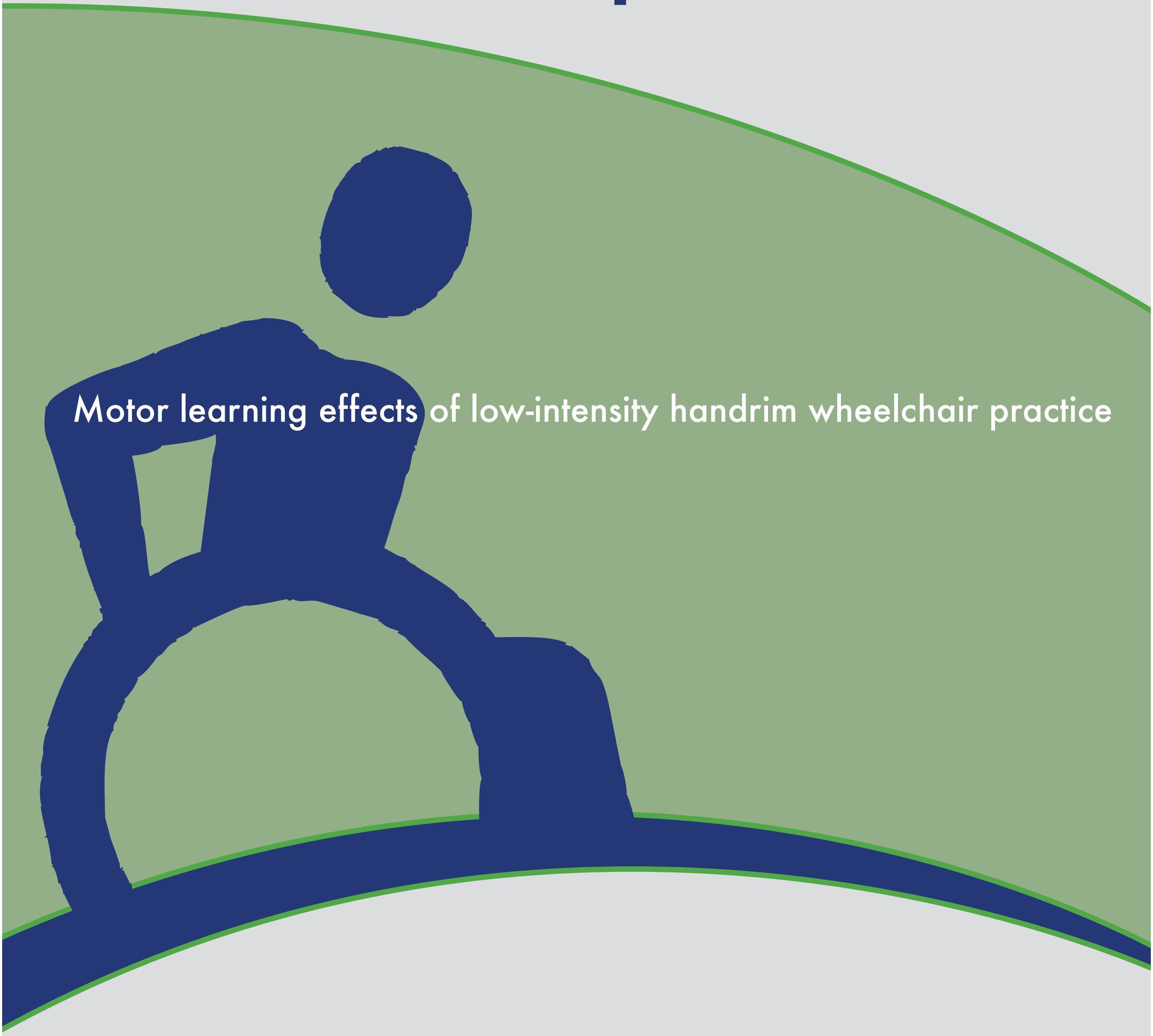
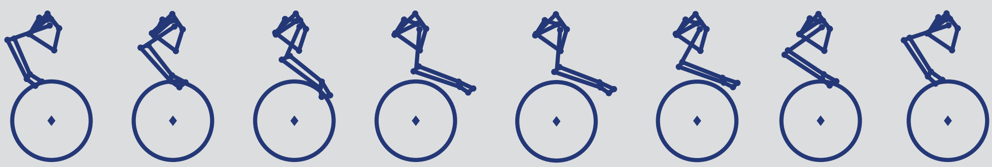
Wheelchair Skill Acquisition



Motor learning effects of low-intensity handrim wheelchair practice



Riemer JK Vegter

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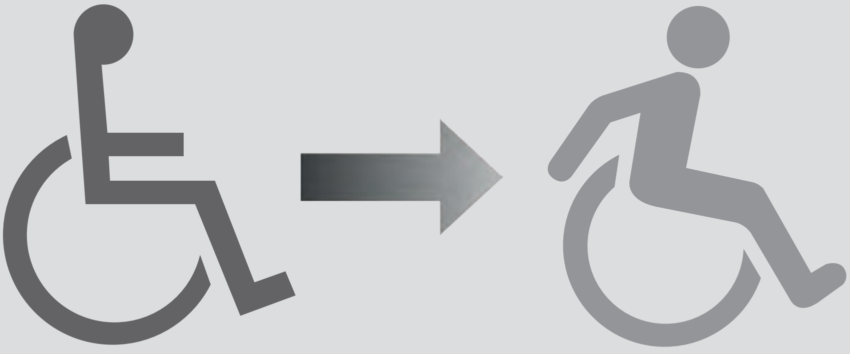
Contents

chapter 1

Introduction



Introduction

The process of skill acquisition is a key element of human functioning during daily life and an essential part of rehabilitation after disease or injury. In the process of rehabilitation people need to adapt to new situations constantly, like learning to walk with a leg-prosthesis or, as in the case of this thesis, learning to use the upper-body for ambulation while seated in a wheelchair. The research described in this thesis aims to increase our knowledge about the acquisition of wheelchair propulsion technique for the rehabilitation setting and to improve our understanding of natural motor learning processes.

1.1 Clinical relevance

Numerous persons are dependent on a wheelchair for their mobility. Worldwide it is estimated to be around 70 million people [1]. In the Netherlands approximately 82 percent of individuals with spinal cord injury (SCI), admitted for inpatient rehabilitation, are wheelchair users [2]. Being wheelchair dependent will cause limitations in functioning and impede functions of daily living, as is exemplified for persons with a spinal cord injury through the conceptual framework of the International Classification of Functioning, Disability and Health (ICF) (fig 1.1) [3]. The ICF model shows that activities and participation are influenced by the impairment, as well as by personal (internal) factors and environmental (external) factors. In the context of this framework, wheelchair mobility was shown to be of importance for participation, with peak aerobic wheelchair power output and wheelchair skill performance as significant predictors for return to work [4].

Manual wheelchair propulsion is a straining form of ambulation. Unlike the hip joint the function of the shoulder is more oriented towards joint mobility at the cost of a lower stability, requiring considerable muscular effort for stability control [5]. Furthermore, the upper body has a limited work capacity compared to the lower limbs. Because of the relatively smaller muscle-mass of the upper-body, that is also often untrained for prolonged physical activity, wheelchair propulsion can lead to high levels of relative mechanical strain during daily life (ADL) [6] and simultaneous low levels of mechanical efficiency [7]. Consequently, the continued mismatch between the high physical strain of daily wheelchair propulsion and the low physical capacity of wheelchair users increases the risk of serious secondary health problems in the long-term. Among the most important health problems are overuse injuries of the shoulder and wrist [8-11] that may express in chronic impairments and pain; shoulder complaints are estimated to be present among 50-70% of the wheelchair user population with a SCI after 10-15yrs of wheelchair use [12]. In addition, experiencing physical strain, fatigue and even (temporary) pain, will influence wheelchair use and daily activity patterns. Inactivity may develop, which in turn may deteriorate physical work capacity, and as such this downward spiral may in the long term introduce overweight, obesity and general health problems such as diabetes, metabolic syndrome and cardiovascular problems, but also other secondary impairments such as pressure sores and urinary tract infections [13,14].

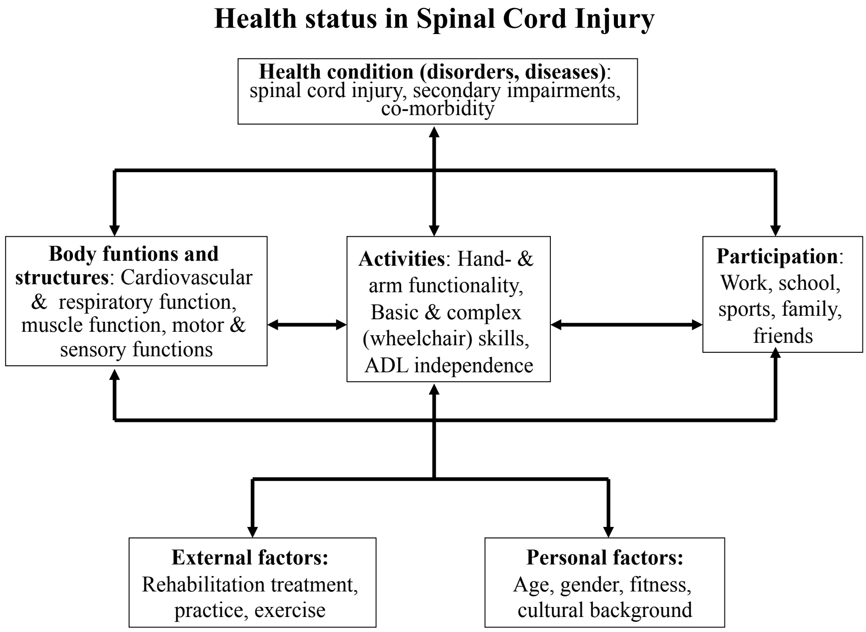


Figure 1.1: International Classification of Functioning, Disability and Health (ICF) [3], as applied for persons with a spinal cord injury [15].

Part of the strain of wheelchair propulsion might be associated with its low mechanical efficiency, which is the ratio of external power output (W) over internal energy expenditure (W) (Figure 1.2) [7,16,17]. During optimal conditions and dependent on power output, roughly less than 10% of the internally liberated energy actually goes into effective propulsion of the wheelchair. The mechanical efficiency is determined by multiple factors, among which the wheelchair, the wheelchair-user interface and the user characteristics. All these factors should be optimized to obtain a higher mechanical efficiency and subsequently better mobility and participation. In chapter 2 of this thesis, the most important aspects to improve the wheelchair and the wheelchair-user interface are explained. From there on the main focus of this thesis is on the wheelchair user.

At the level of the user, apart from aerobic and anaerobic wheelchair work capacity, the physical capacity is also dependent on task, skill and talent. The skill level of the wheelchair user is a critical performance aspect that can be trained, but so far has had only little attention in rehabilitation practice, adapted sports, or in experimental research for that matter. Therefore the present thesis will focus on the level of the user, especially to better understand how wheelchair skill (i.e. propulsion technique) is optimized during practice.

The observation that wheelchair users are able to increase their mechanical efficiency as a result of practice and skill acquisition has been established [18-23] but which skill changes underlie the increased mechanical efficiency or how these are acquired is not yet clear. Therefore it is hard to make evidence-based decisions about practice interventions during rehabilitation, or to give individual advice about the wheelchair propulsion skills of wheelchair users or athletes. To that end the research in this thesis aims to contribute to our understanding about motor learning processes in the context of manual wheelchair propulsion.

Measured at the wheels

$$ME(\%) = \frac{P_{output}}{P_{metabolic}} * 100$$

calculated from Cardio-respiratory measurements

Figure 1.2: Mechanical efficiency is the ratio of external power output (W) over internal energy expenditure (W). Mechanical efficiency (%) is used in the current thesis as the main outcome measure to evaluate the status and process of motor learning.

1.2 Natural motor learning

Studies on other cyclic tasks have also found that people tend to show increased mechanical efficiencies because of practice, without instruction or specific feedback. For instance learning studies using a ski-simulator or a rowing-ergometer showed a reduction of energy expenditure over two weeks of practice, while maintaining the same submaximal power output [24,25]. These findings are consistent with the constraints-based framework of metabolic energy expenditure, motor coordination and control that was proposed by Sparrow and Newell (figure 1.3) [26]. They suggested that the observed movement pattern emerges from the interaction between different constraints, with metabolic energy being the currency of the interaction. In other words, motor learning is the process of acquiring a movement pattern that minimizes the energy expenditure within the constraints that act on the task that needs to be performed. The increase in mechanical efficiency is expected to be the consequence of a change in task performance. The reduced energy cost in the above-mentioned cyclic skills of skiing and rowing coincided with an increase in movement amplitude and a decrease of movement frequency, described as a longer-slower movement pattern. Similar to those observations for wheelchair propulsion a change in wheelchair propulsion technique is expected to relate to an increased mechanical efficiency.

The constraints on handrim wheelchair propulsion can be defined within the constraints-based framework on the level of the organism, task and environment. First, the organismic constraints have to do with the user. For instance the physical characteristics of wheelchair users vary widely in control and function of the upper extremities and trunk-musculature, thus the most efficient movement pattern will be very dependent on the available body functions under control of the wheelchair user [27]. Besides such differences in physical capacity, there is also increasing evidence for individual differences in learning style and previous movement experiences [28], as well as talent [29-31]. Second, the task constraints can be defined on the level of the wheelchair-user interaction. Whether a movement pattern is efficient also depends on the interaction of the musculoskeletal system with the form and geometry of the propulsion mechanism, like the diameter and form of the handrim [32-34] or the seat and backrest configuration [35-37]. Finally, in daily life the environment imposes many different constraints. For instance, a cross-slope on a sidewalk will impose considerable impact on the emerging movement pattern, as will other pedestrians, or traffic lights for that matter.

Within the constraints on performance it is up to the wheelchair user to acquire an optimal movement pattern. One possible way of exploring the available motor solutions is found in the intrinsically variable way people perform a task. This intra-individual movement variability can for instance be found between limbs performing the same action (i.e. inter-limb variability), or in one limb repeating a cyclic movement over time (i.e. intra-limb variability), even in a very constraint environment (figure 1.4). Such variability is assumed to not only be the reflection of noise and/or error, but also to be functional and to contain informative features that may provide insight in motor learning [38-40] and/or pathological processes [41-45]. From this perspective the intra-individual variability allows the performer to subconsciously explore different motor solutions, facilitating the discovery and adoption of individualized optimal patterns of coordination, possibly reducing the energetic cost [26,46,47]. Therefore in this thesis the intra-individual variability of wheelchair propulsion is evaluated over practice (chapter 5) to study its relation to the motor learning process of the wheelchair user.

Natural motor learning in the current thesis is defined as the changes in task proficiency as a consequence of low-intensity steady state handrim wheelchair propulsion on a motor driven treadmill without any verbal or additional visual feedback.

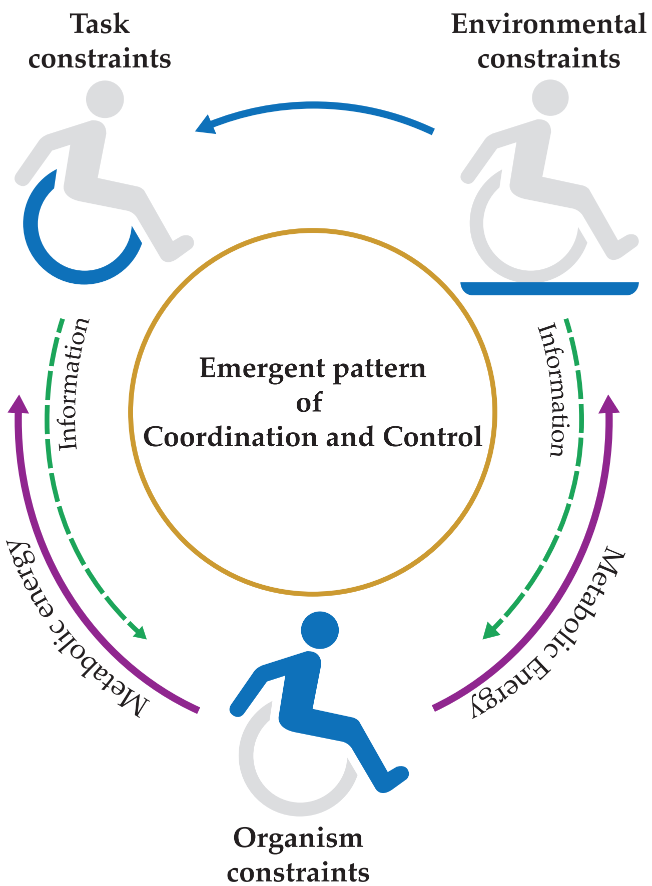


Figure 1.3: Natural motor learning: A constraints-based framework of metabolic energy expenditure and motor coordination and control (adapted from Sparrow & Newell (1998) [26]).

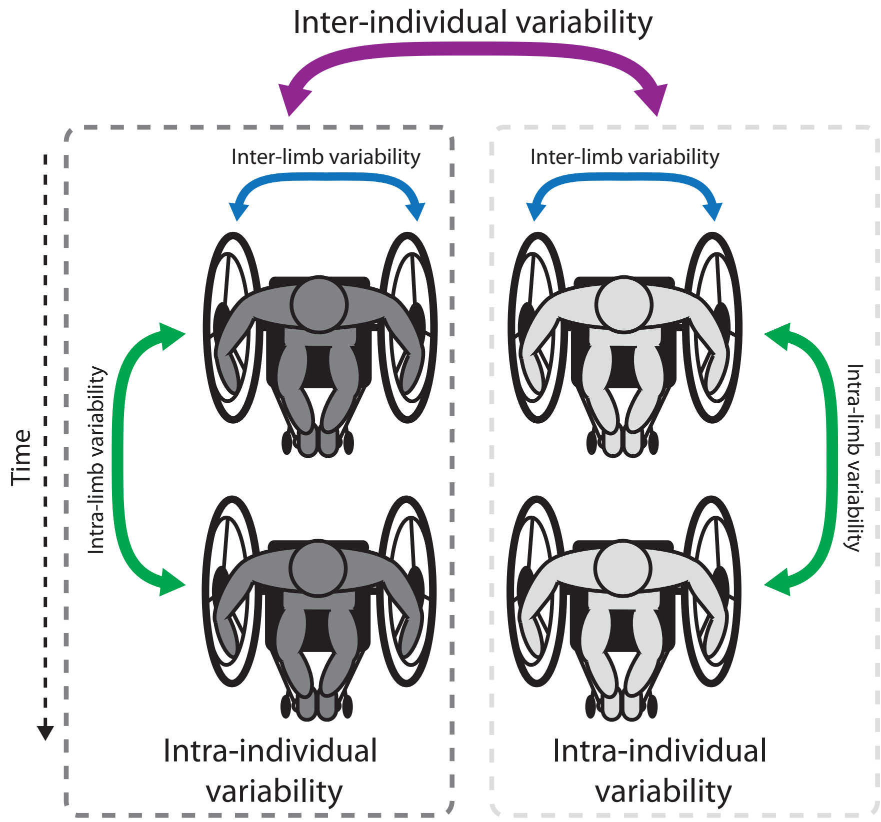


Figure 1.4: Wheelchair propulsion – even on a motor driven treadmill at a constant submaximal speed - is a cyclic skill that varies within and between participants.

1.3 Propulsion technique

A handrim propelled wheelchair generally has two large rear wheels equipped with a fixed handrim. Speed and direction of locomotion is determined by bi-manually application of propulsion forces by the user. The movement pattern of wheelchair propulsion consists of two distinct phases, the push phase and the recovery phase (Figure 1.5) [48]. In the push phase the hands are in contact with the handrim and power is transferred from the upper body to the wheels. This energy transfer phase can be subdivided in three subphases, initial contact, propulsion and release. At initial contact the hands must connect to the handrims, while matching the speed of the already rotating handrim outside the visual field. After the connection is made, the forces and moments applied to the handrims accelerate the wheelchair-user combination. Finally, at the end of handrim contact the joints reach their final positions and the hands must release the handrim and start the recovery phase. During the recovery both upper-limbs and possibly the head and trunk have to travel back to the starting position and the cycle repeats itself [49].

1.4 Measurement wheels

The experiments in this thesis make use of instrumented wheels on a motor-driven treadmill (figures 1.4 & 1.6), which makes it possible to study changes in wheelchair

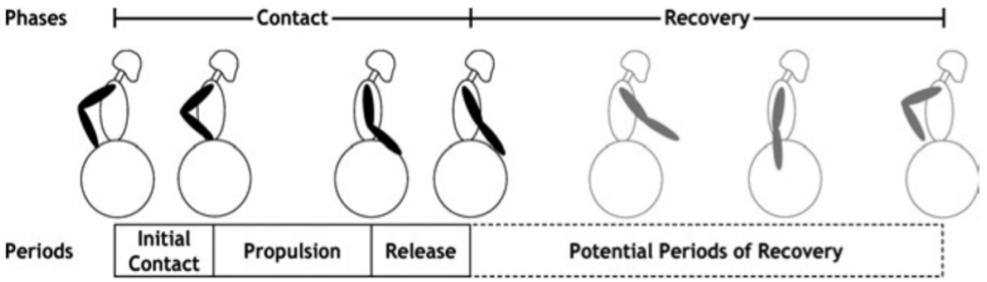


Figure 1.5: The stroke cycle. The stroke is divided into the contact phase and the recovery phase; and the contact phase is subdivided into the initial contact, propulsion, and release periods. (Figure from Kwarciak et al (2009) [48])

propulsion technique under controlled conditions. When using the term 'propulsion technique' in the present thesis, it comprises timing variables (e.g. cycle frequency and cycle time) and force application variables (e.g. direction and magnitude of the applied force vector) together with the inter-cycle variability of these measures (i.e. how similar the subsequent pushes are). Figure 1.7 exemplifies the data collection and some of the derived propulsion technique variables from these instrumented handrim wheel data.

1.5 Upper-extremity load

Although mechanical efficiency is used as the primary measure to evaluate motor learning, the local strain on the shoulder complex is also very important. It has been theorized that a change in propulsion technique and an increased efficiency reduces some of the mechanical strain on the wheelchair-user. However, it is currently not clear in what way these changes in propulsion technique impact the load on the shoulder complex. To evaluate the load on the shoulder complex over the stroke cycle, external forces, and upper-body kinematics can be used as input for a musculoskeletal model to estimate muscle activity and joint reaction forces. For experienced wheelchair users the Delft Shoulder and Elbow Model [50] estimated peak glenohumeral reaction forces between 300 to 1400N during each push cycle at speeds between 0.4 and 1.5 m.s-1, with concomitant high relative forces of the rotator cuff muscles, especially of the subscapu-

Figure 1.6: Smartwheel[a] (left) and Optipush[b] (right). Both wheels measure three-dimensional forces and torques on the handrim, combined with the angle under which the wheel rotates.

laris and infraspinatus muscles [51-54]. When taking into account that wheeling an hour a day with a typical push frequency of 45 pushes per minute may already add up to some 2700 repetitions, the associated load on the shoulder complex might be considered a risk factor for overuse injury of the rotator cuff [55] and shoulder complex in general. Yet, these analyses are mostly of a cross-sectional nature and not much is known about the change in the loading of the shoulder complex over practice time. To understand in what way motor learning processes affect the load on the shoulder complex, it is important to investigate whether motor learning-associated changes in mechanical efficiency and propulsion technique are also related to a reduction of the muscle forces and joint reaction forces of the shoulder complex.

1.6 Able-bodied participants

Wheelchair propulsion is a skill that is new to persons who just lost their walking ability and to many able-bodied participants as well. Therefore, in the study of motor learning able-bodied participants can serve as a model to study the early acquisition of this skill, without being too heterogeneous as a group, because of for instance spinal cord lesion level or upper-body asymmetries and without being hindered by the recent trauma early in rehabilitation. This way we'll get a better understanding about the principles underlying the skill acquisition of wheelchair propulsion so that we can develop better interventions that promote the skill proficiency of wheelchair users that might protect them from overuse injury.

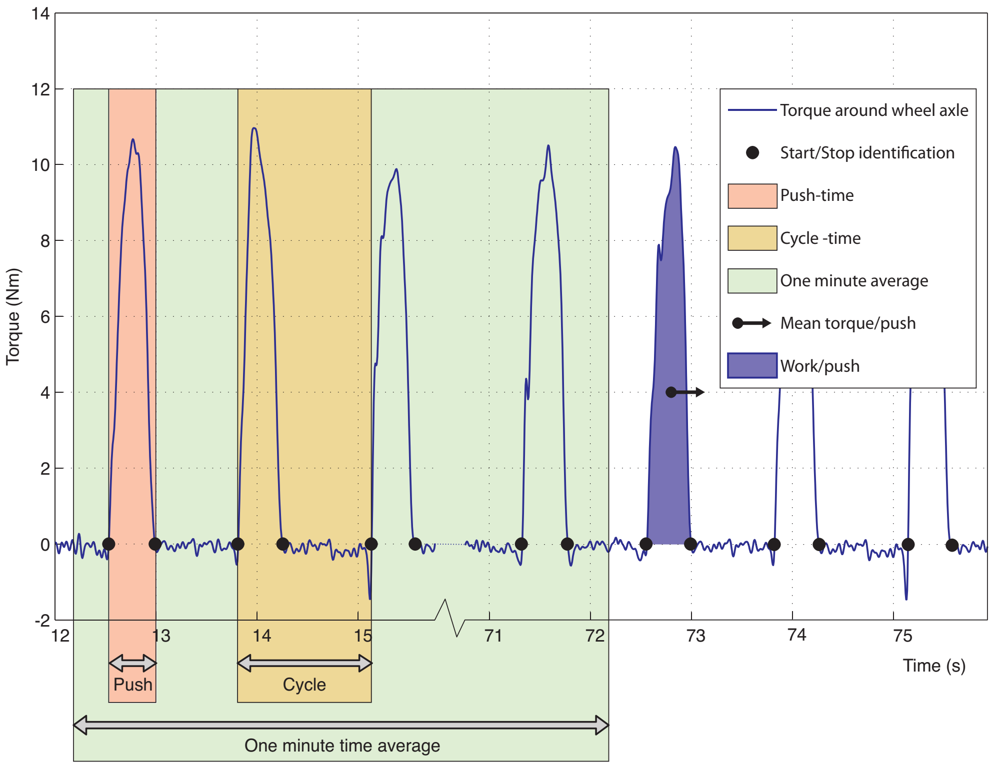


Figure 1.7: Definitions of handrim push variables. Push identification, push-time, cycle-time, work per push, and mean torque. Variables are calculated per push or over all full stroke cycles within one minute.

1.7 Aim of this thesis

The aim of this thesis is to better understand the natural motor learning processes underlying the skill acquisition of submaximal steady state handrim wheelchair propulsion on a motor driven treadmill in novice able-bodied individuals. To monitor improvements in task performance over time mechanical efficiency is used as the primary outcome measure. The biomechanical measures from instrumented wheels are used to study the propulsion technique changes underlying the improved mechanical efficiency and are in chapter 6 used in combination with three-dimensional kinematics and a musculoskeletal model to gain insight in changes in the strain on the shoulder complex.

1.8 Thesis outline

Chapter 2 gives an overview of the different aspects that influence mechanical efficiency, propulsion technique and shoulder load of wheelchair propulsion, with the main focus on the importance of optimizing the wheelchair and the wheelchair-user interface. The subsequent chapters focus on the user level; these experimental studies add to our understanding of motor learning and wheelchair propulsion in different ways. **Chapter 3** introduces our most important research tools, the measurement wheels capable of measuring forces and torques applied to the handrim during wheelchair propulsion. By comparing two commercially available wheels both their reliability and the variability of human coordination are studied. **Chapters 4 and 5** are directly focused on the motor leaning process, as is instigated through practice over different time scales. **Chapter 4** goes into the relation of mechanical efficiency and propulsion technique and their change as a consequence of the initial early (12min) practice period. Interestingly, individual motor learning differences were found in this fourth chapter. **Chapter 5** evaluates these individual differences over a longer practice time (80min). **Chapter 6** focuses on motor learning in relation to the load on the shoulder complex using inverse dynamics combined with a musculoskeletal model (Delft Shoulder and Elbow Model). **Chapter 7** transfers some of the research methods presented in this thesis to clinical practice and gives some initial norm values and studies the smallest detectable differences for individual application. **Chapter 8** serves as the general discussion and as a future outlook on research and clinical application.

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chapter 2

Design of a manually propelled wheelchair: optimizing a wheelchair-user combination.

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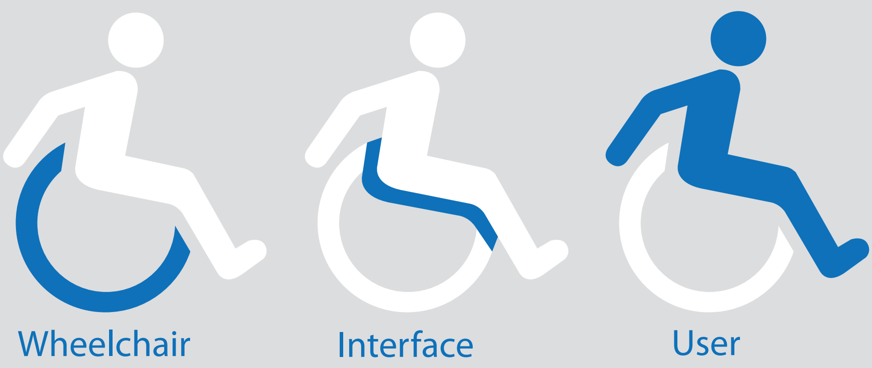
Hettinga FJ,

Veeger HEJ

van der Woude LHV

International Encyclopedia of Rehabilitation. 2010



Abstract

Being wheelchair dependent may cause limitations in functioning, i.e. impede functions of daily living (ADL) and/or participation, as is exemplified in the International Classification of Functioning (ICF). To improve ambulation, the wheelchair-user combination can be optimized at three levels. On the level of the user one can optimize physical capacity and technique by training. The second level focuses on the wheelchair-user interface, i.e. the interaction between the human system and the geometry of both the seating orientation and propulsion mechanism, aiming for a higher efficiency. This is operationalized as a better ratio of internal power from the user to the external power required for propulsion. Finally, at the level of the wheelchair the focus lies on minimizing power loss of the wheelchair-user system by reducing frictional forces and optimizing the vehicle mechanics. To advance wheelchair design, better insight in the working mechanisms of our biological system in combination with a mechanical extension, such as the wheelchair, is necessary. Changes to the design need to be theory driven and the research evaluating those changes is of great importance and needs to be available. Ongoing development of research and research methods allows for better insight and better design as well as wheelchair prescription today.

Introduction

Numerous persons are dependent on a manually propelled wheelchair for their mobility. In the Netherlands for instance approximately 82 percent of individuals with spinal cord injury (SCI), admitted for inpatient rehabilitation, are wheelchair users [1]. Being dependent on a wheelchair for mobility may cause limitations in functioning, i.e. impede functions of daily living (ADL) and/or participation, as is exemplified in the International Classification of Functioning (ICF) [2]. In figure 2.1 the ICF model of human functioning and disability is presented, adapted to persons with SCI. The model shows that activities and participation are influenced by the impairment, and by both personal (internal) factors and environmental (external) factors. In this model the use of assistive technology, such as a wheelchair, is part of the external factors influencing mobility and more important functioning and participation of the wheelchair user. A recent study showed wheelchair mobility to be of influence on participation: peak aerobic power output, and wheelchair skill performance were significant predictors of return to work for a group of wheelchair dependent persons with SCI one year after discharge of inpatient rehabilitation [3].

Despite its importance manual wheelchair propulsion is a straining form of ambulation. Unlike the hip joint the function of the shoulder complex is more oriented towards joint mobility at the cost of a lower stability, requiring considerable muscular effort for stability control [4]. Furthermore the upper body has limited work capacity, due to a small (untrained) muscle mass, which leads to high levels of relative mechanical and cardio-respiratory strain during daily life (ADL) [5]. The long-term consequence of a continued mismatch between the stress of daily wheeling, the physical strain and the

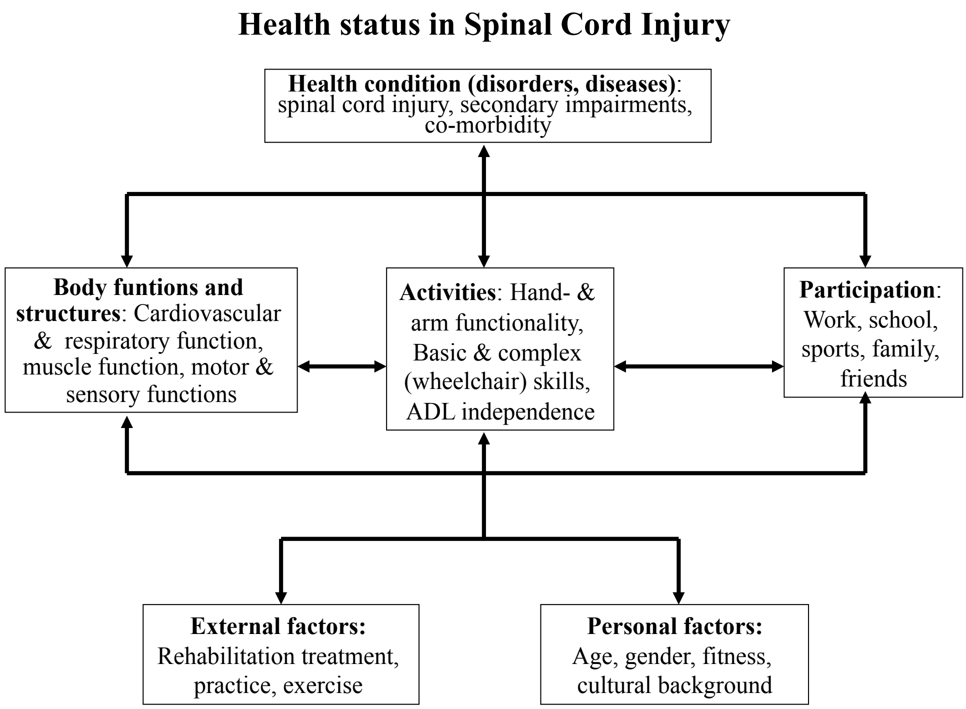


Figure 2.1: International Classification of Functioning, Disability and Health (ICF) [2],

overall physical wheelchair capacity is the chance of serious secondary health problems. Most important are overuse problems in the upper extremities [6-9] that may lead to chronic impairments and pain. Upper extremity overuse problems are estimated to be present after 10-15yrs of wheelchair use among 50-70% of the wheelchair user population with a SCI [10]. In addition, experiencing physical strain, fatigue and even (temporary) pain, will influence wheelchair use and daily activity patterns. Inactivity may emerge, which in turn may deteriorate physical work capacity, and turn into a downward spiral of deconditioning. This may in the long term introduce overweight and obesity, and general chronic health problems such as diabetes, metabolic syndrome and cardiovascular problems, but also secondary impairments such as decubitus and urinary tract infections [11,12].

Part of the physical strain of wheelchair use on the user is caused by the low efficiency of handrim propulsion [13-16]. Only 6-11% of the internally liberated energy actually goes to effective propulsion of the wheelchair. Well-trained wheelchair athletes may get efficiencies up to 18% in a racing wheelchair [17]. However, these values are still lower compared to lower extremity activities such as cycling (20-25%) [18]. To improve efficiency, alterations in wheelchair design and set-up are imperative and one has to deal with the user-wheelchair combination as a whole, as well as on its different components Therefore, the HAAT model [19] , which describes the interaction between human (H), activity (A) and assistive technology (AT)in the environment is instructive. Thus optimizing the wheelchair-user combination must be achieved by adjusting any or a combination of the following elements: the user, the wheelchair-user interface and the wheelchair (figure 2.2). This article will give a short overview of the practical implications of the research developments regarding these aspects over the last decades and will take a look forward at the research requirements for further design improvements in the near future.

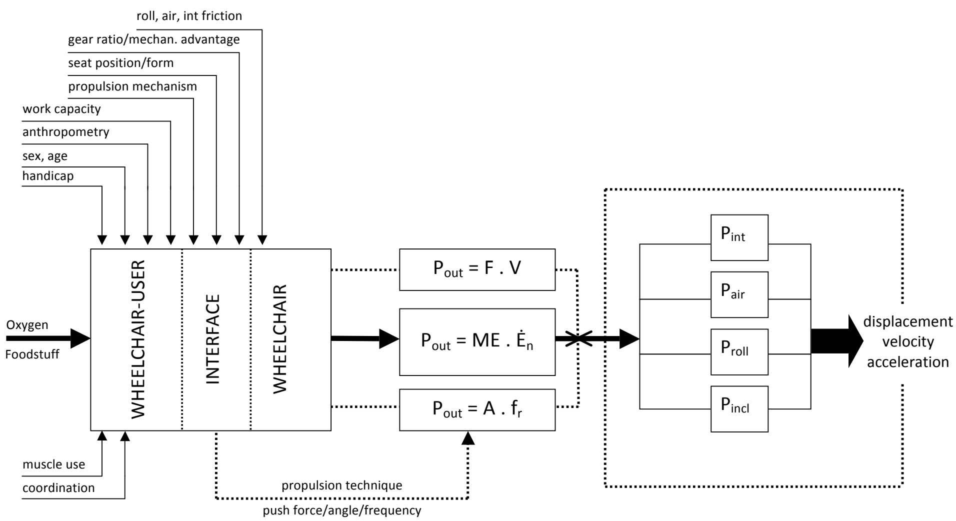


Figure 2.2: Factors influencing the power balance (van der Woude and others 1986)

2.1 User

There is not one ultimate design in general, but always one adjusted to the user. It is important to understand the possibilities and needs of the user to adequately design and fit the correct wheelchair. Physical characteristics such as upper-extremity muscle force and peak power output have been shown to closely relate with wheelchair skill performance [20]. The physical capacity varies widely between users and these inter-individual differences impact the design. Knowing the peak power output of a person can give a good indication of how intensive propelling a wheelchair for him/her is. A recent study of Haisma et al. [21] reports on the large differences in peak power output, measured in a wheelchair, of persons with tetraplegia (mean: 26 W) and paraplegia (mean: 74 W) during inpatient rehabilitation. If we combine this knowledge with research of Van der Woude et al. [22] on wheelchair rolling resistance of different floor surfaces we can get some insight in the clinical impact of these values. For instance vinyl as found in a gym of a rehabilitation center had a rolling resistance for a certain wheelchair of ca. 20 N. At a speed of 4 km/h this takes a power output of 22 W, which is 85% of the peak power output for the person with tetraplegia, while only 30% for persons with paraplegia. This example stresses the need for a different view on design for different user groups. Besides design, training physical capacity of the user has proven its worth and is of great importance to improve mobility [23]. A higher peak power output will reduce the relative strain of daily wheeling. In this perspective a physically active lifestyle is important since 'exercise is medicine' (http://www.exerciseismedicine.org).

Another important user-related aspect is that of motor skill learning, since rehabilitation involves learning the new task of propelling the wheelchair. Propelling the wheelchair takes more than being physically able to exert a certain amount of power; correct coordination patterns will for a larger part determine functioning of any wheelchair-user combination and is expressed in a higher mechanical efficiency. Positive effects of low-intensity training were shown on mechanical efficiency, metabolic cost and the propulsion technique of inexperienced able-bodied participants [13,24]. This shows that even without adaptations to the interface or wheelchair it is still possible to obtain a higher efficiency with training. To optimize training, knowledge about motor learning and insight in proper propulsion technique is of the utmost importance. For instance only recently it was shown that training a person to push as tangentially to the circle of the handrim as possible does not improve efficiency, as has been speculated in the past [25,26].

2.2 Wheelchair-user interface

Power production is not only determined by the user but also depends on the interaction of the human system with the form and geometry of the propulsion mechanism and the seat configuration; therefore design of a proper interface is of importance [27]. Experiments focusing on the interface have proven the possible role of optimization of interfacing on propulsion technique, efficiency and peak power output. Apart from different propulsion mechanisms (levers, cranks etc), possible design venues are handrim characteristics and seat-position. Both aspects will briefly be discussed here, together with a short sidestep to alternative propulsion mechanisms.

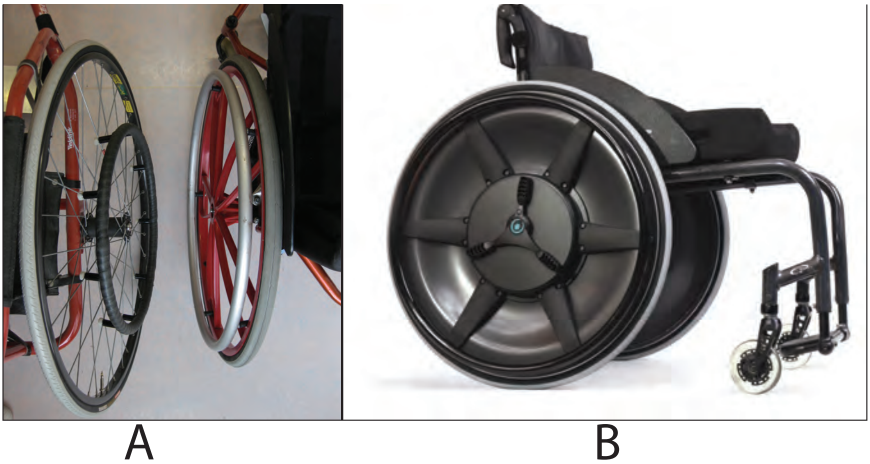


Figure 2.3: Different gearing and rim sizes (a) athletic and normal handrim (b) 2-geared wheel.

