Development of a wheelchair propulsion laboratory

R. de Klerk ○ Development of a wheelchair propulsion laboratory

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Research Institute
SHARE

Propositions belonging to the thesis

Development of a wheelchair propulsion laboratory

Rick de Klerk

1. The heterogeneity of lab-based equipment and the lack of consensus in the field of wheelchair propulsion research limit the generalizability of results and the applicability of the existing knowledge base. – this thesis

2. The measurement of power output is critical to the study of handrim wheelchair propulsion. – this thesis

3. The implementation of the Esseda – an instrumented servo-driven wheelchair roller ergometer – in wheelchair propulsion laboratories will benefit future research, rehabilitation, and sports practice. – this thesis

4. A low-cost and portable calibration system is essential for stable and valid calibration testing within and between ergometers. – this thesis

5. The Esseda is a valid alternative to overground and treadmill testing. – this thesis

6. Wheelchair racing practice on an ergometer improves efficiency and technique in novices, but consistent training, guidance, and feedback are needed to reach those of an experienced athlete. – this thesis

7. Optimization of power-assisted wheelchairs requires algorithms that account for individual propulsion characteristics, estimates of shoulder loading, and environmental conditions. – this thesis

8. Merely quantitative differences beyond a certain point pass into qualitative changes. – Karl Marx

9. If the implementation is hard to explain, it's a bad idea. If the implementation is easy to explain, it may be a good idea. – Tim Peters

Development of a wheelchair propulsion laboratory

Rick de Klerk

Colophon

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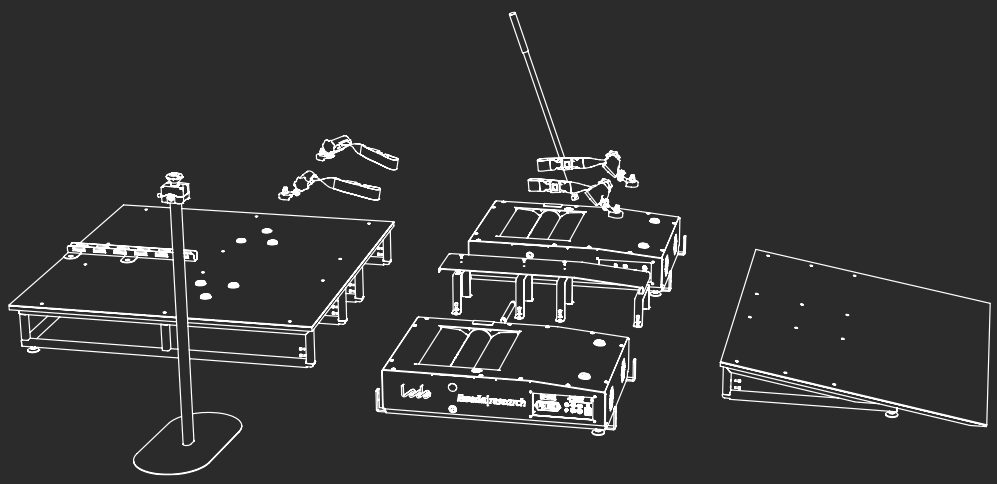
General introduction



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Background

Manual wheelchair propulsion has fascinated scientists since the late sixties due to its integration of the human and technical sciences [1,2]. Indeed, Voigt and Bahn (1969) [3] and Brättgard et al. (1970) [4] already stressed the strenuous character of handrim wheelchair propulsion and research has been going on ever since. Research on the topic of wheelchair propulsion is, however, convoluted by the heterogeneity of the population, the limited availability thereof, and the specialized equipment and methods that are required [5]. Nevertheless, the systematic study of wheelchair propulsion has led to a wealth of knowledge on the cardio-respiratory responses to this mode of exercise [6–9], the training in rehabilitation and sports [10–13], and wheelchair design [14]. Early research mainly concerned the wheelchair and the wheelchair user separately, while later research also emphasized the ergonomics perspective and the wheelchair-user interface [15].

As wheelchair propulsion is a cyclic movement, a power balance [16] can be applied to illustrate the flow of energy of the wheelchair-user combination in its environment (Figure 0.1). The external power output (i.e. the task demands) during steady-state ambulation can be expressed as the sum of all energetic losses [15]. The wheelchair (e.g. mass, tyre pressure) and user (e.g. fitness, propulsion technique) need to be in prime condition to prevent overuse injuries and maintain mobility [17]. In addition to this, the wheelchair-user interface (e.g. seat height, rim and wheel diameter) needs to be ergonomically optimized for every individual given their specific physical, environmental, and task-specific constraints [18,19]. Still, today, handrim wheelchair propulsion is characterized as a straining and inefficient mode of propulsion that more often than not results in complications due to overuse of the upper extremities [20,21]. As such, research to better understand and prevent this strain is ever important and ongoing. Yet, given its rich history, the translation of existing knowledge on and methods for the assessment of wheelchair propulsion into clinical and sports practice is perhaps even more important. Central to this notion is the adoption of 'Wheelchair Propulsion Laboratories' in rehabilitation and adapted sports environments.

Wheelchair propulsion laboratory

Following the development of clinical exercise physiology laboratories ('labs') in cardiology and sports medicine during the last decades, clinical gait labs are an increasingly common phenomenon in hospitals and rehabilitation centres around the globe. The goal of these labs is to investigate and optimize the walking pattern and capacity of patients [17]. For gait, there is a large body of literature that provides evidence for the treatment efficacy of clinical gait analysis [22]. Yet, for wheelchair propulsion, similar concepts have not yet been embedded in clinical practice and are

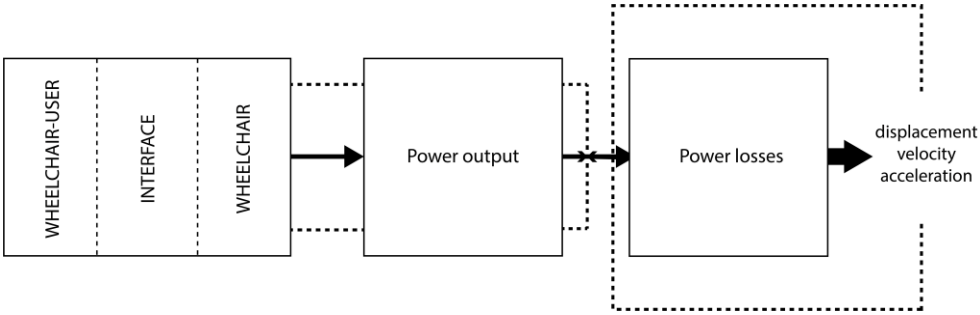

Figure 0.1. Power generation and energy transfer in handrim wheelchair propulsion, adapted from [15], full version is shown in Chapter 4.

still mostly limited to research institutes. The combined physiological, biomechanical, ergonomic, and technical approach used in the wheelchair propulsion lab could create the comprehensive biophysical fingerprint needed to optimize the wheelchair, user, and wheelchair-user interface [23,24]. While considerable effort already went into the development of such labs by researchers in The Netherlands (using skills and treadmill-based testing) [17,25] and the United States of America (using overground testing) [26,27], their implementation in clinical and sports practice has been limited. As such, the systematic biomechanical analyses of the wheelchair, the user, and the wheelchair-user interface using evidence-based approaches and experimental strategies, akin to those used in gait analysis, are not yet commonly available in clinical and adapted sports settings.

The state of the art

Wheelchair propulsion biomechanics has always been a rather arcane art belonging to a relatively small field of experts. In order for wheelchair propulsion labs to become established, human capital in the form of adept movement scientists and/or clinical exercise physiologists is needed, but the necessary equipment also needs to be present. These issues were also observed to be the greatest barriers to the systematic monitoring of wheelchair fitting and motor learning in rehabilitation in the Wheelchair Expert Evaluation Laboratory – implementation (WHEEL-i) project [17]. Specifically, the interpretation of test outcomes, and the lack of individualized recommendations and reference data were mentioned as limiting factors. The latter two can only exist after widespread adoption by building a shared repository that eventually could provide normative data, which causes a causality dilemma. Performing and interpreting tests could, however, be facilitated through hardware, software, as well as end-user training. Undoubtedly, the specificity of the measurement tools has also harmed the valorisation of past research findings. In

terms of general usability, Goosey-Tolfrey and Leicht [28] recently stated the following:

> *[…], it must be noted that the limited availability of specialized laboratory equipment for testing wheelchair-dependent participants, the technical expertise for conducting the tests and interpretation of test results are a drawback of laboratory testing.*

The need for a solution for the lab-based assessment of wheelchair users that is widely available and can be used by multiple institutes is discernible from this statement. Besides, it highlights the importance of a system that can be used by trained rehabilitation and healthcare professionals and adapted sports coaches as well as (embedded) scientists [17], which could provide a more evidence-based and (cost-)effective research, treatment, and training environment on the long term.

Stationary lab-based testing

The most ecologically valid ('realistic') testing environment is simple overground propulsion in the field [28], as all elements of the power balance and contextual factors are present. Recent technological advancements in the form of Inertial Measurement Units (IMUs) have made gathering kinematic data in the field increasingly more accessible [29], but obtaining detailed physiological and kinetic data still requires complicated and expensive systems. By extension, one could perform overground tests in the lab, as is common in clinical gait labs. However, this is also the least standardized condition because it is difficult to control the experimental conditions and procedures such as speed and power output. Moreover, since the wheelchair-user combination is moving through space, it requires relatively heavy ambulant measurement systems. Resultingly, stationary lab-based testing is often chosen for pragmatic reasons to increase data reliability and make the interpretation of results easier [5]. In the lab, one must choose between wheelchair treadmills and ergometers. Wheelchair treadmills are similar to regular treadmills, but are usually wider and have additional safety features. The term 'wheelchair ergometer' is used to describe a range of distinct devices such as roller systems and simulators, with a common goal: to apply a load and measure the effort of the wheelchair user. Wheelchair treadmills are second best compared to field-based testing in terms of ecological validity [5,30], but they do have some notable drawbacks when compared to ergometers (see below).

The case for ergometer-based testing

The current thesis is focused on ergometer-based testing which provides a number of advantages over other testing modalities (overground and treadmill). Similar to the treadmill, an ergometer provides a stationary platform for wheelchair propulsion testing. It is usually not possible to perform acceleration or sprint testing on a treadmill, whereas this is a feature on most wheelchair ergometers. The importance of these (anaerobic) tests is especially apparent when one considers that most activities of in the daily life of wheelchair users involve short intensive bouts of physical activity like ascending a ramp or crossing the street [31,32]. Additionally, doing any sort of turning or asymmetric propulsion on a treadmill would be problematic. Instead, wheelchair ergometers can facilitate virtual turning movements [33] when they have independent wheels or rollers (Figure 0.2), providing a relatively boundless (virtual) environment. Furthermore, ergometers can provide detailed kinetic information without the need for additional equipment such as measurement wheels, thereby preserving the individual wheelchair-user interface and offering a single solution for simulating and measuring wheelchair propulsion, reducing the amount of equipment needed. Finally, these data could be used as input for a screen-based virtual world, creating an immersive experience during training or testing, which provides the ability to add complexity such as curb-climbing, obstacle negotiation or avoidance, dual-tasking or biofeedback [13,34]. Ergometers present an ideal platform for individualised training, optimization, and testing as data gathered by the ergometer can be used to optimize current and future sessions using a data driven approach.

A novel wheelchair ergometer

In turn, the landscape of wheelchair ergometry is sparse and diverse with various different solutions for many specific situations. Wheelchair ergometers are almost always custom-made and one-of-a-kind devices, most often not commercially available. Unlike the treadmill, considerable development effort needs to be put in an ergometer to simulate wheelchair propulsion [35,36] and not all devices attempt to do this in the same way. Previously mentioned concerns regarding the standardised and realistic measurement of wheelchair propulsion have led to the development of a novel wheelchair roller ergometer. The ergometer originated from the combined expertise from over thirty years of wheelchair propulsion research [5,37] and ergometry [38] in The Netherlands together with an industry partner with almost seventy years of experience in ergometry [39]. The aim was to create an ergometer that provides a realistic simulation of overground wheelchair propulsion, measures propulsion technique, and maintains the wheelchair-user interface. Moreover, the ergometer had to be widely available, and the control and analyses



Figure 0.2. A render of the first iteration of the Esseda wheelchair ergometer with its dual rollers with a wheelchair on top. Hidden inside are the two servomotors and force sensors that power the ergometer.

had to be presented in an easy-to-use manner to facilitate use in research, clinical, and sports environments. The dual roller ergometer (Esseda) was developed by Lode BV, Groningen, The Netherlands in collaboration with the University Medical Centre Groningen (Figure 0.2). The ergometer was equipped with two servomotors that can reduce or increase resistance for the participant to simulate different surfaces, inertia, and task elements. As it is an active system, it could perhaps even be used to simulate and optimize power-assisted wheelchair configuration [40] using "human-in-the loop" optimization [41]. Effective applied force is measured with a load cell on each side without the need for additional equipment. The ergometer could be used in a wheelchair propulsion laboratory for wheelchair fitting, training, and testing of propulsion technique and exercise capacity.

Testing the ergometer

With the ergometer as a potential central component in the wheelchair propulsion laboratory, its overall functioning and performance to a large extent determine the efficacy of the lab as a whole. The ergometer should not only adequately measure wheelchair propulsion, but must also simulate wheelchair propulsion to a satisfying degree. In this sense, the ergometer as a virtual surface, adds another 'interface' to the wheelchair-user combination (Figure 0.3). It connects the user to the 'real' world using a model of what wheelchair propulsion looks like. In this specific case, the ergometer is modelled as an admittance-controlled feedback system. That is to say, for a certain force input the ergometer adjust the roller velocity to one that corresponds with the user-input and its conception of how a wheelchair should

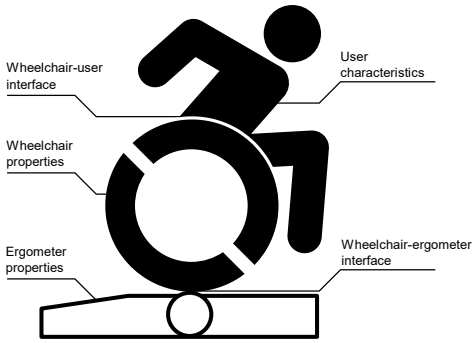

Figure 0.3. Wheelchair propulsion on a wheelchair ergometer adds a second interface: the wheelchair-ergometer interface.

move based on a mechanical model of real-life wheelchair propulsion (i.e. the power balance). Therefore, the first step in the validation process is to ensure that forces are measured with sufficient accuracy. Thereafter, the quality of the simulation can be assessed. Ideally, outcomes derived from speed and effective force measured by the ergometer should be close or equal to those obtained from measurement wheels, which have been shown to have a high accuracy [34,42] and are currently often used to gather kinetic data. Moreover, results should resemble those obtained during overground and treadmill propulsion, as they are more ecologically valid by default.

Aim and outline of the thesis

The aim of the current thesis is to evaluate a newly developed wheelchair ergometer on its ability to measure and simulate wheelchair propulsion. First, the thesis will provide an overview of the currently available state of the art in research literature and propose a conceptual model of real-life wheelchair propulsion (**Chapter 1 – A review of wheelchair ergometers**). Second, it will introduce a novel wheelchair ergometer, provide a thorough description of its design, and compare it to existing solutions for measuring wheelchair propulsion (**Chapter 2 – A new wheelchair ergometer**). A new accessible calibration method for wheelchair roller ergometers is proposed (**Chapter 3 – A low-cost portable calibration method)**. Fourth, a measurement protocol to perform overground, treadmill, and ergometer tests is presented in a combined video and text format, together with the software required to perform these measurements (**Chapter 4 – A guide to lab-based testing**). Fifth, measurements performed on the ergometer are compared to those in overground and treadmill testing to juxtapose the different environments (**Chapter 5 – A comparison of wheelchair modalities**). Sixth, an application of the ergometer in sports practice is presented in a longitudinal study on motor learning (**Chapter 6 – Measuring wheelchair racing propulsion**). Seventh**,** a similar study on power-assisted wheelchair propulsion was included (**Chapter 7 – Measuring power-**

assisted propulsion), which demonstrates potential avenues for the application of wheelchair ergometers in wheelchair propulsion labs. Finally, a general discussion of the results and future perspectives on the implementation of ergometers in wheelchair propulsion laboratories are discussed (**Chapter 8 – General discussion**).

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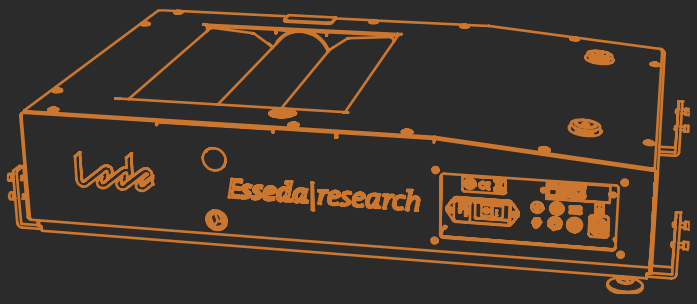
A review of wheelchair ergometers



Chapter 1

Measuring handrim wheelchair propulsion in the lab: a critical analysis of stationary ergometers

R. de Klerk, R.J.K. Vegter, V.L. Goosey-Tolfrey, B.S. Mason, J.P. Lenton, H.E.J. Veeger, and L.H.V. van der Woude - IEEE Reviews in Biomedical Engineering

1

Abstract

There are many ways to simulate handrim wheelchair propulsion in the laboratory. Ideally, these would be able to — at least mechanically — simulate field conditions. This narrative review provides an overview of the lab-based equipment used in published research and critically assesses their ability to simulate and measure wheelchair propulsion performance. A close connection to the field can only be achieved if the instrument can adequately simulate frictional losses and inertia of real-life handrim wheelchair propulsion, while maintaining the ergonomic properties of the wheelchair-user interface. Lab-based testing is either performed on a treadmill or a wheelchair ergometer (WCE). For this study WCEs were divided into three categories: roller, flywheel, and integrated ergometers. In general, treadmills are mechanically realistic, but cannot simulate air drag and acceleration tasks cannot be performed; roller ergometers allow the use of the personal wheelchair, but calibration can be troublesome; flywheel ergometers can be built with commercially-available parts, but inertia is fixed and the personal wheelchair cannot be used; integrated ergometers do not employ the personal wheelchair, but are suited for the implementation of different simulation models and detailed measurements. Lab-based equipment is heterogeneous and there appears to be little consensus on how to simulate field conditions.

Introduction

To improve wheelchair design and the quality of life of handrim wheelchair users in both a daily and sports setting, reliable and valid measures of wheelchair propulsion are necessary [1]. These measurements can either be made in the field (e.g. in everyday propulsion or on the sports court) or in the laboratory on a treadmill or Wheelchair Ergometer (WCE). The equipment used by researchers to measure handrim wheelchair specific performance in the laboratory is diverse. This diversity in itself has implications for the generalizability of results and the applicability of the existing knowledge base [2], yet no critical overview currently exists.

Field-based testing present researchers with the least standardized but most externally valid conditions in which to study wheelchair propulsion [3,4]. It allows for the subject to be tested in their natural environment and personal wheelchair [5]. The latter is especially important as wheelchair settings greatly influence performance and modern wheelchair technology has become increasingly more individualized [6]. It is, however, problematic to collect physiological, kinetic, or kinematic data without changing the wheelchair in terms of mass and configuration, and it is further complicated by the non-stationary position of the wheelchair-user combination with respect to the environment. Additionally, in field testing, experimental conditions, friction or power output are difficult to control, reducing the reliability of any such measures [2].

Hence, wheeled mobility research today is still predominantly conducted inside the laboratory. Lab-based research allows detailed physiology and biomechanics studies to be conducted under controlled conditions [7], while the wheelchair-user combination is stationary on a treadmill or WCE. However, lab-based equipment is often customized and may vary in reliability and validity [2]; in fact, no commercial line-up of wheelchair ergometry, as for instance in the bicycling domain, is available for manual wheelchair testing. Moreover, while lab-based wheelchair testing protocols have been described in the literature [8,9], information on diversity and reliability details of the equipment is sparse. Certain choices during the design process might offer advantages on realism, ease of use, specific measurement capabilities, or cost. Yet, if the limitations of the equipment are not understood, interpretation errors can be made.

The choice of equipment is important as a lab-based modality should not only allow for accurate measurements but must also simulate wheelchair driving as realistically as possible in relation to the research at hand [7]. There are three main factors in wheelchair propulsion that decide the eventual behaviour: the wheelchair, the user,

and the interaction thereof [10]. All three should be considered when assessing the validity of a lab-based testing instrument. A lab-based modality should thus not only be mechanically realistic, but also ergonomically (e.g. seat height or camber), and ecologically (e.g. visual or proprioceptive feedback). In general, to perform a valid simulation of overground manual wheelchair propulsion, the ergometer set-up used to evaluate wheelchair propulsion should thus ideally be able to:

- Simulate frictional losses, environmental conditions, and translational inertia of the wheelchair-user system;
- Facilitate valid and reliable measurements of power output on the wheels;
- Respect the ergonomic properties of the wheelchair-user interface and provide adequate sensory feedback to the user;
- Facilitate different testing protocols (i.e. submaximal, anaerobic and aerobic exercise testing and training)

The aim of this narrative review is therefore twofold. First, to create an overview of the available lab-based equipment in the research literature. Second, to assess the equipment on their ability to simulate and measure wheelchair propulsion in the laboratory based on the four indicators mentioned above. The current review starts off by providing a simple mechanical model of wheelchair propulsion as a conceptual framework for mechanically realistic wheelchair propulsion and simulation. Then the simulation and measurement capabilities of the available equipment and how researchers have approached this in international literature are discussed. Subsequently, ergonomics and sensory feedback on the equipment is examined and finally the testing capabilities are considered. The information from the current review is useful when comparing results of different studies, the standardization thereof, and could aid in the design of new (calibration procedures of) lab-based equipment.

Search strategy

For this narrative review, an overview of the existing literature was made by performing a semi-structured search using the PubMed, CINAHL, and Web of Science internet databases with the query "wheelchair AND (ergomet* OR dynamomet*)" on 2017-05-22 (n=842 results, 333 duplicates) and the consequent snowball method (11 additional papers). Thereafter, articles were first screened by one author on title, then on abstract, and then on full content (if still available/accessible). Articles were screened in chronological order. If two similar devices were found from the same research group or if a newer article referenced a previous article, they were assumed to use the same device. Like any literature study this study relies on the previously published literature and the availability thereof.

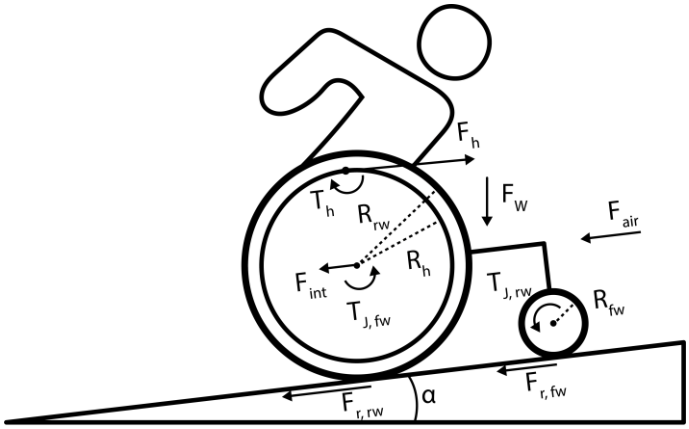


Figure 1.1. Diagram of a wheelchair rolling over a flat surface with an incline; user force and frictional losses are illustrated as force vectors in handrim wheelchair propulsion.

Due to the nature of this review, it often depended on relatively old literature which could not always be accessed.

Overground vehicle mechanics

To simulate the mechanics of handrim wheelchair propulsion in the field, a model [11–13] of wheelchair propulsion with the governing equations of motion is required (Figure 1.1). In this paper a reductionistic translational model for wheelchair propulsion is proposed which can be considered as a minimum requirement for the study of manual wheelchair propulsion and is limited to straight forward motion. The wheelchair can be modelled as a linear system, where the acceleration of the centre of mass of the wheelchair-user combination is equal to the total acting force divided by the mass:

$$F_{sum} = m_{tot} * a \tag{1}$$

Where m_{tot} is the combined mass of the user (m_{user}) and wheelchair (m_{wc}), and the combined moments of inertia (J) of the wheels:

$$m_{tot} = m_{user} + m_{wc} + \sum_{i=1}^{n_wheels} \frac{J_i}{r_i^2} \tag{2}$$

Where F_{sum} includes the user generated forces (F_s), rolling resistance (F_{roll}), air drag (F_{air}), gravitational forces on an incline (F_α), and internal friction F_{int} experienced during wheelchair propulsion [7]. The forces acting on the wheelchair can then be expressed by:

$$F_{sum} = F_s - F_{roll} - F_{air} - F_\alpha - F_{int} \tag{3}$$

The driving force in wheelchair propulsion are the forces and torques generated by the user, where the effect (F_s) of the tangential force on the handrim (F_h) and the local torque at the hand (T_h) can be calculated with (4):

$$F_s = \frac{F_h * R_h + T_h}{R_{rw}} \tag{4}$$

During everyday propulsion (low to medium speeds), rolling resistance is the largest resistive force [2]. It is determined by the mass (m_{user} + m_{wc}), the distribution thereof on the front and rear wheels (N_{fw} & N_{rw}), the radius of the wheels (R_{rw} and R_{fw}), and the characteristics of the wheels and floor surface (μ_{rw} & μ_{fw}). It can be expressed by the following equation:

$$F_{roll} = \left(\frac{\mu_{rw}}{N_{rw} * R_{rw}} + \frac{\mu_{fw}}{N_{fw} * R_{fw}}\right) * \cos(\alpha) \tag{5}$$

Where α is the angle of inclination and N_{fw} and N_{rw} dynamically change during propulsion and are dependent on the length of the wheelbase (L_{wb}) and the distance of the center of mass from the rear wheels (R_{cg}):

$$N_{fw} = \frac{R_{cg} * m * g}{L_{wb}} \tag{6}$$

$$N_{rw} = m * g - N_{fw} \tag{7}$$

At high speeds the air drag becomes an important source of friction [14]. It is a velocity dependent friction (v^2) that is influenced by the velocity of the wheelchair (V_{wc}) and wind (V_w), the frontal plane area of the wheelchair user combination (A), the air density (D), and the aerodynamic drag coefficient (C_d) [15]. The frontal plane area is dependent on the posture of the wheelchair user. Moreover, the drag coefficient can also be influenced by the characteristics of the wheelchair user combination:

$$F_{air} = 0.5 * D(V_{wc} - V_w)^2 A C_d \tag{8}$$

When the wheelchair is going up or down a slope (α) there will be a force acting on the system as a result of gravity:

$$F_\alpha = m * g * \sin(\alpha) \tag{9}$$

The internal friction as a result of the bending of and localized deflections in the bearing rings is defined as a function of the velocity in Cooper's model [13]. The internal friction is therefore equal to the constant K multiplied by the velocity:

$$F_{int} = K * v \quad (10)$$

The contribution hereof is not entirely clear. The hubs typically have annular sealed bearings and the friction coefficient will not exceed 0.001 if the bearings are properly maintained and lubricated [16,17].

Simulations & Measurements: Treadmills

Simulation

Extra-wide treadmills have been used in wheelchair research as early as 1969 [18,19] and allow wheelchair propulsion at various speeds and/or slopes accommodating both every day and sports wheelchairs. They provide a realistic, safe, and stationary environment to measure wheelchair propulsion during a range of constant velocities and loads. Propulsion on a treadmill provides a mechanically accurate simulation of straight-line regular wheelchair wheeling [20,21]. Small steering corrections are necessary, while rolling friction and inertia are realistic due to Galilean invariance [20]. Moreover, the contribution of trunk movement to the wheelchair dynamics is also realistic. However, air drag is not simulated in treadmill propulsion (which only becomes an issue at high speeds [15,22]), turning is not possible, there is different/limited feedback on speed, and due to practical and safety concerns acceleration tasks cannot easily be performed. Many treadmills are fitted with safety systems like sliders or rubber bands [23] which could influence the wheelchair user, but limited information is available on their effect [24]. However, such systems could limit the steering requirements of treadmill propulsion and reduce the required power output, which would hurt the validity.

Few studies have compared overground wheelchair propulsion with propulsion on treadmills. A recent study demonstrated that, similar to gait, self-selected speed on a treadmill is lower in experienced wheelchair users [25]. The authors attributed this to differences in feedback and the higher cadence needed for treadmill propulsion as participants feel a sense of urgency to control the wheelchair. At matched speed conditions and similar power output they still found that spatiotemporal variables were different from overground propulsion [25]. Stephens and Ensberg [26] showed that hand trajectories for overground and treadmill propulsion were significantly different. However, in later studies Kwarciak et al. [21] and Mason et al. [27] found correlations for physiological and biomechanical parameters in treadmill propulsion and over ground propulsion at specific treadmill settings.

Measurements

Mean power output (11) during steady-state wheelchair propulsion can be relatively accurately determined by performing a drag test [10,19]. The treadmill allows for

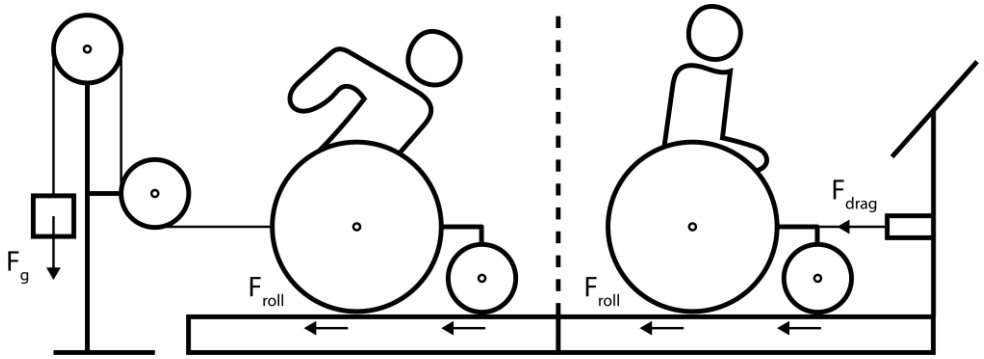


Figure 1.2. On the right-hand side the wheelchair drag test to determine the workload is shown. On the left-hand side a pulley system to increase the workload is shown.

power output to be varied through belt inclination or the application of resistance to the back of the wheelchair via a pulley system (12, Figure 1.2). The importance of determining power output before testing is highlighted by the findings of De Groot et al. [28]. Treadmill speed was often inaccurate and that power output could differ even among identical treadmill models. The source of the difference in power output between institutes in their study was not due to calibration, the wheelchair occupant, or experimenter errors but rather related to small manufacture-based differences in treadmill characteristics. As in any measurement device, regular calibration (of speed and inclination) is crucial in using a treadmill [28]. In a study of Vegter and colleagues [29] the drag test slightly underestimated the required power output. This could be because during the drag test the user assumes a constant immobile upright position and a perfectly straight heading, while in reality the rolling resistance fluctuates and small steering corrections are necessary.

$$P_{ext} = F_{drag} * v_{belt} \quad (11)$$

$$P_{ext} = (F_{roll+int} + m_{pulley} * g) * v_{belt} \quad (12)$$

Detailed kinetics

More detailed kinetic information can be obtained with the use of measurement wheels (e.g. [30–37]) of which two systems have been commercially available [37,38], but today are no longer available on the market. Most measurement wheels acquire 3D forces and torques around the handrim (though some 2D systems also exist [33]). This is valuable data which, when combined with 3D kinematics, can be used for inverse dynamics [39]. Moreover, information from these systems can be used to assess wheelchair propulsion technique by calculating spatio-temporal variables such as contact angle or push time, and kinetic variables such as peak torques or

fraction of effective force. Measurement wheels can be used on treadmills, but can also be used overground and on WCEs and have even been used to control WCEs [40]. Alternatively, ground reaction forces from an instrumented dual-belt treadmill can be used to estimate kinetic measures [41]. This will not result in 3D kinetics, but it does give more detailed kinetic and temporal information than a simple drag test alone.

Simulation & Measurements: Wheelchair ergometers (WCEs)

WCEs provide the most constrained wheelchair testing environment as the wheelchair is fixed and no steering is required to keep the wheelchair on the WCE. They offer some noticeable advantages over treadmills as power output can be easily adjusted, simulated turning is often possible, and acceleration tasks (e.g. a Wingate) can be safely performed. The importance of acceleration tasks is apparent considering that most motor activities of daily living that are practiced in the everyday life of wheelchair users are usually of short duration and of relatively high intensity [42,43], thus taxing the anaerobic energy system.

However, in contrast to treadmills, WCEs are mechanically heterogeneous and there are various different approaches to designing WCEs. The first WCE found was the device of Brouha and Krobath in 1967 [44], with 50 unique WCEs found in the literature in 2017. They can roughly be grouped into three categories: roller ergometers, flywheel ergometers, and integrated ergometers (Figure 1.3). Some hybrid designs (roller attached to flywheel) were also found. An overview of the ergometers found in the literature and their specifications is presented in Table I. In general, WCEs use a simplified model of wheelchair propulsion close to the one in this paper. It should be noted that (most of) these ergometers make some additional assumptions about wheelchair propulsion which could hurt the relation with the field [45]:

- The wheelchair is propelled on a strictly straight line;
- The movements of the subject on the wheelchair do not contribute to the dynamics;
- The rolling resistance force is constant;
- The wheels do not slip on the floor;
- The castors do not contribute to the dynamics of the wheelchair.

Roller ergometers

The majority of WCEs found in literature could be categorized as roller ergometers (Table 1.1). To be considered a roller ergometer, the WCE had to have at least one roller on which a wheelchair could be fixated. Similar to the TM, the advantage of roller ergometer is that they can be used with the personal wheelchair of the user.

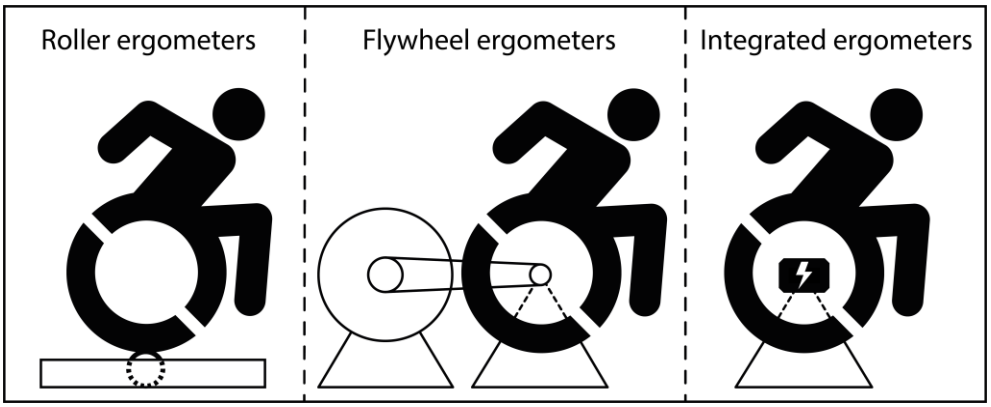


Figure 1.3. Schematic drawing of the three types of wheelchair ergometers found in literature. Roller ergometers use the personal wheelchair on a roller, flywheel ergometers use an integrated wheelchair coupled with an ergometer, integrated ergometers use an integrated wheelchair with a braking/accelerator system.

This typically allows for fast testing as no provisions have to be made to match the ergonomics of the WCE with the regular wheelchair of the user. Roller ergometers range from fully passive rollers to highly advanced computer-controlled systems with electronic brakes or motors for the individual rear wheels.

In their most basic form, roller ergometers consist of one or more rollers that have a moment of inertia that is (or should be) similar to the translational inertia of a wheelchair-user system and which provides passive friction. The most straightforward and common method of simulating wheelchair propulsion on a WCE is to use a static friction and an inertia that is matched to the participant or a 50th percentile equivalent mass. The inertial properties of the roller can be calculated, obtained from CAD-models, or determined with an acceleration or trifilar pendulum test [46].

If the inertia is too low the wheel speed at the start of the push cycle will be low and it is easy to accelerate the wheel. On the other hand, if the inertia is too high it will resist changes to speed more and it will be very difficult to accelerate or decelerate the wheelchair. Two different approaches to simulate the translational inertia of wheelchair propulsion on a roller ergometer have been found:

- Mechanical: choosing a roller with a rotational inertia that matches with the translational inertia of the wheelchair and the subject combined; attach the roller to a flywheel; use weighted disks to adjust the rotational inertia of the roller; change the moment of inertia by adjusting the inner diameter of the roller, thereby changing the inertia experienced by the user [47].

- Electronic: using an electronically controlled motor or brake.

The change in velocity of the wheels (α_{rw}) is dictated by the torque applied by the user (T_s), inertia, the frictional torque in the bearings of the ergometer and wheelchair (T_{int}), and the contact friction (T_{fr}) of the wheel on the roller (13). These frictional characteristics are influenced by the weight pressing down on the wheelchair and roller.

$$\alpha_{rw} = \frac{T_s - T_{int} - T_{fr}}{J_{roller} * \left(\frac{r_{rw}}{r_{roller}}\right)^2 + J_{rw}} \quad (13)$$

The power output required for propulsion on these ergometers can be determined using the Theisen et al. method [48] when the inertia is known. The total internal torque ($T_{tot,int}$), which consists of T_{int} and T_{fr}, can be determined by performing a coast-down test. The deceleration test provides data of time and velocity. The calculation of a linear regression line on values that lie within the range of the velocities performed represents the linear acceleration from which the angular acceleration of the roller can be derived. The external torque delivered to the wheels during this period is zero therefore; a reflection of the total resistive forces of the specific wheelchair ergometer can be calculated.

$$T_{tot,int} = \frac{J_{roller} * \left(\frac{r_{rw}}{r_{roller}}\right)^2 + J_{rw}}{\alpha_{rw}} \quad (14)$$

Assuming a constant frictional torque, the work output at a constant velocity can then be determined by multiplying the total internal torque with the distance travelled.

$$W_s = T_{tot,int} * s \quad (15)$$

The first problem encountered with roller ergometers is that the coast-down test assumes a constant total internal torque. However, during actual overground wheelchair propulsion the weight on the rear wheels shifts. At the end of the push phase the weight is shifted to the castor wheels of the wheelchair, which would increase the rolling friction in regular overground propulsion [49,50]. However, on roller ergometers the opposite happens, the total internal torque is reduced when the weight is shifter to the castor wheels. No information on the impact of this discrepancy between overground propulsion and WCE propulsion is available at this time.

Another problem encountered with roller systems is that the base resistance of the system can be too high for some individuals as rolling friction on a roller system is considerably higher than overground. If this friction is too high for the participant

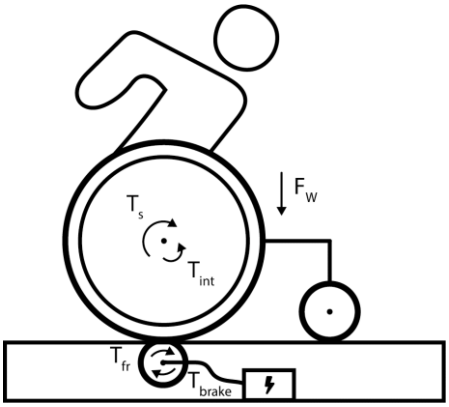


Figure 1.4. Graphic representation of a wheelchair roller ergometer with an active system to generate resistance. Translational inertia is simulated with the roller mass [13].

there is little that can be done as friction can usually only be increased (e.g. with a brake). Aissaoui et al. [51] solved this by reducing the weight on the rollers, thereby also increasing the risk of slipping. High base resistance becomes even more problematic considering the effect of camber on roller ergometers [52], which could further increase the base friction of the WCE. This effect is likely bigger than during overground propulsion, especially on two-roller systems as camber might lead to a misalignment on one of the rollers [53]. Consequently, the ergometer presented by Faupin et al. [52] can be adjusted to rear wheel camber to reduce camber induced friction. Use of low-friction bearings, adaptable camber [52], using a different tie-down system, or reducing the weight on the rollers [51] could reduce internal friction. Similar to the friction on treadmills, the friction in roller ergometers is not constant, even within the same model [54]. Between models there can be even larger differences as the wheelchair setup or method of fixation differs.

The frictional torque on the roller can also be increased with an external mechanism such as a brake or a motor. Equation 13 can be adjusted to account for this friction (16). The Theisen et al. method [48] can then be used to determine the braking torque of the braking mechanism. With multiple tests the characteristics of the brake can be identified.

$$\alpha_{rw} = \frac{T_s - T_{int} - T_{fr} - T_{brake}}{J_{roller} * \left(\frac{r_{rw}}{r_{roller}}\right)^2 + J_{rw}} \quad (16)$$

However, even if adjustments are made for assumed constant forces like rolling resistance and slope, a variable external torque is still required to simulate air resistance [13]. This requires the use of an electronically controlled braking system. Another advantage of such a system is the increased flexibility for simulations or adjusting workload during a test. For example, it can be used to simulate transitions

between surfaces [55]. An example of a WCE with an advanced braking system are the ergometers described by Theisen et al. [48], Wu et al. [55] or the VP100H [56]. Roller ergometers with an electronic control system offer more flexibility to mechanically emulate wheelchair propulsion, but they still have to adjust for system friction.

Finally, some roller ergometers use a motor to simulate overground propulsion. The advantage of using a motor is that it cannot only generate braking torques, but it can also generate assistive torques. It allows for, but also requires, more advanced calibration methods [57]. The ergometer can be modelled as a haptic feedback system [40], in which it uses the torque applied on the rollers and an internal model of a wheelchair to simulate propulsion. Moreover, slopes can be more accurately simulated as the rollers can roll back or accelerate on their own, though this was only found in the ergometer described by Brauer [58]. Examples of motorized ergometers are the Ergotronic 9000 [59], and the ergometers presented by Devillard et al. [56], Harrison et al. [60], and Klaesner et al. [61].

The only roller ergometer that does not have to adjust for system friction is the ergometer presented by Chenier et al. [40]. It uses admittance-control and uses force obtained from a measurement wheel (SMARTwheel, Three Rivers Holding, USA) as input. This is fed into a simulation model and a motor with a controller is used to set the roller speed. This method avoids the calibration problem of other roller ergometers, but it sacrifices the flexibility of being able to use the personal wheelchair (or at least the wheels) of the user. Though, for all motorized ergometers the aforementioned additional assumptions for roller ergometers still apply.

Flywheel ergometers

Ten different flywheel ergometers were found in the older (1960-1990) literature. In flywheel ergometers a chair or wheelchair is mounted on a frame. The wheels of the wheelchair are coupled to a flywheel assembly through a chain and sprocket system. The main advantage of this approach is that a commercially available bicycle ergometer can be used. Moreover, frictional torques as a result of the rear wheel pressing down on a roller is not a problem. The flywheel ergometer design was first implemented by Brattgard in 1970 and was later adopted by researchers at various other universities (Table 1.1).

