

Measuring Handrim Wheelchair Propulsion in the Lab: A Critical Analysis of Stationary Ergometers

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(Methodological Review)

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.

Index Terms—Dynamometer, Ergometry, Ergonomics, Simulation, Mechanics, WERG.

I. 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-

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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:

- 1) Simulate frictional losses, environmental conditions, and translational inertia of the wheelchair-user system.
- 2) Facilitate valid and reliable measurements of power output on the wheels.
- 3) Respect the ergonomic properties of the wheelchair-user interface and provide adequate sensory feedback to the user.
- 4) 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.

II. 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).

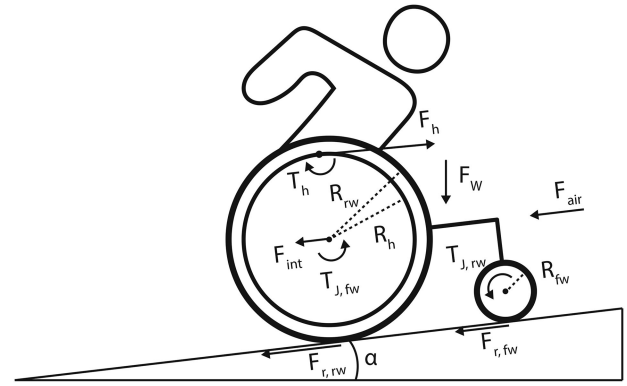


Fig. 1. Diagram of a wheelchair rolling over a flat surface with an incline; user force and frictional losses are illustrated as force vectors in handrim wheelchair propulsion.

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

III. 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 (Fig. 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 modeled as a linear system, where the acceleration of the center of mass of the wheelchair-user combination is equal to the total acting force divided by the mass:

$$F_{\text{sum}} = m_{\text{tot}} * a \quad (1)$$

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:

$$m_{\text{tot}} = m_{\text{user}} + m_{\text{wc}} + \sum_{i=1}^{n\text{-wheels}} \frac{J_i}{R_i^2} \quad (2)$$

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:

$$F_{\text{sum}} = F_s - F_{\text{roll}} - F_{\text{air}} - F_\alpha - F_{\text{int}} \quad (3)$$

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):

$$F_s = \frac{F_h * R_h + T_h}{R_{rw}} \quad (4)$$

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:

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

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}):

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

$$N_{rw} = m * g - N_{fw} \quad (7)$$

At high speeds the air drag becomes the most 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.

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

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.

$$F_\alpha = m * g * \sin(\alpha) \quad (9)$$

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].

IV. SIMULATION & MEASUREMENTS: TREADMILLS

A. 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

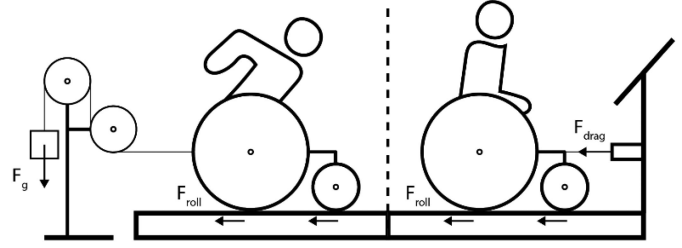


Fig. 2. On the right-hand side the wheelchair drag test to determine the workload is shown. On the left-hand side a pulley system to increase the workload is shown.

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 overground propulsion at specific treadmill settings.

B. 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 power output to be varied through belt inclination or the application of resistance to the back of the wheelchair via a pulley system (12, Fig. 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.

