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Dynamometer prototype to assess muscle strength in wheelchair users. First part: literature review and simulation of the dynamic model

Protótipo de dinamômetro para avaliação da força muscular em cadeirantes. Primeira parte: revisão da literatura e simulação do modelo dinâmico

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ABSTRACT

This paper presents a detailed literature review on the main aspects of a low-cost experimental dynamometer model designed to assess the muscular strength of wheelchair users. In this context, the literature review covers the propulsion cycle of the manual wheelchair, typically segmented into contact and recovery phases, with biomechanical studies investigating the forces applied, joint movements and propulsive efficiency in various techniques and conditions. Additionally, the review explores the movement of wheelchair users in real environments, emphasizing the diversity of terrains and the demands on muscular strength in daily activities, which underscores the importance of a dynamometer that reflects the practical use of the wheelchair. Finally, the application of MATLAB®/Simulink in the simulation of dynamometer models and biomechanical systems is discussed, highlighting its potential to model, analyze and optimize the design of the proposed dynamometer before its physical implementation. In summary, this review establishes the theoretical basis for the development of an accessible dynamometer, integrating knowledge about propulsion, real movements and computational simulation tools, aiming at a more pertinent assessment of the muscular strength of wheelchair users and, consequently, contributing to their physical assessment, rehabilitation and quality of life.

Keywords: Dynamometer; Manual wheelchair; Propulsion cycle; Movement of wheelchair users in real environments

RESUMO

Este artigo apresenta uma detalhada revisão bibliográfica sobre os principais aspéctos de um modelo experimental de dinamômetro de baixo custo destinado à avaliação da força muscular de usuários de cadeira de rodas. Nesse contexto, a análise da literatura abrange o ciclo de propulsão da cadeira de rodas manual, tipicamente segmentado em fases de contato e recuperação, com estudos biomecânicos investigando as forças aplicadas, os movimentos articulares e a eficiência propulsiva em variadas técnicas e condições. Adicionalmente, a revisão explora o movimento de cadeirantes em ambientes reais, enfatizando a diversidade dos terrenos e as demandas de força muscular em atividades cotidianas, o que sublinha a importância de um dinamômetro que reflita o uso prático da cadeira de rodas. Por fim, discute-se a aplicação do MATLAB®/Simulink na simulação de modelos de dinamômetros e sistemas biomecânicos, evidenciando seu potencial para modelar, analisar e otimizar o design do dinamômetro proposto antes de sua implementação física. Em síntese, esta revisão estabelece o fundamento teórico para o desenvolvimento de um dinamômetro acessível, integrando o conhecimento sobre a propulsão, os movimentos reais e as ferramentas de simulação computacional, visando uma avaliação mais pertinente da força muscular de cadeirantes e, consequentemente, contribuindo para sua avaliação física, reabilitação e qualidade de vida.

Palavras-chave: Dinamômetro; Cadeira de rodas manual; Ciclo de propulsão; Movimento de cadeirantes em ambientes reais

1 INTRODUCTION AND MOTIVATION

A cost-effective dynamometer for measuring muscle strength in wheelchair users is an innovative tool that could improve functional assessment, rehabilitation monitoring, and clinical research in this group. The ability to measure muscle strength in an affordable way could have a profound impact on the quality of life of people with limited mobility, providing objective data to help design personalized interventions and track progress over time (Oliveira et al., 2017). The aim of this paper is to develop a functional, low-cost dynamometer using affordable materials and open-source technologies, aiming to make muscle strength assessment more accessible to wheelchair users in different settings, from resource-poor rehabilitation clinics to homes. The effectiveness and accuracy of this device could open new possibilities for improving care and promoting functional autonomy in this segment of the population.

Measuring muscle strength in wheelchair users goes beyond a simple physical assessment, directly impacting their quality of life and autonomy (Wieczorek and Sydor, 2024; Wieczorek and Warguła, 2019). Upper limb and trunk strength is crucial for daily activities such as transfers (bed-to-chair, chair-to-toilet), wheelchair propulsion, reaching objects, personal hygiene, and participation in social and professional activities. Regular and accurate assessments allow for the identification of muscle deficits, monitoring the progression of neuromuscular conditions, quantifying the effectiveness of rehabilitation interventions, and providing objective data for the development of individualized treatment plans. Furthermore, information on muscle strength can be an important indicator of overall health, influencing the prevention of secondary injuries, such as shoulder pain frequently reported by wheelchair users due to muscle imbalances.

In the context of the Brazilian health system, there is a significant gap in the availability of specific equipment for assessing muscle strength in wheelchair users. Conventional dynamometers, often designed for individuals without mobility restrictions, may be inadequate or inaccessible for this population. The lack of low-cost and portable equipment further aggravates this situation, especially in resource-limited rehabilitation clinics and in research settings with restricted budgets. This scarcity of assessment tools contributes to an underestimation of the functional capacity of wheelchair users, making it difficult to implement optimized rehabilitation strategies and collect robust data for clinical research in Brazil. The creation of a low-cost dynamometer adapted to the needs of wheelchair users could therefore fill an important gap in the health system, democratizing access to an assessment essential for improving the quality of life of this population.