2.2.1 Handrim

Considering the interface the first thing that comes to mind is the handrim, since this is where the coupling between the biological system and the wheelchair takes place. Regarding the handrim there are several design options; the handrim radius, the angle under which it is placed (camber) and the diameter, surface material and shape of the tube. An additional aspect to be considered is the use of gloves, changing the handrim-hand interface. It is a common feature in certain wheelchair sports (i.e. quadrugby) and may help optimizing functioning [28].

The hand rim radius is in fact a gearing level [16,29]. Smaller hand rims (figure 2.3a) will require a larger propulsion force and lower hand velocity at a given traveling speed. Different task conditions will require different radii i.e gearing levels: groups of well-trained subjects may want a gearing which enables them to compete at high velocities, whereas a steep incline for physically less able subjects will demand a low gear. A study on the use of a handrim with two gears incorporated in the wheel (figure 2.3b) found pain reductions already 2 weeks after the participants started using the device, indicating the potential for shoulder pain reduction by the use of different gears [30]. In line with these findings the use of an electromotor in the wheel hub, which only aids in the propulsion when the user exerts force on the handrim, can be of use. Such hand-rim activated power assisted wheelchairs aid the user who can no longer cope with the strain on the upper limbs during manual wheelchair propulsion. Compared to standard handrim propulsion such a device lowers the overall cardiorespiratory strain () and shoulder muscle activity [31], yet it still allows the user to be an active participant of society.

The second design possibility does not solely relate to the rim, but to the wheel as a whole, camber in the rear wheels places the handrim under a certain angle to the user. The positive effect of camber on stability of the wheelchair user system is relevant [32]. Furthermore the hands are protected when passing along objects because the width at the base of the wheelchair is larger than at the height of the hands [33]. Besides the positive effects on stability, hand comfort on the push rims and maneuverability, changes of camber do not seem to be associated with changes in efficiency of wheelchair propulsion [34].

Beside the radius of the rim the diameter of the tube, its shape and the use of different materials is of influence. A conventional handrim tube is a 19 mm diameter circle; new alternatives have been developed to make a better interface between hand and wheel. Modifications have to do with the shape of the tube and the attachment to the wheel (figure 2.4). A recent study between four different handrims with respect to shape and material in able-bodied subjects did not find significant effects on any of the physiological parameters and force application characteristics [35]. On the other hand two other specific handrim designs were found to be beneficial and are currently commercially available. First the Flexrim, a flexible handrim (figure 2.4a) that allows some freedom of movement between handrim and wheel showed significant reductions in both peak and total forearm muscle activation. The flexible handrim required less finger and wrist flexor activity than a standard uncoated handrim for the same propulsion conditions [36].

Second the use of the Natural-Fit contoured handrim (figure 2.4b) was surveyed among its users. This handrim has a larger oval shape than the conventional 19 mm circle and uses different materials for the place of the thumb and finger. The majority of participants reported improvements in upper-extremity symptoms, ease of wheelchair propulsion, and functional status [37]. The possible benefits of a larger oval shape as in the Natural-fit handrim were also reported earlier by Van der Linden et al [38].

A B

Figure 2.4: Different tube shapes of the handrim, (a) flexrim (b) natural-fit

2.2.2 Seat height and position in the chair

Besides the connection between hand and wheel the body position with respect to the wheel axle will be of influence on propulsion. Van der Woude et al. [39] operationalized seat height as the elbow angle while sitting in the wheelchair with hand on top dead centre of the wheel. In their research they found a tendency for mechanical efficiency and mechanical strain to optimize seat height at an elbow angle of 100–130° in persons with a spinal cord injury during rehabilitation.

Cowan et al [40] used this angle to maintain seat height and study two different horizontal axle positions. Their findings suggest more anterior axle positions to be beneficial in reducing peak resultant forces exerted by the hand on the rim. Yet this position of the axle leads to a lower rolling resistance confounding possible interface effects. Another investigation on the relation of wrist kinematics and horizontal position did not find effect on wrist kinematics for horizontal displacement [41]. Kotajarvi et al [42] studied both vertical and horizontal changes of the wheel axes, they only found vertical changes e.g. changes in seat height to be of influence. It can be concluded that the effect of seat height is clear, while horizontal changes of the axle still need to be better understood. Furthermore changes in seat to backrest angle and system tilt angle were analyzed [43] but were found not to be of influence in shoulder average and peak moments, and can be optimized towards other goals such as comfort and pressure modulation.

A B C

Figure 2.5: Alternative propulsion mechanisms (a) handcycle (b) lever propulsed tricycle (c) hubcrank wheel.

2.2.3 Other propulsion mechanisms

Handbikes and other forms of propulsion, like hubcranking or lever mechanisms (figure 2.5), are found to have better efficiencies than handrim propulsion [44]. Overall the limited muscle mass and function of the upper body are more cautiously and effectively used in levers or cranks as opposed to the handrim. Especially handbikes are a good alternative for outdoor wheeling, sports and recreation, given their higher efficiency [45]. Certain drawbacks like maneuverability, and added width and weight make these devices especially useful outdoors. These new developments have an effect on the use of the handrim wheelchair. Through the years the use of the handrim wheelchair has proven itself and as such shall probably not be replaced for most ADL tasks. Apart from fine-tuning one needs to consider the task specific use of handrim wheelchairs as is typically done in different sports disciplines. The handrim-propelled wheelchair can be seen as the walking means of the wheelchair user and should be optimized towards that goal. This shift in use should be reflected in the design.

2.3 Wheelchair

If a wheelchair is kept at a constant speed, the wheelchair user has to produce a certain amount of energy per unit time, or power. With each push work is produced and the product of work and push frequency equals the mean power produced by the user. This so-called external power is produced by the user, but requires a much higher amount of internal power. The external power output is necessary to overcome energy losses in the system and environment. The wheelchair–user combination will lose energy in the form of rolling resistance, air resistance and internal resistance in the mechanical structures of the chair. When more external power is produced than needed to overcome these losses, the chair will accelerate. In the following the different contributors to power loss will briefly be discussed, since they will determine for the larger part the experienced strain of wheeling. The main message is that these external forces must be minimized through wheelchair design improvements and maintenance, as well as through environmental changes.

2.3.1 Rolling resistance

The magnitude of the friction is related to the amount of deformation of tire and floor surface [22]. This deformation dissipates energy [46]. Deformation is dependent on tire pressure, tread and profile, wheel diameter, but also on wheel alignment, mass of the wheelchair and the user, and of course the surface on which one wheels.

2.3.2 Air resistance

The second contributor to the frictional forces is air resistance. At high velocity like in wheelchair racing this factor is by far the most important source of energy loss. Air resistance is dependent on the drag coefficient, frontal plane area, air density and velocity of the airflow relative to the object. Air resistance will be of minor importance at low speeds, but at high speeds and/or wind velocities air resistance will be the most important source of resistance. Following Abel and Frank [47], at slow speed (3.6 km/h) air drag will be below 1 N, while at 18 km/h the drag force due to air resistance is ±14 N, which implies an average power output of 70W for wind resistance only at that wheelchair speed. It is obvious that the frontal plane area is dependent on the posture of the user. Although a wind tunnel experiment has been performed [48] as well as empirical measurements [49,50], no recent figures on air resistance have been published in association with contemporary wheelchair sitting posture and propulsion technique. However, from hand cycling or speed skating many new developments were transferred to wheelchair racing. Next to frontal area reduction, adaptation of the seat position and orientation of the segments of the body, and the application of skin suits will influence the drag coefficient.

2.3.3 Internal friction

Energy losses within the wheelchair are caused by bearing friction around the wheel axles and in the wheel suspension of the castor wheels and possibly by the deformation of the frame in folding wheelchairs during the force exertion in the push phase. Bearing friction generally is very small, and given that the hubs have annular bearings and are well maintained and lubricated, this friction coefficient will not exceed 0 [51]. However, the losses in ill-maintained bearings can be considerable.

An unknown aspect of internal energy dissipation is the loss of propulsion energy due to deformation of the frame elements. This will clearly be possible in folding wheelchairs, but has not been addressed empirically. The use of levers and cranks does introduce a chain, chain wheel and gearbox related friction. Whitt and Wilson [52] indicate a possible loss of energy of 1.5% in chain transmission.

2.3.4 Slope and acceleration

Although body mass and wheelchair mass have a small effect on rolling resistance, they have a considerable effect on the slope component and the acceleration component. Acceleration potential is inversely related to total mass at a given power output (acceleration will be slower when the mass of the system is larger). Also, mass is linearly related to power output in climbing. Of course, this extra investment will be returned partially during descents, but will still lead to higher losses. Wheelchair mass can be influenced through proper technology and lightweight materials. Important to realize though is that the major attributor to the total mass is the mass of the user. On the other hand, besides reducing mass for propulsion purposes it is also good to note the amount of times a wheelchair needs to be picked up, to for instance get in a car, here reduced mass of the wheelchair will be of great benefit.

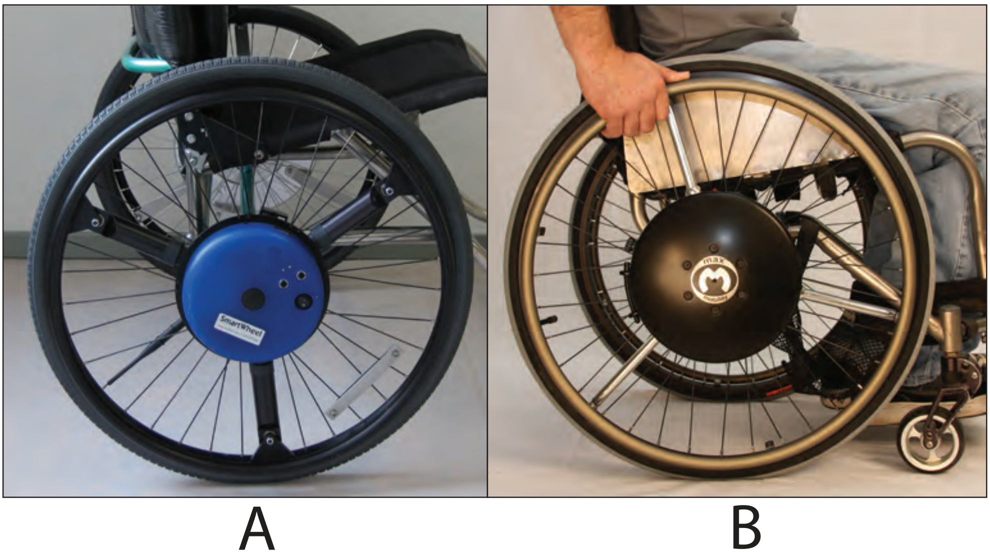


Figure 2.6: Two ambulant 3D force measurement devices, (a) the SmartWheel (b) the Optipush.

2.4 Evaluation of wheelchair design

Cooper [53] stated: "The greatest engineering challenges in manual wheelchair design are optimizing interaction between the user and the wheelchair, which requires knowledge of materials, biomechanics, ergonomics, anthropometrics and human physiology, as well as motor learning to train the user in the skills necessary to achieve maximum mobility." If one is to gain knowledge of the above-named fields research possibilities need to be available. Developments of research tools have made it possible to advance in experimental setup towards the use of a manual wheelchair. For instance an instrumented wheel (SMARTwheel or Optipush) (figure 2.6) that measures three dimensional forces and moments, together with the angle of the wheel [54]. Knowledge of these forces combined with position registration of important anatomical positions can be used as input for an inverse-dynamic model that calculates net moments around the human joints, as well as internal load, using a detailed shoulder-arm model [55]. Furthermore the measurement of oxygen uptake is used to estimate metabolic power and thus calculate efficiency. Electromyography (EMG) can give more insight in muscle activation, like identifying co-contractions. Important note is that technological developments can only attribute to design if they are used to address the proper questions. The link between technology and design needs to be made by proper theory-driven research to give insight in new design venues towards better functioning.

Conclusion

Although handrim wheelchair propulsion has been the focus of quite some research it is not yet fully understood, therefore design cannot yet be fully optimized. Major issues maintain concerning efficiency and shoulder overuse. In the future development in materials and production technique can help in improving the design. Moreover advances in research methods to get insight in the working of our biological system in combination with assistive technology might provide better solutions to optimize mobility in terms of efficiency and preservation of the physical capacity of the user over time.

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chapter 3

Variability in bimanual wheelchair propulsion: consistency of two instrumented wheels during handrim wheelchair propulsion on a motor driven treadmill.

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Abstract

Background:

Handrim wheelchair propulsion is a complex bimanual motor task. The bimanually applied forces on the rims determine the speed and direction of locomotion. Measurements of forces and torques on the handrim are important to study status and change of propulsion technique (and consequently mechanical strain) due to processes of learning, training or the wheelchair configuration. The purpose of this study was to compare the simultaneous outcomes of two different measurement-wheels attached to the different sides of the wheelchair, to determine measurement consistency within and between these wheels given the expected inter- and intra-limb variability as a consequence of motor control.

Methods:

Nine able-bodied subjects received a three-week low-intensity handrim wheelchair practice intervention. They then performed three four-minute trials of wheelchair propulsion in an instrumented hand rim wheelchair on a motor-driven treadmill at a fixed belt speed. The two measurement-wheels on each side of the wheelchair measured forces and torques of one of the two upper limbs, which simultaneously perform the push action over time. The resulting data were compared as direct output using cross-correlation on the torque around the wheel-axle. Calculated push characteristics such as power production and speed were compared using an intra-class correlation.

Results:

Measured torque around the wheel axle of the two measurement-wheels had a high average cross-correlation of 0.98 (std=0.01). Unilateral mean power output over a minute was found to have an intra-class correlation of 0.89 between the wheels. Although the difference over the pushes between left and right power output had a high variability, the mean difference between the measurement-wheels was low at 0.03 W (std=1.60). Other push characteristics showed even higher ICC's (>0.9).

Conclusions:

A good agreement between both measurement-wheels was found at the level of the power output. This indicates a high comparability of the measurement-wheels for the different propulsion parameters. Data from both wheels seem suitable to be used together or interchangeably in experiments on motor control and wheelchair propulsion performance. A high variability in forces and timing between the left and right side were found during the execution of this bimanual task , reflecting the human motor control process.

Introduction.

Handrim wheelchair propulsion is the means of ambulation for a large group of people with a disability. However, handrim wheelchair propulsion is straining and (overuse) injuries to the upper extremities, e.g. shoulder pain or carpal tunnel syndrome, among wheelchair-dependent persons are common [1-3]. Therefore, a better understanding of wheelchair skill, physical capacity and the impact of wheelchair mechanics and fitting on performance are important [4-6]. Research over the past 30 years has led to a number of studies on the physiology and biomechanics of wheeled mobility [7, 8]. Due to the complexity of instrumentation this research was primarily lab-based. Only more recently ambulant instrumentation for both physiological and biomechanical outcomes became available, which today even evolved into commercially available clinical tools [9, 10].

Measurements of forces and torques on the handrim of a wheelchair are important to study change of propulsion technique due to learning, training or the effect of changes to the wheelchair. From a scientific point of view this provides a deeper understanding of the universal principles regarding the motor control of wheelchair propulsion, while from a clinical perspective it can help to better tailor the properties of a wheelchair to a patients needs' and develop intervention protocols with respect to propulsion technique and strategy [11]. Over time, different studies have used different ways to instrument the wheels to gain insight in the forces and timing involved in wheelchair propulsion, varying from instrumented ergometers to specialized wheels [9, 10, 12-16]. These measurement systems have been used to describe unilaterally the cyclic nature of handrim propulsion analogous to gait analysis. For example, frequency of pushes, peak forces and torques and the wheel angle covered within a push have been used to describe the motor learning process of novel wheelchair users [17]. Besides propulsion technique these wheels are able to measure the power output of the wheelchair-user combination, making it possible to calculate mechanical efficiency when combined with cardio-respiratory measurements and energy calculations [18].

Most studies on propulsion technique measured this essentially bimanual propulsion task unilaterally and focused on the description of propulsion characteristics in dependence of a variety of different interventions. Yet, due to both internal control processes and external perturbations interlimb variation is expected [19]. Studying unilateral wheelchair propulsive mechanics provide biomechanical information about propulsion technique measures like peak forces and push time. However, wheelchair propulsion is a bimanual task, and studies in the area of bimanual motor control have shown that the limbs are not controlled independently, but are coupled to each other. This implies that principles of interlimb coordination cannot be derived from the study of single-limb movements. [20].

Only few studies addressed bimanual upper limb consistency or the variability for that matter of motor performance in this task by using two instrumented wheels simultaneously [16, 21, 22]. Moreover, the provided data on reliability or validity of these measurement-wheels or related ergometer technology in literature are very scarce [10, 23-25], let alone the comparability of different measurement-wheels.

Figure 3.1: Smartwheel[a] (left) and Optipush[b] (right). Both wheels measure three-dimensional forces and torques on the handrim, combined with the angle under which the wheel is rotated. The local coordinate systems are defined as: x- and y-axis, orthogonal to each other in the plane of the handrim, z axis orthogonal to the plane of the handrim and pertruding from the wheel axle. The direction of positive torque is mirrored for the wheels so that they measure positive torque when pushing forward.

In order to evaluate the consistency of such measurement systems during steady-state wheelchair propulsion on a motor driven treadmill, the current study simultaneously assesses two commercially available instrumented wheelchair wheels: the Smartwheel[a] and the Optipush[b] (figure 3.1). The Smartwheel uses instrumented beams to measure torques and forces, whereas the Optipush uses a commercial force-torque sensor at the center, which attaches to the rim through rigid beams. Both come with a clinical software package that can be used for guidance and evaluation of wheelchair adaptations and training programs in clinical or adapted sports practice. An important question is how these measurement-wheels compare to each other, and whether they consistently measure similar technique phenomena since they are based on different measurement approaches, yet are suggested to measure the same variables in the same range of accuracy. Thus from a clinical as well as a scientific perspective, it is important to know whether these wheels are interchangeable and whether we can compare the studies using these different wheels.

One way to compare the measurement-wheels is to fit them to both sides of the same wheelchair, during a steady-state propulsion task on a treadmill. Although both wheels examine the same performance of wheelchair propulsion, the motor control process involves a combined movement of the two upper limbs, which determines speed and direction. When comparing the pushes on both wheels, one faces the problem that human movement is intrinsically variable, both within and between individuals [26]. At present, there is a growing recognition that this variability (e.g. intra- and interlimb variability) is not simply the reflection of noise, but contains features that provide insight about normal learning and pathological processes [27-29]. Consequently for comparison of the wheels it is necessary to isolate the consistency of measurement from the variability inherent to the motor control process. Therefore a time dependent one-on-one comparison between pushes of the left and right arm are not expected to provide a suitable outcome measure for comparing the measurement-wheels against each other. Namely at this relatively small time-scale variability due to motor control and task variability, despite propelling at a constant speed on a motor driven treadmill, is to be expected. Indeed, one study using two Smartwheels specifically looked at the asymmetry of wheelchair propulsion and showed side-to-side differences when matching three pushes left and right [21].

Yet, the set task of straightforward steady-state propulsion on a level treadmill should be intrinsically stable over a larger time-scale and should lead to comparable mean outcome values for the left and right side, resulting in a constant mean power output (product of torque and angular velocity) over time. In the current study it is therefore assumed that systematic differences in unilateral mean power output between both wheels, when propelling at constant speed on a motor driven treadmill, should indicate differences in measurement systems rather then motor variability. For instance van der Woude et al showed a high correlation (r=0.97) of left- and right-hand sprint power averaged over 30 seconds for 67 wheelchair athletes on a computer controlled wheelchair simulator [30]. Although steering on this ergometer is not critical like on a motor driven treadmill, this high interlimb consistency in power production still exists.

Traditionally power output on a motor-driven treadmill is determined through a separate drag test [31]. In the current study, the drag test, combined with the use of a pulley system is used to impose an additional drag force of known magnitude to the wheelchair-user combination on the treadmill [32]. The outcomes of the measurement wheels are compared with this other form of measuring power output. .

Specifically we studied 1) if the two measurement-wheels (Optipush[b], Smartwheel[a]) provided comparable time-averaged data for the left and right hand side during steady-state wheelchair propulsion on a motor-driven treadmill in a group of trained able-bodied subjects and 2) if the power output values for the measurement-wheels were comparable to the power output based on a separate drag test.

Answering these questions will enable researchers and practitioners to better interpret results from both measurement-wheels published in previous studies, and use both wheels in the same evaluation setup in the future. Furthermore it gives information on how earlier estimations of power output, using a drag test and a pulley system, compare to the outcomes of the measurement-wheels. Finally this study will help to further our understanding of details of bimanual variability in propulsion technique during steady-state handrim wheelchair propulsion.

Methods

3.1 Subjects

After having given written informed consent, 9 able-bodied subjects participated in the study. Criteria for inclusion were male, between 18-65 years, no prior experience in wheelchair propulsion, and absence of any medical contra-indications. To compare with earlier research in our laboratory only male subjects were selected. The study was performed according to the guidelines of the Ethics Committee of the Faculty of Human Movement Sciences, VU University Amsterdam (ECB 2011-46).

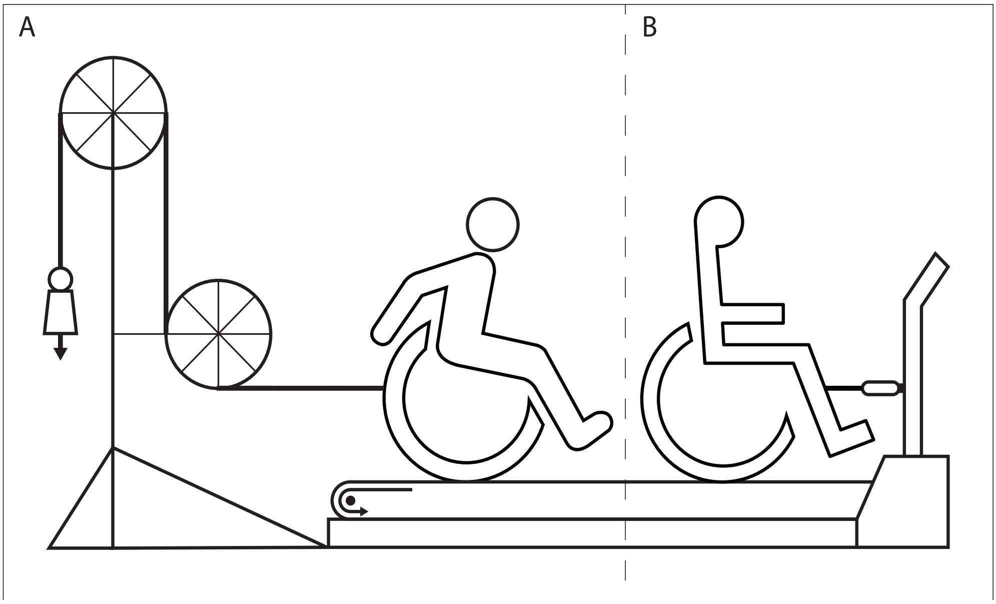

Figure 3.2: Experimental setup. A) To impose the desired power output a pulley-system attaches to the instrumented wheelchair on the treadmill. B) A dragtest is performed beforehand, to determine power output of the user-wheelchair combination.

3.2 *Protocol*

Prior to our study, subjects practiced wheelchair propulsion in 9 practice trials over 3-weeks. Every trial comprised two 4-min exercise blocks at variable low-intensity levels of external power output. The first and the last trial were used as a pre- and post-test and were both extended with one 4-min exercise block. 2-Min rest was given between any two adjacent exercise blocks. Subjects received no specific instructions other than to stay on the treadmill using the handrims. The data for the current study were taken from the post-test that thus consisted of three four-minute blocks (T1, T2, T3) at 1.11 m/s and 0.18 W/kg. Figure 3.2 shows how the power was imposed by adding mass to a pulley system after having performed an individual drag test [31, 32]. Experiments and practice sessions were all conducted on a level treadmill of 2.4 m length by 1.2 m width (Forcelink[c]) in the same experimental wheelchair (Double Performance[d]) with 24-inch measurement-wheels.

3.3 *Measurement-wheels*

The regular rear 24 inch wheels of the standardized wheelchair were replaced with two instrumented wheels; on the left the OptiPush[b] (6.0 kg, Max Mobility) and on the right the Smartwheel[a] (4.9 kg, 3-Rivers). Both wheels measure 3-dimensional forces and torques applied to the handrim, combined with the angle under which the wheel is rotated. These variables were the only ones used in this experiment for data processing; further data processing and interpretation as done by the respective software packages was not included in the current study. Data were wirelessly transferred to a laptop at 200 Hz (Optipush) and 240 Hz (Smartwheel). Both wheels were synchronised by an electronic pulse at the start of each measurement.

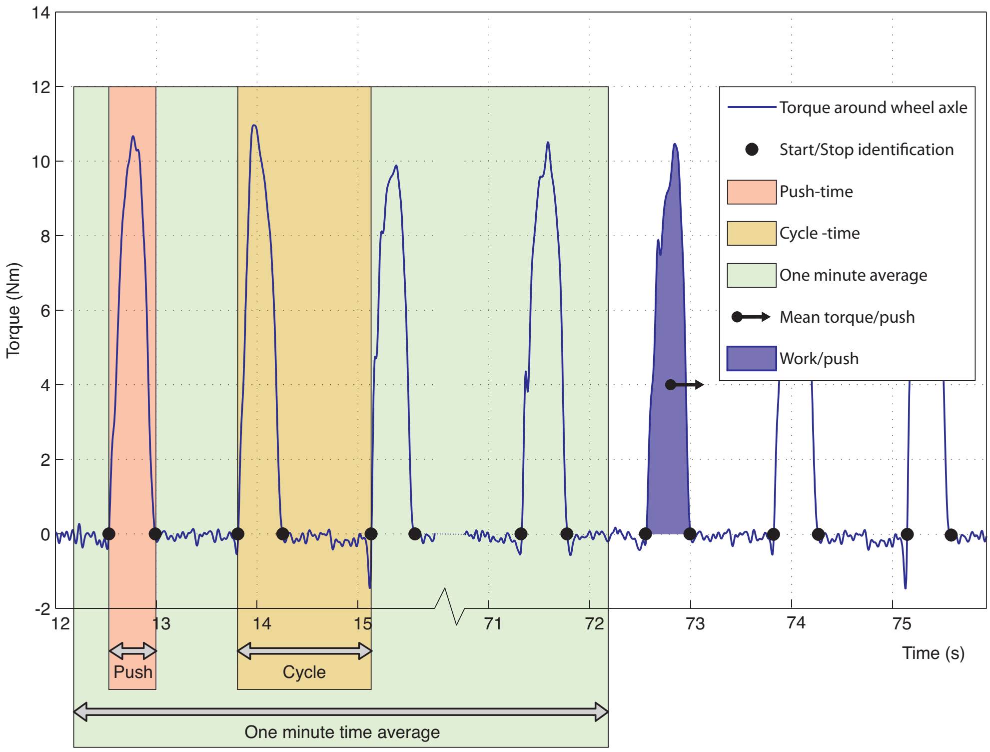

Figure 3.3: Definitions of push variables. Push identification, push-time, cycle-time, work per push, and mean torque. Variables were calculated per puss or over all full push cycles within one minute.

3.4 Data analysis

The rawest data from the instrumented wheels available to the researchers were further analysed using custom-written Matlab routines. Data of all three practice blocks including the rests in between were collected in one continuous measurement. To be sure of stable, steady-state propulsion, each last minute from the three 4-min exercise blocks (T1-T3) was used for the analysis. After data collection the Smartwheel output (240 hz) was downsampled to the frequency of the Optipush (200 hz) using a cubic spline. Per subject and exercise block, nine columns of data output were further used in the comparison between the measurement-wheels. These were the x, y and z components of force (N) and torque (Nm) as expressed by the wheels in their local coordinate systems (fig 3.1), angle (rad), time (s) and sample number. First, individual pushes were defined as each period of continuous positive torque with a minimum of at least 1 Nm.

Table 3.1: Propulsion variables

Variable:	Description:	Equation:
Push time (s)	Time from the start of positive torque to the stop of positive torque for an individual push.	$t_{end}(i) - t_{start}(i)$
Cycle time (s)	Time from the start of positive torque to the next start of positive torque.	$t_{end}(i) - t_{end}(i-1)$
Contact angle (rad)	Angle at the end of a push minus the angle at the start.	$Ø_{end}(i) - Ø_{start}(i)$
Fpeak (N)	3d peak force during the push phase	$Max_{start:end}(Fx^2 + Fy^2 + Fz^2)^{0.5}$
Mean Power/push (W)	The mean power during the push phase.	$Mean_{start:end}(Tz \cdot \Delta Ø)$
Work/push (J)	The power integrated over the duration of the push.	$\sum_{start:end}(Tz \cdot \Delta Ø)$
Frequency (push/min)	Pushes per minute.	$N_{pushes}/\Delta t$
Mean power/minute (W)	The mean total (unilateral) power (Tz*Angular velocity) during a complete series of cycles in a minute multiplied by 2.	$2 \cdot (\sum_{start(i):start(n)}(Tz \cdot \Delta Ø))$

Abbrevations: t, time(s); $_{start}$(i), start of the current push (sample); $_{end}$(i), end of the current push (sample); Ø, angle (rad); Fx, Fy and Fz, force components (N); Tz, torque around wheel axle (Nm).

Over the identified pushes biomechanical characteristics were calculated and later averaged over all pushes within the fourth minute of each practice block per subject. Calculated characteristics are defined in table 3.1 and figure 3.3.

3.5 *Statistics*

First, a cross-correlation was performed between the torque signals around the axle of each of the wheels for each subject, over the whole last minute of each practice block. We were specifically interested in correlation and the time lag between the two measurement-wheels. Possible differences in correlation and time lag were evaluated with repeated-measures Anova.

Secondly, the different biomechanical variables averaged over a minute were compared between the measurement-wheels with an intra-class correlation (ICC) over the different trials. A case 3 ICC was used to compare the degree of absolute agreement of the measurements that are the averages of the three independent 4-minute blocks with the two measurement-wheels as fixed judges [33]. A case 1 ICC was also performed over the three 4-minute blocks within each wheel to relate these within wheel outcomes to the between wheel outcomes. ICC values higher than 0.85 are considered good and measures below 0.7 as poor [34]. To further inspect the differences in power output Bland-Altman plots and limits of agreement were used [35].

Finally the total mean power output measured from the wheels was compared to that estimated by the drag test - pulley combination using an Anova for the different mean power outputs of the different measurement systems. Overall statistical significance was set at $p<0.05$.

Table 3.2: Cross correlation and the corresponding phase lag between the two torque signals around the wheel axis, for the different trials.

	T1		**T2**		**T3**	
Subjects	**Cross correlation**	**lag**	**Cross correlation**	**lag**	**Cross correlation**	**lag**
1	0.97	8	0.98	21	0.97	33
2	0.98	4	0.97	17	0.98	31
3	0.98	7	0.98	17	0.98	28
4	0.96	7	0.97	20	0.98	33
5	0.96	11	0.97	23	0.97	32
6	0.98	8	0.97	17	0.98	32
7	0.98	4	0.98	17	0.98	29
8	0.97	8	0.97	20	0.97	32
9	0.96	8	0.96	22	0.97	35
Mean:	**0.97**	**7.2**	**0.97**	**19.3**	**0.98**	**31.7**
Std:	**0.01**	**2.2**	**0.01**	**2.4**	**0.01**	**2.2**

Table 3.3: Means and standard deviation (between brackets) of propulsion characteristics for the different measurement-wheels (Optipush (Op) en Smartwheel (Sw)) over the three 4-minute blocks (n=9 AB subjects).

Variable:	T1 Op	T1 Sw	T2 Op	T2 Sw	T3 Op	T3 Sw	ICC Betw.	ICC Op Within	ICC Sw Within
Push time (s)	0.37 (0.05)	0.38 (0.05)	0.38 (0.06)	0.37 (0.06)	0.37 (0.06)	0.38 (0.07)	0.97	0.96	0.95
Cycle time (s)	1.24 (0.24)	1.29 (0.27)	1.29 (0.31)	1.24 (0.25)	1.27 (0.27)	1.28 (0.31)	1	0.94	0.95
Contact angle (rad)	1.36 (0.16)	1.37 (0.19)	1.38 (0.2)	1.33 (0.2)	1.34 (0.21)	1.37 (0.24)	0.97	0.96	0.95
Fpeak (N)	66.87 (14.75)	68.92 (16.72)	67.47 (16.34)	63.47 (16.25)	65.25 (19.72)	62.79 (17.3)	0.94	0.92	0.94
Work/push (J)	10.47 (1.76)	10.66 (2.15)	10.72 (2.34)	10.21 (1.88)	10.38 (2.78)	10.57 (2.59)	0.89	0.91	0.94
Mean Power/push (W)	27.73 (5.21)	27.8 (5.35)	27.67 (5.34)	27.82 (5.82)	27.79 (7.01)	27.84 (6.25)	0.9	0.93	0.96
Frequency (push/min)	50.69 (8.87)	49 (9.5)	49.55 (10.16)	50.33 (8.92)	49.27 (9.2)	49.47 (10.18)	1	0.94	0.95
Mean power/ minute (W)	16.4 (2.48)	16.05 (2.63)	16.11 (2.48)	16.28 (2.39)	16.08 (2.61)	16.31 (2.47)	0.89	0.97	0.98

The first ICC is between the wheels. Reported ICC's in the last 2 columns are calculated between the mean per subject of the three blocks for each of the wheels.

Results

The nine male subjects had a mean age of 25.9 years (std = 9.6), a mean body mass of 90.3 kg (std = 12.5) and a mean height of 1.90 m (std = 0.04). All subjects enrolled in the study after 8 sessions (1 first session of 12min and 7sessions of 8min) of low-intensity steady-state wheelchair exercise on the motor driven treadmill.

3.6 *Cross-correlation*

Table 3.2 shows the cross-correlation between the torque signals around the wheel-axle (fig 3.3) of both measurement-wheels for the three different practice blocks for each subject separately as well as the mean over n=9 subjects. For all three blocks we found a high cross correlation (respective means: 0.97, 0.97, 0.98) that did not change significantly ($p=0.46$). However, the lag between the two signals (see table 3.2) for the three blocks did change significantly from a mean of 7.2 samples on T1 to 19.3 on T2 and 31.7 on T3 ($p<0.001$), indicating a shift in time of 0.06 ms between the signals obtained by both wheels between each practice trial.

3.7 *Intra-class correlation*

Mean power output between both wheels had a good intra-class correlation of 0.89 (table 3.3). Within the different wheels the ICC for mean power output over the three 4-minute blocks was considerably higher, 0.97 and 0.98 for the Optipush and Smartwheel respectively. For the other biomechanical variables ICC's between the wheels are high (>0.9), indicating good agreement between both wheels. The variables that took more calculation steps (Work/push, mean power output two-sided and speed) had lower ICC's (table 3.3).

3.8 *Bland Altman plots*

The results on mean power output per push are shown in the Bland Altman plot in figure 3.4. In this plot over n=9 subjects, each individual push of the Optipush has been matched to a time-synchronized push of the Smartwheel and the difference of these pushes is plotted against the mean of those two pushes. As expected, differences between left and right occurred, but the mean difference over the group and measure-

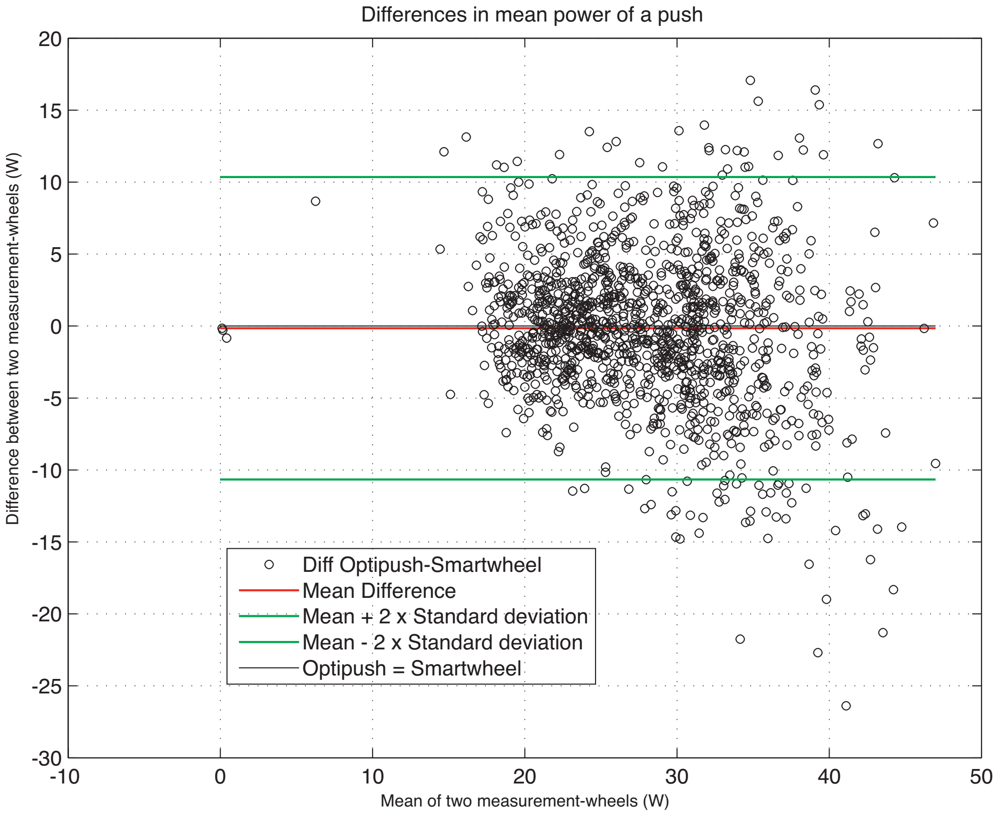

Figure 3.4: Bland-Altman plot of all pushes, for all practice blocks of all subjects. Each push of the Optipushb is matched to the simultaneous one of the Smartwheela. For each push the difference of the two is plotted against the mean of the two (black circles). It is clearly seen that they differ a lot on individual pushes, but on average (red line) there is no clear systematic difference between the wheels.

ment period was very close to zero (-0.03 W). This low mean difference over the group and time, exemplifies that mean power left and right did not differ significantly. Further, figure 3.5 shows the average power output over one minute for each subject (displayed on the x-axis) for each 4-minute block (displayed as different markers). The figure shows that for the individual subjects' differences in power did occur as a consequence of the human motor control process and assumed detailed elements of task variation. Yet on group level (average difference over the group) no systematic differences were found (Mean difference -0.03 W).

3.9 *Measurement-wheels and drag test*

Mean total power output was compared for both wheels and with an external criterion; the calculations of a drag test in combination with a pulley system shown in table 3.4, column 3. The measurement-wheels did not significantly differ from each other (means 16.2 and 16.2 W, $p=0.73$), but measured a significantly higher power output than estimated from the drag test – pulley combination (mean 14.0 W, $p<0.00$).