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

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

C. 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.

V. 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 (Fig. 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

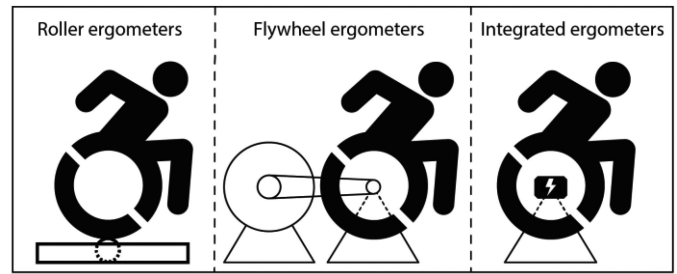


Fig. 3. Schematic drawing of the three types of wheelchair ergometers found in literature. Roller ergometers use the personal wheelchair on a roller, flywheel ergometers use an integrated wheelchair coupled with an ergometer, integrated ergometers use an integrated wheelchair with a braking/accelerator system.

(most of) these ergometers make some additional assumptions about wheelchair propulsion which could hurt the relation with the field [45]:

- 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.

A. Roller Ergometers

The majority of WCEs found in literature could be categorized as roller ergometers (Table I). 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. 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:

TABLE I
CHARACTERISTICS OF ALL WHEELCHAIR ERGOMETERS FOUND IN THE LITERATURE. AN OVERVIEW OF THEIR MEASUREMENT AND SIMULATION CAPABILITIES IS PROVIDED