The development of a low-cost dynamometer prototype is a powerful pedagogical and research tool within the university environment, offering significant benefits for the training of future engineers. For undergraduate students, building a dynamometer represents a unique opportunity to solidify the theoretical concepts learned in the classroom through practical application. Theory on mechanics of materials, electronics, signal processing and instrumentation comes to life when students are faced with the real challenge of designing and building a functional measurement system. They learn to translate abstract principles into tangible solutions, understanding the limitations and challenges inherent in the real world of engineering. The team project inherent in the development of a complex prototype such as a dynamometer fosters collaborative skills essential for professional life. Students learn to divide tasks, communicate ideas, negotiate solutions, manage conflicts and integrate different expertise to achieve a common goal. This teamwork experience simulates the professional environment, preparing them for the challenges of multidisciplinary collaboration.

Direct contact with sensors and actuators, which are fundamental components of measurement and control systems, provides invaluable practical learning. Students experience the operation of different types of sensors for force measurement, understand the need for calibration and treatment of noisy signals, and explore the possibility of incorporating actuators to automate testing. This hands-on interaction with the hardware demystifies the technology and develops engineering intuition.

Instrumentation, which involves the selection, connection and configuration of sensors, the development of signal conditioning circuits and data acquisition, becomes a tangible skill through this project. Students learn how to use data acquisition and analysis software, interpret results and evaluate the accuracy and reliability of measurements. This practical experience with instrumentation is essential for working in several areas of engineering.

In short, the development of a low-cost dynamometer prototype offers a rich hands-on learning experience that complements theory, develops teamwork skills, provides direct contact with instrumentation technologies, and reinforces the importance of practical application in the training of competent and innovative engineers. In addition, the project can generate valuable research results and contribute to solving real-world problems, such as the lack of affordable equipment for assessing muscle strength in specific populations.

2 LITERATURE REVIEW

The wheelchair dynamometer is a specialized piece of equipment designed to measure the forces applied by the user when propelling the wheelchair (Martin et al., 2002) . It has become an indispensable tool in a variety of areas, from biomechanical research and the development of assistive technologies to clinical assessment and training of Paralympic athletes.

Manually propelling a wheelchair is a complex activity that involves interaction between the user, the chair, and the environment. The efficiency and effectiveness of this propulsion depend on several factors, including the user's muscular strength, the pushing technique, the characteristics of the wheelchair (weight, wheel bearing, ergonomics) and the terrain conditions. Understanding and quantifying the forces involved in this process are crucial to optimizing performance, preventing injuries, and improving the quality of life of wheelchair users. It is in this context that the manual wheelchair dynamometer stands out as a fundamental measuring instrument.

Basically, a dynamometer is a device used to measure forces. In the specific case of manual wheelchairs, the dynamometer is designed to quantify the propulsion forces that the user applies to the propulsion rims.

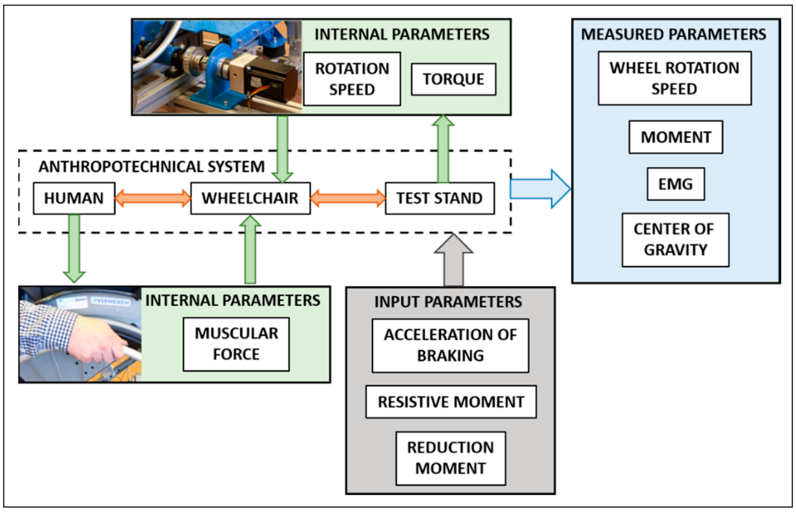
There are different types of dynamometers for manual wheelchairs, varying in terms of measurement technology, complexity, and specific applications. Some models are attached directly to the wheelchair, replacing or integrating with the propulsion rims. Others are platforms or rollers where the wheelchair is positioned to perform the tests Brouha and Krobath (1967); Glaser et al. (1979); Horvat et al. (1984).

