: American Soc of Mechanical Engineers (ASME), Dynamic Systems and Control Division; 1988. 1-16 p.
32. Smit G, Plettenburg DH. Efficiency of voluntary closing hand and hook prostheses. Prosthetics and Orthotics International. 2010;34(4):411–27.
33. Smit G, Bongers RM, Van der Sluis CK, Plettenburg DH. Efficiency of voluntary opening hand and hook prosthetic devices: 24 years of development? The Journal of Rehabilitation Research and Development. 2012;49(4):523.
34. Leblanc M, Setoguchi Y, Shaperman J, Carlson L. Mechanical work efficiencies of body-powered prehensors for young children introduction. Journal of the Association of Children's Prosthetic-Orthotic Clinics. 1992;27(3):70.
35. Biddiss E, Beaton D, Chau T. Consumer design priorities for upper limb prosthetics. Disability and Rehabilitation Assistive Technology. 2007;2(6):346–57.
36. Carlson LE, Radocy B, Marschall PD. Spectron 12 cable for upper-limb prostheses. JPO Journal of Prosthetics and Orthotics. 1991;3(3):130.
37. Carlson LE, Veatch BD, Frey DD. Efficiency of prosthetic cable and housing. Journal of Prosthetics and Orthotics. 1995;7(3):96–9.
38. Monod H. Contractility of muscle during prolonged static and repetitive dynamic activity. Ergonomics. 1985;28(1):81–9.
39. Taylor CL. The biomechanics of the normal and of the amputated upper extremity. In: Klopsteg P, Wilson P, editors. Human limbs and their substitutes. New York, NY: McGraw-Hill; 1954. p. 169–221.
40. Delft Prosthetics BV: the WILMER Open Fitting [Internet]. [cited 2016 Jun 20]. Available from: http://www.delftprosthetics.nl/nl/producten/open-fitting
41. Churchill E, Churchill T, Downing K, Erskine P, Laubach LL, McConville JT. Anthropometric source book volume II - A handbook of anthropometric data. Project S of AR, Associates W, Yellow Springs O, editors. NASA Reference Publication 1024. NASA; 1978.
42. Kejlaa GH. Consumer concerns and the functional value of prostheses to upper limb amputees. Prosthetics and Orthotics International. 1993;17(3):157–63.
43. Biddiss E, Chau T. Upper-limb prosthetics: critical factors in device abandonment. American Journal of Physical Medicine & Rehabilitation. 2007 Dec;86(12):977–87.

44. Micera S, Carpaneto J, Raspopovic S. Control of hand prostheses using peripheral information. IEEE Reviews in Biomedical Engineering. 2010;3:48–68.
45. Wiener N. Cybernetics - Or control and communication in the animal and the machine. New York, USA; Paris, France: The Technology Press - John Wiley & Sons, Inc., New York Hermann et Cie, Paris; 1948.
46. Simpson D. The choice of control system for the multimovement prosthesis: Extended physiological proprioception (e.p.p.). In: Herberts P, Kadefors R, Magnusson R, Petersen I, editors. The Control of Upper-Extremity Prostheses and Orthoses. Springfield; 1974. p. 146–50.
47. Biddiss E, McKeever P, Lindsay S, Chau T. Implications of prosthesis funding structures on the use of prostheses: Experiences of individuals with upper limb absence. Prosthetics and Orthotics International. 2011 Jun;35(2):215–24.
48. Smit G. Natural grasping -Design and evaluation of a voluntary closing adaptive hand prosthesis. [Delft]: Delft University of Technology; 2013.
49. Hichert M, Plettenburg DH, Vardy AN. A need for a more used-centred design in body powered prostheses. Proceedings of the Myo Electric Control Symposium - MEC '14 Redefining the Norm, 2014 Aug 18-22. Fredericton, Canada: University of New Brunswick; 2014;91–5.
50. Plettenburg DH, Hichert M, Smit G. Feedback in voluntary closing arm prostheses. In: Proceedings of the Myo Electric Control Symposium - MEC '11 Raising the Standard, 2011 Aug 14-19. Fredericton, Canada: University of New Brunswick; 2011. p. 74–8.
51. Lum PS, Black I, Holley RJ, Barth J, Dromerick AW. Internal models of upper limb prosthesis users when grasping and lifting a fragile object with their prosthetic limb. Experimental Brain Research. 2014;232(12):3785–95.
52. Bos RA, Mahmood MN, Kooren PN, Verros S, Paalman MI, Koopman HFJM, et al. Measuring dynamic friction properties of Bowden cable systems. In: The Anthology of the ISPO European Congress on Prosthetics and Orthotics, 6-8 October 2016, Rotterdam, The Netherlands. Rotterdam: International Society of Prosthetics and Orthotics; 2016. p. 45.
53. Latour D, Sabolevski T, Lajoie-Weaver K. Ipsilateral scapular cutaneous anchor. In: Proceedings of the 12th World Congress of the International Society for Prosthetics and Orthotics, 2007 July 9 –Aug 3. Vancouver, Canada: International Society of Prosthetics and Orthotics; 2007. p. 555.
54. Hichert M, Abbink DA, Kyberd P, Plettenburg DH. High cable forces deteriorate pinch force control in voluntary- closing body-powered prostheses. PLoS ONE. 2017;12: e0169996. doi:10.1371/journal.pone.0169996.
55. Onneweer B, Mugge W, Schouten AC. Force preproduction error depends on force level, whereas the position reproduction error does not. IEEE Transactions on Haptics. 2016;9(1):54–61.
56. Shergill SS, Bays PM, Frith CD, Wolpert DM. Two eyes for an eye: the neuroscience of force escalation. Science. 2003;301(July):187.
57. Walsh LD, Taylor JL, Gandevia SC. Overestimation of force during matching of externally generated forces. The Journal of Physiology. 2011;589:547–57.
58. Jones LA, Hunter IW. Force sensation in isometric contractions: a relative force effect. Brain Research. 1982;244(1):186–9.
59. Schmidt R a, Zelaznik H, Hawkins B, Frank JS, Quinn JT. Motor-output variability: a theory for the accuracy of rapid motor acts. Psychological review. 1979;47(5):415–51.
60. Todorov E. Cosine tuning minimizes motor errors. Neural computation. 2002;14(6):1233–60.
61. Hamilton AF de C, Jones KE, Wolpert D. The scaling of motor noise with muscle strength and motor unit number in humans. Experimental brain research. 2004;157(4):417–30.
62. Mugge W, Schuurmans J, Schouten AC, van der Helm FCT. Sensory weighting of force and position feedback in human motor control tasks. Journal of Neuroscience. 2009 Apr 29;29(17):5476–82.
63. Shaperman J, LeBlanc M. Prehensor grip for children: A survey of the literature. Journal of Prosthetics and Orthotics. 1995;7(2):61–4.
64. Van Der Niet O, Reinders-Messelink HA, Bongers RM, Bouwsema H, Van Der Sluis CK. The i-LIMB hand and the DMC plus hand compared: A case report. Prosthetics and Orthotics International. 2010 Jun;34(2):216–20.