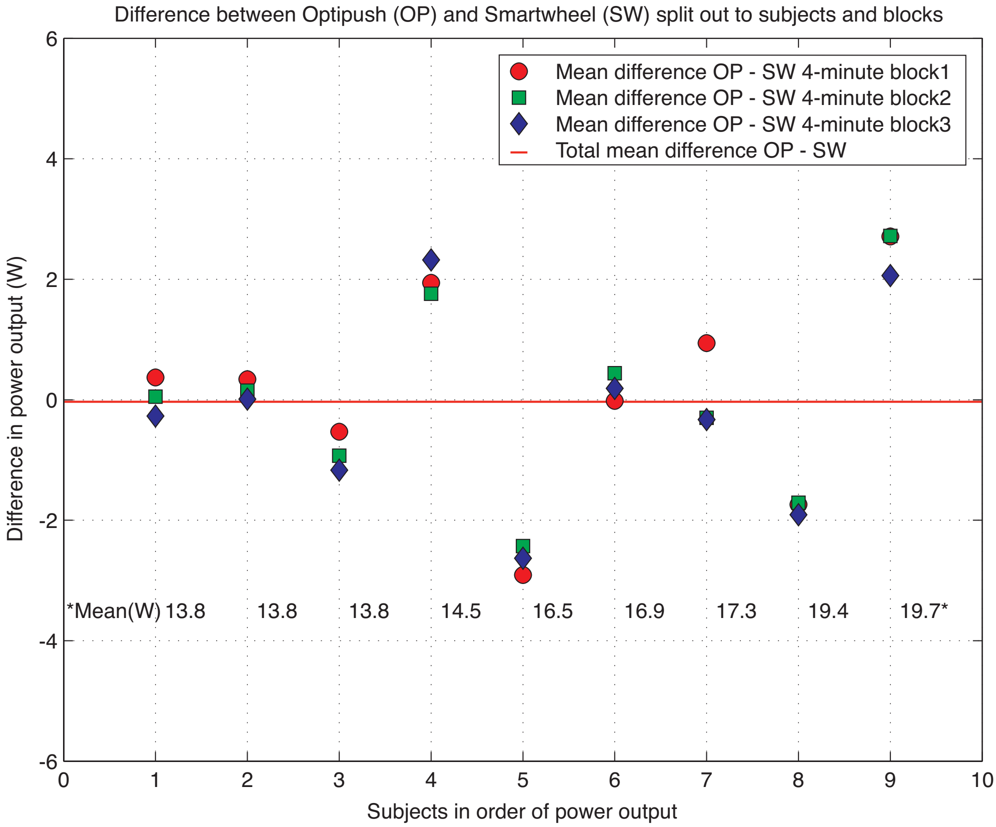

Figure 3.5: Adjusted Bland Altman plot. The differences between Optipushb and Smartwheela are split out for each subject (x-axis) for each 4-minute block (symbols). Subjects are ranked according to their mean power output level.

Discussion

The aim of the present study was to compare two different measurement-wheels with supposedly the same data output under a real life dynamic, yet standardized sub-maximal wheeling condition. The results will help interpretation in future research regarding wheelchair propulsion with the use of these wheels, and will help to further investigate the intricate interlimb coupling during this bimanual task. The results showed good agreement between both wheels during steady-state propulsion on a treadmill. Both in time (cross-correlation) and in amplitude (intra-class correlation) a high correlation between the wheels was found (tables 3.2 and 3.3). With regard to the power output both wheels showed comparable and consistent results.

3.10 Cross correlation

The directly measured torque signals had a high cross-correlation, but over the different 4-minute blocks the time lag between the two signals became larger, from 0.04 s after 4 minutes to 0.16 s after 12 minutes. While the wheels were synchronized at the start a synchronous stop was not possible for the Optipush. The two internal clocks of both devices probably differed, yet in the current setup it was not possible to say in which way since a third source of known reliability was not available. Had a synchronous stop been possible we could have corrected for this phenomenon. Despite the small

Table 3.4: Power output (Pout) as determined by three different methods. There is a significant difference between the measurement-wheels and the drag test.

Subject	Handedness	Mean Pout x2 Op(Watt)	Mean Pout x2 Sw(Watt)	Pout dragtest (Watt)
1	Right	18.5	20.3	15.8
2	Left	17.0	16.8	13.2
3	Right	13.9	13.7	13.7
4	Left	15.5	13.5	11.1
5	Right	13.4	14.3	11.6
6	Right	15.2	17.9	16.0
7	Right	17.4	17.3	15.9
8	Right	13.8	13.8	11.9
9	Right	21.0	18.5	16.5
Mean:		**16.2**	**16.2**	**14.0**

magnitude such options would be greatly appreciated in the future, especially if these measures are to be combined with other measurement systems like EMG or position registration.

3.11 Intra-class correlation

Most push characteristics had an ICC higher than 0.9. The timing parameters push time, cycle time and frequency approach an ICC of 1.00 and so did the contact angle. These results all added to the conclusion that the provided signals by both wheels were highly comparable in the time and space profiles. The force-related parameters peak force, work per push and power had slightly lower ICC's, but were still sufficiently high given that they measured different limb actions, which force profiles had to be produced individually and might also depend on hand dominance [36, 37]. Since the data were collected within a larger framework of experiments it was decided not to change the sides of the different measurement-wheels, which could have shown differences due to hand dominance. Yet the studies that did look into the interlimb coupling and relationship between dominant and non-dominant hand in wheelchair propulsion did not yet show a clear effect of handedness [16, 21]. Future experiments using both wheels could further investigate this possible confounder of the results.

3.12 Bland Altmann plots

The coupling of propulsion to steering in real life and on the motor driven treadmill seems to make wheelchair propulsion an intrinsic variable task requiring continuous coordination by the human motor system. Considerable left-right differences were found when comparing single pushes, showing variation in power output between the different sides, as was expected from a motor control perspective [19] The Bland Altman plot in figure 3.4 is alternatively visualized in figure 3.6. This figure illustrates how left-right differences influence direction and how eventually subjects manage to stay on

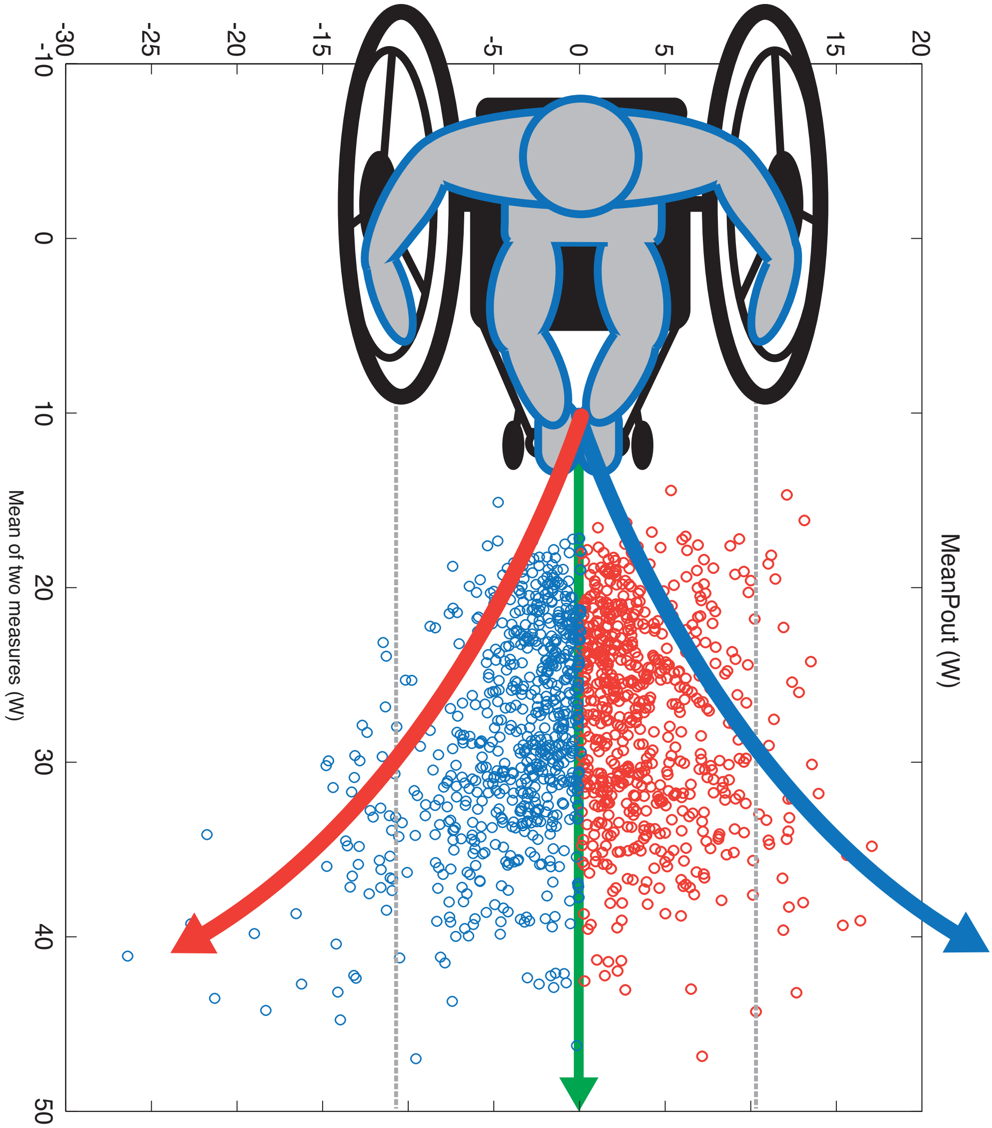

Figure 3.6: Top view visualization of steering and propulsion. This adapted version of figure 4 shows the effect of the differences in mean power output of individual pushes with regard to steering. Blue circles are pushes where the Smartwheel measured more power leading to a change in the direction of the blue arrow. Vice versa for the Optipush the red circles lead to a change in the direction of the red arrow. Result is the green arrow, which was the overall outcome (staying on the treadmill).

the treadmill. Whether wheelchair propulsion is considered an asymmetrical act depends on the research interest. Clearly pushes left and right are not exactly the same and even differ considerably from time to time. As such, research fields like motor learning would greatly benefit from the use of two wheels to see how this variability changes because of a practice or feedback intervention. On the other hand over a larger time scale in a straightforward steady state submaximal propulsion task both limbs show very comparable propulsion performance allowing for generalization of findings from one side to the other. For instance seat height changes might be studied with just one measurement-wheel[38].

3.13 Left-right power output

Power output is a complex output measure, using different components of a measurement-wheel, in this case torque multiplied by angular velocity averaged over time. Assuming a balanced, well-maintained and good quality wheelchair system on a stable and level treadmill, average power output at the left and right side should be identical over time during steady-state wheelchair propulsion on a motor driven treadmill. Although on group level the average difference in power output indeed was almost zero, individual subjects clearly showed differences between the mean power output left and right (figure 3.5). Accordingly the ICC for mean power output was lower than for some of the other propulsion characteristics and remarkably lower than the ICC within the measurement-wheels for the three different trials. This means that as expected mean power output of the three different trials was more consistent within the wheels than between wheels.

The individual differences in power output left and right might have different causes. First, dependent on the weight distribution of the subject in the chair and the position fluctuations of the subject-wheelchair combination on the treadmill belt, rolling friction in both the rear wheels and the front castor wheels might be different left and right. More weight on the left or right front castor wheel will increase friction on that side, leading to a higher necessary power output on that side. Second, the weight distribution by the subject between both wheels also proportionally influences the power output; leaning over to one side makes the power output on that side necessarily larger. Third, the pulley system was positioned in line with the center of the treadmill, however subjects propel an approximately 0.75 m wide wheelchair on a 1.20 m wide treadmill, which allowed for movement toward either side of the treadmill. Propelling the chair more to the side will give a force component from the pulley system orthogonal to the wheels resulting in more power output on the outer side with respect to the pulley system. Also, the belt tension of the treadmill is somewhat different at the sides versus the center of the belt, which may also lead to slightly higher levels of rolling resistance when coasting left or right on the treadmill instead of in the middle.

Finally, the wheels under study might be of influence on the measured power output. The differences in mass and inertia of both wheels could potentially have influenced left-right power output. The suppliers did unfortunately not make inertial properties available, which would be greatly appreciated in the future development. Secondly the two wheels could measure torque and angle differently resulting in different power output. While the first three arguments are assumed to be distributed equally over all subjects, the argument of the measurement-wheels is a systematic difference. Although the other reasons might have masked a difference due to the measurement-wheels, the absence of a systematic difference in power output or any of the other outcomes in our view supports the conclusion that differences between subjects seem to be caused rather by their own propulsion behavior and geometrical characteristics than by the measurement systems used.

The agreement in power output as found in the present study is in line with earlier studies [30, 39]. In different experimental setups these studies also found good agreement on power output for both sides. Important in this respect is also the operationalization of the term power output. As mentioned in the introduction Hurd et al [21] averaged the power over only three push cycles while the other studies averaged power over more cycles. Therefore their finding of asymmetry in power output, averaged over just three consecutive pushes, seems to be in line with our finding of high variability in the left-right difference in power output (figure 3.4).

3.14 Measurement-wheels and Dragtest

The wheels measured more power than estimated from the dragtest-pulley combination. In addition to the aforementioned consequence of task variation on power output other factors might have contributed to this difference. First the drag test was performed without speed changes, but at a constant speed. Secondly, additional losses when going to the front and back of the treadmill or left-right are not measured by the drag test, but are measured by the wheels. Thirdly, the friction on the front wheels is also dependent on weight distribution. During the drag test subjects were seated in a uniform position (sitting upright with the trunk; hands on the lap), while during propulsion they were free to move in their wheelchair (e.g. with trunk flexion/extension) possibly leading to more rolling friction and thus a higher power. The measurement-wheels seem to measure power in a more accurate way, because they are sensitive to change of torque and angular velocity, still the drag test is the only external comparison currently possible and is relatively cost effective and easy to use

Conclusion

A good agreement between both measurement-wheels was found in this study. Data from both wheels seem consistent and suitable to be used together in experiments on wheelchair propulsion. Both wheels measure a higher mean total power output compared to the estimation of power output using a drag test. If a standardization of an experiment is done using a drag test this should be taken into account. Variability in the execution of wheelchair propulsion seems an essential part in the motor control of this bimanual task. Further research into the interlimb coupling during this bimanual task might use bilateral measurement-wheels to explain and understand the variability between and within the push cycles of both wheels.

Endnotes:
a Three Rivers Holdings, Mesa, AZ, USA
b MAX Mobility, LLC, Antioch, TN, USA
c Forcelink b.v, Culemborg The Netherlands
d Double Performance BV, Gouda, The Netherlands

3.15 References

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chapter 4

Initial Skill Acquisition of Handrim Wheelchair Propulsion: A New Perspective.

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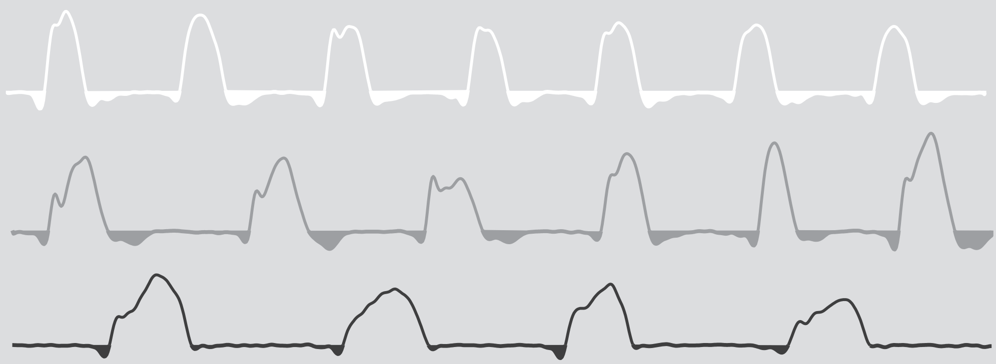

Abstract

To gain insight into cyclic motor learning processes, hand rim wheelchair propulsion is a suitable cyclic task, to be learned during early rehabilitation and novel to almost every individual. To propel in an energy efficient manner, wheelchair users must learn to control bimanually applied forces onto the rims, preserving both speed and direction of locomotion. The purpose of this study was to evaluate mechanical efficiency and propulsion technique during the initial stage of motor learning. Therefore, 70 naive able-bodied men received 12-minutes uninstructed wheelchair practice, consisting of three 4-minute blocks separated by 2 minutes rest. Practice was performed on a motor-driven treadmill at a fixed belt speed and constant power output relative to body mass. Energy consumption and the kinetics of propulsion technique were continuously measured. Participants significantly increased their mechanical efficiency and changed their propulsion technique from a high frequency mode with a lot of negative work to a longer-slower movement pattern with less power losses. Furthermore a multi-level model showed propulsion technique to relate to mechanical efficiency. Finally improvers and non-improvers were identified. The non-improving group was already more efficient and had a better propulsion technique in the first block of practice (i.e. the 4th minute). These findings link propulsion technique to mechanical efficiency, support the importance of a correct propulsion technique for wheelchair users and show motor learning differences.

Introduction

When confronted with a new motor task the performance of this task will usually improve through practice. This process of skill acquisition is a key element of human functioning during daily life and is an essential element during rehabilitation after disease or injury. A typical example of a totally new motor skill to be learned during rehabilitation is handrim wheelchair propulsion. Despite advances in technology and possibilities of other propulsion mechanisms the hand rim-propelled wheelchair is still the most often used form of mobility for those who lost their walking ability [1]. However, compared to other forms of ambulation the mechanical efficiency, i.e. the ratio of external power output over internal power production, of hand rim propulsion is low, while at the same time overuse problems are common [2-6]. Increased proficiency of the wheelchair propulsion skill is implied to improve mobility and reduce risks of injury, where literature specifically advices to use long smooth strokes leading to a reduced frequency of movement [7].

To gain insight into motor learning processes of cyclic motor tasks, the study of hand rim wheelchair propulsion, as a form of ambulation in daily life and rehabilitation is very suitable, because it entails several unique features. First, the cyclic nature of steady-state wheelchair propulsion makes it possible to evaluate performance using energy consumption as a generic outcome measure of motor learning [8]. Second, during the push and recovery phase there are multiple degrees of freedom enabling the user to perform the task in different ways, allowing propulsion technique to change between the left and right wheel and over time within one side [9]. Finally wheelchair propulsion is a task that is new to persons who just lost their walking ability and to many able-bodied participants as well. Therefore, in the study of motor learning able-bodied participants can serve as a model to study the early acquisition of this skill, without being too heterogeneous as a group, because of for instance spinal cord lesion level or upper-body asymmetries and without being hindered by the recent trauma early in rehabilitation.

With regard to motor learning in every day cyclic tasks [10], Sparrow and Newell proposed a constraints-based framework of metabolic energy expenditure, motor coordination and control [11]. Central to their model is the suggestion that observed movement patterns emerge from the interaction between different (external, task and internal) constraints, with metabolic energy being the currency of the interaction. In other words, motor learning is the process of acquiring a movement pattern that minimizes the energy expenditure within the constraints of the task. In line with this model, several learning studies using different cyclical upper- or lower-body tasks found a reduction in energy expenditure through practice [8,12-15]. For instance learning studies using a ski-simulator or a rowing-ergometer showed a reduction of energy expenditure through practice while maintaining the same power output [8,12]. The reduced energy cost in these different cyclical tasks coincided with an increase in movement amplitude and a decrease of movement frequency described as a longer-slower movement pattern. For handrim wheelchair propulsion this corresponds with a longer stroke, both in time and space when pushing and thus with a reduction in the frequency of these pushes.

Indeed motor learning studies in wheelchair propulsion using either an instrumented ergometer or using ambulant measurement-wheels have shown an increase in mechanical efficiency in combination with a longer stroke and reduced frequency for both able-bodied participants and people with a disability [16-23]. In these studies practice interventions ranged from three to twelve weeks and one study followed persons with spinal cord injury observationally over the course of rehabilitation [24]. Two (combined) studies evaluated the initial first 12 minutes of wheelchair propulsion practice performed by nine novice able-bodied participants on a wheelchair ergometer [25,26]. These two studies showed that propulsion technique measures, like a reduced push frequency and an increased peak torque, changed already during the first 12 minutes of practice, however a reduction of energy expenditure was not found.

The current study will revisit the initial motor learning process and evaluate the changes in energy expenditure and propulsion technique over this short 12 minutes period. Three key differences with respect to the earlier studies will further our understanding of changes in mechanical efficiency and its relation to propulsion technique. The first is the use of a treadmill in combination with ambulant measurement wheels instead of a stationary ergometer [27]. Due to the necessity to combine both steering and propulsion the use of a motor driven treadmill is expected to be more demanding, leading to an increased movement variability and subsequently having more degrees of freedom that need to be learned during practice and thus being more similar to overground wheelchair propulsion. Second, the availability of a large sample of 70 able-bodied participants, will make it possible to not only examine the changes over time of energy expenditure and propulsion technique, but also to examine the interaction(s) between propulsion technique and biomechanical variables using multi-level regression analyses. Finally, the larger groups allows for studying possible differences in motor learning capacity between participants [28-30].

Therefore the objective of the current study was to establish whether the motor learning process during the first 12 minutes of handrim wheelchair propulsion would lead to 1) an increased mechanical efficiency and a longer-slower movement rhythm; 2) an association of propulsion technique to mechanical efficiency within and between participants; 3) differences between participants in the motor learning process based on the degree of improvement in mechanical efficiency.

The typical changes of propulsion technique that are found after longer practice interventions such as a reduction in frequency and increase in contact angle and reduction of negative work are expected to already be seen within the twelve minute practice intervention [16-23,25,26]. As a consequence of a more effective propulsion technique a directly increased mechanical efficiency is expected.

Methods

4.1 Participants

After having given written informed consent, a convenience sample of 70 able-bodied men participated in the study. To compare our results with previous research the criteria for inclusion were male, between 18-65 years, no prior experience in wheelchair propulsion, and absence of any medical contra-indications [16,18,19,25,31]. The partici-

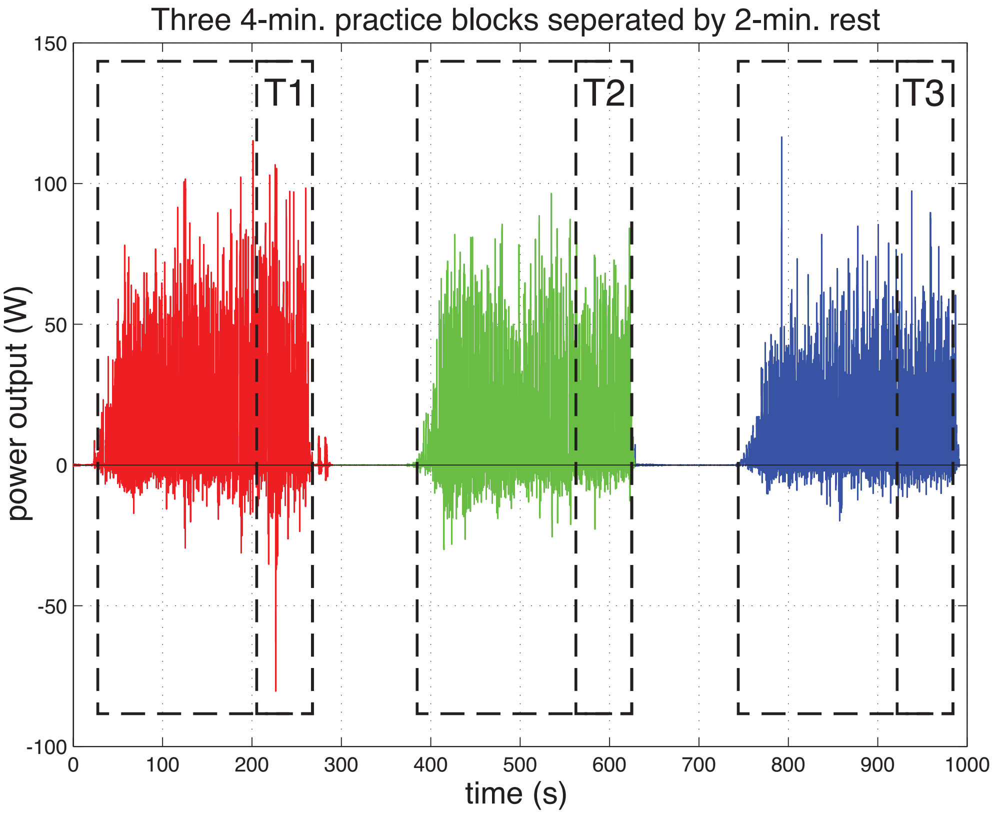

Figure 4.1: Example of the power output over the whole practice protocol for one participant. Three 4-minute practice blocks are separated by two minutes of rest. Data was analyzed at the last minute of each practice-bock, shown as T1, T2 and T3, being the 4th, 8th and 12th minute of practice

pants had a mean age of 22.8 years (std = 3.6), a mean body mass of 80.2 kg (std = 11.4) and a mean height of 1.87 m (std = 0.07). All participants signed an informed consent. The study was performed according to the guidelines of the Local Ethics Committee of the center for Human Movement Sciences, University Medical Center Groningen, University of Groningen.

4.2 Protocol

The single session 16-minute experiment was conducted on a level treadmill of 2.4 m length by 1.2 m width (Forcelink[a]) in the same experimental wheelchair (Double Performance[b]) with 24-inch measurement wheels. Each participant performed three 4-minute exercise blocks at a fixed submaximal power output of 0.20 W/kg body weight, with two minutes of rest in between blocks. This low intensity was chosen to minimize fatigue or training effects and focus primarily on motor learning. The first 40 seconds were used to get the treadmill up to a speed of 1.11 m/s (4 km/h). Participants received no specific instructions other than to stay on the treadmill using the hand rims. Apart from rolling resistance, the required power output was imposed by adding mass to a pulley system. Pulley mass was determined from the results of an individual wheelchair drag test [5,32].

Table 4.1: Propulsion technique variables

Variable:	Description:	Equation:
Energy expenditure (W)	Calculated from the oxygen uptake and respiratory exchange ratio according to Garby and Astrup [33]	((4.94*RER+16.04)*(1000*VO2))/ 60
Mechanical efficiency (%)	The percentage of internal power used for external power delivered at the wheels	Mean power output/ Energy expenditure
Push time (s)	Time from the start of positive torque to the stop of positive torque for an individual push.	$t_{end}(i)$ - $t_{start}(i)$
Cycle time (s)	Time from the start of positive torque to the next start of positive torque.	$t_{end}(i)$ - $t_{end}(i\text{-}1)$
Frequency (push/min)	The number of complete pushes per minute.	$N_{pushes}/\Delta t$
Pos. Work/push (J)	The power integrated over the Contact angle of the push.	$\sum_{start:end}(Tz \bullet \Delta Ø)$
Neg. Work/cycle (J)	The power integrated over the wheel rotation angle during the recovery phase	$\sum_{end:start}(Tz \bullet \Delta Ø)$
Net Work/cycle (J)	The mean power output divided by the push frequency	Mean(P_{out}) / Frequency
%NegWork/cycle (%)	The Neg. work per cycle relative to the Net Work/cycle	(Neg. Work/cycle / (Net Work/cycle))*100
PnegS (W)	The minimum power preceding the push phase	$Min_{<start}$ (Power)
PnegE (W)	The minimum power following the push phase	$Min_{>end}$(Power)
Contact angle (º)	Angle at the end of a push minus the angle at the start.	$Ø_{end}(i)$-$Ø_{start}(i)$
Slope (Nm/s)	The rate of rise from the start of the push phase to the maximum delivered torque around the axle	MaxTorque/Δt
$Ftot_{mean}$ (N)	3d mean force within the push phase	$Mean_{start:end}(Fx^2+Fy^2+Fz^2)^{0.5}$
$Ftot_{peak}$ (N)	3d peak force within the push phase	$Max_{start:end}(Fx^2+Fy^2+Fz^2)^{0.5}$
FeF_{mean} (%)	Mean Fraction effective Force	$Mean_{start:end}(F_{tangential}/F_{total})$
FeF_{peak} (%)	Peak Fraction effective Force	$Max_{start:end}(F_{tangential}/F_{total})$

Abbrevations: t, time(s); $_{start}(i)$, start of the current push (sample); $_{end}(i)$, end of the current push (sample); Ø, angle (rad); Fx, Fy and Fz, force components (N); Tz, torque around wheel axle (Nm);

4.3 Measurement wheels

The experimental wheelchair was kept constant (e.g. tire pressure was inspected before testing), and no individual changes were made to the wheelchair for the different participants. The regular rear wheels of the standardized wheelchair were replaced with two instrumented wheels; on the left the Optipush[c] (Max Mobility) and on the right the Smartwheel[d] (3-Rivers). Both wheels measure 3-dimensional forces and torques applied to the hand rim, combined with the angle under which the wheel is rotated. Data were wirelessly transferred to a laptop at 200 Hz (Optipush) and 240 Hz (Smartwheel). Both wheels were synchronised by an electronic pulse at the start of each measurement [27]. Data from the Optipush were primarily used in the analyses, only when the Optipush data were lacking they were replaced with Smartwheel data. Data of both wheels show good comparability, with an intra-class correlation (ICC) of 0.89 for mean power output and ICC's higher then 0.90 for propulsion technique characteristics [27].

4.4 Energy expenditure

Oxygen consumption (VO2) was continuously measured during the 16-minute experiment using breath-by-breath open circuit spirometry[e]. The gas analyzer was calibrated using a Jaeger 5l syringe, room air and a calibration gas mixture. Data collected over the fourth minute of each exercise block were averaged and taken to reflect physiological steady-state wheelchair propulsion. From the VO2 (L/min), VCO2 (L/min) and respiratory exchange ratio (VCO2/VO2) the energy expenditure was determined using the formula proposed by Garby and Astrup [33].

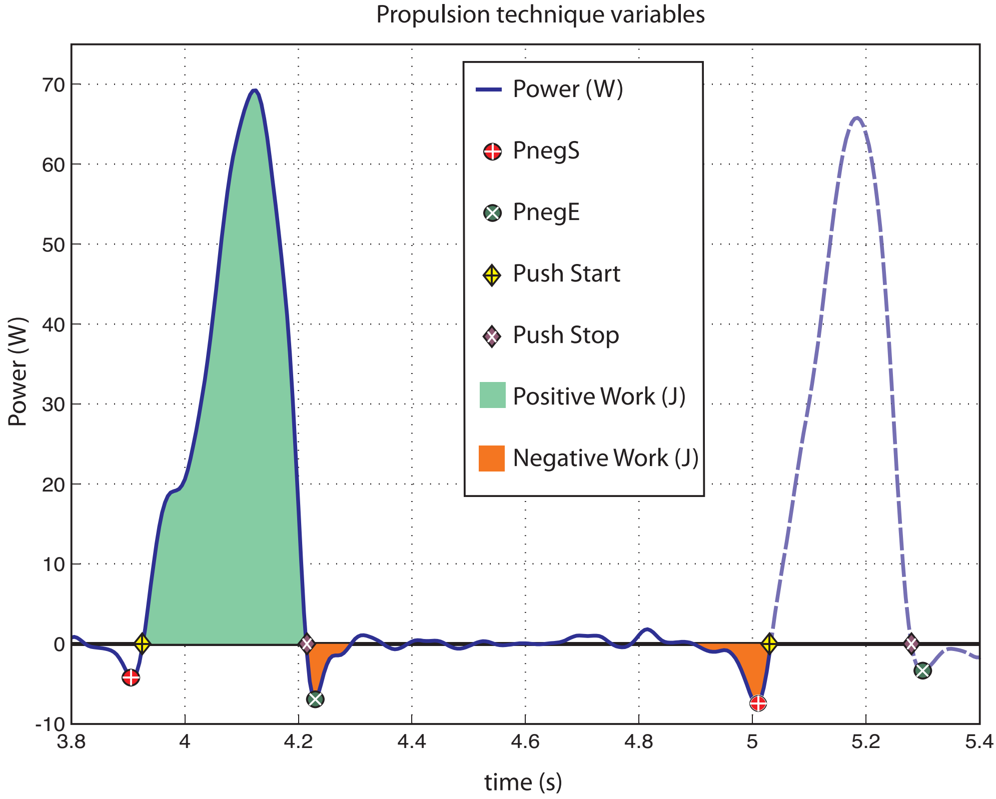

Figure 4.2: Definition of the analyzed propulsion technique variables. Plotted is the power production over time. Note the push detection of the start and end of a push and the negative dips before (PnegS) and after the push (PnegE).

4.5 *Data analysis*

The data from the instrumented wheels were further analysed using custom-written Matlab routines. Data of all three blocks including the rest periods were collected in one continuous measurement (figure 4.1). To be sure of stable , steady-state propulsion, each last minute from the three 4-min blocks (T1-T2-T3) was used for the analysis. Per participant and block, nine parameters of data output were further used in the analysis. These were the x, y and z components of force (N) and torque (Nm) as expressed by the wheels in their local coordinate systems, angle (rad), time (s) and sample number. Individual pushes were defined as each period of continuous positive torque around the wheel axis with a positive minimum of at least 1 Nm. Over the identified pushes biomechanical variables were calculated and subsequently averaged over all pushes within the fourth minute of each practice block per participant. Calculated variables are defined in table 4.1 and figure 4.2. Over these variables the coefficient of variation (CV), i.e. the ratio of the individual standard deviation to the mean for each practice-block, was calculated to see if participants would reduce in variability because of motor learning.

4.6 *Statistics*

For each propulsion-technique variable and its CV a repeated-measures Anova was performed, followed by a post-hoc analysis to see which blocks differed from each other. Significance for the repeated measures Anova was set at a $p<0.05$ and by use of

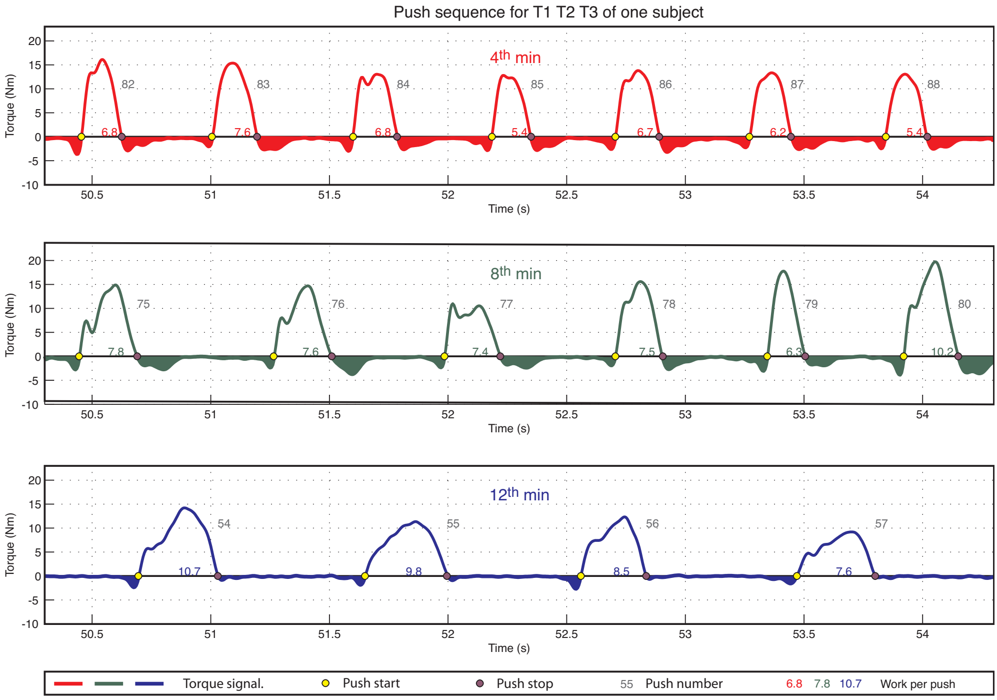

Figure 4.3: Typical example of the individual torque signal for the 4th (T1), 8th (T2) and 12th (T3) minute. Over the whole group participants show a reduction in frequency and increase in work per push. Further the negative work per cycle is reduced for each next measurement period, depicted as the filled surface below zero.

the Bonferroni correction the significance for the post hoc t-test between any two different blocks was $p<0.017$ [34]. Effect size was calculated using partial eta-squared and interpreted as small (≥ 0.01), medium (≥ 0.06), or large (≥ 0.14) [35].

To evaluate the relationship between propulsion technique and mechanical efficiency multi-level analysis was performed using MLWin [36]. The different propulsion technique variables (table 4.1) were first studied univariate in relation to the dependent variable mechanical efficiency. The variables that related significantly with $p<0.05$ to mechanical efficiency were used for multivariate analysis. Since the different propulsion variables are not all independent, but are theoretically linked to each other they are not expected to all remain in the multivariate model. First, all the variables that were significant in the univariate model were added to the multivariate model and then, using a backward regression procedure, one-by-one the non-significant terms were removed to come to the final model. This final model shows the relation of the resulting propulsion technique variables in the model to mechanical efficiency over all observations of participants and blocks.

To examine whether a change in propulsion technique relates to a change in mechanical efficiency a second multi-level analysis was done on the delta scores, i.e. the differences between the blocks (T2-T1 and T3-T2). Here the same method was applied as above to see which variables fitted the model best. The final delta model shows if changes in propulsion technique within participants relate to changes in mechanical efficiency.

Table 4.2: Means and standard deviations of all analyzed technique parameters for the final minute of each of the three 4-minute practice blocks (T1,T2,T3). Last column shows the p-value of the repeated measures Anova. * Notes a significant post-hoc difference between all three blocks. +Represents a significant value for the main effect, but not for all post-hoc differences. Trends of significant changes over time are shown with arrows.

Wheels + Oxycon (N=70)	T1		T2		T3		F	P	η_p^2
	Mean	Std	Mean	Std	Mean	Std	(2,138)		
Energy expenditure (W)	371	108	345	100	332	85	19.46	<0.001*,↓	0.22
Mechanical efficiency (%)	4.8	1.2	5.3	1.3	5.5	1.1	33.27	<0.001*,↑	0.33
Push time (s)	0.26	0.06	0.29	0.06	0.31	0.06	45.89	<0.001*,↑	0.40
Cycle time (s)	0.91	0.29	1.00	0.32	1.05	0.31	28.69	<0.001*,↑	0.29
Frequency (push/min)	73.0	20.9	66.0	18.6	62.2	17.2	27.44	<0.001*,↓	0.28
Pos. Work/push (J)	8.56	2.94	9.36	3.05	9.76	2.93	28.67	<0.001*,↑	0.29
Neg. Work/cycle (J)	-0.85	0.89	-0.68	0.86	-0.51	0.69	19.04	<0.001*,↓	0.22
Net Work/cycle (J)	7.67	3.07	8.62	3.19	9.19	3.09	34.59	<0.001*,↑	0.33
PnegS (W)	-8.1	4.4	-6.1	3.3	-5.5	3.1	43.18	<0.001*,↓	0.38
PnegE (W)	-5.0	5.4	-3.9	4.8	-2.8	3.4	21.49	<0.001*,↓	0.24
Contact angle (°)	55.1	13.0	61.1	12.5	64.5	12.9	41.87	<0.001*,↑	0.38
Slope (Nm/s)	106.2	42.8	90.1	31.6	83.6	30.4	25.43	<0.001*,↓	0.27
F_{mean} (N)	47.2	14.1	45.3	12.2	45.0	11.5	4.18	0.02^{+}, ↓	0.06
F_{peak} (N)	76.4	24.2	73.7	21.5	73.8	19.9	2.35	0.1	-
FeF_{mean} (%)	67.19	8.46	69.19	9.28	69.00	8.97	4.05	0.02^{+}, ↑	0.06
FeF_{max} (%)	96.3	13.3	98.3	15.5	98.5	15.7	1.47	0.23	-

Finally, we examined motor learning differences between participants. The group was split in two, based on a relative increase in mechanical efficiency of larger then 10% between T1 and T3, to ensure that differences in learning were above the natural expected variation. The two groups were subsequently compared on their mechanical efficiency and the most important propulsion technique variables (as shown by the multi-level model) over all three practice-blocks using repeated-measures Anova, with the interaction between group (≤10% or >10%) and practice-blocks as the most important outcome.

Results

All participants were able to complete the whole protocol without incidents. The Optipush data (left side) were used 66 times, while Smartwheel data (right side) were used 4 times. On average participants practiced at a power output of 17.4W (std = 3.67). Figure 1 shows a typical example of the power production over the whole measurement period of one participant, while figure 4.3 gives a more detailed view of the changes in torque production at the three practice-blocks. Table 4.2 lists the results for mechanical efficiency and the propulsion technique variables over time (T1-T3).

4.7 Energy expenditure

The energy expenditure as calculated from the oxygen consumption significantly reduced (from 371 to 345 to 332W), accounting for an increased mechanical efficiency (from 4.8 to 5.3 to 5.5%) over the three blocks (Table 4.2). For both measures the post-hoc comparison showed statistically significant changes over time, i.e. a higher mechanical efficiency each next block.

4.8 Propulsion technique

A significant increase in push time (from 0.26s to 0.29s to 0.31s) and cycle time (from 0.91s to 1.00s to 1.05s) was found (Figure 4.3.). The increase in cycle time over the three practice blocks was associated with a reduced frequency (from 73.0 to 66.0 to 62.2 pushes per minute). The positive work per push went up (from 8.56J to 9.36J to 9.76J) from T1 to T3, while the amount of negative work per cycle reduced (from -0.85J to -0.68 to -0.51J) with practice, which leads to an increased net work per cycle. The reduced amount of negative work was achieved by significantly reducing both the negative phases before the push (from -8.1W to -6.1W to -5.5W) and after the push (from -5.0W to -3.9W to -2.8W). Figure 3 shows the change in propulsion technique of one participant over the three blocks.