Ergometer (name or affiliation) + (first mention found)	Category	Ergonomics	Measurements	Inertia	Rolling resistance	Air drag	Slope [†]	Left/right	Feed-back	Validity/reliability	Remarks
Abilities Incorporated, 1967 [44]; University of Turin, 2001 [82]	Roller ergometer	++ personal wheelchair	+/- distance and time	- roller mass not matched	+/- internal friction of WCE	- No	- No	- No	+/- V	[44] [†]	
Gothenburg, 1970 [83]; McMaster, 1977 [84]; Wright State, 1979 [85]; Boston, 1982 [86]; Southern Illinois, 1981 [87]; Vandoeuvreles-Nancy, 1991 [88]; Wisconsin, 1995 [89]; Maryland, 1998 [90]; Auckland, 2000 [91]; University Of Illinois, 1972 [58]	Flywheel ergometer	- integrated wheelchair	+ power output, speed	+/- fixed mass flywheel	+ brake	- No	- No	- No	+/- V	[75], [76], [83], [85], [92] [†] , [93] ^b	Commercially available bicycle ergometers
Cardrei, 1978 [63]	Roller ergometer	++ personal wheelchair	+ power output, speed	+ variable mass flywheel	+ brake	+ motor	++ motor	- No	+/- V	[58] ^b	
University of Melbourne, 1983 [80]	Integrated ergometer	- integrated	++ torque, speed	+ variable mass flywheel	+ brake	- No	- No	+ Yes	+/- V		Commercial
University of British Columbia, 1983 [94]	Roller ergometer	++ personal wheelchair	+ power output, speed	- roller mass not matched	+ brake	- No	- No	- No	+/- V		Some bicycle ergometer parts
Arizona State University, 1987 [71]	Roller ergometer	++ personal wheelchair	+ power output, speed	+/- fixed mass rollers	+ motor	- No	- No	- No	+/- V		
Liverpool Polytechnic, 1988 [95]; University of Alberta, 1989 [96]	Integrated ergometer	+/- integrated, limited	+ power output, speed	+ variable mass flywheel	+ brake	- No	- No	- No	+/- V	[71] ^b	
Lingköping University, 1989 [64]	Hybrid ergometer	?	+ power output	+/- fixed mass flywheel	+ brake	- No	- No	- No	+/- V	[95] [†]	
Erasmus Rotterdam, 1990 [7]	Integrated ergometer	- integrated wheelchair	+ power output, speed	- n.a.	- n.a.	- n.a.	- n.a.	- n.a.	+/- V		Isokinetic ergometer
California State University, 1990 [97]	Integrated ergometer	+/- integrated, adjustable	++ 3D-handrim and seat kinetics	+ motor	+ motor	+ motor	++ motor	+ Yes	+/- V		Simulator
San Diego State University, 1991 [98]; Lavel University, 1991 [62]	Roller ergometer	++ personal wheelchair	+ speed	+ variable mass roller	+ ?	- No	- No	- No	+/- V		
Shenandoah, 1996 [92]; Mayo Clinic, 2006 [99]; Utrecht, 2014 [100]	Flywheel ergometer	- integrated wheelchair	+ power output, speed	+/- fixed mass flywheel	- n.a.	- n.a.	- n.a.	- No	+/- V		Watt controlled
California State University 2, 1993 [101]	Hybrid ergometer	++ personal wheelchair	+ power output, speed	+/- fixed mass flywheel	+ brake	- No	- No	- No	+/- V		Rollers attached to flywheel
Hines Hospital "WAFI", 1993 [68]	Hybrid ergometer	- integrated wheelchair	+ power output, speed	+/- fixed mass	+/- Cateye CS1000	- No	+ Yes	+ Yes	- No		Turbo trainer
Ergotronic 4000, 1993 [102]; Ergotronic 9000, 1993 [59]	Roller ergometer	++ personal wheelchair	+ power output, speed	+/- fixed mass flywheel	+ electromagnetic brake	- No	- No	+ Yes	+/- V	[68] ^b	
Catholic University of Louvain 1, 1996 [48]	Roller ergometer	++ personal wheelchair	++ tangential force, speed	+ motor	+ motor	+ motor	++ motor	- No	+/- V		Commercial
Rancho Los Amigos Medical Centre, 1996 [103]	Roller ergometer	++ personal wheelchair	+ power output, speed	+ variable mass roller	+ electromagnetic brake	+	+	+ Yes	+/- V + P	[48] ^b	
Loughborough "Bromakin", 1998 [104]	Hybrid ergometer	++ personal wheelchair	++ tangential force, speed	+ variable mass roller	+ Velodyne	- No	- No	+ Yes	+/- V		
Wingate Institute "WILLY", 2000 [69]	Roller ergometer	++ personal wheelchair	++ torque, speed	+/- fixed mass rollers	?	- No	- No	- No	+/- V		
VP100H, 2001 [56]; VP100HANDI, 2008 [52]; VP25A, 2015 [105]	Roller ergometer	++ personal wheelchair	++ tangential force, speed;	+ motor	+ motor	+ motor	+	+ Yes	+/- V	[69] ^b	
University of Pittsburgh, 2001 [57]	Roller ergometer	++ personal wheelchair	+ power output, speed	+ equal to 35 or 75 kg	+ hysteresis brake, adaptable to camber	- No	+ brakes	- No	+/- V	[56] ^b	Commercial
University of Pittsburgh "Gamewheels", 2001 [106]	Roller ergometer	++ personal wheelchair	++ torque, speed	+/- fixed mass rollers	+ motor	+ motor	++ motor	+ Yes	+ V + P	[57] ^b	
École de technologie supérieure Montreal, 2002 [51]	Roller ergometer	++ personal wheelchair	+/- distance and time	+/- fixed mass rollers	+ electromagnetic brake	- No	- No	+ Yes	+ V + G		Exergaming, portable
The Hong Kong Polytechnic University, 2003 [107]	Roller ergometer	++ personal wheelchair	+ power output, speed	+/- fixed mass rollers	+ adjusting internal friction	- No	- No	- No	+/- V + P		Adjustable incline
University of Strathclyde, 2004 [60]	Roller ergometer	++ personal wheelchair	++ distance and time	?	?	?	?	+ Yes	+/- V		
University of Pittsburgh 2, 2007 [54]; University College London/University of Alberta 1, 2012 [108]	Roller ergometer	++ personal wheelchair	+ power output, speed	+ variable mass rollers	+ mechanical brake	- No	+	+ Yes	++ VR	[60] ^b	Immersive simulation
Marmara University, 2008 [109]	Roller ergometer	++ personal wheelchair	+ power output, speed	+/- fixed mass rollers	+/- passive	- No	- No	+ Yes	+/- V	[70] ^b	
Louvain 2, 2011 [110]	Roller ergometer	++ personal wheelchair	+/- distance and time	- roller mass not matched	+ brake	- No	- No	- No	?		
Yonsei University, 2013 [111]	Hybrid ergometer	++ personal wheelchair	+ power output, speed	+ flywheel ratio adjustable	+ brake	- No	- No	- No	+/- V		Rollers attached to flywheel
University of Alberta 2, 2013 [55]	Roller ergometer	++ personal wheelchair	++ torque, speed	+ variable mass rollers	+/- passive	- No	- No	+ Yes	+/- V		
Washington University "Wheelmill", 2014 [61]	Roller ergometer	++ personal wheelchair	+ power output, speed	+/- fixed mass rollers	+ pneumatic brake	- No	- No	+ Yes	+/- V	[55] ^b	
École de technologie supérieure Montreal, 2015 [40]	Roller ergometer	++ personal wheelchair	+ power output, speed	+ motor	+ motor	- No	- No	+ Yes	+ V + P	[61] ^b	Inclines and slopes
	Roller ergometer	++/- personal wheelchair	++ 3D handrim forces	+ motor	+ motor	+ motor	++ motor	+ Yes	+/- V	[40] ^b	Measurement wheels