65. Berning K, Cohick S, Johnson RE, Miller LA, Sensinger JW. Comparison of body-powered voluntary opening and voluntary closing prehensor for activities of daily life. Journal of rehabilitation research and development. 2014 Jan;51(2):253–61.
66. Sensinger JW, Lipsey J, Thomas A, Turner K. Design and evaluation of voluntary opening and voluntary closing prosthetic terminal device. Journal of Rehabilitation Research & Development. 2015;52(1):63–76.
67. Collier M, LeBlanc M. Axillar bypass ring for shoulder harnesses for upper-limb prostheses. Journal of Prosthetics and Orthotics. 1996;8(4):130–1.
68. Jonathan Kuniholm. Improved body-powered harness [Internet]. Open Prosthetics Project Wiki. 2008. Available from: https://web.archive.org/web/20150630141228/http:/openprosthetics.wikispot.org/Improved_Body_Powered_Harness
69. Latour DA. Patent: US8821588 B2. 2006.
70. Antfolk C, D'Alonzo M, Rosén B, Lundborg G, Sebelius F, Cipriani C. Sensory feedback in upper limb prosthetics. Expert Review of Medical Devices. 2013;10(1):45–54.
71. Bongers RM, Kyberd PJ, Bouwsema H, Kenney LPJ, Plettenburg DH, Van der Sluis CK. Bernstein's levels of construction of movements applied to upper limb prosthetics. Journal of Prosthetics and Orthotics. 2012 Apr;24(2):67–76.
72. Hichert M, Abbink DA, Vardy AN, Sluis CK van der, Janssen W, Brouwers MAH, et al. Perception and control of low cable operation forces in voluntary closing body-powered upper-limb prostheses. IEEE Transactions of Human Machine Systems (submitted).
73. Radocy B. TRS - Cutaneous Anchor with Debbie Latour [Internet]. 2015 [cited 2016 Jun 24]. Available from: https://www.youtube.com/watch?v=y2v7BNKJ-aA
74. Hichert M, Vardy AN, Plettenburg DH. Fatigue-free operation of most body-powered prostheses not feasible for majority of users with trans-radial deficiency. Prosthetics and Orthotics International (submitted).
75. Keller AD, Taylor GL, Zahm V. Studies to determine the functional requirements for hand and arm prostheses. The final report covering work during the year 1946-1947. 1947.
76. Belter JT, Reynolds BC, Dollar AM. Grasp and force based taxonomy of split-hook prosthetic terminal devices. In: Engineering in Medicine and Biology Society (EMBC), 2014 36th Annual International Conference of the IEEE. 2014. p. 17–21.
77. Pursley RJ. Harness patterns for upper-extremity prostheses. Artificial limbs. 1955;2(3):26–60.
78. Gudfinnsdottir T. Design and evaluation of a new harnessing system for body-powered prostheses. Delft University of Technoloy; 2013.
79. LeBlanc M a. Current evaluation of hydraulics to replace the cable force transmission system for body-powered upper-limb prostheses. Assistive Technology. 1990;2(3):101–7.
80. TRS Prosthetics Inc.: TRS SURE-LOK [Internet]. [cited 2016 Sep 27]. Available from: http://www.trsprosthetics.com/product/sure-lok-cable-system/
81. LeBlanc M, Parker D, Nelson C. New designs for prosthetic prehensors. In: Proceedings of the 9th International Symposium on External Control of Human Extremities, August 31-September 5. 1987. p. 475–81.
82. Meeks D, Leblanc M. Preliminary assessment of three new designs of prosthetic prehensors for upper limb amputees. Prosthetics and orthotics international. 1988;12:41–5.
83. Veatch BD. A combination VO/VC terminal device with variable mechanical advantage. American Academy of Orthotics and Prosthetics. 2004;
84. Jonathan Kuniholm. Voluntary opening voluntary closing prehensor [Internet]. Open Prosthetics Project. [cited 2016 Sep 29]. Available from: https://web.archive.org/web/20160310024122/http:/openprosthetics.org/body-powered
85. Sullivan T, Teh KS. Design and fabrication of a hybrid body-powered prosthetic hand with voluntary opening and voluntary closing capabilities. Proceedings of the ASME 2011 International Mechanical Engineering Congress & Exposition. 2011;1–8.
86. Smit G, Plettenburg DH, Van Der Helm FCT. The lightweight Delft Cylinder hand: First multi-articulating hand that meets the basic user requirements. IEEE Transactions on Neural Systems and Rehabilitation Engineering. 2015;23(3):431–40.

