Can an ergometer based virtual reality environment reproduce typical real-world wheelchair manoeuvring?

by

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Abstract

Virtual reality together with a wheelchair ergometer provide the perfect setting for conducting research involving wheelchair users, rehabilitation for inexperienced wheelchair users, and training wheelchair athletes. The target of this study was to develop a wheelchair ergometer-based virtual reality system that could mimic the biomechanics of real-world wheelchair manoeuvring. In an attempt to do so, a linear inertia system based on a roller ergometer was built based on a thorough biomechanical analysis, and was further equipped with a cube virtual reality environment. In addition, in a developmental manner, three approaches were taken to simulate wheelchair turning, forming three final systems. System I only accommodates linear inertia, assuming the additional inertial effects associated with rotation can be neglected, while system II and III provide compensation for rotational inertia, mechanically and perceptually, in addition to linear inertia. The mechanical system uses a pneumatic braking to add real time resistance against turning, while system III slows down the rotations in virtual reality, corresponding to the rate of turning that happens in the real world where rotational inertia is present.

To test the validity and reliability of the systems developed, 15 able-bodied participants were recruited, each performing a series of Illinois agility tests both in RW and in the VR environment. The test-retest reliability of using this agility test was also examined in this study, based on a comprehensive biomechanical analysis

assessing 53 variables in total. 94% of the variables tested showed good to excellent ICC, and none of the single-value parameters tested showed a meaningful difference from session one to session two, where the two sessions were at least one week apart (10 days in average).

The main objective of this study was to show the reliability and validity of the virtual reality systems developed. Since inexperienced, able-bodied persons using a wheelchair are less likely to be consistent in the way they propel a wheelchair, recruiting them for this study was justifiable. If we could show that inexperienced wheelchair users were consistent from one session to another, and between real and virtual worlds, we could also assume that experienced wheelchair users would also be consistent between these two environments.

Test-retest reliability and construct validity of the three systems were examined, using, respectively, ICC for VR systems tried in two different sessions, and Pearson correlation between the data obtained from VR system tests and real-world tests, conducted in the same session. 53 biomechanical variables were tested in total, and as a result, all three systems showed good validity for many of the variables tested. Also, a very good reliability for the System II and good reliability factors for Systems I and III was observed for most of the variables tested. In general, System II was biomechanically the best system while System III delivered a better VR experience. This study also revealed that to finish a manoeuvring task in VR, people use more time, shorter pushes, less pushes, more tangential forces, and less backward pushes relative to real world.

To prepare the participants for the experiments, they underwent up to four training sessions to acclimatize to motion sickness. The results of this study showed that a

maximum of four sessions held on different days was enough to resolve motion sickness or reduce the susceptibility to it to a well tolerated level. Furthermore, the Motion Sickness Assessment Questionnaire and the Igroup Presence Questionnaire were used to assess participants' motion sickness and feeling of presence in the VR. It was observed that all of the three systems presented relatively low motion sickness and high virtual presence. Also, a meaningful inverse relationship between motion sickness and VR presence was observed.

Preface

The materials presented in this work are original research conducted in Rehabilitation Robotics Lab, University of Alberta, led by Dr. Ferguson-Pell, except for Chapter 3 which is a review paper. This thesis is formatted as "paper-based". The systems developed for this thesis were designed and developed by me with the help of Dr. Martin Ferguson-Pell and Kenton Hamaluik.

The original idea for Chapter 3 was from a coursework instructed by Dr. Susan Armijo Olivo. This chapter has been published as "Zohreh Salimi and Martin W. Ferguson-Pell (2017). Validity in Rehabilitation Research: Description and Classification, Physical Disabilities - Therapeutic Implications, Prof. Uner Tan (Ed.), InTech, DOI: 10.5772/67389. Available from: https://www.intechopen.com/books/physical-disabilities-therapeutic-implications/validity-in-rehabilitation-research-description-and-classification". Dr. Pamela Bentley helped with editing this chapter.

Chapter 4 summarizes the results of a study conducted in the Rehabilitation Robotics laboratory. People who helped with the study are: Dr. Martin Ferguson-Pell, Dr. Liping Qi, Zohreh Salimi, Dr. Ailar Ramadi, Dr. Robert Haennel, Jiajie Wu, and Zhifu Liu. The paper was written by Dr. Liping Qi. I helped with the study design, preparing the experiment setup, conducting the experiments, data analysis, and reviewing the paper. This study received research ethics approval from University of Alberta Research Ethics Board, Project: "Shoulder muscle function during wheelchair propulsion, comparing performance in real world and laboratory settings", No. Pro00003315_REN4, Date: 5/14/2012. This paper has been published as "Wheelchair users' perceived exertion during typical mobility activities" by Liping Qi, Martin Ferguson-Pell, Zohreh Salimi, Robert Haennel, and Ailar Ramadi in Spinal Cord (Nature). 2015;53(MARCH):1-5. doi:10.1038/sc.2015.30".