In their 2019 study (De Klerk et al., 2019) , De Klerk et al. compared 50 wheelchair ergometers, emphasizing the benefits and drawbacks of various ergometers employed for laboratory-based wheelchair propulsion testing. Their findings indicated that each ergometer type presents both advantages and limitations, suggesting that the selection of equipment should be guided by specific research objectives and needs. The authors also underscored the existing diversity and lack of agreement in this area, pointing to the necessity for continued research and standardization initiatives to enhance the utility of laboratory assessments of wheelchair propulsion. Nevertheless, it is important to note that roller ergometers permit the use of an individual's own wheelchair, which can be particularly valuable when investigating the interaction between a specific user and their wheelchair (Wieczorek and Sydor, 2024).

Wheelchair treadmills are frequently employed in the investigation of wheelchair propulsion performance. Distinct from conventional treadmills intended for runners, wheelchair treadmills incorporate a platform designed to accommodate a wheelchair, allowing users to propel themselves against the moving belt while seated. These specialized treadmills offer adjustability to replicate diverse terrains, encompassing inclines and level surfaces (Baumgart et al., 2020). Wheelchair treadmills present an effective substitute for instrumented push-rims in the measurement of temporal and kinetic parameters. Research has established significant correlations between data acquired from treadmills and during real-world wheelchair use (Gagnon et al., 2016).

As part of the scientific endeavors at Poznan University of Technology in Poland, a stationary roller dynamometer was created for the purpose of measuring wheelchair drive characteristics, including torque, speed, and energy consumption as they change over time. This specialized testing apparatus was patented, constructed, and its validity confirmed (Wieczorek et al., 2019). This test stand records the rotational parameters of a wheelchair's driving wheels and possesses the capability to simulate inclined surfaces. Furthermore, the stand is equipped with sensors to measure key biomechanical parameters associated with propelling the wheelchair. Figure 1 provides an illustration of the information flow within the dynamometer system.

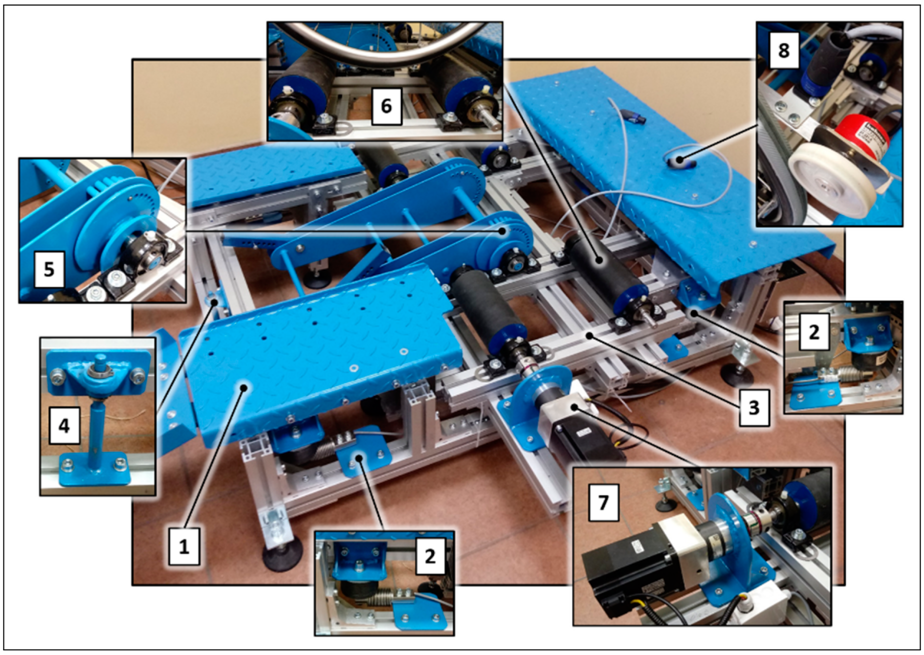
Figure 1 – Information flow diagram for wheelchair propulsion biomechanics study on the stationary roller dynamometer



Source: (Wieczorek and Sydor, 2024)

The stationary roller dynamometer illustrated in Figure 2 comprises a support frame (1) to which a weighing platform (3) is affixed utilizing strain gauges (2). These strain gauges measure the pressure exerted under each wheel of the wheelchair, and from this data, the position of the center of mass (COM) of the combined human–wheelchair system is calculated. Given the susceptibility of strain gauge scales to longitudinal forces, three linear guides (4) are incorporated to permit only vertical movement of the weighing scales relative to the support frame. The weighing platform is equipped with a wheelchair frame securing mechanism (5) and two double traction rollers (6) that ensure a slip-free interface with the wheelchair's driving wheels, a crucial element for precise testing.