87. Belter JT, Dollar AM. Progress towards the Yale Body-powered anthropomorphic prosthetic hand, mechanical coupling methods. In: MEC "14 Redefining the Norm University of New Brunswick"s Myoelectric Controls/ Powered Prosthetics Symposium. Fredericton, Canada: University of New Brunswick; 2014. p. 22–6.
88. Belter JT, Leddy MT, Jr KDG, Dollar AM. Comparative clinical evaluation of the Yale multigrasp hand. 2016;
89. Kruit J, Cool JC. Body-powered hand prosthesis with low operating power for children. Journal of Medical Engineering & Technology. 1989;13(1/2):129–33.
90. van Eibergen Santhagens RA. The ellbow controlled handprosthesis for children [in Dutch: De ellebooggestuurde kinderhandprothese]. Delft; 1985.
91. Frey DD, Carlson LE. A body powered prehensor with variable mechanical advantage. Prosthetics and Orthotics International. 1994;18:118–23.
92. Bruinekool RA, Plettenburg IDH. Pneumatic pinch force magnification in voluntary closing hand prostheses. Delft University of Technology; 2012.
93. Rombout D. Design of a power-assistance system for body-powered hand prosheses. Delft University of Technology; 2016.
94. Plettenburg DH. A sizzling hand prosthesis: On the design and development of a pneumatically powered hand prosthesis for children. Delft University of Technology; 2002.

APPENDIX

Appendix A. Extent and locations of sensations provoked by the exertion of cable forces.

Subject #	Touchiness (=green)	Irritation (=orange)	Pain (=red)
1	–	–	–
5	–	–	–
8	–	–	–
12	–	–	–
3	left and right armpit	–	–
17	left and right neck & upper & lower back	–	–
10	left armpit	–	–
21	left armpit & back (sensor) & stump	–	–
2	left neck	–	–
18	left neck	–	–
9	right shoulder	–	–
13	right upper arm	–	–
23	upper back	–	–
15	upper back (sensor) & left armpit	–	–
20	back (harness)	left armpit	–
6	–	left elbow & right armpit	–
14	–	right armpit	–
16	–	right armpit	–
11	left shoulder & neck	right stump	–
19	–	–	back (harness)
22	right armpit	–	left armpit
7	right shoulder	left upper back	left neck
4	upper back (harness)	left elbow	right armpit

Appendix B. Subject characteristics and anthropometric measures. Subjects are sorted by gender and the cause of their arm defect (indicated by the horizontal lines).