Flywheel ergometers are dependent on the properties of the chosen bicycle ergometer. Friction is simulated by a standard friction belt or other braking system. This setup can only simulate a constant friction which is sufficient for simulating rolling resistance, but usually not for velocity-dependent friction such as air drag. Additionally, the inertia of the flywheel cannot be easily adjusted for participants of

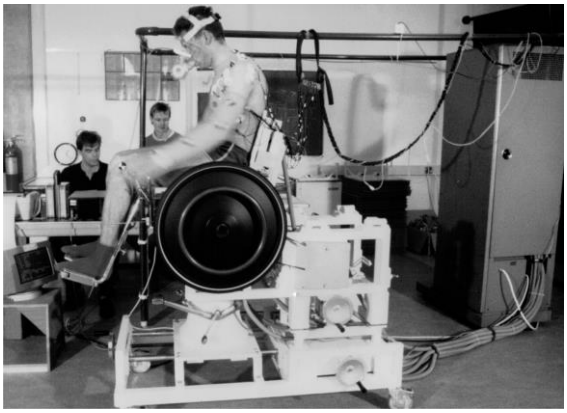


Figure 1.5. Example of an integrated ergometer at the VU Amsterdam. The wheelchair is integrated with the stationary ergometer. This ergometer uses a haptic feedback model based on user torque and simulated frictional forces [7].

different body mass without making adjustments to the original bicycle ergometer. As a result, flywheel ergometers can generally only use basic simulation models of wheelchair propulsion. The acceleration of the rear wheels is dependent on the braking torque and the inertia of the flywheel ($J_{flywheel}$) (gear ratio of 1:1):

$$\alpha_{rw} = \frac{T_s - T_{int} - T_{brake}}{J_{flywheel} * \left(\frac{r_{rw}}{r_{flywheel}}\right)^2 + J_{rw}} \quad (17)$$

Some researchers have chosen for an ergometer that has a work rate which is independent of the turning frequency. The flywheel ergometer at Lavel University [62] uses a bicycle ergometer with these properties. Work rate can be very tightly controlled with such an ergometer, but it could be less realistic than the other approaches.

Integrated ergometers (simulators)

In integrated ergometers or simulators, the simulation and measurement capabilities of an ergometer are integrated in a wheelchair-like device. The first integrated ergometer found in the literature was described by Dreisinger in 1978 and was patented by Cardrei Corporation [63]. The advantage of this approach, over roller ergometers, is that the system friction and inertia are almost negligible. More importantly, they do not change between users or when a user shifts their weight.

One decade later, Niesing et al. [7] presented a more advanced integrated ergometer. An electronic control system simulates frictional losses on the basis of feedback with software based on the power balance [10]. A force measuring system allows for the measurement of forces applied to the handrim for each individual wheel. The design

of this ergometer ensures provision of an accurate simulation of frictional losses and the ability to simulate slopes. In addition to this, the translational inertia of the wheelchair-user system is simulated. The force transducers in both the seat and the wheels allow for biomechanical analysis of wheelchair configuration and wheelchair propulsion. Finally, it allowed for wheelchair adjustments to be tested.

The final integrated ergometer that was found was presented by Samuelsson et al. [64] and used an isokinetic dynamometer on a wheelchair attached to a frame. The use of an isokinetic dynamometer can provide insight in the torque-velocity relation of the propulsion movement during wheelchair propulsion. This method, also facilitated by the previous ergometer [7], provides data not disclosed by other methods, though it does not fit in the framework set in this paper.

Measurements on a wheelchair ergometer

As WCE designs differ, their measurement possibilities also differ. Some estimate power output with a coast-down test, while others can measure torque directly or indirectly. Measurement validity is closely tied to the validity of the mechanical strain the WCE imposes on the user, yet this is not often reported in the literature. Potentially because this data is considered to be internal data. Due to the variation in WCEs, a variety of methods are employed to validate the measurement validity, there is currently no 'gold standard'. In general, most authors provide limited information on the calibration and validation of their devices.

Flywheel ergometers are dependent on the validity of the bicycle ergometer they are based on. Wheelchair propulsion is of much lower intensity than cycling. While readily available, bicycle ergometers should be treated with care. There is variation between bicycle ergometers [65] and the validity of bicycle ergometers during incremental power tests can be lower than expected compared to steady-state exercise tests [66]. The determination of power output could be inaccurate [7], as was demonstrated in some "turbo trainers" [67]. Moreover, some bicycle ergometers do not include the power required to accelerate the flywheel in their calculations [66].

Langbein et al. [68] used a calibration rig to compare the ergometer output with a torque sensor. They fully characterized the eddy-current brake of their WCE with this device. Similar techniques have also been used in the calibration of bicycle ergometers [66]. A test with intermittent speeds, akin to those in wheelchair propulsion, was not performed. Alternatively, the VP100H ergometer was validated with the use of a two-dimensional force transducer platform [56]. Errors in force and power ranged from 0.89% to 7.56% and from 0.41% to 6.74%, respectively. The higher error rates were attributed to trunk movements during the tests.

In another study, Hutzler et al. [69] also used a force transducer to validate the force readings of their ergometer. They performed a static test against a simulated external load and reported errors of "below 5%". No other information about this test was presented. The WheelMill system, presented by Klaesner et al. [61], showed similarities with a measurement wheel, but measurement error was substantial

Alternatively, the emphasis can be put on the outcome parameters. The measurement capabilities of the device could be defined as the concurrent validity of the WCE with an established field protocol, which mainly depends on the validity of the simulation and testing protocols. Mason et al. [27] compared physiological outcomes between TM, overground, and WCE propulsion. They found significant differences in oxygen consumption and heart rate between modalities. However, they did not standardize the load between exercise modalities. Koontz et al. [70] compared handrim kinetics in a quantitative and qualitative analysis. They found significant correlations between overground and WCE propulsion, but large differences between modes were found. Again, inertia and friction between modalities was not standardized.

One comparison of peak aerobic performance in the field and on a WCE ergometer was found. Burkett et al. [71] found similar physiological response patterns and magnitudes in a field and WCE test. Van der Scheer et al. [72] found a weak relationship between 15m overground sprint outcomes and a Wingate test on a WCE. However, the purpose of this test was not necessarily to compare the two modalities, but to compare the two tests. If anything, these results suggest that the WCE testing environment is not yet able to closely emulate overground conditions.

Reliability of outcome parameters was extensively studied by researchers. It is not only dependent on the mechanical reliability of the WCE but is also influenced by the reliability of the testing environment, protocols, and biological variance. Bhambhani et al. [73] specifically looked into the reliability of a maximal graded exercise test on a WCE in wheelchair users with cerebral palsy [74] and spinal-cord injury [73]. They concluded that physiological responses during graded exercise tests on a WCE ergometer are highly reliable. Similar results were found by Burkett et al. [71] in spinal-cord injury patients on their ergometer. The ergometer of Theisen et al. [48] was tested with sedentary patients and sportsmen. They also concluded that results are reproducible on their ergometer. In another study, Keyser et al. [75] performed 30-minute bouts of constant work-rate wheelchair ergometry. They showed that, in able-bodied participants, the oxygen uptake and heart rate is highly reliable. In a similar test, Finley et al. [76] showed that most kinetic and kinematic variables obtained during wheelchair ergometry are reliable unless when fatigued.

Ergonomics & Sensory feedback

Ergonomics

Treadmills and roller ergometers allow the use of the personal wheelchair. However, in flywheel ergometers the chair is attached to the flywheel and cannot be exchanged with another chair. A similar problem is found in integrated ergometers. Yet, to compensate for this, it was found that integrated ergometers, like the WCE described by Burkett et al [71], and Niesing et al. [7], use highly adaptable seating, making integrated ergometers ideal for experimenting with ergonomic settings, but not for testing the original wheelchair-user configuration. For example, the apparatus described by Niesing et al. [7] consists of a frame with two independently mounted wheels and an adjustable seat and backrest that are mounted on a console with a hydraulic foot. Wheels, camber, handrim form and configuration could all be altered and were evaluated in the past decades [2].

Proprioceptive feedback

Proprioceptive feedback in a wheelchair consists of the force on the handrim, but also of the feeling of motion, and response of the seating and backrest. The validity of the first components is determined by the mechanical validity of the simulation. It is also influenced by the configuration of the wheelchair-user interface, which affects the mechanical advantage of the user [13]. Another addition that can be made is to allow the wheels to roll backward when on a virtual incline or accelerate when on a decline [58]. Finally, due to the absence of wind, there is also no sensation of air drag. This could be added with the addition of a fan, but has not yet been done in wheeled mobility studies before and the contribution might be negligible.

A second component, vestibular perception, is more difficult to simulate on a stationary platform. As the wheelchair is fastened or integrated in a WCE, there is no risk of tipping or toppling. Treadmills provide realistic sensory feedback to the user. In contrast to WCEs, the wheelchair is not fixated and the user needs to 'stabilize' the wheelchair. De Groot et al. [77] found a small difference in mechanical efficiency between the computer-controlled ergometer and overground propulsion. This difference is likely due to the fact that the trunk needs to be stabilized during overground wheelchair propulsion but not as much during wheelchair ergometry, which is a problem that arises with almost every form of ergometry. Indeed, Veeger et al. found a small difference in trunk movement during treadmill propulsion and wheelchair propulsion on the ergometer [78].

One example was found of a system that simulated up- and downslopes, and cross-slopes [61]. Moreover, the platform of the ergometer at Human Engineering Research Laboratories [79], the University of Melbourne [80], and University of

Louvain [48] can simulate slopes in the forward-backward direction. However, these systems are the exceptions to the norm as the majority does not include any form of vestibular feedback.

Visual feedback

Wheelchair propulsion in the field is accompanied by optical flow. The subject uses visual cues to steer the wheelchair in the right direction. On a treadmill the user receives some feedback on heading as they need to stay in the centre of the belt [81], but they are more or less stable in the environment. On a WCE the user needs some form of feedback to know where they are (in virtual space). Most ergometers provide limited feedback in the form of speed and/or direction. Other ergometers provide a moderate form of feedback on a screen. Very little is known about what feedback to use. For example, the ergometer presented by Wu et al. [55] provides visual feedback of the surface that is currently being simulated and when a transfer is made.

Finally, the most extensive feedback of the position of the wheelchair can be given by employing virtual reality (VR) on semi-immersive 180° screens or head mounted displays. This implementation was only found in one WCE for regular handrim wheelchairs [60]. A study [82] with powered wheelchairs has shown that self-chosen speed is lower when the environment is more immersive. The incorporation of haptics into VR simulations of the built environment provides a powerful tool that should allow wheelchair users to directly participate in the design and testing of accessible environments, and it is a motivational tool [83]. VR technology has rapidly improved over the past few years, but it is scarcely used in this line of research. Newer systems no longer need extensive setups with multiple screens, but could use commercially available head mounted displays.

The WCE described by Harrison et al. [60] is specifically used as a simulator. It is used to test how wheelchair users interact with the built environment. Hence, they implemented a more advanced system for visual feedback. This allows users to participate in the design and testing of accessible environments. Another example of a setup with visual feedback is the treadmill at Pittsburgh [79]. Finally, visual feedback can be used to enhance or induce motor learning [84], or as a motivational tool [85].

Auditory feedback

Auditory cues could be used in learning tasks or to increase immersion [83]. Additionally, they can be used to enforce a preferred cadence. No examples of auditory feedback were found in the literature regarding WCEs. Nevertheless, all WCEs and treadmills could be a suitable platform for studies that include a form of auditory feedback would it fit the needs of the researcher.

Testing capabilities

Biomechanics & motor learning

All lab-based modalities provide considerable advantages for researchers when compared to overground testing. Both the treadmill and ergometer environments are relatively easy to standardize. Moreover, they can be expanded with extra equipment that is necessary for biomechanics and motor learning studies. Indeed, motor learning during overground propulsion [115,116] and treadmill [81,84,117] or ergometer propulsion [77] show somewhat similar results. However, in overground and ergometer propulsion, it is possible for the self-selected speed to change during motor learning. It has also been argued that wheelchair propulsion on an ergometer is less complex and therefore might produce different learning outcomes, but De Groot et al. [77] found similar results for ergometer, treadmill, and overground modalities after a 3-week practice period. As discussed in the previous sections, there is little information available on the ecological validity of biomechanics testing on treadmills [21,25,26] and ergometers [27,70–72].

Aerobic

Exercise testing and training benefit from task specificity. Submaximal aerobic tests (or training protocols for that matter), often used in motor learning and biomechanics studies, can be performed on any wheelchair treadmill or WCE. Submaximal tests are either predictive or performance tests, where predictive tests are used to estimate peak aerobic capacity and performance tests involve measuring responses to typical physical activities or interventions [118].

Peak oxygen uptake and power output are used to indicate peak physical capacity [119] which can be used to evaluate the effect of training programmes. Historically, the majority of studies performed were using arm crank ergometers (ACEs) in favour of wheelchair ergometry [120]. ACEs were used as early as 1971 by Stoboy et al. [121] to measure exercise capacity in wheelchair users. They offer some noticeable benefits as they are a low-cost, porTable 7nd non-specific measuring tool for upper body work capacity [122]. They allow for the measurement of upper body fitness in isolation of context, which is useful for inter-group comparisons or for the assessment of individuals who do not use their wheelchair during activities of daily living, rehabilitation, or their sports activities [122]. Peak oxygen uptake is similar in ACE and wheelchair ergometry tests, but external power output is higher in the ACE condition due to a better (bio-)mechanical transmission of internal power and lower skill requirement [123].

Table 1.1. Characteristics of all wheelchair ergometers found: an overview of their measurement and simulation capabilities[†]

Article (year + reference)	Category	Ergonomics	Measurements	Inertia	Rolling resistance	Air drag	Slope[‡]	Left/ right	Feed-back	Validity/ reliability	Remarks
1967 [44], 2001 [86]	Roller ergometer	++ personal wheelchair	+/- distance and time	- roller mass not matched	+/- internal friction of WCE	- No	- No	- No	+/- V	[44][a]	
1970 [87], 1977 [88], 1979 [89], 1981 [90], 1982 [91], 1991 [92], 1995 [93], 1998 [94], 2000 [95]	Flywheel ergometer	- integrated wheelchair	+ power output, speed	+/- fixed mass flywheel	+ brake	- No	- No	- No	+/- V	[75,76,87,89,96][a], [97][b]	Commercially available bicycle ergometers
1972 [58]	Roller ergometer	++ personal wheelchair	+ power output, speed	+ variable mass flywheel	+ brake	+ motor	++ motor	- No	+/- V	[58][b]	
1978 [63]	Integrated ergometer	- integrated	++ torque, speed	+ variable mass flywheel	+ brake	- No	- No	+ Yes	+/- V		Commercial
1983 [80]	Roller ergometer	++ personal wheelchair	+ power output, speed	- roller mass not matched	+ brake	- No	- No	- No	+/- V		Some bicycle ergometer parts
1983 [98]	Roller ergometer	++ personal wheelchair	+ power output, speed	+/- fixed mass rollers	+ motor	- No	- No	- No	+/- V		
1987 [71]	Integrated ergometer	+/- integrated, limited	+ power output, speed	+ variable mass flywheel	+ brake	- No	- No	- No	+/- V	[71][b]	
1988 [99], 1989 [100]	Hybrid ergometer	?	+ power output	+/- fixed mass flywheel	+ brake	- No	- No	- No	+/- V	[99][a]	
1989 [64]	Integrated ergometer	- integrated wheelchair	+ power output, speed	- n.a.	- n.a.	- n.a..	- n.a.	- n.a.	+/- V		Isokinetic ergometer
1990 [7]	Integrated ergometer	+/- integrated, adjustable	++ 3D-handrim and seat kinetics	+ motor	+ motor	+ motor	++ motor	+ Yes	+/- V		Simulator

Article (year + reference)	Category	Ergonomics	Measurements	Inertia	Rolling resistance	Air drag	Slope[†]	Left/ right	Feed-back	Validity/ reliability	Remarks
1990 [101]	Roller ergometer	++ personal wheelchair	+ speed	+ variable mass roller	+ ?	- No	- No	- No	+/- V		
1991 [102], 1991 [62]	Flywheel ergometer	- integrated wheelchair	+ power output, speed	+/- fixed mass flywheel	- n.a.	- n.a.	- n.a.	- No	+/- V		Watt controlled
1993 [103]	Hybrid ergometer	- integrated wheelchair	+ power output, speed	+/- fixed mass	+/- Cateye CS1000	- No	+ Yes	+ Yes	- No		Turbo trainer
1993 [68]	Roller ergometer	++ personal wheelchair	+ power output, speed	+/- fixed mass flywheel	+ electromagnetic brake	- No	- No	+ Yes	+/- V	[68][b]	
1996 [96], 2006 [104], 2014 [105]	Hybrid ergometer	++ personal wheelchair	+ power output, speed	+/- fixed mass flywheel	+ brake	- No	- No	- No	+/- V		Rollers attached to flywheel
1993 [106], 1993 [59]	Roller ergometer	++ personal wheelchair	++ tangential force, speed	+ motor	+ motor	+ motor	++ motor	- No	+/- V		Commercial
1996 [48]	Roller ergometer	++ personal wheelchair	+ power output, speed	+ variable mass roller	+ electromagnetic brake	+	+	+ Yes	+/- V + P	[48][a,b]	
1996 [107]	Hybrid ergometer	++ personal wheelchair	++ tangential force, speed	+ variable mass roller	+ Velodyne	- No	- No	+ Yes	+/- V		
1998 [108]	Roller ergometer	++ personal wheelchair	++ torque, speed	+/- fixed mass rollers	?	- No	- No	- No	+/- V		
2000 [69]	Roller ergometer	++ personal wheelchair	++ tangential force, speed;	+ motor	+ motor	+ motor	+	+ Yes	+/- V	[69][a,b]	
2001 [56], 2008 [53], 2015 [109]	Roller ergometer	++ personal wheelchair	+ power output, speed	+ equal to 35 or 75 kg	+ hysteresis brake, adaptable to camber	- No	+ brakes	- No	+/- V	[56][a,b]	Commercial
2001 [57]	Roller ergometer	++ personal wheelchair	++ torque, speed	+/- fixed mass rollers	+ motor	+ motor	++ motor	+ Yes	+ V + P	[57][b]	

Article (year + reference)	Category	Ergonomics	Measurements	Inertia	Rolling resistance	Air drag	Slope†	Left/ right	Feed-back	Validity/ reliability	Remarks
2001 [85]	Roller ergometer	++ personal wheelchair	+/- distance and time	+/- fixed mass rollers	+ electromagnetic brake	- No	- No	+ Yes	+ V + G		Exergaming, portable
2002 [51]	Roller ergometer	++ personal wheelchair	+ power output, speed	+/- fixed mass rollers	+ adjusting internal friction	- No	- No	- No	+/- V + P		Adjustable incline
2003 [110]	Roller ergometer	++ personal wheelchair	+/- distance and time	?	?	?	?	+ Yes	+/- V		
2004 [60]	Roller ergometer	++ personal wheelchair	+/- distance and time	+ variable mass rollers	+ mechanical brake	- No	+	+ Yes	++ VR	[60]ᵇ	Immersive simulation
2007 [54], 2012 [111]	Roller ergometer	++ personal wheelchair	+ power output, speed	+/- fixed mass rollers	+/- passive	- No	- No	+ Yes	+/- V	[70]ᵇ	
2008 [112]	Roller ergometer	++ personal wheelchair	+/- distance and time	- roller mass not matched	+ brake	- No	- No	- No	?		
2011 [113]	Hybrid ergometer	++ personal wheelchair	+ power output, speed	+ flywheel ratio adjustable	+ brake	- No	- No	- No	+/- V		Rollers attached to flywheel
2013 [114]	Roller ergometer	++ personal wheelchair	++ torque, speed	+ variable mass rollers	+/- passive	- No	- No	+ Yes	+/- V		
2013 [55]	Roller ergometer	++ personal wheelchair	+ power output, speed	+/- fixed mass rollers	+ pneumatic brake	- No	- No	+ Yes	+/- V	[55]ᵇ	
2014 [61]	Roller ergometer	++ personal wheelchair	+ power output, speed	+ motor	+ motor	- No	- No	+ Yes	+ V + P	[61]ᵇ	Inclines and slopes
2015 [40]	Roller ergometer	++/- personal wheelchair	++ 3D handrim forces	+ motor	+ motor	+ motor	++ motor	+ Yes	+/- V	[40]ᵇ	Measurement wheels

† Full table in article; ‡ Mechanical simulation of slope (i.e. constant torque on roller); a: reliability study; b: validity study; c: more detail possible with instrumented treadmill; VR: virtual reality; V: visual; P: proprioceptive; G: game

Nevertheless, peak physical capacity can also safely be determined on treadmills or WCEs [120]. Wheelchair ergometry provides additional insights on top of peak oxygen uptake on performance and mobility in daily life [118] as the wheelchair settings and propulsion technique influence external power output. To reach peak physical capacity the load has to be incremented in small steps. On a treadmill this can be achieved by increasing resistance with a pulley or by increasing the slope inclines. Increasing gradient is less safe and has some ergonomic issues at high slopes. By using a pulley system, the posture of the participant does not change and the resolution of the increments is higher [120]. On a WCE the same tests can be performed if a variable braking system is implemented. As the wheelchair is tethered to the WCE the test is also safer than on a treadmill and the participant does not have to adjust their posture for inclines.

Anaerobic

A recent systematic review on anaerobic exercise testing in rehabilitation by Krops et al. [124] showed that for the upper extremities valid ACE, WCE, and overground tests are available. Overground testing is limited to sprint tests, while ACE and WCE can also be used for modified Wingate (mWAnT) protocols. In contrast, there appear to be no tests available for anaerobic testing on treadmills. The main outcome parameters for mWAnT and sprint tests are power output and peak velocity. Peak velocity may be useful in WCEs that cannot measure power output as it moderately correlates with peak power [125]. Again, the specificity of using WCEs can be seen as an advantage or disadvantage based on the goal of the measurement [122].

Synopsis & Perspectives

The aim of this review was twofold: first, to create an overview of the available lab-based equipment used in research. This was done by collecting and examining existing literature from internet databases and resulted in 50 unique wheelchair ergometers. Second, to assess this equipment on their ability to simulate and measure wheelchair propulsion in the laboratory. This was based on a number of criteria: accurate simulation of friction and inertia, reliable and valid measurement capabilities, realistic feedback and ergonomic soundness.

In general, treadmills were found to provide a mechanically realistic simulation of wheelchair propulsion, with the exception of air drag. Other advantages were: limited realistic steering (if no sliders/rubber bands are used), realistic contribution of trunk movement, and being able to use the personal wheelchair. WCE design was found to be more heterogeneous than treadmills. A surplus of different designs was found in the research literature. The WCEs were divided into three groups: roller-, flywheel-, and integrated ergometers. Each approach was found to have its own

advantages and disadvantages (Table 1.2). An advantage shared by all ergometers is the possibility of performing acceleration tasks.

With a vast array of different ergometers found in the literature, it is evident that there is little standardization among research centres. While having options allows researchers to choose the most pertinent device for their specific research questions, the comparability of results and applicability of existing knowledge remains somewhat limited without the standardization of measurement equipment [126]. This exemplifies the need for consensus among institutions on what lab-based research should look like.

The impact of using different WCE designs on propulsion technique and physiological parameters is not known as few comparison studies are available. Moreover, there is currently no overview of what device fits best in what situation. The diversity in equipment is especially troubling as research in this field often relies on small sample sizes, which increases the importance of combining evidence from different studies. In wheelchair sports this is even more apparent as restrictions on data sharing are further limit the availability of information [127]. Studies should therefore always include the specific settings and power output that was used.

In this paper the simulation aspect of an instrument was defined as the realistic simulation of the frictional and inertial components of (translational) wheelchair propulsion. A WCE does not by default simulate the translational inertia of overground propulsion. This implies that in passive systems the inertia of the roller or flywheel has to be matched to the weight of the wheelchair-user combination and active systems need to simulate the required inertia with a brake or motor. Most researchers aimed for a simulation of static friction to simulate rolling friction and a fixed inertia for their system. Some also incorporated the ability to simulate slopes and air drag. The ability to produce a valid and reliable mechanical strain is also closely tied to the measurement capabilities of the device.

It should be noted that the model used as a reference can be seen as a minimal model that describes most of the important forces acting on a wheelchair during straight forward motion. For physiological testing, a minimal model for straight-line wheelchair propulsion is probably sufficient. However, to accurately simulate the biomechanics of real-life situations, especially when turning is involved, requires more advanced models that include additional inertial and frictional properties [45]. From that perspective, most ergometers do give the ability to turn, but the simulation might be inaccurate. Another important factor is the effect of movements by the participants [50]. On a treadmill, these movements realistically influence the

Table 1.2. Generalization of the advantages and disadvantages of different ergometer types

	Treadmills	**Roller ergometer**	**Flywheel ergometer**	**Integrated ergometer**
Advantages	Can use personal wheelchair Mechanically realistic Realistic slope Good for fixed speeds	Can use personal wheelchair Anaerobic/sprint testing feasible	Uses commercially available parts Can be suitable for ergonomic testing	Suited for implementation of different simulation models Suitable for ergonomic testing
Disadvantages	Additional instruments required No air drag No acceleration tasks Difficult for fixed power outputs	Internal friction and inertia influences simulation Potentially high base load	Cannot use personal wheelchair Inertia is fixed Wheels not independent Not always suited for acceleration tasks	Cannot use personal wheelchair Electronic control system required Most expensive to build

wheelchair, but as most ergometers cannot incorporate kinematics in their simulation models, which might impact their validity somewhat.

There is a need for international collaboration to define the standards that WCEs should adhere to. If only the mechanics of the ergometers are considered, there are already large differences to be observed. It is unreasonable to expect that studies where a different mechanical strain is imposed on users can yield similar results. Concurrent validity of different WCEs is a research topic that has not yet been explored. The difference in equipment observed in this study adds to the variation already present in the testing protocols, further increasing diversity in research. All in all, there is a need for a commercially available line-up of wheeled mobility ergometry that allows a standardized protocol of wheeled mobility and testing in the clinical and adapted sports setting around the world [1].

Conclusion

The kinetic, kinematic, and physiological components of wheelchair propulsion can be studied in the laboratory on a treadmill or a wide variety of wheelchair ergometers. The simulation that these instruments provide is not always the same. Moreover, different levels of feedback are provided for the subjects. Different calibration methods were reported in the literature. In addition to this, researchers also employed different validation procedures. Often nothing was reported. Many questions about the measurement instruments that are used in studies are still left unanswered. Consequently, the evidence-base of performance enhancing factors and risks associated with wheelchair propulsion is limited due to the diversity in equipment, testing and measurement principles. Comparison studies are needed to evaluate the differences between approaches. There is an increasing need for ergometric standardization and general agreement to enable proper comparison of results.

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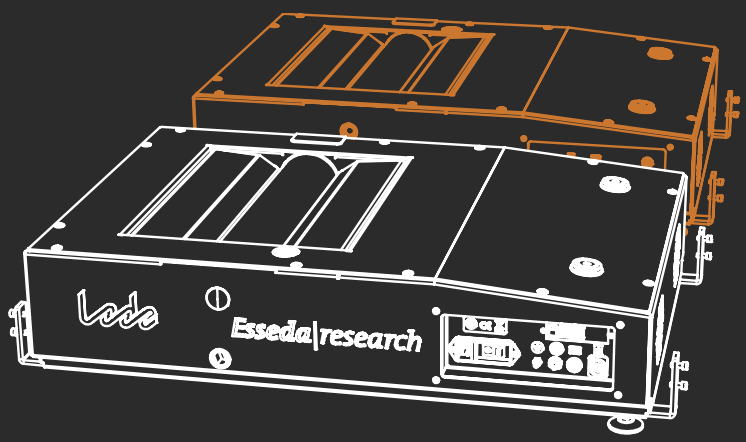
A new wheelchair ergometer



Technical note: a novel servo-driven dual-roller handrim wheelchair ergometer

R. de Klerk, R.J.K. Vegter, H.E.J. Veeger, L.H.V. van der Woude - IEEE Transactions on Neural Systems and Rehabilitation Engineering

Abstract

The measurement of handrim wheelchair propulsion characteristics and performance in the field is complicated due to the non-stationary nature of wheelchair driving. In contrast, the laboratory provides a constrained and standardisable environment to conduct measurements and experiments. Apart from wheelchair treadmills, dynamometers or ergometers for handrim wheelchairs are often custom-made, one-of-a-kind, expensive, and sparsely documented in the research literature. To facilitate standardised and comparable lab-based measurements in research, as well as in clinical settings and adapted sports, a new wheelchair ergometer was developed. The ergometer with instrumented dual rollers allows for the performance analysis of individuals in their personal handrim wheelchair and facilitates capacity assessment, training and skill acquisition in rehabilitation or adapted sports. The ergometer contains two servomotors, one for each rear wheel roller, that allow for the simulation of translational inertia and resistive forces as encountered during wheelchair propulsion based on force input and a simple mechanical model of wheelchair propulsion. A load cell configuration for left and right roller enables the measurement of effective user-generated torque and force on the handrim and the concomitant timing patterns. Preliminary results are discussed.

Introduction

Handrim wheelchair propulsion is characterized as a straining and inefficient activity that produces high repetitive strain on the upper extremities, resulting in an increased risk of musculoskeletal problems [1,2] and/or inactive lifestyle [3]. To improve physical capacity, wheelchair technology, individual propulsion technique, wheelchair propulsion biomechanics has been extensively studied [4–7]. Oftentimes studies use a combination of physiological, kinematic, and kinetic outcomes, requiring highly specialized equipment. Research is generally conducted on systems that simulate (to some extent) overground wheelchair propulsion, such as dynamometers, ergometers, or treadmills as they allow for measurements and training interventions to be performed in a controlled environment.

The validity of these lab-based experiments relies not only on the instrumentation and its measurement capabilities, but also on the simulation quality that these systems provide [8]. In addition to allowing for valid and reliable measurements of power output on the wheels, a simulator should provide an adequate simulation of the frictional losses, environmental conditions, and inertia of the wheelchair-user system. Finally, this should all be possible without altering the existing wheelchair-user interface [9] while facilitating other measurements such as metabolic cost, kinematics or electromyography.

Wheelchair propulsion on a motorized treadmill is probably the most realistic way of simulating wheelchair propulsion in the lab, but also has a number of drawbacks. On a treadmill the frictional forces and inertia are comparable to overground propulsion [10], while steering as well as visual position control are a necessity [11]. However, air resistance is absent, which becomes an important resistive force at speeds beyond 2m/s [12,13]. Although a standardised drag test can determine average power output without altering the wheelchair-user interface [14,15], it conveys no information about propulsion technique. Additionally, a treadmill mostly does not allow for sprinting or acceleration tasks to be performed and athletes can often exceed the maximum treadmill speed [16]. Other than that, treadmills can be too narrow and some labs and/or manufacturers therefore require a sidebar connected to the wheelchair to secure the stability of the wheelchair on the treadmill [17], which in turn could impact testing validity.

To cope with the aforementioned problems, a variety of different ergometers have been developed in the past [18]. Some are purely mechanical and others have electrical components to simulate overground propulsion mechanics. Inertia is generally simulated with a fixed-sized flywheel or weighted rollers and cannot be adjusted to an individual wheelchair-user combination. Few systems allow for the

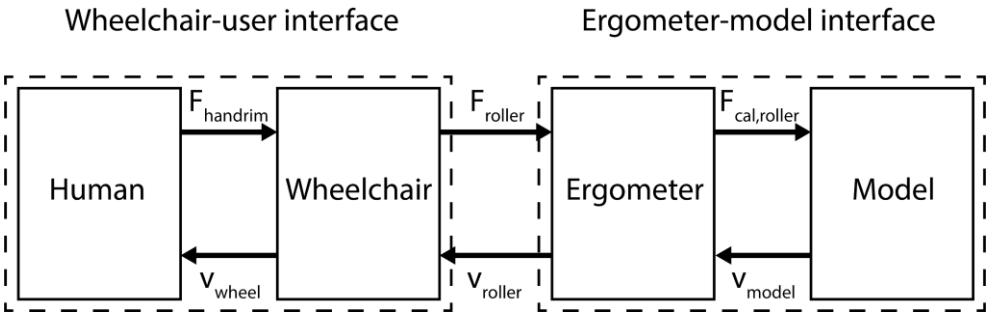


Figure 2.1. Schematic overview of ergometer feedback control scheme [38].

determination of push technique or even power output. Finally, they are often highly specialized pieces of equipment [8] that can only be operated in a few specialized labs. Widespread implementation of wheelchair biomechanical measurements can only be achieved when an easy-to-use system is available. Such a system should have standardized protocols and produce standardized results. Currently, no such system is available.

In the current paper, a newly developed commercial handrim wheelchair ergometer is presented that aims to provide the researcher, clinician, coach or athlete with a system that allows for accurate measurements for independent left and right-hand propulsion actions during exercise testing or training, while providing a mechanically realistic simulation of overground propulsion without adjusting the wheelchair-user interface. The ergometer includes a real time simulation of wheelchair propulsion based on a simple mechanical model of wheelchair propulsion [8], with user requirement dependent options such as inertia, rolling friction, air friction, and slopes. The system provides feedback on propulsion speed, power output, as well as push characteristics, if need be in a virtual environment. The purpose of this technical note is to provide a thorough description of the ergometer and to demonstrate its functionalities.

System overview

The Esseda is a novel wheelchair ergometer developed and produced by Lode BV (Groningen, The Netherlands) in collaboration with the University of Groningen, University Medical Centre Groningen, (UMCG), Centre for Human Movement Sciences, Groningen, The Netherlands. The ergometer measures the force production of the wheelchair user with a load cell configuration and simulates overground wheelchair propulsion in an admittance-control feedback loop (Figure 2.1). In the following sections the ergometer design, the measurement system, simulations, and calibrations will be further discussed.

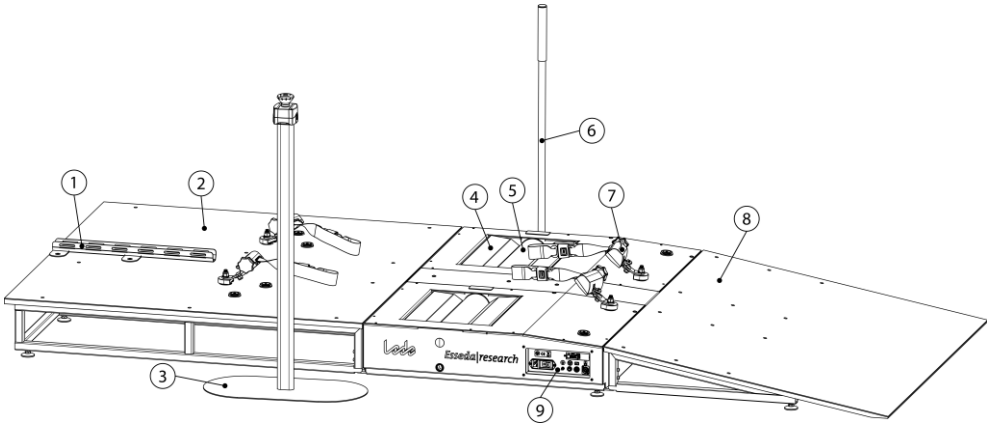


Figure 2.2. The Esseda wheelchair ergometer: 1. Wheeler extension; 2. Castor support board; 3. Emergency stop; 4: Alignment flaps (4x); 5. Roller (2x); 6: Alignment handle; 7: Straps (4x); 8: Ramp; 9: Communication module

Physical design

The ergometer (l=3.01m w=1.39m) consists of two nearly identical modules; one for the left wheel and one for the right wheel (Figure 2.2). Both modules have their own roller (l=0.43m, d=0.10m), servomotor, and control unit. The gap between the rollers is 0.36m wide. It includes a ramp (~10°) to allow wheelchairs to access the ergometer and a front panel to support the castor wheels with an extension for wheelers. A wheelchair can be centred on the rollers with the alignment system. Four tie downs are used to strap the wheelchair to the ergometer.

Electronics

Most electronic components are designed by Lode BV and custom-made for the ergometer. Both modules contain a loadcell (Utilcell Model 300, 50kg, combined error=0.017%, creep=0.016%, temperature drift=0.02% per 5ºK) with a Force Measure Print PCBA (printed circuit board assembly), a wheelchair control module PCBA, and a brushless servomotor (Schneider Electric BMH0703T11F2A, nominal output=900W, nominal torque=2.9Nm, peak stall torque=10.2Nm) with controller (Schneider Electric Lexium 32C). The motor:roller gear ratio is 2:1, resulting in a top speed of 12.5ms-1. The left module also contains the net filter and a communication module (Fieldbus Interface Module) in addition to the previously mentioned electronics. An overview of the electronics setup can be found in Figure 2.3. It is possible to send a 24V synchronisation pulse to the ergometer to synchronise data with other measurement devices.

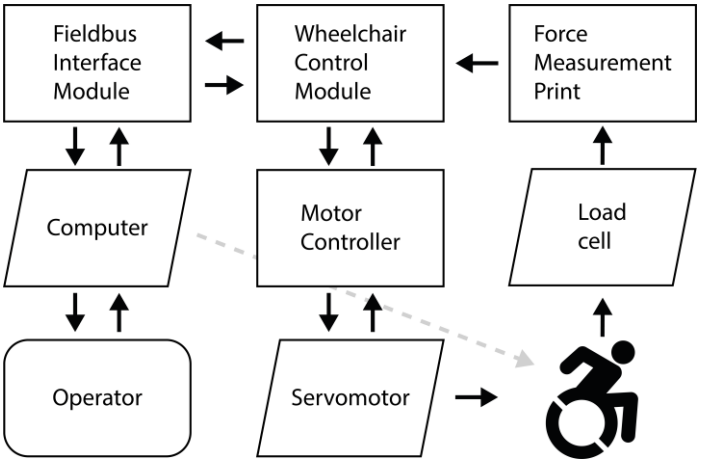


Figure 2.3. Overview of the electronics layout for the wheelchair ergometer. The wheelchair control module, motor controller, servomotor, force measurement print, and load cell are present in both sides of the ergometer.

Control software (firmware)

The ergometer is controlled with closed-source embedded firmware developed and maintained by Lode BV in the C programming language. Calculations in the control loop were verified by researchers at the UMCG. The control loop receives force data, calculates a new control speed, and sends the new control speed to the motor controller at 100Hz [19] in an interrupt-controlled loop. If the firmware receives erroneous values the ergometer automatically stops.

Associated software

The ergometer can be controlled with a (Windows) computer using the proprietary Lode Ergometry Manager (LEM) software. As of yet there is no Application Programming Interface (API) available. Friction and inertia can be set in LEM and they can be adjusted manually during a measurement or by setting a predetermined protocol. Feedback to the participant is also managed through LEM. Currently, LEM only supports showing speed, target speed, and basic heading to the participant. The default interface is a line-plot of left and right velocity with respect to a target line. Participants can infer directionality and speed from this plot in a natural way. Finally, LEM includes an analysis screen for the operator. This screen provides basic feedback on the distance travelled (m), speed (m/s), amount of work performed (J), and power output (W). The current version does not yet provide a detailed push-by-push analysis as used with measurement wheels and previous research ergometers [20–22]. A report (.pdf) can be generated with some precalculated common outcome parameters or raw force and velocity data (100Hz) can be exported to a Microsoft

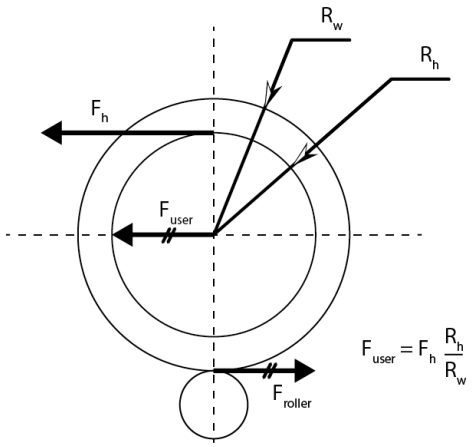


Figure 2.4. User generated effective component of the total handrim forces and torques (not shown) are indirectly measured through the roller [4]. This force can be inferred when information about the wheel size and handrim size is available.

Excel (.xls) file to allow for further processing by a researcher with their preferred data processing software (e.g., Python, R, MATLAB).

Measurement principles

The effective component of the user generated force (and torques) exerted in the handrim can be calculated from the force measured at the roller (F_{roller}) when information about the wheel size relative to the handrim size is available (Figure 2.4).

However, user force is not the only force acting on the roller. Two additional components need to be considered, namely: the internal friction of the wheelchair-ergometer system ($F_{frict,int}$) and the combined inertia ($F_{inertia,int}$) of the wheelchair wheel (usually unknown constant) and the roller of the ergometer (known constant).

$$F_{user} = F_{roller} = F_{meas} + F_{frict,int} + F_{inertia,int} \quad (1)$$

System friction

System friction is determined and adjusted for with a dynamic calibration [23]. During a dynamic calibration the ergometer samples the measured force from 0 to 2.5 m/s in steps of 0.28 m/s. Finally, a linear interpolation [24] function is used to estimate intermediate values and friction is assumed to be constant above 2.5 m/s. As an indication of friction found in the field: in a recent trial, 34 wheelchair users were measured in their personal everyday wheelchair. All tires were inflated to 6 bars (600kPa) before the measurement. A dynamic calibration was performed with the participant in an upright stationary natural position after the wheelchair had

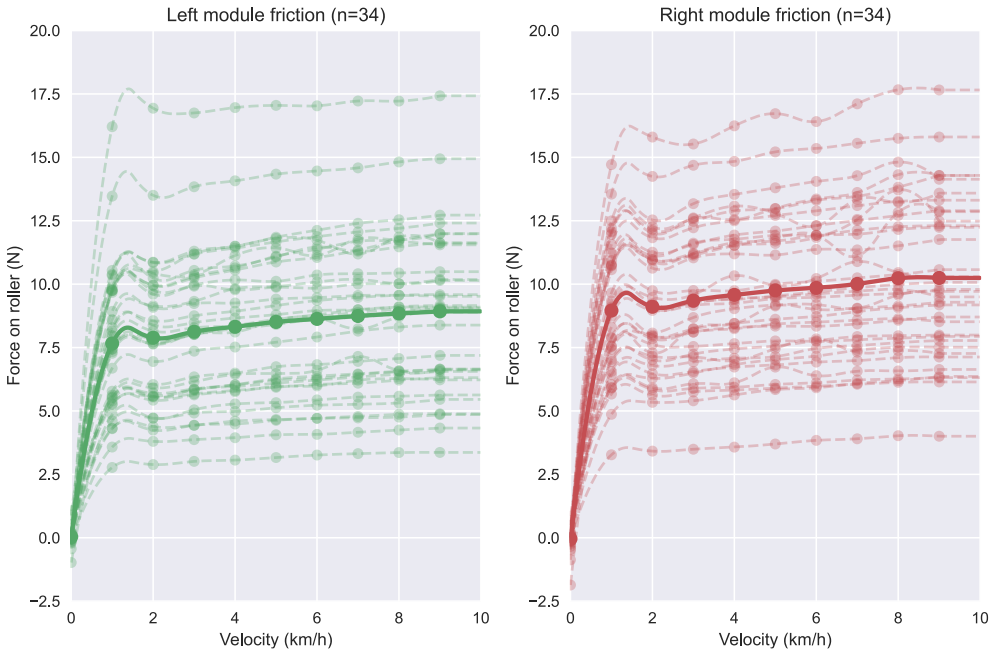


Figure 2.5. Individual and mean calibration curves for 34 participants in their personal wheelchair. Every calibration is different due to differences in mass, the distribution thereof, and the tension on the fastening straps. The spline (as shown) was replaced with a linear interpolation (firmware version 1.0.4).

been fastened on the ergometer. The resulting calibration curves are shown in Figure 2.5 for the left and the right module respectively. These calibration curves show that individual calibration curves are different for each wheelchair-user combination, in dependence of the wheelchair design/condition, fastening, and total weight, but follow a similar pattern. As such, they highlight the importance of a proper individual calibration before the onset of each measurement to compensate for inter- and intra-individual differences.