Figure 2 – View of the stationary roller dynamometer: 1 – support frame, 2 – strain gauges, 3 – weighing pan, 4 – linear guides, 5 – clamping system, 6 – traction rollers, 7 – BLDC motor, 8 – encoder

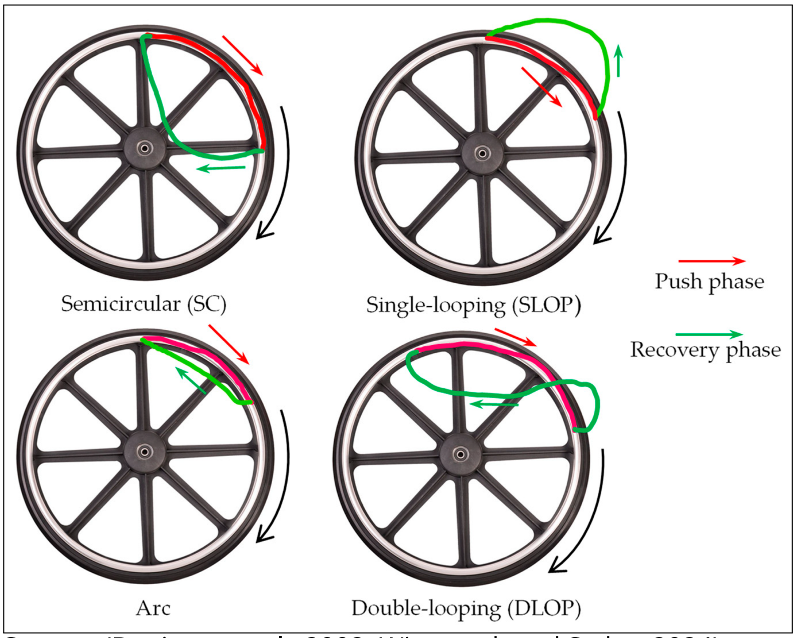


Source: (Wieczorek and Sydor, 2024)

2.1 The propulsion cycle of a manual wheelchair

The kinematics of human body segments represents a fundamental aspect of biomechanics (Boninger et al., 2002; Masse and Lamontagne, 1992; Rao et al., 1996; Silva et al., 2019). The propulsion cycle of a manual wheelchair can be delineated into two distinct phases: the push phase and the recovery phase (Rodrigues da Silva et al., 2023). During the push phase, mechanical energy is transmitted to the wheel's push-rim via hand contact; conversely, during the recovery phase, the hand is repositioned in anticipation of the subsequent push. Throughout the push phase, the hand moves in conjunction with the rim. In the recovery phase, the hand can follow various trajectories, which can be categorized based on the form of the hand's projection onto the sagittal plane (Shimada et al., 1998). The existing literature identifies four primary stroke patterns: semicircular, single-looping, arc, and double-looping (Boninger et al., 2002). These stroke patterns are visually presented in Figure 3.

Figure 3 – Stroke patterns of the wheelchair propulsion

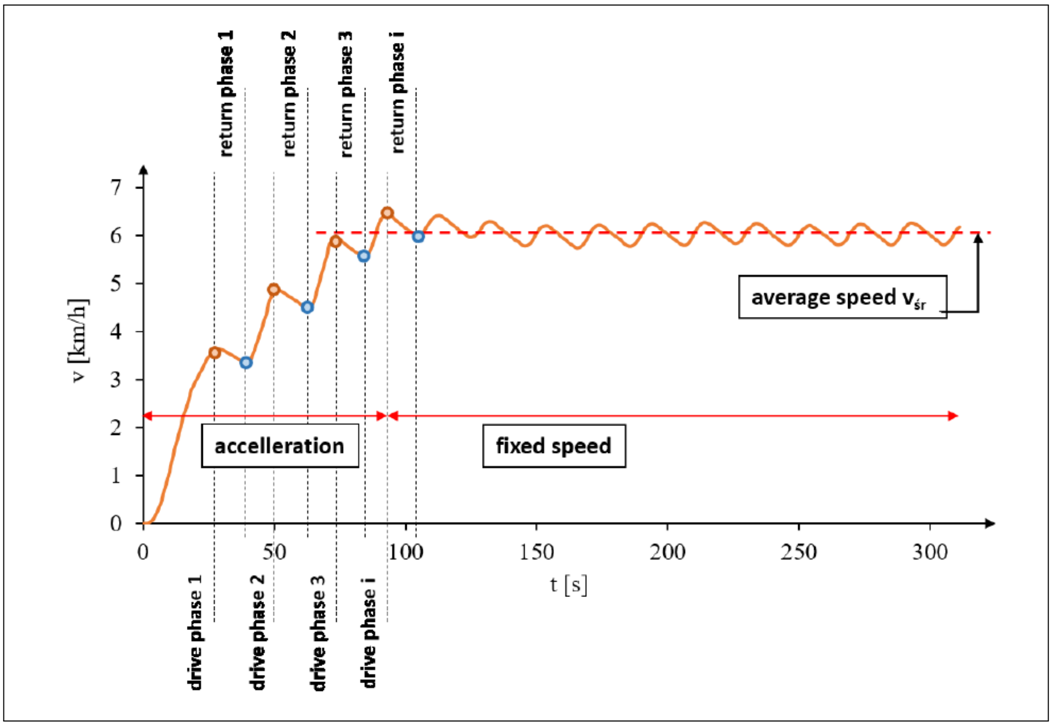


Source: (Boninger et al., 2002; Wieczorek and Sydor, 2024)