Subject no.	Gender	Age	Acquired/ congenital defect	Affected side	Dominant side	Max. cable force [N]	Body-maps‡	Shoulder width*	Affected upper-arm length**	Sound upper-arm circumference ***	Affected upper-arm circumference***	Remnant length****	Weight [kg]	Height [cm]
11	f	60	acq.	r	r	87	2	39,7	31,7	25,0	23,6	20,9	57	164
9	f	54	acq.	l	r	134	1	35,1	33,9	27,3	26,0	24,1	58	172
14	f	69	acq.	l	r	198	2	40,3	33,3	36,8	34,3	15,4	93	168
12	f	52	cong.	r	r	100	0	36,8	33,2	26,7	22,5	9,1	60	169
7	f	49	cong.	l	r	118	3	38,2	36,5	31,5	29,0	10,0	70	177
19	f	58	cong.	l	r	148	3	39,3	31,4	28,3	27,2	14,7	64	161
8	f	46	cong.	l	r	164	0	35,7	33,3	26,5	24,5	16,7	57	161
2	f	20	cong.	l	r	165	1	36,4	35,5	26,0	23,2	10,5	64	168
4	f	43	cong.	l	l	182	3	40,0	35,4	33,4	27,2	9,7	94	173
18	f	34	cong.	l	l	290	1	41,0	33,0	33,0	31,0	18,0	89	168
10	f	54	cong.	r	l	308	1	37,5	33,5	32,0	26,8	10,5	68	168
17	f	43	cong.	r	l	360	1	39,4	28,4	23,8	34,0	14,9	91	170
23	m	49	acq.	l	r	229	1	48,0	35,4	45,5	46,5	23,5	-	186
1	m	57	acq.	l	r	360	0	42,0	37,5	34,0	30,5	17,0	103	187
3	m	68	acq.	r	r	369	1	42,3	35,4	34,5	32,0	18,4	81	170
16	m	50	acq.	l	l	442	2	45,0	35,0	32,0	28,0	29,0	98	184
6	m	35	acq.	l	r	490	2	41,2	35,0	27,8	27,5	31,9	68	178
20	m	68	acq.	r	r	538	2	44,2	34,7	31,0	26,0	20,0	85	179
22	m	55	cong.	l	l	199	3	41,5	34,5	28,3	24,9	23,2	62	176
5	m	63	cong.	l	r	213	0	41,0	32,0	31,5	25,5	10,7	74	172
21	m	37	cong.	r	r	259	1	45,5	35,1	31,0	30,0	12,9	82	185
15	m	25	cong.	r	r	273	1	40,9	34,0	32,0	28,5	12,3	76	178
13	m	47	cong.	r	r	278	1	45,0	37,5	31,2	24,3	8,9	82	186

‡ Body-maps: 0=none, 1= mild sensation, 2=irritation, 3=pain

anthropometric data, in cm, were taken following the instructions of the NASA Reference Publication 1024

* 103. Biacromial Breadth

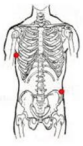
** 751. Shoulder-Elbow Length

*** 113. Biceps Circumference, Relaxed

**** 381. Forearm-Hand Length (the fingertips are represented by the far end of the subjects' stump)

Appendix C. Feasible movements and their attainable cable excursions and excursion related cable forces for a two-way controlled prosthesis.

The movements are ordered on decreasing operation force. The virtual line between the two landmarks indicates the line of action. Values are indicated as mean (standard deviation) over all subjects. F1 – F4 are the mean forces measured at 0% displacement – 75% displacement.

Movement	Landmarks	Main muscles involved		Displ. (cm)	F1 (N) -0%-	F2 (N) -25%-	F3 (N) -50%-	F4 (N) -75%-
Elevation 		-M. trapezius, pars descendens -M. levator scapulae -M. rhomboideus minor -M. rhomboideus major	Left	8.05 (2.26)	98.57 (28.48)	75.95 (28.04)	53.95 (18.72)	29.77 (13.19)
			Right	8.55 (1.99)	103.59 (32.11)	86.53 (31.37)	57.30 (25.64)	34.83 (21.46)
Flexion trunk 		-M. quadratus lumborum -M. psoas major -M. obliquus externus abdominis -M. obliquus internus abdominis	Left	5.45 (1.70)	77.11 (26.35)	56.89 (22.94)	30.99 (11.78)	12.50 (8.07)
			Right	5.35 (1.73)	84.81 (31.90)	57.20 (26.74)	30.52 (16.05)	13.27 (9.53)
Protraction 		-M. serratus anterior -M. pectoralis minor	Left	3.70 (0.66)	46.41 (10.54)	37.27 (11.05)	25.67 (10.20)	16.11 (8.71)
			Right	3.75 (0.85)	57.74 (11.46)	42.14 (13.54)	29.95 (10.56)	20.29 (11.66)
Rotation trunk 		-M. transversus abdominis -M. obliquus externus abdominis -M. obliquus internus abdominis	Left	4.70 (1.87)	32.73 (8.84)	23.72 (8.81)	15.72 (7.43)	10.88 (6.48)
			Right	5.45 (1.61)	40.36 (12.36)	29.75 (8.34)	18.77 (7.94)	13.72 (6.77)
Flexion toe 		-M. flexor hallucis brevis -M. flexor hallucis longus -M. flexor digitorum brevis -M. flexor digitorum longus -Mm. lumbricales	Left	1.75 (0.50)	22.15 (6.47)	-	12.67 (5.10)	-
			Right	1.75 (0.50)	22.39 (7.26)	-	13.34 (5.25)	-

SUMMARY

In the Netherlands approximately 3750 persons have an arm defect: they miss (part of) their hand, forearm or even their entire arm. The majority of these people are in the possession of a prosthesis. This prosthesis can be purely cosmetic, or offer the user some grasping function. The latter can either be a body-powered or a myo-electric prosthesis. A myo-electric prosthesis is controlled by electrical signals originating from the contraction of muscles of the user and is powered by electric motors. Body-powered prostheses are operated by body movements, which are captured by a harness and transmitted through a Bowden cable to the prehensor. Unfortunately, 23-45% of the users are so dissatisfied with their chosen prosthesis that they decide not to wear it. Thus, prostheses need to be improved.