Chapter 5 has been published as "Ergometers Can Now Bio-mechanically Replicate Straight-Line Floor Wheelchair Propulsion: Three Models Are

Presented” by Zohreh Salimi and Martin Ferguson-Pell (2013, Nov. 15-21), in Proceedings of the 2013 ASME international mechanical engineering congress and exposition (IMECE), San Diego, CA, USA.

Chapter 6 has been submitted as “Development of three inertia-compensated wheelchair ergometers for curvilinear manual wheelchair propulsion using virtual reality” by Zohreh Salimi and Martin Ferguson-Pell to IEEE Transactions on Neural Systems and Rehabilitation Engineering. Chapters 7 to 9 are produced using results from experiments performed by me. The ethics application for these experiments was approved by University of Alberta Research Ethics Board, Project Name “Can simulation be used to accurately represent real-life wheelchair propulsion?”, No. Pro00003315 _AME5, Date: 9/25/2015.

Chapter 7 has been submitted as “Investigating the test-retest reliability of Illinois Agility Test for wheelchair users” by Zohreh Salimi and Martin Ferguson-Pell to Journal of Disability and Rehabilitation.

Some of the material presented here were presented in conferences, as below:

Peer-Reviewed Conference Abstracts and Extended Abstracts, Posters and Oral Presentations

Main author and the presenting author is <u>underlined</u>, supervisors are shown in *Italic,* corresponding author is indicated by a star (*), and equal contributions indicated using (†).

1. **<u>Zohreh Salimi</u>**, *Ferguson-Pell M**, (2016, Oct, 21-23) “Reliability of Illinois Agility Test for wheelchair users”, in 17th Biomedical Engineering conference (BME) Program and Proceedings, Banff, Alberta, Canada.

   I wrote and planned the manuscript, formulated ideas and methods, conducted the experiments, and analyzed the data. Abstract and poster presentation.

2. **Zohreh Salimi**, *M. W. Ferguson-Pell**, Ailar Ramadi, Robert Haennel, Farhood Mohammadi, and Liping Qi, (Oct, 24-26, 2014) “Investigating validation of physiological cost index for wheelchair users”, in 15th Biomedical Engineering conference (BME) Program and Proceedings, Banff, Alberta, Canada, p. 73.

I wrote and planned the manuscript, helped formulate ideas and methods, helped conducting experiments, and mainly analyzed the data. Abstract and poster presentation.

3. **Zohreh Salimi**, *M. W. Ferguson-Pell**, (2012, Oct. 19-21) “Ergometer Model for Representing Straight-Line Floor Wheelchair Propulsion”, in 13th Annual Alberta Biomedical Engineering Conference (BME) Program and Proceedings, Banff, Alberta, Canada, P.79.

   I wrote and planned the manuscript, formulated ideas and methods, did the calculations and designing. Abstract and poster presentation.

Non-Peer-Reviewed/Invited Abstract/talk

1. **Zohreh Salimi**, *Ferguson-Pell M**, (2016, Oct, 20) “Reliability of Illinois Agility Test for wheelchair users”. Glenrose Rehabilitation Hospital Spotlight on Research Breakfast & Symposium, Edmonton, Alberta, Canada.

   I wrote and planned the manuscript, formulated ideas and methods, conducted the experiments, and analyzed the data. Abstract and poster presentation.

2. **Zohreh Salimi**, *Ferguson-Pell M**, (2015, Oct, 21) “Learning effect in performing wheelchair slalom courses, in real world and in virtual reality environments”. Glenrose Rehabilitation Hospital Spotlight on Research Breakfast & Symposium, Edmonton, Alberta, Canada.

   I wrote and planned the manuscript, formulated ideas and methods, conducted the experiments, and analyzed the data. Abstract and poster presentation.

3. **Zohreh Salimi**, *M. W. Ferguson-Pell**, Ailar Ramadi, Robert Haennel, Farhood Mohammadi, and Liping Qi, (May, 2015) “Investigating validation of physiological cost index for wheelchair users”, in 3rd summer camp at University of Alberta.

   I wrote and planned the manuscript, helped formulate ideas and methods, helped conducting experiments, and mainly analyzed the data. Poster presentation.

4. **Zohreh Salimi**, *M. W. Ferguson-Pell**, (2014, Oct, 15) "Physiological Cost Index (PCI): is it valid to assess wheelchair propulsion?" Glenrose Rehabilitation Hospital Spotlight on Research Breakfast & Symposium, Edmonton, Alberta, Canada.

   I wrote and planned the manuscript, helped formulate ideas and methods, helped conducting experiments, and mainly analyzed the data. Poster presentation.