System inertia

During dynamic conditions the system inertia of both the wheelchair and ergometer significantly impacts the force readings of the ergometer. The inertia was estimated to be equal to 0.021 kgm^2 based on the CAD model of the roller, this was later corroborated through empirical testing (Figure 2.6). Likewise, the moment of inertia of the wheelchair wheel should be considered which adds approximately 0.004 kgm^2 to the system based on an assumed gear ratio of 1.0:5.3 (roller:wheel) and a moment of inertia of 0.117 kgm^2 [25]. As such, a compensation factor on the force signal for the combined system and wheel inertia (0.025 kgm^2) was included in the control loop.

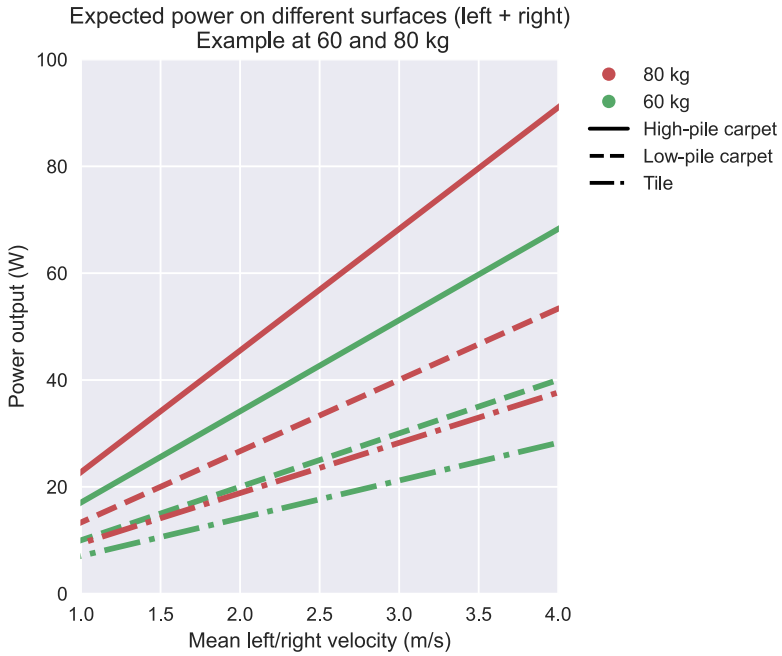


Figure 2.7. Example of rolling friction simulation effect on expected power output. Friction coefficients taken from Van der Woude et al. [39] for High-pile carpet (μ=0.028), Low-pile carpet (μ=0.20), and Tile (μ=0.15).

Simulation principles

The speed of the rollers is controlled by the servomotors. To calculate the speed at which the rollers should turn (v_{model}) the user generated effective force components (F_{user}) and simulated frictional components of wheelchair propulsion have to be considered together with the simulated mass for each side of the ergometer. The frictional components during wheelchair propulsion consist of the rolling resistance (F_{rr}), air resistance (F_{air}), and gravity when on a slope (F_{angle}). The resulting force (F_{res}) on one side of the wheelchair can then be calculated with (2):

$$F_{res} = F_{user} - \frac{F_{rr} - F_{air} - F_{angle}}{2} \quad (2)$$

The frictional forces are calculated separately for each wheel to allow both rollers to function independently of each other. With Newton's second law [26] we can then derive the equation of motion, given the combined mass of the wheelchair-user system (m_{tot}):

$$F_{res} = \frac{m_{tot}}{2} * a \quad (3)$$

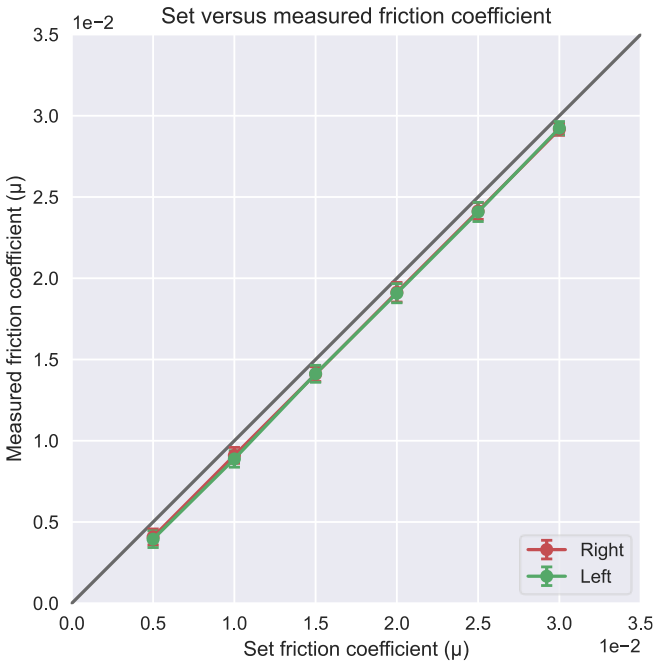


Figure 2.8. Comparison between set friction in the ergometer software and friction as determined by a coast-down test with a single wheelchair-user combination. Mean and standard deviations of five trials with five separate calibrations.

Where a is the acceleration of the wheelchair-user combination. Mass is divided by two to simulate both wheels independently. The rotational inertia of the wheels is not included. From this equation we can derive the acceleration of the wheelchair:

$$a = \frac{2 * F_{res}}{m_{tot}} \tag{4}$$

To compute the momentary speed of the wheelchair (v_{model}) for the motor controllers the acceleration is integrated:

$$v_{model} = \int a \, dt \tag{5}$$

The force on the wheelchair as a result of rolling resistance (Figure 2.7) can be modelled as a simple static friction that depends on the mass of the participant and the wheelchair combined multiplied by the rolling resistance coefficient (μ), the gravitational constant (g), and adjusted for inclination (α):

$$F_{rr} = m_{tot} * g * \mu * \cos(\alpha) \tag{6}$$

Air drag is a velocity dependent friction and at high speeds it becomes the most important source of friction. It is influenced by the velocity of the wheelchair (V_{wc}) and wind (V_w), the frontal plane area of the wheelchair user combination (A), the air density (D), and the aerodynamic drag coefficient (C_d):

$$F_{air} = 0.5 * D(V_{wc} - V_w)^2 A C_d \quad (7)$$

Finally, when the wheelchair is going up or down a slope (α) there will be a force acting on the system as a result of gravity, this is expressed in (8):

$$F_{angle} = m_{tot} * g * \sin(\alpha) \quad (8)$$

The advantage of using servomotors is that they can simulate realistic slope conditions (with exception of actual tilting of the wheelchair-user combination) where the wheelchair accelerates or decelerates accordingly, which is not possible with a braked or passive system.

Coast-down results demonstrating the simulation for a single participant in a regular handrim wheelchair are shown in Figure 2.8.

Measurement modes

The ergometer currently supports four different ways to evaluate wheelchair propulsion performance:

a) Isoinertial
b) Isokinetic
c) Isospeed
d) Isometric

The *isoinertial* mode allows for the realistic simulation of wheelchair propulsion with the previously described simulation model. It allows for the study of wheelchair biomechanics in realistic circumstances, but higher or lower (than overground) values for friction or inertia can also be used for specific research designs. Friction can be adjusted during a test or in a predefined protocol.

The *isokinetic* mode allows for realistic propulsion up to a set speed which acts like a ceiling. Participants can push the ergometer to this ceiling, but the motors will prevent going beyond that speed. In wheelchair propulsion this mode is still unexplored, but it could for example be used for max-testing.

This is similar to the *isospeed* mode in which the wheels turn at a constant speed that is not affected by external forces, which, for example, could allow for the study of

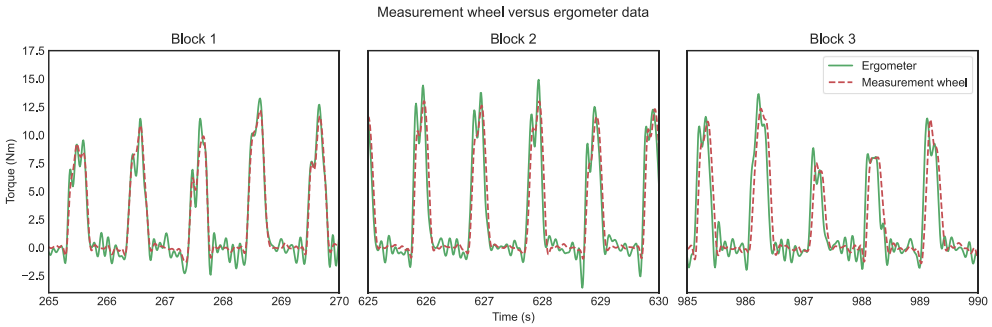


Figure 2.9. Comparison of measurement wheel and ergometer data. Both signals are low-pass filtered with a 10Hz 8th order Butterworth filter. Typical individual example is for the last five seconds of each four-minute block for the participant with the median cycle time.

speed-force relations in wheelchair propulsion. Additional safety precautions are strongly advised for this mode.

Finally, the *isometric* mode locks the breaks so a maximum force isometric strength test can be performed. This could be used to scale friction for sprint testing to the personal needs of the participant [27,28].

Processing data

Data can be exported from LEM to a Microsoft Excel file. When exported, one can use their own algorithms to extract meaningful variables. As only the effective force can be measured, the outcome measures that can be computed are more limited than with an instrumented wheel (e.g. Table 2.1). Examples of often used outcome variables are given below. Data should first be filtered before performing further calculations [19]. The ground frequency of the measurement system is approximately 19Hz. Moreover, irregularities in the rollers and wheelchair tyres, and changes in posture all add noise to the measurement signal. As such, data in the next section were first filtered in Python (Python Software Foundation, https://www.python.org) with a 10 Hz cut-off 8th order Butterworth filter for the force data and a 5 Hz cut-off 4th order Butterworth filter was used on the velocity data [29]. Software for the analysis of ergometer and measurement wheels is available at [30].

Preliminary results

To assess the measurement capabilities of the ergometer an initial comparison was made between the ergometer and an existing commercially-available Optipush (Max Mobility, Antioch, TN, USA) measurement wheel. The measurement wheel was used as a gold standard as it has a low error [31], obtains similar results as other measurement wheels [21], and can directly measure force on the handrim, whereas

Table 2.1. Common outcome parameters that can be calculated with only effective force, speed, and their derivatives [21]

Variable	Description	Equation
Spatio-temporal		
Push time (s)	Time from the start of positive torque to the end of positive torque for a push	$t_{end}(i)-t_{start}(i)$
Cycle time (s)	Time from the start of one push till the start of the next push	$t_{start}(i+1)-t_{start}(i)$
Push frequency (pushes/min)	Amount of pushes per minute	n_{pushes}/min
Contact angle (deg)	Amount of the handrim used per push	$\varpi_{end}-\varpi_{start}$
Kinetic		
Mean torque (Nm)	The mean torque (during an individual push)	$Mean_{start;end}(T)$
Peak torque (Nm)	The maximum torque (during an individual push)	$Max_{start:end}(T)$
Mean power (W)	Mean power (during the push phase)	$Mean_{start;end}(\omega*T)$
Smoothness (%)	Mean relative to the peak of push torque	Mean /Peak*100
Slope (Nm/s)	Peak toque divided by the time it takes to achieve peak torque	$Peak/t_{start:peak}$

Table 2.2 Comparison between ergometer and measurement wheel outcome parameters during steady-state propulsion (N=19)

Variable	Measurement wheel + (sd)	Ergometer value + (sd)	Difference + (%)	Absolute difference + (%)	Root mean square difference	Intraclass correlation coefficient (95% CI)[a]
Push time	0.32 (0.07)	0.32 (0.07)	0.00 (-0.07)	0.01 (2.96)	0.01	0.99 (0.97-0.99)
Cycle time	1.29 (0.54)	1.29 (0.54)	0.00 (-0.32)	0.00 (0.32)	0.01	1.00 (1.00-1.00)
Push angle	66.59 (14.17)	66.24 (13.84)	-0.35 (-0.52)	2.00 (3.00)	2.43	0.98 (0.96-0.99)
Mean torque	1.15 (0.30)	1.28 (0.27)	0.13 (11.18)	0.15 (13.41)	0.18	0.91 (0.53-0.97)
Mean power	4.24 (1.11)	4.68 (1.00)	0.45 (10.53)	0.55 (12.98)	0.64	0.92 (0.61-0.97)
Mean torque p.p	4.82 (1.30)	5.06 (1.41)	0.24 (4.98)	0.24 (4.98)	0.30	0.98 (0.62-0.99)
Peak torque p.p	8.33 (2.64)	9.08 (2.86)	2.39 (7.70)	2.39 (7.70)	2.88	0.95 (0.17-0.99)
Mean power p.p	17.78 (4.78)	18.52 (5.13)	0.74 (4.17)	0.74 (4.17)	0.97	0.98 (0.76-1.00)
Peak power p.p	31.06 (10.01)	33.46 (10.74)	2.39 (7.70)	2.39 (7.70)	2.88	0.96 (0.34-0.96)
Slope	61.88 (33.51)	79.57 (47.94)	17.68 (28.59)	17.69 (28.59)	24.05	0.84 (0.21-0.95)

a. Two-way, mixed effect, single rater, absolute agreement; p.p = per push; sd = standard deviation

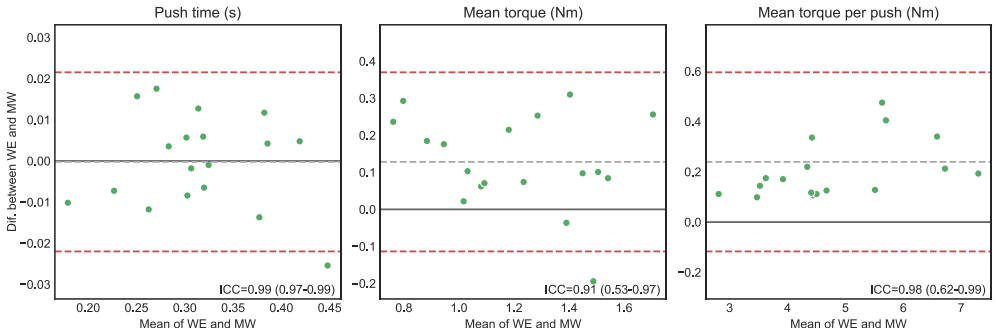


Figure 2.10. Bland-Altman plots for a spatio-temporal, kinetic, and push-by-push parameter. Intraclass correlations with 95% confidence intervals are included in the right bottom of the figure. The ergometer slightly overestimates torque.

the ergometer measures force more indirectly. The study was approved by the internal ethical committee (Ethische Commissie Bewegingswetenschappen). Nineteen trained able-bodied participants provided written informed-consent. They performed three blocks of four-minutes of submaximal wheelchair propulsion on the ergometer at 1.11 ms-1 (Figure 2.2) with friction standardized to the results of an overground coast-down test in a smooth hospital hallway. The last minute of each block was used for analysis, assuming steady state propulsion.

The results are shown in Table 2.2 and Figure 2.9. As expected, the ergometer signal is significantly noisier than the measurement wheel data. Spatio-temporal parameters (i.e. push time, cycle time, contact angle) all showed excellent agreement (Figure 2.10). However, mean power, torque, and torque per push all had excellent agreement with a moderate-excellent confidence interval and mean power per push showed an excellent agreement with a good-excellent confidence interval. Peak torques per push were higher for the ergometer and showed excellent agreement with the measurement wheel, however, the confidence interval ranged from poor-excellent. Slope showed good agreement, but the confidence interval ranged from poor-excellent.

Discussion

The current system can provide an appropriate simulation of overground wheelchair propulsion across a large range of frictional settings and speeds. In addition to this, it allows for push-by-push analysis of wheelchair kinetics and temporal characteristics through indirect force measurements.

The results of the coast-down tests show that, while consistent over multiple trials/calibrations, the simulated friction of the ergometer is slightly lower than expected based on the input settings of the model. This could be due to sensor drift

or due to slight changes in posture of the participant. Another possibility could be that some factors that were not calibrated for influence the force readings of the ergometer and thereby the simulation. The current calibration implementation of the ergometer includes static and viscous frictional components. However, second order (or higher) frictional components are not included. Moreover, the inertia compensation is fixed and might not be appropriate for all wheelchair wheels (especially measurement wheels). In this regard, a calibration based on ramp or sinusoidal input could provide information on the nonlinear properties of the wheelchair-ergometer combination [23].

The comparison between the ergometer and the measurement wheel showed similar results and good-excellent agreement for most outcome parameters. The ergometer appears to slightly overestimate the force/torque production of the participants. More detailed calibrations with reference weights should determine the origin of this difference. Moreover, further studies using a wider range of speeds and loads should be performed to provide a broader picture of measurement accuracy. The noise could be reduced by improving the balance of the rollers, using better wheelchair wheels and construction, and investigating the effect of adjustments by the motor controller on the measurement signal.

Another difference between the ergometer and a measurement wheel is that roller ergometers by definition cannot measure handrim propulsion force. Instead, they measure the effect of the combined hand torques and propulsion force, which is pooled into one component: the effective-force component. While most wheelchair biomechanics propulsion outcomes do not require detailed (3D) information (Table 2.1), it is essential for modelling segmental torques like in the Delft Shoulder and Elbow Model [32].

Nevertheless, the advantage of using the ergometer instead of the measurement wheels apparent [1]. The ergometer can be used with any wheelchair configuration without changing the wheelchair-user interface. This allows for faster testing, which is especially important in a clinical/sports environment. It also makes the ergometer more suitable for measuring in a sports environment as measurement wheels are usually not built for sports wheelchairs and the high strains experienced during testing of athletes [33]. Finally, the ergometer allows for the testing of different wheelchair configurations (e.g. handrim types), whereas in most measurement wheels the configuration is fixed.

The ergometer in this paper uses a simple mechanical model to simulate wheelchair propulsion. It assumes straight-line propulsion, that the wheels do not slip on the floor, the movements of the subject do not influence the wheelchair, and that the

castors do not contribute to the dynamics of the wheelchair [34]. Very detailed models of wheelchair propulsion include several wheelchair parts and the centre of mass [35,36]. However, it is not feasible to include kinematic information of body segments in a model that also directly controls the ergometer without impacting the ease-of-use. There are also models specifically built for simulating curvilinear propulsion [34], however, the majority of ergometer-based testing is on straight-line propulsion and the accuracy increase is only within the range of a few percent. The difference is between models is more pronounced for the inner wheel than for the outer wheel. Resultingly, when turning movements are of interest (e.g. for simulating tight spaces) a curvilinear model should be adopted.

The ergometer will be improved by collaborating with various research institutes, rehabilitation centres, and sports organizations. In our future work with the system the measurement performance of the ergometer will be tested with reference weights during static and dynamic conditions. For widespread adoption, standardized protocols for use in all application areas will have to be designed and validated together with researchers and clinicians. As wheelchair users form a very heterogeneous group ranging from people that seldomly propel the wheelchair themselves to Paralympic athletes with a variety of different impairments [4], it is important that the ergometer can facilitate a wide range of users, wheelchairs, and protocols.

The ergometer facilitates researchers by providing access to raw data through LEM. Moreover, the ergometer can be used in rehabilitation centres to allow for the training and assessment of patients. In this case, submaximal exercise at low loads will be more common than in research or sports settings. Practitioners should be able to quickly identify inefficient propulsion patterns (i.e. high push frequency, low contact angle [37]) and have patients adjust where necessary. In contrast to researchers, they are less interested in the raw data and they want LEM to provide the analysis for them. In the future, efficient propulsion techniques might be promoted with the use of exergames or different forms of (visual) biofeedback. Having an API available could aid in this process. Finally, in athlete assessment and training the athletes and their coaches need reliable data to monitor the status of the exercise capacity and performance ability of the athlete. The system needs to be able to withstand the high forces that a wheelchair athlete can exert on the ergometer during peak or sprint testing. Athletes also need to be able to train with standardized protocols. Again, the ergometer should provide quickly interpretable data and standardized protocols for this group of users based on input from researchers.

Conclusion

The current wheelchair ergometer is able to measure and simulate wheelchair propulsion. It is presently being used for different experiments. At first, these are related to the validation of the ergometer and associated measurement protocols. Then, for studying both the physiology and biomechanics of wheelchair propulsion from early rehabilitation to sports. A variety of use-cases were presented in this paper. Our preliminary results demonstrate the functionality of the ergometer. Future research should focus on reducing the signal noise and producing standardized protocols with easy to interpret input and output.

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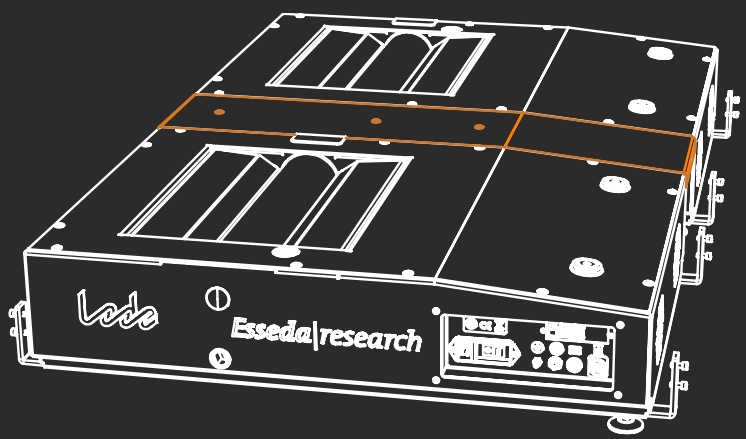
3

A low-cost portable calibration method



Development of a portable low-cost system for the validation of wheelchair roller ergometers

R. de Klerk, P. Spada, R. Muhametaj, M. Lancini, L.H.V. van der Woude, R.J.K. Vegter – Submitted

Abstract

Wheelchair ergometers are widely used in research, clinical practice, and sports environments. The majority of wheelchair ergometers are roller systems that allow for wheelchair propulsion in the personal wheelchair on one or two (instrumented) rollers. Oftentimes these systems are only statically calibrated. However, wheelchair propulsion is dynamic by nature, requiring a dynamic validation process. The aim of the current project was to present a low-cost portable system for the dynamic validation of wheelchair roller ergometers. The uncertainty of the tangential force on the roller as modelled by the validation system is determined. The force measured by the ergometer can be compared with the force predicted based on the characteristics of the validation system. Uncertainty of the validation system was found to be lower than the reference ergometer, indicating that the validation system can be used to validate ergometers.

Introduction

State of the art

The ergometric assessment of handrim wheelchair propulsion is essential for research [1,2], clinical practice, and sports environments [3–5] as it can be used to optimize the wheelchair, wheelchair user, and the wheelchair-user interface [6]. As a result, wheelchair ergometry and wheelchair ergometers are an active topic of research [2,6]. Ergometers provide a constrained environment in the sense that it can be tightly controlled with regards to speed, load, and feedback [6]. Yet, this environment is also unconstrained in the sense that it is boundless, allowing for the collection of a high number of push cycles.

A wide variety of ergometers has been developed over the past decades, however, in recent years, there is a renewed interest in roller ergometers as they provide a number of advantages over other methods [2]. The main advantage of these ergometers is that they provide an opportunity to measure the wheelchair user in their personal wheelchair while simultaneously collecting data on their power output. In contrast to treadmills, they also allow for sprint testing [7]. The main outcome measures for a wheelchair ergometer, like most power meters, are the torque and velocity. Which are used to compute power as an indirect measurement. For the purpose of this paper the measured torque applied by the wheelchair-user on the roller of the ergometer is viewed as the primary outcome measure of the system.

Calibration and validation of an ergometer is always required regardless of the origin or type of the ergometer [8]. The force component at the interface between wheel and ground, is particularly relevant for most studies, since it is involved in propulsion, and could be considered an alternative expression of the torque measured. Using this variable, most of the variables (e.g. peak/mean force, push/cycle time, etc.) relevant to handrim wheelchair propulsion research can be computed [9,10]. Force transducers are usually calibrated and validated in static condition, since a high accuracy at low-cost is achievable using gravity and known masses. The working conditions of ergometers, however, are far from static: rolling resistance and inertial parameters affect the measurement principle used, therefore a mechanical validation system for such ergometers should involve testing in dynamic conditions.

Devices for cycling applications were subject to numerous studies, focusing more on repeatability than traceability; an unbroken chain of comparisons to stated references. The calibration of bicycle ergometers has been shown to contain errors for different types of ergometers [11,12]. They are usually validated by comparison

with a reference mobile power meter for cycling, comparing only power output measured by both under steady-state conditions. However, these tests cannot be directly applied to wheelchair ergometers as calibration or reference devices are not widely available.

The current paper proposes a low-cost validation system for the force measurements of a roller ergometer since measuring the rotational speed of the wheel could be easily accomplished using robust and affordable sensors (encoders, optical, magnetic detectors, etc.) with a good accuracy [13,14], the target of the validation system proposed was focused mainly on a single outcome measure: the contact force at the wheel-roller interface, along the direction of motion. The uncertainty of the approach is determined and is compared to an existing wheelchair ergometer.

Proposed approach

A cantilever system containing a wheel with a high inertial mass is proposed as a dynamic validation system. The system should be low-cost, to allow any lab to reproduce the setup without having a manufacturer involved. It also must be portable, to allow for inter-laboratory comparison of different ergometers. The system should also be completely independent from the ergometer data and guarantee an uncertainty lower than the desired uncertainty of the ergometer being tested.

The proposed approach focuses on a reference device, able to apply a known tangential force on the ergometer roller at different rotational speeds. The idea is based on three different applications of a rolling test, with a wheel of known inertial properties, without the need of a costly reference transducer such as a power meter. On every test, the wheel is to be driven up to a known speed, then decoupled from its motor source, and slowed down without external contribution. This will allow for the assessment of the forces inducing a deceleration from the profile of the rotational speed of the wheel over time given the inertial properties of the wheel and its dimensions.

The first test is used to characterize the validation system. During the first test the wheel is free-standing, only to be slowed down by air and internal friction associated with the wheel itself. In the second test, the wheel will be in contact with the roller of the ergometer, but no resisting torque is applied by the ergometer. In this way, the contributions of the rolling resistance associated with the viscous deformation of the pneumatic wheel, as well as the bearing friction, can be assessed. Finally, in the third test, the wheel will be counteracted by the ergometer, which can be juxtaposed to the behaviour of the wheel in the first two tests.

The uncertainty of the tangential force measured by the ergometer can then be assessed by comparing the measurements from the ergometer with the estimates given from the validation system, which can be considered as reference values.

Methods

Mathematical model

An external motor is used to drive the wheel up to the desired velocity, then it is disconnected using a freewheel system, which is suddenly engaged. This limits the forces acting on the wheel other than those dependent on the mechanical properties of the wheel (friction, inertia, etc.) and those associated with the tangential force applied by the ergometer.

The proposed method relies on the solution of the dynamical equilibrium of the wheel, in order to describe the wheel angular acceleration $\dot{\omega}(t)$ as a polynomial function of the rotational velocity $\omega(t)$, as shown in Eq. 1.

$$\dot{\omega}(t) = -k\,\omega(t)^2 - c\omega(t) - \alpha \quad (1)$$

Where k is a parameter related to the squared rotational velocity, c to the linear rotational velocity, and α is independent of the velocity. These variables assume different meaning in different test configurations and are used to assess the desired quantities. The value of k, c and α for each test are obtained with a numerical solver (a Gauss-Newton algorithm which implements a non-linear least square regression, developed in MATLAB), minimizing the error between the measured angle and the estimated one.

Stand-alone test

The mechanical model used is depicted in Figure 3.1 with the wheel modelled as a rigid disc of radius R_w in a bi-dimensional space. Three forces are considered acting on the wheel: T_a, the torque associated with air friction, T_{bw}, the torque due to the rolling friction of the wheel bearings, and $J_w\dot{\omega}$, due to the moment of inertia J_w of the wheel.

Since no other forces are acting, the equilibrium of the torques is given by Eq. 2, where $T_{bw} = f_b R_b g M_w$ is the bearings friction torque and $T_a = C_w\omega(t)^2$ the torque associated with the drag.

$$J_w\dot{\omega} = -T_a - T_{bw} \quad (2)$$

If we define M_w as the wheel mass (including the masses added to increase inertia), R_b the bearing radius, C_w an aerodynamic factor (combining drag factor, air density and area), f_b the bearing friction coefficient, we can rewrite Eq. 2 as Eq. 3:

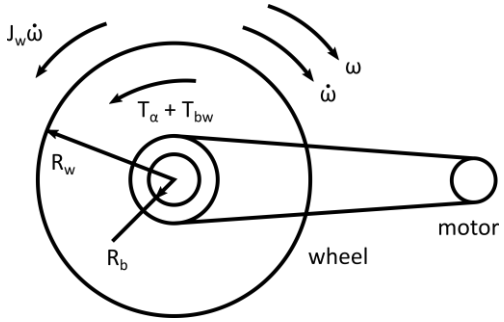


Figure 3.1. Mechanical model of the wheel in standalone configuration. T_a: resistance torque due to the air friction, T_{bw} : resistance torque of the wheel bearings, ω: wheel angular velocity, ώ: wheel angular acceleration, J_w: wheel moment of inertia.

$$\dot{\omega} = -[C_w/J_w] * \omega(t)^2 - [f_b R_b g M_w/J_w] \tag{3}$$

This allows us to define the parameters k_{SA} and α_{SA}, which will be the results of the first set of tests, as specified in Eq. 4, where the 'SA' subscript stands for parameters assessed during stand-alone tests.

$$k_{SA} = C_w/J_w \quad \alpha_{SA} = f_b R_b g M_w/J_w \tag{4}$$

Zero-torque mode

The second set of tests is defined to assess the dissipative actions that are due to internal elements and rolling resistances [15] caused by the interaction between the ergometer and the wheel. The roller of the ergometer in this configuration is set as completely passive, with no active torque except the resisting torque offered by the friction of the roller bearings and its inertia.

The mechanical model associated with this configuration is shown in Figure 3.2, with the same nomenclature used in the standalone model, and the addition of T_{br}, the torque due to the roller bearings, J_r, the inertia of the roller and T_{rr}, the rolling resistance of the wheel associated with the deformation of the tyre.

The dynamic equilibrium at the wheel can be expressed by transferring the T_{br} and J_r terms from the roller to the wheel, as shown in Eq. 5:

$$-J_w\dot{\omega} = C_w\omega^2 + f_b R_b g M_v + J_r\dot{\omega}\left(\frac{R_w}{R_r}\right)^2 + f_{rr}\omega + f_{br}R_{br}(M_w + M_v + M_r)g\frac{R_w}{R_r} \tag{5}$$

T_{rr} considers the rolling resistance as a dissipative-only phenomenon related linearly to the speed. While this is usually not the case in car and cycle application, this assumption is valid at low speed [16]. T_{rr} could be rewritten as a factor f_{rr} to be

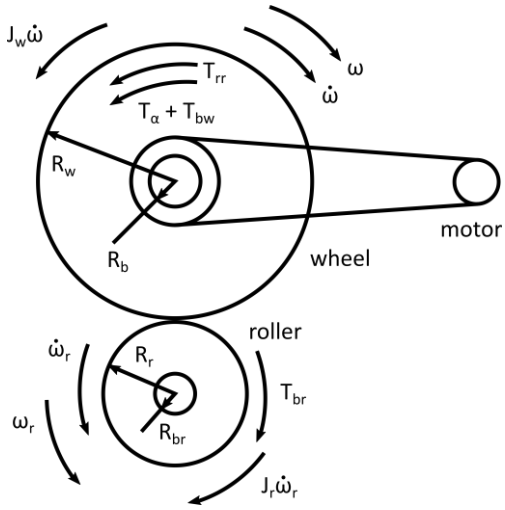


Figure 3.2. Mechanical model of the wheel in zero-torque configuration. T_a: resistance torque due to the air friction, T_{bw} : resistance torque of the wheel bearings, ω: wheel angular velocity, ω̇: wheel angular acceleration, J_w: wheel moment of inertia, ω_r: roller angular velocity, $\dot{\omega}_r$: roller angular acceleration, T_{rr}: rolling resistant torque, T_{br}: resistant torque due to roller bearing's friction

multiplied by ω(t). A separate friction factor f_{br} is instead assumed on the roller bearings.

Concerning the forces transmitted, the wheel bearings are working in a different way: the wheel mass M_w is fully supported by the roller, and the wheel bearings are pushed down by an external additional load M_v. The force acting on the wheel bearing is now just $M_v g$, the vertical force between wheel and roller is $(M_v + M_w)g$ and the force acting on the roller bearing is given by $(M_v + M_w + M_r)$ where M_r is the mass of the roller and g is gravity.

Following the same approach as earlier, the dynamic equation could be rewritten as a differential equation governing angular acceleration, as shown in Eq. 6:

$$\dot{\omega}(t) = -\frac{C_w}{J_w + J_r\left(\frac{R_w}{R_r}\right)^2}\dot{\omega}(t)^2 - \frac{f_{rr}}{J_w + J_r\left(\frac{R_w}{R_r}\right)^2}\omega(t) + \frac{T_{bw} + f_{br}R_{br}(M_w + M_v + M_r)g\frac{R_w}{R_r}}{J_w + J_r\left(\frac{R_w}{R_r}\right)^2} \quad (6)$$

With the parameters of Eq. 1 assuming the values shown in Eq. 7, where the 'ZT' suffix indicated that they are assessed using the numerical optimization of experimental data obtained during zero-torque tests.

$$k_{ZT} = \frac{C_w}{J_w + J_r\left(\frac{R_w}{R_r}\right)^2} \quad c_{ZT} = \frac{f_{rr}}{J_w + J_r\left(\frac{R_w}{R_r}\right)^2}$$
$$\alpha_{ZT} = \frac{T_{bw} + f_{br}R_{br}(M_w + M_v + M_r)g\frac{R_w}{R_r}}{J_w + J_r\left(\frac{R_w}{R_r}\right)^2} \tag{7}$$

Full model (Variable Torque)

The last configuration used is the closest to the actual working condition of the ergometer: the wheel is applied on the roller, which provides a resistive torque thanks to its motor. This configuration, shown in Figure 3.3, is similar to the zero-torque mode, with the only addition of a constant resistive torque T_r actively generated by the ergometer, which can be set at different constant values thanks to the ergometer controller. The dynamic equilibrium, similarly to the previous case, could be described by Eq. 8:

$$J_w\dot{\omega} = -T_a - T_{bw} - J_r\dot{\omega}\left(\frac{R_w}{R_r}\right)^2 - (T_{br} + T_r)\frac{R_w}{R_r} - T_{rr} \tag{8}$$

Applying the same substitutions as before, the differential equation describing angular acceleration can be rewritten as shown in Eq. 9:

$$\dot{\omega}(t) = -\frac{C_w}{J_w + J_r\left(\frac{R_w}{R_r}\right)^2}\omega(t)^2 - \frac{f_{rr}}{J_w + J_r\left(\frac{R_w}{R_r}\right)^2}\omega(t) + \\ -\frac{T_{bw} + f_{br}R_{br}(M_w + M_v + M_r)g\frac{R_w}{R_r} + T_r\frac{R_w}{R_r}}{J_w + J_r\left(\frac{R_w}{R_r}\right)^2} \tag{9}$$

With the parameters of Eq. 1 assuming the values shown in Eq. 10, where the 'VT' suffix indicated that they are valid for the variable-torque tests.

$$k_{VT} = \frac{C_w}{J_w + J_r\left(\frac{R_w}{R_r}\right)^2} \quad c_{VT} = \frac{f_{rr}}{J_w + J_r\left(\frac{R_w}{R_r}\right)^2} \tag{10}$$

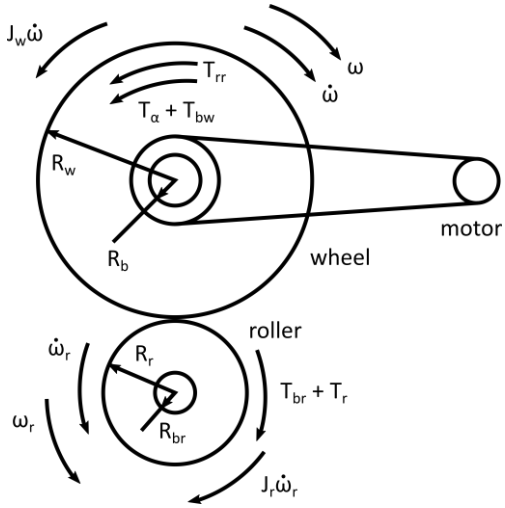


Figure 3.3. Mechanical model of the wheel in variable torque mode. T_a: resistance torque due to the air friction, T_{bw} : resistance torque of the wheel bearings, ω: wheel angular velocity, ώ: wheel angular acceleration, J_w: wheel moment of inertia, ω_r: roller angular velocity, $\dot{\omega}_r$: roller angular acceleration, T_{rr}: rolling resistant torque, T_{br}: resistant torque due to roller bearing's friction, T_r: resistive torque actively generated by the ergometer

$$\alpha_{VT} = \frac{T_{bw} + f_{br}R_{br}(M_w + M_v + M_r)g\frac{R_w}{R_r} + T_r\frac{R_w}{R_r}}{J_w + J_r\left(\frac{R_w}{R_r}\right)^2}$$

It can be noticed that, given the same vertical load condition, and the same inertia, $k_VT = k_ZT$, as well as $c_VT = c_zT$.

Tangential force assessment

The tangential force F_t, that is exchanged between wheel and roller, can be obtained by considering that F_t, multiplied by the wheel radius, must equal the torques acting on the wheel, as shown in Eq. 11:

$$F_tR_w = J_w\dot{\omega}(t) + T_{bw} + T_a + f_{rr}\omega(t) \tag{11}$$

$$F_t = [J_w\,\omega(t) + f_bR_b\,gM_v\; +\; C_w\,\omega^2(t) + f_{rr}\omega(t)]\frac{1}{R_w} \tag{12}$$

Therefore, it is possible to compute the value of the tangential force, as shown in Eq. 12, starting from the rolling resistance (computed from the c value the zero-torque mode), the air friction and the wheel bearings friction (computed from k_SA and α_SA), the wheel radius R_w and its inertia J_w.

The values of α, k, c, obtained by fitting of experimental data in all three different configurations can be used to assess different properties of the wheel-roller system. The wheel bearings friction f_b can be estimated using the value of $\alpha_S A$, the $c_Z T$ values lead to determine the rolling resistance coefficient f_{rr}; finally, in variable-torque mode, from the values of T_{bw} combined with $c_V T$ were computed the active resistance torque of the roller T_r.

Testing

Protocol

Following a standardized protocol, trained observers (PS, RM) performed all tests in the three described conditions (Standalone, Zero Torque, Variable Torque), with different working parameters. The parameters taken into account were:

1. the inertia of the wheel, set by adding 0, 4, 8 or 16 masses of about 0.3 kg each to the wheel
2. the initial speed of the wheel, set to 5, 10, 15, or 20 km/h
3. the vertical mass on the wheel axis, set using only the wheel weight or adding an additional 4 kg to its axle
4. the active resistance of the roller, set to 0.1, 0.2, 0.3 or 0.5 Nm

For each combination of parameters, the test was repeated 5 times.

During tests in standalone configuration there was no contact between the wheel and the roller, so only the inertia and the initial speed of the wheel were taken into account, totalling 16 combinations repeated 5 times each. During tests in zero-torque mode the resistance torque of the roller is null, therefore only the other three parameters were changed, for a total of 32 combination repeated 5 times each. In the full test with variable torque all the four parameters were taken into account, with a total of 128 combinations repeated 5 times each.

For each variable-torque test, the quantities depending on previous tests, were estimated by averaging the α, k, c values resulting from the curve fitting of the zero-torque and standalone tests having the same parameters combination.

Estimation of inertia

The moment of inertia of the wheel, in its 4 different configurations, was measured using the trifilar pendulum method [17] and a motion capture system (Optotrak, NDI, Canada). All the masses (the inertial masses of the wheel, the wheel mass, the vertical mass) were weighted using a laboratory weighing scale. The dimensions of the wheel and of the roller were measured directly, while the friction coefficients

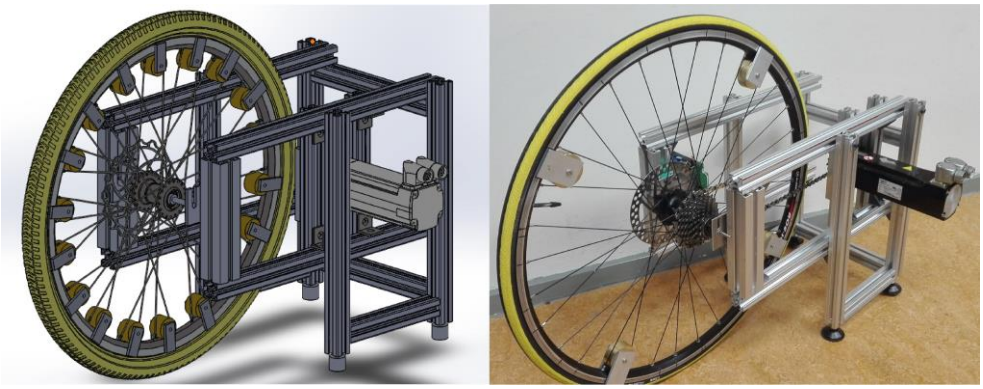


Figure 3.4. Instrumented cantilever wheel for ergometer validation.

and the dimensions of the bearing were extracted from their data sheets and their relative uncertainty was assumed to be about 1%.

Ergometer testing

Since the ergometer datalogger and the validation system are synchronized, the error was computed as the difference between the instantaneous values of the tangential force $\hat{F}_T$ measured by the ergometer and the same value F_T assessed by the instrumented wheel.