This thesis focuses on the improvement of body-powered prostheses, which offer several advantages compared to myo-electric prostheses: they are much lighter, cheaper and more reliable and – perhaps most importantly – offer the user extended proprioceptive feedback about the prehensor's movements and exerted grip force. On the down side, body-powered prostheses currently require high operation forces, causing pain and fatigue during or after use, and potentially limiting the inherent advantages in perception and control. Additionally, users complain about the comfort and outer appearance of the harness, the design of which still looks like that of the Count of Beaufort in 1860.

Lowering the operation forces will most likely increase the pinch force control accuracy and reduce fatigue and pain during or after operation and therefore improve the prosthesis' functionality. To which level cable forces need to be lowered is up till now unknown; it is assumed that lowering operation forces is effective, but only up the point where the control forces are still clearly distinguishable from noise (like inefficiencies in prehensor or cable friction).

The goal of this thesis is to quantify the perception and control capabilities of prosthesis users as a function of body-powered prosthesis design elements, such as mechanical properties of the prehensor, or an alternative harness. The obtained quantified understanding is intended to guide improvements in body-powered prosthesis design, to enhance the quality of life of upper-limb prosthesis users and to prevent (repetitive strain) injuries.

First, a range of maximum cable operation forces between 87 N and 538 N was established for a representative group of prosthesis users (Chapter 2). When the corrected values for fatigue-free operation (20% of the individually measured maximum force) were compared to the required operation forces of ten commercially available body-powered prostheses, it was concluded that only one of these could be operated fatigue-free. Based on the available results, cable forces should not exceed 38 N for the average female, and 66 N for the average male for most activities in daily life, to enable users to operate their prosthesis fatigue-free.

A second study investigated the effect of cable operation forces (15 N versus 51 N) on the ability to transport a test object (Chapter 3). The object was a mechanical egg: too high cable forces would 'break' the object; too low cable forces would cause the operator to drop it. The results indicated that the egg was transferred successfully more often at the low cable operation force settings than at the high force setting.

A third study investigated users' perception and control abilities by utilizing a force reproduction task (Chapter 4). For successful object manipulation we desire a small difference between the intended and actually applied force on an object, as well as only minor fluctuations in the applied force level. In a force reproduction task the force reproduction error resembles the difference between the intended and actually applied force, whereas the force variability indicates the force fluctuations. The results showed a decreasing force reproduction error with increasing cable excursions for force levels of 10 and 20 N, and a decreasing force variability for decreasing operation force levels varying between 10 and 40 N. Thus, low force levels and large cable excursions contribute to improved force perception and control.

In the fourth and final study an alternative harness design, the Ipsilateral Scapular Cutaneous Anchor System, was compared with the traditional figure-of-nine harness, as comfort of the harness was identified as being an issue in body-powered prosthesis (Chapter 5). In terms of perception and control capacities of users no differences between the two systems were found for operation forces ranging from 10 to 40 N. It could thus be concluded that the Anchor system appears to be a valid alternative to the traditional harness at low operation force levels as performance is comparable while comfort is reportedly better.

In conclusion, this thesis shows that the operation forces which prosthesis users are required to exert are an important factor in body-powered prosthesis design. For most commercially available body-powered prostheses, the control cable forces are too high to be used on a daily basis. To enable users to operate a body-powered prosthesis fatigue-free during the day – every day – with the provision of high quality feedback and adequate prehensor control, operation forces should not exceed 38 N for the average female and 66 N for the average male user. A long operation movement stroke and thus a large cable excursion does contribute to increased prehensor control. For the suggested low operation force levels the Ipsilateral Scapular Cutaneous Anchor System provides a good alternative for the traditional harness.

SAMENVATTING

In Nederland hebben ongeveer 3750 personen een armdefect: zij moeten een (deel van) hun hand, onderarm of zelfs de hele arm missen. De meesten van deze mensen hebben een prothese. Deze prothese kan puur cosmetisch van aard zijn, of een grijpfunctie hebben die de gebruiker in staat stelt dagelijkse activiteiten uit te voeren. De laatstgenoemde prothese kan lichaamsbekrachtigd of myo-electrisch zijn. Een myo-electrische prothese wordt bestuurd door middel van elektrische signalen die geassocieerd zijn met spiercontracties van de gebruiker en is aangedreven door elektromotoren. Lichaamsbekrachtigde prothesen worden met lichaamsbewegingen aangestuurd. Deze bewegingen worden via een harnas en Bowden kabel doorgegeven aan het grijpmechanisme (prehensor). Helaas is zo'n 23–45% van de gebruikers dusdanig ontevreden met hun prothese dat ze besluiten deze niet te dragen. Om kort te gaan, prothesen moeten verbeterd worden.