To my husband

Meisam, you are the meaning of a true life-companion!

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First and foremost, I need to express my sincere gratitude to my supervisor, Professor Martin Ferguson-Pell, for helping me throughout my PhD program. Without his help and continuous support and guidance, this project would have never been completed this way. Not only he made sure that I have the support I needed for my research, but also he genuinely tried to train me to be an independent researcher and a future supervisor. Other than all this, the kindness of him and his lovely wife, Peggi, has always been heart-warming.

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Beside my committee members, I would like to thank my fellow lab-mates, especially Kenton Hamaluik who greatly helped me in development of the VR systems. Dr. Pamela Bentley, Dr. Liping Qi, Jiajie Wu, Maliheh Hadizadeh, Alain Clark, Dr. Martha Funabashi, Yolanda Casciaro, Peter Jun, and Nooshin Jafari are the others I particularly would like to name and thank, as they made memorable days for me during my PhD. I also appreciate all the help that the faculty members and staff in Faculty of Rehabilitation Science gave me throughout this program, especially Angela Libutti, Jodi Lopetinsky, Dr. Tammy Hooper, and Dr. Susan Armijo-olivo. Finally, I thank all my participants who took the time to take part in several sessions of my experiments.

Last but not the least, I would like to send my special thanks to my loving husband, Meisam Nouri. Without him completing this thesis in this way and in this time frame would have been impossible. He was always there for me, especially in taking care of our lovely newborn: Nicou! I also would like to profoundly thank my parents and my siblings for supporting me spiritually throughout writing this thesis and my life in general, and also for their patience during these 68 months while I was far away for my PhD.

The world encaptured by those nine blue heavens, is like a chaff floating in the ocean.

Take a time and have a look! Where are you on this chaff? One should sarcastically smile at self!

- Avecina

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LIST OF ABBREVIATIONS AND ACRONYMS

ADL	Activities of Daily Living
BOM	Bill of Material
cRio	compact Rio
CTS	Carpal Tunnel Syndrome
DAQ	Data Acquisition
EMG	Electromyography
FBD	Free Body Diagram
IPQ	Igroup Presence Questionnaire
MSAQ	Motion sickness Assessment Questionnaire
RAP	Rear Axle Position
RPE	Rating of Perceived Exertion
RR	Rolling Resistance
RW	Real World
SCI	Spinal Cord Injury
SLP	Straight-Line Propulsion
VIMS	Visually-Induced Motion Sickness
VR	Virtual Reality
VR_sysI	Virtual Reality System I
VR_sysII	Virtual Reality System II
VR_sysIII	Virtual Reality System III

The following chapter provides an introduction to the thesis; some background and motivation to conducting the research, the research question and the hypotheses, and description of chapters are included. This chapter explains how the chapters of this thesis are related and explains the rational behind them.

1 INTRODUCTION

1.1 Motivation

Wheelchair users comprise about 1% of the global population [1] and many of them suffer from shoulder injuries due to over-use of the structures of the upper extremities during wheelchair propulsion and transfers. These problems, however, could be reduced through improved wheelchair design, training in propulsion techniques and changes to the built environment. Researchers have conducted wheelchair-related studies for decades to improve the quality of life for wheelchair users, including ways of preventing and reducing their shoulder pain. Much of this research has been conducted in highly constrained environments, and most particularly in laboratories using ergometers, which are treadmill-like devices that are specialized for wheelchair users. Unfortunately, these studies only hint at the biomechanical and physiological challenges of wheelchair propulsion while undertaking typical real-world activities. One of the consequences has been that nearly all studies on wheelchair propulsion biomechanics have concentrated on straight-line propulsion on flat surfaces against modest resistance. Little is known about the joint forces generated during non-straight-line manoeuvres, yet manoeuvrability is considered by many wheelchair users to be a very important functional skill and key performance requirement for their wheelchair.

Research on wheelchair propulsion to date has involved simple environments pushing in a straight line using a battery of instrumentation to provide a comprehensive biomechanical analysis. The next step is to study test environments that are more typical of real life, including non-straight-line propulsion. Conducting studies in the real world (RW) is difficult due to the need to use a synchronized cluster of complex instrumentation in outdoor environments. An alternative, that can capture the physical challenges and task complexity of RW propulsion, is to conduct these studies using Virtual Reality (VR) environments that include sophisticated wheelchair ergometers.

VR is a recent technology that is helpful in the study of many different human activities, including wheelchair propulsion. For instance, it is increasingly being used for safely training for operating powered wheelchairs for newly injured wheelchair users, to prepare them for using their wheelchairs in real-life situations [2], [3]. There is a wide range of different VR settings, comprising immersive [2], [4] and non-immersive [3], [5], [6] virtual reality environments, including systems that engage participants as they undertake physical activities, such as CAREN (Computer Assisted Rehabilitation Environment) system [7] and the CEAL (Challenging Environment Assessment Laboratory) environment [8]. However, these systems do not provide a VR environment that accurately reproduces the biomechanical demands of wheelchair propulsion, including the effects of different surfaces, slopes and inertia.