2.2 Wheelchair user movement in real life conditions

In typical real-world scenarios, wheelchair movement encompasses three distinct stages: acceleration, a period of constant velocity, and deceleration (Goosey-Tolfrey and Moss, 2005). As depicted in Figure 4, these phases influence the wheelchair's speed at any given moment. Furthermore, the acceleration and constant motion phases can be further divided into drive and return phases (Shimada et al., 1998; Vanlandewijck et al., 2001). During the drive phase, the wheelchair user's upper limb is in contact with the propulsion mechanism, simultaneously transferring muscular power (de Barros Lombardi Jr and Dedini, 2009) to the drive wheel, which generates the wheelchair's propulsion force. The application of this propulsion force results in positive acceleration of the system.

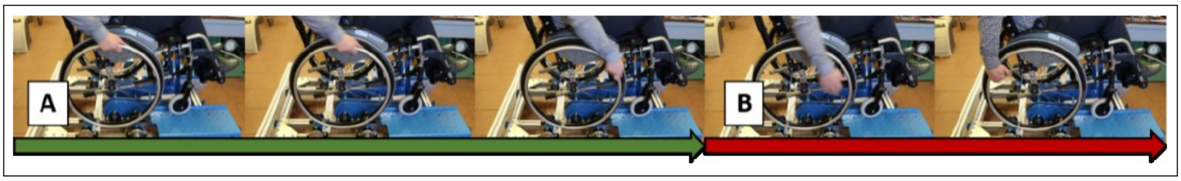
Figure 4 – Change of speed over time during wheelchair propulsion



Source: (Wieczorek and Warguła, 2019)

During the return phase, the wheelchair user's upper limb moves back to its initial position (Figure 5). The drive phase concludes once the propulsion element has rotated through a specific angle. The magnitude of this angle is determined by the operator's upper extremity range of motion and the geometric specifications of the wheelchair's propulsion system. Following the drive phase is the return phase, during which the operator's hand is brought back to its starting point.

Figure 5 – Time-lapse image of propulsion movement with an indication of the drive phase (A) and return phase (B)



Source: (Wieczorek and Warguła, 2019)

In the return phase, no propulsive force is produced; the wheelchair's movement is sustained by inertia and the energy stored during the preceding drive phase. It is also during this phase that the wheelchair experiences deceleration due to resistive forces. The resulting deceleration a_h reduces the wheelchair's velocity in the return phase. Consequently, the operator must initiate another drive phase to increase speed or maintain a consistent average driving speed v_{sr}. It is important to emphasize that in real-world conditions, the magnitude of this deceleration and the duration of the return phase do not cause the wheelchair to come to a complete stop. Instead, they result in only a small percentage decrease in speed compared to the wheelchair's peak speed achieved during the drive phase. As a result, the operator can accelerate the wheelchair to a desired average speed and then sustain this speed over an extended period.

If the drive phase is not repeated for a sufficiently long interval during the return phase, the wheelchair's speed can potentially decrease to zero. This is attributable to the decelerating forces eventually bringing the wheelchair to a stop 1. The duration of the return phase t_{fp} can increase depending on the magnitude of this deceleration. The extent of this deceleration is influenced by the environment in which the wheelchair is used, as this determines the strength of the opposing forces (Kwarciak et al., 2009):

$$v(t) = v_0^i - a_h^i t_{fp} \tag{1}$$

where:

$v(t)$ - actual speed of the wheelchair,

v_0^i - linear speed of the wheelchair on completion of the i-Th, propulsion phase,

a_h^i - acceleration of deceleration in the i-Th return phase,

t_{fp} - duration of propulsion phase.