Dit proefschrift richt zich op het verder verbeteren van lichaamsbekrachtige prothesen, welke verschillende voordelen bieden ten opzichte van myo-electrische prothesen: ze zijn lichter, goedkoper, betrouwbaarder en -wellicht het belangrijkst- bieden de gebruiker proprioceptieve terugkoppeling over de bewegingen van het grijpmechanisme en de krachten die hiermee worden uitgeoefend. Daar staat tegenover dat de huidige lichaamsbekrachtige prothesen hoge bedieningskrachten vereisen, wat tot pijn en vermoeidheid tijdens of na het gebruik kan leiden, en mogelijk de inherente voordelen op het gebied van krachtwaarneming en besturing kan beperken. Daar komt bij dat gebruikers klachten hebben over het comfort en uiterlijk van het harnas, waarvan het ontwerp niet veel verschilt van dat van de Graaf van Beaufort uit 1860.

Een verlaging van de vereiste bedieningskrachten zal naar verwachting de precisie van de knijpkrachtbesturing ten goede komen, en vermoeidheid en pijn tijdens of na gebruik verminderen. Daarmee wordt de functionaliteit van de prothese verbeterd. Tot op welk niveau de kabelkrachten moeten worden teruggebracht is tot nu toe niet bekend; aangenomen wordt dat een verlaging van de bedieningskrachten alleen effectief is zolang deze nog duidelijk te onderscheiden zijn van ruis (zoals inefficiënties in het grijpmechanisme of wrijvingsverliezen in de Bowden kabel).

Het doel van dit proefschrift is het kwantificeren van de waarnemings- en besturingscapaciteiten van prothesegebruikers ten aanzien van ontwerpaspecten van lichaamsbekrachtigde prothesen, zoals de mechanische eigenschappen van het grijpmechanisme, of het type harnas. Het verkregen kwantitatieve begrip is bedoeld om als leidraad te dienen voor verbeteringen in het ontwerp van lichaamsbekrachtigde prothesen, om de levenskwaliteit van armprothesegebruikers te verbeteren en om lichamelijke klachten en blessures te voorkomen.

Als eerste is een bereik van maximaal haalbare kabelkrachten vastgesteld voor een representatieve groep prothesegebruikers (Hoofdstuk 2). Dit blijkt tussen de 87 N en 538 N te liggen. Als deze krachten, gecorrigeerd voor vermoeidheidsvrij gebruik (20% van de individueel

gemeten maximum kracht), worden vergeleken met de vereiste bedieningskrachten van tien commerciëel verkrijgbare lichaamsbekrachtigde prothesen, luidt de conclusie dat slechts één van deze zonder vermoeidheid gebruikt kan worden. Op basis van de verkregen resultaten zouden kabelkrachten niet hoger moeten zijn dan 38 N voor de gemiddelde vrouw, of 66 N voor de gemiddelde man voor de meeste dagelijkse activiteiten, zodat de gebruikers hun prothese zonder vermoeidheid kunnen gebruiken.

Een tweede studie heeft de invloed onderzocht die kabelkrachten (15 N en 51 N) hebben op de vaardigheid om een testvoorwerp te verplaatsen (Hoofdstuk 3). Dit voorwerp was een 'mechanisch ei': te hoge kabelkrachten laten het ei 'breken'; bij te lage kabelkrachten valt het ei op de grond. De resultaten laten zien dat het ei vaker succesvol verplaatst werd bij lage kabelkrachten dan bij hoge.

Een derde studie onderzocht de waarnemings- en besturingscapaciteiten van gebruikers door middel van een kracht-reproductie taak (Hoofdstuk 4). Voor succesvolle hantering van een voorwerp is het wenselijk dat het verschil tussen de voorgenomen en de uitgeoefende kracht op het voorwerp klein is, evenals schommelingen in het uitgeoefende krachtniveau. In een kracht-reproductie taak wordt de reproductiefout weergegeven door het verschil tussen de voorgenomen en de werkelijk uitgeoefende kracht. De krachtvariabiliteit geeft de schommelingen in de kracht weer. Uit de resultaten blijkt dat de reproductiefout afneemt bij toenemende kabel uitslagen voor krachten van 10 en 20 N, en dat de krachtvariabiliteit afneemt bij afnemende bedieningskrachten tussen de 10 en 40 N. Lage krachtniveaus en grote kabel uitslagen dragen dus bij aan een verbetering van krachtwaarneming en besturing.

In de vierde en laatste studie is een alternatief harnasontwerp, het Ipsilateraal Scapulair Huid Anker Systeem, vergeleken met een traditioneel harnas, aangezien draagcomfort van het harnas belangrijk is bij het gebruik van lichaamsbekrachtigde prothesen (Hoofdstuk 5). Er werden geen verschillen gevonden in waarnemings- en besturingscapaciteiten van gebruikers tussen de twee systemen voor krachten tussen de 10 en 40 N. De conclusie is dus dat bij deze lage kabelkrachten het Anker Systeem een volwaardig alternatief is voor het tranditionele harnas, aangezien het vergelijkbare prestaties levert maar meer comfort biedt.