The increased work per push was achieved by an increase of the contact angle on the hand rim (from 55.1 to 61.1 to 64.5 degrees), rather than an increase of force application, i.e. no increase of either Ftotmean or Ftotpeak was found. The mean push force Ftotmean actually went down (from 47.2N to 45.3N to 45.0N), which was a significant main effect, but post-hoc tests only showed a significant change between the first and last block. The slope, i.e. the rise of torque per second, significantly reduced (from 106.2Nm/s to 90.1 Nm/s to 83.6 Nm/s) showing that the peak torque was reached over a longer range of both time and angle. The mean fraction effective force showed a significant main effect (from 67.2% to 69.2% to 69.0%), but post-hoc tests only showed a significant difference between the first and second block. The peak fraction effective force did not change significantly.

4.9 Within subject variability

Participants significantly reduced the coefficient of variation for the positive work per push, slope, contact angle, Ftotmean, cycle time, Ftotpeak, and push time. The largest reduction was found in the positive work per push (from 24.9 to 22.1 to 20.1%), which is a 19.2% reduction of the between cycle variability.

4.10 Relationship of propulsion technique to mechanical efficiency

Table 4.3 lists the univariate relation of the different propulsion technique variables to mechanical efficiency. Table 4 shows the final multivariate models for the three practice-blocks and their delta values. In the final multivariate model the percentage neg-

Table 4.3: Univariate multi-level models, with mechanical efficiency as the dependent variable.

	Constant	SE	Beta	SE	p-value	Explained Var. (%)
Empty	5.20	0.13	-	-	-	-
Neg. Work/cycle (J)	5.89	0.13	1.01	0.09	<0.001	31.43
PnegE (W)	5.85	0.13	0.17	0.02	<0.001	30.97
Frequency (push/min)	7.53	0.27	-0.04	0.00	<0.001	27.95
PnegS (W)	6.36	0.16	0.18	0.02	<0.001	27.89
Contact angle (º)	2.11	0.31	0.05	0.01	<0.001	25.72
Push time (s)	2.23	0.31	10.31	1.00	<0.001	23.23
Slope (Nm/s)	6.61	0.21	-0.02	0.00	<0.001	18.37
Cycle time (s)	3.14	0.29	2.08	0.26	<0.001	18.04
Net Work/cycle (J)	2.99	0.25	0.26	0.03	<0.001	17.32
Pos. Work/push (J)	3.01	0.31	0.24	0.03	<0.001	4.72
FeF_{mean} (%)	3.90	0.66	0.02	0.01	<0.05	1.57
F_{mean} (N)	6.04	0.38	-0.02	0.01	<0.05	0.39

Table 4.4: Multivariate multi-level models, with mechanical efficiency as the dependent variable.

ME	Normal values				Delta values			
	Beta	SE	p-value	R^2	Beta	SE	p-value	R^2
Constant	4.231	0.356			0.122	0.068		
% Neg work/cycle	-0.063	0.007	<0.0001		-0.060	0.013	<0.0001	
Contact angle	1.414	0.305	<0.0001	47%	1.318	0.666	<0.05	
Freq(min)	-	-	-		0.028	0.009	<0.05	
Net Work/cycle	-	-	-		0.163	0.071	<0.05	35%

ative work per cycle and the contact angle related significantly to mechanical efficiency, explaining together 49% of the variance in mechanical efficiency (Table 4.4).

The change model based on the delta-scores showed that a change of percentage negative work per cycle, contact angle, frequency and net work per cycle related to a change in gross ME, together explaining 35% of the variance in change of mechanical efficiency (Table 4.4).

4.11 Individual differences in motor learning

From the 70 participants 46 increased their mechanical efficiency with more then 10 percent between T1 and T3 whereas 24 did not. The repeated measures Anova (figure 4.4, table 4.5) showed interaction effect between group and practice-blocks ($p<0.0001$). The 24 non-improvers had a significant higher mechanical efficiency at T1 compared to the improving group (5.6% vs. 4.4%, $p<0.001$). At T2, because of the Bonferroni correction, the difference between groups almost reached significance (5.7% vs. 5.1%, $p=0.026$). At T3 the non-improving and improving group were equal (5.5% vs. 5.5%, $p<0.97$). The four propulsion technique variables, i.e. percentage negative work per cycle, contact angle, frequency and net work per cycle, that were identified by the multilevel analysis as being strongly related to mechanical efficiency, also showed an interaction effect between group and practice-blocks ($p<0.001$).

Table 4.5: Means and standard deviations of mechanical efficiency and the most important technique parameters for the final minute of each of the three 4-minute practice blocks (T1,T2,T3). The t-test shows the differences between the groups within a practice block. The interaction effect shown by the repeated measures anova shows the learning differences. * Notes a significant effect of p<0.001.

		T1			T2			T3			F time		F group		F time*group	
		Mean	Std	p-ttest	Mean	Std	p-ttest	Mean	Std	p-ttest	(2, 136)	η_p^2	(1, 136)	η_p^2	(2, 136)	η_p^2
Mechanical efficiency (%)	<10%	4.37	1.10	0.000	5.06	1.26	0.042	5.52	1.20	0.970	48.61*	0.42	5.15*	0.42	32.79*	0.33
	>10%	5.62	1.00		5.71	1.16		5.51	1.05							
Neg. Work/cycle (J)	<10%	-0.44	0.42	0.005	-0.45	0.57	0.102	-0.40	0.64	0.317	21.07*	0.24	4.07*	0.35	8.35*	0.11
	>10%	-1.07	1.00		-0.80	0.95		-0.57	0.71							
Contact angle (°)	<10%	61.90	11.89	0.001	63.03	11.47	0.344	66.03	12.78	0.465	46.86*	0.41	3.23	0.22	9.22*	0.12
	>10%	51.55	12.30		60.03	12.93		63.64	12.98							
Frequency (push/min)	<10%	65.06	17.89	0.020	63.00	18.05	0.340	60.99	18.16	0.677	29.53*	0.30	1.96	0.16	6.26*	0.08
	>10%	77.15	21.31		67.50	18.84		62.82	16.85							
Net Work/cycle (J)	<10%	8.89	2.94	0.015	9.09	2.98	0.379	9.43	3.01	0.652	38.60*	0.36	1.69	0.24	9.00*	0.12
	>10%	7.04	2.96		8.38	3.30		9.07	3.16							

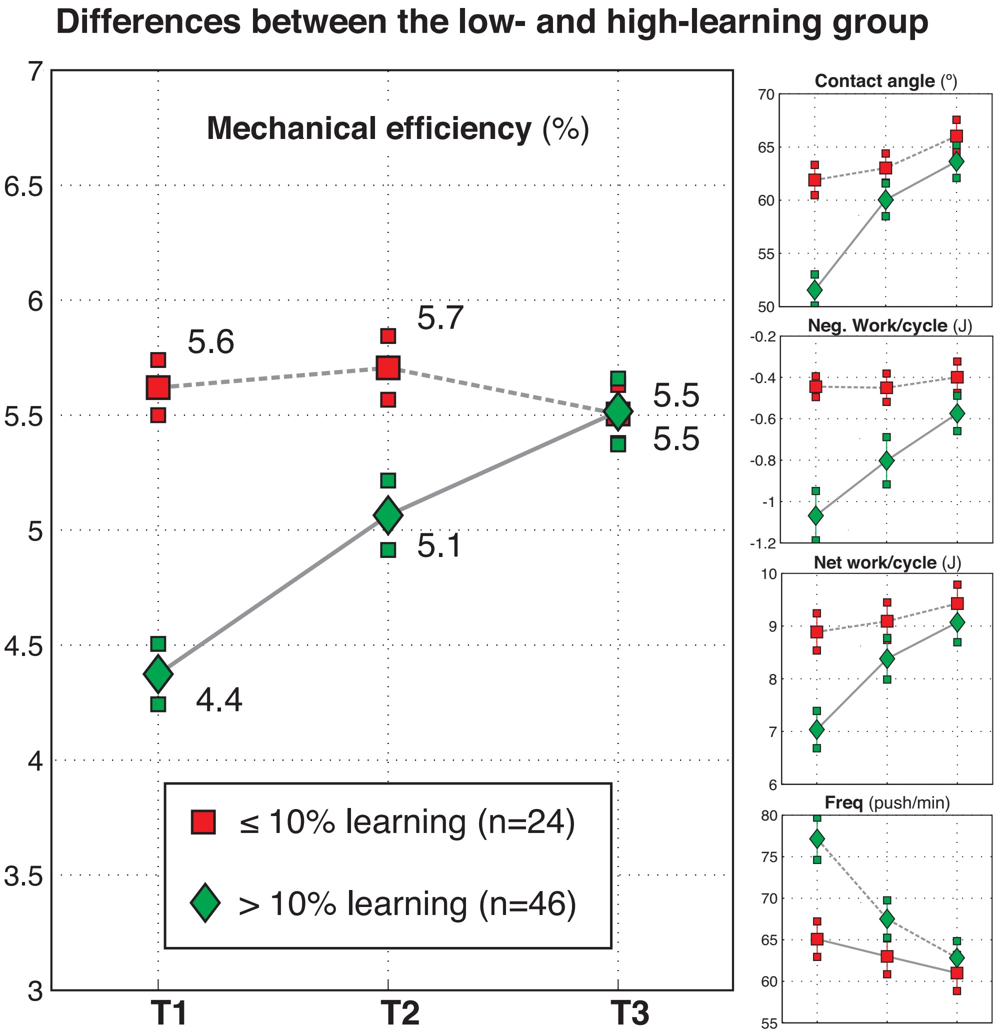

Figure 4.4: The interaction effects of group and practice-blocks for mechanical efficiency and the most important propulsion technique variables. The ≤ 10%-increase group already had a higher mechanical efficiency and a better propulsion technique at the start. The error bars depict the standard error of measurement. T1,T2,T3 on the x-axis represent the 4th 8th and 12th minute of practice in all graphs.

Discussion

Aim of the present study was to evaluate the change in mechanical efficiency and propulsion technique during the initial skill acquisition of a steady-state wheelchair propulsion task, using able-bodied participants. Within the 12 minutes of practice participants learned to deliver the same power output using less energy and concomitantly changed their propulsion technique. Furthermore the increased mechanical efficiency related to the changed propulsion technique of the participants. Finally, it was shown that two different learning groups could be identified, a group that not or only slightly improved their mechanical efficiency and one that improved much more during the three four-minute practice-blocks. The no-improvers already had a higher mechanical efficiency and a better propulsion technique compared to the improving group at the first time of measurement.

Where previously the study of De Groot et al. [25] did not observe a reduction of energy cost, i.e. an increased mechanical efficiency, the current study did find these changes over a very short practice period. Important differences of the current study with that of De Groot et al. [25] is the much larger number of participants (N=70 vs. N=9) and the use of a wheelchair on a treadmill instead of an ergometer. It was hypothesized that the combination of propulsion with steering would make propulsion on the treadmill a more challenging task then pushing on a stationary ergometer. One clear difference that in our view relates to the increased difficulty of the treadmill is the higher push frequency of the participants. Compared to the push frequency on the ergometer (57-53-51 Pushes/min) the frequency on the treadmill was higher (73-66-62 Pushes/min), despite the lower power output (22.5W vs. 17.2W respectively). This is contrary to the findings of a different study with two levels of intensity on the ergometer, which found that a higher power output (0.15-0.25 W) showed a higher push frequency (41.7-46.4 Pushes/min) [16]. Apparently participants propel at a higher frequency on the treadmill to maintain a better control over the directional change of the wheelchair, which has to be aligned with the 1.2m width of the treadmill. Since this extra steering component relies more on control it might be more susceptible to learning processes that reduce and compensate for bilateral asymmetries, explaining the increased learning effects found in the current study.

The larger sample size leads to more statistical power. The group of 70 participants offered a unique opportunity to find group level effects, allowed the use of multi-level analysis within and between subjects and gave the possibility to discriminate between differences in motor learning. The changes in propulsion technique that were found together with the reduced energy cost are discussed in relation to each other below.

At steady-state wheelchair propulsion with a fixed speed of the treadmill, the average power output remains constant. Propulsion technique can change but in the end the average power output must be maintained to keep rolling on the treadmill. Because of this constant power output the propulsion technique parameters are linked to each other and change in one will be reflected in the other.

First power output is performed through the multiplication of work per cycle and the frequency of the pushes [37]. Any reduction of push frequency will have to go along with increased work per cycle and vice versa to maintain the power output necessary at a certain treadmill speed. As expected from earlier work on wheelchair propulsion and motor learning the participants indeed learned a 'longer-slower' movement pattern [16-23]. An increased work per push was associated with a reduced push frequency.

Second, the work per push is the integration of positive torque around the axle over the angle through which it rotates. Any combination of angle and torque can account for the work done within a push. Although this gives a large range of performance possibilities probably some are more suitable to perform in a less straining, more energy efficient way. As expected from the 'longer-slower' movement hypothesis [11] participants learned to use a prolonged trajectory of the hand in contact with the hand rim, resulting in a longer push time. Interestingly this is opposite to the results found in the de Groot et al study, where the increased work per push was attributed to an increase in

4

peak torque and no significant change in push time was found [25].

Third, the increase of contact angle led to a reduction of the slope, the rise of torque per second, which meant that the build up of force became more gradual, possibly reducing stress on the upper extremity and reducing the risk of repetitive strain [38,39]

Finally, participants learned to reduce the amount of negative work during the coupling and decoupling of the hand to the rim. Thus, in combination with the reduced frequency, the amount of (de)couplings reduced in both number and magnitude, leading to less negative work done by the participants. Because the negative work did not have to be compensated with positive work in total less work needs to be done to maintain the same power output.

For a number of propulsion technique variables the coefficient of variation reduced. In our view the reduction of variability in the positive work per push is the most important one since it combines others variables like contact angle and mean and peak forces, of which the coefficient of variation also reduced. The reduced variability between cycles might reflect motor learning, leading to less error within cycles (matching the speed of the treadmill) and possibly less error between left- and right-hand push differences (compensations for directional change).

The above-described changes in propulsion technique theoretically imply a reduction in the energy cost of the user. Using multi–level modeling this relationship was further explored to see which technique changes related most to mechanical efficiency. Both multi-level models indeed showed a relation between propulsion technique and mechanical efficiency. Although this relation was assumed in earlier studies [16-23] the current model results make a further step in understanding the relation between the components of skill of execution and energy cost. In light of the variability in personal characteristics and the fact that the wheelchair was not adapted to the individual anthropometry of each participant the explained variance of 47% by propulsion technique in mechanical efficiency is a meaningful result. The propulsion technique variables that together related significantly to mechanical efficiency were the percentage negative work per cycle and contact angle. Reduced losses because of negative work and a larger contact angle relate to a higher mechanical efficiency as was expected.

For the delta scores the change in propulsion technique predicted 35% of the observed change in mechanical efficiency. Besides the variables percentage negative work per cycle and contact angle, the push frequency and the net work per cycle also contributed to mechanical efficiency. The percentage negative work per cycle and contact angle changed in the same direction as the previous model. The direction of frequency on the other hand is counterintuitive because here also an increase is predicted to contribute positively to mechanical efficiency. However the other changes should already have led to a reduction in frequency, which makes this outcome harder to interpret. Finally an increase in the net work per cycle increases the mechanical efficiency as expected. The change in both models was nearly identical, and therefore we conclude that the relationship between propulsion technique and mechanical efficiency was mainly based on the within-participant variance instead of between-participant variance. This implicates that persons who are able to improve their propulsion technique can expect an improvement in their mechanical efficiency.

To identify different types of learners two groups were formed on the basis of change in mechanical efficiency (>10%) between T1 and T3, and compared on their mechanical efficiency and propulsion technique over all practice-blocks. The improvers, with about 2/3 of the participants started with a lower mechanical efficiency and a less optimal propulsion technique then the non-improvers. Possibly the improving group still had more room to increase in proficiency of the propulsion skill, while the more proficient group at the start, i.e. the low-learning group was already closer to their optimum [40]. Whether the low-increase group learned faster and already had adapted in the 4th minute, or that they had this higher mechanical efficiency from the start cannot be concluded from the present study. How individual differences impact motor learning of a cyclic task like wheelchair propulsion is an important topic for future research so rehabilitation programs can be better tailored to the needs of novice individual wheelchair-users.

Although the clinical relevance lies with people in a wheelchair during early rehabilitation, it was thought necessary to use able-bodied participants to get a better view on technique changes in this early stage of learning a new task, since the results are not confounded by the heterogeneity of wheelchair-users like lesion level or comorbidities after trauma. The current study shows that a better propulsion technique relates to energy cost, which is an important factor in daily life for those with a limited physical capacity [24]. However, the relatively young age of our participants might make inferences for wheelchair-users harder. Furthermore 12 minutes at a fixed speed of 4 km/h at 0.20 W/kg might be too high a load to be a feasible practice method during early rehabilitation especially for those with a tetraplegia [41].

Since only male participants were recruited, we do not know whether the found changes in mechanical efficiency and propulsion technique would be of the same order and magnitude in female participants. We expect similar trends in female participants, yet at relative different levels of timing and kinetics as well as metabolic cost. Overall the motor learning differences found in our relatively homogeneous group of male participants only further stresses the need for more individualized assessment of motor learning, where female participants should also be studied.

Altogether, over the course of 12 minutes of wheelchair propulsion participants learned a more favorable push strategy. It is an important finding that participants, without getting any specific feedback or modified training program already find a more optimal wheelchair propulsion technique during the initial minutes of practice. This further supports the view of Sparrow and Newell that the human system is continuously in search of the most energy efficient solution within the constraints of the task [11]. The observed transition to a longer-slower movement pattern found in other cyclical motions is also observed as a consequence of motor learning in hand rim wheelchair propulsion over this short practice period. Future research should take these early learning adaptations into account when evaluating different interventions on motor learning over longer timescales.

Conclusion

Over the first 12 minutes of practice naive able-bodied participants increased their mechanical efficiency and changed their propulsion technique. The propulsion technique of the participants changed from a high frequency mode with a lot of negative work to a longer-slower movement pattern with less power loss. This change in propulsion technique related to an increased mechanical efficiency of the participants and thus a lower physical strain. These findings link propulsion technique to mechanical efficiency supporting the importance of a correct propulsion technique for wheelchair users. Furthermore differences in baseline efficiency and propulsion technique were shown to affect the motor learning process. Individual motor learning differences are important to take into account for rehabilitation programs.

Endnotes:
a Forcelink b.v, Culemborg The Netherlands
b Double Performance BV, Gouda, The Netherlands
c Three Rivers Holdings, Mesa, AZ, USA
d MAX Mobility, LLC, Antioch, TN, USA
e Oxycon Pro-Delta, Jaeger, Hoechberg, Germany

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chapter 5

Inter-individual differences in the initial 80 minutes of motor learning of handrim wheelchair propulsion.

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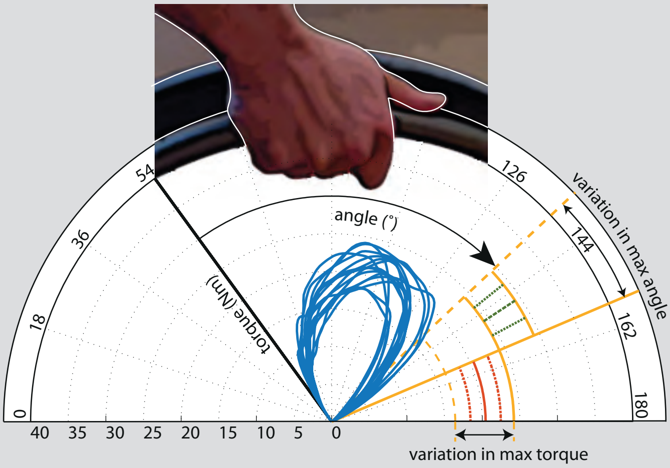

Abstract

Handrim wheelchair propulsion is a cyclic skill that needs to be learned during rehabilitation. Yet it is unclear, how inter-individual differences in motor learning impact wheelchair propulsion practice. Therefore we studied how early-identified motor learning styles in novice able-bodied participants impact the outcome of a low-intensity wheelchair-practice intervention. Over a 12-minute pre-test 39 participants were split in two groups based on a relative 10% increase in mechanical efficiency. Following the pretest the participants continued one of four different low-intensity wheelchair practice interventions, yet all performed in the same trial-setup with a total 80-minute dose at 1.11 m/s at 0.20 W/kg. Instead of focusing on the effect of the different interventions, we focused on differences in motor learning between participants over the intervention. Twenty-six participants started the pretest with a lower mechanical efficiency and a less optimal propulsion technique, but showed a fast improvement during the first 12 minutes and this effect continued over the 80 minutes of practice. Eventually these initially fast improvers benefitted more from the given practice indicated by a better propulsion technique (like reduced frequency and increased stroke angle) and a higher mechanical efficiency. The initially fast improvers also had a higher intra-individual variability in the pre and posttest, which possibly relates to the increased motor learning of the initially fast improvers. Further exploration of the common characteristics of different types of learners will help to better tailor rehabilitation to the needs of wheelchair-dependent persons and improve our understanding of cyclic motor learning processes.

Introduction

Handrim wheelchair propulsion is a cyclic bimanual form of ambulation that needs to be learned during early rehabilitation by people with a lower-limb disability. Compared to other forms of ambulation the gross mechanical efficiency of handrim propulsion, i.e. the ratio of external power output over internal power production is low, while at the same time overuse problems are common [1-5]. Yet, different intervention studies have shown that, through low-intensity practice both mechanical efficiency and propulsion technique of handrim wheelchair propulsion can improve, possibly reducing the load on the wheelchair user [6-13]. However, it is unknown how inter-individual differences in motor learning impact the outcomes of wheelchair propulsion practice in such an early stage.

Within the rehabilitation environment, using the International Classification of Functioning, Disability & Health (ICF) framework, there is appreciation for inter-individual differences in outcomes of health and disability [14]. An important domain in this framework is 'personal factors' such as age, gender, physical ability, self-efficacy and motivational level [15]. Other important personal factors related to motor learning are trainability and talent, i.e. the individual response to exercise [16,17] and the ability to adopt and optimize motor skills [18,19]. For instance, inter-individual differences were found in the effect of regular physical activity on maximal oxygen consumption, submaximal heart rate response, cholesterol and systolic blood pressure [20]. Correlations of these variables with age, gender or ethnic background were low. In contrast, baseline values of heart rate and blood pressure strongly correlated with the effect of the intervention. Individuals with higher baseline values and thus a worse physical condition showed larger reductions in heart rate and blood pressure due to training [20].

Analogous to exercise programs that focus on improving physical capacity, low-intensity practice sessions aim to improve the motor skill of individuals. On a group level it has been shown that inexperienced individuals improve their wheelchair propulsion skills through practice [6-13]. Yet, this improvement over the group may not fully apply to each member of that group. Although there is increasing evidence of inter-individual differences in learning a new motor task, this notion is still rarely assessed [18,21-24].

Not only between, but also within individuals, human movement is intrinsically variable [25,26]. This intra-individual movement variability can for instance be found between limbs performing the same action (i.e. interlimb variability), or in one limb repeating a cyclic movement over time (i.e. intralimb variability). Such variability is assumed to not only be the reflection of noise and/or error, but also to be functional and to contain features that may provide insight in motor learning [27-29] and pathological processes [30-34]. From this perspective, intra-individual variability is seen as a mechanism allowing individuals to adapt their movements as a function of organismic, environmental and task constraints [35,36]. Variability allows the performer to explore different motor solutions, facilitating the discovery and adoption of individualized optimal patterns of coordination, possibly reducing the energetic cost [37]. In the current study, changes in the intra-individual variation in learning wheelchair propulsion are studied based on the coefficient of variation (CV) defined as the percentage standard deviation of the mean of a given technique parameter.

Because of several unique features, the study of handrim wheelchair propulsion is suitable to gain insight into inter- and intra-individual differences in early motor learning processes of cyclic motor tasks in novice able-bodied individuals. Firstly, wheelchair propulsion is cyclic, which makes it possible to evaluate steady-state submaximal performance using energy consumption and thus mechanical efficiency as a generic outcome measure of motor learning [38]. Secondly, the movement is sufficiently unconstrained to allow for performance of the task in different ways, allowing propulsion technique to change between the left and right wheel and over time within one side [39,40]. Finally, for most people, wheelchair propulsion is a new task. Therefore, in the study of motor learning, learning wheelchair propulsion is highly suitable as a model to study the initial acquisition of a cyclic skill. Wheelchair skill acquisition in early rehabilitation can well be studied with able-bodied participants, thus reducing heterogeneity within the participant group, which might be expected in for instance a group of participants with a spinal cord injury due to the level and completeness of the lesion, health history or upper-body asymmetries beyond age, gender and training status [41]. On another note, researchers do not have to burden patients early on as they are learning to cope with the far-reaching effects of a new SCI.

In our previous work, early inter-individual motor learning differences were found in 70 novice able-bodied wheelchair users [42]. Two different groups were formed based on a relative 10% increase in mechanical efficiency between the 4th and 12th minute of practice. The Initially Slow Improvers (ISI) already demonstrated a significantly higher mechanical efficiency and more skilled propulsion technique at the first steady-state measurement (the 4th minute) compared to the Initially Fast Improvers (IFI). However, the ISI did not further increase in proficiency in the next 8 minutes, whereas the IFI, despite starting at a lower level of mechanical efficiency, were able to improve in mechanical efficiency each next trial. After 12 min of practice the groups showed a similar absolute level of mechanical efficiency [42].

For rehabilitation it is important to know how these short-term inter-individual differences in motor learning impact the outcome of an intervention over a longer timescale. From the 70 participants in the previously discussed twelve-minute study, 39 continued in four different low-intensity interventions. Instead of focusing on the effect of the different intervention types, the main aim of the current study is to follow the two designated motor learning groups (ISI / IFI) over time, to find out whether their initially different motor learning styles still differed after 80 min practice.

The research question of the current study is therefore: how do early-identified motor learning styles among two different groups of able-bodied novice participants impact the outcome of an 80 min low-intensity wheelchair-practice intervention? The early motor learning differences will again be assessed during the 12-minute pretest based on a relative increase of either less or equal to 10% or higher than 10% in gross mechanical efficiency [42]. The two identified groups will then be analyzed over the pre -and post-test to see how the early differences between the groups impact the eventual intervention outcome.

It is hypothesized from earlier work [42] that the same types of early differences in motor learning between individuals will be present over the follow-up period. Also, it is hypothesized that the mean outcomes of both groups shall differ in the coefficient of variation, showing differences in the variability of task execution between the groups. These initial motor learning differences are expected to impact the final outcome of the intervention, where those participants that learn more in the pretest are expected to be the ones who benefit most from the given practice [20] .

Table 5.1: Personal characteristics of the four practice groups: One–day monotonous (ODM), three-week monotonous (TWM), one-day seat-height (ODS) and one-day feedback (ODF).

Study	Duration	Dose:	Test-nature	Age (yrs)	Height (m)	Weight (kg)	n-total	n-IFI	n-ISI
ODM	1-day	80 min	Monotonous	22.0 (2.0)	1.89 (0.11)	81.1 (19.4)	10	8	2
TWM	3-week	80 min	Monotonous	22.8 (3.9)	1.89 (0.07)	83.8 (11.6)	13	8	5
ODS	1-day	80 min	Variable (seat height)	23.3 (2.6)	1.86 (0.07)	80.4 (14.0)	10	6	4
ODF	1-day	80 min	Variable (feedback)	23.3 (4.1)	1.84 (0.04)	74.8 (6.1)	6	4	2
total				22.9 (3.2)	1.87 (0.07)	80.0 (12.8)	39	26	13

Methods

5.1 Participants

After written informed consent was provided, 39 able-bodied men spread over four experimental groups fulfilled our criteria for participation in this study (table 5.1). To compare our results with previous research the criteria for inclusion were male, between 18-65 years, no prior experience in wheelchair propulsion, and absence of any medical contra-indications [6,8,9,43,44].. The study was approved by the Local Ethics Committee, of the Center for Human Movement Sciences, University Medical Center Groningen, University of Groningen, the Netherlands.

5.2 Protocol

Each of the 39 participants were involved in one of four intervention formats. The four wheelchair interventions were different in nature (table 5.1), but were performed in the same experimental and trial-setup (figure 5.1) and had the same dose of 80 min propulsion at a relative power output of 0.20 W/kg. Although the four intervention studies had a common design, allowing the combination of the data at a more global level (see Statistics), each intervention had their own question beyond the main aim of the present study (manuscripts under preparation). The low intensity was chosen to minimize fatigue or training effects and focus primarily on motor learning. The first key difference between the interventions was the time-scale over which the 80 min practice was performed; the participants either participated in a single-day or a three-week experiment. During the single-day experiment the intervention shown in figure 1 was completed in one continuous experiment with 30 min rest between each 8-min practice session, whereas during the three-week experiment each 8-min practice session was separated by 48 hours. The second key difference was practice variation. One single-

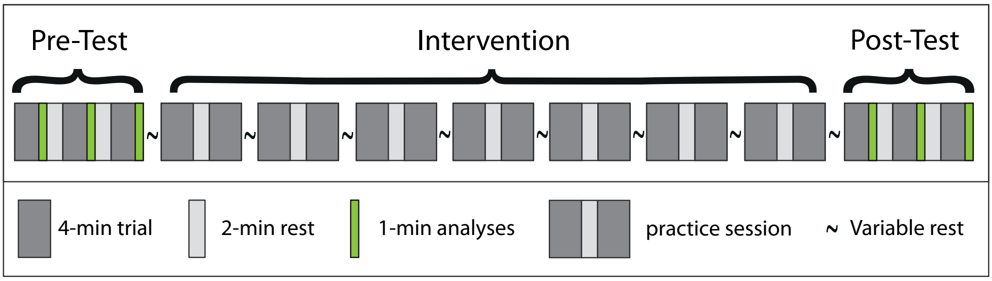

Figure 5.1: Setup of practice. A practice session consists of two 4-min trials separated by 2-min rest at 0.20 W/kg. The pre and post-test had one trial extra and the last minute of each trial was used for analyses. The time between practice sessions (variable rest) was either 30 min or 48 hr.

day group (ODM) and one three-week group (TWM) trained monotonously during the intervention. Two other single-day studies trained with variations. Participants in the first study practiced with four different absolute seat-heights as provided by the experimental wheelchair. The seat-height counterbalanced over the 7 blocks of the intervention (ODS). The participants of the second single-day study received real-time feedback (ODF) on seven propulsion technique variables, also in a counterbalanced order, earlier described by Richter et al. [45]. These seven propulsion technique characteristics were individually presented as a bar graph on a monitor in each of the seven practice blocks. Participants were free to use this feedback, but never got any specific instruction on how to manipulate any of these parameters. Thus for all groups technique improvements over time are assumed to have occurred as a function of practice. Eventually, n=26 participants were identified as IFI and n=13 were ISI.

5.3 Experimental setup

All trials were performed on a level treadmill of 2.4 m length by 1.2 m width (Forcelink[a]) in the same experimental wheelchair (Double Performance[b]) with 24-inch measurement wheels. Each participant practiced according to the schedule presented in figure 1. The first 40 seconds of a trial were used to get the treadmill up to a speed of 1.11 m/s (4 km/h). The required power output to get to 0.20 W/kg was imposed by adding mass to a pulley system. For each participant a drag test was performed prior to the start of the experiment. Based on the calculated drag force of the wheelchair-user combination at the required constant speed of the treadmill (1.11 m/s) and the participants' body mass, the added mass to the pulley was calculated [4,46]. For data analysis the last minute of each trial in the pre-test and post-test were used (i.e. trial 1, 2, 3 and 18, 19, 20).

5.4 Measurement wheels

One standardized experimental wheelchair was used and no individual adjustments were made for individual participants. The regular rear wheels of the standardized wheelchair were replaced with two instrumented wheels; on the left the Optipush[c] (Max Mobility) and on the right the Smartwheel[d] (3-Rivers). Both wheels measure 3-dimensional forces and torques applied to the handrim, combined with the angle under which the wheel is rotated. Data were wirelessly transferred to a laptop at 200 Hz (Optipush) and 240 Hz (Smartwheel). Both wheels were synchronized by an electronic pulse at the start of each measurement [40]. Data from the Optipush were primarily used in the analyses, only when the Optipush data were lacking they were replaced with Smartwheel data after mirroring those data. Time averaged data of both wheels attached to the left and right side of the same wheelchair placed on a treadmill showed good comparability, with an intra-class correlation (ICC) of 0.89 for mean power output and ICC's higher than 0.90 for propulsion technique characteristics [40]. Therefore, the time averaged outcomes of the left and right wheel in this experiment were assumed to be comparable.

5.5 *Energy expenditure*

Oxygen consumption (VO2) was continuously measured during each practice session using breath-by-breath open circuit spirometry[e]. The gas analyzer was calibrated using a Jaeger 5l syringe, room air and a calibration gas mixture. Data collected over the fourth minute of each exercise trial were averaged and taken to reflect physiological steady-state wheelchair propulsion. From the VO2 (L/min), VCO2 (L/min) and respiratory exchange ratio (VCO2/VO2) the energy expenditure was determined using the formula proposed by Garby and Astrup [47].

5.6 *Data analysis*

The data from the instrumented wheels were further analyzed using custom-written Matlab routines. To be certain of stable, steady-state propulsion, each last minute from the 4-min trials was used for the analysis. Per participant and trial, the torque (Nm) around the wheel-axle and the rotation angle (rad) were used to calculate the propulsion technique variables of interest. Individual pushes were defined as each period of continuous positive torque around the wheel axis with a positive minimum of at least 1 Nm. Over the identified pushes the propulsion technique variables were calculated and subsequently averaged over all pushes within the fourth minute of each practice trial per participant. The studied propulsion technique variables are defined in table 5.2 and figure 2. They were chosen based on their previously found association with mechanical efficiency (frequency, contact angle and negative work per cycle [42]) and two other variables were added because variability in them was expected to change (positive work per push and max torque/push (figure 5.2)).

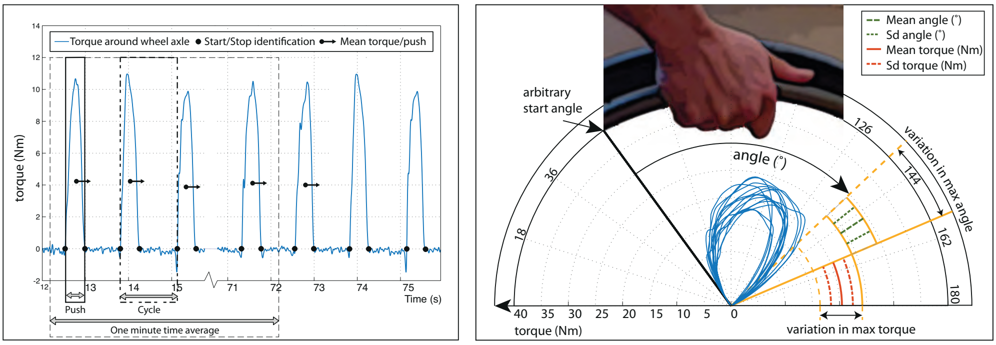

Figure 5.2: Two visualizations of propulsion kinetics. a) Time history of the torque signal showing the push identification, push-time, cycle-time, work per push, and mean torque. b) Alternative Polar plot of the torque against the angle for 12 pushes, showing the intra-individual variation in contact angle and maximum torque. Since no position data were recorded each push is started from the same arbitrary angle.

5

Table 5.2: Definitions of propulsion technique variables

Variable:	Description:	Equation:
Energy expenditure (W)	Calculated from the oxygen uptake and respiratory exchange ratio according to Garby and Astrup [47]	((4.94*RER+16.04)*(1000*VO2))/60
Gross mechanical efficiency (%)	The percentage of internal power used for external power delivered at the wheels	Mean power output/Energy expenditure
Frequency (push/min)	The number of complete pushes per minute.	$(\text{Cycletime})^{-1} \cdot 60$
Contact angle (°)	Angle at the end of a push minus the angle at the start.	$Ø_{end(i)} - Ø_{start(i)}$
Max Torque/push (Nm)	The maximum torque generated around the wheel-axle within a push	$\text{Max}_{start(i):end(i)}(Tz)$
Pos. Work/push (J)	The torque around the wheel-axle integrated over the contact angle of the push.	$\sum_{start(i):end(i)} (Tz \cdot \Delta Ø)$
Neg. Work/cycle (J)	The torque around the wheel-axle integrated over the wheel rotation angle during the recovery phase	$\sum_{end(i):start(i+1)} (Tz \cdot \Delta Ø)$

Abbrevations: $_{start}$(i), start of the current push (sample); $_{end}$(i), end of the current push (sample); Ø, angle (rad); Tz, torque around wheel axle (Nm).

5.7 *Statistics*

Two groups (ISI and IFI), were formed based on a higher or lower than 10% relative increase in mechanical efficiency between the first and the third 4-min trial in the pretest, common for all interventions [42]. To replicate the results of the previous study for this smaller subset of participants the initial 12 minutes were pre-analyzed. Differences between the groups during the 12 min pretest were assessed on mechanical efficiency and propulsion technique using a repeated-measures Anova with the factors time, group and the interaction of time*group. Since not only the propulsion technique values, but also the variation therein is of interest, this process was repeated for the coefficient of variation, i.e. the percentage of standard deviation with respect to the mean. Significance level was set at $p < 0.05$ for all statistical procedures.

Analysis of the inter-individual differences between the pre- and the posttest for the different learning trajectory groups was the main aim of this paper. To control for the different intervention types multi-level modeling was applied, [48]. The differences between the ISI and IFI were examined over all trials of the pre- and post-test to evaluate whether they were differently influenced by the longer practice period (i.e. an interaction effect of test*group). To correct for the different natures of the four interventions two extra terms were added to the model, namely 'Duration' (1-day or 3-wk) and 'Variation' (monotonous or variable). The model thus consisted of five terms: Test (pre=0, post=1), Learning group (ISI=0, IFI=1), Test*Learning group interaction, Duration (1day=0, 3wk=1) and Variation (no=0, yes=1). This model was applied to the dependent variables mechanical efficiency and selected propulsion technique variables (see table 2), as well as to the accompanying coefficient of variation of these outcome variables.

Results

All participants were able to complete the protocol. The Optipush data (left side) were used for 35 participants and Smartwheel data (right side) were used for the other 4 participants. On average participants practiced at a power output of 17.6W (s.d.= 4.2).

The differences between the ISI and IFI are presented for the first twelve minutes (repeated measures Anova) in table 5.3 and for the total 80 minutes (multi-level regression) in table 5.4. Changes in mechanical efficiency, propulsion technique and intra-individual variability for both groups over the first 12 and total 80 minutes are described below.

Table 5.3: Results of the repeated measures Anova on the three trials of the 12 min pretest (n=39). P-values < .05 are interpreted as statistically significant results.

Values pre-test Anova		mean 1	sd 1	mean 2	sd 2	mean 3	sd 3	p-Time	p-Group	p-Interaction
Mechanical Efficiency (%)	ISI	5.45	0.81	5.53	1.06	5.39	1.01	0.000	0.075	0.004
	IFI	4.38	0.94	4.99	1.03	5.25	1.20			
Frequency (/min)	ISI	68.22	13.30	69.27	12.59	68.42	14.96	0.000	0.910	0.002
	IFI	75.41	18.37	66.99	17.96	61.71	17.65			
Frequency CV (%)	ISI	8.81	4.87	7.51	4.53	5.94	3.02	0.075	0.010	0.918
	IFI	12.30	7.10	11.19	5.06	10.26	5.42			
Neg work/cycle (J)	ISI	−0.51	0.45	−0.50	0.62	−0.45	0.75	0.000	0.226	0.039
	IFI	−1.09	1.08	−0.84	1.06	−0.60	0.82			
Neg work/cycle CV (%)	ISI	116.07	175.98	78.17	60.21	82.90	50.02	0.800	0.892	0.036
	IFI	68.71	46.23	97.20	78.72	100.85	86.46			
Contactangle (°)	ISI	60.73	10.89	59.59	9.17	60.73	10.89	0.000	0.804	0.003
	IFI	53.29	10.89	61.31	12.61	63.60	13.18			
Contactangle CV (%)	ISI	12.72	4.73	9.85	1.80	9.02	1.89	0.000	0.008	0.784
	IFI	14.78	4.00	12.90	4.73	11.21	2.32			
max Tz/push(Nm)	ISI	13.84	2.59	13.55	2.37	13.58	2.62	0.526	0.825	0.933
	IFI	14.02	3.40	13.88	3.07	13.72	3.03			
max Tz/push CV (%)	ISI	18.37	3.10	16.25	2.95	15.37	3.23	0.000	0.004	0.600
	IFI	21.65	4.98	20.01	3.83	17.71	3.45			
work/push (J)	ISI	9.02	2.57	8.79	2.24	8.99	2.41	0.016	0.975	0.046
	IFI	8.13	2.72	9.22	2.99	9.37	2.88			
work/push CV (%)	ISI	21.96	4.47	19.09	2.75	17.58	3.98	0.000	0.004	0.435
	IFI	26.63	5.78	23.82	5.51	20.22	4.69			

P-values <.05 are interpreted as statistically significant results.