Materials

Instrumented wheel

The designed system is a device which produces forces at the ergometer roller edge using its own inertia during deceleration. The guidelines followed during the design of the system were:

- cost: to limit the overall cost of the system
- availability: to use off-the-shelf components
- portability: to ensure the system could travel between different labs

This structure (Figure 3.4) consists of a 24-inch (0.61 m) rear bike wheel (model Alexrims ace 19) positioned on vertical guides attached to a support aluminium structure. The wheel is connected to a brushless servomotor (model BMH0703T11F2A, Schneider Electric, France) with a nominal speed of 2000 rpm and nominal torque of 3.1 Nm. However, almost any motor would suffice. The motor controls the speed of the bike wheel and drives it through a freewheel system which allows for a mechanical decoupling of the motor torque from the wheel. At the rim, brass masses were clamped to allow varying the inertia of the wheel. The design of the masses is mostly attributable to allow for easy installation on the system. In fact,

it is possible to attach and detach them without removing the wheel from the cantilever system.

The system is equipped with a position sensor based on a low-cost custom-made phonic wheel. A standard bike brake disc coupled with a photocell was used for this purpose (model MOCH25A, Kingbright, Taiwan). The rising and falling edges of the photocell signal were recorded and treated as an encoder signal. The acquisition system used was a cDAQ 9172 unit (by National Instruments, USA) equipped with an encoder acquisition module (NI9411), to continuously acquire the phonic wheel signal at 25 kHz, and a signal generation module (NI9475) to provide a trigger signal to synchronize the data with the ergometer data logger. The phonic wheel was calibrated against the servomotor encoder, used as reference, during a set of constant speed tests. The systematic errors due to physical imperfections (hole position accuracy, eccentricity, ellipticity) were corrected using a lookup table, as suggested in [18]. The software used to convert these signals into position information was developed in Labview (by National Instruments) and is provided in the additional materials.

Esseda ergometer

The Esseda is a computer-controlled servo-driven dual-roller ergometer developed by Lode BV (Groningen, The Netherlands) in close collaboration with the University Medical Centre Groningen (UMCG, Groningen, The Netherlands). Every day and sports wheelchairs can be mounted on the ergometer and secured with four tie

Table 3.1. Values and uncertainties of the evaluated parameters.

Parameter	Unit	Value	Uncertainty
Roller b. r.	m	$16.64 \cdot 10^{-3}$	$0.02 \cdot 10^{-3}$
Wheel b. r.	m	$47.10 \cdot 10^{-3}$	$0.47 \cdot 10^{-3}$
Wheel r.	m	$299.5 \cdot 10^{-3}$	$0.29 \cdot 10^{-3}$
Roller r.	m	$50.1 \cdot 10^{-3}$	$0.29 \cdot 10^{-3}$
Vertical masses	kg	$390.5 \cdot 10^{-3}$	$1.4 \cdot 10^{-3}$
Brass masses	kg	$399 \cdot 10^{-3}$	$0.5 \cdot 10^{-3}$
Roller mass	kg	13.96	0.14
Wheel mass	kg	2.04	0.02
Roller b. fr.		0.05	0.0005
Moment of i. 1	kgm^2	0.11	0.1%
Moment of i. 2	kgm^2	0.22	0.9%
Moment of i. 3	kgm^2	0.34	0.3%
Moment of i. 4	kgm^2	0.56	0.2%

b.: bearings; fr.: friction; r.:radius, i.:inertia

downs. User generated bi-manual forces are transferred through the roller-motor assembly to a load-cell. The ergometer has a number of different modes to measure wheelchair specific performance. In the current paper the iso-inertial and zero-torque mode are used. In the zero-torque (freewheel) mode, which was added for this validation setup, the motor controller is set to torque mode with torque set to zero. In this mode the ergometer is fully passive with only internal friction and inertia acting on the roller. In the iso-inertial mode the ergometer simulates overground propulsion according to a simple mechanical model of wheelchair propulsion in an admittance-control feedback loop. This is the most commonly used mode and allows for the simulation of different wheeling environments or different artificial loads. As such, this mode was used for testing in the variable torque condition.

Results

Measurement uncertainty

In order to provide a reliable validation, the uncertainty associated with the reference measurement should be less than the uncertainty of the measurement system under test. To guarantee this condition, an uncertainty budget was computed for every test condition following the standard approach suggested by [19]. Table 3.1 shows all known and measured parameters' values, along their standard uncertainties.

The uncertainty associated with the parameters α, k, c obtained by fitting experimental data were assessed using the residuals of the fit, following the approach suggested in [20]. For all three parameters the relative standard uncertainty ranged from 3% to 23%, depending on the configuration and the parameters combination.

The uncertainty associated with the tangential force was assessed as combined uncertainty of all parameters present in Eq. 12, following the standard approach suggested in [19]. The combined uncertainty of the tangential force, $U_{95\%}(F_T)$, was extended to a 95% level of confidence. In all tests this extended uncertainty ranged between 0.05 and 0.15 N, depending on the actual speed and the test condition.

The value of the residual is consistently higher than the value of the uncertainty of the measurement wheel, indicating that the cantilever system is able to serve as a reference system to validate the ergometer.

Sensitivity with respect to investigated variables

Since different conditions lead to similar results in term of errors, the sensitivity of the error with respect to different parameters used to define the tests was

investigated. Figure 3.5 shows the variation of the residuals of the tangential force with respect to the different testing conditions. As can be noticed all conditions are compatible and depict a similar performance of the measurement system.

Ergometer testing

A typical result of this comparison is shown in Figure 3.6, which shows the tangential force measured by the ergometer ($\sigma = 4 - 2\text{N}$) and by the cantilever system ($\sigma = 0.15 - 0.05\text{N}$) along with its confidence interval. To provide an indicator of the overall quality of the encoder measurement the root mean square of the error was determined for each test condition, taking into account data recorded from the when the wheel starts to decelerate to its complete stop. Table 3.2 reports the average results for all the tests performed. Root mean square error of the tangential force measured by the ergometer, with respect to the force assessed by the cantilever wheel assumed as reference.

To provide an even more synthetic evaluation of the quality of the measurement of the tangential force by the ergometer, an ordinary linear regression between all the $\hat{F}_T$ values in every test and the same F_T values were computed, resulting in a regression standard error of $\sigma_0 = 1.9N$.

Tests with a high resistant torque and a low top speed display the highest values, probably due to the short time needed for the wheel to completely stop, resulting in less data points available for evaluation, and a less stable behaviour. For the same reason some tests were not considered valid, since not enough data were available for fitting to provide an estimate.

Discussion

The constructed device achieved its aim of being low-cost, using available components, and being portable while proving its use to determine the validity of a wheelchair roller ergometer. The current iteration is expensive due to the motor that was used. However, the motor was used because it was available at the time. The estimated cost of the system without this specific motor is therefore significantly lower at €400.

The combined uncertainty of the validation system was approximately ten times lower than the uncertainty of the ergometer, which is the basic criterion for a metrological validation. This allows to neglect the uncertainty contribution of the validation system in the comparison between the two systems. This requirement is obviously limiting the validation system applicability: if a better measurement of the tangential force were to be evaluated, a lower uncertainty would also be required by

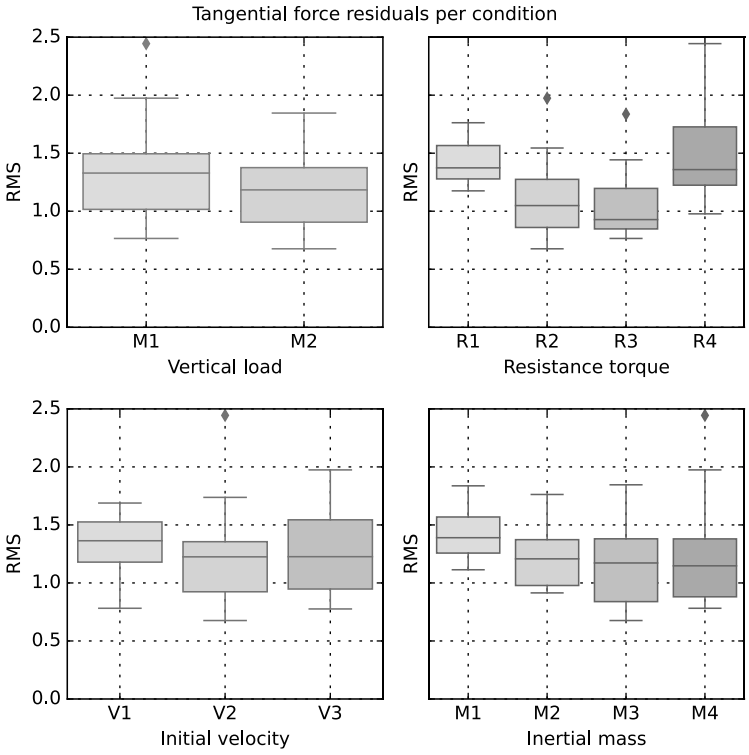


Figure 3.5. Sensitivity with respect to the variables investigated: vertical load, resistant torque, initial velocity, and inertial mass.

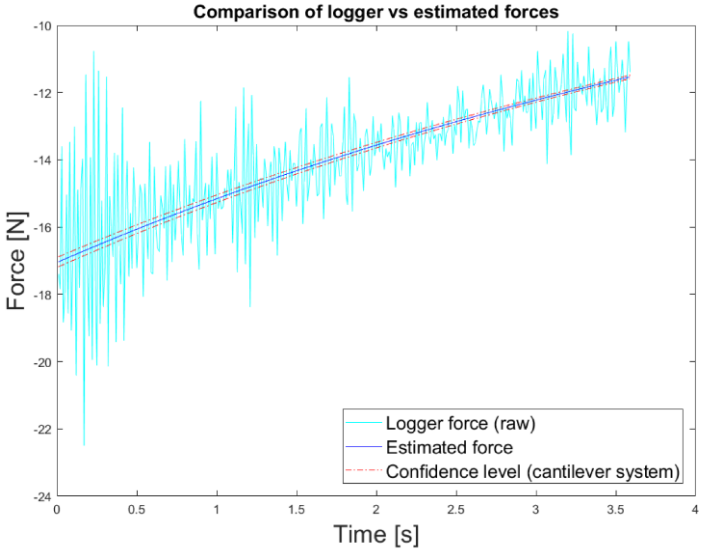


Figure 3.6. Example of a comparison between ergometer and validation system data at an initial velocity of 20km/h, no masses added and a resistant torque of 0.1Nm.

Table 3.2. Root mean square error of the tangential force measured by the ergometer with respect to the force assessed by the cantilever wheel assumed as reference.

Condition			**Resistant Torque [Nm]**			
IM	VL [N]	TS [km/h]	0.1	0.2	0.3	0.4
0	0	20	1.3	1.2	1.8	1.5
0	0	15	1.5	1.3	1.4	1.7
0	0	10	1.6	1.5	*	*
0	39	20	1.2	1.1	1.1	1.2
0	39	15	1.7	1.3	1.4	1.2
0	39	10	1.7	*	*	*
4	0	20	1.2	1.0	1.0	1.1
4	0	15	1.4	0.9	1.0	1.2
4	0	10	1.6	1.3	1.4	*
4	39	20	1.8	1.3	0.9	1.2
4	39	15	1.4	1.0	0.9	*
4	39	10	1.6	1.5	*	*
8	0	20	1.8	0.8	0.8	1.8
8	0	15	1.2	0.8	0.8	1.3
8	0	10	1.4	1.5	*	*
8	39	20	1.6	0.9	0.9	1.8
8	39	15	1.3	0.7	0.8	1.0
8	39	10	1.3	*	1.2	*
12	0	20	1.5	2.0	0.8	1.4
12	0	15	1.4	1.1	0.9	2.4
12	0	10	1.4	0.9	1.3	*
12	39	20	1.4	0.9	0.8	1.8
12	39	15	1.2	0.9	0.9	1.0
12	39	10	1.2	0.8	0.8	*

IM: Additional Inertial Masses; VL: Vertical Load; TS: Top Speed; *: the test was too short to perform an assessment (the wheel stopped too early)

the validation system. If, such in the case of the ergometer used in this work, the manufacturer introduces a filter to reduce the noise and improve results, then the validation system should be improved as well. Applying a low-pass filter to the

ergometer signal would be recommended in this case as the frequency content of wheelchair propulsion kinetics is less than 10Hz [21].

The freewheel mode of the ergometer was achieved by setting the motor to torque mode and applying zero torque. While it is assumed that the motor does not influence the system it is possible that the controller was not able to maintain exactly zero torque throughout the testing. While this could have influenced the determination of the added bearing and rolling friction, it is certainly more valid than disconnecting the motor entirely. Moreover, it is unlikely that any errors attributed to the inability to maintain zero-torque would change between the tested conditions.

Another limitation is the applied vertical load during the tests, which is lower than a wheelchair-user combination on the ergometer. However, the loading of the system did not appear to influence the uncertainty of the system itself. Moreover, increasing the applied vertical load would require a bigger and heavier validation system which would increase the cost and decrease portability. Systems that allow for heavier loading can be based on this system if cost and portability are deemed less important.

The main contributors to the combined uncertainty of the tangential force are the speed measurement, which has an uncertainty percentage contribution of 90% in slower conditions, and the inertia measurement, which reached a similar level at higher speeds. Hence, these parameters are opportunities to improve the measurement wheel. For example, the validation system could be improved by using a different encoder. The custom-made encoder was low-cost, but introduced variance due to eccentricities in the disk brake. A new disk or an industrial encoder could avoid irregularities in the speed data. The inertia of the wheel as determined by the trifilar pendulum had a relatively large uncertainty compared to its value (about 1% to 5% depending on the number of masses). Therefore, any improvements to the determination of that inertia could reduce the combined uncertainty of the validation system. This could be achieved by accelerometric measurements, employing torque meters or by replacing the wheel with a homogeneous disc.

The validation system can be used to validate roller ergometers. This can be done within one lab to monitor the consistency of a single roller ergometer over time, but this could also be done between labs as the system is portable and low-cost.

Conclusion

The validation system allows for the assessment of (wheelchair) roller ergometers within labs, but also between labs as it is both low-cost and portable. This is a

prerequisite for comparing or combining results obtained from different ergometers/institutes, which is especially relevant for the field of wheelchair propulsion where sample sizes are generally low. It can be used to assess the long-term stability of an ergometer by performing multiple measurements over the span of multiple weeks or months. Another possible application is pre-post testing, where an ergometer is assessed before and after an experiment to determine if the experiment was detrimental to the performance of the ergometer.

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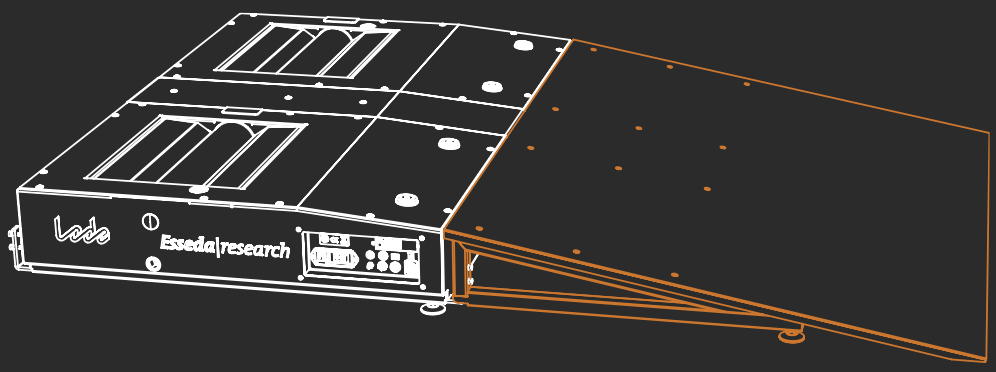
4

A guide to lab-based testing



Determining and Controlling External Power Output During Regular Handrim Wheelchair Propulsion

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Abstract

The use of a manual wheelchair is critical to 1% of the world's population. Human powered wheeled mobility research has considerably matured, which has led to improved research techniques becoming available over the last decades. To increase the understanding of wheeled mobility performance, monitoring, training, skill acquisition, and optimization of the wheelchair-user interface in rehabilitation, daily life, and sports, further standardization of measurement set-ups and analyses is required. A crucial stepping-stone is the accurate measurement and standardization of external power output (measured in Watts), which is pivotal for the interpretation and comparison of experiments aiming to improve rehabilitation practice, activities of daily living, and adapted sports. The different methodologies and advantages of accurate power output determination during overground, treadmill, and ergometer-based testing are presented and discussed in detail. Overground propulsion provides the most externally valid mode for testing, but standardization can be troublesome. Treadmill propulsion is mechanically similar to overground propulsion, but turning and accelerating is not possible. An ergometer is the most constrained and standardization is relatively easy. The goal is to stimulate good practice and standardization to facilitate the further development of theory and its application among research facilities and applied clinical and sports sciences around the world.

The video accompanying this article and supplementary materials are available on JoVE: https://www.jove.com/t/60492

Introduction

With an estimated 1% of the world's population being dependent on wheeled mobility today [1,2], a consistent flow of international research work increasingly emerges into international peer-reviewed journals in diverse fields such as rehabilitation [1,3], engineering [4], and sport sciences [5,6]. This leads to a growing knowledge base and understanding of the complexities of this common mode of human ambulation. Yet, for continual development and implementation in rehabilitation and adapted sport practices, there is a need for further international exchange and collaboration in research. Integral to such collaborative networks are improved standardization of experimental and measurement procedures and technology. Furthermore, consistent implementation of accurate monitoring of performance of the wheelchair-user combination in the laboratory and/or in the field is important for an optimal individual functioning and participation, while a healthy and active lifestyle is maintained over the individual's lifespan [7–9].

Experimentally, manual wheelchair propulsion during steady-state or peak exercise conditions [10,11] is often approached as cyclical upper body motion for the purposes of examining the wheelchair-user interface [12,13], musculoskeletal loading [14–16], and motor learning and skill acquisition [17,18]. The combined biomechanical and physiological notions of cyclic motions allow the use of the "Power balance", a modelling approach that was initially introduced by Van Ingen Schenau [19] for speed skating and cycling, and later introduced in manual wheeled mobility [8,20,21]. Figure 4.1 shows a power balance diagram for manual wheelchair propulsion. It converges from a selection of critical performance determining factors for the wheelchair-user combination and its three central components (the wheelchair, user, and their interface), at the left-hand side into the layout of (bio)mechanical and physiological power denominators and equations.

Power output is an important outcome parameter in the contexts of sports and daily life where peak power output can represent both increased performance in adapted sports or ease of functioning during activities in daily living [22]. Moreover, in combination with energy consumption it can be used to evaluate performance in terms of gross mechanical efficiency [17,18,23] (i.e., where a more skilled individual would require less internal energy to produce the same amount of external power output). From an experimental perspective, power output is a parameter that needs to be tightly controlled during a test, because changes in power output are of direct influence on all performance outcomes such as push time, recovery time [24], and mechanical efficiency [25]. Consequently, controlling and reporting power output is essential for all studies related to manual wheelchair propulsion.

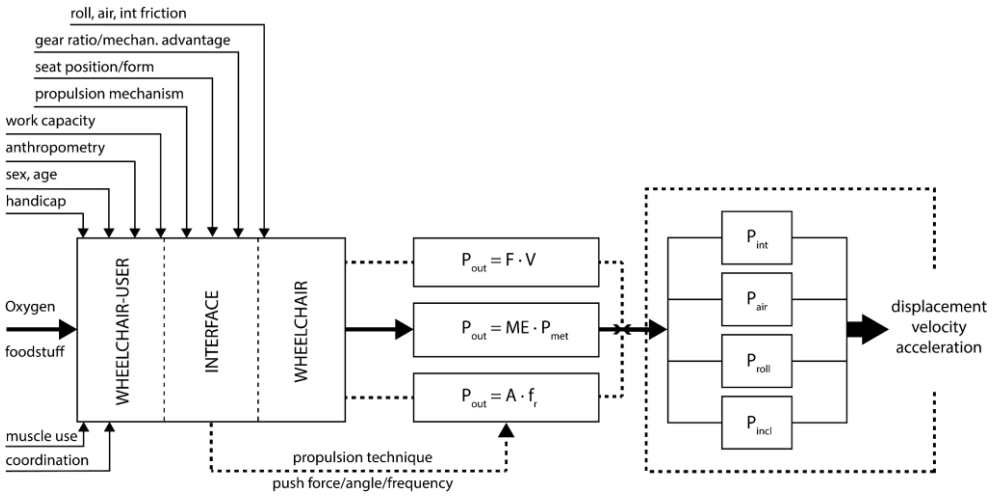

Figure 4.1. Power balance applied to manual wheelchair propulsion: P_{out}: external power output (W); ME: gross mechanical efficiency (%); F: mean resisting force; V: mean coasting velocity; A: work per push or cycle (J); f_r: frequency of pushes or cycles (1/s); P_{int}: internal losses (W); P_{air}: aerodynamic resistance (W); P_{roll}: rolling friction (W); P_{incl}: losses due to inclination (W). This figure is reprinted from van der Woude et al. [20].

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Overground testing is the gold standard in terms of validity (i.e., inertia, air friction, optical flow, and dynamic movement) [26], yet standardization of external power output, speed, and associated environmental conditions is much more difficult, and repeatability over time suffers. Overground wheelchair-related studies started in the 1960s [27,28] and focused on the physical strain of wheeled mobility. Although crucial in data interpretation and understanding [8,20], notions on external power output were limited to observation of the internal metabolic cost when performing different activities on different surfaces. Nowadays, measurement wheels can be used to measure power output [29,30] and coast-down tests [31,32] can be performed to infer the frictional losses during propulsion and thereby power output.

Different laboratory-based technologies were developed for wheelchair-specific exercise testing [33], ranging from a multitude of ergometers to differently sized and brands of treadmills. Treadmills are considered to be closest to overground testing in terms of validity [34] and have been used since the 1960s for wheelchair exercise testing [35,36]. Prior to testing, the slope and speed of the treadmill must be checked regularly. Even treadmills from the same brand and make may differ considerably and change in their behaviour over time [37]. For the determination of external power output, a drag test [20,36] is used for the individual wheelchair-user combination's total of rolling and internal drag force [38]. The force sensor for the drag test also has to be periodically calibrated. For the experimental

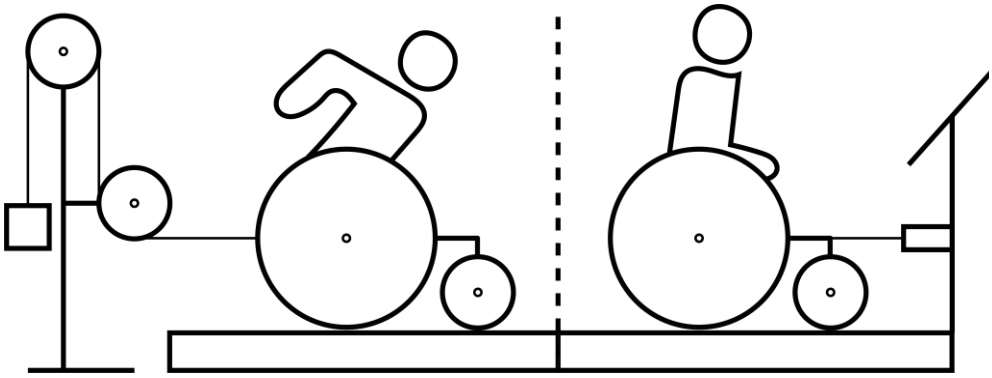

Figure 4.2. Treadmill setup. Left: Pulley setup to increase the external power output on a treadmill during propulsion. Right: Drag test setup to measure the frictional forces during treadmill wheelchair propulsion.

individualization of the protocol in terms of overall external load of wheeling over time and between subjects, a pulley system (Figure 4.2) has been designed as an alternative for the previous slope-dependent gradients of loading [36].

Another alternative for standardized wheelchair exercise testing has been the use of stationary ergometers [33], from simple off-the shelf ergometer solutions [39] towards highly specialized computer-based and instrumented ergometers [40]. Very few are commercially available. The enormous diversity in ergometer technology and mechanical characteristics introduces large unknown degrees of variability among the test outcomes [8]. Ergometers and wheelchairs need to be connected or inherently fused by design. Air friction is not present and perceived inertia is limited to the simulated inertia on the wheels, and movement experienced in the trunk, head, and arms during propulsion, while the wheelchair user is essentially stationary. The ergometer does allow for sprint or anaerobic testing as well as isometric testing, if the wheels can be adequately blocked.

A basic methodology for manual wheeled mobility research in lab-based studies is presented. Also, a brief outlook on field-based wheelchair research methodology and its potential outcomes is provided. The central focus is controlling and measuring external power output (W) in both field and laboratory-based experiments. The determination of internal power output through spirometry is also added, as this is often used to determine gross mechanical efficiency. Apart from the implementation of good practice, the goal is to produce discussions on experimental standardization and international information exchange. The current study will primarily address handrim wheelchair propulsion and the measurement thereof because it is the most prominent form of manually wheeled mobility in scientific

literature. However, notions discussed below are equally valid for other wheelchair propulsion mechanisms (e.g., levers, cranks [41]).

The current protocol describes the standardization and measurement of power output during overground, treadmill, and wheelchair ergometer-based testing during steady-state propulsion at 1.11 m/s. As an example, rolling friction will first be determined in overground testing with a coast-down test. Using this estimate of friction, power outputs will be set in the treadmill and ergometer tests using available protocols from the research literature. For the treadmill condition, friction will be determined with a drag test, and power output will be adjusted using a pulley system. For the ergometer condition a computer-controlled ergometer is used to match external power output with the overground test.

Protocol

1. Study design and setup

1.1. Instruct the participant and obtain informed consent in line with the Ethical Committee of the institution.

1.2. Determine the readiness for physical activity of the participants by performing a baseline assessment with the Physical Activity Readiness Questionnaire [42,43].

1.3. Perform in-patient screening with a medical doctor.

1.4. Decide on a fixed power output for all participants (e.g., 10–20 W at 1.11 m/s), a relative power output (e.g., 0.25 W/kg bodyweight at 1.11 m/s), or a "realistic" individual power output based on a surface of interest (based on a coast-down test).

1.5. Allow the participant to become familiar with the overground, treadmill, and ergometer conditions before testing.

1.6. Check the tire pressure and overall wheelchair mechanics before each measurement and inflate the tires to 600 kPa if needed.

NOTE: To obtain valid steady-state outcomes for cardiopulmonary upper body work and gross mechanical efficiency (ME), one should adhere to a minimum duration of 3 min per submaximal (up to 70% peak exercise capacity) exercise block to achieve steady-state exercise with a respiratory exchange ratio below 1 [44,45]. *Especially in handrim propulsion, the wheelchair speed must remain within a comfortable or feasible range (0.56–2.0 m/s) to exclude motor control issues* [46–48], *which means that power increments are preferably controlled by increments in resistance.*

2. External power output during overground testing

2.1. Perform the coast-down test on the surface of interest. Position the participant in an active position and as standardized as possible: feet on the footrest, hands on the lap, and looking straight ahead (the position should be reflective of the position during propulsion).

NOTE: Every movement changes the centre of mass, which changes the rolling resistance.

2.2. Accelerate the wheelchair to a high velocity.

NOTE: This can also be done by the participant.

2.3. Let the wheelchair decelerate to a complete standstill without interference.

2.4. Record the time and velocity data during the deceleration (e.g., with measurement wheels or inertial measurement units). See sections 2.4.1 and 2.4.2.

2.4.1. Record data using measurement wheels.

2.4.1.1. Replace the wheels of the wheelchair with a measurement wheel and the inertial dummy, preferably while the participant is not in the wheelchair.

NOTE: This example is for the OptiPush wheel. Other wheels might have different calibration requirements.

2.4.1.2. Turn on the measurement wheel using the on/off switch.

2.4.1.3. Turn on the laptop with the USB Bluetooth receiver and associated software.

2.4.1.4. Open the software on the computer.

2.4.1.5. Connect the wheel with the software by selecting the correct communication port (COM). If the correct COM port does not appear in the list, press refresh to update the list and try again. Press `Next`.

2.4.1.6. Fill in the required fields on the Client Setup screen. Press `Next`.

NOTE: Pay special attention to the Wheel Size and Wheel Side settings.

2.4.1.7. Gather offset data by pressing `Start` in the wheel setup and slowly spinning the wheel without touching the handrim until the red circle turns to green. Alternatively, press `Skip` to skip this step if the procedure has already been performed since the last wheel installation. Press `Next`.

2.4.1.8. To collect data press `Record` in the Data Collection screen. Resume the regular coast-down protocol from here on.

NOTE: Scripts for the analysis of measurement wheel data are available in the Supplementary Material 1.

2.4.2. Record the data using inertial measurement units (IMUs).

2.4.2.1. Attach the IMUs to the wheelchair: one on each wheel hub and one in the centre underneath the seat. Write down which IMU is attached where and in what orientation for later reference.

2.4.2.2. Turn on the IMUs and connect the IMUs to the computer using the NGIMU Synchronized Network Manager executable.

2.4.2.3. To collect data, go to `Tools`, then select `Data Logger`, and press `Start`. Resume the regular coast-down protocol from here on.

NOTE: Scripts for the analysis of IMU data are available in the Supplementary Material 2.

2.5. Repeat the coast-down procedure (2.1–2.4) and collect back-and-forth data to reduce the influence of uneven surfaces.

2.6. Open the `coast_down_test` software (Figure 4.3) on a computer. Press `Import data` to import the coast-down data file (measurement wheel or IMU). Select coast-down sections in the data by using the slider in the graph on the right and pressing `Grab selection`.

2.7. Set participant and wheelchair weight in the `Settings` section. Press `Calculate outcomes`. Write down the mean rolling friction (N) and rolling friction coefficient. Press `Export` to save all (meta)data for later reference.

NOTE: When constant friction cannot be assumed due to air drag (i.e., in most sports environments) the analysis becomes a little more complex. The protocol is identical, but the initial speed should probably be higher. In this case, a non-linear differential equation needs to be solved [49] and that equation needs to be fit with a curve fitter (e.g. Levenberg-Marquardt).

$$v_t = \sqrt{\frac{c_2}{c_1}} \tan\left[tan^{-1}\left(v_0 \sqrt{\frac{c_1}{c_2}} - t\frac{\sqrt{c_1 c_2}}{M} \right) \right] \quad (1)$$

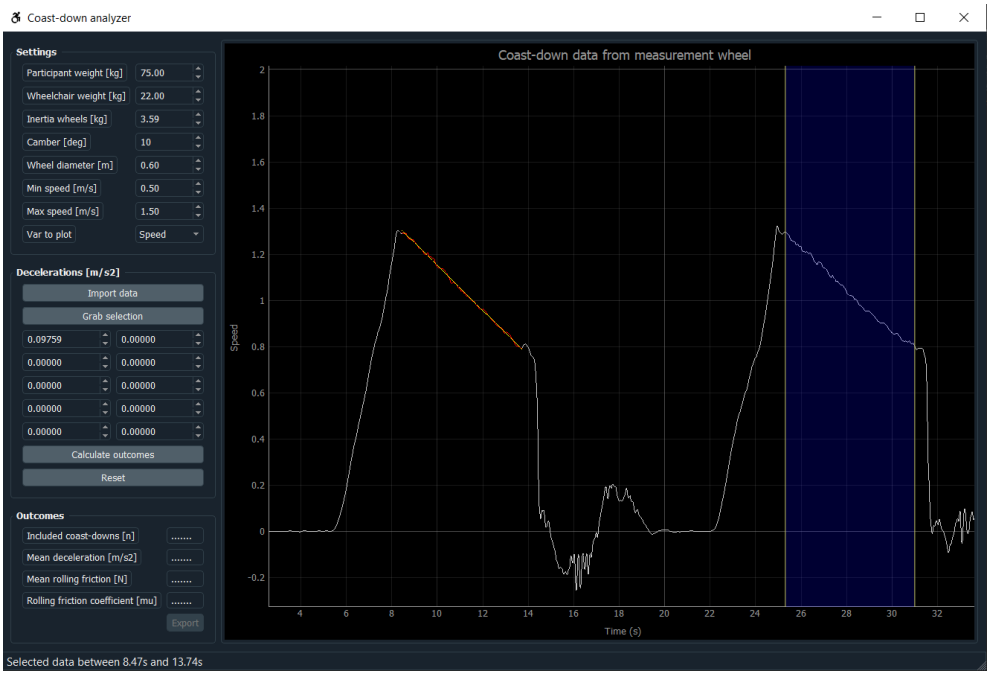

Figure 5.3. Coast down analyser: A Graphical User Interface to select coast-down sections in data and determine the friction based on measurement wheel or Inertial Measurement Unit data.

In this equation v_t is the instantaneous velocity and v_0 is the initial velocity at the start of deceleration. c_1 reflects the velocity dependent friction and c_2 reflects the velocity independent friction (mgμ). Scripts for the analysis of coast-down tests are available in the Supplementary Material 3 and the graphical user interface (GUI) for the analysis of coast-down tests used in step 2.7 is available in the Supplementary Material 4.

3. External power output during treadmill testing

3.1. Treadmill characterizationMeasure the belt velocity of the loaded treadmill with a calibrated tachometer to determine what treadmill setting needs to be used (e.g., for 1.11 m/s the treadmill needs to be set to 4.1 km/h on the display instead of 4.0 km/h).

NOTE: Alternatively, determine the belt velocity by measuring the belt length and counting ten rotations while recording the time with a stopwatch/video camera.

3.1.2. Measure the treadmill angles using an angle sensor. Check for consistency by repeating the measurements and check for hysteresis by repeating the measurements in descending order.

NOTE: Check treadmill velocity with a tachometer and angles with an angle sensor during every measurement if their reliability is low.

3.2. Drag test: calibration

3.2.1. Turn on the power supply of the drag test force sensor at least 30 min before calibration.

3.2.2. Suspend the force transducer vertically and align with a self-levelling laser or angle sensor.

3.2.3. Set up the drag test computer and connect the force sensor with the computer. Open the ADA3 software on the drag test computer and press `Calibrate force sensor`.

3.2.4. Attach known (calibrated) weights (1–10 kg with 1 kg increments) to the sensor and register the digital values.

3.2.5. Fit a linear regression equation to determine the relation between applied load and measured voltage by continuing in the ADA3 software.

3.2.6. Repeat (3.2.1–3.2.5) if the root-mean-square error (RMSE) exceeds 0.13 N [37].

3.3. Performing a drag test

3.3.1. Turn on the power supply at least 30 min before measuring.

3.3.2. Set up the drag test computer and connect the force sensor with the computer. Open the ADA3 software on the drag test computer and press `Power table measurements`.

3.3.3. Place the wheelchair-user combination on the treadmill. Position the participant in an active position and as standardized as possible: feet on the footrest, hands on the lap, and looking straight ahead (the position should be reflective of the position during propulsion). Instruct the participant to maintain the same position throughout the test.

3.3.4. Measure the offset of the load cell by recording force with no rope attached using the ADA3 software. Press `OK`.

3.3.5. Connect the wheelchair to the force transducer with a lightweight rope. Ensure the load cell and rope are horizontally aligned with the rear wheel axle of the wheelchair.

3.3.6. Accelerate the belt to the desired speed, in this case 1.11 m/s (4.1 km/h on display).

3.3.7. Increase the inclination of the treadmill, wait until the position of the treadmill and wheelchair-user combination is stable, and record the force and angle. Repeat for 10 increasingly steep angles (1.5–6% in 0.5% increments).

3.3.8. Fit a linear regression using the angle and force using the ADA3 software by clicking **Next**. Calculate the force at the zero-angle of the treadmill.

NOTE: The intercept of the regression equation cannot be used if the treadmill angle has an offset.

3.3.9. Repeat the drag test (3.3.3–3.3.8) if the RMSE of the linear regression line exceeds 0.5 N [37].

3.4. Setting power output on a treadmill

3.4.1. Calculate the desired power output and determine the testing speed.

NOTE: For the current protocol, this is equal to the results obtained in step 2.7.

3.4.2. Calculate the required pulley weight by subtracting the drag test friction (from step 3.3.8) from the target friction (from step 2.7).

3.4.3. Position the pulley in front or behind the treadmill and make sure it is centred. Attach the pulley to the wheelchair and make sure the rope is level. Instruct the participant that the weight in the pulley might move the wheelchair.

3.4.4. Attach the weight (usually between 0–1 kg) to the pulley system using a basket of known low mass and a carabiner. Slowly increase the weight if necessary, until the desired power output is achieved.

NOTE: Alternatively change the power output by changing the angle of the treadmill based on the power table from a drag test.

4. External power output during ergometer-based testing

4.1. Turn on the ergometer at least 30 min before measuring. Start the associated software on the computer.

4.2. Press the **Participant** widget, then press **Add…**. Give the participant an ID and enter the body weight of the participant. Press **OK**.

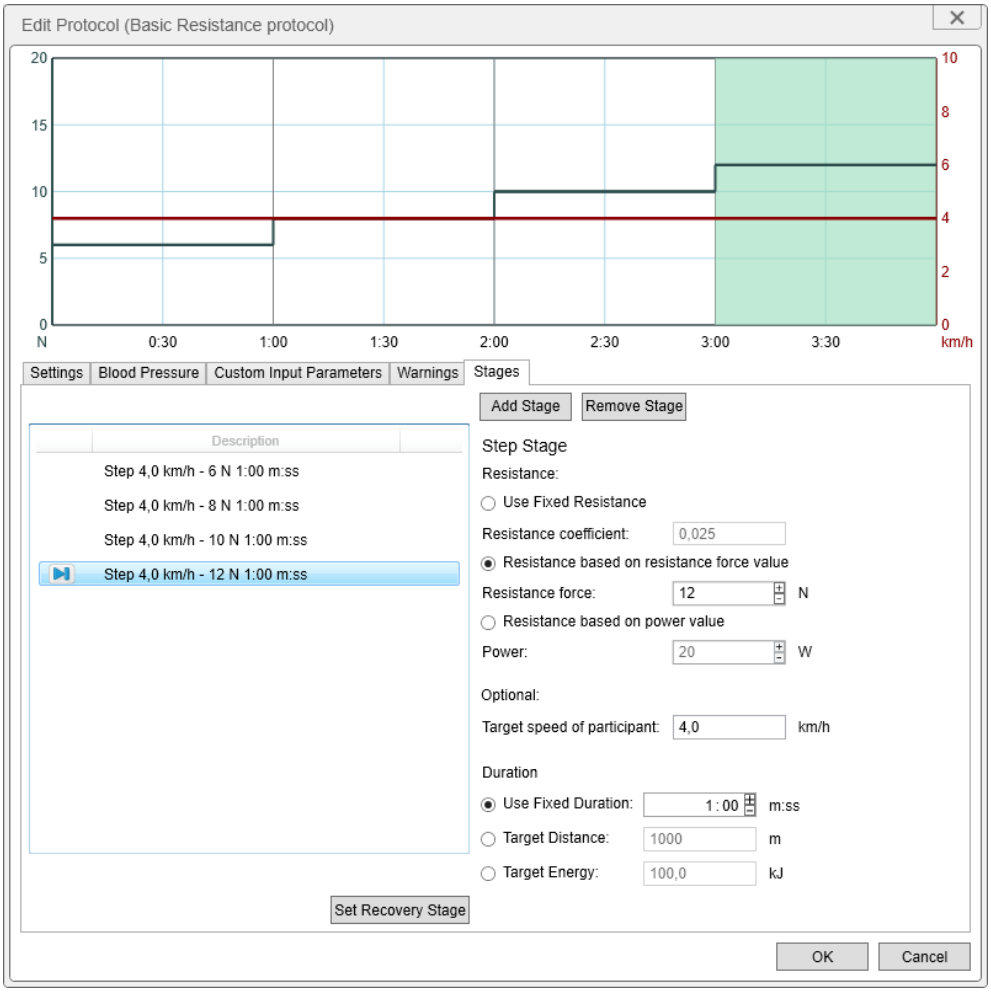

Figure 4.4. Protocol settings window for the wheelchair ergometer: Power output can be set by choosing a power output and a target speed or a rolling friction and a target speed.

4.3. Press the wheelchair icon on the Device menu. Fill in the wheelchair specifications in the form. Press `OK`.

NOTE: The body weight variable is important, because it will influence the simulation provided by the ergometer.

4.4. Press the protocol widget. Create a custom protocol by selecting **`Add…`**. Select **`Custom Protocol`** and press **`Next`**. Give the protocol an appropriate name and press **`Create`**.

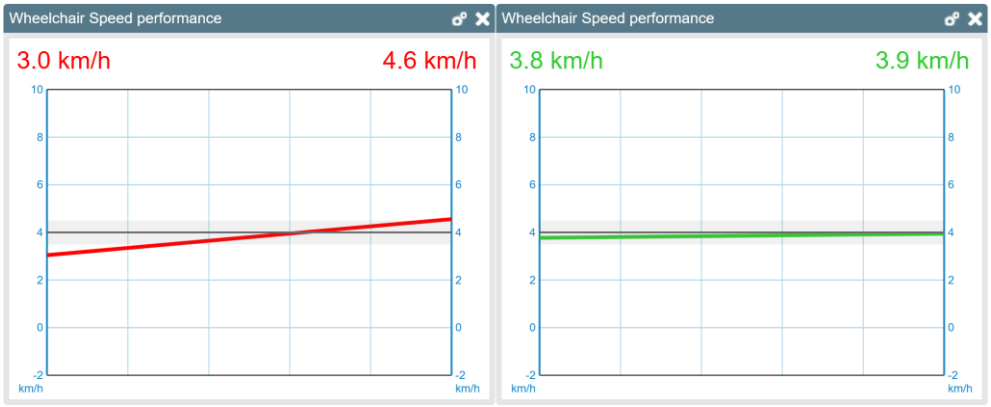

Figure 4.5. Feedback on the wheelchair ergometer in the form of a line plot: Left and right roller speeds are plotted. Participants should try to keep a steady speed while going in a straight line (by keeping the on-screen line horizontal). Speed data is smoothed with a sliding window that can be changed in the settings.

4.5. Select **Stages** and click **Add Stage** and **Resistance**. Set the resistance to the friction coefficient obtained with the coast-down test in section 2. Set the target speed at 4 km/h and press **OK** (Figure 4.4).

4.6. Set up the participant screen. Remove all widgets from the screen. Click **Add widget** and select the **Wheelchair direction** widget and drag it into the screen (Figure 4.5).

4.7. Align the wheelchair on the rollers using the alignment system. Fasten the wheelchair using the four-belt system. Check that the wheels are not touching the ergometer and are properly aligned.

4.8. Position the participant in an active position and as standardized as possible: feet on the footrest, hands on the lap, and looking straight ahead (the position should be reflective of the position during propulsion). Instruct the participant to maintain the same position throughout the test.

4.9. Calibrate the ergometer with the associated software by pressing the crosshair button in the Device menu.

NOTE: Scripts for the analysis of ergometer data are available in the Supplementary Material 5.

5. Internal estimates of power output

5.1. Turn on the spirometer for at least 45 min before any calibrations or testing.

5.2. Calibrate the spirometer according to the factory guidelines using the associated software, including calibrations for turbine, reference gas, room air, and delay.

NOTE: Room air and reference gas calibrations should be performed before each test.

5.2.1. Perform a turbine calibration.

5.2.1.1. Press `Turbine` in the calibration menu. Connect the turbine with an optoelectronic reader to the spirometer. Connect the calibration syringe with a known volume to the turbine.