Concluderend, dit proefschrift laat zien dat de bedieningskrachten die prothesegebruikers moeten uitoefenen een belangrijke factor in het ontwerp van lichaamsbekrachtigde prothesen zijn. Voor de meeste commercieel verkrijgbare lichaamsbekrachtigde prothesen geldt dat de vereiste kabelkrachten te hoog zijn voor dagelijks gebruik. Om gebruikers in staat te stellen hun prothese de hele dag -en elke dag- vermoeidheidsvrij te gebruiken, en om goede terugkoppeling en besturing van het grijpmechanisme mogelijk te maken, zouden bedieningskrachten niet hoger moeten zijn dan 38 N voor de gemiddelde vrouwelijke gebruiker, en 66 N voor de gemiddelde mannelijke. Een grote aansturingsbeweging, en dus een grote kabel uitslag, draagt bij aan een verbeterde besturing van het grijpmechanisme. Voor de voorgestelde lage krachtniveaus biedt het Ipsilateraal Scapulair Huid Anker Systeem een goed alternatief voor het traditionele harnas.

CURRICULUM VITAE

Mona Hichert was born on October 17th, 1981 in Northeim, Germany. She earned her Abitur (university entrance diploma) at the Berufsbildenden Schulen I Northeim – Fachgymnasium Wirtschaft in 2001. During her school time Mona was playing Handball on top level and was part of the selection team of North Germany. In 2001 she started her studies in Mechanical Engineering at the Technische Universität Berlin and conducted internships at Alcan International Banbury Laboratory (United Kingdom) in 2001 and at Alcan Deutschland GmbH in Göttingen (Germany) in 2002 on aluminum testing and manufacturing procedures. In 2002 she moved to The Netherlands where, after learning the Dutch language, she continued her BSc in Mechanical Engineering at the Delft University of Technology. Besides her studies, she was training the student handball teams of Torius over the years, organizing numerous activities for the club and was Chairwomen in 2007-2008. For all this work she was awarded the first 'member of merit' title of Torius. In the scope of her MSc studies in Biomechanical Engineering she worked in Reykjavík in 2009 at Össur, an Icelandic company that develops and manufactures noninvasive orthopedics equipment. Here she worked on a project to "Reduce the necessity of human testing of Össurs Bionic products". In 2010 she was awarded her Master of Science degree in Mechanical Engineering from the Delft University of Technology with her thesis on perception and control of cable operation forces in body-powered prostheses. After a period of traveling she started her PhD project in 2011 on developing design requirements for body-powered upper-limb prostheses at the Delft University of Technology. In 2012 she participated in the Global Center of Excellence for Mechanical Systems Innovation (GMSI) Summer Camp in Tokyo/Shizuoka (Japan). In the same year she was Co-founder of the 'Handenstichting' or 'Hand Foundation', a foundation that aims to raise funds to support applied scientific research on upper-limb prosthetics. As part of her PhD-thesis work, she visited the University of New Brunswick in Fredericton (Canada) in 2014 were she built test equipment and conducted experiments on the users' abilities to control pinch forces with a body-powered upper-limb prosthesis. In 2014 she got married, gave birth to her son in 2015 and at the time of writing she is expecting her second child. In her free time she likes to practice sports (running, swimming, cycling, yoga), to be in the outdoors and to combine travelling with sports like sailing, diving, skiing, mountainbiking, and backpack hiking.

LIST OF PUBLICATIONS

Journal articles

Hichert M., Abbink D.A., Kyberd P.J., Plettenburg D.H., "High cable forces deteriorate pinch force control in voluntary closing body-powered prostheses". PLoS ONE. 2017;12: e0169996. doi:10.1371/journal.pone.0169996

Hichert M., Plettenburg D.H., "Ipsilateral Scapular Cutaneous Anchor System: an alternative for the harness in body-powered upper-limb prostheses". Prosthetics and Orthotics International (Accepted for publication)

Hichert M., Vardy A.N., Plettenburg D.H.; "Fatigue-free operation of most body-powered prostheses not feasible for majority of users with trans-radial deficiency". (submitted)

Hichert M., Abbink D.A., Vardy A.N., van der Sluis C.K., Janssen W., Brouwers M.A.H., Plettenburg D.H., "Perception and control of low cable operation forces in voluntary closing body-powered upper-limb prostheses". (submitted)

Conference articles

Plettenburg D.H., **Hichert M.**, Smit G., "Feedback in voluntary closing arm prostheses". Proceedings of the "Myo Electric Control Symposium – MEC '11, Raising the Standard", Fredericton, Canada, 2011, August 14-19, p.74-78.

Hichert M., Vardy A.N., Plettenburg D.H., "A need for a more user-centred design in body-powered prostheses". Proceedings of the "Myo Electric Control Symposium – MEC '14, Redefining the Norm", Fredericton, Canada, 2014, August 19-22, p.91-95.