5.8 *Gross mechanical efficiency*

5.8.1 First 12 minutes of practice

Based on a 10% relative change in mechanical efficiency out of the 39 participants 13 were classified as ISI and 26 as IFI. Concomitant with this selection an interaction effect was found between the two groups based on the repeated measures Anova on the pretest, where the ISI already had a higher mechanical efficiency in the first 4-min trial than IFI (ISI 5.5% vs. IFI 4.4%, $p<0.002$).

5.8.2 Total 80 minutes of practice

Over the whole 80-minute intervention the interaction effect on mechanical efficiency remained consistent between the two groups over time (figure 5.3). Based on the multilevel regression analysis and controlling for the nature of the intervention the IFI, despite starting lower in the pretest, benefitted more from the intervention and had a significantly higher mechanical efficiency compared to the ISI at the posttest (ISI 5.5-5.5% vs. IFI 4.9-5.9%, $p<0.001$).

5.9 *Propulsion technique*

5.9.1 First 12 minutes of practice

Similar to the mechanical efficiency also an interaction effect was found during the pretest for the propulsion technique variables frequency, contact angle, work per push and negative work per cycle. For each of these variables the ISI had a significant better outcome in the first 4-min trial than the IFI but did not further improve in technique over the next 8 minutes. For the maximum torque per push no significant effect of trial, group or interaction within the two groups was found.

Table 5.4: Multi-level regression results for the 80-minute intervention for both the propulsion technique variables and their coefficient of variation (n=39). S.e. is the standard error of the multi-level model result. P-values < .05 are interpreted as statistically significant results.

Values intervention MLWin		Cons	Time	P-Time	Group	P-Group	Interaction	P-Interaction	Duration	P-Duration	Variation	P-Variation
Mechanical Efficiency (%)	result	5.48	0.04	0.79	−0.55	0.04	0.90	0.00	0.58	0.06	−0.52	0.07
	s.e.	0.30	0.17		0.27		0.21		0.31		0.29	
Frequency (/min)	result	67.25	−15.19	0.00	0.82	0.85	−4.28	0.19	4.72	0.32	−0.94	0.83
	s.e.	4.74	2.60		4.15		3.26		4.79		4.45	
Frequency CV (%)	result	6.90	3.23	0.01	3.84	0.02	−3.50	0.01	−1.32	0.47	2.23	0.19
	s.e.	1.84	1.26		1.65		1.58		1.81		1.68	
Neg work/cycle (J)	result	−0.43	0.34	0.01	−0.39	0.02	0.41	0.01	−0.23	0.24	0.06	0.73
	s.e.	0.19	0.11		0.17		0.14		0.19		0.18	
Neg work/cycle CV (%)	result	91.54	142.40	0.00	−11.89	0.78	−105.48	0.04	−34.87	0.40	30.88	0.42
	s.e.	44.03	41.04		41.45		51.34		40.88		37.97	
Contactangle (°)	result	62.45	7.22	0.00	−1.43	0.67	7.96	0.01	−0.63	0.87	−4.07	0.25
	s.e.	3.78	2.12		3.32		2.69		3.78		3.50	
Contactangle CV (%)	result	10.14	−0.77	0.39	2.54	0.01	−1.88	0.08	−1.49	0.08	2.10	0.01
	s.e.	0.93	0.89		0.88		1.11		0.86		0.80	
max Tz/push(Nm)	result	11.87	0.94	0.02	0.76	0.39	−0.54	0.29	3.11	0.01	1.27	0.20
	s.e.	1.03	0.41		0.88		0.51		1.06		0.99	
max Tz/push CV (%)	result	16.36	−2.86	0.00	3.17	0.00	−1.21	0.26	−1.49	0.16	1.90	0.06
	s.e.	1.10	0.85		1.00		1.07		1.07		0.99	
work/push (J)	result	8.05	1.90	0.00	0.22	0.82	0.65	0.29	1.90	0.10	0.33	0.76
	s.e.	1.13	0.49		0.97		0.61		1.16		1.07	
work/push CV (%)	result	18.62	−3.26	0.01	4.26	0.00	−2.86	0.05	−1.67	0.23	3.39	0.01
	s.e.	1.44	1.17		1.32		1.46		1.38		1.28	

S.e. is the standard error of the multi-level model result. P-values <.05 are interpreted as statistically significant results.

5.9.2 Total 80 minutes of practice

For all propulsion technique parameters a significant effect of 'Test' was shown (figure 5.4). During the posttest participants of both groups had decreased their push frequency, reduced their amount of negative work, increased their contact angle, increased their maximum torque and finally increased their work per push.

Over the 80-minute intervention and after controlling for the nature of the intervention an interaction effect for 'Test'*'Learning Group' was only found for the contact angle and the negative work per cycle (figure 5.4). The IFI increased significantly more in contact angle than the ISI and had a larger contact angle in the posttest (ISI 62.5º-69.7º vs. IFI 61º-76.2º, p<0.01). The IFI decreased significantly more in the negative work per cycle than the ISI (ISI 0.43J-0.09 J vs. IFI 0.82 J-0.07 J, p<0.01).

5.10 Intra-individual variability

First 12 minutes of practice

During the pretest the IFI had a higher coefficient of variation for the frequency, contact angle, maximum torque and the work per push compared to the ISI (i.e. Anova effect of group). The coefficient of variation for the negative work per cycle showed an interaction effect; the ISI decreased, while the IFI increased in intra-individual variability.

5.11 Total 80 minutes of practice

A significant reduction in the coefficient of variations of frequency, maximum torque per push and work per push was shown for all participants in the posttest, i.e. a significant of 'Test' (figure 5.5). For the negative work the coefficient of variations significantly increased for all participants in the posttest.

Over the whole intervention a group effect was found for the coefficient of vari-

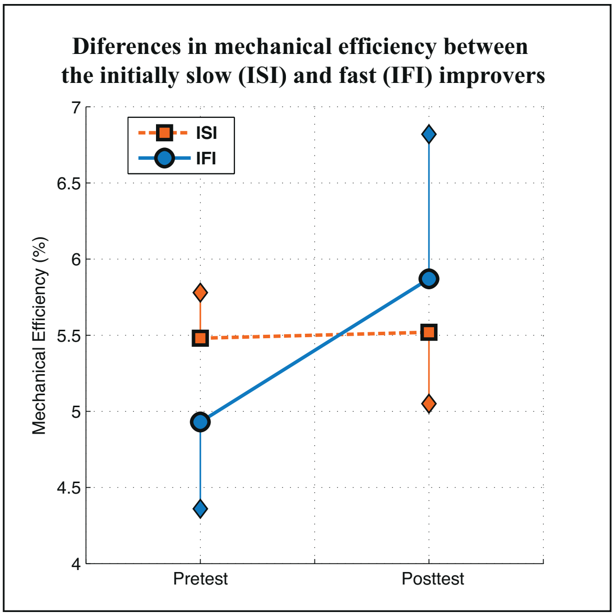

Figure 5.3: Effect of 80 minute practice time on the mechanical efficiency (mean and standard-error of multi-level model) for both groups between the pre- and posttest. While controlling for the nature of the intervention the n=26 IFI (>10%) started with a lower mechanical efficiency but benefitted more from the intervention and had a higher mechanical efficiency in the posttest compared to the n=13 ISI (≤10%).

ations of the variables frequency, contact angle, maximum torque per push and work per push, where the IFI had higher coefficient of variations compared to ISI. An interaction effect was found for the coefficient of variations of frequency, negative work per cycle and positive work per push, but not all in the same direction. For frequency and negative work per cycle it were the ISI that increased more in the coefficient of variations, while for the work per push it were the IFI that decreased more.

Discussion

Aim of the present study was to evaluate differences between individuals in learning low-intensity steady-state wheelchair propulsion on a motor-driven treadmill. Therefore two groups of learners were first identified, based on a higher (IFI) or lower (ISI) than 10% relative increase in mechanical efficiency, during the first twelve minutes of practice. Concomitant with this pretest difference in mechanical efficiency the ISI and IFI also differed in the change of propulsion technique and intra-individual variation during the first 12 minutes of practice. Over the total 80 minutes of low-intensity wheelchair-practice the two groups maintained different motor learning styles. Despite starting at a lower mechanical efficiency during the first minutes of practice, the IFI benefitted most of the given practice in terms of increased mechanical efficiency and better propulsion technique like an increased contact angle and reduced negative work [49].

Increased mechanical efficiency following practice is frequently found and thought to be indicative of motor learning [37,38,50]. Most of these studies have assessed motor learning by studying a single group as a whole. However, an indication for individual differences in the initial mechanical efficiency and change thereof was found in an earlier study, only analyzing the first 12 min of practice [42] . In the current study the effects of

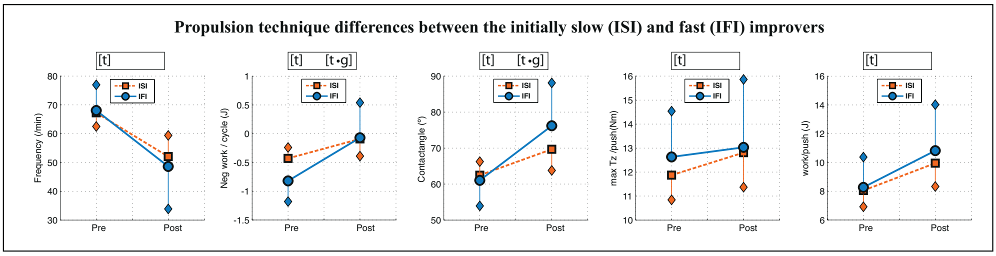

Figure 5.4: Effect of 80-minute practice on the propulsion technique (mean and standard-error of multi-level model) for the initially slow learners and the initially fast learners between the pre- and posttest. Contact angle and negative work per cycle showed an interaction effect for the groups over time. [t]=effect of time, [g]=effect of group, [t•g]=interaction effect of time with group.

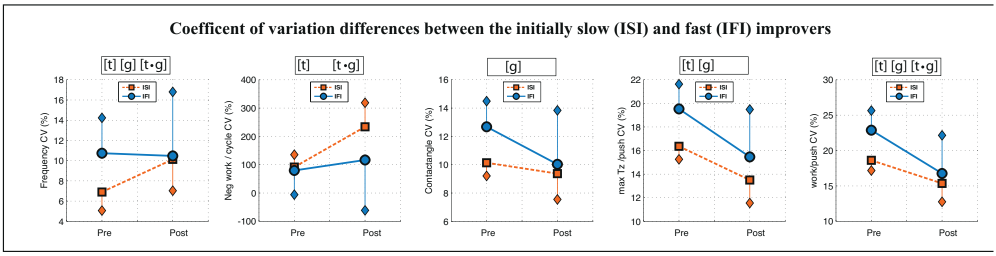

Figure 5.5: Effect of 80-minute practice on the intra-individual variability of the propulsion technique parameters (mean and standard-error of multi-level model) for the initially slow learners and the initially fast learners between the pre- and posttest. The frequency, contact angle, maximum torque per push and work per push showed an interaction effect for the groups over time. [t]=effect of time, [g]=effect of group, [t•g]=interaction effect of time with group

extended practice were studied, taking into account the individual differences in learning. The results indicate that the group of participants (IFI) that increased more in mechanical efficiency on a short term (during the pretest) also increased more over the long term, implying differences in the motor learning process between the two groups. Since all the interventions were low in intensity and total practice time, the changes in mechanical efficiency are presumably attributed to a changed propulsion technique instead of physiological adaptations expected from an extended high intensity dose [51].

The ISI started with better scores for the propulsion technique parameters, i.e. a larger contact angle, more work per push and less negative work than the IFI [49]. Yet, the IFI changed more in these parameters and in the twelfth minute they were on the same level as the ISI. For two variables, the contact angle and the negative work per cycle, this effect continued over the 80 minutes practice period. The contact angle of the IFI increased more and was higher in the posttest compared to the ISI. Since the work per push is the integration of positive torque around the axle over the angle through which it rotates, using a larger contact angle helps to increase the work per push and might help reduce peak forces and make the build up of force more gradual, possibly decreasing the risk of overuse injury [49,52-54]. The IFI also reduced more in the negative work per cycle than the ISI. Because this negative work did not have to be compensated with positive work, in total less work is needed to maintain the same power output. As found in previous wheelchair learning studies, an effect of time was present for all propulsion technique variables, showing the effect of motor learning on propulsion technique for both learning groups [6-13].

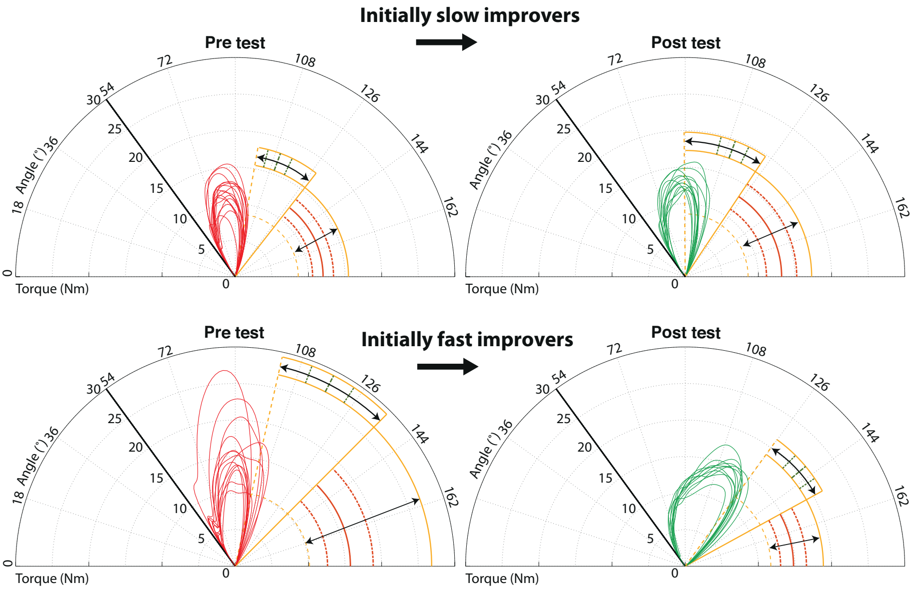

Figure 5.6: The first and last trial of a typical participant for both groups. The push-curves of the torque against the angle show the intra-individual variation in contact angle and maximum torque within a trial. The change over time because of practice is shown by the push-curves over time between the pre and posttest. Since no position data were recorded each push is started from the same arbitrary angle.

Besides the means of the propulsion technique parameters, also the intra-individual variation in these parameters was studied. It was found that for all propulsion technique parameters the IFI had a significantly higher intra-individual variability during the 12-minute pretest than the ISI. Over the 80-minute practice the IFI continued to be more variable in frequency, contact angle, maximum torque per push and work per push. Possibly the IFI were more active in exploring different motor solutions, to find a more optimal pattern of coordination [28,36,37]. Besides the differences in intra-individual variability between the learning groups, a reduction in the intra-individual variation for both groups over time was found for the maximum torque and work per push. Contrary to our expectations, the reduction in intra-individual variation was not shown for the frequency, which would have been expected on basis of the decreased variability in work per push, since these two together should lead to an average constant power output over time in each trial, as required by the constant speed of the motor driven treadmill.

An earlier study on motor learning with the same practice dose and trial set up, but performed on a wheelchair ergometer, did not find reductions in the coefficient of variation of different propulsion technique variables [6]. Possibly, the higher freedom with the continued need to maintain a straight course and a mean fixed speed on the treadmill introduces extra elements to the learning task, which can be minimized over time [40].

To illustrate the total change in propulsion technique, figure 5.6 shows the first and last trial of one typical participant for both groups. During the last 15 seconds for each push the torque around the wheel-axis is plotted against the wheel-angle in a polar plot. The difference in motor learning between the two participants can be visualized by both the amount of variation in the push-curves within a trial and the change of the push-curves over time. During the pretest the variation in peak torque and contact angle is much larger for the initially fast improver. Over the intervention the change of the shapes between the pre- and the posttest is much larger for the initially fast improver compared to the change of the initially slow improver. The post-test propulsion technique of the initially fast improver shows a larger contact angle and a much more gradual build up of torque than the initially slow improver, implying a more optimal propulsion technique [54].

Our findings suggest motor learning differences between able-bodied individuals regarding the acquisition of a low-intensity steady-state wheelchair propulsion skill. For rehabilitation practice it is important to appreciate that these motor learning differences between individuals exist, beside those differences caused by an individual's specific impairment. Ideally, exercise programs with a focus on improving skill should be individually tailored to the motor learning style and capacity of the participants. Such a program may be beneficial to reduce external and internal mechanical loading of the upper limbs [55,56], next to increasing the mechanical efficiency. Thus, the task load of handrim propulsion might be reduced and overuse injury may be prevented during early rehabilitation. More specific focus on motor learning is therefore necessary during the early rehabilitation of actual wheelchair–dependent persons, to further improve their rehabilitation outcomes.

In that sense the higher intra-individual variability found in the IFI gives some insight into the differences in motor learning strategy between the two groups. Further research on the link of inter-individual differences in intra-individual variation with motor learning processes might help to design more individualized and efficient rehabilitation programs. There is increasing evidence for an association between intra-individual variation and overuse injury [57]. A recent study showed that wheelchair-users with shoulder pain showed a lower intra-individual variability in peak resultant forces of the shoulder joint [58]. Possibly, the ISI in our study, showing a lower intra-individual variability over the 80 minutes of practice, are at a higher risk of developing overuse injury than the IFI. Thus, it may be beneficial from both a motor learning and an injury prevention perspective to develop interventions that try to elicit more intra-individual variation from the participants. In that sense the control variable Variation showed a significant increase in the coefficient of variation in contact angle and work/push, giving a possible direction for future research on increasing intra-individual variation.

Several limitations should be taken into account when interpreting the results of the current study. First, the different interventions were not originally intended to discriminate between the two learning groups, but were focused on other motor learning related research questions. Using a multi-level model we have tried to correct for practice variability and total duration to make a comparison between the different possible interventions. Fortunately the ratio between the initially slow and fast learners was pretty comparable for the different interventions (table 1). Secondly, all subjects practiced in a standardized wheelchair without adjustments for the participant's anthropometry. It could be that this setup gave more room for improvement for some participants compared to others. Finally, the groups were split on a pre-set criterion of 10% increase in mechanical efficiency during the pre-test. This is a first attempt to identify different groups of learners in a cyclic steady-state low-intensity wheelchair propulsion intervention. However, whether there are only 2 groups of learners or more cannot be certain from the current research. Perhaps in the future more data-driven methods like cluster analysis can be used to explore what kind of groups can be logically put together [22].

Conclusion

The IFI, about two thirds of the able-bodied novice participants, started the pretest with a lower mechanical efficiency and a less optimal propulsion technique. However already during the pretest the IFI learned more and this effect continued over the total 80 minutes of practice, while controlling for differences in the practice format. Eventually the IFI benefitted more from the given practice compared to the ISI and learned a better propulsion technique, performed at a higher mechanical efficiency. Over the given practice the IFI had a higher intra-individual variability in the pre and posttest. Possibly this higher variability relates to the increased motor learning of the IFI. Individual motor learning differences are important to take into account for rehabilitation programs. Further exploration of the common characteristics of different types of learners will help to better tailor rehabilitation to the specific needs of wheelchair dependent persons.

Endnotes:
a Forcelink b.v, Culemborg The Netherlands
b Double Performance BV, Gouda, The Netherlands
c Three Rivers Holdings, Mesa, AZ, USA
d MAX Mobility, LLC, Antioch, TN, USA
e Oxycon Pro-Delta, Jaeger, Hoechberg, Germany

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5

chapter 6

Early motor learning changes in upper-limb dynamics and shoulder complex loading during handrim wheelchair propulsion

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Abstract

Background:

To propel in an energy-efficient manner, handrim wheelchair users must learn to control the bimanually applied forces onto the rims, preserving both speed and direction of locomotion. Previous studies have found an increase in mechanical efficiency due to motor learning associated with changes in propulsion technique, but it is unclear in what way the propulsion technique impacts the load on the shoulder complex. The purpose of this study was to evaluate mechanical efficiency, propulsion technique and load on the shoulder complex during the initial stage of motor learning.

Methods:

15 naive able-bodied participants received 12-minutes uninstructed wheelchair practice on a motor driven treadmill, consisting of three 4-minute blocks separated by two minutes rest. Practice was performed at a fixed belt speed (v=1.1m/s) and constant low-intensity power output (0.2W/kg) relative to body mass. Energy consumption, kinematics and kinetics of propulsion technique were continuously measured. The Delft Shoulder Model was used to calculate net joint moments, muscle activity and glenohumeral reaction force.

Results:

With practice mechanical efficiency increased and propulsion technique changed, reflected by a reduced push frequency and increased work per push, performed over a larger contact angle, with a more tangentially applied force and reduced power losses before and after each push. Contrary to our expectations, the above mentioned propulsion technique changes were found together with an increased load on the shoulder complex reflected by higher net moments, a higher total muscle power and higher peak and mean glenohumeral reaction forces.

Conclusions:

It appears that the early stages of motor learning in handrim wheelchair propulsion are indeed associated with improved technique and efficiency due to optimization of the kinematics and dynamics of the hand and arm. This process goes at the cost of an increased muscular effort and mechanical loading of the shoulder complex. This seems to be associated with an unchanged stable function of the trunk and could be due to the early learning phase where participants still have to learn to effectively use the full movement amplitude available within the wheelchair-user combination. Apparently whole body energy efficiency has priority over mechanical loading in the early stages of learning to propel a handrim wheelchair.

Introduction

Persons with a lower-limb disability often depend on a handrim-propelled wheelchair for mobility during daily life. Handrim wheelchair propulsion is a physically straining form of ambulation as a consequence of a low mechanical efficiency and a high mechanical load on the shoulder complex, which might be associated with the frequent over-use injuries of the shoulder in people with a spinal cord injury [1-8].

Different studies on motor learning of wheelchair propulsion have shown that on a group level low-intensity practice can change the propulsion technique of handrim wheelchair propulsion and improve the mechanical efficiency [9-16], which is the ratio of external power output over internal power production. Furthermore, it was found that the propulsion technique changes because of practice, towards a longer-slower movement pattern with an increased angle of hand to rim contact and more net work per cycle, consequently reducing the push frequency [17,18]. However, it is currently not clear in what way these changes in propulsion technique impact the load on the shoulder complex.

To evaluate the load on the shoulder complex during a push cycle, inverse dynamics can be used as input for a musculoskeletal model to estimate muscle activity and joint reaction forces. For experienced wheelchair users the Delft Shoulder and Elbow Model [19] estimated peak glenohumeral reaction forces between 300 to 1400N during each push cycle at speeds between 0.4 and 1.5 m.s-1, with concomitant high relative forces of the rotator cuff muscles, especially of the subscapularis and infraspinatus muscles [3,20-22]. When taking into account that wheeling an hour a day with a typical push frequency of 45 pushes per minute may already add up to some 2700 repetitions, the associated load on the shoulder complex might be considered a risk factor for overuse injury of the rotator cuff [23] and shoulder in general. Therefore, it is important to investigate whether motor learning-associated changes in propulsion technique are related to a reduction of the muscle forces and joint reaction forces of the shoulder complex.

In the present study the effect of natural motor learning on propulsion technique, shoulder load and mechanical efficiency will be studied in a group of novice able-bodied participants during the first twelve minutes of low-intensity wheelchair practice. Previously, this relatively short time frame of practice already showed improvements in mechanical efficiency and propulsion technique while at the same time also showing motor learning differences between a group of slow and fast improvers [17,18]. The slow and fast improvers were identified based on a relative 10% increase in mechanical efficiency over a 12 min practice period. The fast learning group increased more in mechanical efficiency and propulsion technique over the whole practice intervention. The current study will enroll a group of able-bodied novices in the same experimental protocol and - by adding three-dimensional position registration - will also be able to use the Delft Shoulder and Elbow Model [19] to evaluate the consequences of three bouts of 4 min low-intensity natural steady state wheeling practice on a motor driven treadmill on mechanical efficiency, propulsion technique, and on the modeled loading of the shoulder complex.

Therefore the objective of the current study was to establish whether the motor learning process during the first 12 minutes of handrim wheelchair propulsion would lead to 1) an increased mechanical efficiency and propulsion technique; 2) a reduction of mean and peak net moments around the glenohumeral shoulder joint and elbow; 3) a reduction of muscle activation and glenohumeral joint reaction force of the shoulder complex; 4) differences in the effect of practice between two groups of learners based on mechanical efficiency and reflected in propulsion technique and load on the shoulder complex.

It is hypothesized that because of practice the participants will change their propulsion technique towards a less straining mode of wheelchair propulsion [17,18], i.e. an increase in mechanical efficiency, adaption of a longer-slower movement pattern and a reduction in muscle forces and consequent glenohumeral reaction forces. In line with the results of our previous study we expected to identify two different groups of learners.

Methods

6.1 Participants

Fifteen able-bodied novices, with a mean age of 27.4 ± 11.9 years, mean mass of 70.6 ± 13.6 kg and mean height of 1.78 ± 0.09 m, participated in the research after giving informed consent. Criteria for inclusion were: being able-bodied and having no previous experience with wheelchair propulsion. The exclusion criterion was the presence of any severe medical conditions that could have an influence on parameters measured in this study, based on a questionnaire (PAR-Q, ACSM (2009)). The study was approved by the Local Ethics Committee, of the Center for Human Movement Sciences, University Medical Center Groningen, University of Groningen, the Netherlands.

6.2 Protocol

The single session 16-minute experiment was conducted on a level treadmill of 2.4 m length by 1.2 m width (Forcelink[a]) in an experimental wheelchair (Double Performance[b]) with 24-inch measurement wheels (figure 6.1, top). Each participant performed three consecutive 4-minute exercise blocks at a fixed submaximal power output of 0.20 W/kg body weight with two minutes of rest in between blocks. This low intensity was chosen to minimize fatigue or training effects and focus primarily on motor learning. The first 40 seconds were used to get the treadmill up to a speed of 1.11 m/s (4 km/h). Participants received no specific instructions other than to stay on the treadmill using the handrims. Apart from rolling resistance, the required power output was imposed by adding mass to a pulley system (figure 1, top). Pulley mass was determined from the results of an individual wheelchair drag test [24,25].

6.3 Energy expenditure

Oxygen consumption (VO2) was continuously measured during each practice session using breath-by-breath open circuit spirometry[e]. The gas analyzer was calibrated using a Jaeger 5l syringe, room air and a calibration gas mixture. Data collected over the fourth minute of each exercise trial were averaged and taken to reflect physiological

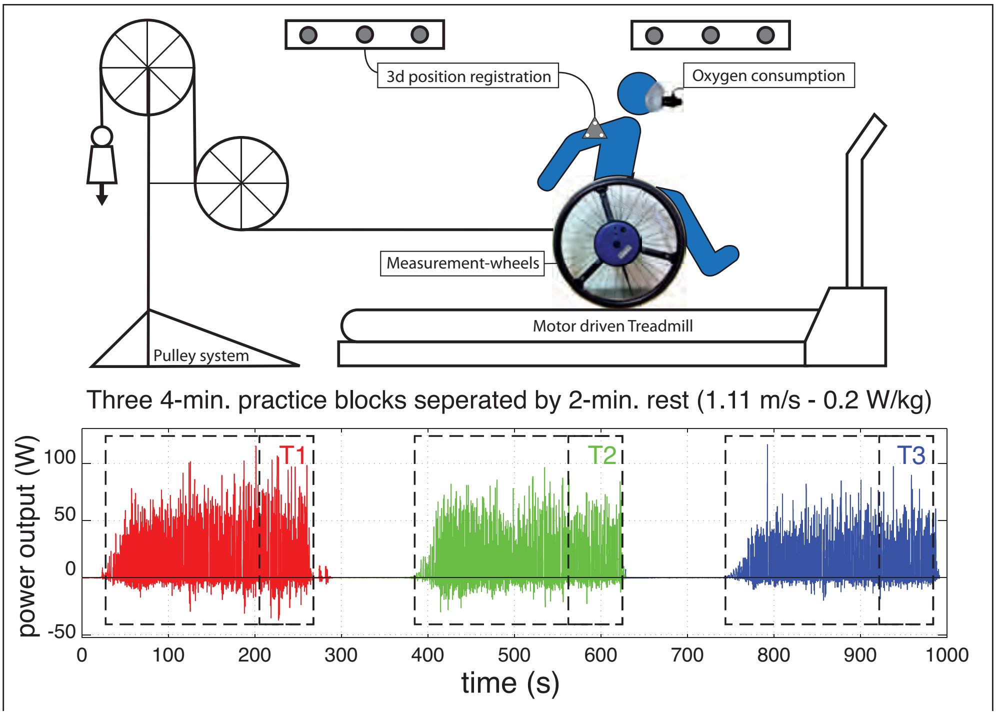

Figure 6.1: Top: Experimental measurement and test set-up for low-intensity steady state (v=1.1m/s; 0.2W/kg body weight) treadmill wheeling, using a pulley system, measurement wheels, mobile oxygen uptake and 3D motion capture on a motor driven treadmill. Bottom: Example of the momentary power output (W) over the whole 12 min practice protocol (3x4min; 2min rest in between) for one participant. Data of the last minute of each practice block was analyzed (T1, T2, and T3).

steady-state wheelchair propulsion. From the VO2 (L/min), VCO2 (L/min) and respiratory exchange ratio (VCO2/VO2) the energy expenditure was determined using the formula proposed by Garby and Astrup [26]. Mechanical efficiency was derived from the ratio between the external power output (W) and the energetic equivalent of oxygen uptake (W) and (Table 1).

6.4 Measurement wheels

The regular rear wheels of the standardized wheelchair were replaced with one of two instrumented wheels, the Optipush[c] (Max Mobility) or the Smartwheel[d] (3-Rivers). Both wheels measure 3-dimensional forces and torques applied to the handrim, combined with the angle under which the wheel is rotated. Data were wirelessly transferred to a laptop at 200 Hz. An electronic pulse at the start of each measurement synchronized both wheels. Data of both wheels show good comparability, with an intra-class correlation for absolute agreement (ICC) of 0.89 for mean power output and ICC's higher than 0.90 for propulsion technique characteristics [27].Table 6.1 Propulsion technique variables and their definitions, automatically processed from the wheel signals using custom written Matlab code [27].

6.5 Propulsion technique

The data from the instrumented wheels were further analyzed using custom-written Matlab routines. To be certain of stable, steady-state wheelchair propulsion, each last minute from the 4-min trials was used for the analysis. Per participant and exercise block

Table 6.1: Propulsion technique variables and their definitions, automatically processed from the wheel signals using custom written Matlab code [27]

Variable	Description	Equation
Mechanical efficiency (%)	The percentage of internal power used for external power delivered at the wheels	Mean power output/Energy expenditure
Push time (s)	Time from the start of positive torque to the stop of positive torque for an individual push.	$t_{end}(i) - t_{start}(i)$
Cycle time (s)	Time from the start of positive torque to the next start of positive torque.	$t_{end}(i) - t_{start}(i - 1)$
Frequency (push*min^{-1})	The number of complete pushes per minute.	$N_{pushes}/\Delta t$
Work/push (J)	The power integrated over the Contact angle of the push.	$\Sigma_{start:end}(Tz * \Delta Ø)$
PnegS (W)	The minimum power preceding the push phase	$Min_{<start}(Power)$
PnegE (W)	The minimum power following the push phase	$Min_{>end}(Power)$
Contact angle (°)	Angle at the end of a push minus the angle at the start.	$Ø_{end}(i) - Ø_{start}(i)$
$Ftot_{mean}$ (N)	3d mean force within the push phase	$Mean_{start:end}(Fx^2 + Fy^2 + Fz^2)^{0.5}$
$Ftot_{peak}$ (N)	3d peak force within the push phase	$Max_{start:end}(Fx^2 + Fy^2 + Fz^2)^{0.5}$
FeF_{mean} (%)	Mean Fraction effective Force	$Mean_{start:end}(F_{tangential}/F_{total})$
GH start position (mm)	Horizontal position of the glenohumeral joint (GHx) at the start of the push with respect to the wheel-axle (WAx)	$GHx_{start}(i) - WAx_{start}(i)$
GH displacement (mm)	The position difference between GH at the start and end of the push phase	$GH_{end}(i) - GH_{start}(i)$

Abbreviations: *t* time(s), $_{start}$(i), start of the current push (sample); $_{end}$(i), end of the current push (sample); Ø, angle (rad); Fx, Fy and Fz, force components (N); Tz, torque around wheel axle (Nm).

the measured force (N), torque (Nm), angle (rad) and time (s) were used for further analyses. Individual pushes were defined as each period of continuous positive torque around the wheel axis with a positive minimum of at least 1 Nm [27]. Over the identified pushes the propulsion technique variables (Table 6.1) were calculated and subsequently averaged over all pushes within the fourth minute of each practice trial per participant.

6.6 *Kinematics*

Kinematic data were collected using an optoelectronic camera system (Optotrak, Northern Digital, Waterloo, Canada) at 100Hz with technical cluster markers attached to the right side of the participants' body and to the wheelchair (Figure 6.2, left). Prior to the actual experiment, a calibration measurement was performed to determine the location of anatomical landmarks in relation to their technical clusters. From these calibrations, the positions of the anatomical landmarks were reconstructed during the experiment (Figure 6.2, right), which in turn were used to construct joint coordinate systems of the shoulder, elbow and wrist [28]. The location of the glenohumeral (GH) rotation point was calculated using the regression method proposed by Meskers et al [29].

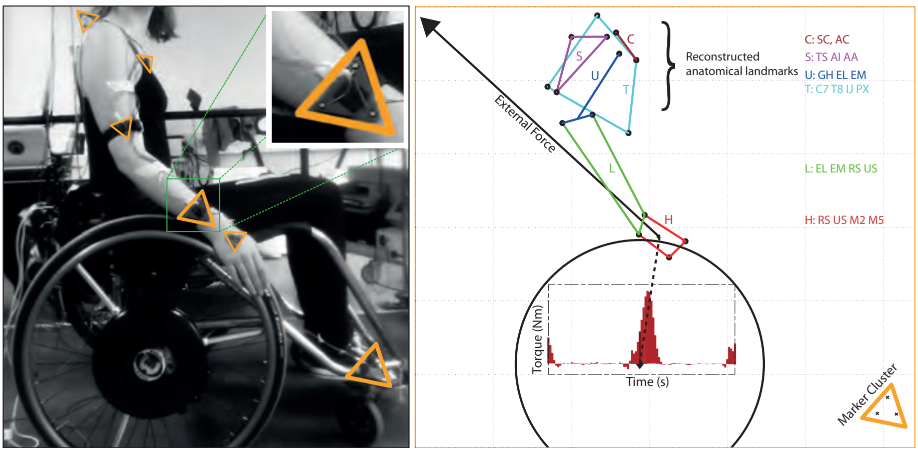

Figure 6.2: Left: Placement of the technical marker clusters during active wheeling on the motor driven treadmill. Right: Combination of kinematics and wheel kinetics showing a sample of the individual external reaction force and resulting torque around the wheel-axle, during the push phase.

Table 6.2: DSEM outcome variables

Variable	Description
GH mean Net Moment/Push (Nm)	The mean external net moment of the reaction force around the glenohumeral joint
GH peak Net Moment/Push (Nm)	The peak external net moment of the reaction force around the glenohumeral joint
HU mean Net Moment/Push (Nm)	The mean external net moment of the reaction force around the Humeroulnar joint
HU peak Net Moment/Push (Nm)	The peak external net moment of the reaction force around the Humeroulnar joint
Muscle Power total mean/Push (W)	The mean sum of all muscle powers during the push
Muscle Power total peak/Push (W)	The peak sum of all muscle powers during the push
Muscle Work total/Push (J)	The total muscle work performed per push
GH Reaction force mean/Push (N)	The mean glenohumeral reaction force per push
GH Reaction force peak/Push (N)	The peak glenohumeral reaction force per push
GH Reaction force peak/Cycle (N)	The peak glenohumeral reaction force per cycle

6.7 *Delft Shoulder and Elbow model*

The Delft Shoulder and Elbow Model (DSEM) is a finite-element, inverse dynamic model describing musculoskeletal behaviour of the upper extremity. Kinematic input was the position of the incisura jugularis, the orientations of the thorax, scapula, humerus, forearm and hand. The 3-dimensional external forces applied by the hand on the handrim served as kinetic input. Five regular consecutive pushes were selected for data analysis. The output of the model is twofold (table 6.2). First inverse dynamical calculation takes into account the external forces and accelerations to calculate net moments around the glenohumeral shoulder joint and humeroulnar joint. From this input the model simulates the activity of 31 muscles, divided in 155 elements and the consequent joint reaction forces. The non-individualized anthropometric parameters are based on two cadaver studies [30]. Muscle forces were calculated by an energy related cost function [31]. To enable interpretation and comparison of muscle forces, forces were also expressed as percentages of their maximum based on a force per physiological cross-sectional areas of these muscles of 100 N*cm-2, taking into account that the physiological cross-sectional area was measured in an older specimen [30], while the task is performed by young participants.

6

6.8 *Statistics*

All data were checked for normal distribution and qualified for parametric statistical testing. To evaluate the effect of practice time repeated-measures ANOVA was used to compare mechanical efficiency, propulsion technique parameters, net joint moments of the glenohumeral and humeroulnar joint and the resulting muscular activity and glenohumeral joint reaction forces. Significance for the repeated-measures ANOVA was set at a $p<0.05$ and by use of the Bonferroni correction the significance for the post hoc t-tests between any of the three different blocks was set at $p<0.017$.

The relationship between the mean net joint moment and the mean glenohumeral joint reaction force was evaluated using a linear least square regression.

To examine motor learning differences between participants, the group was split in two sub-groups, based on a relative increase in mechanical efficiency of more than 10% between T1 and T3, to ensure that differences in learning were above the natural expected variation [18]. The two groups were subsequently compared on the main outcome measures over all three practice-blocks using repeated-measures Anova, with the interaction between group (≤10% or >10%) and practice-blocks as the most important outcome.

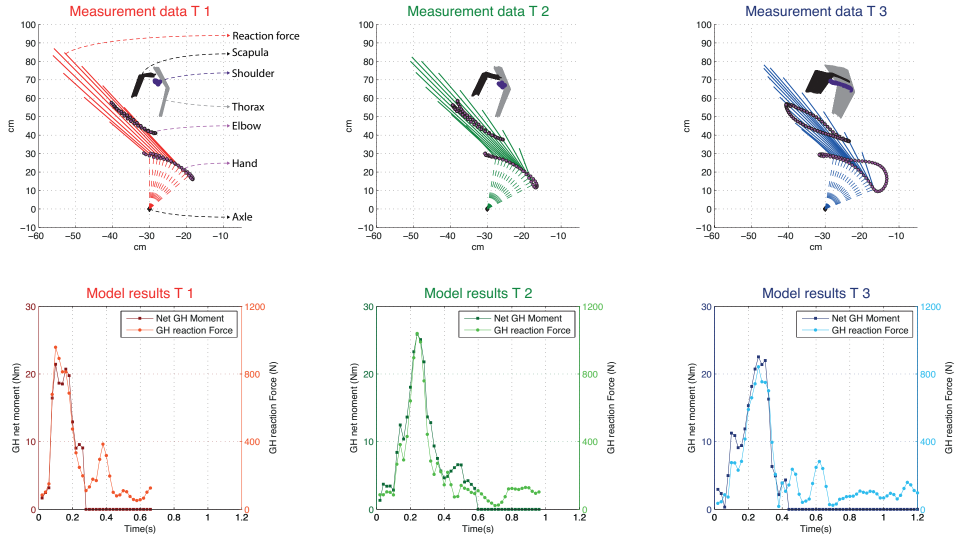

Figure 6.3: Typical example of the measured data (row 1) and consequent DSEM outcomes (row 2) over the 12 min natural learning period, derived from the input from the T1,T2 and T3 measurements. The 1st (top) row shows the kinematic and kinetic input (reaction force vector) for the model in relation to the trajectories of the shoulder, elbow and hand over the push and recovery phase (Cycle) at T1,T2 and T3.The 2nd row shows the shoulder loading as expressed in both total net moment (Nm) and total joint reaction force (N) over this same cycle at T1,T2 and T3.

Results

Participants practiced at an average power output of 16.5 ± 3.4 W. Figure 6.3 shows a typical example of the data collections and outcomes for a push cycle at T1, T2 and T3.

6.9 *Effect of motor learning on mechanical efficiency and propulsion technique*

The mechanical efficiency significantly increased (T1: 5.5%, T2: 5.9%, T3: 6.0%) over the practice time (Table 6.3). The post-hoc comparison however only showed a significant difference between T1-T3.