5.2.1.2. When the unit is ready, perform six controlled and complete strokes with the piston. Press the `Exit` icon.

5.2.2. Perform a reference gas calibration.

5.2.2.1. Press `Reference gas` in the calibration menu. Connect the pressure regulator to the calibration cylinder with a known concentration of mixed gas.

NOTE: The cylinder must be open, but the pressure regulator must be closed.

5.2.2.2. Connect the sampling line to the sampling connector of the spirometer and leave the other end disconnected. Let the spirometer flush the analysers. Make sure that the sampling line is far from any exhaled gas.

5.2.2.3. When prompted by the spirometer, connect the free end of the sampling line to the pressure regulator on the calibration cylinder and open the regulator. Exit with the `Exit` icon after the calibration ends.

5.2.3. Perform a room air calibration.

5.2.3.1. Connect the sampling line to the sampling connector on the spirometer and leave the other end free. Exit with the `Exit` icon after the calibration ends.

5.2.4. Perform a delay calibration.

5.2.4.1. Connect the turbine to the optoelectronic reader and connect the sampling tube. Ensure that both are connected to the spirometer.

5.2.4.2. Synchronize breathing with the acoustic signal. This can be performed by the operator.

NOTE: This procedure needs to be repeated every time a sampling tube is changed. Clean or switch the mask used for this procedure before giving it to the participant.

5.2.4.3. Exit with the `Exit` icon after the calibration ends.

5.3. Put the spirometer mask on the participant. Adjust the elastic bands on the head cap to create a tight seal around the face of the subject.

NOTE: Optionally connect a heart-rate monitor to the spirometer and let the participant wear the heart-rate belt.

5.4. Fixate the hose of the spirometer so it does not interfere with movement.

5.5. Enter a `New subject` on the display of the spirometer.

5.6. Start the recording by pressing the test icon on the spirometer. For submaximal exercise testing choose the breath-by-breath mode. To start recording press the record key on the spirometer.

NOTE: Scripts for the analysis of spirometer data are available in the Supplementary Material 6.

6. Testing procedure

6.1. Instruct the participant to perform 4 min of steady-state exercise at the desired speed (1.11 m/s).

6.1.1. Instruct the participant to use velocity feedback to stay (on average) at the desired speed.

NOTE: Speed can be shown from the measurement wheel or the IMUs from their respective laptops in the overground condition. The laptops have hook-and-loop straps that allow for fixation on the legs.

6.1.2. Instruct the participant to stay (on average) in the centre of the treadmill for the treadmill condition.

6.1.3. Instruct the participant to look at the velocity and heading feedback on the ergometer screen in the ergometer condition and keep it (on average) within the target range.

6.2. Start a stopwatch and the spirometer (step 5.6) at the same time.

NOTE: This can be done without an external trigger as the timing difference from pressing start is negligible while using breath-by-breath spirometry.

6.3. After 30 s, start the wheelchair propulsion.

NOTE: For the treadmill and ergometer conditions this implies starting the treadmill or ergometer. When using a measurement wheel (step 2.4.1.8) or IMUs (step 2.4.2.3) start those as well.

6.3.1. Use the `lap` button in the overground condition to mark the corners of the track.

6.4. After another 4 min during the test, without prior notice, instruct the participant to stop pushing the wheelchair.

NOTE: In the treadmill condition a few additional pushes are needed before the belt stops.

Representative results

Using the aforementioned procedure, power output was determined for 17 familiarized (two 30 min sessions of practice) able-bodied participants with an overground back-and-forth coast-down test (mean of five trials). The coast-down profile was characterized with a measurement wheel in a smooth hospital hallway. Afterwards, participants were measured during overground (25.0 x 9.0 m circuit), treadmill (2.0 x 1.2 m), and ergometer wheelchair propulsion. The power output in the treadmill and ergometer modalities were matched with the overground condition using the protocols described in this paper.

Power output was obtained from the same measurement wheel during three blocks of 4 min of wheelchair propulsion after a familiarization block of equal length. Only the last minute of each block was used for analysis, assuming steady-state propulsion. For the overground propulsion data only the long straights (25 m) were used. All data (pre-)processing was performed in Python 3.7 (Python Software Foundation). ICC estimates and their 95% confidence intervals were calculated in R 3.3.4 (R Core Team), using a single-rating, absolute-agreement, random-effects model.

The mean combined weight of the wheelchair-user system was 92.6 kg (± 8.3). The mean expected power output from the coast-down test was 9.7 W (± 1.6). Power output as calculated from the measurement wheel was lower for overground 8.1 W (± 1.4), treadmill 7.8 W (± 1.9), and ergometer 8.7 W (± 2.2) wheelchair propulsion. The average difference between target power output and measured power output were -1.6 (± 1.6), -1.8 (± 1.4), -1.0 (± 1.0) W for overground, treadmill, and ergometer propulsion, respectively. These results are also shown in Table 4.1, Figure 4.6, and Figure 4.7.

Power output for overground propulsion showed a poor-to-moderate (ICC: 0.38, CI: 0.00–0.73) agreement with the target output. In contrast, treadmill propulsion

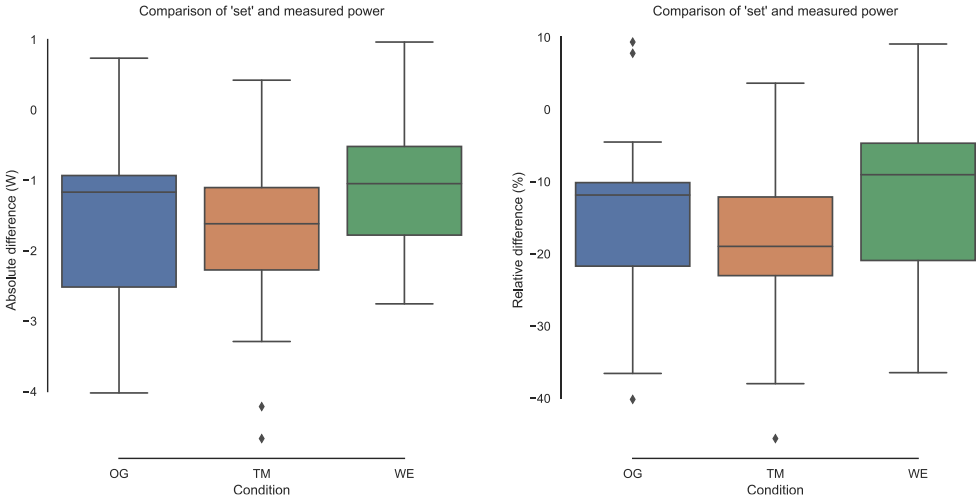

Figure 4.6. Relative and absolute difference distributions between coast-down friction and measured power output during overground (OG), treadmill (TM), and ergometer (WE) wheelchair propulsion: The whiskers show 1.5x the interquartile range.

showed poor-to-good (ICC: 0.45, CI: 0.00–0.79) agreement and ergometer propulsion showed poor-to-excellent (ICC: 0.77, CI: 0.11–0.93) agreement. Absolute error was negatively correlated with power output for propulsion on the ergometer (r=-0.55, p=0.02), but not for the other two conditions (overground: r=0.47, p=0.06; treadmill: r=0.22, p=0.40).

Agreement between conditions was poor-to-moderate (ICC: 0.49, CI: 0.20–0.74). Within-modality (between the three 4 min blocks) reliability was good-to-excellent for overground (ICC: 0.91, CI: 0.82–0.97) and treadmill (ICC: 0.97, CI: 0.93–0.99) propulsion and moderate-to-excellent for ergometer propulsion (ICC: 0.97, CI: 0.71–0.99). The ergometer appeared to perform worse over time, which was confirmed by a repeated-measures ANOVA ($F(2, 32)=64.7$, $p<0.01$), but there was no time effect for overground ($F(2, 32)=0.9$, $p=0.418$) and treadmill ($F(2, 32)=0.9$, $p=0.402$) propulsion.

Table 4.1. Comparison of set power output and actual power output measured with a measurement wheel.

	Value two-sided (W)[2]	**Difference with target**	**Difference with target (%)**	**Difference with target (abs)**	**Agreement with target PO (ICC)[3]**
Target PO[1]	9.68(±1.57)	n.a	n.a	n.a.	n.a
Overground PO	8.12(±1.41)	-1.56(±1.57)	-15.30(±13.70)	1.72(±1.57)	0.38 (0.00-0.73)*
Treadmill PO	7.84(±1.92)	-1.84(±1.38)	-18.98(±13.42)	1.91(±1.16)	0.45 (0.00-0.79)*
Ergometer PO	8.65(±2.24)	-1.02(±0.97)	-11.82(±11.94)	1.16(±0.78)	0.77 (0.11-0.93)*

1. Calculated from coast-down friction. 2. Determined with measurement wheel. 3. Two-way, absolute agreement, fixed raters with 95% confidence intervals. * p<0.001

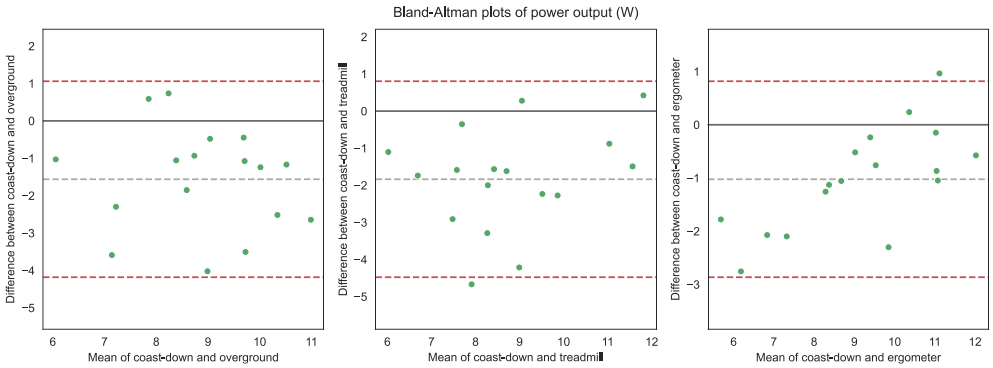

Figure 4.7. Bland-Altman plot for coast-down friction and measured power output during overground (left), treadmill (middle), and ergometer (right) wheelchair propulsion. The dark grey dotted lines indicate the pooled mean for a combination and the red dotted lines are the mean ± 1.96 standard deviations.

Discussion

In the previous sections an accessible methodology for determining and standardizing power output for different laboratory-based modalities was presented. Additionally, a comparison between set power output and measured power output during steady-state propulsion was made. While systematic error was present as well as some variability, the tools presented are better than the alternative: not standardizing at all. These results are similar to another study that reported measured power output and set power output [50]. Moreover, agreement between conditions was poor-to-moderate, indicating that extra attention should be paid when comparing studies using different modalities. As expected, the ergometer condition presented the easiest environment to standardize from the perspective of the operator. The ergometer performed better in the high friction settings. The blocks (3 x 4 min) within one modality showed good-to-excellent and moderate-to-excellent agreement. Interestingly, the ergometer performed worse over time, possibly due to sensor drift. Therefore, it might be prudent to recalibrate the ergometer between every block. Note that these results are for low-intensity steady-state exercise and could differ for different protocols.

Minor mechanical or ergonomic changes in the wheelchair-user combination can have a large impact on experimental outcomes [12,51]. Material maintenance and a full awareness of vehicle mechanical principles are essential for performance outcomes and the validity of the experiment. The vehicle mechanics (e.g., mass, wheel sizes, tire type and pressure, alignment) and fit (e.g., fore-aft position, centre of mass, mass, frontal plane) of wheelchair-user combination will determine rolling and air drag in combination with environmental conditions. The mass and the

orientation of the centre of mass will affect rolling drag with respect to the larger rear wheels and the smaller castor wheels in front. A summary of factors influencing rolling friction is presented in [8]. Moreover, the wheelchair is often individualized. Apart from the intervention conditions (e.g., vehicle mechanics or interface) at each test, the wheelchair conditions must also be constant and its vehicle mechanics, including frame, seat, and tires should be checked. The tires need to be at a fixed pressure over tests and among individuals. Important checkpoints [52] are possible friction points, rear wheel position, and potential changes in wheel alignment [36,53–55].

Overground testing also requires ambulant technology for each of the indicators for cardiopulmonary strain, kinematics, or kinetics outcomes. This can be met, but the practicality of complex measurements is limited in a non-research environment. Coast-down tests are specific for the individual wheelchair-user combination and rolling surface. However, they are static, so they might not capture all the characteristics of the wheelchair-user combination [56]. They are especially sensitive to changes in the centre of mass, which might explain the small differences between the coast-down test and the measured overground power output. These limitations are also found in the drag test and ergometer calibration, which also assume a static position of the wheelchair user.

The drag test measures the resisting forces of rolling and internal drag of each individual wheelchair-user combination. It is clearly sensitive to vehicle mechanics of the wheelchair, but also position and body orientation of the user. A standardized procedure is essential [20,36], where at a constant belt speed, the user-wheelchair combination is pulled over the belt being connected to a unidimensional calibrated force transducer on the frame of the treadmill at a series of slope angles (Figure 4.2). A treadmill adaptor for load cells that can be adjusted to the height of the centre axis of the wheelchair is required. Using linear regression analysis provides a static estimate of the mean drag force on the treadmill belt at zero inclination for a given wheelchair-user combination, which provides the mean external power output with the product of belt speed and drag force. The drag test is robust with regards to small differences in the execution of the test by different operators (e.g., position of the rope) [37].

Although sometimes assumed an apparently simple test, each of the testing elements of the drag test requires understanding of the underlying theory and training on all the details of the procedures [8]. Similar to the coast-down test, this test is especially sensitive to changes in centre of mass. Moreover, the behaviour and sensitivity of the strain gauge-based force transducers, their consistent calibration

(i.e., precision of calibration weights, sequence of mounting) [20,36,37], as well as any of the procedures of the drag test that are sensitive to changes in speed or inclination angle of the treadmill all have to be considered. This means that the treadmill itself needs to be checked and calibrated as well [37]. Consistent awareness of such noise generating phenomena must be tracked and executed in day-to-day experimentation.

Precision of power output-based simulations and their outcomes are fully dependent on the standardization, practice, and training of those who conduct the experiments. Diversity of treadmills, ergometers, or any other electronically motor driven device can be an issue, as shown by De Groot et al. [51]. In exchange of population-based data, one should be aware of the potential role of such differences on the test outcomes. In any wheelchair experiment, a proper explanation of the testing conditions and open presentation of the actual values for speed, resistance, and power output should be presented for any subgroup or measurement condition.

In wheelchair experimentation, heterogeneity of the test sample is hard to escape from when focusing on the actual wheelchair users. Among those, people with a spinal cord injury are most frequently subject to research, because they tend to have a stable spinal cord lesion for the rest of their lives. Lesion level, completeness, gender, age, talent, and training status determine the heterogeneity of such study groups [57]. Increasing the number of participants through multicentre collaboration is an important way to circumvent this and increase the power of experimentation [57], even in the early stages of rehabilitation [10]. This paper is hopefully a stepping-stone to a broad discussion on wheelchair experimenting in rehabilitation and adapted sports communities that hopefully leads to international collaboration and knowledge exchange through the existing and new networks of researchers. Availability of adequate testing infrastructure allows consistent monitoring and evaluation of progress in clinical rehabilitation, adapted sports, and beyond.

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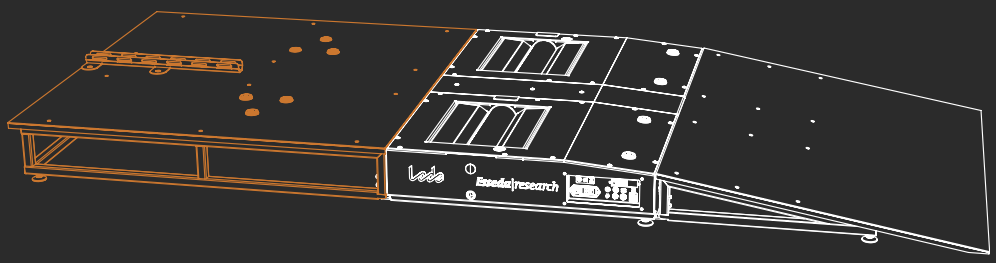
5

A comparison of wheelchair modalities



Physiological and biomechanical comparison of overground, treadmill, and ergometer handrim wheelchair propulsion in able-bodied subjects under standardized conditions

R. de Klerk, V. Velhorst, H.E.J. Veeger, L.H.V. van der Woude, R.J.K. Vegter - Journal of NeuroEngineering and Rehabilitation

Abstract

Handrim wheelchair propulsion is often assessed in the laboratory on treadmills (TM) or ergometers (WE), under the assumption that they relate to regular overground (OG) propulsion. However, little is known about the agreement of data obtained from TM, WE, and OG propulsion under standardized conditions. The current study aimed to standardize velocity and power output among these three modalities to consequently compare obtained physiological and biomechanical outcome parameters obtained. Seventeen able-bodied participants performed two submaximal practice sessions before taking part in a measurement session consisting of 3x4min of submaximal wheelchair propulsion in each of the different modalities. Power output and speed for TM and WE propulsion were matched with OG propulsion, making them (mechanically) as equal as possible. Physiological data and propulsion kinetics were recorded with a spirometer and a 3D measurement wheel, respectively. Agreement among conditions was moderate to good for most outcome variables. However, heart rate was significantly higher in OG propulsion than in the TM condition. Push time and contact angle were smaller and fraction of effective force was higher on the WE when compared to OG/TM propulsion. Participants used a larger cycle time and more negative work per cycle in the OG condition. A continuous analysis using statistical parametric mapping showed a lower torque profile in the start of the push phase for TM propulsion versus OG/WE propulsion. Total force was higher during the start of the push phase for the OG conditions when compared to TM/WE propulsion. Physiological and biomechanical outcomes in general are similar, but possible differences between modalities exist, even after controlling for power output using conventional techniques. Further efforts towards increasing the ecological validity of lab-based equipment is advised and the possible impact of these differences -if at all- in (clinical) practice should be evaluated.

Introduction

The repetitive and relatively high loads on the upper-extremities during handrim wheelchair propulsion, associated with an increased risk of pain and pathology [1–3], are a continued concern addressed in wheelchair research [4,5]. Ideally, research would assess the user during overground testing in the environment they are daily exposed to, as this has the highest ecological validity [5]. During overground propulsion the power output necessary at a certain velocity is dependent on a number of uncontrollable factors such as floor-type, slope, cross-slope and air resistance, besides individual factors of the wheelchair-combination as a whole like weight, frontal area tire-pressure and internal frictional losses. Moreover, there are effects of optical flow, and additional requirements such as braking and cornering [6]. Therefore, experimental conditions overground are difficult to control and it can be challenging to consistently collect enough consecutive push cycles without a sufficiently spacious laboratory environment [7].

Various other options for conducting studies on wheelchair propulsion exist, such as, motorized treadmills or wheelchair ergometers with each having their own advantages and disadvantages [8]. The advantage of these lab-based systems, in general, is the better standardization and the ability to collect multiple subsequent push cycles, increasing data reliability [9]. However, stationary systems offer no visual flow or meaningful context which reduces task complexity and might confound data obtained from these methods. In fact, ergometers mostly remove the need for steering and balancing as task elements in handrim wheelchair propulsion, making it the most abstract measurement modality [6].

Research in gait has shown that, while treadmills are mechanically valid [10], differences between overground and treadmill modalities exist [11–19], while others have argued that differences are minimal [14,16,20]. Similar studies for wheelchair propulsion, however, are lower in number and have also yielded mixed results [6,21–25]. Stephens and Ensberg [25] found that biomechanical outcome variables for overground propulsion and treadmill propulsion were significantly different. Moreover, Chénier et al. found that wheelchair users perceive speed differently on treadmills compared to overground propulsion [23]. However, in different studies Kwarciak et al. [24] and Mason et al. [21] found that physiological and biomechanical parameters in treadmill propulsion highly correlate with overground propulsion at specific treadmill settings. Koontz et al. [22] also found correlations between overground and ergometer wheelchair propulsion kinetics ranging from poor-good depending on the outcome parameter.

A possible explanation of these mixed results could (in part) be the lack of standardization for power output and/or speed in those experiments. As of yet, there are no studies that compared overground, treadmill, and ergometer wheelchair propulsion when power output and speed are matched, even though the required methodology has been well described and adopted in the research literature [5,26] and is available to most labs [4]. Spatiotemporal variables are known to be dependent on factors influencing power output such as speed and slope [27]. Finally, only qualitative [22] and discrete quantitative [21,22,24] comparisons have been made, whereas continuous analysis of handrim biomechanics might also yield useful information as the biomechanical context is immediately apparent [28].

The goal of this study is therefore to compare the physiological, spatiotemporal, and kinetic characteristics of wheelchair propulsion between overground, treadmill, and ergometer handrim wheelchair propulsion while controlling for power output and speed using available standardization methods [26]. Results from this study can be used to better translate research to the field or to improve existing testing protocols.

Methods

Participants

A convenience sample of seventeen able-bodied subjects, age(21.6±2.4yrs), mass(69.6±8.2kg), height(1.74±0.07m), sex(4M/13F), handedness(16R/1L), volunteered in the study. Participants had no previous experience in wheelchair propulsion and did not have any contraindications for exercise (Par-Q, [29]). Written informed consent was obtained before participants were enrolled. The study was approved by the local ethics committee of the Centre for Human Movement Sciences Groningen, University Medical Centre Groningen, The Netherlands.

Protocol

The current study employed a within-subjects design with three sessions (two training sessions and one measurement session on separate days) in which participants did a trial of overground, treadmill, and ergometer handrim wheelchair propulsion in a randomized order (Figure 5.1). Two training sessions at a fixed speed of 1.11m/s were performed to familiarize participants with wheelchair propulsion in general [30,31] and with the specific testing modalities [20]. Before the measurement session, a coast-down test [32] was performed to determine the static rolling resistance overground which was used to standardize power output among conditions. During the measurement session, the first block was used to familiarize the participant with the modality again. Thereafter, three blocks of four minutes of steady state propulsion were performed at 1.11m/s in each of the three testing modalities.

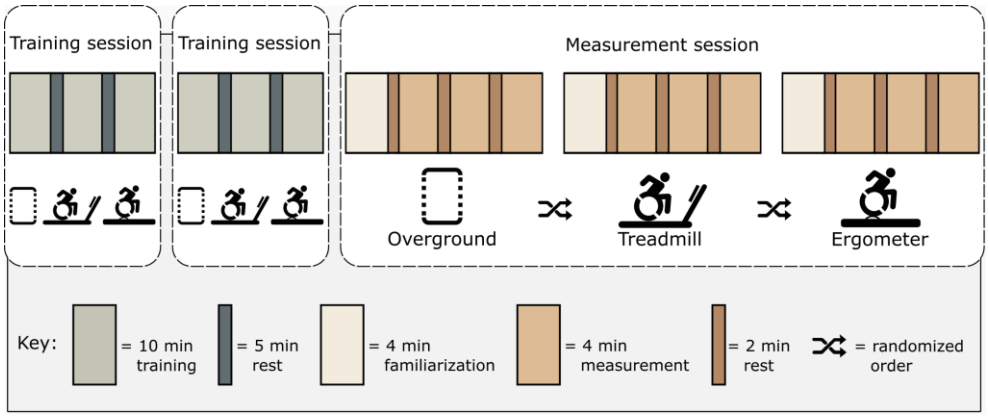


Figure 5.1. Research design: two training sessions were followed by a longer measurement session. Each session was independent and on a different day. During the measurements, participants performed three blocks of wheelchair propulsion per modality (overground, treadmill, and ergometer) at 1.11m/s. The last minute of data were used for analysis.

All tests were performed in the same non-individualized Küschall k-series wheelchair with instrumented wheels. The right wheel was replaced with a force sensing Optipush Biofeedback System (MAX Mobility, USA, [33]) and the left wheel was replaced with an inertial dummy wheel (total weight= 22kg). Tires were inflated before every session to 6 bars (600kPa). Oxygen consumption was measured with a mobile breath-by-breath spirometer (K5, Cosmed, Italy). Heart rate was determined with a heart rate monitor (Garmin, USA) connected to the same system.

Standardization of power output

To ensure all conditions were (at least mechanically) as equal as possible, the power output and speed were standardized for all subjects [26]. The overground condition served as a reference for the treadmill and ergometer condition. Frictional forces during overground propulsion were estimated with a back-and-forth coast-down test on the same surface as the overground condition, using velocity data from the measurement wheel [32]. Friction was assumed to be constant and independent of velocity. Treadmill rolling friction was individually determined with a drag test and matched with the friction obtained from the coast-down tests using a pulley system [5]. Ergometer power output was set in the ergometer software by changing the rolling friction coefficient in the associated software [34] for each individual wheelchair-user combination.

Overground propulsion

The participants propelled the wheelchair in an empty rectangular hospital hallway (long straights: 25m, short straights: 9m, width: 2.5m) with a smooth linoleum floor (Figure 5.2). All practice and measurement sessions were performed in clockwise

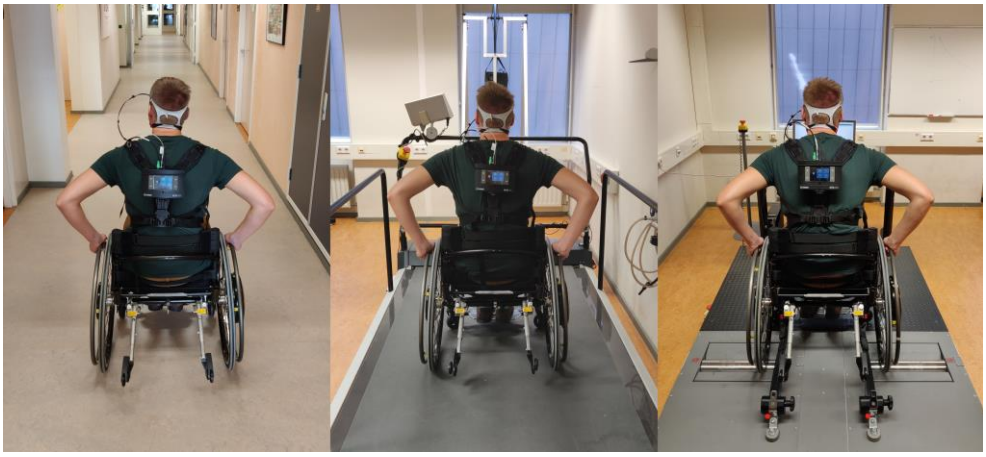


Figure 5.2. Overview of the three different conditions: overground, treadmill, and ergometer wheelchair propulsion.

laps. The subjects were instructed to propel the wheelchair at a constant speed using feedback from the measurement wheel. They received visual feedback that was updated each push cycle from the measurement wheel on a laptop screen attached to their lap with hook-and-loop fasteners. No adjustments to rolling resistance were made during the overground blocks. A timer was used to mark the long sections of the course as only the straight corridor – and not the corners – were used for analysis.

Treadmill propulsion

Participants performed the same exercises on a wide wheelchair treadmill (1.20 x 2.00m, Forcelink, Culemborg, The Netherlands). A safety system that stopped the treadmill if a participant were to roll too far to the back of the treadmill was present, as well as one operator in front of the treadmill with access to a safety stop button and one operator at the back of the treadmill. After performing a drag-test [5], a pulley-system was attached to the wheelchair to match the power output requirements of overground propulsion. Participants were instructed to stay in the middle of the treadmill and received no additional (visual) feedback of their speed. Treadmill velocity was verified with a CDT-2000HD calibrated tachometer (Checkline, USA).

Ergometer propulsion

The ergometer trials were performed on a computer-controlled Lode Esseda wheelchair ergometer (Lode B.V., Groningen, The Netherlands, described in [34]). The wheelchair was secured on the ergometer with four straps. Before each trial a static and dynamic calibration were performed to null the system. The simulated inertia on the ergometer was set to the combined weight of the user and the

wheelchair. The simulated friction on the ergometer rollers was set as a constant value based on the coast-down test in the overground condition. Participants received real-time visual feedback of their current speed and heading (speed left, speed right) on a monitor in front of the ergometer. They were instructed to maintain an average speed of 4.0 km/h (1.11m/s).

Analysis

Data were collected during each trial of the measurement session and only the last minute was used for analysis, assuming a steady-state had been reached. Physiological, spatiotemporal, and kinetic outcome variables were obtained from the spirometer, and measurement wheel data. Data (pre-)processing and calculations [4,35], and subsequent statistics were performed using Python 3.7 (Python Software Foundation). Data were first cut so the last minute of each block for each test could be used for analysis. For overground propulsion, only data from the long straights were included. Kinetic data were then filtered using an 4th-order zero-phase Butterworth filter with a 10Hz cut off frequency and discrete outcomes [4] were calculated (push time, cycle time, contact angle, fraction of effective force, mean torque per push, max torque per push, work per push, and negative work per cycle). Heart rate, energy expenditure, and gross mechanical efficiency were calculated based on the spirometer data.

Statistical analysis

Agreement among conditions was determined with a two-way random effects, single rater, absolute agreement Intraclass Correlation (ICC(2,1) [36]). ICC values below 0.50 were considered as poor, values between 0.50 and 0.75 as moderate, values between 0.75 and 0.90 as good, and values higher than 0.90 as excellent [37]. Thereafter, a repeated measures analysis of variance (RM-ANOVA) was performed to test for significant differences between modalities. A Friedman test was used if data were not normally distributed. Planned pairwise comparisons were performed after a significant main effect was detected using paired t-tests (or Wilcoxon signed-rank test) without correction to maximize sensitivity, and effect sizes were given. Effect sizes lower than 0.2 were considered small, values between 0.2 and 0.5 as medium, and higher than 0.5 as large [38]. For the continuous analysis, the mean torque and total force of the last twenty push cycles (normalized from 0-100% with cubic interpolation) for each participant were compared with paired Statistical Parametric Mapping (SPM) t-tests using the spm1d package [28]. All values are presented as is, without (statistical) correction for power output.

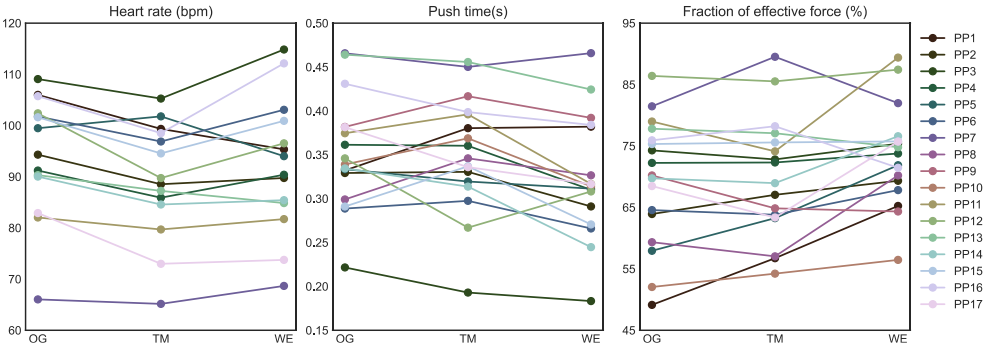


Figure 5.3. Individual responses on three variables with moderate-excellent agreement but significant differences between modalities.

Results

The results and statistical outcomes are provided in Table 5.1. The measured power output by the measurement wheel was significantly lower than expected based on the coast-down test for all modalities ($p<0.001$), -16%, -19%, -11%, for overground, treadmill, and ergometer propulsion, respectively. Please refer to [26] for a more detailed analysis of external power output. Agreement for power output among conditions was not significant for overground versus treadmill propulsion and moderate for overground versus ergometer and treadmill versus ergometer propulsion. However, there was no significant difference between conditions in power output $F(2,32)=1.614$, $p=0.215$. There was, however, a small difference in speed (which did not follow a normal distribution) $\chi^2(2)=6.706$, $p=0.035$ between conditions, specifically for overground/treadmill versus ergometer propulsion (2%). Intraclass correlations for speed were non-significant (due to the low variation in results, i.e. their coefficients of variation were lower than 0.02.

Physiology

Heart rate showed good agreement among all modalities (Table 5.1, Figure 5.3). A significant main effect of modality was found $F(2,26)=7.998$, $p=0.002$. Heart rate was significantly different between overground and treadmill propulsion with a large effect size. Energy expenditure showed moderate agreement among overground and treadmill/ergometer propulsion, and good agreement among treadmill and ergometer propulsion. No significant differences were found between the three modalities $F(2,32)=1.526$, $p=0.233$. Gross mechanical efficiency showed good agreement among overground and treadmill propulsion, but only moderate agreement among overground and ergometer, and poor agreement among treadmill and ergometer propulsion. No significant main effect was found $F(2,32)=0.362$, $p=0.699$.

Spatiotemporal outcomes

Agreement for push time was good among all modalities (Table 5.1, Figure 5.3). A significant main effect was found $F(2,32)=7.323$, $p=0.002$. Participants had a significantly higher push time in the overground/treadmill condition than on the ergometer. Cycle time also showed good agreement among all conditions. Data were not normally distributed and a significant main effect was found $\chi^2(2)=12.118$, $p=0.002$. Post-hoc comparisons showed that cycle time in the overground condition was higher than in treadmill/ergometer propulsion. Contact angle also showed good agreement among all conditions and no significant main effect was found $F(2,32)=2.547$, $p=0.094$.

Kinetics

Fraction of effective force (FEF) showed excellent agreement among overground and treadmill propulsion (Table 5.1, Figure 5.3). However, only moderate agreement was found between overground and ergometer, and treadmill and ergometer propulsion. A significant main effect was found $F(2,32)=5.377$, $p=0.010$. Post-hoc analysis showed significant differences between overground/treadmill and ergometer propulsion with large effect sizes. Mean and peak torque (T_z) per push showed good agreement among all conditions and no significant main effects were found $F(2,32)=1.703$, $p=0.198$, and $F(2,32)=2.628$, $p=0.088$, respectively. A good agreement among all conditions was found for work per push without a significant main effect $\chi2(2)=3.647$, $p=0.161$. However, negative work per cycle only showed moderate agreement between treadmill and ergometer propulsion. A significant main effect was found $\chi^2(2)=26.941$, $p=0.000$ and overground propulsion was found to be different from treadmill/ergometer propulsion.

One part of the push phase (11-26%) that differed significantly between overground and treadmill propulsion ($p=0.012$) was identified by the SPM (Figure 5.4) for total force (F_{tot}). Moreover, a part of the push phase (0-8%) was identified that differed between overground and ergometer propulsion. The SPM also identified one part of the push phase (6-29%) that differed significantly between overground and treadmill propulsion ($p<0.001$) for torque (Figure 5.5). Additionally, a part of the push phase (10-22%) was found to differ significantly between treadmill and ergometer propulsion.

Discussion

The aim of the current study was to compare the physiological and biomechanical characteristics of handrim wheelchair propulsion in able-bodied participants in overground, treadmill, and ergometer conditions at similar submaximal power

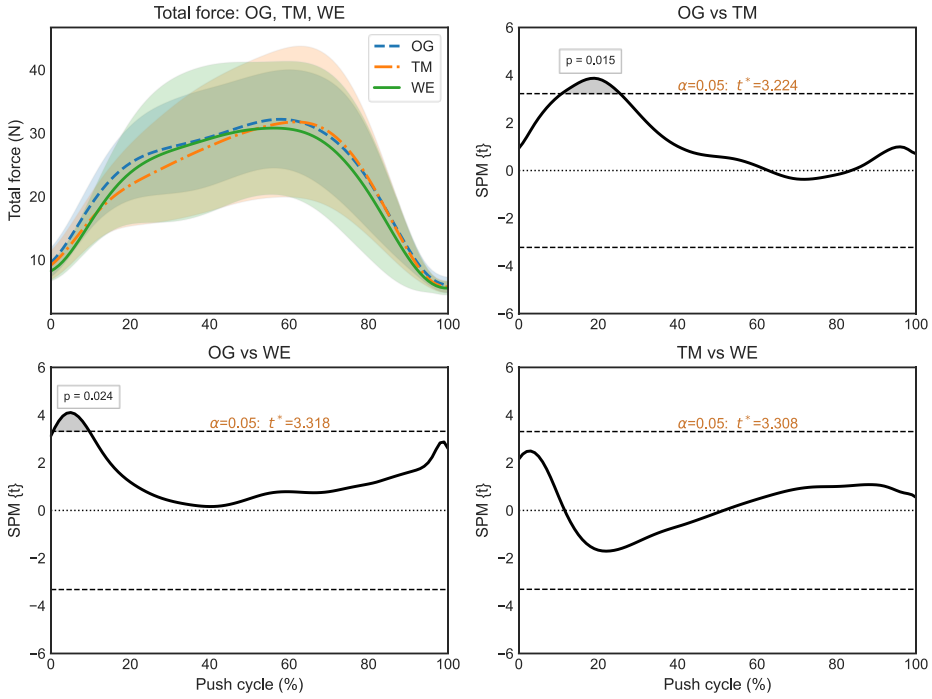


Figure 5.4. Total force (μ±σ) and results of SPMs with pointwise t-statistics and p-values (n=17).

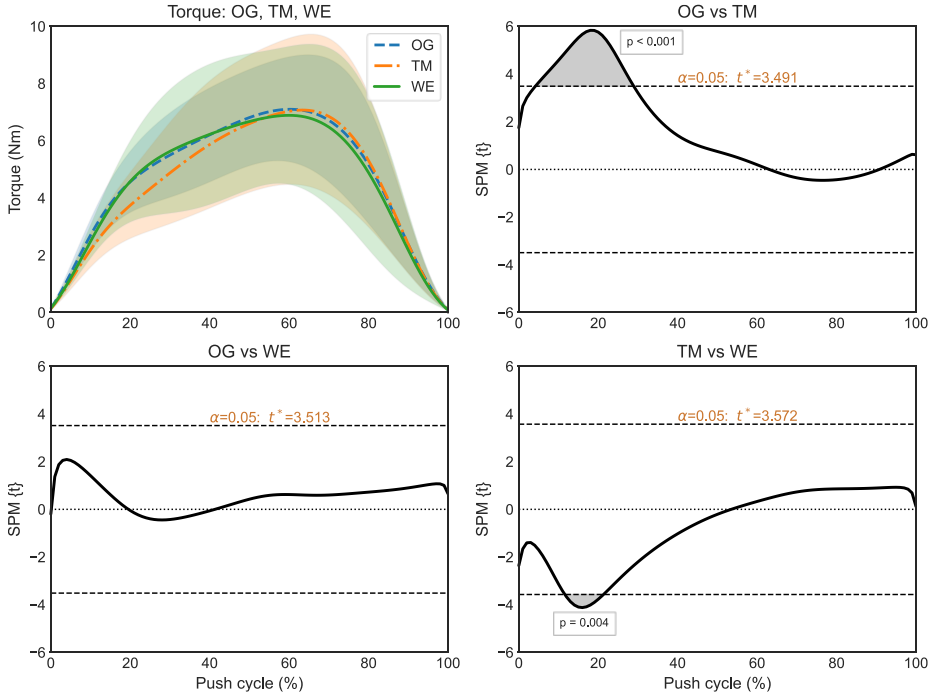


Figure 5.5. Torque (μ±σ) and results of SPMs with pointwise t-statistics and p-values (n=17).

Table 5.1. Physiological and biomechanical characteristics for overground (OG), treadmill (TM), and ergometer (WE) propulsion (n=17).

Variable		Mean (±SD)	Contrast	ICC (95% CI)[a]	p-value[b]	p-value	Effect size
Experimental							
Power output (W)	OG	8.12 (1.41)	OG vs. TM	n.s.[e]	0.10	n.s.[f]	
	TM	7.84 (1.92)	OG vs. WE	0.57 (0.15-0.82)	**0.01**	n.s.[f]	
	WE	8.65 (2.24)	TM vs. WE	0.52 (0.10-0.79)	**0.00**	n.s.[f]	
Speed (m/s)	OG	1.12 (0.02)	OG vs. TM	n.s.[e]	0.48	0.20[g]	
	TM	1.12 (0.00)	OG vs. WE	n.s.[e]	0.57	**0.02**[g]	0.7[i]
	WE	1.14 (0.02)	TM vs. WE	n.s.[e]	0.52	**0.03**[g]	-0.6[i]
Physiological							
Heart rate (bpm)[h]	OG	94.49 (11.80)	OG vs. TM	**0.86** (0.05-0.97)	**0.00**	**0.00**[c]	-1.4[d]
	TM	89.30 (11.31)	OG vs. WE	**0.91** (0.73-0.97)	0.13	0.27[c]	
	WE	92.24 (13.16)	TM vs. WE	**0.88** (0.64-0.96)	0.07	0.07[c]	
Energy expenditure (W)	OG	208.88 (50.00)	OG vs. TM	0.74 (0.42-0.90)	**0.00**	n.s.[f]	
	TM	195.97 (40.08)	OG vs. WE	0.68 (0.31-0.87)	**0.00**	n.s.[f]	
	WE	206.31 (42.47)	TM vs. WE	**0.78** (0.49-0.91)	**0.00**	n.s.[f]	
Gross mechanical efficiency (%)	OG	4.07 (1.07)	OG vs. TM	**0.75** (0.44-0.90)	**0.00**	n.s.[f]	
	TM	4.12 (1.10)	OG vs. WE	0.44 (0.00-0.75)	**0.03**	n.s.[f]	
	WE	4.28 (1.15)	TM vs. WE	0.55 (0.11-0.81)	**0.01**	n.s.[f]	
Spatiotemporal							
Push time (s)	OG	0.35 (0.06)	OG vs. TM	**0.85** (0.64-0.94)	**0.00**	0.96[c]	
	TM	0.35 (0.07)	OG vs. WE	**0.80** (0.34-0.93)	**0.00**	**0.00**[c]	-0.8[d]
	WE	0.32 (0.07)	TM vs. WE	**0.83** (0.35-0.95)	**0.00**	**0.00**[c]	0.8[d]
Cycle time (s)	OG	1.43 (0.47)	OG vs. TM	**0.85** (0.60-0.95)	**0.00**	**0.02**[g]	0.6[i]
	TM	1.31 (0.44)	OG vs. WE	**0.79** (0.49-0.92)	**0.00**	**0.01**[g]	0.8[i]
	WE	1.28 (0.55)	TM vs. WE	**0.75** (0.44-0.90)	**0.00**	0.39[g]	

Variable		Mean (±SD)	Contrast	ICC (95% CI)[a]	p-value[b]	p-value	Effect size
Contact angle (deg)	OG	71.43 (12.34)	OG vs. TM	**0.86** (0.66-0.95)	**0.00**	0.83[c]	
	TM	72.28 (13.64)	OG vs. WE	**0.87** (0.60-0.95)	**0.00**	**0.00**[c]	-1.2[d]
	WE	68.74 (14.01)	TM vs. WE	**0.87** (0.60-0.95)	**0.00**	**0.00**[c]	0.9[d]
Kinetics							
Fraction of effective	OG	69.28 (10.33)	OG vs. TM	**0.92** (0.81-0.97)	**0.00**	0.69[c]	
force (%)	TM	69.67 (9.86)	OG vs. WE	0.71 (0.29-0.89)	**0.00**	**0.02**[c]	0.6[d]
	WE	73.38 (8.11)	TM vs. WE	0.69 (0.30-0.88)	**0.00**	**0.03**[c]	-0.7[d]
Mean torque per	OG	4.63 (1.06)	OG vs. TM	**0.76** (0.47-0.91)	**0.00**	n.s.[f]	
push (Nm)	TM	4.34 (1.33)	OG vs. WE	**0.81** (0.55-0.93)	**0.00**	n.s.[f]	
	WE	4.65 (1.33)	TM vs. WE	**0.81** (0.55-0.93)	**0.00**	n.s.[f]	
Max torque per	OG	7.83 (2.02)	OG vs. TM	**0.81** (0.57-0.93)	**0.00**	n.s.[f]	
push (Nm)	TM	7.43 (2.41)	OG vs. WE	**0.82** (0.57-0.93)	**0.00**	n.s.[f]	
	WE	8.25 (2.73)	TM vs. WE	**0.77** (0.46-0.91)	**0.00**	n.s.[f]	
Work per push (J)	OG	5.94 (1.94)	OG vs. TM	**0.82** (0.57-0.93)	**0.00**	n.s.[f]	
	TM	5.61 (2.17)	OG vs. WE	**0.85** (0.63-0.94)	**0.00**	n.s.[f]	
	WE	5.64 (2.20)	TM vs. WE	**0.80** (0.53-0.92)	**0.00**	n.s.[f]	
Negative work per	OG	-1.44 (0.51)	OG vs. TM	n.s.[e]	0.23	**0.00**[g]	-1.0[i]
cycle (J)	TM	-0.50 (0.20)	OG vs. WE	n.s.[e]	0.21	**0.00**[g]	-1.0[i]
	WE	-0.42 (0.21)	TM vs. WE	0.52 (0.10-0.79)	**0.01**	0.07[g]	

a. two-way random-effects model, single rater, absolute agreement, ≥good values are bold; b. p-values for the intraclass-correlations, significant results are bold. c. paired-comparison with t-test without correction, significant results are bold; d. effect size: Cohen's d; e. non-significant intraclass-correlation; f. non-significant main effect; g. Wilcoxon signed-rank test; h. n=14; i, rank-biserial correlation

output, using conventional standardization methods. In general, almost all physiological, spatiotemporal, and kinetic variables showed moderate to good agreement among modalities without significant differences in level of agreement at the group level. About half of the variables were significantly different between conditions

This was the first study that tried to actively standardize power output and speed between modalities. While no significant difference in power output was found, there was some within-subject variance as could be observed from the poor-moderate intraclass correlations and their large confidence intervals. Given the low mean power output, the absolute differences (up to 0.8W) are small [4], but the relative differences (up to 17%) are rather large. Standardizing power output on the treadmill and ergometer at these low power outputs is almost impossible using current techniques [26] as small shifts in the, already dynamic, weight distribution will affect the rolling resistance and power output as a result [39]. The difference in speed was small and only present for the ergometer. It could be that the actual radius of the wheel/tyre was slightly lower in the ergometer condition due to the high pressure on the small tyre surface, leading to a slight mismatch between measured velocity by the measurement wheel and displayed velocity by the ergometer.