[†]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.

1) 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].

2) 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

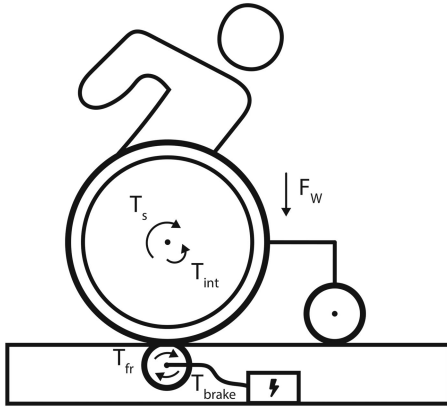


Fig. 4. Graphic representation of a wheelchair roller ergometer with an active system to generate resistance. Translational inertia is simulated with the roller mass [13].

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 shifted 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 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

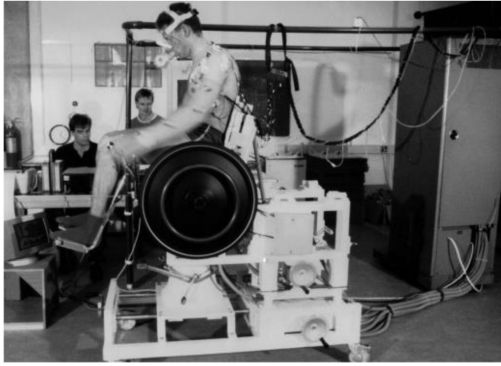


Fig. 5. Example of an integrated ergometer at the VU Amsterdam. The wheelchair is integrated with the stationary ergometer. This ergometer uses a haptic feedback model based on user torque and simulated frictional forces [7].

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.

B. 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 I).

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 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 (I_{flywheel}) (gear ratio of 1:1):

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

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.

C. 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.

D. Measurements on a WCE

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 15 m 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. Bhamhani *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.

VI. ERGONOMICS & SENSORY FEEDBACK

A. 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].

B. 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.

C. 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 [112] 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 [113]. 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 [114], or as a motivational tool [106].

D. Auditory Feedback

Auditory cues could be used in learning tasks or to increase immersion [113]. 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.

VII. TESTING CAPABILITIES

A. 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], [114], [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 chapter IV and V, there is little information available on the ecological validity of biomechanics testing on treadmills [21], [25], [26] and ergometers [27], [70]–[72].

B. 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 and 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].

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.

TABLE II
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

C. 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].

VIII. SYNOPSIS AND 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 II). 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 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].

IX. 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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