For the timing of propulsion technique significant increases in push time (T1: 0.31s, T2: 0.34s, T3: 0.34s) and cycle time (T1: 0.97s, T2: 1.15s, T3: 1.15s) were found with significant post-hoc differences between T1-T2 and T1-T3. The increase in cycle time was also reflected by the reduced push frequency (T1: 66.6, T2: 55.5, T3: 55.0 pushes per minute) with similar significant post-hoc differences between T1-T2 and T1-T3. The positive work per push went up (T1: 8.7J, T2: 10.3J T3: 10.3J), but again showing post-hoc effects only between T1-T2 and T1-T3. The negative phases before the push (T1: -8.1W, T2: -6.1W, T3: -5.5W) and after the push (T1: -5.0W, T2: -3.9W, T3: -2.8W) significantly reduced each next trial.

The increased work per push was performed over a larger contact angle on the handrim (T1: 63.5, T2: 69.6 T3: 70.4 degrees), rather than by an increase of force application. The latter is expressed in the absence of change in both Ftotmean (T1: 41.5N, T2: 41.8N, T3: 40.4N) and Ftotpeak (T1: 68.0N, T2: 69.6N T3: 66.5N). The mean fraction effective force showed a significant increase (T1: 69.4%, T2: 75.4%, T3: 75.3%), but again showing post-hoc effects only between T1-T2 and T1-T3.

The start position of the glenohumeral joint in the sagittal plane at the start of

Table 6.3: Mean (+/- sd) outcomes for all participants (n=15) over the three consecutive practice blocks and outcomes of statistical analyses (levels of significance: P-Anova: <0.05; Bonferonni tests: <0.017).

		T1 Mean	Sd	T2 Mean	Sd	T3 Mean	Sd	p-Anova	p-T1T2	p-T1T3	pT2T3
	Mechanical Efficiency	**5.5**	1.1	**5.9**	1.1	**6.0**	1.3	**0.047**	0.039	0.014	0.266
Propulsion Technique	Push time (s)	**0.31**	0.06	**0.34**	0.06	**0.34**	0.06	**0.003**	0.006	0.007	0.202
	Cycle time (s)	**0.97**	0.24	**1.15**	0.26	**1.15**	0.24	**0.004**	0.004	0.012	0.471
	Frequency (push/min)	**66.6**	17.5	**55.5**	12.8	**55.0**	11.9	**0.004**	0.007	0.010	0.371
	Work/push (J)	**8.7**	2.1	**10.3**	3.0	**10.3**	3.0	**0.009**	0.009	0.019	0.599
	PnegS (W)	**−6.1**	3.3	**−4.8**	2.3	**−4.3**	2.1	**0.003**	0.011	0.007	0.027
	PnegE (W)	**−3.7**	2.8	**−2.6**	1.9	**1.9**	1.9	**0.012**	0.048	0.011	0.014
	Contact angle (°)	**63.5**	12.1	**69.6**	10.9	**70.4**	10.8	**0.002**	0.005	0.005	0.216
	Ftotmean (N)	41.5	10.3	41.8	14.2	40.4	13.1	0.672	0.877	0.562	0.167
	Ftotpeak (N)	68.0	17.2	69.6	24.1	66.5	22.2	0.553	0.632	0.650	0.082
	FeFmean (%)	**69.4**	10.4	**74.8**	11.6	**75.3**	9.9	**0.000**	0.002	0.001	0.342
	GH start position (mm)	−41	33	−53	35	−47	47	0.247	0.081	0.454	0.388
	GH displacement (mm)	27	13	37	31	39	21	0.245	0.297	0.116	0.770
Net Moments	GH mean Net Moment per Push (Nm)	**12.4**	3.4	**16.1**	5.9	**15.3**	3.4	**0.010**	0.010	0.000	0.700
	GH mean Net Moment per Push (Nm)	26.3	8.1	31.0	12.7	28.5	7.0	0.270	0.080	0.120	0.770
	HU mean Net Moment per Push (Nm)	1.6	2.6	0.9	2.1	0.8	2.7	0.423	0.284	0.309	0.779
	HU peak Net Moment per Push (Nm)	7.9	4.2	6.6	3.2	7.0	3.5	0.452	0.268	0.443	0.636
Model Results	Muscle Power total mean per push (W)	**25.1**	11.0	**35.0**	14.7	**37.5**	15.4	**0.012**	0.008	0.018	0.561
	Muscle Power total peak per push (W)	110.7	49.0	120.5	52.8	134.5	108.6	0.566	0.543	0.222	0.645
	Muscle Work total mean per push (J)	**11.3**	4.0	**15.1**	6.7	**16.1**	7.8	**0.040**	0.014	0.005	0.333
	GH Reaction force mean per push (N)	**315**	107	**419**	171	**439**	165	**0.006**	0.001	0.003	0.337
	GH Reaction force peak per push (N)	690	320	790	390	901	560	0.265	0.119	0.031	0.268
	GH Reaction force peak per cycle (N)	**239**	41	**266**	58	**277**	68	**0.007**	0.006	0.005	0.187

Significant results are presented in bold font style.

the push did not increase significantly over time (T1: -41mm, T2: -53mm, T3: -47mm). Also the following displacement during the push did not increase significantly over time (T1: 27mm, T2: 37mm, T3: 39mm), suggesting a rather inert trunk position during the propulsion cycle.

6.10 *Effect of motor learning on shoulder complex loading*

The mean net moment of the external force over the push phase around the glenohumeral joint significantly increased (T1: 12.4Nm, T2: 16.1Nm, T3: 15.3Nm) with significant post-hoc differences between T1-T2 and T1-T3. The peak net moment of the external force around the glenohumeral joint did not increase significantly over time (T1: 26.3Nm, T2: 31.0Nm, T3: 28.5Nm). Around the humeroulnar joint no significant changes in mean net moment (T1: 1.6Nm, T2: 0.9Nm, T3: 0.8Nm) or peak net moment (T1: 7.7Nm, T2: 6.6Nm, T3: 7.0Nm) were present over time.

In line with the increased net moments around the glenohumeral joint, the total mean muscle power per push, as estimated from the DSEM, increased significantly (T1: 25.1W, T2: 35.0W, T3: 37.5 W), with post-hoc difference seen for T1-T2 and T1-T3. No significant increase in peak power was observed (T1: 110.7W, T2: 120.5W, T3: 134.5W). Also, the total muscle work per push increased over time (T1: 11.3J, T2: 15.1, T3: 16.1J), with post-hoc differences for T1-T2 and T1-T3.

A significant increase was found for the mean glenohumeral reaction force per push (T1: 315N, T2: 419N, T3: 439N) with post-hoc differences for T1-T2 and T1-T3. This increase per push also resulted in an increased mean glenohumeral force per cycle (T1: 239N, T2: 266N, T3: 277N), with post-hoc differences again seen between T1-T2 and T1-T3 (Table 3). The peak glenohumeral reaction force did not significantly increase over time (T1: 690N, T2: 790N, T3: 901N).

The increase in net moments and glenohumeral reaction force indicate an increased load on the shoulder complex. Over all observations of all participants a linear relationship was found between the net joint moments (M_dsem) and total compression forces (F_dsem) in the GH joint, shown by the following regression equation: F_dsem=33.4*M_dsem+112.1, with $p<0.01$, e=102.1N and $r=0.73$ (Figure 6.4).

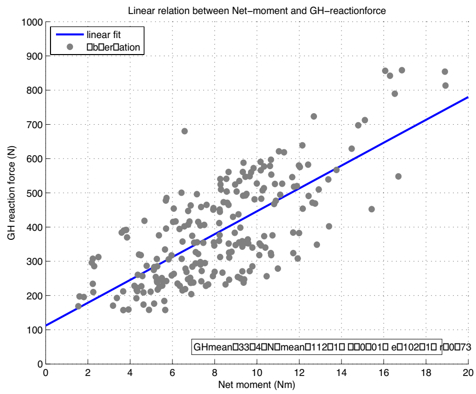

Figure 6.4: Linear relationship between the net joint moments (M_dsem) and total compression forces (F_dsem) in the GH joint, shown by the following regression equation: F_dsem=33.4*M_dsem+112.1, with $p<0.01$, e=102.1N and $r=0.73$.

6.11 Moment-Balances

Figure 6.5 shows a typical example of the different muscle contributions that counteract the external moment around the glenohumeral joint for each of the three global axes. Around the global x-axis, mainly the infraspinatus, subscapularis and biceps muscles are responsible for the ‘flexion’ moment, with smaller contributions of the coracobrachialis and pectoralis major. Around the global y-axis the supraspinatus, subscapularis and biceps mostly account for the ‘adduction’ moment. The moment around the global z-axis is mainly expressed by pectoralis major, biceps and coracobrachialis activity, but besides the external moment these muscles also have to counteract the vector components of the infra- and supraspinatus in this plane. The potential consequences of motor learning for this typical pattern over time are described below.

6.12 Effect of motor learning on individual muscle activity

6.12.1 Main drivers

The triceps showed the highest mean forces over time (T1: 176N, T2: 184N, T3: 185N), with a large positive contribution to power development around the elbow (T1: 5.1W, T2: 5.6W, T3: 5.7W), but both force and power did not change significantly over time (figure 6.6). The highest mean forces leading to positive power development around the shoulder were found in the rotator-cuff muscles subscapularis (T1: 106N, T2: 129N, T3: 132N), infraspinatus (T1: 86N, T2: 120N, T3: 114N), and supraspinatus (T1: 75N, T2: 106N, T3: 105N), of which only supraspinatus expressed a significant change over time at group level (T1-T2 and T1-T3). The mean force of the serratus

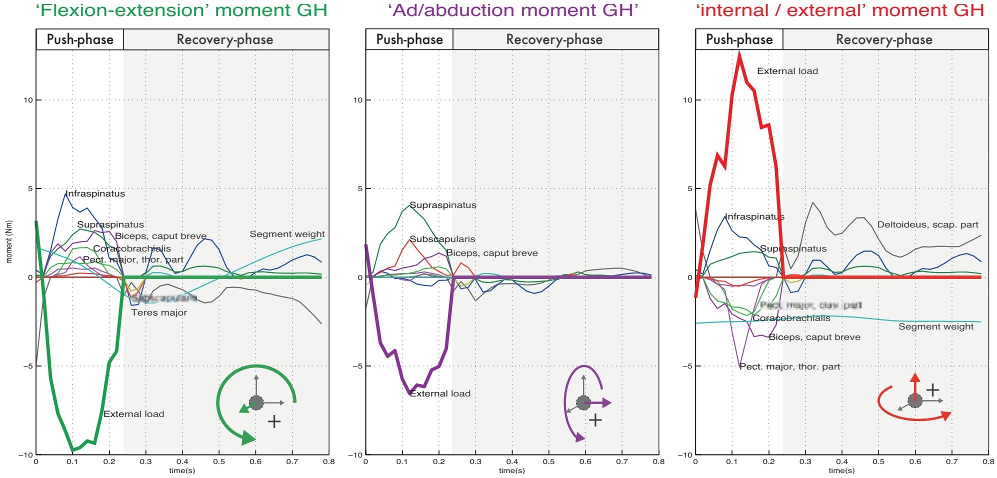

Figure 6.5: Example of the muscles- vs. external moment-balance around the Glenohumeral joint for the three global axes, as determined by the DSEM for an individual push at T3.

anterior (T1: 65N, T2: 83N, T3: 87N) increased significantly between T1-T2 and T1-T3. Although this did not lead to a significant change in power output it is noticeable that its mean power contributions are negative at T1 and positive at T3 (T1: -0.9W, T2: 0.6 W, T3: 1.1 W). The mean force of the biceps (T1: 44N, T2: 68N, T3: 74N) increases significantly between T1-T2 and T1-T3. Figure 5 shows the positive contribution of the biceps to flexion/extension and ad/abduction around the shoulder, but since the biceps is a bi-articular muscle, its force delivers a moment around both shoulder and elbow and a negative power contribution is found (T1: -0.7W, T2: -1.5W, T3: -1.5 W), which did not increase over time. Although a trend was present (p<0.1), the mean force of the pectoralis major (T1: 45N, T2: 65N, T3: 61N) did not increase significantly, but a significant increase of positive power (T1: 3.5W, T2: 6.2W, T3: 5.4W) was found for T1-T2. The mean force of the trapezius (T1: 45N, T2: 45N, T3: 54N) and scapular part of the deltoideus (T1: 49N, T2: 42N, T3: 47N) did not increase over time. The power production of these two muscles was negative and also did not change significantly (T1: -0.4W, T2: -0.8W, T3: -0.1W) and (T1: -0.3W, T2: -0.9W, T3: -0.1W). The muscle force of the brachialis (T1: 47N, T2: 41N, T3: 38N) significantly decreased between T1-T2 and T1-T3, but no significant change was found for the power production (T1: -0.6W, T2: -0.4W, T3: -0.3W).

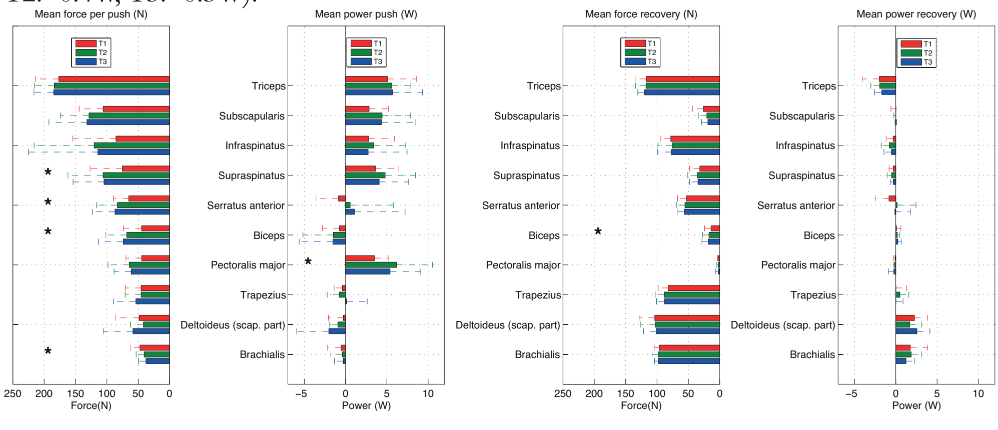

Figure 6.6: Outcomes (n=15; mean +/-sd) of the DSEM for individual muscle forces (N) and powers (W) during the push (left) and recovery (right) phases over time (T1-T3).

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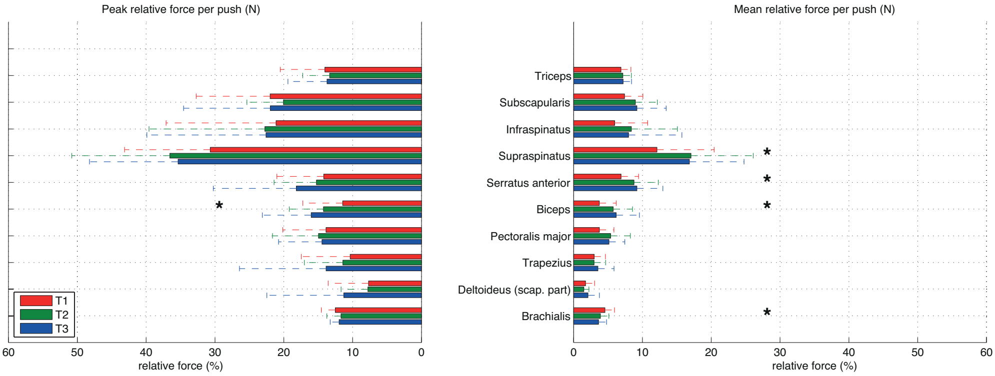

Figure 6.7: Relative mean forces (n=15) of individual muscles during the push phase. The error bars represent the standard deviation

6.12.2 Relative muscle activity

The contributions of individual muscles relative to their theoretical maximum force (figure 6.7) gives a perspective on those muscles that may be at risk for overuse. The supraspinatus is the most taxed muscle during the push phase, of which the mean relative force (T1: 12.1%, T2: 17.1%, T3: 16.8%) significantly increased for T1-T2 and T1-T3, but with no significant increase in the maximum relative force (T1: 30.7%, T2: 36.6%, T3: 35.4%). The biceps was the only muscle to significantly increase in peak relative muscle force (T1: 11.4%, T2: 14.2%, T3: 16.0%), with significant increase between T1-T2 and T1-T3.

6.13 Individual differences in learning

Seven participants could be classified as Initially Fast Improvers and the other eight as Initially Slow Improvers. A significant interaction was found for mechanical efficiency (figure 6.8), but not for the propulsion technique variables or the net moments or the model results.

Discussion

Because of practice an increase was found in mechanical efficiency over time, indicating that overall less energy was used to maintain a constant speed and power output in the wheelchair on the motor driven treadmill. A concomitant change in propulsion technique was expressed in a reduced push frequency and increased amount of work per push, performed over a larger contact angle with reduced power losses before and after a push, where mean and peak total force in the push remained constant over time. Simultaneously, the fraction effective force increased, indicating a more tangential direction of the applied forces around the wheel-axle. Contrary to our expectations, the above-mentioned propulsion technique changes were found together with an increased net moment, increased total muscle power and increased total muscle work around the glenohumeral shoulder joint. Consequently, this resulted in higher local strains in the shoulder complex as expressed in higher mean and peak glenohumeral reaction forces during both the push-phase as well as the full propulsion-cycle over time.

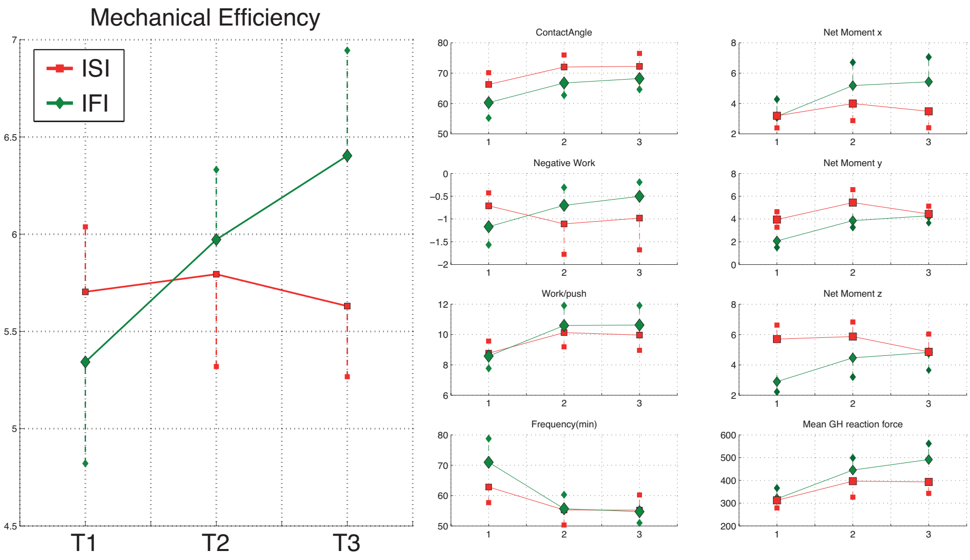

Figure 6.8: Differences between initially fast improvers (n=7; IFI; green) and initially slow improvers (n=8; ISI, red). Only mechanical efficiency shows a significant interaction effect between Initially Slow Improvers (ISI) and Initially Fast Improvers (IFI) over practice time (T1-T3).

The current study evaluated the same motor learning process of a steady-state cyclical task on three distinct levels of task execution; the mechanical efficiency encompasses the whole body physiological outcome, the propulsion technique reflects the wheelchair-user interaction at the hand and handrim and the DSEM gives the most detailed description of changes on the level of the shoulder complex. The relations among these three levels are discussed below in the context of the constant experimental conditions and task of maintaining an average power output (0.2W/kg) and treadmill speed (1.11m/s) over time; given this common task different relations can be presumed among the different outcomes of these different levels of measurement.

6.14 *Effect of motor learning on mechanical efficiency and propulsion technique*

The increased mechanical efficiency indicates a more optimal task performance, i.e. energy efficient changes within the body as a consequence of task execution characteristics, among others propulsion technique. The propulsion technique changes that were previously reported to relate most to the increased mechanical efficiency over the initial 12 minutes indeed changed in the current study, i.e. a reduced negative work per cycle, an increased contact angle, an increased work per cycle and consequently a reduced push frequency [18].

In other cyclical tasks the reduced energy cost also coincided with an increase in movement amplitude and a decrease of movement frequency, described as a longer-slower movement pattern [32-36]. Similar to those observations the reduction of the push frequency as a consequence of motor learning is thought to be key to all other propulsion technique changes seen in this cyclic synchronous upper body task [37]: it reduces the repetitiveness of arm motions, which leads to less moments of peak strain and less negative work because of the reduction of the number of (de)coupling of the hands onto the handrim per time unit. An additional increase in movement amplitude and performed work might have been achieved by use of the trunk muscles [38]. However, no increase of trunk motion, i.e. no increase in GH displacement in the sagittal plane, was observed with practice. Possibly, in this early phase of learning the users are still solving the control problem of wheelchair propulsion by maintaining a fairly rigid trunk orientation, instead of already fully using the movement amplitude of the trunk as can be observed in more trained wheelchair-users with adequate trunk control [39].

The Fraction effective Force increased on average 5% between T1 and T3 in the current experiment, which indicates a more tangential orientation of the total force vector of the hand on the rim. This is more than in our previous study on natural learning of handrim propulsion [18], where in a larger group only an increase of 2% was found. As this increase is the consequence of non-instructed natural motor learning, this change in FeF is seen as beneficial because less non-propulsive force needs to be applied.

6.15 Effect of motor learning on shoulder complex loading

Contrary to our expectations, the mean net moment per push of the external force around the glenohumeral joint increased over time, indicating a higher load on the shoulder complex. This implies that the force of the hand on the handrim increased in vector length and/or in moment arm with respect to the glenohumeral joint over time. However, no changes in mean or peak total force of the hand on the handrim were found over time. Therefore, the change in the mean net moment is mainly attributed to changes in moment arm, which is in accordance with the observed increase of the fraction effective force. Another potential factor that might have influenced the moment arm is the position of the glenohumeral joint with respect to the external applied force, but no changes were found in the position or displacement of the glenohumeral joint over time.

Following the same trend as the net moments, the total muscle power and total muscle work around the glenohumeral shoulder joint increased with practice. Given the reduced push frequency, by definition an increased work per push on the wheel is necessary to maintain power output [25]. From our results the increase in total muscle work is larger then the increase in work per push at the wheel. Possibly, for an increase in positive work of the muscles extra work is necessary to stabilize the joint, since the shoulder joint unlike the hip needs more active muscle control for joint stability [40].

The higher estimated muscle activity, as expressed by the increased muscle power and muscle work, resulted in higher mean glenohumeral reaction forces during both the push-phase and the whole push-cycle over time. The average glenohumeral peak force at T3 was around 900 N, which is in accordance with previously reported values [3].

The net moments and the joint reaction force of the glenohumeral joint showed a moderately strong linear relationship. This was previously reported for abduction in static tasks [41] with a fairly similar slope (33.4 vs. 35.3), but with a different intercept (112.1N vs. 8.12N). The net glenohumeral joint moment appears to be a good indicator for mechanical load in the glenohumeral joint for the dynamic wheelchair propulsion task.

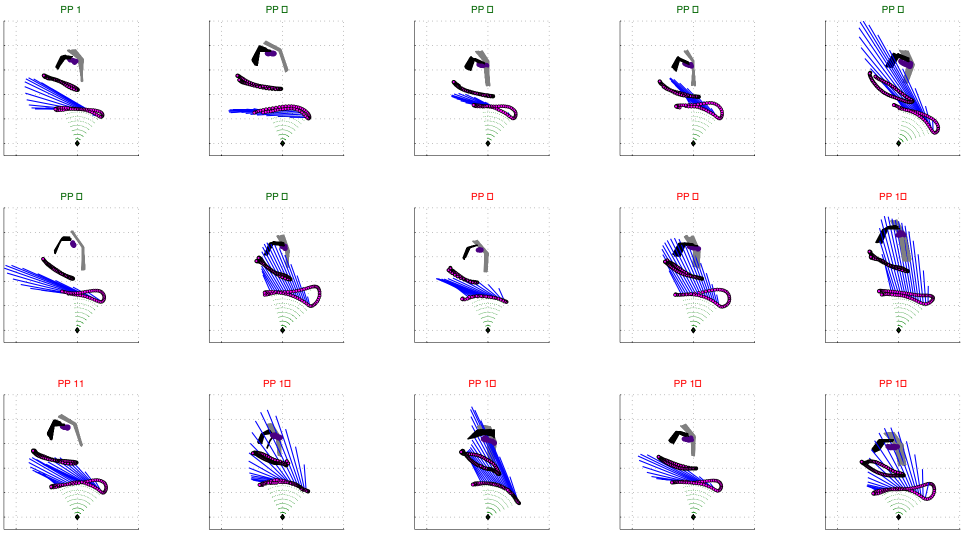

Figure 6.9: One push cycle for each individual in the 12th minute of practice. Participants are categorized on the classification of initially fast Improvers (IFI, green title) and initially slow improvers (ISI), red title).

6.16 *Effect of motor learning on individual muscle activity*

The activity of the triceps in this group of young able-bodied novices is higher than reported in an EMG study during this initial phase of learning on an ergometer [42] and also higher than reported in more experienced users [3,43]. The triceps as a group have the highest physiological cross sectional area of all muscles and during this initial phase of learning appear to be the prime muscle power producers [44].

The rotator-cuff muscles subscapularis, infraspinatus and supraspinatus, three prime stabilizers of the glenohumeral joint, are highly active during the push phase, especially relative to their limited muscle mass; their activity is comparable to the activity reported by other studies with more experienced users [3,43]. Moreover, because of practice even an increase in the supraspinatus activity is seen that contributes to positive power. The only other muscle that significantly increased in mean force over time and contributed to positive power is the serratus anterior. Even though no significant change in power of the serratus anterior was shown, it is a muscle that at T1 had a mean negative power and at T2 and T3 a mean positive power. The muscle helps to protract the scapula around the thorax and might depending on the timing be able to deliver more positive power.

An increase in biceps and decrease in brachialis activity was observed with practice. Both deliver negative power around the elbow, i.e. increase in muscle length, but the biceps also has an important contribution to counteract the net moment around the shoulder (Figure 6.5). The negative power contributions of the elbow flexors are in line with the previously stated suggestions for a low mechanical efficiency [45]. Although increased biceps activity might have helped with the more tangential force direction, because the negative power observed in the biceps allows the direction of the external force to come closer to or cross over the elbow, its function can be described as a balance between cost and effect, since the mechanically required and biomechanically preferred force directions are not in accordance with each other [46].

An increase in the power of the pectoralis major was found with practice, with a trend of increased muscle force. Also in other studies the pectoralis major was shown to be one of the major power contributors [43,47,48]. Finally, the contribution of the clavicular part of the deltoideus was very low in these novice wheelchair-users, while previously this was reported to be a main contributor [42,43,47].

6.17 Individual differences in learning

Seven initially fast improvers and eight initially slow improvers were identified; this is relatively more slow learners than found in a larger group of 70 participants with 46 vs. 24 respectively [18]. The curves for mechanical efficiency in the current study look fairly similar compared to a previous study with the slow learners showing a steady line around 5.7 % and the fast learners initially starting lower and increasing over time. However, apart from mechanical efficiency, in the current experiment no significant interactions were present in any of the propulsion technique measures or in the load on the shoulder complex. This might be due the high standard deviations in performance outcomes within the limited sample-size of this group. The goal of the experiment was to look at common motor learning changes because of practice at different levels of observation. However, since no individual adjustments were made to the wheelchair every participant was confronted with a slightly different wheelchair-user interface as a consequence of body size vs the constant wheelchair configuration for all participants, while seat-height, chair-width and weight distribution are considered important factors for wheelchair propulsion [49-56]. On a group level an increase in mechanical efficiency showed that individuals were able to optimize within the task constraints, however the optimal solution is suspected to be different based on the constraints-based framework proposed by Sparrow and Newell [57]. Although Figure 6.9 needs to be interpreted with caution, given the large intra-individual variability it gives a view on the large inter-individual differences still present during the final minute of practice.

6.18 Clinical relevance

Little is known about the upper-body strain of wheelchair propulsion during the initial stages of wheelchair propulsion during rehabilitation, while at the same time shoulder pain is already present at the start of active rehabilitation [58] and at discharge was recently reported as high as 39% of 138 of persons with a newly acquired spinal cord [59]. The inexperienced able-bodied group in the current study showed a high load on the rotator-cuff muscles subscapularis, infraspinatus and supraspinatus, possibly placing them at risk for over-use injury. Novice wheelchair-users during rehabilitation that are still recovering from the recent trauma are expected to be more vulnerable and although the chosen intensity had low impact on the cardio-respiratory system it may cause a high local risk for overuse of the rotator-cuff muscles already in the very first stage of rehabilitation wheelchair practice. Moreover, with practice the load on the shoulder complex increased instead of reduced. Therefore the design of practice interventions aimed at improving propulsion technique and physical capacity should be evaluated on their impact on the shoulder, balancing stress and recovery.

Continued practice over a longer time scale by able-bodied participants [9-18] and by wheelchair-dependent persons [60] has been shown to further improve mechanical efficiency and propulsion technique, however the findings of the current study emphasize the need to further explore the consequences of motor learning and possible physical adaptations for the local strain on the shoulder complex, using a combination of modeling, kinematics and kinetics.

6.19 Limitations

The Delft shoulder model does not individualize to the anthropometrics of an individual but translates the measured values onto a cadaver based model. Although the values of the model showed reasonable agreement with EMG and an instrumented shoulder joint [19,61], the absolute values should be taken with caution. Fortunately the entire data recording was done in a single session, so each next trial was performed with the same placement of technical markers and calibrations of the measurement devices. Therefore, the same input was used on the same model to say something about change over time on a group level.

Conclusion

Over the first 12 minutes of practice naive able-bodied participants increased their mechanical efficiency, indicating that less energy was used to maintain a constant speed and power output. A change in propulsion technique was shown by a reduced push frequency and increased work per push, performed over a larger contact angle with reduced power losses before and after a push and a more tangentially applied force. Contrary to our expectations, the above-mentioned propulsion technique changes were found together with an increased net moment, increased total muscle power and increased total muscle work around the glenohumeral joint. Consequently, this resulted in higher mean and peak glenohumeral reaction forces. This could be due to the early learning phase where participants still have to learn to effectively use the full movement amplitude available within the wheelchair-user combination. Apparently whole body energy efficiency has priority over mechanical loading in early stages of learning to propel a handrim wheelchair.

Endnotes:
a Forcelink b.v, Culemborg The Netherlands
b Double Performance BV, Gouda, The Netherlands
c Three Rivers Holdings, Mesa, AZ, USA
d MAX Mobility, LLC, Antioch, TN, USA
e Oxycon Pro-Delta, Jaeger, Hoechberg, Germany

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chapter 7

WHEEL-i: the development of a wheelchair propulsion lab for rehabilitation and sports.

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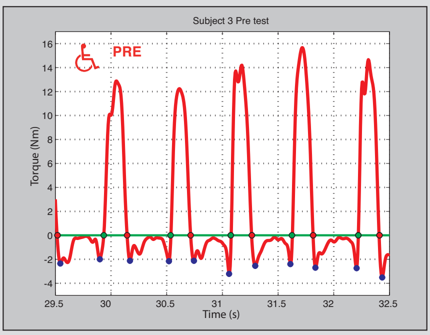

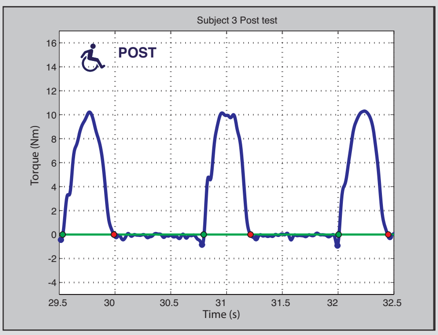

Abstract

Objective To describe the enabling factors and barriers experienced in the "Wheelchair Expert Evaluation Laboratory – implementation" (WHEEL-i) project, in which scientific knowledge, tools and associated systematic analyses of hand rim wheelchair propulsion technique, the user's wheelchair propulsion capacity, the wheelchair-user interface as well as the wheelchair mechanics were implemented in two rehabilitation centers.

Design Implementation project.

Patients Spinal cord injury.

Methods In this implementation project standardized tests were performed: wheelchair skills tests, two questionnaires, and a steady-state exercise test on a treadmill in which propulsion technique (forces and torques) and physical strain (oxygen uptake, heart rate and mechanical efficiency) were measured.

Results Good interpretation of the test outcomes was the most important barrier. To discuss individual wheelchair performance results with patients and clinicians, reference data were developed, smallest detec differences (SDD) were calculated and software to simultaneously show video recordings and force and torque signals was developed.

Conclusion Based on pilot results, the largest barrier for systematic monitoring of the individual wheelchair fitting and learning process in rehabilitation with, among others, instrumented measurement wheels was the interpretation of the outcomes. For proper interpretation of individual outcomes, the availability of reference data, SDDs, and visualisation of the outcomes is of utmost importance.

Introduction

Wheeled mobility is of crucial importance to a growing population of lower-limb impaired and often ageing individuals worldwide. The vast majority of this population in the Western world will use hand rim wheelchairs. Upper-body exercise – especially hand rim propulsion - is far more strai ning and less mechanically efficient than leg work (1;2). This low efficiency, together with the often low physical capacity of the user, leads to high physical strain in daily life and subsequently a limited radius of action. Furthermore, hand rim wheelchair propulsion often leads to upper-body overuse complaints. For example, 30-40% of people with a spinal cord injury indicated shoulder pain during and in the year after rehabilitation (3). This is probably due to the high mechanical strain on the glenohumeral joint during wheelchair propulsion and wheelchair-related activities such as making a transfer, which might lead to joint damage in the long term (4).

To prevent overuse injuries and to obtain and maintain mobility and develop an active lifestyle both wheelchair (e.g. mass, tire pressure) and user (e.g. fitness, skills in terms of negotiating a slope or mounting a curb, propulsion technique in terms of force application on the rim) must have the best condition. Furthermore, the wheelchair-user interface (e.g. seat height, rim and wheel diameter) needs to be ergonomically tuned to the best wheeling performance in different environments for the specific individual (5) (Figure 7.1).

Despite that several research groups conducted a considerable number of investigations into wheelchair propulsion over the last thirty years (6-11), until today wheelchair fitting is a personalized professional skill that has still little scientific foundation. Furthermore, monitoring the propulsion technique and physical strain of new wheelchair users in Dutch rehabilitation is not common practice. However, this approach can be very useful to optimize the wheelchair, interface, and educate the user to establish the best propulsion technique, to improve physical capacity and skill and to prevent upper-extremity overuse injuries. This essentially points towards a global aim whereby wheelchair users can engage more actively in society.

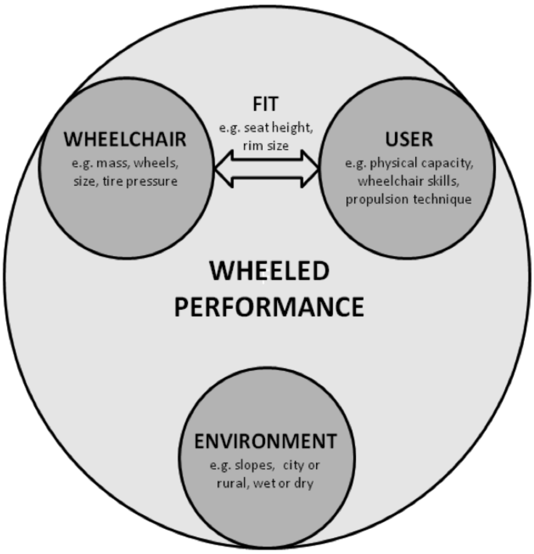

Figure 7.1: Model indicating that wheelchair performance is dependent on different factors.

The use of objective standardized measurements at an individual level, to quantify results of rehabilitation and as part of evidence-based rehabilitation practice, is seen as an increasingly important part of good clinical practice. Therefore, monitoring patients with different tests is more and more common in rehabilitation practice (12), even with very sophisticated measurement techniques. Nowadays, many rehabilitation centers have a gait analysis laboratory to investigate and optimize the walking pattern (e.g. ground reaction force, muscle activity, joint angles) and capacity of individual patients. Similar measurements can be performed in a wheelchair propulsion analysis lab, the only difference is that the focus is on the upper body and that the applied forces and torques are, therefore, measured on the hand rim. Today, these forces can be quite easily measured with commercially available measurement wheels, such as the SmartWheel and the Optipush (Figure 7.2). These wheels can easily be attached to most user's own wheelchairs.

In several clinics in the USA measurements with an instrumented wheel (SmartWheel) are implemented (13). They use a standard protocol that consists of four basic elements (propulsion over tile, carpet, up a ramp, and through a figure 8). Unfortunately, that protocol is not standardized regarding velocity and power output, which makes it hard to interpret possible differences in propulsion technique and physical strain due to interventions on the level of the wheelchair or user. For this reason the clinical SmartWheel User Group (SWUG) protocol was not employed; instead a standardized protocol with a steady-state wheelchair exercise test on a treadmill besides skill testing and questionnaires to measure shoulder pain and self-efficacy in wheeled mobility was employed. All Dutch specialized spinal cord injury rehabilitation units have a wheelchair-specific treadmill since the start of this century (14).

In the context of today's scientific knowledge and understanding in wheelchair propulsion linked to the availability of measurement wheels, the WHeelchair Expert Evaluation Laboratory – implementation (WHEEL-i) project was undertaken. The objective of this project was to implement a systematical analysis of the user, the wheelchair-user interface as well as the wheelchair mechanics in the clinical setting of two Dutch rehabilitation centers analogous to gait analysis, using evidence-based techniques and experimental strategies. The aims of the present study were to describe 1) the test protocol that is used in WHEEL-i and 2) the enabling factors and barriers of successful implementation of WHEEL-i.

METHODS

7.1 Implementation

Two Dutch rehabilitation centers participated in the WHEEL-i 'innovation in rehabilitation' project. Within each center a WHEEL-i work group was formed to discuss how the project could be implemented best in the specific center. These work groups consisted of a rehabilitation physiatrist, physical therapist, an occupational therapist, a rehabilitation technician, a human movement scientist in the field of wheelchair propulsion and a professional in wheelchair fitting. These work groups defined the test protocol, how the project could be implemented best in the rehabilitation centers and discussed the implementation process.

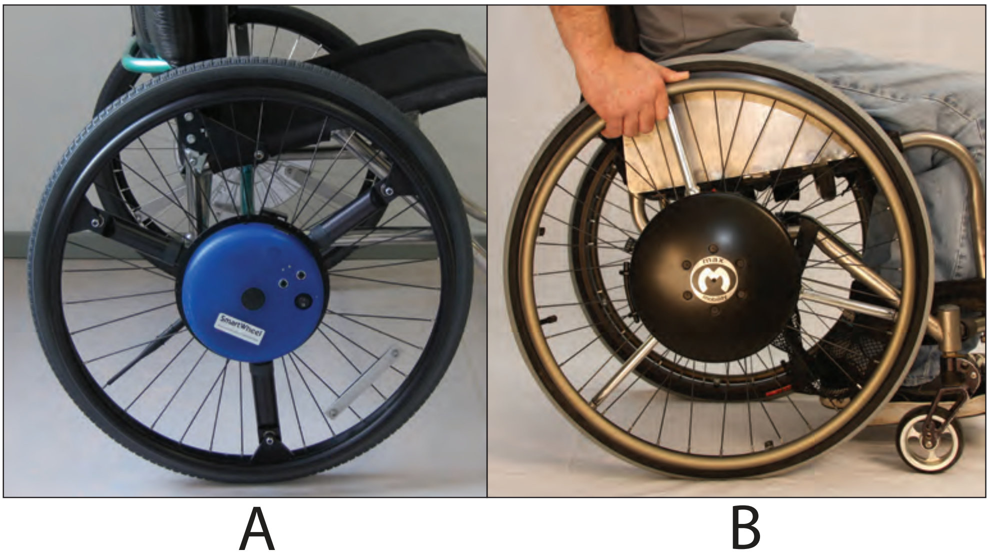

Figure 7.2: Two commercially available measurement wheels, on the left side (A) the SmartWheel and on the right side (B) the Optipush.

7.2 *Tests protocol*

After several meetings with the work group, the following testing were chosen for WHEEL-i: a wheelchair circuit, two questionnaires to assess shoulder pain and self-efficacy of wheeled mobility and a submaximal exercise test on a treadmill to measure the propulsion technique and physical strain. Since it is important to perform the test in a standardized manner on each test occasion (e.g. regarding speed and resistance), a manual with information about the test procedure and the execution of the different test protocols was specifically developed for the WHEEL-i project (to be downloaded from www.scionn.nl). Furthermore, clearly defined explanations for the meaning of each test variable are also provided in the manual, examples of these can be seen in Figures 7.3 and 7.4 and Table 7.1.

Tests were always performed twice to investigate the effect of change over the course of an intervention. When the intervention was a learning or training period then the time between the pre- and post-test could be weeks (Figure 5). When different wheelchair configurations were tested, the time between test conditions was shorter, i.e. on the same day or within a few days, to diminish the possible confounding effect of learning or training on the outcome measures (Figure 5). Prior to all testing, the wheelchair users were screened by a physician for any contra-indications regarding the exercise tests. Furthermore, the wheelchair users signed an informed consent before undertaking any testing.