Conference abstracts

Hichert M., Plettenburg D.H., Smit G., "Optimal control cable forces in shoulder controlled prosthesis operation". Proceedings of the "Trent International Prosthetic Symposium", Loughborough, United Kingdom, 2012, May 21-23, p.14.

Smit G., Plettenburg D.H., **Hichert M.**, "Mechanical performance of VO upper limb terminal devices: Do we really give our patients the best they can get?". Proceedings of the "Trent International Prosthetic Symposium", Loughborough, United Kingdom, 2012, May 21-23, p.27.

Smit G., Plettenburg D.H., **Hichert M.**, "Mechanical differences between a silicone and a pvc glove". Proceedings of the "Trent International Prosthetic Symposium", Loughborough, United Kingdom, 2012, May 21-23, p.46.

Hichert M., “Arm prosthesis control: perception of cable activation forces and displacements”. Proceedings of the “Global Center of Excellence for Mechanical Systems Innovation (GMSI) Summer Camp, M. Mitsuishi (ed.)”, Tokyo/Shizuoka, Japan, 2012, July 23-27, p. 52.

Hichert M. and Plettenburg D.H., “Body-powered arm prosthesis control: perception of cable forces and displacements”. Proceedings of the “4th Dutch Bio-Medical Engineering Conference”, Egmond aan Zee, The Netherlands, 2013, January 24-25, p.90.

Plettenburg D.H., Smit G., **Hichert M.**, “Voluntary closing prostheses – an update”. Proceedings of the “4th Dutch Bio-Medical Engineering Conference”, Egmond aan Zee, The Netherlands, 2013, January 24-25, p.49.

Hichert M., Plettenburg D.H., Smit G., “Force perception during Ipsilateral Scapular Cutaneous Anchor System versus figure-of-nine shoulder harness operation”. Proceedings of the “ISPO World Conference”, Hyderabad, India, 2013, February 4-7, p.194.

Hichert M., Valk M., Plettenburg D.H., Smit G., “Arm prosthesis control: perception of cable forces and displacements”. Proceedings of the “ISPO World Conference”, Hyderabad, India, 2013, February 4-7, p.233.

Hichert M., “Designing a physiological arm prosthesis control system” in the Symposium: Technology – ICS Code 17 – “Body powered prostheses: they are old school, right?”. Proceedings of the “ISPO World Conference”, Hyderabad, India, 2013, February 4-7, p.455.

Hichert M., van Mil E.H.A., Plettenburg D.H., “Body-powered prostheses: moving towards two-way control”. Proceedings of “OT World”, Leipzig, Germany, 2014, May 13-16, Abstract no. 5037.

Hichert M., Vardy A.N., Plettenburg D.H., “A need for a more user-centered design in body-powered prostheses”. Proceeding of “5th Dutch Bio-Medical Engineering Conference”, Egmond aan Zee, The Netherlands, 2015, January 22-23, p.163-164.

Hichert M., Kyberd P.J., Abbink D.A., Plettenburg D.H., “High cable operation forces deteriorate pinch force control in voluntary closing body-powered upper-limb prostheses”. Proceedings of “ISPO World Conference”, Lyon, France, 2015, June 22-25, p.216. (Nomination for best oral presentation)

Versluis E., **Hichert M.**, Plettenburg D.H., “Normative values for maximum voluntary applied grip and pinch forces in healthy subjects as a baseline for hand prosthesis design – a systematic review”. Proceedings of “ISPO World Conference”, Lyon, France, 2015, June 22-25, p.498.

Pogosian M., **Hichert M.**, Plettenburg D.H., “Hand opening and grip force – a literature study that investigates healthy humans ability to control hand opening and grip force with proprioceptive feedback”. Proceedings of “ISPO World Conference”, Lyon, France, 2015, June 22-25, p.554.

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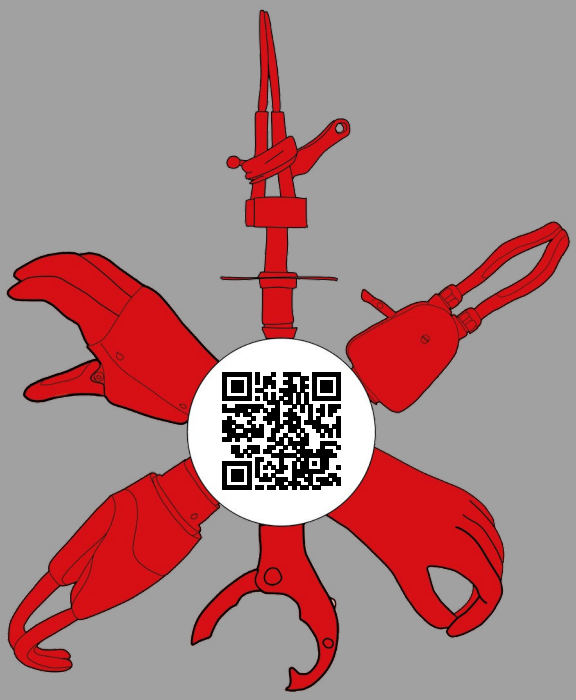

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