Heart rate was significantly higher during overground propulsion than during treadmill propulsion. While heart rate is more sensitive to emotional and social environmental distractions of the 'open field' [40], the overground modality also contained corners which require more effort to traverse than straight-line propulsion. Although the corners and short-straights were excluded from the data, the increased effort might have been reflected in the heart rate as it does not change instantaneously. This corresponds with the (non-significant) higher energy expenditure and somewhat lower mechanical efficiency that were observed when compared to treadmill and ergometer propulsion.

Push time was significantly different between conditions, but only for the ergometer. Contact angle was also different between overground and ergometer, and treadmill and ergometer propulsion. It is reasonable to expect both push time and contact angle to be highest on the ergometer as there is a reduced emphasis on steering compared to overground and treadmill propulsion [6,23]. However, in this particular case, the real-time feedback of velocity as two numbers on a screen could have motivated participants to make fast adjustments to their speed to keep on target. Accordingly, future studies should look into the effect of different forms of speed/directional feedback on wheelchair ergometers. The somewhat larger cycle time (and cadence) of overground versus treadmill/ergometer propulsion is in line

with earlier studies on gait [11] and wheelchair propulsion [23]. It was expected that the relatively large familiarization time would reduce any possible differences. Cycle time is known to increase after practice on a treadmill [30,41] and it was shown that familiarization reduces differences between modalities in gait [20].

In general, the kinetics obtained from all modalities were highly similar. Fraction of effective force was higher on the ergometer compared to the other two modalities which is in line with the reduced emphasis on steering and balancing. No significant differences and good agreement were found for discrete force related parameters (mean and maximum). However, the continuous analysis showed a different build-up of force/torque for the treadmill condition, highlighting the importance of a more detailed continuous analysis [28]. This difference could be attributed to a more cautious pushing approach as treadmill propulsion requires the most finesse due to its stringent steering requirements, though differences appear to be very small and should not be overstated.

A limitation in the study was that, despite standardizing power output between modalities, within-subject variance in power output was still present. However, as the conditions were not significantly different from each other at the group level, the approach did appear to work to an extent. Additionally, the feedback provided to participants of their speed was specific for the different conditions, which could have influenced participant behaviour. Adding similar task constraints to treadmill/ergometer as to overground propulsion in the form of Virtual Reality could provide a higher degree of standardization. Overground and ergometer propulsion will, however, still have different task requirements than treadmill propulsion as the latter does not generally allow for self-selected speeds. Furthermore, the study did not look at the kinematics among modalities. It could very well be that similar outcomes are reached with different kinematic solutions. Moreover, involvement of the trunk was attributed to differences found between treadmills and ergometers in the past [42]. Indeed, some of the differences found between ergometer and overground /treadmill propulsion could be explained by the involvement of the trunk as rotational stability is higher on an ergometer.

Finally, changes that have been reported for walking have been small and the differences that reached statistical significance may not be functionally meaningful [14]. Most of the differences reported here are below the smallest detectable differences reported in [4], indicating that this also could be the case for wheelchair propulsion. It should also be noted that the results in the current study were obtained in able-bodied participants that form a more homogenous group but are also less experienced in wheelchair propulsion [30]. Possible effects of experience

and impairment remain unexplored. As such, future studies should look into the clinical relevance of these differences and their effects on specific measurement protocols used in (clinical) practice within patient populations.

Conclusions

In general, overground, treadmill, and ergometer wheelchair propulsion provide quite similar physiological and biomechanical outcomes after standardizing power output, speed, and a familiarization period. Some small differences were found in a number of physiological, spatiotemporal, and kinetic parameters. Moreover, the build-up of the push phase is different between modalities. However, differences are small and unlikely to be functionally meaningful. Future studies should look into the possible sources of differences and the clinical relevance of differences between modalities by comparing testing protocols. Under the constraints of identical standardized power output and velocity, the studied modalities of handrim wheelchair propulsion appear quite comparable.

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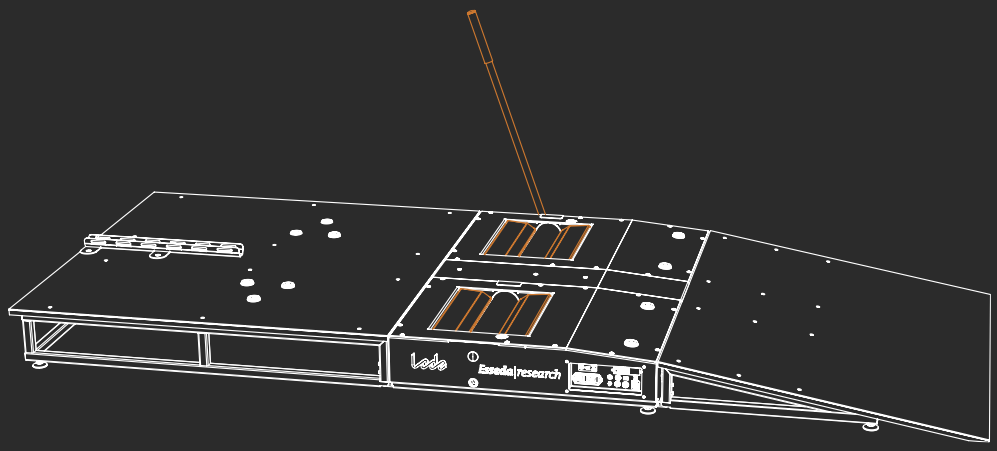
6

Measuring wheelchair racing propulsion



Evolution of wheelchair racing propulsion technique over three weeks of wheeling practice on an instrumented ergometer in able-bodied novices

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Abstract

The acquisition of daily handrim wheelchair propulsion skill as a multi-layered phenomenon has been studied in the past. Wheelchair racing, however, is considerably different from daily handrim wheelchair propulsion in terms of propulsion technique as well as interfacing and underlying equipment. Understanding wheelchair racing skill acquisition is important from a general motor learning and skill acquisition perspective as well as from a performance and injury prevention perspective in wheelchair athletics. The aim of the current lab-based study was twofold: to investigate the evolution of racing wheelchair propulsion technique and wheeling skill among a sample of novices and to compare their technique with an experienced wheelchair racer under comparable conditions. A convenience sample of 15 able-bodied novices (8 male, 7 female, 22 yrs old) completed a standardised three-week submaximal uninstructed practice protocol (3 weeks, 3 sessions per week, 3x4 min per session) in a racing wheelchair on a wheelchair ergometer. Required wheeling velocity was set at 2.78 m/s (10 km/h) and a rolling friction coefficient of 0.011, resulting in a mean target load of 21W, was used. For comparison, an experienced T54 Paralympic athlete completed one block of the same protocol. Kinetics, kinematics, and physiological data were captured. A mixed effects regression analysis was used to examine the effect of practice for the novices, while controlling for speed. All participants finished the protocol successfully. However, not all participants were able to achieve the target velocity during the first few sessions. Statistically significant improvements over time were found for all outcome measures with the exception of mean power and torque per push (i.e. higher GME, lower metabolic strain, longer push and cycle times). The athlete used a significantly greater contact angle and significantly better outcomes (i.e. less straining) on most metabolic and kinetic variables were seen. While the athlete used a semi-circular propulsion technique, most participants used a double looping over technique. Three weeks of uninstructed wheelchair racing practice significantly improved efficiency and technique among a group of novices. The results from this study are in line with previous studies on daily handrim wheelchair propulsion. A comparison with an experienced athlete showed expectedly that there is still a large performance (and knowledge) gap to be conquered. Future studies should emphasize the use of feedback and a realistic learning environment, and include perspectives of coaches and trainers.

Introduction

Wheelchair racing was part of the first international wheelchair sporting competition for people with disabilities in 1952 [1]. Since then, wheelchair racing and racing wheelchairs have greatly evolved, the latter now consisting of a long-base three-wheel lightweight configuration with one large wheel in the front and two large rear wheels with relatively small handrims in order to reach and maintain high speeds [2]. Races are organised for field and track events and include sprints, middle-long distances, and long distances, including the marathon. Athletes compete in their own class to ensure that athletes with similar impairment race against each other [3]. Involvement in sports, such as wheelchair racing after rehabilitation has a positive influence on physical [4] and psychological health and well-being [1,5]. Therefore, it is important that patients with lower-limb impairments get involved in new (disability) sports, such as wheelchair sports, during and after rehabilitation. This requires them to learn new propulsion (and game) skills, especially assumed to be present in wheelchair racing, where the wheelchair design and interface require different postures and propulsion technique. Although there is existing knowledge on skill acquisition during daily wheelchair propulsion [6–8], mechanisms of learning wheelchair racing are still unclear.

To become more proficient in wheelchair racing, an athlete either needs to increase the physical work capacity or become more efficient [1]. Experienced wheelchair racing athletes have been studied to gain insight in their propulsion technique and corresponding mechanical efficiency. Compared with regular handrim wheelchair propulsion, athletes use a larger contact angle of approximately 180° and start at 20° past the top-dead centre of the handrim [9,10]. Starting further on the handrim allows athletes to be in a more horizontal position in the racing wheelchair, reducing air resistance. Moreover, wheelchair racers use gloves as coupling is infeasible at racing speeds. During racing conditions as segmental velocities increase, the push is performed as a stroke against the rims. Whereas, during regular handrim wheelchair propulsion one can grab the handrims, making push-pull action possible [11]. To increase wheelchair racing performance, athletes need to learn this new movement, requiring different movement patterns and adaptations in both physiology and technique. Little scientific research has focused on wheelchair racing skill acquisition so far.

The acquisition of daily wheelchair propulsion skill has been extensively studied for regular handrim wheelchairs in wheelchair users [12] and (novice) able-bodied participants [6–8]. These studies generally examined steady-state submaximal performance at low speeds, using gross mechanical efficiency as the main outcome

measure [13]. Experienced participants have a higher mechanical efficiency, meaning they are able to produce the same amount of external power output at a lower energetic cost. This is in line with the framework of Sparrow and Newel [14,15] and Almåsbakk et al. [16], where cyclic movement patters are thought to emerge through the interaction of different constraints, with metabolic energy as an optimization parameter. The increase in mechanical efficiency can be due to improvements in propulsion technique and/or physiological adaptation [12]. A high mechanical efficiency in wheelchair propulsion was linked to increased wheeling proficiency, expressed as greater contact angles and a decreased push frequency, which is especially beneficial as this is insinuated to improve mobility and reduce risk of injury [17,18].

A better technique and higher efficiency are also beneficial to racing performance [9,10] and could reduce injury sensitivity. However, racing wheelchair and regular handrim wheelchair propulsion techniques are distinct and there is no information available on the acquisition of wheelchair racing skill. One challenge specific to wheelchair racing is to maintain extreme high velocities. Smaller sized handrims help to meet the required speeds to some extent, since linear hand speed can be kept lower with smaller rim diameters which was shown to be more efficient and less straining in experienced wheelers [19]. Yet, the majority of wheelchair racing performance probably still comes down to underlying coordination and skill. Like regular handrim wheelchair propulsion, wheelchair racing can be approached as a cyclical skill where motor learning can be quantified as a decrease in energy expenditure [8,14]. As such, mechanical efficiency is expected to increase, as mastering this task should result in more optimal kinetic and kinematic solutions [7,16].

The current study focused on the initial motor learning process of three weeks of lab-based natural wheelchair racing propulsion practice in inexperienced able-bodied participants on a wheelchair ergometer. Able-bodied participants were chosen as they are full novice to the task and form a relatively homogenous group (similar age, lack of wheelchair experience and no comorbidities), minimizing the inter-individual variation which allows to better isolate the effect of natural learning on the outcomes of the motor learning process [20]. Additionally, an experienced athlete performed a similar protocol to provide a reference for propulsion technique and efficiency of actual wheelchair racing propulsion.

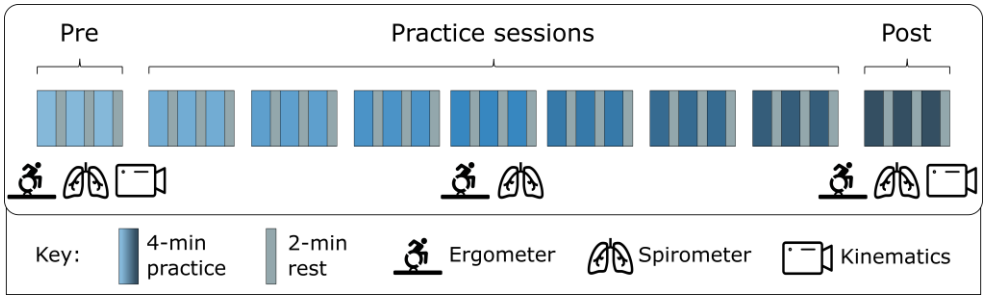

Figure 6.1. Overview of the protocol: participants were tested on 9 occasions spread over three weeks with three blocks of practice each. Data were captured during all sessions, but kinematics were only recorded during the first- and last (pre and post) session.

Methods

Participants

The current study used a convenience sample of 15 inexperienced able-bodied participants (7 female/8 male, 22.0 (±1.35) years old, 69.3 (±9.87) kg). Participants were eligible for inclusion if they had no previous experience with wheelchair propulsion and no contraindications for exercise (PARQ, [21]). Additionally, one high-level T54 middle-distance athlete, was included for comparison (male, 67kg). All participants provided written informed consent after receiving detailed information about the study. The study was approved by the local ethical committee of the Centre for Human Movement Sciences, University Medical Centre Groningen, University of Groningen.

Study design

Participants received a total practice load of 108-minutes consisting of nine sessions (three sessions per week for three weeks) of 3x4 minutes of submaximal manual racing wheelchair exercise (Figure 6.1) on an instrumented wheelchair roller ergometer (Lode, Groningen, The Netherlands, [22]). They received no advice on propulsion technique prior to the experiment and no feedback during the sessions, resulting in a natural learning process [23]. Before the start of an exercise block, the sole instruction was to propel at a required speed of 2.78 m/s (10 km/h) and to hit/push the handrim with the soft hand gloves. The required velocity was based on a pilot determining a feasible speed for untrained participants, yet fast enough to somewhat approach wheelchair racing propulsion. A computer screen in front of the participants provided visual feedback on the actual and target speeds [24].

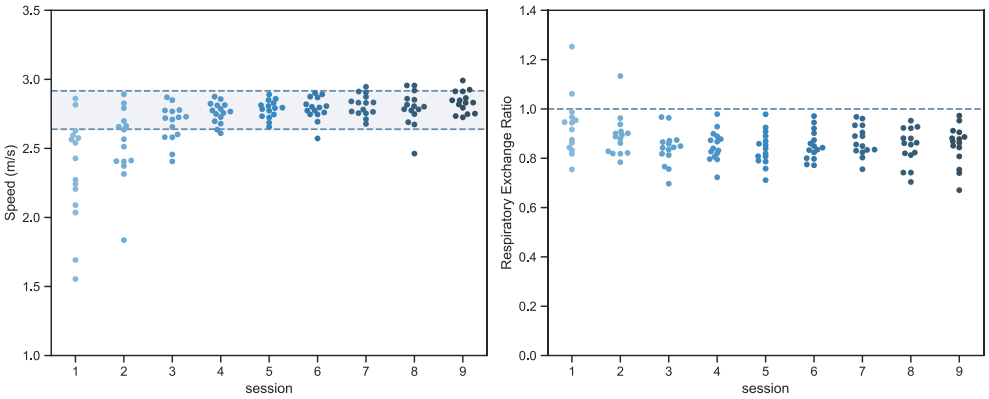

Figure 6.2. Swarmplot of the ability of individual participants to match the target speed (±5%) and the RER (<1.0) during each session (n=15).

Materials

Wheelchair

All tests were performed in the same experimental Amasis racing wheelchair (Wolturnes, Nibe, Denmark) with 0.71m (28-inch) wheels and 0.38m (15-inch) handrims on the roller ergometer. The wheelchair was not individually accommodated. Participants used soft hand gloves to push the handrim. The athlete performed in his personal racing wheelchair with 28-inch wheels and 37cm diameter handrim. Tire pressure of the rear wheels was set at 800 kPa (8 bar) before every session.

Physiology

Metabolic data were collected using a K5 Cardio-Pulmonary Exercise Testing (CPET) spirometer (COSMED, Rome, Italy) in breath-by-breath mode. Turbine, room air, reference gas, and delay calibrations for the spirometer were performed before each session. Heart rate was measured with a Garmin HRM Dual (Garmin International Inc, Kansas, USA) connected with the CPET. Participants were asked to rate their perceived exertion on a 6-20 Borg scale [25].

Kinetics

Force and velocity data were collected with an Esseda (Lode BV, The Netherlands) wheelchair ergometer at 100Hz [22,26]. The ergometer was calibrated to account for static and dynamic friction before each session. A rolling friction coefficient of 0.011 was set, resulting in a theoretical power output of 21W at the mean body weight of the novice participants in this study. The coefficient was based on eight coast-down tests [27] with two athletes at the outdoor athletics track at the Olympic Training

Centre Papendal. The athlete, originally part of another study, performed at a power output of 28W.

Kinematics

Finally, an active cluster marker was placed on the right-hand glove and tracked by an optoelectronic camera system (Optotrack, Northern Digital, Waterloo, Canada) at 100Hz. The cluster was used to determine the location of second and fifth metacarpal (M2 and M5) during propulsion.

Analyses

All analyses were performed in Python (The Python Foundation, [28]) using a custom package available on the Python Package Index [29]. To examine the motor learning process over time, all blocks were included and the last minute of each block was used, assuming steady-state propulsion. Finally, the mean of the three blocks per session were used for statistical analysis. Pre-processed data are included as a supplementary file in a comma separated values (.csv) format.

Physiology

Heart rate, respiratory exchange ratio (RER), and energy expenditure (EE) were obtained from the CPET system. Gross mechanical efficiency (GME) was calculated from the EE and the external power output (PO) obtained from the ergometer:

$$\mathrm{GME}(\%) = \mathrm{EE} * \mathrm{PO}^{-1} * 100 \quad (1)$$

GME for sessions where the mean RER was higher than 1.0 were discarded, which was the case for three samples (Figure 6.2).

Kinetics

Kinetic data (force on roller) from the ergometer were first filtered using a 15Hz 4th-order zero-phase Butterworth filter. Propulsion technique variables (contact angle, push & cycle time, mean & peak torque and power per push, and work per push) were then determined based on the speed and force data from the ergometer for the left and the right side. Afterwards, the mean of the left and right side was used for further (statistical) analysis.

Kinematics

The last fifteen seconds of the M2 virtual marker location were plotted for each block of the first- and last session. Three raters (GJ, RK, and PW) qualitatively rated the propulsion technique using the definitions of Boninger et al. [30]: arcing (ARC), single looping over propulsion (SLOP), semicircular (SC), and double looping over propulsion (DLOP). The most frequent technique among blocks was identified as the session technique. In the case of a tie, the observed technique of the last block

was used, this was done for each rater individually. Finally, the most frequent technique among raters was determined and reported.

Statistics

Physiology & kinetics

A linear mixed effects analysis of the effect of time (session 1-9) was performed using R (R Core Team, [31]) and lme4 [32]. Time and speed (without interaction term) were included in the model as fixed effects. Speed was added as not all participants were able to achieve the target velocity during the first sessions (Figure 6.2). Separate random intercepts and slopes were added for participants for the effect of time. The final model was defined as:

$$\text{outcome} \sim \text{session} + \text{speed} + (1 + \text{session}|\text{subject}) \quad (2)$$

There were no obvious deviations in the residual plots with regards to homoscedasticity or normality. P-values were obtained with a likelihood ratio test of the full model versus a model without the effect of time. Data from the last session were compared with the athlete using a one-sample t-test.

Kinematics

Fleiss' Kappa was calculated to determine the agreement between raters with regards to the propulsion technique and were interpreted based on the suggestions of Landis and Koch [33]. A contingency table was produced to describe the development of propulsion technique, but no further statistical analysis was performed due to the sparsity of the data.

Results

All participants completed the experiment successfully. Yet, not all novices were able to achieve the desired velocity (±5%) during the first three sessions (Figure 6.2). Resultingly, speed and power output significantly increased between subsequent sessions as participants were increasingly able to achieve the target speed (Figure 6.2). Concomitantly, the average respiratory exchange ratio was higher than 1.0 during the first two sessions for some of the participants (2/15 in session 1 and 1/15 in session 2).

Physiology & kinetics

Physiological and kinetic aggregates and statistical outcomes are displayed in Table 6.1 and Figure 6.3. A statistically significant improvement (i.e. higher GME, lower metabolic strain, higher push and cycle times) over time was found for all outcome measures with the exception of mean power and mean torque per push. Moreover, the perceived exertion also significantly lowered over time from "hard" to "fairly

light". The athlete showed significantly better outcomes (i.e. less straining) on most metabolic and kinetic variables.

Kinematics

Results of the qualitative assessment of propulsion technique during the pre- and post-test are displayed in Table 6.2 and Figure 6.4. Agreement among the three raters was substantial during the pre-test κ=.790, $p < .001$ and almost perfect during the post-test κ=.813, $p < .001$. Most participants started with a SLOP (53%) technique, but the majority gravitated towards a DLOP technique in the post-test (73%). The athlete used an SC propulsion technique.

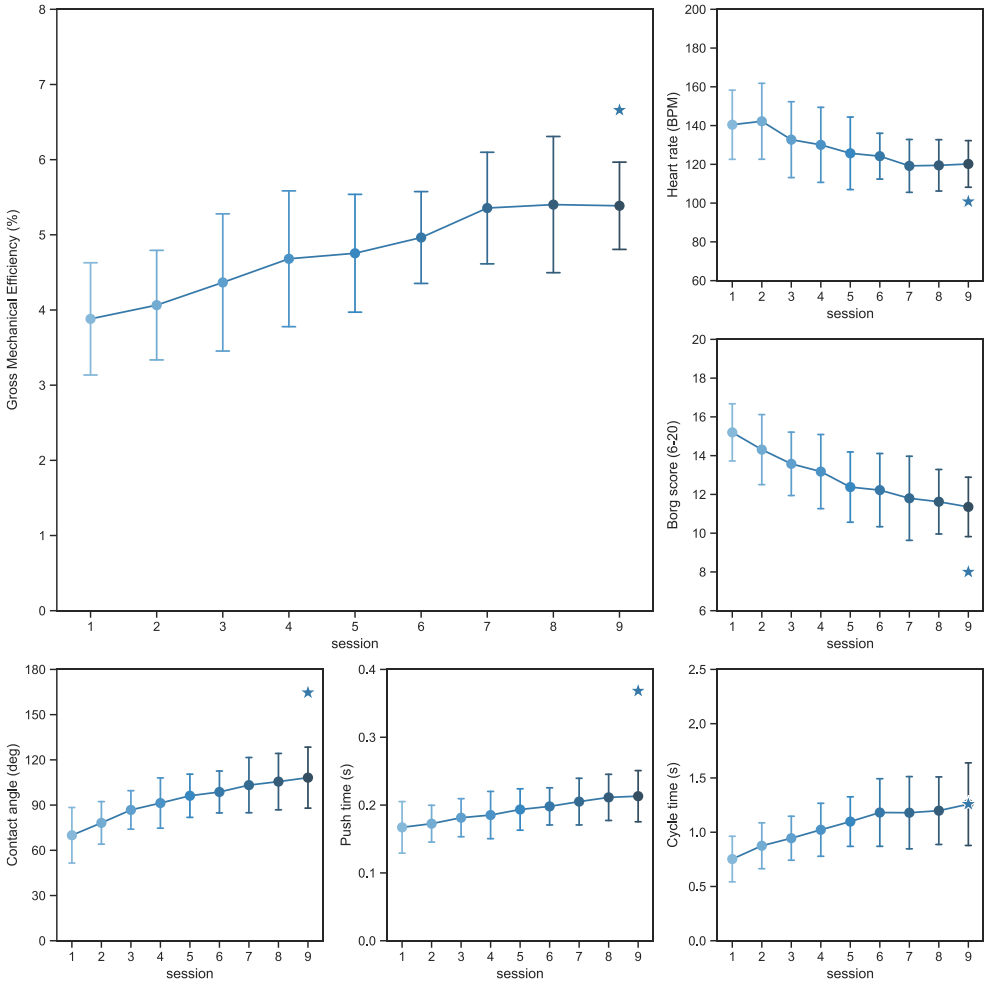

Figure 6.3. Outcomes for six metabolic and kinetic parameters aggregated (mean and standard deviations) by session (★=athlete). All variables shown significantly changed over time. The athlete scored significantly 'better' on all parameters shown, except for cycle time.

Table 6.1. Outcomes: last minute of each block aggregated by session and reference data of a single wheelchair athlete with mixed effects regression and one-sample t-test results (n=15).

Variable	Session									Statistics					
	1	2	3	4	5	6	7	8	9	A[a]	speed[c]	time[c]	$\chi2$	p[d]	t[e]	p
Protocol																
Speed	2.34 (0.39)	2.53 (0.27)	2.68 (0.15)	2.76 (0.08)	2.78 (0.08)	2.79 (0.09)	2.81 (0.08)	2.79 (0.14)	2.84 (0.08)	2.78	n.a.	2.45 (0.07)	13.987	<0.001	2.60	0.02[f]
Power	19.7 (4.49)	21.6 (3.55)	22.7 (3.15)	23.6 (3.2)	23.6 (3.04)	23.7 (3.03)	23.8 (2.97)	23.7 (3.20)	24.1 (2.97)	28.4	n.a.	0.43 (0.09)	13.195	<0.001	-5.50	<0.001[f]
Physiological																
RPE (6-20)	15.2 (1.8)	14.3 (2.14)	13.6 (1.85)	13.2 (2.09)	12.4 (2.03)	12.2 (1.98)	11.8 (2.23)	11.6 (1.77)	11.4 (1.60)	8.0	-2.04 (0.47)	-0.36 (0.05)	22.937	<0.001	8.20	<0.001
HR (BPM)	141 (19.5)	142 (20.7)	132 (20.2)	130 (20.5)	126 (19.8)	124 (12.9)	119 (14.3)	120 (14.1)	120 (12.8)	101	2.36 (4.95)	-3.21 (0.56)	19.128	<0.001	6.04	<0.001
EE (W)	552 (152)	543 (115)	544 (134)	522 (123)	510 (105)	482 (72.7)	451 (71.1)	445 (61.4)	450 (58.2)	425	112 (31.4)	-21.7 (3.71)	19.105	<0.001	1.52	0.06
GME (%)[b]	3.88 (0.78)	4.06 (0.76)	4.37 (0.94)	4.68 (0.93)	4.75 (0.81)	4.96 (0.63)	5.36 (0.77)	5.4 (0.94)	5.39 (0.60)	6.66	0.74 (0.32)	0.18 (0.83)	18.276	<0.001	-8.20	<0.001
Kinetics (per push)																
Contact angle (deg)	70.0 (20.4)	78.2 (15.0)	86.8 (13.5)	91.3 (17.7)	96.2 (15.7)	98.7 (14.6)	103 (18.8)	106 (19.3)	108 (20.7)	165	28.7 (4.80)	3.12 (0.84)	10.216	<0.01	-10.4	<0.001
Push time (s)	0.17 (0.04)	0.17 (0.03)	0.18 (0.03)	0.19 (0.04)	0.19 (0.03)	0.20 (0.03)	0.21 (0.04)	0.21 (0.04)	0.21 (0.04)	0.37	0.003 (0.001)	0.006 (0.002)	8.9818	<0.01	-15.4	<0.001
Cycle time (s)	0.75 (0.23)	0.88 (0.22)	0.94 (0.21)	1.02 (0.26)	1.10 (0.25)	1.18 (0.32)	1.18 (0.34)	1.2 (0.32)	1.26 (0.39)	1.26	0.32 (0.07)	0.04 (0.01)	8.8434	<0.01	0.00	0.50
Mean torque (Nm)	6.76 (2.31)	7.1 (2.39)	7.15 (2.08)	7.56 (2.44)	7.79 (2.48)	8.07 (2.92)	7.55 (1.75)	7.69 (1.85)	7.73 (1.87)	6.17			0.0242	0.87	3.22	0.03
Peak torque (Nm)	12.4 (3.85)	12.9 (3.97)	13.2 (3.92)	13.9 (4.24)	14.4 (4.58)	15.1 (4.93)	14.6 (3.76)	15.0 (4.44)	15.0 (4.10)	13.0	2.35 (0.81)	0.24 (0.10)	4.8888	0.03	1.89	0.04
Work (J)	8.59 (3.73)	10.2 (3.85)	11.3 (3.68)	12.7 (5.21)	13.7 (5.00)	14.5 (5.8)	14.2 (4.71)	14.9 (4.87)	15.3 (5.41)	18.3	6.25 (1.01)	0.49 (0.16)	7.5477	<0.01	-2.07	0.03
Mean power (W)	51.0 (23.7)	57.2 (22.8)	60.3 (19.1)	65.5 (22.1)	68.0 (22.8)	70.5 (26.0)	66.5 (16.0)	67.2 (17.2)	68.6 (17.2)	48.1			0.0768	0.78	4.61	<0.001
Peak power (W)	93.0 (40.0)	104 (38.4)	111 (36.3)	120 (38.5)	126 (42.1)	132 (44.3)	128 (34.0)	131 (40.7)	133 (37.7)	102	54.8 (7.07)	2.06 (0.85)	5.2618	0.02	3.26	<0.01

a. athlete, single block at higher resistance; b. cases with RER < 1.0; c. unscaled estimates ± standard errors; d. p-value from likelihood ratio test; e. one-sample t-test (df=14); f. two-sided p-value

Discussion

This is the first study to examine the motor learning process within the first three weeks of practice of inexperienced able-bodied participants learning wheelchair racing propulsion on a wheelchair ergometer. Participants became more proficient in wheelchair propulsion in a racing wheelchair, which was reflected in the successful completion of the practice bouts in terms of speed and power output, and significant improvements in propulsion technique and corresponding reductions in metabolic cost and perceived exertion. However, the novice participants still had a significantly different propulsion technique compared to the professional athlete.

Physiology

Lower heart rates and energetic cost suggest that the propulsion technique became less strenuous for the inexperienced participants over time, which is corroborated

Table 6.2. Contingency table of propulsion technique during the first session and last session n(%).

		Before			
		ARC	DLOP	SC	SLOP
After	ARC	0 (0%)	0 (0%)	0 (0%)	0 (0%)
	DLOP	3 (20%)	2 (13%)	0 (0%)	6 (40%)
	SC	0 (0%)	0 (0%)	2 (13%)	0 (0%)
	SLOP	0 (0%)	0 (0%)	0 (0%)	2 (13%)

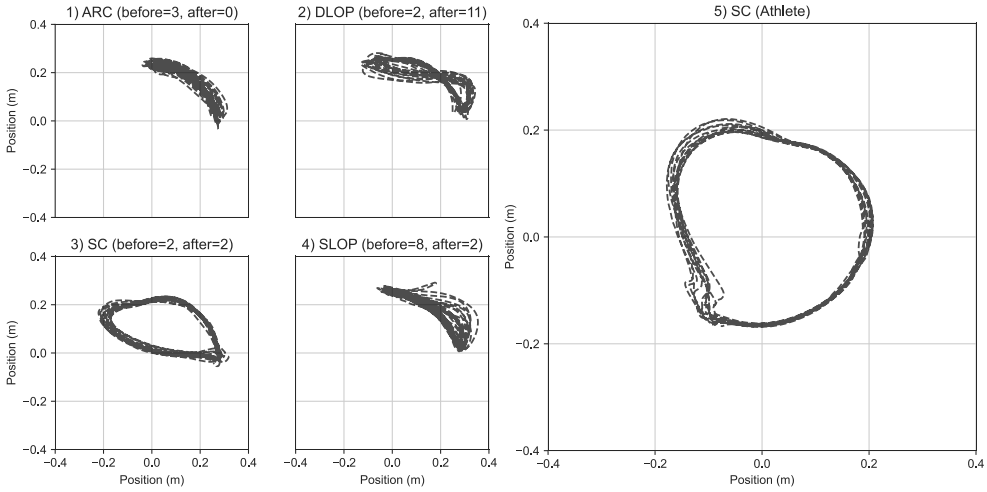

Figure 6.4. Typical kinematics examples of the last fifteen seconds of the M2 virtual marker position in the sagittal plane displayed by different participants (each subplot shows one): 1) arcing, 2) double looping over, 3) semicircular, 4) single looping over propulsion, 5) semicircular, athlete. All data are from the last block of the last session with the exception of 1).

by the decrease in perceived exertion (RPE). While these lower heart rates may have been the result of improved cardiorespiratory fitness, the American College of Sports Medicine (ACSM) states that 150 min of moderate exercise, or 75 min of vigorous exercise per week are required [34]. Since these requirements are not met with 108 min of exercise and as energy expenditure also decreased, the lower heart rates were more likely caused by improvement in neuromuscular coordination and thus a reduction in cardiometabolic requirements with improved coordination and skill level [20]. Accordingly, gross mechanical efficiency follows an inverse pattern, increasing from 3.9% to 4.5% (39%). This is relatively low compared to other studies in regular handrim wheelchair propulsion [6–8,12,20,24]. Gross mechanical efficiency was expected to be higher compared to daily wheelchair propulsion, due to the higher power output requirements [35,36] and since the use of small hand rims improves conditions at muscle level, resulting in lower energetic costs [11]. Experienced wheelchair racing athletes generally have a more efficient propulsion technique [10], as was the case in the current study. Yet, the results of the experienced athlete were not comparable to those of experienced wheelchair racing athletes in previous studies [9,10,37]. However, the speed and power output of the current study (2.78 m/s) were much lower than those of previous studies (3.60-7.20 m/s), which could explain the difference in mechanical efficiency [9,37].

Kinetics

Coordination of wheelchair racing propulsion is complicated due to the use of gloves, a small hand rim and a fast-spinning wheel [38]. Coupling happens outside of the visual field which makes it harder to start the push with the same hand velocity compared to the wheel velocity [24]. As a result of practice, participants were able to increase their contact angle and decrease their push frequency. This is in line with previous studies in regular handrim wheelchairs where similar changes were observed [6–8,20,24] and the longer-slower hypothesis as proposed by Sparrow and Newell [15]. In contrast to the other variables, mean power did not significantly change. However, using the same mean power on a longer push means that the participants were able to increase the amount of work delivered per push. The wheelchair athlete used an even larger contact angle, resulting in an even longer push time. Even though the athlete performed at a higher external power output, this still allowed for a lower mean and peak power per push.

Kinematics

Only two (13%) participants adopted a semi-circular propulsion pattern which is ubiquitous in competitive wheelchair racing. All other participants used different techniques with the majority (73%) gravitating towards a double looping over propulsion. This propulsion technique is often associated with regular handrim

wheelchair propulsion [30,39], though the single looping over propulsion is more common. On the other hand, athletes use a near horizontal trunk position during wheelchair racing which limits the available range of motion for the recovery pattern and forces a starting position on the handrim that is beyond top-dead centre. As the current study was performed in a lab setting, where no wind or air resistance was present, there is no need to employ a more horizontal position and reduce the exposed surface area. This might have encouraged a different propulsion pattern as the task/environment constraints are different than those of actual wheelchair racing, leading to a different movement solution. However, it is still unclear whether a longer attenuation period may lead to the same kinematic solutions. Finally, while pattern classification is subjective, the inter-rater agreement in this study was high. Nevertheless, some quantitative measures are available and should be further developed to provide a more robust objective method of describing propulsion patterns [39,40].

Limitations

Despite piloting beforehand, not all participants seemed able to achieve the desired velocity during the first three sessions. Wheelchair racing propulsion is a relatively hard task which takes a certain amount of skill to even begin the process of mastery. To borrow terminology from the electronic-sports domain: it has a high skill floor. However, as speed was included in the mixed effects regression model, the statistical outcomes "account" for the effect of speed. The inclusion of one experienced athlete provided information about the reference technique of racing wheelchair propulsion. Yet, one athlete does not provide propulsion technique representable for all wheelchair racing athletes across all disciplines. The athlete also performed at a higher external power output than the novice participants. Which is known to influence wheelchair propulsion technique parameters and mechanical efficiency [35,36,41]. However, for other parameters such as RPE and heart rate the differences found are even more pronounced when considering the higher power output. Finally, it must be noted that all results were obtained on a wheelchair ergometer and in able-bodied participants. The ergometer provides a more constrained, yet standardised, environment than a wheelchair racing track or other training environments. Previous studies in regular handrim wheelchair propulsion, however, have not found any differences between treadmill/ergometer and overground propulsion practice [20,42]. Whether this is also the case for the more complicated wheelchair racing task is an avenue for future research.

Perspectives

The current study examined the effects of a natural learning setup, to improve our understanding of the learning process of wheelchair racing propulsion. Yet,

previous studies in daily wheelchair propulsion have also examined the effects of variable practice [20] and feedback [8]. Exploring those setups would be especially interesting since learning processes in sports are generally guided or supervised by trainers or coaches. The effect of feedback or variable practice could therefore provide them with valuable input. Perhaps one of the most essential parts of wheelchair racing technique is the coupling of the gloved hand with the handrim. Therefore, studies that specifically examine this coupling are needed (e.g. using hand velocity and 3D kinetics). Finally, since the sport is only eligible for athletes with an impairment, this seems crucial for understanding wheelchair racing. As these athletes usually already have some wheelchair experience, but might have a reduced physiological capacity or other impairments that influence the learning process. Therefore, future research should also include experienced wheelchair users that are new to wheelchair racing propulsion.

Conclusion

The current study on motor learning processes found similar results for wheelchair racing and previous research in daily wheelchair propulsion. Similar to previous studies, participants show larger contact angles and a decreased push frequency. Using only uninstructed practice, participants increased their mechanical efficiency by 39% (1.5%-point). A comparison with an experienced athlete showed that both the propulsion pattern, and physiological and kinetic outcomes are still different. The performance gap between the participants and the experienced athletes shows that much can still be learned about the difficult task that is wheelchair racing.

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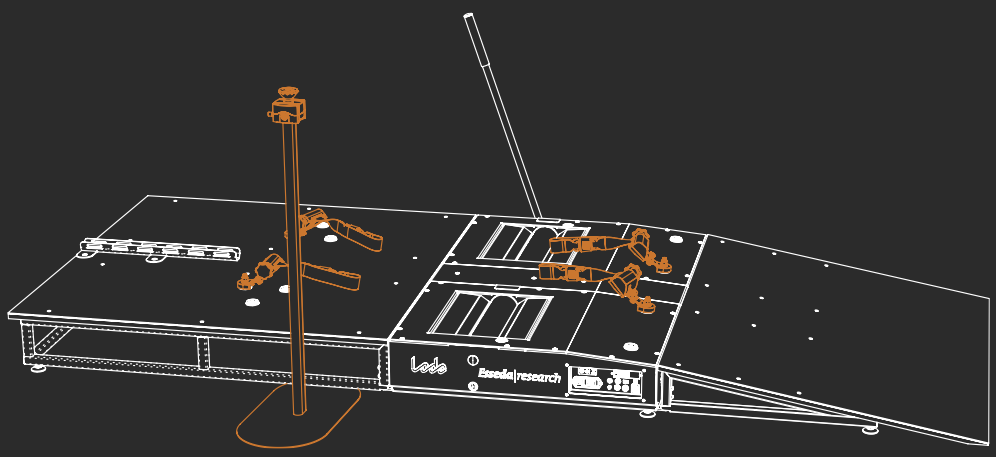
7

Measuring power-assisted propulsion



Practice-Based Skill Acquisition of Pushrim-Activated Power-Assisted Wheelchair Propulsion Versus Regular Handrim Propulsion in Novices

R. de Klerk, T. Lutjeboer, R.J.K. Vegter, L.H.V. van der Woude
– Journal of NeuroEngineering and Rehabilitation

Abstract

Regular handrim wheelchair (RHW) propulsion is straining for the upper extremities and wheelchair users often experience overuse problems. A recent advancement in wheelchair technology that could assist users is the pushrim-activated power-assisted wheelchair (PAPAW). PAPAWs are challenging to control, yet it is unclear how people learn to use a PAPAW. The purpose of this study is to examine early skill acquisition through practice in PAPAWs and compare it with RHWs. Twenty-four able-bodied novices were randomly allocated to either the RHW group or the PAPAW group. The experiment consisted of five sessions with three blocks of 4 min steady-state practice at 1.11 m/s and 0.21 W/kg. Finally, a transfer to the other mode was made. Data were collected with a drag-test, breath-by-breath spirometry, and a motion capture system. The last minute of each four-minute block was used for analysis. A mixed analysis of variance (ANOVA) was used to compare group, time, and interaction effects. Both groups improved their (assisted) mechanical efficiency, reduced their stroke rate, right-left and forward-backward deviation on the treadmill, and had a lower rate of perceived exertion (RPE) over time. (Assisted) mechanical efficiency was higher for the PAPAW group than for the RHW group and RPE was lower. However, left-right and forward-backward deviation was also found to be higher in the PAPAW group. At the group level the energetic cost of RHW and PAPAW propulsion can be lowered through low-intensity practice in novice users. The PAPAW is more 'efficient' than the RHW given the reduced energy requirement of the user from the motor assist, but more difficult to control. Future studies on PAPAWs should focus on the control needs of the user and their interaction with the power-assist technology.