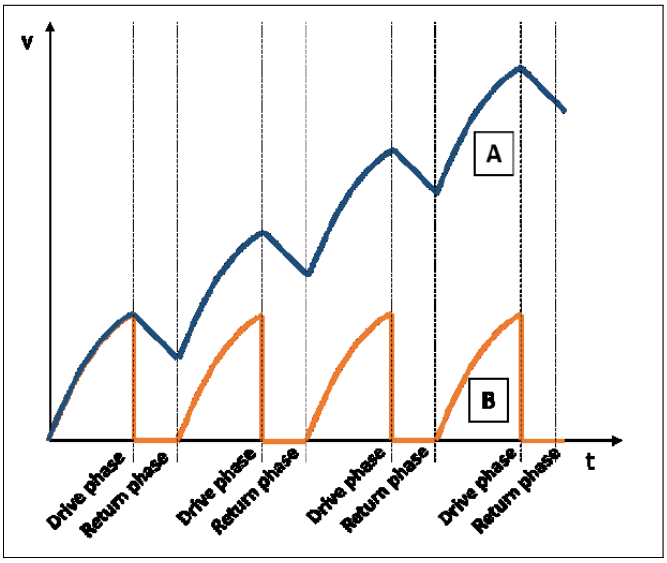
It is important to emphasize that when a wheelchair travels on a level surface, the deceleration becomes zero once the wheelchair has stopped. However, in the scenario of uphill movement, after forward motion ceases, deceleration will cause the wheelchair to roll backwards.

Investigating wheelchair propulsion on a test stand allows for the propulsion of the vehicle's wheels without influencing its linear displacement. Consequently, no inertia force resulting from the wheelchair's acceleration during the drive phase is present in the system. Inertia force is one of the forces that act on the wheelchair, maintaining its movement during the return phase. During wheelchair testing on a test stand, the sole force sustaining the drive wheel's rotation during the return phase is the energy derived from the inertia of the rotating drive wheel itself. The design of the test stand incorporates elements that introduce additional resistive forces to the human-machine system beyond those encountered in real-world conditions, while simultaneously eliminating the effect of the wheelchair's inertia force. As a result, during the drive phase, the system cannot accumulate sufficient energy to maintain the drive wheel's revolutions throughout the return phase, as occurs during actual use equation 2.

$$v(t) = \omega(t) \cdot R_D \cdot 60 \quad (2)$$

The influence of the absence of inertia force during testing and the diminished internal resistances of the test stand is illustrated in the graph showing the theoretical speed variation over time (Figure 6). The graph depicts a scenario where speed increases during the drive phase and then abruptly decreases to zero immediately upon its completion (Figure 6 - B). This phenomenon is attributed to the presence of internal resistances within the test stand system and the lack of inertia force in the linear movement.

Figure 6 – Graph of the wheelchair's theoretical speed under real life conditions (A) and a the test stand with no compensation for the lack of inertia (B)



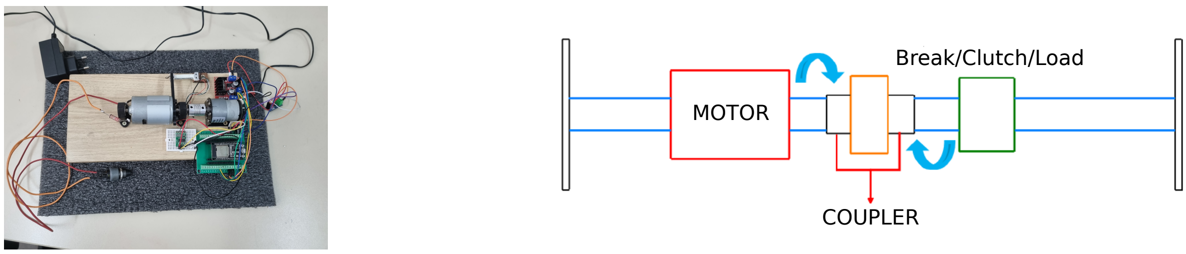
Source: (Wieczorek and Warguła, 2019)

3 RESULTS AND DISCUSSION

3.1 Prototype characteristics

The prototype of the dynamometer can be seen in Figure 7. A DC motor with an encoder is coupled to a second DC motor to which a load is added. Sensors to measure force (load cell) and current (ACS712) are added. An H-bridge (L298D) is used to control the speed of the DC motor with the encoder. A push-button switch is used to simulate the load.

Figure 7 – Dynamometer prototype



(a) Dynamometer prototype (b) Dynamometer schematic

Source: Authors and (Ashwindran et al., 2023)

3.2 Speed Control

In our experiment, it is considered that the speed of the wheelchair user is constant; for this stage, a digital PI control was implemented in the DC motor with encoder.