7.3 *Wheelchair circuit*

The wheelchair circuit (15;16) is a test to assess manual wheelchair skill performance. The research version of the wheelchair circuit consists of 8 different standardized tasks. The 8 tasks are 1) figure-of-8 shape, 2) crossing a doorstep (height, 0.04m), 3) mounting a platform (height, 0.10m), 4) 15-m sprint, 5) transfer, 6) negotiating a 3% slope on a treadmill, 7) negotiating a 6% slope on a treadmill, 8) propelling the wheel-

Table 7.1: Possible outcome variables from measurement wheels

Variable	Description
Push time, s	Time from the start of positive torque to the stop of positive torque for an individual push
Cycle time, s	Time from the start of positive torque to the next start of positive torque
Relative push time, %	Push time expressed as percentage of the cycle time
Push frequency, push/min	Number of complete pushes per minute
Contact-angle, °	Angle at the end of a push minus the angle at the start
Fpeak, N	3D peak force within the push phase
Fmean, N	3D mean force within the push phase
FEFmean, %	Mean fraction effective force
FEFmax, %	Maximum fraction effective force
Slope, Nm/s	Rate of rise from the start of the push phase to the maximum delivered torque around the axle
Negative dip before push phase, N	Minimum torque preceding the push phase
Negative dip after push phase, N	Minimum torque following the push phase
Work/push, J	The power integrated over the push.
POmean push phase, W	Mean power output within the push phase
POmax push phase, W	Maximum power output within the push phase
Power output 2-sided, W	Mean power output of 2 wheels during the sample period
Energy expenditure, W	Calculated from the oxygen uptake and respiratory exchange ratio according to Garby & Astrup (21)
Mechanical efficiency, %	Percentage of internal power used for external power delivered at the wheels

chair for 3 minutes on a treadmill at a constant velocity of 0.56, 0.83, or 1.11 m.s-1, depending on the participant's ability. The wheelchair circuit leads to an ability score and performance time score. All standardized tasks are scored on the ability to perform the task. When the task is performed independently and within a certain time 1 point is assigned, otherwise the score is 0. Three items (crossing a doorstep, mounting a platform, transfer) can also be scored as partially able and can be given half a point. Points for all 8 tasks are summed to give an overall ability score, ranging from 0 to 8. The performance time score is the sum of the performance times of the figure-of-8 shape and the 15-m sprint. Participants are instructed to perform these 2 tasks at their maximum speed. The wheelchair circuit outcome measures have been shown previously to be sensitive to change (17).

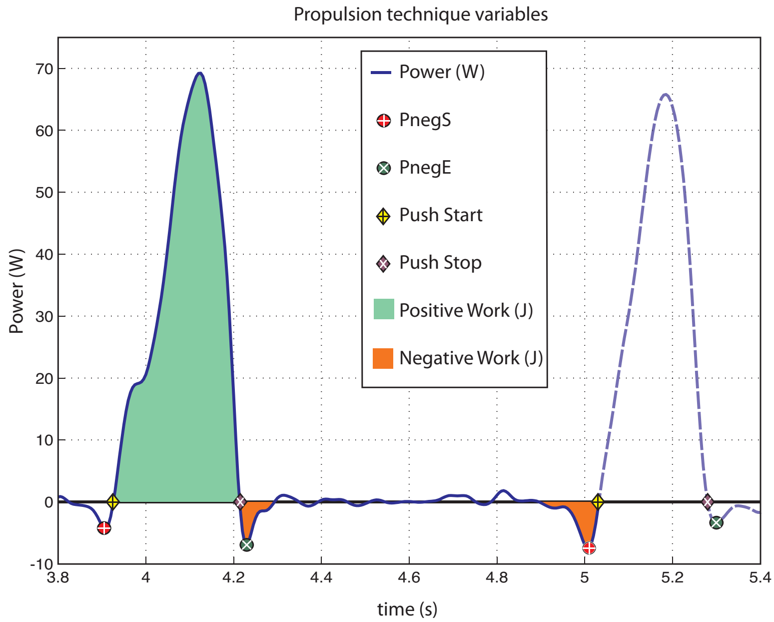

Figure 7.3: Illustration of the definition of push time (from push start to push end), cycle time (from push start to push start), and power loss before (PnegS) and after (PnegE) the push time (27).

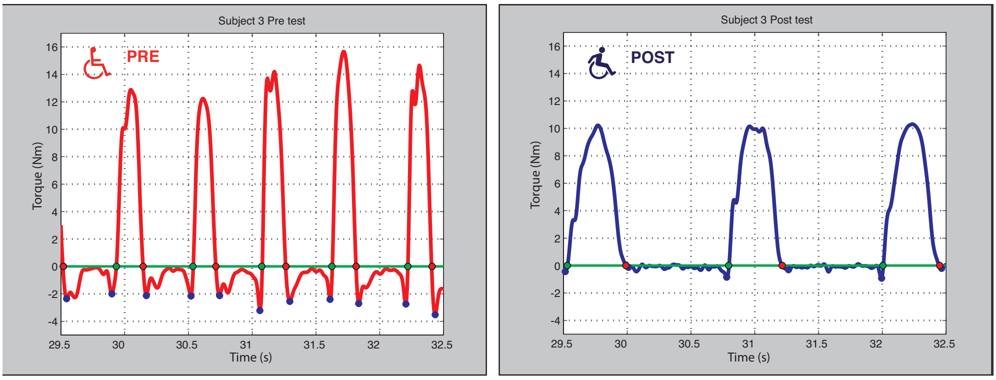

Figure 7.4: Visualised is the torque around the wheel axle. Typical example of changes from pre- to post-test over a learning period (3 weeks, 3 times a week, 80 minutes in total): e.g. a lower push frequency, higher push time and cycle time, less braking torque before and after the push, and a lower peak torque.

7.4 Questionnaires

The Dutch versions of the Wheelchair User Shoulder Pain Index (WUSPI) (18) and Self-Efficacy in Wheeled Mobility Scale (SEWM) (19;20) were selected and administered during the pre- and post tests when a longer (i.e. learning/training) intervention is evaluated. The WUSPI is a 15-item self-report survey specifically designed to assess shoulder pain in wheelchair users during daily functional activities with a 10-point visual analogue scale (18).

The SEWM is a 10-item scale and instructs respondents to rate how confident they are with regard to the performance of specific and general wheeled mobility skills on a 4-point Likert scale (1 = not at all true, 2 = rarely true, 3 = moderately true, 4 = always true) (19;20).

7.5 Steady-state exercise test

With a steady-state exercise test the propulsion technique and physical strain can be determined. It was chosen to perform this test on a treadmill because the velocity and resistance, which both have an effect on propulsion technique and physical strain, can be standardized under different test conditions.

The submaximal exercise test protocol of the Dutch multi-center prospective cohort study 'Restoration of mobility in SCI rehabilitation' was used (14). This protocol consists of two 3-minutes submaximal exercise blocks with 2 minutes of rest in between. Velocity of the exercise blocks is dependent on lesion level as well as overall functional status and is set to 0.56, 0.83 or 1.11 m·s-1. For each person the same test condition is applied on all test occasions.

In the first 3-min exercise block, the person propels the wheelchair with a predetermined velocity and 0º slope of the treadmill. After completion, the person rests for two minutes before starting with the second 3-min exercise block, which is performed at the same velocity and a 0.36º slope of the treadmill. Metabolic cost and heart rate are continuously measured during the exercise blocks with a metabolic cart (Oxycon delta, CareFusion, San Diego, USA) and Polar sport testers (Polar Electro Oy, Kempele, Finland), respectively. Calibration with standardized gases and a 3L volume syringe is performed prior to testing.

During the last minute of each exercise block the forces and torques applied on the right hand rim are measured by the Optipush. The patient is performing the test in his own wheelchair with the Optipush wheel attached (24, 25 and 26 inch options available) on the right side and a regular wheel with extra mass, with similar inertia as the Optipush wheel, on the left side. After the measurement, the Optipush software can automatically generate a report with the averaged values of several propulsion technique variables such as cadence, braking torque, peak force and torque, contact angle and power. This report was used by the rehabilitation professionals.

The energy expenditure (En) is calculated from the oxygen uptake and the respiratory exchange ratio (RER) according to Garby and Astrup (21). Energy expenditure is calculated over the last minute of each exercise block. To obtain the gross mechanical efficiency (ME) of wheelchair propulsion, the ratio power output (PO) / energy expenditure (En) is calculated according to:

1. $ME = PO \cdot En^{-1} \cdot 100\ (\%)$

The power output is calculated from the Optipush, as the product of the torque around the wheel axle and the angular velocity, and expressed as the average power output during the last minute of the exercise block (from start first push until start last push in that minute). This power output is multiplied by two to calculate the overall power output (for two wheels).

The measures of physical strain are the submaximal oxygen uptake, heart rate and ME during the last minute of the exercise blocks.

Table 7.2: Characteristics of the wheelchair–user combination and their influence on rolling friction (26)

Characteristics	Effect on rolling friction
Body mass↑	↑
Wheelchair mass↑	↑
Tyre pressure↓	↑
Wheel size↑	↓
Hardness floor↓	↑
Camber angle ↑	?
Toe-in/out↑	↑↑
Castor shimmy↑	↑
Centre of mass over large rear wheels	↓
Folding frame (vs box frame)	↑
Maintenance↓	↑

7.6 Reference values and smallest detectable difference

To interpret the individual test results, previously collected data were used for calculating reference values and the smallest detectable differences (SDD) of different test outcomes (table 7.2).

Reference values regarding wheelchair skills were calculated from the data of the Dutch multi-center study 'Restoration of mobility in SCI rehabilitation' (22). Ethics approval for the multi-center SCI study was received from the medical ethics committee of SRL/iRv Hoensbroeck. The protocols for the wheelchair circuit (15;16) and gross mechanical efficiency (23) in WHEEL-i were identical to the protocols in the multi-center SCI study. Reference values were calculated as percentiles (20th, 40th, 60th, and 80th

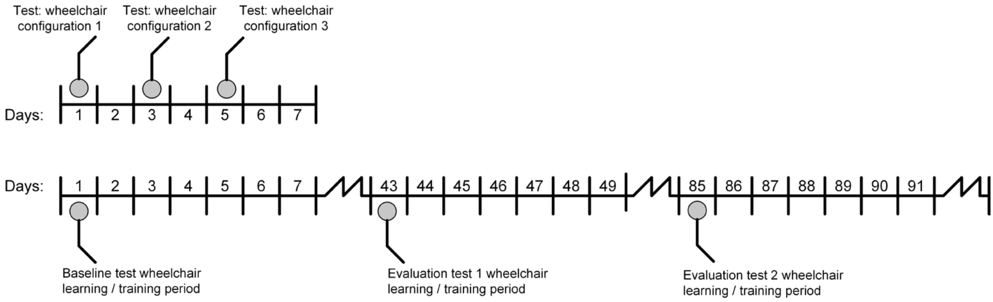

Figure 7.5: Difference in time schedule when testing different wheelchair configurations (within days, upper time line) or the effect of a learning / training intervention (over weeks, lower time line).

percentile) besides the mean and standard deviation for different lesion groups (motor complete and incomplete paraplegia or tetraplegia).

For the calculation of the SDD, data of a study on able-bodied participants was used. This study was approved by the Local Ethics Committee, of the Center for Human Movement Sciences, University Medical Center Groningen, University of Groningen, the Netherlands.

The intraclass correlation (ICC), standard error of the measurement (SEM) and SDD were determined with data of experienced able-bodied person (N=56) who practiced wheelchair propulsion on a treadmill for three weeks (68 minutes in total). Physiological and propulsion technique data were collected after this practice period with a protocol that consisted of three 4 minutes wheelchair propulsion on a treadmill at a power output of 0.18 W/kg with 2 minutes of rest in between. With variance component analysis the ICC, SEM and SDD were calculated (24) when using one exercise test, or two or three exercise tests and using the average.

Participants of both studies signed an informed consent.

RESULTS

7.7 Reference data

For the wheelchair circuit (performance time score (Table 7.3a) and ability score (Table 7.3b)) and gross mechanical efficiency (Tables 7.3c-d) reference values for four test occasions during and after rehabilitation are shown in Table 7.3. The reference values are shown for people with a tetraplegia or paraplegia and with a motor complete or incomplete lesion at the start of active rehabilitation (when people can sit for ≥ 3 hours), 3 months after the start, at discharge of inpatient rehabilitation and 1 year after discharge. Important to note is that not every participant was able to perform the test at each test occasion. The percentage of participants within the lesion groups that was able to perform the test at a specific test occasion is visualised in the table by the column '% all part'. For example, only 10 persons with a complete tetraplegia, which is 19% of all participants with a complete tetraplegia in the study, were able to perform the submaximal exercise block at the start of active rehabilitation (Table 7.3c). So, when a person with a complete tetraplegia is able to perform the exercise test at the start of active rehabilitation that means that he is already very good and the reference data should be used with caution.

Table 7.3: Reference values for persons with SCI for the performance time score of the wheelchair circuit at different times after the start of active spinal cord injury rehabilitation (22).

	n	% all part	20th percentile Excellen–Good	40th percentile Good–Average	60th percentile Average–Fair	80th percentile Fair–Poor	Mean (SD)
Performance time score start active rehabilitation							
Tetraplegia							
Complete	18	34.0	28.8	40.4	48.0	68.4	48.9 (21.9)
Incomplete	16	51.6	22.4	27.0	38.0	66.2	43.3 (28.0)
Paraplegia							
Complete	66	77.7	17.4	21.0	24.2	33.8	26.2 (11.7)
Incomplete	30	83.3	18.0	22.4	28.0	36.8	30.5 (19.5)
Performance time score 3 months after the start of active rehabilitation							
Tetraplegia							
Complete	21	60.0	24.2	31.0	37.6	56.6	39.6 (17.9)
Incomplete	14	51.9	25.0	27.0	33.0	42.0	34.3 (12.7
Paraplegia							
Complete	49	92.5	15.0	17.0	20.0	26.0	20.7 (8.3)
Incomplete	19	95.0	16.0	21.0	26.0	38.0	25.1 (10.3)
Performance time score at discharge of inpatient rehabilitation							
Tetraplegia							
Complete	24	72.7	19.0	25.0	29.0	38.0	28.3 (10.3)
Incomplete	22	73.3	18.0	24.2	27.0	34.8	28.7 (13.1)
Paraplegia							
Complete	64	91.4	14.0	16.0	18.0	22.0	19.1 (7.8)
Incomplete	26	92.9	13.4	15.0	17.0	20.6	17.2 (4.6)
Performance time score at 1 year after discharge of inpatient rehabilitation							
Tetraplegia							
Complete	9	75.0	13.0	29.0	35.0	43.0	30.3 (13.7)
Incomplete	10	71.4	22.0	25.0	30.2	33.8	27.2 (7.2)
Paraplegia							
Complete	39	90.7	13.0	15.0	17.0	19.0	18.9 (10.4)
Incomplete	17	81.0	13.0	15.4	17.0	24.4	19.9 (10.7)

% all part: percentage of participants who were able to perform the test from the total number of participants (within the specific lesion group) of the study. E.g. only 34% of all participants with a complete tetraplegia in our study were able to perform the 15-m sprint and figure-of-8 at the start of active rehabilitation; SD: standard deviation.

Table 7.4: Reference values for persons with SCI for the ability score of the wheelchair circuit at different times after the start of active spinal cord injury rehabilitation (22).

Ability score start active rehabilitation							
Tetraplegia							
Complete	37	69.8	0.0	0.0	2.0	2.7	1.3 (1.3)
Incomplete	20	64.5	1.2	2.0	3.3	4.0	2.7 (1.3)
Paraplegia							
Complete	71	83.5	2.0	3.0	3.6	4.5	3.2 (1.3)
Incomplete	30	83.3	3.0	3.2	4.0	4.4	3.6 (0.8)
Ability score 3 months after the start of active rehabilitation							
Tetraplegia							
Complete	30	85.7	0.0	2.0	2.3	3.4	2.0 (1.6)
Incomplete	20	74.1	0.0	2.2	3.0	3.9	2.3 (1.7)
Paraplegia							
Complete	49	92.5	3.0	4.0	4.5	5.0	4.0 (1.3)
Incomplete	19	95.0	3.0	4.0	4.5	5.0	4.0 (1.0)
Ability score at discharge of inpatient rehabilitation							
Tetraplegia							
Complete	28	84.9	2.0	2.5	3.0	4.5	2.8 (1.4)
Incomplete	27	90.0	1.6	3.6	4.0	4.5	3.3 (1.6)
Paraplegia							
Complete	68	97.1	3.5	4.5	5.0	5.0	4.2 (1.3)
Incomplete	25	89.3	4.5	4.5	5.0	5.0	4.7 (0.4)
Ability score at 1 year after discharge of inpatient rehabilitation							
Tetraplegia							
Complete	10	83.3	2.0	2.7	4.3	4.9	3.4 (1.6)
Incomplete	11	78.6	0.8	2.0	4.1	4.5	3.0 (1.8)
Paraplegia							
Complete	40	93.0	4.0	5.0	5.0	5.0	4.5 (1.1)
Incomplete	17	81.0	4.3	4.5	5.0	5.0	4.7 (0.4)

% all part: percentage of participants who were able to perform the test from the total number of participants (within the specific lesion group) of the study; SD: standard deviation.

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Table 7.5: Reference values for persons with SCI for the gross mechanical efficiency during submaximal exercise block 1 at different times after the start of active spinal cord injury rehabilitation (22). The velocity was set at 0.56 m/s for people with a tetraplegia, at 1.11 m/s for those with paraplegia and if that was too fast, it was set at 0.83 m/s. The velocity was standardized within a patient for all test occasions. The mean power output is shown in the table.

	n	% all part	20th percentile Poor–Fair	40th percentile Fair–Average	60th percentile Average–Good	80th percentile Good–Excellent	Mean (SD)
Gross mechanical efficiency at start of active rehabilitation, mean: 6W							
Tetraplegia							
Complete	10	19	2.1	2.8	3.0	4.1	3.0 (1.1)
Incomplete	11	37	1.8	2.6	3.9	4.0	3.0 (1.1)
Paraplegia							
Complete	54	56	3.4	4.7	5.0	5.4	4.6 (1.4)
Incomplete	28	72	2.5	3.4	4.1	5.6	4.0 (1.8)
Gross mechanical efficiency 3 months after the start of active rehabilitation, mean: 6W							
Tetraplegia							
Complete	16	33	1.6	2.2	3.0	4.6	2.9 (1.4)
Incomplete	6	29	2.4	2.5	2.8	3.3	2.7 (0.4)
Paraplegia							
Complete	52	81	3.9	4.8	5.4	6.3	5.1 (2.0)
Incomplete	18	86	3.1	3.5	4.3	5.1	4.1 (1.2)
Gross mechanical efficiency at discharge of inpatient rehabilitation, mean: 11W							
Tetraplegia							
Complete	23	47	2.2	2.9	3.3	4.2	3.2 (1.0)
Incomplete	12	52	1.8	2.2	3.2	4.1	3.0 (1.4)
Paraplegia							
Complete	67	77	3.7	4.5	5.2	6.1	4.9 (2.1)
Incomplete	28	85	2.8	4.2	4.7	5.7	4.4 (1.5)
Gross mechanical efficiency at 1 year after discharge of inpatient rehabilitation, mean: 10W							
Tetraplegia							
Complete	9	24	2.5	2.8	2.9	5.3	3.4 (1.3)
Incomplete	6	38	1.9	2.0	3.0	5.0	3.0 (1.5)
Paraplegia							
Complete	55	76	3.9	4.9	5.5	6.6	5.3 (2.0)
Incomplete	16	57	3.0	4.3	4.9	6.0	4.5 (1.7)

% all part: percentage of participants who were able to perform the test from the total number of participants (within the specific lesion group) of the study; SD: standard deviation.

Table 7.6: Reference values for persons with SCI for the gross mechanical efficiency during submaximal exercise block 2 at different times after the start of active spinal cord injury rehabilitation (22). The velocity was set at 0.56 m/s for people with a tetraplegia, at 1.11 m/s for those with paraplegia and if that was too fast, it was set at 0.83 m/s. The velocity was standardized within a patient for all test occasions. The mean power output is shown in the table.

	n	% all part	20th percentile Poor–Fair	40th percentile Fair–Average	60th percentile Average–Good	80th percentile Good–Excellent	Mean (SD)
Gross mechanical efficiency at start of active rehabilitation, mean: 8W							
Tetraplegia							
Complete	9	17	2.9	3.6	4.4	5.6	4.1 (1.1)
Incomplete	11	37	2.8	3.1	5.0	5.4	4.1 (1.4)
Paraplegia							
Complete	48	49	5.2	6.2	6.5	7.2	6.3 (1.5)
Incomplete	27	69	4.1	5.1	6.4	7.3	5.7 (1.8)
Gross mechanical efficiency 3 months after the start of active rehabilitation, mean: 9W							
Tetraplegia							
Complete	18	38	3.2	4.1	4.4	5.4	4.3 (1.4)
Incomplete	6	29	3.5	3.7	5.1	5.7	4.5 (1.0)
Paraplegia							
Complete	50	78	5.3	6.2	7.0	7.9	6.9 (3.8)
Incomplete	18	86	4.6	5.3	6.1	6.5	5.9 (2.1)
Gross mechanical efficiency at discharge of inpatient rehabilitation, mean: 16W							
Tetraplegia							
Complete	23	47	3.3	3.7	4.6	5.8	4.4 (1.4)
Incomplete	12	52	3.0	3.4	3.9	5.1	4.0 (1.2)
Paraplegia							
Complete	65	75	5.8	6.6	7.2	8.1	7.3 (3.7)
Incomplete	27	82	4.5	6.1	6.8	7.5	6.2 (1.7)
Gross mechanical efficiency at 1 year after discharge of inpatient rehabilitation, mean: 15W							
Tetraplegia							
Complete	10	27	3.4	4.0	4.4	6.4	3.4 (1.3)
Incomplete	6	38	3.1	3.9	5.1	6.8	4.7 (1.8)
Paraplegia							
Complete	55	76	5.6	6.4	7.5	8.9	7.0 (2.1)
Incomplete	16	57	5.1	6.2	7.4	8.0	6.5 (1.8)

% all part: percentage of participants who were able to perform the test from the total number of participants (within the specific lesion group) of the study; SD: standard deviation.

Table VII. *Intraclass correlation (ICC), standard error of the measurement (SEM) and smallest detectable difference (SDD) of the propulsion technique variables and gross mechanical efficiency (ME) in able-bodied wheelchair users (n=56). The test was performed at 1.11 m/s and a resistance of 0.20 W/kg (18.4 W on average for the whole group)*

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Table 7.7: Intraclass correlation (ICC), standard error of the measurement (SEM) and smallest detectable difference (SDD) of the propulsion technique variables and gross mechanical efficiency (ME) in able-bodied wheelchair users (N=56). The test was performed at 1.11 m/s and a resistance of 0.20 W/kg (18.4 W on average for the whole group).

			SEM			SDD			SDD %			Case		
												24-inch	25-inch	
Variables	Mean	ICC	1 block	2 blocks	3 blocks	1 block	2 blocks	3 blocks	1 block	2 blocks	3 blocks	Mean	Mean	Delta
Push time, s	0.35	0.93	0.02	0.01	0.01	0.05	0.03	0.03	13.80	9.76	7.97	0.48	0.52	0.03
Cycle time, s	1.20	0.88	0.12	0.08	0.07	0.33	0.23	0.19	27.32	19.32	15.78	1.10	1.24	0.14
Relative push time, %	29.82	0.84	1.54	1.09	0.89	4.27	3.02	2.46	14.32	10.12	8.26	44.11	42.22	–1.89
Frequency, pushes/min	53.99	0.91	3.83	2.71	2.21	10.63	7.51	6.13	19.68	13.92	11.36	55.12	49.12	–5.99
Contact angle, °	72.30	0.94	3.34	2.36	1.93	9.27	6.55	5.35	12.82	9.06	7.40	92.00	96.00	4.00
Fpeak, N	71.78	0.87	5.47	3.87	3.16	15.17	10.73	8.76	21.14	14.95	12.21	43.69	43.09	–0.60
Fmean, N	43.58	0.89	2.97	2.10	1.71	8.22	5.81	4.75	18.87	13.34	10.89	28.71	29.44	0.73
FEFmean, %	70.08	0.89	3.62	2.56	2.09	10.02	7.09	5.79	14.30	10.11	8.26	47.10	50.03	2.93
FEFmax, %	101.98	0.81	7.80	5.52	4.50	21.62	15.29	12.48	21.20	14.99	12.24	73.33	78.47	5.14
Slope, n/s	69.38	0.83	7.98	5.64	4.61	22.12	15.64	12.77	31.88	22.54	18.41	26.91	27.14	0.22
Negative dip before push phase, *n*	–2.80	0.72	0.56	0.40	0.32	1.55	1.10	0.90	–55.38	–39.16	–31.97	–3.15	–2.53	0.62
Negative dip after push phase, *n*	–1.16	0.89	0.25	0.18	0.15	0.71	0.50	0.41	–61.15	–43.24	–35.31	–1.19	–0.36	0.82
Work per push, J	11.00	0.92	1.07	0.76	0.62	2.97	2.10	1.71	27.01	19.10	15.59	5.92	6.63	0.71
Power output 2-sided, W	18.40	0.99	0.49	0.34	0.28	1.35	0.96	0.78	7.35	5.20	4.24	10.29	10.34	0.05
Energy expenditure, W	310.19	0.91	18.48	13.07	10.67	51.23	36.23	29.58	16.52	11.68	9.54			
Gross ME	5.99	0.81	0.41	0.29	0.24	1.14	0.81	0.66	19.05	13.47	11.00			

Case: a 25-year-old man with a motor complete lesion at C6. Standardized 3-min propulsion test on a treadmill at 1.11 m/s with 24-inch and 25-inch wheels; Delta: difference in outcome between the 24-inch and 25-inch wheels.

7.8 *Smallest detectable difference*

Table 4 shows the results for the ICC, SEM and SDD of the propulsion technique variables and gross mechanical efficiency in able-bodied wheelchair users.

The ICC varied between 0.72 (negative dip before push phase) and 0.99 (power output two sided) for the propulsion technique variables and was 0.81 for the gross mechanical efficiency and 0.91 for the energy expenditure. When measuring a participant once the individual improvement in propulsion technique variables have to be 14% (push time) or even 61% (negative dip) to conclude that the change is larger than the measurement error.

The effects of design optimization are also shown in Table 7.7. Using the average of two exercise blocks instead of one improves the SDD of, for example, the negative dip after the push phase from 0.71 (SDD%: -61%) to 0.50 (SDD%: -43%). Using the average of three exercise blocks improves the SDD of the propulsion technique variable even further to 0.41 (SDD%: -35%).

DISCUSSION

Over the course of this 1-year implementation project, it clearly showed that there are many factors that determine the success of such a project in a positive (enabling factors) or negative (barriers) sense. As said before, the use of objective standardized measurements to quantify results of rehabilitation is seen as an increasingly important part of good clinical practice. However, the most important factor in this project is the understanding of the test results by the clinical professionals involved. Therefore, below a description of some barriers regarding the interpretation of the test outcomes and the subsequent spin-off projects to help overcome these barriers is given.

7.9 Interpretation of test outcomes

For most tests and outcomes it is very important to get experienced with the testing but even more important with the meaning of the test outcomes. However, clinicians do not have much time to get familiar with the testing and interpretation of the test outcomes. In addition, they are not specifically trained on these (bio)mechanical, ergonomic and/or physiological phenomena. Although it is helpful to educate them by for instance presentations, it is also important that they are involved in testing and the discussion about the outcomes under supervision of skilled embedded human movement scientists. On the other hand, for the researchers it is difficult to draw conclusions based on individual data. Current theory is based on group-based experimental data. As such, normally their conclusions are based on modelling and statistical analyses of these group data. Having individual data and the requirement to judge relevant change, requires at least reference data and/or prediction models.

7.10 Evidence

Based on the wheelchair propulsion studies performed in the last 30 years, general (non-individualized) recommendations can be given for optimizing wheelchair propulsion with respect to physical strain and propulsion technique.

Regarding the physical strain it is important to strive for the highest mechanical efficiency or the lowest oxygen uptake and heart rate at a submaximal steady-state exercise test at the same power output and resistance. With training and learning this lower physical strain can be achieved (25), but also by changing the mechanics of the wheelchair or its ergonomic set-up and fitting (26).

Wheelchair users are able to change their propulsion technique due to natural practice, i.e. by just practicing without getting any intervention. Figure 4 illustrates the changes due to a learning process: at the same power output and velocity, the push frequency will diminish and subsequently the push time, cycle time, contact angle and work per push will increase (27). The power losses before and after the push (negative dips), probably caused by unskilled coupling/uncoupling of hands to the rim, will be lower after learning (27). Furthermore, mean forces and torques and the rate of force application (i.e. the slope) can diminish due to learning (27). Recently, it was found that during the early stage of motor learning all these changes in propulsion technique variables relate to the change in mechanical efficiency, with the percentage negative work per cycle and the contact angle showing the strongest relationship (27).

Regarding wheelchair mechanics and the interface between wheelchair and user, in general it can be recommended to strive for the lowest rolling resistance (thus required power output (PO), which has an effect on both the physical strain and the propulsion technique. Table 7.2 gives an overview of the effects of different wheelchair characteristics on the rolling resistance (26). For example, a higher tire pressure leads to a lower power output and oxygen uptake and to a longer cycle time and contact angle and subsequently lower push frequency (28). Seat height also has an effect on physical strain and propulsion technique. The physical strain seems to be optimal at 100-130 degrees elbow angle while increasing the seat height (smaller elbow angle) leads to lower forces in people with a spinal cord injury (29).

In summary, the recommendations of the Consortium of Spinal Cord Medicine (30) can be followed. They recommend, based on direct and indirect evidence, reducing peak forces, decreasing the rate of application of force and minimizing the frequency of propulsive strokes.

7.11 Reference data

The SmartWheel User Group (SWUG) has described a clinical protocol for the objective assessment of manual wheelchair propulsion (13). Furthermore, they collect the wheelchair propulsion data from the different institutes that use the SmartWheel with that clinical protocol. From this data pool, reference values are generated for several groups, e.g. wheelchair users with a high and low spinal cord injury. These reference values will be very helpful for interpretation of the data of an individual patient when the velocity and power output are known and constant. The data collected during our WHEEL-i protocol will also be pooled and used to calculate reference values in the future. For the wheelchair skills and gross mechanical efficiency we have developed reference values (Tables 7.3a-d) and also prediction models for the wheelchair skills (31) (to be downloaded via www.scionn.nl), based on data of a multi-center study in which patients with a spinal cord injury were followed during and after inpatient rehabilitation.

7.12 Synchronously viewing video

Since clinicians are not used to the biomechanical outcomes of the measurement wheel, it is very helpful to see also a synchronized video of the actual test performance and - when possible - simultaneous of the pre- and post-test. Specific software (MoXie Viewer), built by Out et al. (32), for that purpose is available in the Dutch gait labs. The MoXie Viewer allows synchronously viewing video and concurrently acquired signals such as force and torque data (32)(Figure 7.6). Besides that, also a frontal and/or sagital view of the sitting posture of the wheelchair user, a very important aspect when choosing the optimal wheelchair configuration, can be simultaneously viewed on video and discussed. As a result of the WHEEL-i project the MoxieViewer has been adapted and can now show the Optipush data together with video. Electromyograms (EMG) or joint angles, signals that are often analysed in gait analysis and clearly also of relevance in optimizing wheelchair propulsion (33-35), can be added in the future.

7.13 Smallest detectable difference

To determine whether the intervention has led to a real change in propulsion technique or physiology, it is important to determine the smallest detectable differences (SDD) of the different propulsion technique and physiology test outcomes. We missed this information during the implementation of WHEEL-i. Therefore, we assembled the SDDs from already existing data of wheelchair propulsion in experienced able-bodied persons (Table 7.4). The SEM and SDD of many propulsion technique variables are rather large in the able-bodied group but can be improved substantially by performing two or three exercise blocks and taking the average over the blocks. When taking the average over 3 blocks, the timing variables have to change by 7-16%, the forces by 8-12% and the gross ME by 11% to indicate a real improvement. However, the dips before and

Figure 7.6: The MoXie Viewer: software for synchronously viewing video and concurrently acquired signals such as force and torque data.

after the push have to improve by 32-35% to indicate a real improvement.

For example, we have tested a 25 year old man with a motor complete C6-lesion (height: 1.91m; body mass: 81 kg) while he was propelling his own wheelchair on a treadmill with a set of 24 inch or 25 inch wheels and at a velocity of 1.11 m/s. The test outcomes for both wheel sizes are shown in the last columns of Table 4. Since we have an indication for the SDDs of the propulsion technique variables, we can conclude for this individual wheelchair user that there is an improvement, above the measurement error, in the negative dip after the push phase when using the larger wheel. Of course, we have to be very careful with this interpretation since the SEM and SDD were based on a slightly different protocol and on able-bodied participants, i.e. among others with a good arm/hand function in contrast to our patient. Therefore, a project is set up and performed in collaboration with research groups in Miami, USA and Vancouver, Canada to determine the SDDs of the same variables in wheelchair users with a spinal cord injury performing the WHEEL-i exercise protocol.

Although references values and knowing the SDDs for the propulsion technique variables will be helpful, the interpretation of a good or bad propulsion technique as stated in the 'evidence' paragraph will be a combination of research knowledge and clinical experience. An embedded scientist, who really understands the data, signals and outcomes, together with a clinician and/or rehabilitation technician would make the ideal combination to perform and interpret the tests. Since it is important to get familiar with the tests and the outcomes, not too many testers should be appointed for this.

7.14 Future work

Introducing new measurement techniques in clinical practice automatically triggers new dialogues among researchers and clinicians. In the work groups, specific aspects of the wheelchair prescription process were discussed. There are some guidelines for wheelchair prescription (36-38), which can be followed. However, also new research questions were raised by the clinicians, such as what is the effect of tire pressure, tire profile, wheelchair mass, rear wheel diameter, caster wheel material on propulsion technique and physical strain. To answer these questions, we reviewed the literature and started a research project to answer some of these issues (28).

Another question was whether we could educate patients with the outcomes. By testing the patients multiple times and making them aware how they propel the wheelchair, by showing the video and the test outcomes, these tests might be very helpful to educate patients. Furthermore, the Optipush and SmartWheel and their software also have a feedback function. Biofeedback can be given on all test parameters while the patient is propelling the wheelchair (39). Richter et al. (39) found that biofeedback can be used to improve specific aspects of wheelchair propulsion and may be useful for clinical propulsion training. However, they also pointed out that clinicians should be aware that when training with feedback on a single variable that also other propulsion technique variables will change.

Measurements of wheelchair propulsion technique and physical strain are not only important for patients during rehabilitation. Sports for people with a disability, including wheelchair users, is more and more common. Monitoring the propulsion technique of (elite) wheelchair athletes and evaluating changes in their sport wheelchair and interface, such as in the "Practical guidelines for wheelchair selection in the court sports" (38), can be performed in the wheelchair propulsion lab as well. The overall goal is performance enhancement: getting better, yet staying fit. This applies for new wheelchair users in rehabilitation to elite wheelchair athletes.

7.15 Methodological considerations

It is important to realize that when using measurement wheels, this leads to a 3.5-4.5 kg increase in wheel mass or 7-9 kg in total when using a dummy wheel on the other side with the same mass. In one of our spin-off projects we investigated, among others, the effect of the measurements wheels on power output, propulsion technique and physical strain (28). A higher power output was found when the measurement wheels were attached compared to using regular wheels, which led to changes in physical strain and probably propulsion technique. Furthermore, it has to be kept in mind that for a good force and torque measurement the wheelchair users have to apply the force on the rim and not on the tire, which is often seen. Also the rim coating and the distance between rim and tire can be different compared to the user's own wheels. However, when the test conditions between interventions are exactly the same, i.e. with the same measurement wheels, this does not have to be an issue in terms of pre/post comparability.

Conclusion

After describing and implementing the WHEEL-i test protocol, the largest barrier for systematic monitoring of the individual wheelchair fitting and learning process in rehabilitation with, among others, instrumented measurement wheels was interpretation of the outcomes by professionals. Good interpretation of the outcomes is vital and requires an embedded scientist in rehabilitation who continuously collaborates with science. Other important facilitators are the availability of reference data, knowledge about the smallest detectable difference, and visualisation of the outcomes.

7.16 References

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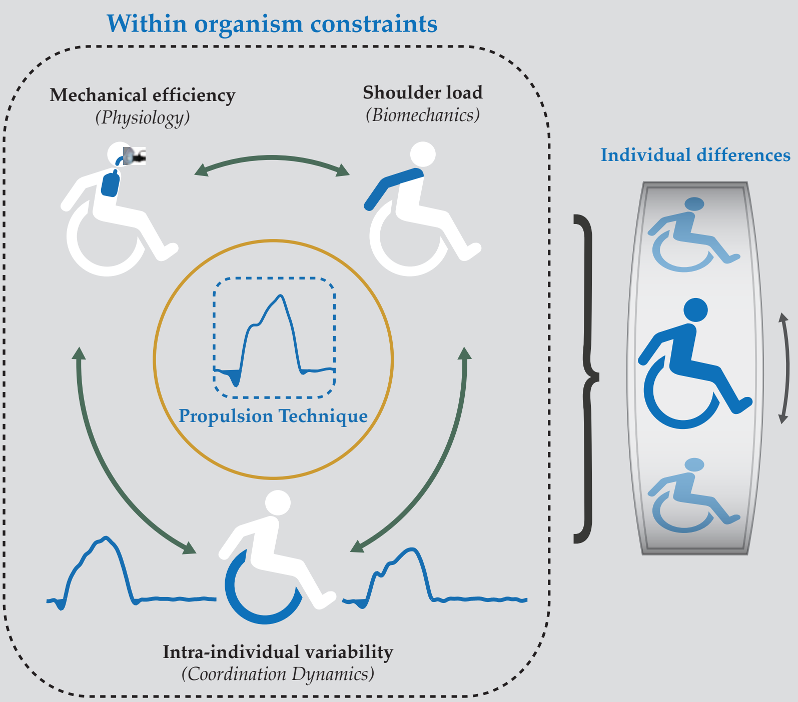
chapter 8

General Discussion



General discussion

The overarching aim of this thesis is to contribute to the participation of wheelchair users, by increasing our understanding of the wheelchair skill acquisition of inexperienced persons. By increasing our insight in the principles that underlie the natural motor learning process of manual wheelchair propulsion in naive individuals, future rehabilitation programs might become more effective and focused. Hopefully this will help wheelchair users to become more proficient at using their upper body for manual ambulation in a wheelchair and therefore increase their mobility and possibly protect them from overuse injury. Other than that the task of wheelchair propulsion can also be viewed as an exemplary cyclic motion often to be learned anew, thus helping to understand more about the learning of cyclic motor skills in a more general context.

The aim of this chapter is to discuss the experimental chapters and reflect on their overall contribution to knowledge about manual handrim wheelchair propulsion and the field of motor learning research. Figure 8.1 shows an expanded conceptual model of within individual constraints on wheelchair propulsion and is an addition to the constraints-based model of coordination and control described in the introduction (fig 1.3, [1]). Since the task and environmental constraints were kept constant we were especially interested in what happens within an individual. The to be learned propulsion technique will be related to several factors, amongst others the energy efficiency,

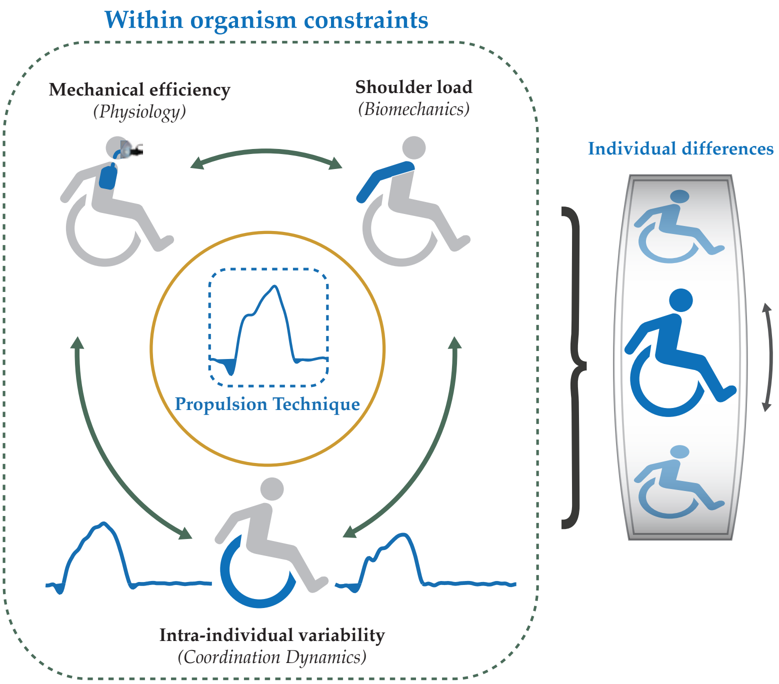


Figure 8.1: In addition to the constraints based model of coordination and control (fig 1.3, [1]) there are several constraints within the organism on coordination that can be distinguished as well. The emergent propulsion technique at any given time will be related to several factors, amongst others the energy efficiency, the strain on the shoulder-complex and the time history of past cycles and planning for future cycles. The chosen solution within the constraints is expected to be dependent on individual differences, such as the pre-existing movement repertoire, body composition and talent of an individual.

the strain on the upper extremities and the time history of past cycles together with the planning for future speed and direction (fig 8.1). The chosen solution within these within organism constraints is expected to be dependent on individual differences, such as the pre-existing movement repertoire, body composition and talent of an individual.