Background

A large section of persons with a disability is dependent on wheelchairs for locomotion with approximately 70 million people worldwide that rely on a wheelchair for their mobility [1]. Regular handrim wheelchairs (RHW) increase independent mobility for people with lower limb impairments, which improves their quality of life [2], while keeping them physically active to prevent a non-active lifestyle [3]. However, wheelchair propulsion is known to have a low mechanical efficiency and wheelchair users often experience overuse problems, especially in the shoulders, but also in the elbows and wrists [2, 4, 5].

Hence, alternatives to RHWs have been developed in the past for those with shoulder pain, limited arm function, or upper body capacity. One such alternative is the fully-powered wheelchair [6]. Although they might reduce strain on the shoulders and arms, they are expensive, require vans and lifts for transportation [7], and they encourage a much less physically active lifestyle [8]. Thus, various other substitutes to handrim and fully-powered wheelchairs have been proposed such as crank and lever propelled wheelchairs. Whereas these systems prove to be less physically straining, there are several practical limitations that prevent regular use [9].

A more recent advancement in wheelchair technology is the pushrim-activated power-assisted wheelchair (PAPAW). A PAPAW can be seen as a hybrid between fully-powered wheelchairs and RHWs. The aim of PAPAWs is to prevent overuse complications, while maintaining a level of physical activity [10], increasing the social participation and independence of the user. They are equipped with an integrated electric motor mounted in the wheels or wheelchair frame [10] that is activated by pushing the handrims, as you would if you were propelling a RHW [11]. Previous research has shown that using a PAPAW can reduce heart rate [11-13] and rate of perceived exertion [7, 11, 14], lower metabolic cost [13, 15, 16], decrease shoulder range of motion during propulsion [13, 17], and increase distance travelled throughout the day [18, 19] when compared with RHWs.

However, handrim wheelchair propulsion is a complex motor task in which the bimanually applied forces determine both the speed and direction of locomotion [20]. Hence, PAPAWs might be more challenging to control since small interlimb differences are amplified by the wheelchair motor assist system, causing changes in direction. Therefore, the motor skill of the user is thought to be important for the effectiveness of PAPAW use. Yet, no research is available that examined the process of skill acquisition of PAPAW propulsion during the initial stages of motor learning.

Wheelchair propulsion is a cyclic motor task, which makes it possible to evaluate performance using energy consumption as a generic outcome measure of submaximal steady-state motor performance [21-23]. Skilled individuals will require less internal energy to produce the same amount of external power, this ratio is called mechanical efficiency. In this paper, we use the term 'assisted mechanical efficiency' for PAPAWs, since the external power output contribution of the motor and user cannot be disentangled.

An increase in this efficiency because of practice has already been demonstrated on ski-simulators, rowing-ergometers and in RHWs [21, 23-26]. Vegter and colleagues showed that healthy novices starting to learn handrim propulsion under steady-state low-intensity practice improve both efficiency and propulsion technique after bouts of 12-80 min practice [21, 27]. These time associated changes through practice have not yet been studied in PAPAWs.

Another commonly used measure in motor learning studies is the amount of endpoint variability [28]. In wheelchair propulsion on a motor driven treadmill this can be translated to the ability to steer the wheelchair where it is required to be steered (in speed and position), akin to a tracing task. This will be referred to as control. Wheelchair propulsion on a motor driven treadmill provides a mechanically valid [29] and representative environment [25]. It also requires the occupant to pay close attention in order to stay on the track given the limited length and width of the belt. Therefore, a second measure for skill is proposed, which consists of the ability to maintain more or less a steady position on the treadmill. It is expected that someone with a higher level of skill will show less position-related variability. Furthermore, RHWs are expected to show less variability than PAPAWs, since small interlimb differences in force application are not amplified by the power-assist motors.

As wheelchair propulsion is novel to persons in early rehabilitation after losing their walking ability and to many able-bodied participants, it allows for the use of able-bodied participants as a model to study the early acquisition of wheelchair propulsion proficiency. Additionally, able-bodied participants form a less heterogeneous group, allowing the study of motor learning in absence of impairment [21].

The aim of the current study is to explore the initial skill acquisition through low-intensity practice of able-bodied participants in PAPAWs and compare them to able-bodied participants that performed the same protocol with RHWs. To that end both groups performed five practice sessions consisting of three blocks of 4 min steady-state practice at 1.11 m/s and 0.21 W/kg on a motorized treadmill. After the practice

sessions, a crossover trial was performed to investigate the transfer from one mode to the other.

It was hypothesized that mechanical efficiency and control improves over time in both propulsion modes. It was also hypothesized that RHWs are easier to control but are more physiologically demanding. Moreover, a positive transfer of skill was expected between the two modes, since PAPAW propulsion is thought to be at least partly similar to RHW propulsion [30]. If the skills obtained during one mode (RHW or PAPAW) are not specific to that mode, a higher performance should be expected from someone that practiced in one mode than someone who has never used a wheelchair. However, if the skills obtained during practice are highly specific to a mode, a negative or zero transfer of skill could take place. Information on the transfer of skill between RHWs and PAPAWs could provide insight on the specificity of the mode.

Methods

Participants

A convenience sample of seventeen able-bodied male and seven female university students was drawn for this randomized controlled trial. The participants had no prior experience in wheelchair propulsion and did not have any contraindications for exercise. Participants were randomly allocated into one of two groups (RHW and PAPAW, Figure 7.1). Characteristics of both groups are given in Table 7.1. Participants were informed with an information letter about the research and provided written informed consent before taking part in the study. The study was approved by the Ethics Committee of the Centre for Human Movement Sciences Groningen (ECB/2016.01.15_1), University Medical Centre Groningen, University of Groningen, The Netherlands.

Protocol

All experiments were performed in a climate-controlled laboratory (20 degrees Celsius, 45% humidity) at the Centre for Human Movement Sciences, University Medical Centre Groningen, Groningen, The Netherlands on a level treadmill with a

Table 7.1. Participant characteristics (mean±SD) per group.

	RHW group (N=12)	PAPAW group (n=12)	p-value
Sex (m/f)	8/4	9/3	p=1.000†
Handedness (r/l)	10/2	11/1	p=1.000†
Age (yrs)	22 (±2.5)	22 (±3.3)	p=0.725‡
Height (m)	1.84 (±0.07)	1.82 (±0.08)	p=0.699‡
Weight (kg)	75.6 (±11.6)	77.4 (±11.6)	p=0.709‡

† 2-sided p-value of a Fisher's exact test. ‡ 2-sided p-value of an independent samples t-test.

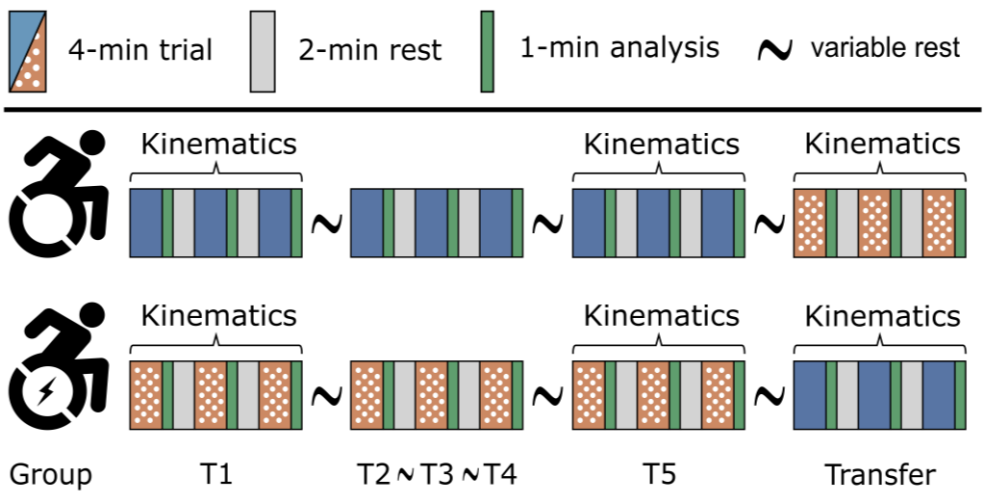


Figure 7.1 Participants were randomly assigned to one of two groups. One group practiced in a RHW (blue) and the other one in a PAPAW (orange/dotted). Participants practiced during five sessions, each consisting of three blocks of four minutes. The transfer took place after T5 and a 10-minute rest. Physiological measurements were performed at each session and kinematics were only collected during the first (T1), final (T5) and Transfer session.

2.0m long by 1.2m wide belt (Motek-Forcelink, Culemborg, The Netherlands). Each group practiced five times at a submaximal external power output of 0.21 W/kg bodyweight (Figure 7.1). Each of the five practice sessions (T1-T5) lasted for sixteen minutes with three blocks of four minutes of wheelchair propulsion (total training dose: 60 minutes) at 1.11 m/s (4km/h) and two breaks of two minutes in between. A resting period of approximately two days separated the individual sessions.

After the fifth session and ten minutes of rest, the wheels were switched and the participants performed one additional session of equal length, the transfer session, in the other mode. Participants received no help or feedback; the only instruction was to stay in the centre of the treadmill. Physiological measurements were performed during every session and participants were asked to rate their exertion after every session. Kinematics were only collected during T1, T5, and the transfer session. Data were collected during each practice block and the fourth minute of each block was used for analysis to ensure steady-state conditions.

Wheelchair

The power-assisted wheels used in the current study for the PAPAW mode were the commercially available 24-inch Twion wheels (Alber, Albstadt, Germany). These wheels contain a sensor in the handrim and an electric motor (60 Watt, 20Nm) that delivers power proportionally to the input of the wheelchair user. The wheels were

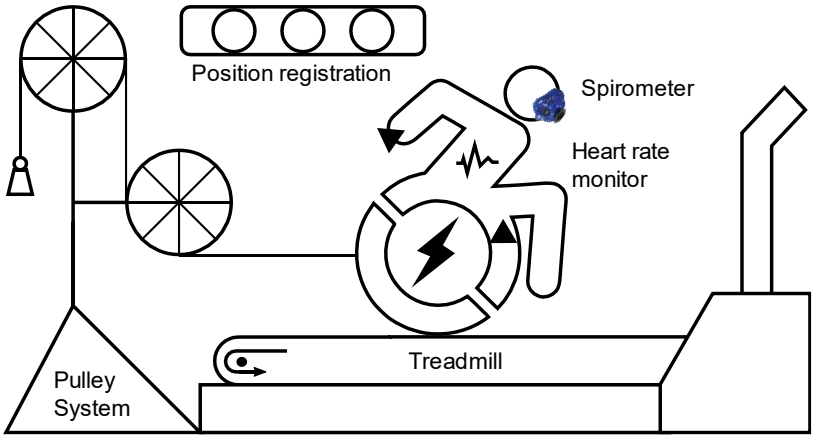


Figure 7.2 Experimental setup for all five sessions. The spirometer and heart rate monitor were used during every session. Position registration was only used during the first and the last session. Treadmill speed was 1.11 m/s and power output was set at 0.21W/kg with the pulley system. Black triangles exemplify the position of the two cluster markers. Modified figure from Vegter et al. [20].

mounted on a Küschall k-series wheelchair (Witterswil, Switzerland). In the RHW mode conventional 24-inch wheels were used on the same wheelchair. Tire pressure was checked before each session by the same researcher and kept constant at six bars (600 kPa). Anti-tip wheels were mounted to prevent backward falls.

(Assisted-) mechanical efficiency & energy expenditure

External power output was approximated with a drag test as described by Van der Woude and colleagues [9]. A weight on a pulley was attached to the wheelchair to ensure all participants performed at 0.21W/kg bodyweight external power output in both groups [31]. The added power output was calculated by adding the acting gravity force of the pulley weight to the force measured by the drag test multiplied by the speed of the treadmill.

Breath-by-breath open circuit spirometry was performed with a Quark CPET (COSMED, Rome, Italy). The internal energy expenditure (EE), assumed to be generated by glucose and lipid oxidation, and lipogenesis (W), was determined with the equation of Garby and Astrup [32].

(Assisted-) mechanical efficiency was derived from the ratio between the external power output (W) and the metabolic energy used internally during steady-state submaximal exercise and expressed as a percentage.

Heart rate & perceived exertion

Heart rate was also measured by the Quark CPET with a sensor placed across the chest area and a wireless connection. Additionally, participants were asked to give

their physical exertion a score between 6 and 20 on the Borg rate of perceived exertion (RPE) scale [33].

Control

An active-marker Optotrak system (NDI, Waterloo, Canada) was used to detect two cluster markers. Position data were sampled at 100Hz with two Optotrak modules, each containing three cameras. The first cluster marker was placed on the wheelchair to determine the position of the wheelchair with respect to the treadmill. The position of the wheelchair was represented by the centre of the rear axles. This was calculated by creating virtual markers on the right and left axle of the wheel in relation to the wheelchair cluster and averaging their position. Control was defined as the standard deviation in the right-left direction and the forward-backward direction of the wheelchair position. Additionally, push frequency was determined from the position of the second cluster marker, which was placed on the right hand of the subject.

Statistical analysis

All data were first processed and cut in MATLAB (Mathworks, Natick, USA) using custom-written scripts. Only the last minute of each four-minute block was used for analysis. The three separate four-minute blocks per session were averaged. Subsequently, outcome variables were exported to IBM SPSS version 23 (SPSS Inc., Chicago, USA). The level for statistical significance was set at 0.05 for all tests. All data were checked for deviations from a normal distribution. To evaluate the effect of practice, a mixed analysis of variance (ANOVA) was used with group and time as independent variables. Data on (assisted) mechanical efficiency, heart rate, RPE, push frequency, and treadmill position were compared for time and for group effects. If an interaction effect was found, a separate repeated-measures ANOVA was conducted for each group. This allowed for the analysis of time effects for each group separately. Finally, a MANOVA was used to check for between-group effects. An independent t-test with Bonferroni correction was used to compare T1 of one group with the transfer session of the other group.

Results

No differences were found in the characteristics of the participants (Table 7.1). All participants completed the protocol at a mean external power output of 17W, though some safety stops were necessary to ensure their safety in the initial bouts, this predominantly happened in the PAPAW group during T1 in the first minute ($\bar{x}$RHW=0.30 stops, $\bar{x}$PAPAW=3.25 stops). No emergency stops were needed after T1 for the RHW group and after T4 for the PAPAW group. During the transfer of the RHW group to the PAPAW mode some emergency stops were needed ($\bar{x}$=0.92

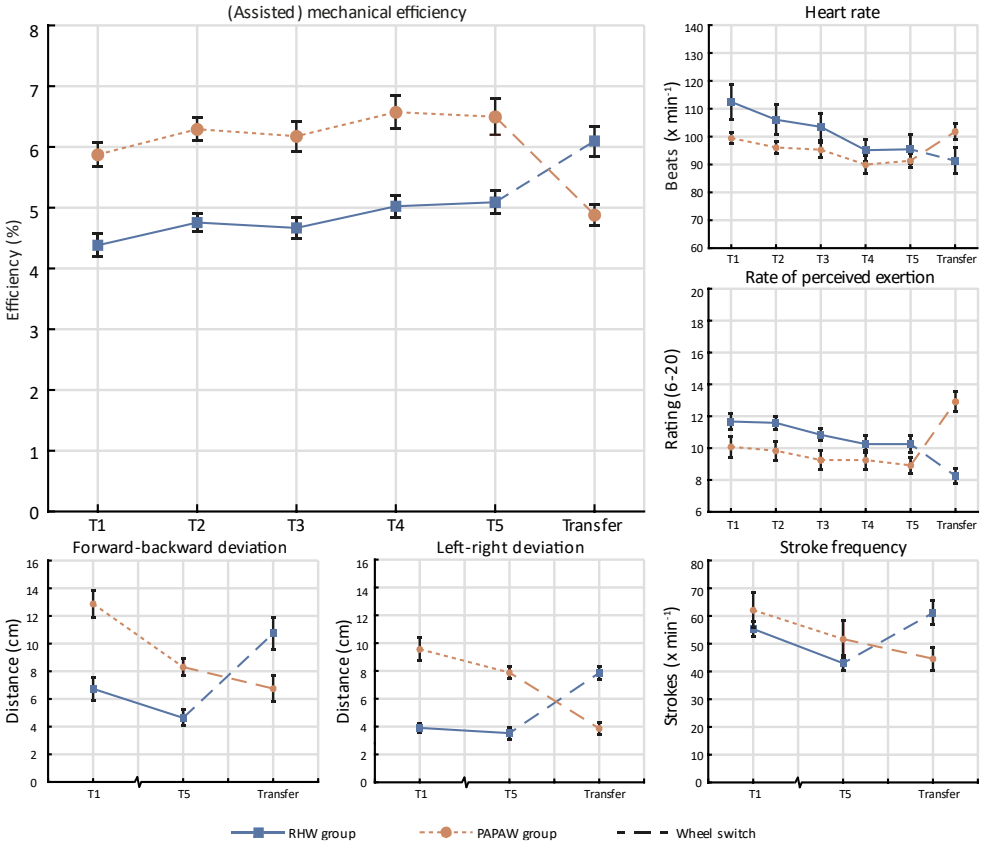


Figure 7.3 Average (assisted) mechanical efficiency, heart rate, RPE, left-right and forward-backward deviation, and stroke frequency of each session for the RHW group (blue, n=12) and the PAPAW group (orange/dotted, n=12). Standard error bars are given for each session.

stops). There were no adverse incidents resulting in injury. If a safety stop was made the protocol was resumed from that point on. Kinematic data were missing on three occasions and could not be included in the analysis. Results (Figure 7.3, Table 7.2) and statistical outcomes (Table 7.3) for (assisted) mechanical efficiency, heart rate, RPE, stroke frequency and position on the treadmill will be discussed below.

Mechanical efficiency & energy expenditure (T1-T5)

(Assisted) mechanical efficiency significantly increased over time (Table 7. 2, Figure 7.3) for both groups (ΔRHW.T1-T5=0.71%, ΔPAPAW.T1-T5=0.62%). A post-hoc analysis with simple contrasts showed that T1 significantly differed from T5 in both groups. Also, a between group difference was found with a higher mechanical efficiency for the PAPAW group compared to RHW (Δ=1.21% at T5), however no significant interaction effect was found.

Energy expenditure (EE) is closely tied to mechanical efficiency as external power output was standardized. A significant reduction in EE was observed over time for both groups (ΔRHW.T1-T5=-57W, ΔPAPAW.T1-T5=-30W), this was corroborated by a post-hoc analysis with simple contrasts between T1 and T5. Additionally, EE was significantly lower in the PAPAW group than in the RHW group (Δ=68W at T5). No interaction effect for EE was found.

Heart rate & perceived exertion (T1-T5)

An interaction effect between time and group was found for heart rate, where heart rate reduced more over time for the RHW group than for the PAPAW group (ΔRHW.T1-T5=16.5 bpm, ΔPAPAW.T1-T5=8.2 bpm). Therefore, a separate repeated-measures ANOVA for each group was conducted so time effects could be isolated for individual groups. Heart rate significantly reduced over time in the RHW group and in the PAPAW group. Post-hoc analysis with simple contrasts revealed that T1 significantly differed from T5 in both groups. A separate between-group analysis was conducted with a MANOVA and did not reveal a significant difference between groups.

Participants rated their physical exertion significantly lower over time (ΔRHW.T1-T5=1.4, ΔPAPAW.T1-T5=1.1). Moreover, the PAPAW group had a significantly lower RPE than the RHW group (Δ=1.4 point on T5). Post-hoc analyses with simple contrasts revealed that participants scored significantly lower on RPE during T1 when compared with T5 and no interaction effects were found for RPE.

Control (T1-T5)

Deviation of the forward-backward position on the treadmill contained an interaction effect as the PAPAW group showed a greater reduction over time, therefore, separate repeated-measures ANOVAs were used to compare the effect of time. It was found that the forward-backward deviation significantly reduced over time for the PAPAW group and for the RHW (ΔRHW.T1-T5=1.4, ΔPAPAW.T1-T5=1.1). Furthermore, a MANOVA, with follow-up ANOVAs revealed that the forward-backward deviation was significantly higher for the PAPAW group than the RHW group during T1 and T5.

Left-right deviation of the treadmill position significantly lowered between T1 and T5 for both groups. The PAPAW group had a significantly higher left-right deviation than the RHW group and no interaction effect was found. Stroke frequency significantly lowered over time for both groups (ΔRHW.T1-T5=12.5, ΔPAPAW.T1-T5=10.4). Additionally, stroke frequency was found to be lower in the RHW group (Δ=8.7 strokes x min-1 at T5), however this difference was not statistically significant. No interaction effects were found.

Table 7.2. Descriptive statistics (mean±sd) for (assisted) mechanical efficiency (ME), energy expenditure (EE), heart rate (HR), rate of perceived exertion (RPE), stroke frequency (stroke), left-right (Dev L-R) and forward-backward (Dev F-B) deviation in the RHW group (n=12) and the PAPAW group (n=12).

Variable	Group	T1	T2	T3	T4	T5	Transfer
ME (%)	RHW	4.38 (0.67)	4.76 (0.53)	4.67 (0.62)	5.02 (0.63)	5.09 (0.66)	6.10 (0.85)
	PAPAW	5.88 (0.67)	6.29 (0.66)	6.18 (0.85)	6.57 (0.96)	6.50 (1.02)	4.88 (0.59)
EE (W)	RHW	380 (75.1)	349 (73.8)	353 (53.4)	329 (58.6)	323 (59.0)	272 (51.8)
	PAPAW	285 (48.5)	265 (41.1)	271 (42.1)	255 (44.9)	255 (45.0)	345 (67.0)
HR (bpm)	RHW	112 (22.0)	106 (18.7)	104 (16.4)	95.1 (13.5)	95.5 (17.8)	91.3 (16.0)
	PAPAW	99.5 (6.70)	96.1 (7.73)	92.3 (9.50)	90.0 (10.9)	91.3 (8.99)	102 (9.95)
RPE (6-20)	RHW	11.7 (1.67)	11.6 (1.38)	10.8 (1.34)	10.3 (1.91)	10.3 (1.91)	8.25 (1.54)
	PAPAW	10.1 (2.23)	9.83 (2.04)	9.25 (2.01)	9.25 (2.10)	8.92 (1.78)	12.9 (2.15)
Stroke (n x min^{-1})	RHW	55.3 (9.90)				43.0 (9.30)	61.1 (14.3)
	PAPAW	62.1 (21.6)				51.7 (21.7)	44.6 (12.9)
Dev F-B (m x 10^{-2})	RHW	6.73 (2.82)				4.63 (2.03)	10.7 (3.78)
	PAPAW	12.9 (3.33)				8.30 (1.94)	6.75 (3.03)
Dev L-R (m x10^{-2})	RHW	3.91 (1.12)				3.54 (1.49)	7.87 (1.56)
	PAPAW	9.57 (2.92)				7.90 (1.51)	3.88 (1.42)

Table 7.3. Statistical outcomes for (assisted) mechanical efficiency (ME), energy expenditure (EE), heart rate (HR), rate of perceived exertion (RPE), stroke frequency (stroke), left-right (Dev L-R) and forward-backward (Dev F-B) deviation in the RHW group (n=12) and the PAPAW group (n=12).

Variable	Group	Interaction-effect		Group-effect		Time-effect		Pre-post[b]	Transfer[f]
		F(df)	p	F(df)	p	F(df)	p	p	p
ME (%)	RHW	0.07(4,88)	0.990	39.1(1,22)	**<0.001**	7.26(4,88)	**<0.001**	**0.003**	0.242
	PAPAW								0.032
EE (W)	RHW	1.44(4,88)	0.227	15.7(1,22)	**0.001**	8.62(4,88)	**<0.001**	**0.001**	0.278
	PAPAW								0.125
HR[a] (bpm)	RHW	2.60(4,88)	**0.041**	1.16(5,18)	0.368[c]	14.6(4,44)	**<0.001**	**<0.001**	0.063[d]
	PAPAW					4.24(4,44)	**0.005**	**0.022**	0.071[d]
RPE (6-20)	RHW	0.45(4,88)	0.770	4.92(1,22)	**0.037**	7.02(3,59)	**0.001**	**0.032**	0.023[e]
	PAPAW								0.076[e]
Stroke (n x min^{-1})	RHW	0.17(1,22)	0.687	1.38(1,22)	0.253	32.1(1,21)	**<0.001**	**<0.001**	0.448
	PAPAW								0.020
Dev F-B[a] (m x 10^{-2})	RHW	6.73(1,22)	**0.017**	11.6(1,23)	**<0.001**[c]	10.3(1,11)	**0.008**	**0.008**	0.082
	PAPAW					26.2(1,11)	**<0.001**	**<0.001**	0.495
Dev L-R (m $x10^{-2}$)	RHW	3.67(1,22)	0.068	40.4(1,22)	**<0.001**	7.04(1,22)	**0.015**	**0.042**	0.048[d]
	PAPAW								0.474

a Separate repeated-measures ANOVAs were performed for each group.
b Tested with simple contrasts.
c Tested with a MANOVA with follow up ANOVAs.
d Equal variances not-assumed
e Mann-Whitney U
f One-sided p-value (α=0.004)

Transfer

Mechanical efficiency was higher for the PAPAW group during their transfer session than for the RHW group during their first session (Δ=0.5%), however, these differences were not statistically significant. Similar results were found for EE (Δ=-35W), heart rate (Δ=-10 bpm), RPE (Δ=-1.2), and stroke frequency (Δ=-10.7 strokes x min-1), but these were also not significant.

Additionally, the RHW group had a lower heart rate (Δ=-7.9 bpm), RPE (Δ=-1.9), left-right (Δ=-2.2 m x 10-2) and forward-backward (Δ=-1.7 m x 10-2) deviation during their transfer session than the PAPAW group during their first session, however, none of these results reached statistical significance.

Discussion

The aim of the current study was to explore early skill acquisition through practice of able-bodied participants in PAPAWs and compare them to able-bodied participants that performed the same protocol with RHWs. Learning effects were found for both groups, as both increased their (assisted) mechanical efficiency, reduced their heart rate, RPE, their right-left and forward-backward deviation on the treadmill, and had a lower stroke frequency over time.

Assisted mechanical efficiency in the PAPAW increased from 5.8% to 6.5%. These values are lower compared to those of Arva [16] and colleagues (9.9-20.6% at different speeds/power outputs), but higher than the findings of Pavlidou [34] and colleagues (4.2%). (Assisted) mechanical efficiency rises as external power output increases [16], which might explain the differences between our study and the study of Pavlidou and colleagues. However, Arva and colleagues had participants perform at a lower power output and still found a higher mechanical efficiency. Of course, the use of different amounts of power assist from the different PAPAWs could be the cause of this. For future studies, it seems important to consider that PAPAW propulsion, in addition to RHW propulsion, changes through learning.

Mean mechanical efficiency in the RHW mode was found to be slightly lower (5.1%) than in other studies (5.5-7.0%) with able-bodied subjects [21, 27, 35] and also increased over time. One possible explanation is that the way external power output is determined between different studies influences the outcomes. When external power output is determined from measurement wheels the effect of movement variability on the treadmill can also be included, increasing the observed external power output. In the current study the power output was determined from a drag-test, which determines the power output required for moving in a straight line at a

constant velocity. In a paper of Vegter et al. where two measurement wheels and the drag test were compared an underestimation of 14% was found [20].

In general, PAPAW propulsion was more efficient than RHW propulsion. As external power output was constant during this study, a higher mechanical efficiency also corresponds with a reduction in EE. The lower EE [7, 13, 15, 16, 34, 36] is in accordance with other studies that compared RHWs with PAPAWs. Likewise, RPE was found to be lower for the PAPAW group and similar results were found in other studies [7, 11, 14]. Correspondingly, heart rate was also expected to be lower for the PAPAW group. Mean heart rates were in the expected direction, but due to the high variance between subjects no statistically significant difference could be found, which is in contrast with previous studies [11-13, 15, 34].

Unfortunately, it is not clear what proportion of the reduced energetic cost of PAPAW propulsion in comparison with RHW propulsion is due to the contribution of the motor. For example, it could be that the technique of the participant is less efficient, but the added power of the motor is enough to compensate for that. The assist provided by the wheels is of proportional grade with no exact knowledge of the algorithm of the assist. The assist being proportional means that the torque provided by the motors during propulsion is dependent on the propulsion characteristics of the user. Knowledge on the algorithm of the assist and the propulsion characteristics of the user are of great importance to objectively determine whether the mechanical efficiency is high or low compared to other studies. The motor power is rated at 60W per wheel, considering the low intensity required for this task (17W, fixed speed and direction) it can be stated that the motors do not use their full potential. Although it is good to see that the user needs to stay somewhat active. Different task constraints might give different results due to their interaction with the control algorithm.

The control parameters, left-right and forward-backward deviation, were also found to be higher in the PAPAW group than in the RHW group. This was in line with the expectation that the PAPAW would be harder to control, especially considering the proportional control of the power assist wheels. This finding is in agreement with other authors that have already stated that a PAPAW may be helpful for specific tasks that require more torque, but the PAPAW is less suited for tasks that require a high amount of control [10, 37]. However, user control did improve over time, indicating that some practice is needed to properly use a PAPAW and evaluate the impact of a PAPAW. Clinicians and dealerships should therefore consider a trial period for potential new users. As the factor of control is not often used in research it is still unclear how different PAPAWs vary in their controllability and what this

effect means for the user in terms of efficiency and ease of use during activities of daily living. Standardized measures like the one proposed in this study could be used to quantify the control over the wheelchair. However, to be able to generalize results to real life propulsion the task might need to be expanded (e.g. with obstacle avoidance).

Stroke frequency reduced over time in both groups and the PAPAW group used a higher stroke frequency than the RHW group on average [17], however, this difference was not significant. Ideally, the PAPAW would require fewer pushes to maintain the same velocity as the motors continue to engage for a short period of time after release. A decrease in stroke frequency over time could show that less adjustments have to be made to maintain a steady position on the treadmill. Moreover, it could also explain why mechanical efficiency in this study was lower than in other studies for the PAPAW, as other studies have also shown that an increase in movement amplitude and a decrease of movement frequency reduces the energy cost of a task [23, 24, 38, 39]. The lack of control in the PAPAW group is the most likely explanation for the higher stroke frequency. Maintaining a straight line on the treadmill, when not in control, urges the user to apply corrections during propulsion resulting in a higher stroke frequency. This study did not evaluate whether these pushes were unilateral (correction) or bilateral. In this experiment, the wheelchair was not attached to the treadmill with a rail or a fixation system. In experiments where such a system is used the factor of control becomes less important, this might give an unfair advantage to PAPAWs in some comparison studies.

As PAPAW propulsion is similar to regular handrim wheelchair propulsion, a skill transfer between these two modes was hypothesized to exist [40]. From a learning perspective, a positive influence on mechanical efficiency might have advocated the use of PAPAWs until a sufficient level of skill is reached. A positive influence in control could have pointed towards the use of RHWs in early practice in order to gain sufficient control before transferring to the more demanding (in terms of control) PAPAW mode. Tendencies of a positive transfer effect were observed in the data, though no statistically significant effects could be found. For now, based on the zero transfer, it is recommended that participants practice or receive training in a new mode before choosing to settle with a configuration. Transfer tendencies were also observed for RPE, where prior experience in a RHW or PAPAW led to a non-significant (after Bonferroni correction) difference in perceived exertion. Even though no statistically significant difference was found, researchers and clinicians should still beware of changes in perception due to prior experience or a response bias.

It should also be noted that there were some limitations in this study. Although significant between-group differences for left-right and forward-backward deviation were already found, the values for left-right and forward-backward deviation are conservative for the PAPAW group, with an average of three emergency stops during the first session. The data during an emergency stop was not used, but an emergency stop shows a high amount of deviation and a high stroke frequency, this would have influenced the outcome for the PAPAW group during the first session. During the transfer session of the RHW group there were fewer emergency stops. It may be that the RHW group was more skilful through their previous experience, but the statistical analysis failed to support this finding.

Moreover, the control task could have been unclear for some participants. We instructed our participants to 'stay in the centre of the treadmill as best as they could'. A projected target on the treadmill could have helped participants in staying in the centre of the belt. These targets can then also be used to produce a more generalizable task (e.g. with obstacle avoidance). Furthermore, large individual differences between participants were observed. Recent research has shown that individual motor learning differences are important to take into account in motor learning studies [21, 27]. In this study, it appeared that some participants were able to capitalize more on the benefits of a PAPAW than other participants, as indicated by the high variability between participants. However, with the current sample size and study design it was not possible to analyse these individual differences.

Finally, mechanical efficiency, RPE, stroke frequency, and right-left deviation did not show any interaction effects, indicating that learning rate between the two modes might be similar. The difference in difficulty between the two modes could have led to differences in motor learning strategies. However, without knowledge of the kinetics of the user and the algorithm of the PAPAW we cannot make inferences about the underlying motor learning principles of (assisted) wheelchair propulsion. Future research into the kinetics and kinematics of PAPAW propulsion is needed to provide more insight in the adaptations that are made during the learning process.

Conclusions

The present study showed that the energetic cost of regular handrim and power-assisted wheelchair propulsion can be lowered through low-intensity practice in users with no previous experience. Power-assisted wheelchair propulsion requires less energy than regular handrim wheelchair propulsion, but power-assist wheels are more difficult to control. Based on these results, users should be able to practice with a power-assisted wheelchair before evaluating whether the user can benefit

from the technology. Additionally, a tendency was found indicating that the use of subjective scales for exertion might be influenced by the previous experience of the user, which should be taken in consideration when presenting a power-assist wheel to a new user. Additional research on the effects of practice with assistive technology is needed to improve efficient use and design. More knowledge about effective power-assist algorithms is needed, especially with respect to the control needs of the user. These studies would benefit from kinetic data measured at the handrim and information on the power-assist algorithms.

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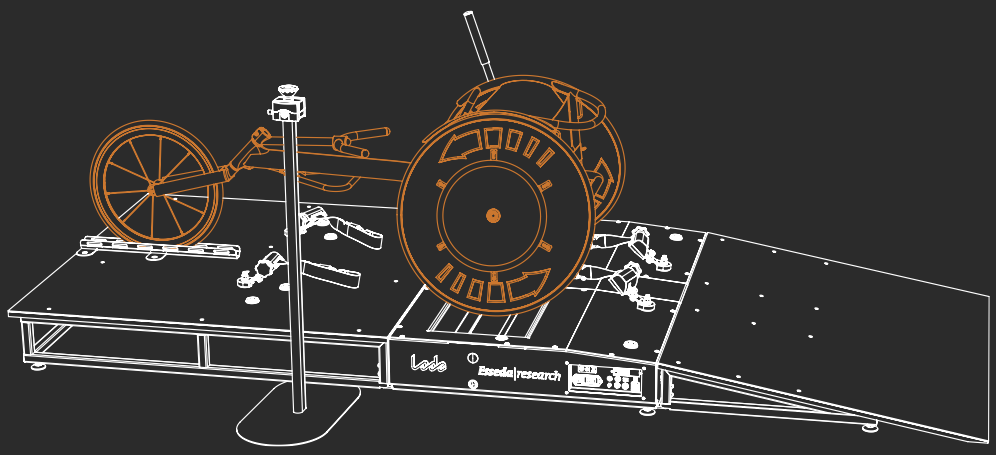
8

General discussion



Development of a wheelchair propulsion laboratory

The current thesis positions wheelchair ergometers as a central component for wheelchair propulsion analysis and, by extension, the wheelchair propulsion lab. It is well established that long-term use of handrim wheelchairs can lead to repetitive strain injuries in the upper extremities [1–3], underlining the importance of wheelchair propulsion monitoring and research. Wheelchair propulsion laboratories (labs) to study the physiological, biomechanical, ergonomic, and technical underpinnings of wheelchair propulsion to optimize the wheelchair, user, and wheelchair-user interface are, however, not yet widely implemented. The upcoming section will discuss the past (main findings of the thesis), as well as the present (current state of the ergometer and wheelchair propulsion labs) and future (opportunities and challenges ahead) of the development of a wheelchair propulsion laboratory.

Main findings

The current thesis highlighted the rich history, starting in the sixties, of wheelchair ergometers (Chapter 1) that were developed to research and optimize all components of the wheelchair-user combination [4]. Researchers around the globe have built devices that allowed for the measurement and simulation of wheelchair propulsion in the lab, each with their own trade-offs for their specific research needs. Nevertheless, wheelchair ergometry has never been as widely implemented as bicycle and arm-crank ergometry did in the sports domain and clinical gait analysis [5] in the clinical domain.

The current thesis presents a new wheelchair ergometer for the physiological and biomechanical testing of wheelchair users in their own wheelchair, thereby maintaining the individual wheelchair-user interface (Chapter 2). The ergometer uses force sensors and servomotors to respectively measure and simulate wheelchair propulsion, allowing for the use in (applied) research, as well as clinical and sports practice. With regards to the measuring aspects of the ergometer, it produces similar informative outcomes (e.g. push-time, cycle-time, contact angle) as a measurement wheel during submaximal steady-state propulsion (Chapter 2). However, the unfiltered force signal does seem to have a relatively high uncertainty (Chapter 3).

A limiting factor to the adoption of wheelchair propulsion labs, for example in rehabilitation [6], have been the complexity of measurements (e.g. power output) and the interpretation of test outcomes [7]. As such, the current state-of-the-art of laboratory wheelchair research methodology regarding overground, treadmill, and ergometer propulsion was compiled and presented in an accessible tutorial format for end-users (online video and Chapter 4). With regards to the simulation that the

ergometer provides, Chapter 5 shows that the biomechanical outcomes measured during overground, treadmill, and ergometer propulsion are comparable (e.g. heart rate, push time, work per push), though some differences were found (e.g. fraction of effective force) which should be further explored. These results are important to consider when comparing (future) research with the existing knowledge base and can be used to further improve simulation and ecological validity.

Chapter 6 showed that wheelchair racing propulsion technique in novices changes over time for the better when practicing submaximal steady-state propulsion on the ergometer. While these results by themselves are novel and valuable, they also demonstrate that the ergometer can be employed for longitudinal and/or learning studies. Moreover, Chapter 7 illustrated the challenges associated with conducting motor learning studies on a wheelchair treadmill, thereby demonstrating potential avenues for the application of wheelchair ergometers as a testing surface and perhaps in the simulation of power-assists. As such, the ergometer proves to be a promising addition to the wheelchair propulsion lab, allowing for wheelchair propulsion testing and training in research, clinical, and sports practice.

Current state-of-the-art

Wheelchair propulsion labs could be used to determine the cause of atypical or asymmetrical propulsion and exercise capacity (both aerobic and anaerobic), aid in the planning of treatment or training, prescribe and fit (power-assisted) wheelchairs, test building configurations, identify those at risk for overuse complications, classify athletes, and expand the knowledge base. To enable wheelchair propulsion analysis, a new wheelchair roller ergometer (Esseda, Lode BV, The Netherlands) was developed with the aim to be the main testing platform in a wheelchair propulsion lab (Chapter 2), while preserving the wheelchair-user interface, and allowing for the use of different simulation parameters based on the power balance [8]. Not shown in the thesis were the many design iterations that predated the current version of the ergometer. Further, developments are still ongoing today and effort is primarily targeted towards improving software-based end-user (i.e. researchers, clinicians, coaches) experience.

For the wheelchair ergometer to become a central component in the wheelchair propulsion lab, it will need regular static and dynamic calibration. A low-cost portable validation system for wheelchair roller ergometers was presented (Chapter 3) which needs slight adaptations before it can be deployed in the field. First, the improvised tachometer should preferably be exchanged for another (off the shelf) sensor, because eccentricities in the disc significantly complicated the analysis for a minimal reduction in cost. Second, the control and analysis software need to be

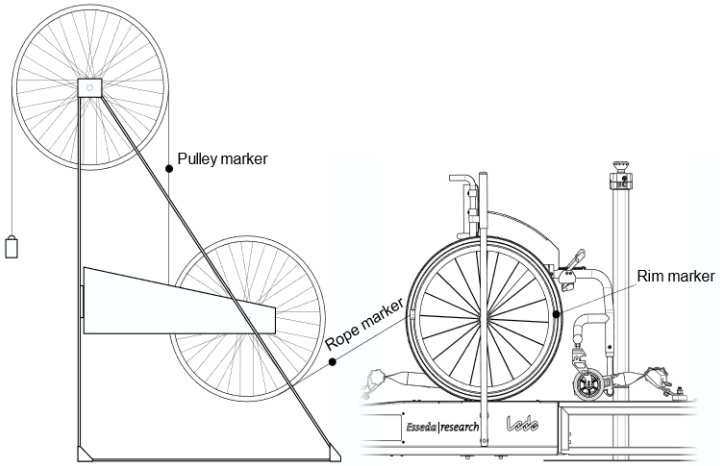


Figure 8.1. A schematic drawing of a pulley and marker setup to apply a known force on the handrim.

further developed so assemblers and distributors can individually perform measurements and calibrations without the need for additional scripting. To allow for a more realistic loading of the rollers, a larger version of the calibration system could be built using the same guiding principles, though this would come at the cost of portability. Thereafter, the calibration system could be used to validate roller ergometers in a variety of settings. Calibration with this system could for example be performed right after assembly, periodically by the distributor, or by the end-user to monitor the consistency of an ergometer.

End-users could also use existing equipment in the wheelchair propulsion lab to verify the status of their ergometer. A simple pulley-based system (Figure 8.1) could be used to apply a known sinusoidal force to the handrim of a wheelchair on the ergometer. This method, originally developed by Leonard Rozendaal at the Free University of Amsterdam for their ergometer (see [9]), only requires an ergometer with an isospeed mode, a pulley system, a motion tracking system, and a series of calibrated weights, all of which are usually present in a wheelchair propulsion lab. It allows for the evaluation of an ergometer in a highly realistic situation, with a loaded wheelchair (e.g. with a dummy), and an applied periodic 3D force of similar magnitude as seen in regular handrim wheelchair propulsion. Moreover, it could be used to apply force at different angles which would also allow for the dynamic evaluation of measurement wheels in multiple dimensions.