3.2.1 Model of DC Motor (Speed Control)

A mathematical modeling of the DC motor is presented in detail in (Maheriya and Parikh, 2016; Rahman and Yahya, 2021), the transfer function that relates the angular velocity to the voltage is

$$\frac{\dot{\theta}(s)}{V(s)} = \frac{K}{LJs^2 + (RJ + LB)s + RB + K^2}. \tag{3}$$

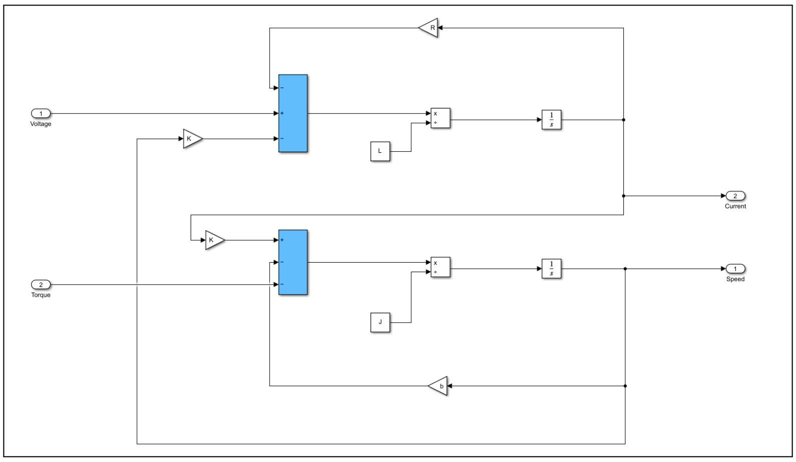
where, J rotor's moment of inertia ($kg \cdot m^2$), B motor's viscous friction constant ($N \cdot m \cdot s$), K electromotive force constant ($V/rad/s$), R electric resistance ((Ω)) and (L) electric inductance (mH).

The DC motor used in the experiment has the following characteristics: $12\ V$ power supply, $30:1$ metal gearmotor, 11 PPR encoder, 285 rpm ($30\ rad/s$).

The values of the parameters of the model defined in 3 are: $K = \dfrac{v(t)}{\dot{\theta}(t)} = \dfrac{12}{30} = 0.40\ V/rad/s$, $R = 4\Omega$ and $L = 2.3\ mH$. The J and B parameters are very difficult to obtain accurately, to get around this, we will use the MATLAB®/Simulink™parameter estimation toolbox (Mathworks, 2025).

To estimate the parameters, the linear model shown in Figure 8 is used to capture the dynamics of the DC motor.

Figure 8 – Linear DC motor model



Source: Authors (May 2025)

After open-loop testing of a real DC motor and using the parameter estimator toolbox, we obtain the following values:

$$J = 3.8335 \times 10^{-6}, \quad B = 0.1868,$$

And the open-loop transfer function of the DC motor after substituting the parameter values in 3 is

$$P(s) = \frac{0.4037}{8.567 \times 10^{-10} s^2 + 5.725 \times 10^{-5} s + 0.918} \tag{4}$$

3.3 Controller Design C(s)

Now we can proceed with the design of the C(s) controller using classical control techniques such as root locus, frequency response analysis, and practical methods.

Most conventional DC motors have their output speed regulated using a PI controller:

$$C(s) = \frac{U(s)}{E(s)} = K_p + \frac{K_i}{s}. \tag{5}$$

In this article, the MATLAB® **pidTuner** function is used to design the PI controller. The transient and steady-state performance criteria that are to be met are:

- settling time less than 2 seconds.
- overshoots and undershoots should be under 5%.
- steady-state error should be less than 3%.

After using the **pidTuner** function to meet the criteria mentioned, the following PI controller is obtained:

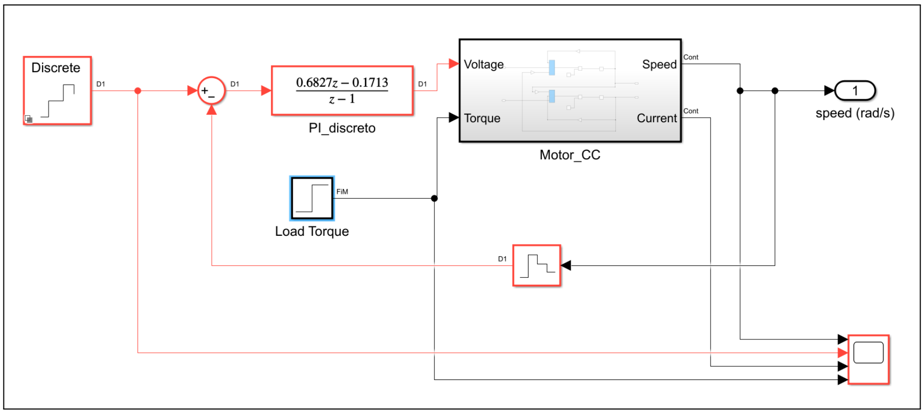
$$C(s) = \frac{U(s)}{E(s)} = 0.427 + \frac{5.114}{s}. \tag{6}$$

For discretization of 6, bilinear transformation is used (also known Tustin's method), with sample time ($T = 0.1$ seconds) resulting in

$$C(z) = \frac{U(z)}{E(z)} = \frac{0.6827z - 0.1713}{z - 1} = \frac{0.6827 - 0.1713z^{-1}}{1 - z^{-1}}. \tag{7}$$