To that end the motor learning process will be discussed in the following order; First, the process of motor learning is inferred from the observable physiological changes measured in our experiments, with mechanical efficiency as the primary outcome measure to describe if the wheelchair user energetically favorable changed his task performance. The second step is to couple this observation to the changes in propulsion technique (i.e. timing, kinetics, kinematics) to evaluate what is being done more energy efficient. These propulsion technique changes were also expected to be of influence on the local strain on the shoulder complex and thus have implications for injury prevention. Third, the individual differences underlying the motor learning results on a group-level are discussed. These relate back to the intra-individual changes in propulsion technique and resulting mechanical efficiency, but are also linked to the fourth section about individual differences in intra-individual variability. The intra individual variability is discussed in this thesis as the possible means of persons to explore the available task solutions within the constraints that bound performance. Finally, some of the clinical implications and future research directions will be discussed, with the Wheel-I project (chapter 7) as a first stepping-stone towards future implementation.

8.1 Motor learning and mechanical efficiency

In our experiments we kept the experimental constraints (more specifically, the wheelchair, environment and task) constant and within that context we observed changes in mechanical efficiency due to changes in performance by the user, because of the natural motor learning process. On a group level chapter 4, 5 and 6 showed the effect of different doses of practice on mechanical efficiency. After 12 and 80 min of practice the participants needed less energy to perform the same task of steady-state wheelchair propulsion at a constant average power output. This is in line with earlier research on natural motor learning of wheelchair propulsion [2-9] and other cyclic motor actions [10,11] and is in accordance with the constraints-based framework of metabolic energy expenditure, motor coordination and control (figure 1.3) [1].

More specifically, our findings support this framework with energy cost as the hypothesized human body's optimization criterion, in different ways: First, almost instantaneous changes in mechanical efficiency in the first twelve minutes were observed, that could also be related to specific propulsion technique changes. Apparently the direct changes in task execution had enough effect to improve the mechanical efficiency of the participants. Thus, we could directly link which learned changes contributed to the reduced energy cost. This was not only apparent within participants as they learned over time, but also in the differences between participants, since the propulsion technique differences between individual were also reflected in their mechanical efficiency, again linking mechanical efficiency to the way the task is executed.

Another argument promoting mechanical efficiency as optimization criterion within our experimental conditions is found in chapter 6. Here we compared the mechanical efficiency to another important outcome measurement, namely the local strain on the shoulder complex. We found that in the very early phase of skill acquisition, improved whole body energy efficiency comes at a cost of increased local load on the shoulder complex during the initial phase of learning. Apparently whole body energy efficiency has priority over mechanical loading in these early stages of learning to propel a handrim wheelchair.

8.2 Motor learning and propulsion technique

Propulsion technique is defined in this thesis as those meaningful properties that can be captured from the measurement wheels and/or position registration in terms of timing and force application. It is thus the observation of how the human body interacts with the mechanics of the wheelchair over time. The laws of physics, the design and anthropometrics of the human body and the assistive technology together determine the optimal energy solution. In the context of motor learning the wheelchair user apparently is driven to find an optimum movement pattern in terms of energy expenditure within these constraints, however it is not at all clear what optimal wheelchair propulsion technique actually is. There are some recommendations about the reduction of push frequency and the use of a semi-circular recovery pattern, but these recommendations are very broad and there is very limited scientific support for these findings so far [12]. Therefore, during the study of motor learning in this thesis we also had to evaluate what good propulsion technique actually is, in relation to the measured optimization criteria.

In relation to mechanical efficiency a desirable change in propulsion technique can be described as changing from a high frequency mode with a lot of negative work to a longer-slower movement pattern with lower power losses at the start and end of the push phase (chapter 4 & 5). Similar changes were found in other cyclical tasks where the reduced energy cost also coincided with an increase in movement amplitude and a decrease of movement frequency, described as a longer-slower movement pattern [10,11,13-15].

On the other hand, from chapter 6 we can conclude that the propulsion technique that is beneficial for mechanical efficiency might negatively impact – at least in the very early stages of skill acquisition - the local mechanical loading of the shoulder complex. Thus from an injury prevention perspective, a different propulsion technique than the adopted longer-slower mode at the end of the motor learning experiments might be better to reduce the load on the rotator-cuff muscles. Here we run into the constraints of the task that might not allow satisfying both energy minimization and reduction of local strain on the shoulder complex. How motor learning relates to the load on the shoulder complex is further discussed below.

8.3 Motor learning and shoulder load

One of the major challenges for wheelchair users is to prevent overuse injuries of the upper extremities, since they rely on their upper body for all activities of daily living, physical activity and daily mobility. Therefore, having a proper wheelchair propulsion technique is thought to be beneficial, and the biomechanical changes in propulsion technique because of motor learning are often speculated to reduce some of the strain on the shoulder complex. However, most studies have evaluated the load on the shoulder complex cross-sectionally. The research described in chapter 6 tried to capture the changes in shoulder load over time as a result of practice. Contrary to our expectation in the early phase of motor learning (12 min) we found an increase rather than a decrease in the activity of the shoulder muscles and glenohumeral compression force after practice, with especially a high strain on the rotator cuff muscles that are vulnerable to injury [16,17].

Whether it is possible to both increase mechanical efficiency as well as reduce the local strain is still unclear. That this might be possible to some extent was exemplified by individual data collected after eight weeks of steady-state wheelchair practice (2x per week 30min at 30% HRR) [18]. This naive able-bodied participant was able to increase in mechanical efficiency, while at the same time reduce glenohumeral contact force (mean and peak) and muscle power (mean and peak). One notable difference compared to the participants of chapter 6 after 12 min of practice was the increased movement amplitude of the trunk over the prolonged practice time. Possibly the increased range of motion of the trunk facilitated a larger contact angle without the need for a large range of motion necessary in the shoulder complex. A possible hypothesis might be that persons solve the motor control problem from distal to proximal and thus the trunk would be one of the last body components to become actively involved with wheelchair propulsion. Yet, even though the glenohumeral contact force and mean muscle power decreased, this still was performed with increased supraspinatus activity, which was recently found to be the most common rotator-cuff tear in an MRI study of manual wheelchair users [19].

However, the biomechanical mechanisms behind shoulder complex damage are still unclear, for instance the dose-response relationship of the shoulder-complex in relation to the mechanical load of wheelchair propulsion is still largely unknown [20]. Future research is needed to understand more about what is 'good' propulsion technique in the context of the load on the shoulder complex. Therefore more longitudinal evaluation of wheelchair users under standardized conditions is necessary, to see how the human motor system adapts propulsion technique to a wheelchair-dependent lifestyle with regard to energy efficiency and loading of the shoulder complex. Such data could be used together with more advanced data analysis techniques to determine what the common propulsion technique characteristics are of proficient wheelchair users, similarly as is currently explored in walking [21]. Also more epidemiological approaches to motor learning might be beneficial, for instance Latent class growth modeling could be used to identify distinct propulsion techniques parallel to shoulder pain trajectories [22].

Another route to gain more insight in how propulsion technique relates to mechanical efficiency and shoulder load might be forward dynamical modeling to see which movement pattern is theoretically optimal and subsequently test if this is possible in real life. However, as such a theoretical pattern might not be preferred over a more energy efficient mode of propulsion, it is also very important to think about optimization and/or redesigning the wheelchair user interface in the context of efficiency as well as mechanical loading to see if both requirements can be addressed at the same time.

8.4 Motor learning and individual differences

Another important finding in this thesis is the clustering of two types of motor learning groups. During the pretest the two groups were split on the basis of a 10% relative increase in mechanical efficiency. It turned out that such a distinction was also reflected in differences in propulsion technique and intra-individual variability, not only during the pretest, but also over prolonged practice. On the one hand the non-improving group started with a higher mechanical efficiency, but did not change much in mechanical efficiency or propulsion technique. On the other hand, the improving group started lower, yet increased more in mechanical efficiency and propulsion technique (chapter 4 and 5). Eventually it was the second group of fast improvers that benefitted most from the extended (3 weeks) practice. Future studies could be designed with appreciation for such individual differences beforehand to further understand the common characteristics of different subgroups like cluster analysis [23] or again latent class growth modeling [24].

An interesting but unanswered question in the current thesis is what caused these individual differences. Probably the motor learning of a new skill also depends on the pre-existing movement repertoire, body composition and talent of an individual [25-27]. There might be a genetic component to motor learning, but additionally motor learning might also in itself be a skill that has to be acquired. For instance, infants who had formerly belly crawled are more proficient crawling on hands and knees than infants who had skipped the belly-crawling period indicating the importance of previous experience [28].

A possibly related observation of chapter 5 was that the group that learned more showed a consistently higher intra-individual variability of their movement pattern. This might be one of the important characteristics relevant for motor learning and is further discussed in the next section.

Motor learning and intra-individual variability

From the experiments a high intra-individual variability was shown between limbs performing the same action (chapter 3) and in one limb repeating a cyclic movement over time (chapter 3 and 5). Since the group that learned more also showed a higher intra-individual variability this is interpreted to not only be the reflection of noise and/or error, but also to be functional and contain features that may provide insight in motor learning. From this perspective, intra-individual variability is seen as a mechanism allowing individuals to explore their movements and eventually to adapt their movements as a function of organismic, environmental and task constraints [29,30].

At the same time chapter 5 was published, another study on motor learning found similar results [31], using a different research paradigm, where they made use of velocity-dependent force-field adaptations in a reaching task. They found that higher levels of task-relevant motor variability predicted faster learning between individuals and that the temporal structure of motor variability was actively reshaped, aligning it with the trained task to improve motor learning. The specific design of their experiment made it possible to determine task relevant motor variability. Which components of the motor variability of wheelchair propulsion are task relevant and which ones would be deemed to be error is currently hard to define. The variability in the propulsion technique measures that are known to change because of practice might be called task relevant, while for instance the inter-limb variability that causes directional changes might be viewed as error. Although the variability was consistently higher in the fast improvers it decreased over time in both groups. This reduction might be viewed as reducing the variability because of error and keeping the task-relevant part.

If the intra-individual (task relevant) variability actually is the info-source that an individual uses to optimize the movement into a more efficient pattern, future studies might try to elicit its potentially functional role during practice. We recently performed an experiment where we tried to elicit extra intra-individual variability through visual feedback of measurement wheel data. Participants received feedback on propulsion technique parameters without being informed about their specific nature [32]. The naïve participants were asked to manipulate the signals presented to them, aimed to help them explore their movement possibilities. This successfully increased the intra-individual variability, but it did rather disrupt their energy optimization process instead of improving it, which was shown by a lack of increase in mechanical efficiency in contrast to a natural learning group. Future studies might look into different ways of supporting the natural motor learning process. Possibly there is an optimum amount of task-relevant variability to improve motor learning and different means to elicit this within participants. Therefore future studies should focus on understanding the task relevant variability of wheelchair propulsion and aim to support this variability by for instance exer-gaming situations with an external focus of attention [33].

8.5 Wheelchair 'gait' analysis

Both our research methodology and the experimental findings are suggested to be of clinical relevance. In chapter 7 we attempted to bring a more systematic evaluation of wheelchair propulsion into the working floor of rehabilitation centers (and adapted sports practice for that matter). Similar to a gait analysis laboratory for assessment of the walking pattern in clinical decision-making (e.g. diagnosis, evaluation and development of interventions), similar methods are suggested to be important for wheelchair-using patient groups. Currently systematic biophysical evaluation of wheelchair fitting, skill acquisition and propulsion technique is not common practice, although some initiatives are being taken [34-36]. To make this possible is both a scientific and a clinical challenge, given the complicated measurement equipment and knowledge about interpretation necessary to measure and interpret biomechanical variables. Also, there is the lack of for instance norm values for patient groups. Yet, in order to prevent overuse injury, such methods might be of crucial importance for choosing the appropriate wheelchair, giving advice on propulsion technique and diagnose people at risk of overuse.

From the science part we need to improve the way information is presented through the proper clinical interfaces and find out what are the critical properties to evaluate. In that sense the figures presented in this thesis were all intended to be relatively easy to interpret. For instance, the shape of torque against wheel angle in a polar plot as shown in chapter 5 and below in figure 8.2 might become a meaningful shape in itself, where in the future we can more easily spot where an individual deviates from an in our view optimal pattern.

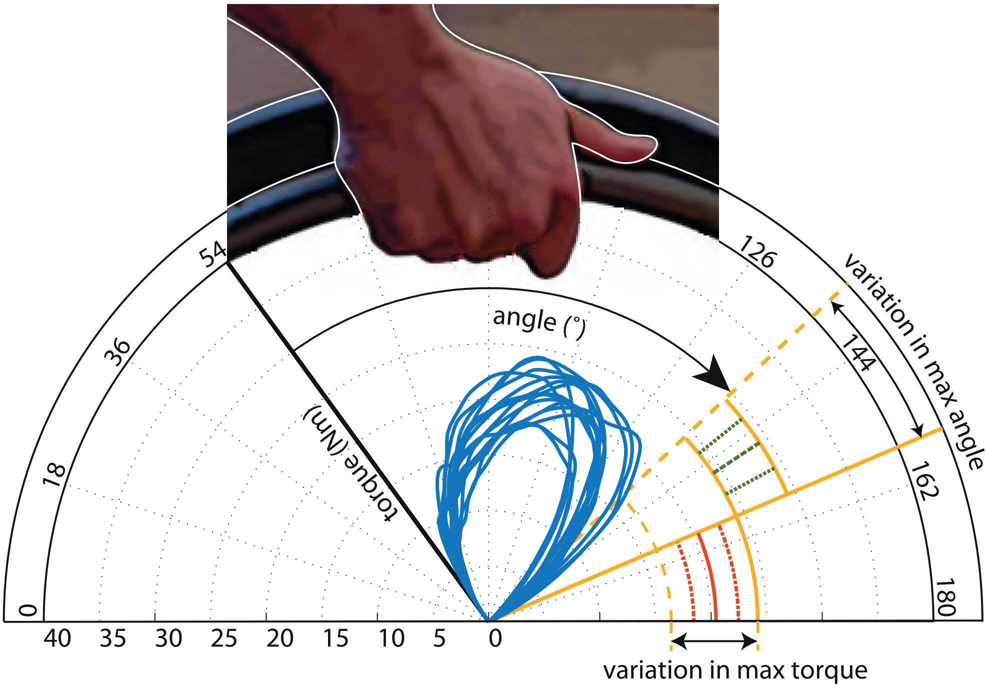


Figure 8.2: Example of the shape of a push-cycle with the torque against the wheel angle in a polar plot. Each shape represents an individual push and by overlaying them differences between cycles can be easily spotted.

An engineering challenge lies in the development of new measurement equipment. The instrumented wheels used in this thesis are still relatively heavy and not that easy and reliable to use in a daily clinical setting. Alternative methods could use for instance a power tap (www.powertap.com) to get a lighter wheel capable of measuring power output, although this would give less detailed information it could be used in a more applied setting and thus form a bridge between lab and field experiments. Another new development is a more advanced ergometer currently in development that can actively simulate rolling resistance and measure power output without the need to replace the wheels of a person's own wheelchair (www.umaco.nl).

On the other hand, the proper implementation will not be possible without getting some of the lab environment into the clinical field. Hopefully within the high pressure on rehabilitation centers, there is room to educate rehabilitation physicians, physical therapists or specialized trained lab-support members in the carrying out of such systematic wheelchair propulsion analyses and the following decision-making based on these data.

8.6 *Methodological considerations*

The translation of the results presented in this thesis to clinical practice should be done carefully, taking the limitations of our work into account. Shortly stated, we measured able-bodied young participants to drive at a constant speed on a motor-driven treadmill at low intensity. How this relates to the generally older population of wheelchair-dependent users in their daily life is uncertain. In that sense we do not aim to interpret the propulsion technique changes literally. Rather the principles underlying motor learning and how individuals explore their available options and subsequently optimize their skills within their own constraints are thought to be critical.

Currently our work should ideally be continued within the rehabilitation centers to get a better understanding of how wheelchair users become accustomed to a wheelchair and evaluate the existing training protocols on their effects. From there hopefully improved understanding will lead to more efficient and effective training protocols that support the wheelchair skill acquisition of participants, so that they may learn to propel in an energy efficient and sustainable way.

8.7 Future perspectives

As previously discussed the results on wheelchair skill acquisition in this thesis have a theoretical and an applied component. In both fields much work is still to be done to better understand the abstract construct of motor learning and its meaning for clinical practice.

From a theoretical perspective it is important to continue the research as performed in chapter 6, where from multiple perspectives is looked at the motor learning process because of practice. As such the conceptual model put forward in this discussion might form the basis for future experiments focused on better understanding the optimization criteria of the human system. A first next step in this direction is to continue these experiments over a longer dose of practice. By simultaneously evaluating mechanical efficiency and shoulder load over longer practice time we might gain a better understanding on how these relate to each other and what 'good' is in the context of the bodies internal motor learning optimization criteria and from our own clinical perspective on injury prevention.

A second lab-based perspective is found in more detailed analysis in the time structure of variability and changes in coordination dynamics. Since wheelchair propulsion is a cyclic skill the variations in propulsion technique from cycle to cycle are expected to be time dependent; previous cycles will influence the present cycle, which at the same time might already be planned towards future events. Therefore, more advanced mathematical methods like principle component analyses [37] might give more direction on how variability is used and changes over time in relation to motor learning.

From a clinical perspective more embedded research in the clinical setting is necessary towards more evidence-based training programs. More explicit programs focused on the acquisition of such a crucial skill as wheelchair propulsion might benefit future wheelchair users. To make such programs become a reality challenges lie in the systematic way of implementation, evaluation and consequent decision-making. Developing norm values for wheelchair users and understanding what changes are clinically meaningful will help monitoring individual wheelchair users and might also find its way back to our broader understanding of wheelchair skill acquisition.

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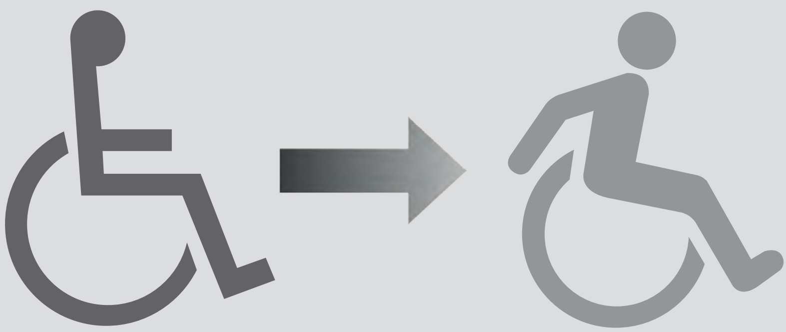
Summary

Long Summary



Summary (long)

The process of skill acquisition is a key element of human functioning during daily life and an essential part of rehabilitation after disease or injury. In the process of rehabilitation people need to adapt to new situations constantly, like learning to walk with a leg-prosthesis or, as in the case of this thesis, learning to use the upper-body for ambulation while seated in a wheelchair. The research described in this thesis aims to increase our knowledge about the acquisition of wheelchair propulsion technique for the rehabilitation setting and to improve our understanding of natural motor learning processes.

Chapter 1 introduces the central hypothesis that because of practice persons learn to adopt a propulsion technique that minimizes the energy cost within the constraints of the task, environment and individual. This is tested in the following experimental chapters by evaluating the short-term effect (4-80 minutes) of steady-state low-intensity handrim wheelchair practice by novice able-bodied subjects, without providing instruction or feedback. The energy expenditure, propulsion technique, shoulder load and measures of variability were continuously measured during the experiments.

Chapter 2 gives a literature overview about the optimization possibilities for the wheelchair-user combination on three levels. On the level of the user one can optimize physical capacity and technique by training. The second level focuses on the wheelchair-user interface, i.e. the interaction between the musculoskeletal system and the propulsion mechanism, aiming for a higher efficiency operationalized as a better ratio of internal power from the user to external power required for propulsion. Finally, at the level of the wheelchair the focus lies on minimizing power loss of the wheelchair-user system by reducing frictional forces. To advance design and performance, better insight in the working mechanisms of our biological system in combination with assistive technology, such as the wheelchair, is necessary.

Chapter 3 compares the outcomes of two different measurement-wheels (Smartwheel and Optipush) attached to the different sides of the wheelchair. A good agreement between both measurement-wheels was found at the level of the power output. This indicates a high comparability of the measurement-wheels for the different propulsion parameters. Data from both wheels seem suitable to be used together or interchangeably in experiments on motor control and wheelchair propulsion performance. A high variability in forces and timing between the left and right side were found during the execution of this bimanual task, reflecting the human motor control process.

Chapter 4 evaluates mechanical efficiency and propulsion technique during the initial stage of motor learning. Therefore, 70 naive able-bodied men received 12-minutes uninstructed low-intensity wheelchair practice on a motor driven treadmill. Participants significantly increased their mechanical efficiency and changed their propulsion technique from a high frequency mode with a lot of negative work to a longer-slower move-

ment pattern with less power losses. Furthermore a multi-level model showed propulsion technique to relate to mechanical efficiency. Finally improvers and non-improvers were identified. The non-improving group was already more efficient and had a better propulsion technique in the first block of practice (i.e. the 4th minute). These findings link propulsion technique to mechanical efficiency, support the importance of a correct propulsion technique for wheelchair users and show motor learning differences.

Chapter 5 studies the natural motor learning process and individual differences over prolonged low-intensity wheelchair practice (80 min) on a motor driven treadmill. The initially fast improvers benefitted more from the given practice indicated by a better propulsion technique (like reduced frequency and increased stroke angle) and a higher mechanical efficiency. The initially fast improvers also had a higher intra-individual variability in the pre and posttest, which possibly relates to the increased motor learning of the initially fast improvers. Further exploration of the common characteristics of different types of learners will help to better tailor rehabilitation to the needs of wheelchair-dependent persons and improve our understanding of cyclic motor learning processes.

Chapter 6 investigates the changes in shoulder load because of the initial phase of low-intensity wheelchair practice on a motor driven treadmill. It appears that during the early stages of motor learning in handrim wheelchair propulsion increased mechanical efficiency comes at the cost of an increased muscular effort and mechanical loading of the shoulder complex. This seems to be associated with an unchanged stable function of the trunk and could be due to the early learning phase where participants still have to learn to effectively use the full movement amplitude available within the wheelchair-user combination. Apparently whole body energy efficiency has priority over mechanical loading in the early stages of learning to propel a handrim wheelchair.

Chapter 7 describes the enabling factors and barriers experienced in the "Wheelchair Expert Evaluation Laboratory – implementation" (WHEEL-i) project, in which scientific knowledge, tools and associated systematic analyses of hand rim wheelchair propulsion technique, the user's wheelchair propulsion capacity, the wheelchair-user interface as well as the wheelchair mechanics were implemented in two rehabilitation centers. Based on pilot results, the largest barrier for systematic monitoring of the individual wheelchair fitting and learning process in rehabilitation with, among others, instrumented measurement wheels was the interpretation of the outcomes. For proper interpretation of individual outcomes, the availability of reference data, SDDs, and visualisation of the outcomes is of utmost importance.

Chapter 8 discusses the most important conclusions of this thesis. The central hypothesis was that because of natural practice under controlled submaximal conditions novice persons learn to adopt a propulsion technique that minimizes the energy cost within the constraints of the task, environment and individual. This is supported by the reduced energy expenditure because of practice, related to propulsion technique. The discussion goes on to suggest that the to be learned propulsion technique will relate to several factors within an individual, amongst others the energy efficiency, the strain on the upper extremities and the time history of past cycles together with the planning for future speed and direction. The chosen solution is expected to be dependent on individual differences, of which the characteristics are an important future research direction. For instance, the higher movement variability of the group that learned more might be an important characteristic to identify different learning types. Attention for motor learning processes and the systematic evaluation of propulsion technique, during clinical rehabilitation, daily life and adapted sports are of great importance to enhance mobility and possibly help protect wheelchair-users from upper body overuse injury.

Summary

Short Summary

Summary (short)

Wheelchair users depend on their upper body for mobility during daily life. However, handrim wheelchair propulsion is a physically straining form of ambulation as a consequence of a low mechanical efficiency and a high mechanical load on the shoulder complex. The research described in this thesis aims to increase our knowledge about the acquisition of wheelchair propulsion technique for the rehabilitation setting and to improve our understanding of natural motor learning processes.

The experiments in this thesis have shown that through practice participants change their propulsion technique, consequently resulting in a lower energy-expenditure. Special attention has been paid to individual differences and the importance of functional variability in the propulsion technique of individuals during practice. Furthermore, biomechanical analysis showed that contrary to the reduced energy expenditure the local load on the shoulder complex increases. Apparently whole body energy efficiency has priority over mechanical loading in the early stages of learning to propel a handrim wheelchair.

Finally we have attempted to translate some of our insights and methods to clinical practice, towards more evidence-based decision-making. Attention for motor learning processes and the systematic evaluation of propulsion technique, during clinical rehabilitation, daily live activities and adapted sports, are of great importance to enhance the mobility of wheelchair users and possibly protect them from overuse injury.

Samenvatting

Uitgebreide samenvatting

Uitgebreide Samenvatting

Rolstoelgebruikers zijn afhankelijk van hun bovenlichaam voor mobiliteit in het dagelijks leven, maar rolstoelrijden heeft een lage mechanische efficiëntie en overbelasting komt veel voor. Het doel van dit onderzoek is daarom bij te dragen aan kennis over en verbetering van het motorische leerproces en de rolstoelvaardigheden van rolstoelgebruikers, om zo mogelijk hun mobiliteit te vergroten en overbelasting te verminderen.

Hoofdstuk 1 introduceert de hypothese dat personen door laag intensief te oef enen zonder specifieke feedback of instructie uit zichzelf op zoek gaan naar de aandrijftechniek die minder energie kost binnen de grenzen van de taak, de omgeving en het individu. Dit hebben we in de volgende hoofdstukken bestudeerd door niet-rolstoelafhankelijke personen zonder instructie te laten oefenen op een loopband met een constante snelheid (1.11 m/s) en gestandaardiseerd relatief vermogen (0.2 W/kg). Daarbij hebben we geanalyseerd hoe mensen over de tijd veranderen in hun energieverbruik, aandrijftechniek, schouderbelasting en de variabiliteit in de uitvoering. De tijdsduur van het oefenen varieerde van 4 tot 80 minuten.

Hoofdstuk 2 geeft een literatuuroverzicht van de optimalisatiemogelijkheden van de rolstoelgebruiker combinatie. Met name aspecten van de rolstoel en de koppeling van de rolstoel met de gebruiker worden besproken. Op deze manier wordt de bredere context van optimalisatie geschetst, waarnaar in de rest van dit proefschrift de gebruiker zelf centraal komt te staan terwijl de technologie constant wordt gehouden.

Hoofdstuk 3 vergelijkt de twee gebruikte geïnstrumenteerde wielen (Smartwheel en Optipush), die de krachten en momenten kunnen meten die de gebruiker op de hoepel levert. Door deze wielen gelijktijdig links en recht op de rolstoel te plaatsen, terwijl proefpersonen op constante snelheid en vermogen op de loopband rolden, konden de uitkomsten van beide meetwielen worden vergeleken. Deze studie vond een goede vergelijkbaarheid van de meetwielen en illustreerde de grote variabiliteit tussen de linker en rechter arm als je deze gelijktijdig bestudeerd, terwijl gemiddeld over de tijd de uitkomsten niet verschilden, omdat er in een rechte lijn op de loopband gereden moest worden.

Hoofdstuk 4 is de eerste experimentele leerstudie. Hier hebben we tijdens de eerste 3x4 minuten van rolstoelrijden het initiële leerproces van niet rolstoelafhankelijke personen bestudeerd. Bij 70 personen zijn tijdens het oefenen het energieverbruik en het aandrijftechniek geanalyseerd. Door laag intensief te oefenen op constante snelheid en vermogen werd men energiezuiniger tijdens het rolstoelrijden. Gelijktijdig werden veranderingen in de aandrijftechniek waargenomen. Deze veranderingen konden hoofdzakelijk beschreven worden als een langzamer bewegingsritme waarbij met een verlaagde duwfrequentie meer arbeid wordt geleverd over een verlengd contact met de hoepel en met minder verlies tijdens het aan- en afkoppelen van de hand met de hoepel. Door multipele regressieanalyse bleken energieverbruik en de verandering in de aandrijftechniek aan elkaar gerelateerd. Naast deze groepsobservaties werden in een nadere analyse twee groepen geclassificeerd op basis van de reductie in energieverbruik, de initieel langzame verbeteraars en de initieel snelle verbeteraars. Deze twee groepen bleken ook te verschillen in de verandering in de aandrijftechniek, waarbij de initieel langzame verbeteraars energetisch efficiënter begonnen maar nauwelijks veranderden, terwijl de initieel snelle verbeteraars minder efficiënt begonnen en sneller vooruitgingen.

Hoofdstuk 5 bestudeert het effect van langer oefenen in de context van de resultaten uit hoofdstuk 4. Hier bleek dat de eerder gevonden groepsverschillen doorwerken over een langere oefenperiode van in totaal 80 minuten. Over de totale oefenduur profiteerde de initieel snelle verbeteraars meer van de aangeboden oefening en eindigden met een lager energieverbruik en een meer veranderde aandrijftechniek. Tevens waren de initieel snelle verbeteraars variabeler in hun aandrijftechniek dan de initieel langzame verbeteraars, mogelijk is dit een eigenschap die motorisch leren bevordert.

Hoofdstuk 6 gaat in op de schouderbelasting tijdens het initiële leerproces. Door positieregistratie van het bovenlichaam te koppelen aan de aandrijftechniek afgeleid uit de meetwielen kon het 'Delftse Schouder en Elleboog Model' worden gebruikt voor een invers dynamische analyse en een modelmatige voorspelling van de belasting rond het schoudergewricht. Hieruit bleek dat door oefening in de eerst 12 minuten de mechanische belasting op de schouder groter werd en met name de m. Supraspinatus meer werd belast, een spier die gevoelig is voor overbelasting. Opvallend was dat in deze fase van het leren een verlaagd energieverbruik samengaat met een hogere lokale belasting van de schouder.

Hoofdstuk 7 poogt de in het lab ontwikkele methoden en verkregen kennis te vertalen naar de praktijk. Doel hierbij was te onderzoeken wat de barrières waren voor systematische evaluatie van rolstoelpassing en -training in de revalidatiepraktijk. De belangrijkste barrières waren met name de interpretatie van de gemeten waarden. Voor individueel advies is het belangrijk dat er normwaarden worden ontwikkeld, dat er meer kennis is over wat klinische relevante veranderingen zijn en dat er goede visualisatie mogelijk is om de meetwaarden met de revalidant en revalidatieprofessionals op inzichtelijke wijze te bespreken.

Hoofdstuk 8 bediscussieert de conclusies van dit proefschrift. Het blikt terug op de hypothese dat personen door te oefenen uit zichzelf op zoek gaan naar de aandrijftechniek die het minste energie kost, binnen de grenzen van de taak, de omgeving en het individu. Dit wordt door dit proefschrift ondersteund door het op groepsniveau gereduceerde energieverbruik als gevolg van oefenen. Als extra suggestie, op basis van de resultaten in dit proefschrift, wordt een dergelijk conceptueel model nogmaals herhaald voor optimalisatie binnen het individu. De gebruiker moet een aandrijftechniek vinden binnen de grenzen van energieverbruik, lokale (over)belasting en de cyclische structuur van rolstoelrijden. Tussen personen zullen deze grenzen verschillen en toekomstig onderzoek zal moeten proberen de gemeenschappelijke karakteristieken van verschillende leergroepen beter te duiden.

Een mogelijk voorbeeld hiervan zijn de verschillen in mate van variabiliteit. De variabiliteit bleek hoger in de groep die meer leerde; mogelijk is dit een eigenschap die mede bepalend is voor het leerproces en groepen van elkaar onderscheid. Toekomstig onderzoek zou kunnen proberen die variabiliteit beter te begrijpen en mogelijk te gebruiken voor meer effectieve oefenprotocollen.

Uit verdere biomechanische analyse blijkt bovendien dat, in tegenstelling tot het verlaagde energieverbruik, de lokale belasting van het schoudergewricht tijdens het beginnende leerproces omhoog gaat. Deze ogenschijnlijke contradictie tussen systeem efficiëntie en lokale belasting behoeft toekomstig onderzoek om te zien hoe ook de schouderbelasting geminimaliseerd kan worden.

Aandacht voor motorische leerprocessen en het systematisch evalueren van de vaardigheid van de rolstoelgebruiker, tijdens zowel de klinische revalidatie als tijdens het dagelijks leven en aangepaste sport, is van groot belang om hen beter te kunnen beschermen tegen overbelasting en prestaties te bevorderen..

Samenvatting

Korte samenvatting

Korte Samenvatting

Rolstoelgebruikers zijn afhankelijk van hun bovenlichaam voor mobiliteit in het dagelijks leven, maar rolstoelrijden heeft een lage mechanische efficiëntie en overbelasting komt veel voor. Het doel van dit onderzoek is daarom bij te dragen aan kennis over en verbetering van het motorische leerproces en de rolstoelvaardigheden van rolstoelgebruikers, om zo mogelijk hun mobiliteit te vergroten en overbelasting te verminderen.

De experimenten in dit proefschrift hebben laten zien dat personen door laag intensief zonder specifieke feedback te oefenen in staat zijn hun aandrijftechniek aan te passen en zo hun energieverbruik te verlagen. Extra aandacht is er hierbij voor de gevonden individuele verschillen tussen personen en het mogelijk belang van variabiliteit in de aandrijftechniek tijdens het oefenen. Uit verdere biomechanische analyse blijkt bovendien dat, in tegenstelling tot het verlaagde energieverbruik, de lokale belasting van het schoudergewricht tijdens het beginnende leerproces omhoog gaat. Deze ogenschijnlijke contradictie tussen systeem efficiëntie en lokale belasting behoeft toekomstig onderzoek om te zien hoe ook de schouderbelasting geminimaliseerd kan worden.

Tot slot werd onderzocht hoe systematische evaluatie van aandrijftechniek in de praktijk zou kunnen worden gebruikt om tot meer 'evidence-based practice' te komen, om zo aanpassingen aan de rolstoel en trainingen voor de gebruiker beter te kunnen onderbouwen. Dit is van groot belang om de hele rolstoel/gebruiker combinatie beter te kunnen optimaliseren, zowel tijdens de klinische revalidatie als tijdens het dagelijks leven en aangepaste sport.

Curriculum Vitae

CV Riemer Vegter

Curriculum Vitae

Riemer JK Vegter was born in Groningen, the Netherlands on the 16th of March 1981 as the first child of Jaap JR Vegter en Janine Vegter-Philbert. He finished high school at the Wessel Gansfort College in Groningen, in 1999. Interested in design and technology he started an education in Architecture, however he changed direction to Human Movement Sciences in 2001.

In 2006 he obtained his MSc in Human Movement Sciences at the University of Groningen, with a major in Rehabilitation Sciences and an internship on the functional modeling of multi-axis prosthetic knee joints under the supervision of Prof. Dr. Bert Otten. This work resulted in an application for PhD-funding to the open technology program of the Dutch Science organization (NWO). Awaiting the funding approval he got the chance to become a lecturer at Human Movement Sciences for the courses Mathematics and Matlab programming in 2007.

In 2009 he switched to the PhD research of the current thesis. The earlier funding was not approved, but a new chance was presented on the motor learning of wheelchair propulsion under the supervision of Prof. Dr. Luc van der Woude, Prof Dr. Dirk-Jan Veeger, Dr. Claudine Lamoth and Dr. Sonja de Groot. This resulted in a combined position of lecturer Mathematics (0.2 FTE) and PhD-student (0.8 FTE). During this period he was able to obtain his University Teaching Qualification and finish the educational program of the Behavioral and Cognitive Neurosciences Graduate School. Since 2014 he has a faculty position as lecturer Mathematics and Biomechanics, which will be continued as Assistant Professor after his successful defense of the current thesis. His research will continue on biophysical aspects of wheeled-mobility during rehabilitation, daily life and adapted sports.

Besides his professional interest in Human Movement Sciences his passion for rock climbing serves as an ongoing introspective study of solving complicated movement problems.

Publications

1 2015 van der Scheer JW, de Groot S, **Vegter RJK**, Hartog J, Veeger HE, van der Woude LH. Low-intensity wheelchair training in inactive people with a long term spinal cord injury: a randomized control trial on propulsion technique. *Am J Phys Med Rehabil, in press.*

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Submitted:

1. 2015 Leving MT, **Vegter RJK,** Hartog J, Lamoth CJ, de Groot S, van der Woude LH. Effects of Visual Feedback-Induced Variability on Motor Learning of Handrim Wheelchair Propulsion. *PLoS ONE, under review*

Conferences

- 2014 Dutch Congress of Rehabilitation Medicine, Rotterdam. ***Oral presentation.***
- 2014 International Congress on Complex Systems in Sports and Healthy Ageing, Groningen. ***Oral presentation***
- 2014 International Spinal Cord Society, Maastricht. ***Poster presentation.***
- 2014 Symposium 'Disability is not Inability', Groningen. ***Oral presentation***
- 2014 Rehabilitation: 'Mobility, Exercise & Sports' Groningen. ***Oral presentation.***
- 2014 VvBN Annual meeting Utrecht. ***Oral presentation.***
- 2013 Vista conference of the International Paralympic Committee (IPC), Bonn. ***Oral presentation***
- 2012 Annual conference of the Netherlands Society of Physical and Rehabilitation Medicine, Noordwijkerhout. ***Oral presentation***
- 2012 Smartwheel user-group conference, Chicago. ***Oral presentation***
- 2012: Symposium Innovation in Rehabilitation, Oegstgeest. ***Poster presentation***
- 2011 AAATE conference of the Association for the Advancement of Assistive technology in Europe. ***Oral presentation***
- 2011: Symposium Movement for Life of the University Medical Center Groningen, Groningen. ***Oral presentation***
- 2011: Symposium Physical strain, work capacity and mechanisms of restoration of mobility in the rehabilitation of individuals with spinal cord injury, Groningen. ***Oral presentation***
- 2011: Skills conference of the SKILLS European Project (IST, Multimodal Interfaces, 2007-2011), december 2011 Montpellier. ***Poster presentation***
- 2010 Annual conference of the Netherlands Society of Physical and Rehabilitation Medicine, Noordwijkerhout. ***Oral presentation***

Dankwoord

Dankwoord

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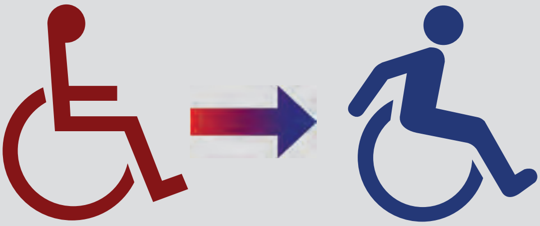

Wheelchair users depend on their upper body for mobility during daily life. However, handrim wheelchair propulsion is a physically straining form of ambulation as a consequence of a low mechanical efficiency and a high mechanical load on the shoulder complex. The research described in this thesis aims to increase our knowledge about the acquisition of wheelchair propulsion technique for the rehabilitation setting and to improve our understanding of natural motor learning processes.

The experiments in this thesis have shown that through practice participants change their propulsion technique, consequently resulting in a lower energy-expenditure. Special attention has been paid to individual differences and the importance of functional variability in the propulsion technique of individuals during practice. Furthermore, biomechanical analysis showed that contrary to the reduced energy expenditure the local load on the shoulder complex increases. Apparently whole body energy efficiency has priority over mechanical loading in the early stages of learning to propel a handrim wheelchair.

Finally we have attempted to translate some of our insights and methods to clinical practice, towards more evidence-based decision-making. Attention for motor learning processes and the systematic evaluation of propulsion technique, during clinical rehabilitation, daily live activities and adapted sports, are of great importance to enhance the mobility of wheelchair users and possibly protect them from overuse injury.

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