Furthermore, a new static calibration arm (Figure 8.2) that can be used to calibrate the force sensor of an assembled ergometer was recently developed (by Umaco BV,

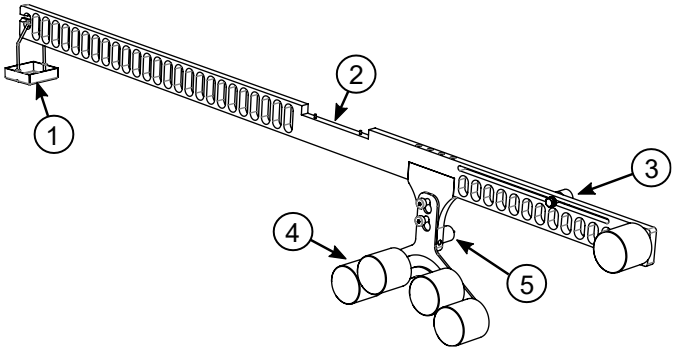


Figure 8.2. The new static calibration arm (lengths in mm). 1) The basket for the M1 calibrated reference weights, 2) mounting spot for the angle sensor, 3) tare weight, 4) compensation weight, 5) roller attachment.

The Netherlands). The calibration arm can be attached to the outside of a roller and is raised, ensuring enough clearance with other ergometer parts and the ground. Weights were added on the bottom to counteract the offset in the centre of rotation. A calibrated angle sensor, which is used to compensate for angular offsets, can be mounted to the top of the arm. Finally, a basket is used to hold the M1 calibrated weights that are used as reference to determine the gain and offset of the loadcell. The development of this new mobile calibration arm is an important step in the life-cycle management of the ergometer as it enables distributors to perform periodic calibrations in the field, thereby ensuring consistent results over time.

Composing wheelchair propulsion laboratories

Wheelchair propulsion labs can be used for scientific research, but should also serve as expertise centres to generate and implement knowledge for clinical and sports purposes [7]. The concept and even the implementation of wheelchair propulsion laboratories is certainly not new. Protocols and guidelines were for example provided by the SMART^{Wheel} User Group (SWUG, [10]), though the wheels themselves are no longer commercially available. More recently, De Groot and colleagues [11] presented their vision of a wheelchair propulsion lab using, among others, measurement wheels and a wheelchair treadmill (Wheelchair Expert Evaluation Laboratory – implementation; WHEEL-i). In a recent application of the WHEEL-I protocol in rehabilitation practice, Leving et al. [6] already noted that the proper selection of technology may be key to generating standardized and clinically meaningful outcomes in a reasonable amount of time. They recommend the use of an ergometer, as it can be pre-programmed to perform standardized protocols and automatically generate standardized outcomes.

Indeed, the current thesis is aligned with WHEEL-i but proposes to replace the treadmill and measurement wheels by a roller ergometer in the wheelchair propulsion lab. This could be an important step in the dissemination of wheelchair propulsion labs and practitioner-accessible technology, but there are also some remaining challenges in the transfer from the academic to a clinical and sports environment. Implementation typically involves a multidisciplinary team of consultants, physiotherapists, embedded scientists, and other technical staff that will use their experience and expertise to make an assessment based on collected data and reference data. As such, the interpretation of test outcomes and the availability of reference data are challenges that still need to be addressed [10], though some work has already been done in this area [7,12]. Moreover, data analysis and interpretation are typically time consuming because of the integration of multiple measurement devices and the complexity of outcomes, and should therefore be facilitated as much as possible with dedicated software. In this regard, the software is equally -if not more- important than the hardware. Clinicians and coaches are often not familiarized with the complex biomechanical outcomes obtained in the lab, so outcomes need to be preselected, pre-processed and analysed, and presented in a proper format. For example, the SWUG protocol uses velocity, average peak resultant force, push frequency, and stroke length as its key outcome measures, and provides standardized visualizations and a flowchart to guide decision making by clinicians [10]. It is also helpful to see a synchronized video [7] of the actual test performance next to the other outcomes to help visualize the impact of certain parameters (for an example see Figure 8.3). Based on these considerations, a software platform that integrates outcomes from all equipment could be an important asset for wheelchair propulsion labs.

To create a complete overview of a wheelchair-user combination, additional information such as metabolic data, 3D kinematics, regular (2D) video, and electromyography (EMG) might also be needed, depending on the specific (research) questions at hand. Future studies should examine which additional equipment is needed for different research questions and (clinical) scenarios. For example, the studies in the current thesis often included a spirometer in the measurement setup as it allows for the determination of energy cost and the calculation of gross mechanical efficiency (Chapter 5, 6, and 7) as a measure of propulsion skill [13–15]. Another example of an extended measurement setup including the ergometer can be found in a recent study by Briley et al. [16] where scapular kinematic variability and its effect on shoulder pain is examined. For kinematics, it would be beneficial if markerless time-of-flight sensors could be used instead of marker-based systems as they require much less time to set up.

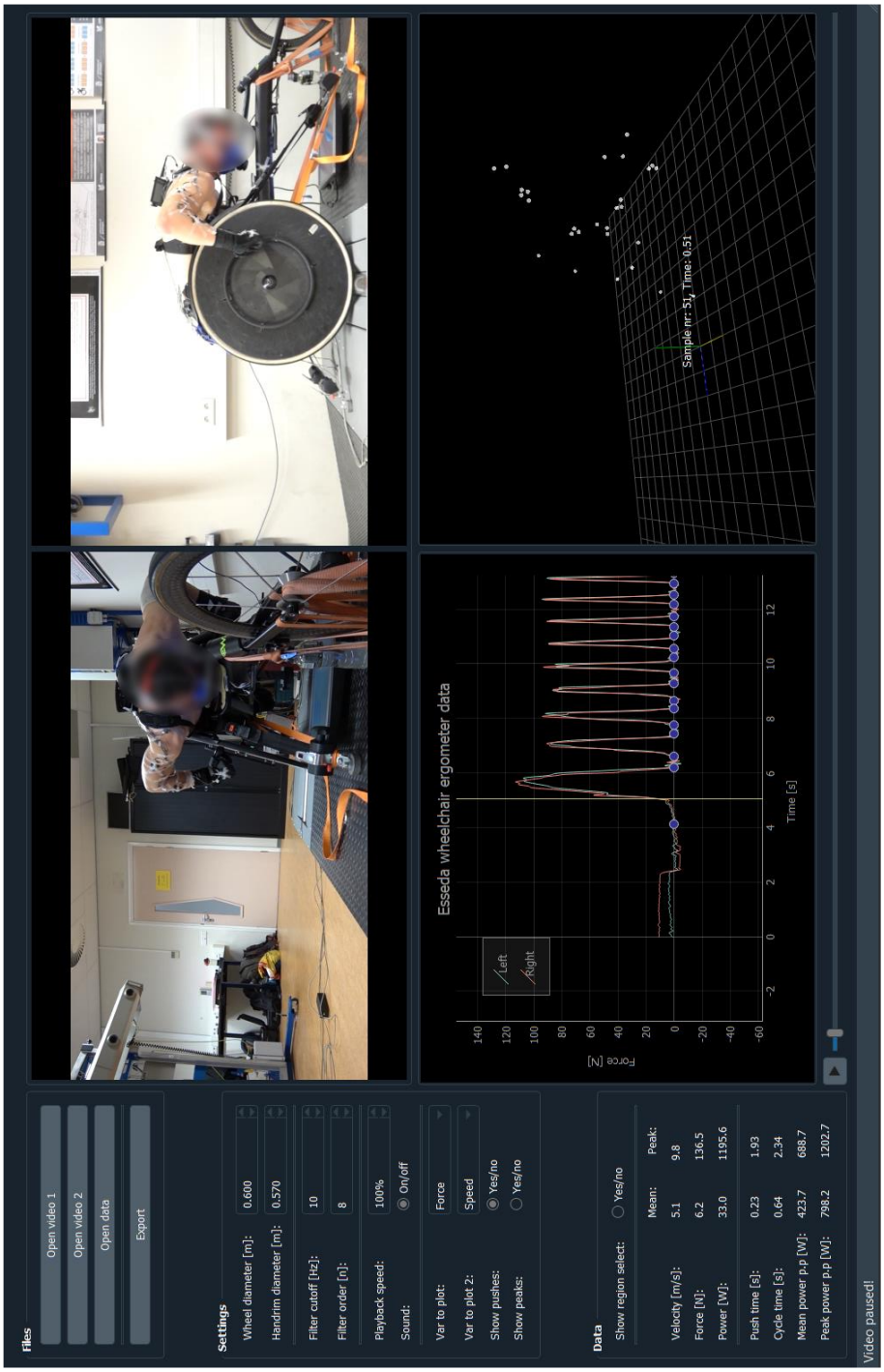


Figure 8.3. A feedback system for clinicians and coaches with synchronised data and video.

Measurement wheels are still required for the determination of 3D forces and thus for advanced inverse-dynamics modelling of the shoulder complex (e.g. [17–19]). Yet, information on glenohumeral joint reaction forces is not always needed and can be predicted to some extent using less information [20]. An ergometer is also not suited for tasks that heavily involve turning and/or balancing the trunk as a task element, as is common in manoeuvrability tests like the Spider test [11], Butterfly-sprint test [21], or the Illinois test [22]. The inertial forces during these tests could be simulated at the level of the handrim (they are currently not), but their effects on the wheelchair-user combination are difficult to simulate and would, for example, require a Stewart (i.e. 6-DoF) platform [23]. Still, overlooking these edge cases, a roller ergometer can meet the majority of testing demands without the need for additional equipment.

The Esseda wheelchair ergometer provides a realistic simulation of the frictional and inertial forces encountered during handrim wheelchair propulsion. However, due to the stationary nature of ergometer-based testing, the visual feedback for the participant is limited. The ecological validity of lab-based testing could be improved by adding contextual richness through scenario and context simulation while maintaining the overall benefits of testing in the lab [24,25]. The combination of physical exertion through the kinetic feedback of the wheelchair ergometer with a virtual world could provide a unique, comprehensive and realistic experience. Simulated visual feedback presents the user with the opportunity to engage in environments that are similar to real-world objects and events [26], while still harnessing all of the advantages of testing on the ergometer. Therefore, a coupling between the ergometer and a visual world, in this case a gym, is proposed (Figure 8.4). This virtual environment can then be displayed on a screen or on a head-mounted display. Another possibility is to use this coupling to add elements of gamification [27] or biofeedback [15].

The role of lab-based testing

Human movement consists of much more than just the physical ability and skills of an individual [28]. It is, instead, the result of a complex interplay of sociological, environmental, psychological, and health-related factors, where the environment can prove to be a limiting or facilitating factor with respect to one's moving pattern, which is especially true for wheeled mobility [29]. The applicability of results obtained from wheelchair propulsion analysis outside the research environment is therefore also dependent on the relevance of those measurements to the daily life and their ability to provide functional information. Consequently, there also have been ongoing efforts to enable measurements outside the lab [30], which has recently picked up momentum due to the increasing availability of low-cost, reliable, easy-

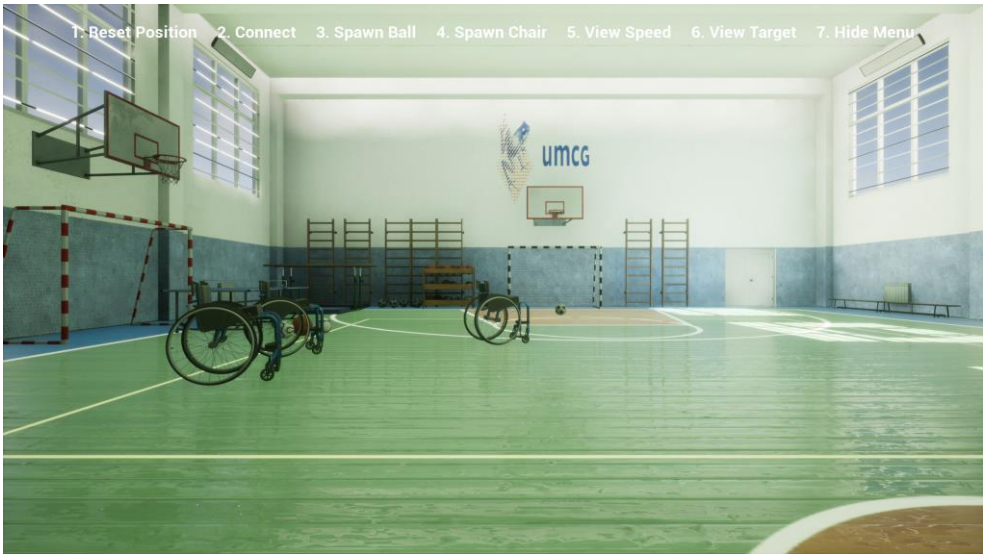


Figure 8.4. Virtual reality environment built in the Unreal Engine controlled by the ergometer.

to-use mobile measurement equipment [31,32], such as Inertial Measurement Units (IMUs). IMU based methods tend to have great ecological validity because they measure the user within their specific context [31,33]. The opportunities presented by these devices are almost limitless and call in a new era of research and treatments.

Yet, this does not mean that these field-based methods will replace lab-based testing. Instead, both lab- and field-based testing have become more important than ever, as research can greatly benefit from combining lab- and field-based measurements. The wheelchair propulsion lab should therefore employ a holistic view of wheelchair propulsion. A comprehensive portfolio of "wheeled mobility" should thus contain lab- based outcomes and functional outcomes determined with wheelchair skills tests [7]. The first can be determined on a wheelchair ergometer (e.g. using the protocols presented in Chapter 4), and the latter could be quantified using IMUs. The Centre for Human Movement Sciences (University Medical Centre Groningen, The Netherlands) is also working on standardized testing and the integration of field-based with lab-based testing in the WheelPower consortium [34]. A project that was launched in 2020 for adapted sports, but is equally relevant for clinical practice as was shown by Leving et al. in early rehabilitation [35].

A wheelchair propulsion lab could also be used to monitor or facilitate the motor learning process for handrim wheelchair propulsion. Two examples of natural learning studies are given in Chapter 6 and 7 for wheelchair racing and power-assisted propulsion, respectively. In line with previous research, an improvement in

mechanical efficiency was seen on both the ergometer and the treadmill [36], even though the task complexity on the ergometer is arguably lower. The current thesis examined natural learning, but the learning process could also be facilitated by the use of biofeedback [15,37]. A target value that is thought to reduce physical strain could be shown on a screen in front of the participant to allow for targeted improvements on selected propulsion metrics (e.g. increasing contact angle or reducing peak forces) [37]. Another target for optimization is the wheelchair-user interface. Regular handrim wheelchair have a multitude of different settings that can be configured (e.g. seat height, wheel size, and camber). Moreover, recent advances such as Pushrim-Activated Power-Assisted Wheelchairs (PAPAWs, Chapter 7) are effective at reducing the strain imposed by wheelchair propulsion [38,39] and increase effective range of wheelchair users [40,41]. Yet, these introduce even more parameters that can be tuned (e.g. motor delay, power, and overrun) to the individual's propulsion technique, complicating the manual tuning of these wheelchairs [42,43]. Using a wheelchair ergometer in the wheelchair propulsion lab could allow for the safe and automatic individual tuning of control parameters based on physiological cost using a "human-in-the-loop" optimization process.

Methodological considerations

The focus of the current project was initially directed towards clinical (rehabilitation) practice. Interestingly, most early interest for the wheelchair ergometer has come from the adapted sports field. While validating the ergometer, a comparison with measurement wheels available at our lab at higher speeds seemed infeasible as they do not allow for the use of racing handrims, they are too 'fragile' to sustain racing kinetics, and they too have not been tested at these more typical racing speeds. As such, the ergometer has been validated for regular daily use, but it remains unclear how extreme increases in velocity and acceleration impact the measurement accuracy of the ergometer. While there is no current indication that these tests would obtain erroneous results, it would be prudent to also perform tests in this area and consider more advanced dynamic calibration procedures, using ramp or sinusoidal input signals [44].

The current thesis uses filter parameters based on previous research with measurement wheels [37,45] and wheelchair ergometers [9,46]. Filtering can be used to remove the high-frequency noise that is present in the force signal that is measured by the ergometer [45] (e.g. Chapter 2 explored the resonance frequency). It is a recurring and sensitive topic among researchers as, without the proper parameters, one could either leave some noise in the signal or remove important information. However, previous studies have used a range of different filter parameters and it is not always clear why a specific combination was used.

Therefore, specific guidelines for the current ergometer are needed, which could be determined using the methods described by Cooper et al. [45,47] or by using the Woltring filter routine [48]. Moreover, given the broad range of applications for the ergometer, filter guidelines should cover the entire spectrum of wheelchair testing.

The current thesis tried to employ a number of practices aimed at making its results more open, to ensure reproducible and valid results [49,50]. Among others, these were the use of open access publications, sharing code in a free and open-source language [51] while using distributed version control [52,53]. However, it must also be conceded that a number of open science aspects were not yet implemented in this thesis. Performing preregistered, well-powered experiments are perhaps the most important step for science in general [50], but the studies in the current thesis were not preregistered. Data, while available on request, were not made available by default, as the intent to publicly share data should have been more explicit during the approval stage. The selected journals for open access publication did not allow for the interactive display of data and still few accept R markdown [54] or Jupyter Notebooks [55]. Investing additional effort in this domain should not be considered as a burden, as it does not only improve science, but also leads to more citations, collaboration, and funding opportunities [56]. A goal worth striving for, as science only works when it is open.

Concluding remarks

Finally, it should be noted that there is already a large existing knowledge base, which is the result of more than five decades of wheelchair propulsion research. Before embarking on new endeavours these studies should be consulted and data collected by new research should be embedded in their results. Nevertheless, it is time to rethink the implementation of wheelchair research. By applying standardized protocols and outcomes in a research community, large sets of data can be collected in a field that desperately needs it due to large heterogeneity and low patient numbers, and a diversity of health-related issues over the life span. Therefore, it seems important to build a community of wheelchair propulsion labs that meet to share best practices, discuss experiences, design protocols, develop graphical references, and share (de-identified) data [10]. This should kickstart the trend to collect data in clinical and applied (sports) environments, which could be the collaborative push needed to accelerate developments even further. The current thesis presents a case for including wheelchair ergometers in wheelchair propulsion labs which will hopefully spark the broader discussion on wheelchair propulsion testing in research, clinical, and adapted sports environments.

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Appendices

Summary

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About the author

Output

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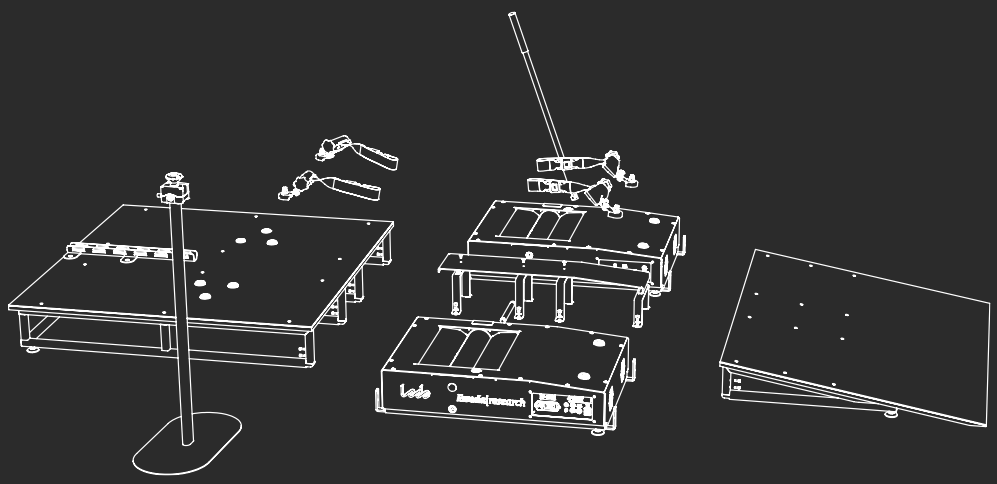

Summary

This thesis **-Development of a wheelchair propulsion laboratory-** aggregates existing methods and equipment, and introduces new innovations for the formation of a wheelchair propulsion laboratory that can be employed in research, clinical practice, adapted sports, and beyond. **Chapter 0** provides a brief introduction on the state of the art in wheelchair research. Wheelchair propulsion more often than not leads to overuse complications, underlining the need for proper methods to assess wheelchair propulsion to find their way from research into clinical and adapted sports environments. The development of a wheelchair propulsion laboratory could provide the information needed to optimize the wheelchair, user, and wheelchair-user interface, but their implementation outside academia has been limited. The wheelchair ergometer introduced in this thesis could significantly lower the barrier to the systematic assessment of wheelchair users and help in the dissemination of wheelchair propulsion laboratories.

Chapter 1 examines the existing state of the art of equipment used to simulate and measure wheelchair propulsion in the laboratory. The assessment of wheelchair users on treadmills and ergometers has allowed for highly controlled and standardized lab-based studies to be performed. However, as ergometers are often not commercially available, many institutes have developed their own device, which have not been collectively examined before. This chapter critically assesses lab-based equipment used in published research on their ability to simulate and measure wheelchair propulsion performance. Special attention was paid to the ability to simulate frictional losses and inertia of real-life handrim wheelchair propulsion, while maintaining the ergonomics of the wheelchair-user interface. The review identified fifty different ergometers with a wide range of strengths and limitations. The heterogeneity of this equipment and the lack of consensus on how to simulate field conditions complicates the interpretation of results and any comparison with the existing knowledge base.

Chapter 2 introduces a novel servo-driven dual-roller wheelchair ergometer which allows for the testing of wheelchair users in their own personal wheelchair. The ergometer was developed with a commercial partner and is thoroughly described in this chapter, ensuring its availability to other researchers and professionals. It employs an admittance-controlled feedback system that can produce different simulations of wheelchair propulsion, thereby providing a flexible testing environment. The ergometer measures the effective component of the user applied force and produces similar results to those of established systems. However, agreement between some variables such as mean- and peak force is lower than others, indicating that these should be further examined. The ergometer is expected

to play an essential role in the wheelchair propulsion laboratory due to its ability to provide an accessible standardized training and testing environment.

Chapter 3 introduces and evaluates a new portable low-cost dynamic calibration method for wheelchair roller ergometers. The dynamic calibration of ergometers is important as wheelchair propulsion is a dynamic activity. Yet, ergometers are often only statically calibrated as this is relatively simple and less resource intensive. The uncertainty of the calibration system was assessed and compared to a reference ergometer. As the uncertainty of the calibration system was found to be lower than the ergometer, it can be used to evaluate roller ergometers. Moreover, as it is a portable and low-cost method, it could be used by distributors to perform annual calibrations, or be used by researchers in their own lab. Some suggestions to improve the performance and general usability of the system are provided. In general, the developed calibration method is suitable for the dynamic calibration of roller ergometers in various contexts.

Chapter 4 aggregates, visualises, and reiterates the current state of the art of measuring and standardising power output in wheelchair propulsion research for overground, treadmill, and ergometer propulsion. The measurement and standardisation of power output (W) is seen as a crucial step needed for the interpretation and comparison of experiments and tests aimed at optimizing the wheelchair-user combination. For overground propulsion, power output can be estimated through coast down testing. To match the power output of overground propulsion on the treadmill, a drag test should be performed in combination with a pulley system. Finally, the ergometer needs to be properly calibrated and the right settings have to be selected. The goal of this chapter was to stimulate good practice and standardization to facilitate the further development of theory and its application. The chapter is accompanied by an online video that demonstrates the procedures to promote their use in practice.

Chapter 5 compares the physiological and biomechanical outcomes between overground, treadmill, and ergometer wheelchair propulsion under standardized conditions. Different labs employ different testing modalities, under the assumption that the results obtained relate to everyday overground wheelchair propulsion. However, little is known about the agreement of data obtained during overground, treadmill, and ergometer propulsion at similar power output and speed. By employing the methods in chapter 4, the power output was standardized, making the task requirements as equal as possible. Agreement among conditions was moderate to good for most outcome variables, but some differences between conditions were also found. The ecological validity of lab-based equipment should

be further improved and the possible impact of these differences on specific protocols and outcomes should still be evaluated.

Chapter 6 and 7 show two implementation studies that demonstrate how the wheelchair propulsion lab could be used. Chapter 6 employs a standardized learning protocol on the wheelchair ergometer to examine the acquisition of wheelchair racing propulsion skill in a group of novices. Wheelchair racing is a highly technical skill that is considerably different from regular handrim wheelchair propulsion. The study investigates novices during uninstructed wheelchair practice and compares them with an experienced athlete. The propulsion technique of the novices improved over time, but not nearly reached that of an experienced athlete. Chapter 7 contrasts this by illustrating the challenges associated with conduction motor learning studies on a wheelchair treadmill, while exploring the acquisition of power-assisted wheelchair propulsion skill. Power assisted wheelchairs are an increasingly common mode of transportation as they can reduce the strain of wheelchair propulsion and increase the range of action of wheelchair users. This chapter shows that power-assisted wheelchairs are initially challenging to control for novices, but this improves with practice. The ergometer could be used in future studies to further optimize the algorithms used in power assisted wheels.

Chapter 8 provides a general discussion of the findings of this thesis, discussing the possibilities for implementing new innovations in the wheelchair propulsion laboratory. Current practice cannot yet fully benefit from the methods developed in the academic environment and researchers struggle to find enough participants. The implementation of wheelchair propulsion laboratories could tackle both problems by allowing for the systematic analysis of wheelchair users in research, but also in clinical- and adapted sports practice. Moreover, they will allow for efficient multi-centre collaborations by providing standardized protocols and outcomes. The availability of the wheelchair ergometer is crucial to the development of wheelchair propulsion laboratories as it allows for and facilitates the simulation and measurement of wheelchair propulsion.

Samenvatting

Dit proefschrift **-Ontwikkeling van een rolstoelpropulsie-lab-** brengt bestaande methoden en instrumenten samen, en introduceert nieuwe innovaties voor het vormen van een rolstoelpropulsie-lab, dat kan worden ingezet binnen onderzoek, de klinische revalidatiepraktijk, aangepaste sport en daarbuiten. **Hoofdstuk 0** biedt met een korte introductie inzicht in de stand van zaken in het handbewogen rolstoelonderzoek. Rolstoelpropulsie leidt vaak tot overbelastingsklachten. Dit onderstreept het belang dat passende methoden en technieken om rolstoelpropulsie te onderzoeken vanuit het onderzoeksveld ook hun weg vinden naar de klinische en aangepaste sportpraktijk. De ontwikkeling van een rolstoelpropulsie-lab kan de informatie leveren die nodig is om de rolstoel, gebruiker en de rolstoel-gebruiker combinatie te optimaliseren en te monitoren. Buiten de onderzoekswereld worden deze labs echter nog beperkt ingezet. De rolstoelergometer die in dit proefschrift wordt geïntroduceerd zou de drempel om rolstoelgebruikers systematisch te onderzoeken significant kunnen verlagen en daarmee de verspreiding en implementatie van rolstoelpropulsie-labs kunnen stimuleren.

Hoofdstuk 1 brengt in kaart welke instrumenten momenteel worden gebruikt om hoepelrolstoelpropulsie te meten in het lab. Het meten van rolstoelgebruikers op loopbanden en ergometers biedt de kans om in een controleerde en gestandaardiseerde lab-omgeving onderzoek uit te voeren. Ergometers zijn echter vaak niet commercieel verkrijgbaar, en daarom hebben veel instituten hun eigen instrumenten ontwikkeld die nog niet eerder gezamenlijk zijn onderzocht. Dit hoofdstuk brengt op basis van gepubliceerd onderzoek in kaart in welke mate instrumenten uit het lab in staat zijn om rolstoelpropulsie te simuleren en te meten. Daarbij wordt nadrukkelijk gelet op de mate waarin deze instrumenten de wrijvingsverliezen en massatraagheid van de rolstoelpropulsie konden simuleren, met handhaving van de rolstoel-gebruiker koppeling. De overzichtsstudie identificeerde vijftig verschillende ergometers met uiteenlopende ontwerpen en toepassingen. De heterogeniteit van deze instrumenten en het ontbreken van overeenstemming over het simuleren van veldcondities, bemoeilijkt de interpretatie van resultaten en het vergelijken met de bestaande kennisbasis.

Hoofdstuk 2 introduceert een nieuwe servo-aangedreven dubbel-rol ergometer die kan worden ingezet bij het testen van rolstoelgebruikers in hun eigen rolstoel. De ergometer werd in samenwerking met een commerciële partner ontwikkeld, en wordt in dit hoofdstuk uitgebreid beschreven, waardoor de (onderliggende) technologie ook beschikbaar is andere onderzoekers en professionals. De ergometer maakt gebruik van een admittantie-gecontroleerd feedback systeem dat verschillende simulaties van rolstoelpropulsie kan produceren, en daarmee een

flexibele testomgeving biedt. De ergometer meet de effectieve component van de kracht die de rolstoelgebruiker op de hoepels levert, en produceert daarin vergelijkbare resultaten als bestaande systemen. Dat de overeenstemming tussen bepaalde variabelen lager is dan anderen geeft aanleiding tot vervolgonderzoek. De ergometer zal naar verwachting een essentiële rol spelen in rolstoelpropulsie-labs omdat het een toegankelijke, gestandaardiseerde train- en testomgeving biedt voor (vroege) revalidanten, in alle daags gebruik en voor (top)rolstoelatleten.

Hoofdstuk 3 introduceert en evalueert een nieuwe dynamische kalibratie methode voor rollerergometers, die verplaatsbaar en goedkoop is. Dynamische kalibratie van ergometers is belangrijk omdat rolstoelrijden een dynamische activiteit is. Ergometers worden vaak alleen statisch gekalibreerd, omdat dat relatief makkelijk is en weinig middelen vereist. De onzekerheid van het ontwikkelde kalibratiesysteem is in kaart gebracht door het systeem te vergelijken met een referentie ergometer. Omdat de onzekerheid zoals die werd gevonden in het kalibratiesysteem veel kleiner bleek te zijn dan de onzekerheid van de ergometer, kan dit kalibratiesysteem worden gebruikt om roller ergometers te valideren. Omdat het systeem verplaatsbaar en goedkoop is, kan het door distributeurs worden gebruikt voor periodieke kalibraties, en door onderzoekers in hun eigen lab. Er worden een aantal suggesties gegeven om de prestaties en inzetbaarheid van het systeem te verbeteren. De nieuw-ontwikkelde dynamische kalibratiemethode is geschikt voor dynamische kalibratie van ergometers in verschillende contexten.

Hoofdstuk 4 benoemd en benadrukt de huidige stand van zaken in het meten en standaardiseren van vermogen in onderzoek naar hoepelrolstoelpropulsie. Het meten en standaardiseren van vermogen wordt gezien als een cruciale stap voor het interpreteren en vergelijken van experimenten en testen die bedoeld zijn om de rolstoel-gebruiker combinatie te verbeteren in onderzoek, revalidatie en aangepaste sport. Voor propulsie over de grond wordt vermogen gemeten door middel van een ‘coast-down’ test. Om het rijden over de grond te benaderen op de loopband, moet er een ‘dragtest’ worden uitgevoerd, in combinatie met een ‘pulley’ systeem. De ergometer, tot slot, moet passend gekalibreerd zijn en de instellingen moeten passen bij die van het grondoppervlak. Het doel van dit hoofdstuk was om ‘good practice’ te stimuleren, om theorieontwikkeling en inzet te faciliteren. Bij dit hoofdstuk hoort een onlinevideo waarin de procedures worden gedemonstreerd, bedoeld om het gebruik van deze methoden toegankelijker te maken.

Hoofdstuk 5 vergelijkt de fysiologische en biomechanische uitkomsten van hoepelrolstoelpropulsie over de grond, op de loopband en de ergometer, onder gestandaardiseerde condities. Verschillende labs gebruiken verschillende testmodi,

onder de aanname dat de gevonden resultaten zich laten vertalen naar rolstoelpropulsie in het dagelijks leven. Er is echter nog weinig bekend over de mate waarin data over propulsie met vergelijkbare vermogen en snelheid daadwerkelijk overeenkomt, wanneer het is verzameld over de grond, de loopband en de ergometer. Door de methoden uit hoofdstuk 4 te hanteren, werd het vermogen gestandaardiseerd, waardoor taakeisen zo gelijk mogelijk waren. Overeenstemming tussen de condities was gemiddeld tot goed voor de meeste uitkomstvariabelen, maar er werden ook enkele verschillen tussen de condities gevonden. De ecologische validiteit van lab apparatuur zal verder moeten worden verbeterd en de mogelijke invloed van deze verschillen op specifieke protocollen en uitkomsten moet worden geëvalueerd.

Hoofdstuk 6 en 7 omvatten twee implementatieonderzoeken die demonstreren hoe het rolstoelpropulsie-lab zou kunnen worden gebruikt. Hoofdstuk 6 hanteert een gestandaardiseerd leerprotocol op de rolstoel ergometer, om het aanleren van vaardigheden voor propulsie in rolstoel racen onder een groep onervaren gebruikers in kaart te brengen. Rolstoelracen is een zeer technische vaardigheid, die duidelijk verschilt van reguliere hoepelrolstoelpropulsie. De studie onderzoekt onervaren deelnemers gedurende onbegeleide rolstoel-oefensessies op de rolstoelergometer en vergelijkt hen met een ervaren atleet. De propulsie techniek van nieuwe gebruikers verbeterde over de tijd, maar kwam nog niet in de buurt van dat van een ervaren atleet. Hoofdstuk 7, daarentegen, illustreert de uitdagingen van motorisch leren in het gebruik van een elektrisch ondersteunde hoepelrolstoel op een rolstoel loopband. 'Power assisted' rolstoelen zijn een steeds vaker voorkomende modus van verplaatsen, omdat ze overbelastingsklachten in de schouders kunnen verminderen en de actieradius van rolstoelgebruikers kunnen vergroten. Dit hoofdstuk laat zien dat het controleren van ondersteunde rolstoelen voor nieuwe gebruikers in het begin een uitdaging vormt, maar dat de controle door oefening verbetert. De ergometer zou in toekomstig onderzoek kunnen worden gebruikt om de algoritmes die in 'power-assist' rolstoelen worden gebruikt te verbeteren.

Hoofdstuk 8 biedt een algemene discussie over de bevindingen van dit proefschrift, met benoeming van de mogelijkheden voor het implementeren van nieuwe innovaties in het rolstoelpropulsie-lab. De huidige praktijk kan nog niet volledig profiteren van de methoden die in de academische omgeving zijn ontwikkeld, en het blijft voor onderzoekers lastig om voldoende participanten te vinden. De implementatie van rolstoelpropulsie-labs kan deze beide problemen oplossen, doordat het mogelijkheden biedt voor het systematisch analyseren van rolstoelgebruikers in onderzoek, maar ook in de klinische praktijk en de

aangepaste (top)sport. Daarnaast bieden deze labs de mogelijkheid voor efficiënte samenwerking tussen verschillende centra, doordat zij gestandaardiseerde protocollen en uitkomsten aanleveren. De beschikbaarheid van de nieuwe rolstoel ergometer is cruciaal voor de ontwikkeling van rolstoelpropulsie-labs, doordat het adequaat simuleren en accuraat meten van rolstoelpropulsiekenmerken ermee mogelijk is, en wordt gefaciliteerd.

Dankwoord

Dit proefschrift voelt als een collectieve prestatie en dat is het ook. Ik ben de mensen die in de afgelopen vier jaar aan dit project hebben bijgedragen erg dankbaar. Een aantal personen zou ik daarbij nog graag in het bijzonder willen noemen.

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C'è stato un momento durante il mio progetto in cui la mia motivazione era piuttosto bassa. Fortunatamente, **Paolo** e **Ridi** si sono uniti alla squadra. Non solo ci siamo divertiti molto in laboratorio, ma siete stati anche preziosi partner di sparring. Quei mesi sono stati tra i più divertenti e produttivi del mio dottorato. **Matteo**, la tua ospitalità quando ho visitato Brescia era indescrivibile. Onestamente non credo che sarei stato in grado di farlo senza di voi ragazzi, e per questo vi ringrazio.

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About the author



Rick de Klerk was born on December 26th, 1992 in Oostvoorne, the Netherlands. He earned his high school diploma (VWO) at the CSG Jacob van Liesveldt in Hellevoetsluis in 2011. He completed the International Baccalaureate English Language A2 higher level program as part of his high school education. Rick moved to Groningen to study Human Movement Sciences at the University of Groningen where he obtained his Bachelor's and Master's degree in 2014 and 2016, respectively. As part of his master program, he completed the Honours College master programme 'Leadership: making the difference!' in 2015. In 2016, he also finished the Teacher Training for Human Movement Sciences program at the Vrije Universiteit in Amsterdam after an internship at Medical Imaging and Radiation Therapy at Hanze University of Applied Sciences in Groningen.

After his Master's, Rick worked at Umaco BV on a wheelchair ergometer prototype. Shortly thereafter, he transferred to the University Medical Centre Groningen (UMCG) to start a research project on the Esseda wheelchair ergometer, a new wheelchair roller ergometer developed by Lode BV in collaboration with the UMCG. During this period, he conducted several research projects which resulted in a number of research papers, twelve Bachelor theses, and one Master thesis. For his work during his PhD, Rick earned three 'SHARE Top Publication Awards'. In his second year, he got a teaching appointment for one day per week to aid in the courses Movement Analysis 1 and Big Data for Human Movement Sciences & Sport Sciences. This earned him his University Teaching Qualification (UTQ) as a proof of didactic competence in academic education. During his final year he joined Axtion BV as a Junior Software Engineer to implement scientific algorithms for the wheelchair ergometer in Lode Ergometry Manager. Currently, Rick is employed at Fontys Hogescholen as a Data Scientist.

Output

Journal publications

- De Klerk R, Lutjeboer T, Vegter RJ, van der Woude LH. Practice-based skill acquisition of pushrim-activated power-assisted wheelchair propulsion versus regular handrim propulsion in novices. Journal of NeuroEngineering and Rehabilitation. 2018
- De Klerk R, Vegter RJ, Goosey-Tolfrey VL, Mason BS, Lenton JP, Veeger HE, Van Der Woude LH. Measuring handrim wheelchair propulsion in the lab: a critical analysis of stationary ergometers. IEEE Reviews in Biomedical Engineering. 2019
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- De Klerk R, Vegter RJ, Leving MT, de Groot S, Veeger HE, van der Woude LH. Determining and Controlling External Power Output During Regular Handrim Wheelchair Propulsion. Journal of Visualized Experiments. 2020
- De Klerk R, Velhorst V, Vegter RJ, Veeger HE, Van der Woude LH. Biomechanical comparison of overground, treadmill, and ergometer propulsion. Journal of NeuroEngineering and Rehabilitation. 2020

Submitted for publication

- De Klerk R, Spada P, Muhametaj R, Lancini M, Van der Woude LH, Vegter JK. Development of a portable low-cost system for the validation of wheelchair roller ergometers. Submitted.
- De Klerk R, Van der Jagt G, Veeger HE, Van der Woude LH, Vegter JK. Evolution of wheelchair racing propulsion technique over three weeks of wheeling practice on an instrumented ergometer in able-bodied novices. Submitted.

Manuscripts in progress

- De Klerk R, Braaksma J, Vegter JK, Veeger HE, Van der Woude LH. Dynamic validation of wheelchair ergometers using a simple pulley-based method. Under construction.

Other publications

- Valent, L, De Groot, S, Kaandorp, E, De Klerk, R, Roebroek, B, Vegter, RJK, Wynants, M. Nuttige informatie op de Supportbeurs: check je zit. Dwarslaesie Magazine. 2019
- De Klerk, R. Worklab: a wheelchair biomechanics mini-package. Zenodo. http://doi.org/10.5281/zenodo.3268671. 2019
- De Klerk, R. De Esseda: met je rolstoel op de rollerbank. Moving Matters. 2019
- De Klerk, R. Pyirr: Coefficients of Interrater Reliability and Agreement, a Python implementation of the R package IRR. https://pypi.org/project/pyirr/. 2020

Other products

- Esseda in UE4: a coupling between the Esseda and a Virtual Reality environment to be used with a normal or head-mounted display, built in the Unreal Engine.
- Dataviewer: an interactive data inspection tool that allows for synchronized display of video, ergometer, and kinematics data.
- Coast-down analyser: a data analysis tool that allows for the selection of coast-down moments in measurement wheel or IMU data to determine rolling friction.
- Drag test analyser: a data analysis tool that allows for the custom processing of drag test data.

Conference contributions

- 2018: Oral Presentation: Science and Engineering Congress on Sports Innovation
- 2018: Oral Presentation: European College of Sport Science
- 2018: Oral Presentation: World Congress of Biomechanics
- 2019: Poster Presentation: RehabMOVE
- 2019: Workshop: VISTA (International Paralympic Committee)
- 2019: Poster Presentation: VISTA (International Paralympic Committee)
- 2019: Oral Presentation: Dutch Congress for Rehabilitation Medicine
- 2020: Workshop: Dutch Congress for Rehabilitation Medicine

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More recent theses can be found in the list below.

(supervisors are between brackets)

2020

Wijnen A
Rehabilitation policies following total hip arthroplasty; across borders
(prof SK Bulstra, prof D Lazovic, dr M Stevens)

Spinder N
Maternal occupational exposure and congenital anomalies
(prof HM Boezen, prof H Kromhout, dr HEK de Walle, dr JEH van Kammen-Bergman)

Driel-de Jong TJW van
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(prof U Bültmann, prof R Tkacova, dr JP van Dijk, dr I Nagyova)

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Total knee arthroplasy among working-age patients
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Donk LJ van der
Cancer survivors' experience with depressive symptoms and their (low) need for psychological care; Lessons learned from a multi-center randomized controlled trial
(dr MJ Schroevers, dr J Fleer, prof R Sanderman)

Hovenkamp-Hermelink A
The long-term course of anxiety disorders; an epidemiological perspective
(prof RA Schoevers, dr H Riese, dr B Jeronimus)

Blikman T
Neuropathic-like symptoms in hip and knee osteoarthritis
(prof SK Bulstra, dr M Stevens, dr I van den Akker-Scheek)

Fard B
Dysvascular lower limb amputation: incidence, survival and pathways of care
(prof JHB Geertzen, prof PU Dijkstra)

Niebuur J
Who volunteers and why? Understanding the role of resources and motivations in participation in voluntary work
(prof AC Liefbroer, prof N Steverink, dr N Smidt)

Thio CHL
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Rocha dos Santos
Effects of age and fatigue on human gait
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Health economics of direct oral anticoagulants in the Netherlands
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Diemen MCJM van
Self-management, self-efficacy, and secondary health conditions in people with spinal cord
(prof MWM Post, prof JHB Geertzen, dr I van Nes)

Jacobs MS
Anticoagulation in atrial fibrillation; consideration for treatment and health economic aspects
(prof MJ Postma, dr M van Hulst, dr RG Tieleman)