A schematic of the control design in MATLAB®/Simulink is shown in Figure 9

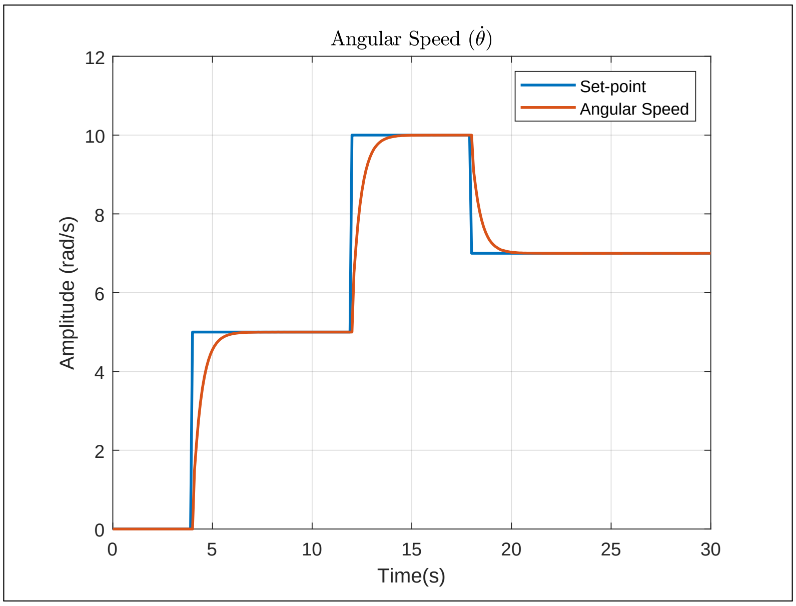
Figure 9 – A schematic of the control design



Source: Authors, May 2025

Applying the digital PI controller defined in 7, the behavior of the closed-loop system can be seen in Figure 10.

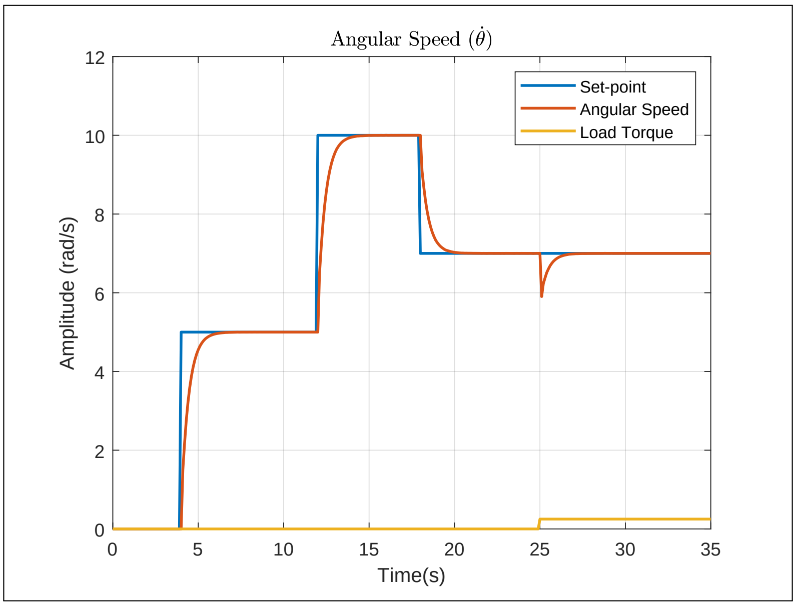
Figure 10 – Closed-loop system behavior for step-type inputs



Source: Authors, May 2025

To simulate the behavior of the wheelchair user, a step-type load simulation of $0.25\ N \cdot m$ was applied at time $t = 25$ seconds; the behavior of the closed-loop system is shown in Figure 11. In future work, a deeper look at the behavior of the load imposed by the wheelchair user will be evaluated and a digital load control will be applied to simulate various scenarios.

Figure 11 – Closed-loop system behavior for an input torque load



Source: Authors, May 2025

4 CONCLUSION

The literature review presented demonstrates the relevance and feasibility of developing a low-cost dynamometer for assessing the muscular strength of wheelchair users. By integrating detailed knowledge about the propulsion cycle of manual wheelchairs, understanding the biomechanical demands imposed by movements in real conditions, and the potential of computer simulation via MATLAB®/Simulink, a solid foundation is established for the creation of an accessible and relevant device. This device would have the potential to positively impact the physical assessment, rehabilitation processes, and, ultimately, the quality of life of wheelchair users, offering a practical and economical tool to quantify crucial aspects of their functional capacity. As future work, all the budgetary part necessary for the construction of the test prototype and the full-size dynamometer will be presented, as well as the comparison with the models present on the market. The entire measurement part of quantities such as torque and power will be evaluated, as well as the instrumentation, amplification, and filtering of sensor signals. A load control will be implemented to simulate various load scenarios that resemble those of the wheelchair user.

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