An immersive virtual reality environment has a substantial influence on the participants' feeling of being in the RW [9] and helps them to act the same in the VR as they would in RW. Only a few examples to date have provided immersive VR [10]. By 2008 only non-immersive VR had been developed [5]. I used a commercial VR system (Eon Icube) that had been programed to represent RW tasks and was interfaced to a sophisticated ergometer.

The main purpose of this research was to determine if a virtual-reality-based laboratory can reproduce the biomechanics of RW wheelchair manoeuvring. For this purpose, I investigated the test-retest reliability and convergent validity of an immersive ergometer-based VR for manual wheelchair users. Using a flexible commercial VR system (Eon Icube Mobile) we are able to provide visual feedback for different tasks that are representative of wheelchair propelling in the RW. Inside the VR, a wheelchair ergometer was employed comprising instrumented rollers that are able to represent the biomechanical conditions of straight-line wheelchair propulsion in RW, including inertial effects and kinetic energy [11]. This system was further equipped to simulate wheelchair turning, using three methods. In the first method, the visual feedback from the VR was used to induce the sense of rotation to the participants, without adding extra resistance against rotation. In the second method, a pneumatic braking system was built to apply resistance against rotations in real time. Finally, the third method utilized a perceptual method to represent rotational inertia. Each of these methods form a VR system that was tested for their reliability and validity in representing wheelchair manoeuvring.

This study used a developmental approach, with participant recruitment (able-bodied subjects) started after developing the first system and as the recruitment went on, systems II and III were developed using the feedback received from the participants. It was intended here to demonstrate the feasibility of using VR to biomechanically replicate complex manoeuvring tasks in the real world using a laboratory environment designed for developing VR systems. The end goal of this research is to recommend the basis for a definitive study with a fully powered experiment design, by recruiting experienced wheelchair users.

Although the primary objective of the VR in Rehabilitation Robotics Lab is to use it for wheelchair biomechanics research, it has the potential to be also used for training manual wheelchair propulsion. These ergometer-based VR environments enable us to test manoeuvrability that is beneficial both for the elite athlete wheelchair users and also for every-day wheelchair users. For instance, VR provides a safe yet accurate environment for clinicians to train manual wheelchair propulsion to new wheelchair users to prepare them before stepping out into RW. Newly injured spinal cord injured (SCI) patients or children could use this motivating, safe, and supervised environment to get prepared to wheel in RW. Activities like being able to cross the traffic road quickly and safely, navigating thorough grainy and rough sidewalks, shopping (navigating through the aisles), and using transportation system (getting on and off a bus and parking in the reserved place for wheelchairs in the bus)), can be trained.

Also, Paralympic athletes may benefit from these ergometer-based VR environments to practice and be trained in a weather-independent and safe environment that can be used to repeat the same scenarios, if needed, for several times with a high precision. Furthermore, wheelchair athletes could be trained in the VR for specific tasks that they need close guidance and supervision for, such as shooting toward the basket in basketball. The coaches could have more effective supervision as they could observe different factors, such as the velocity of athlete's hand and its trajectory, and therefore provide the best guidance.

1.2 Research Question and Objectives

The key underlying question that motivated us to establish this study is:

How well can the biomechanical variables of manual wheelchair propulsion in a real-world obstacle course be reproduced by using an instrumented wheelchair ergometer plus visual feedback?

The objectives of this study are:

- To assess perceived exertion of experienced wheelchair users when performing propulsions on the ergometer with low to medium intensities and also during exertion
- To show the reliability of Illinois agility test in RW for wheelchair users
- To show the reliability and validity of the ergometer-based virtual reality environments developed in this study in reproducing the biomechanics of straight-line propulsion and wheelchair manoeuvring.
- To show the effectiveness of using a maximum of four training sessions in subsiding motion sickness to a well-tolerated level.
- To determine which of the three VR systems developed can better simulate the biomechanics of RW wheelchair manoeuvrings and provide a better VR experience for the users.

- To assess participants' sense of presence in the VR systems developed.

1.3 Hypotheses

The hypotheses of this thesis are listed here. The first one is the main hypothesis of this research.

H1: An instrumented wheelchair ergometer equipped with virtual reality is a reliable and validated device to reproduce bio-mechanical variables of real-world wheelchair manoeuvring.

H2: The Illinois agility test is reliable when testing wheelchair users in real world.

H3: Exposing the participants to a maximum of four training sessions with a comfortable length, held on different days, will substantially decrease the level of visually induced motion sickness to an easily-tolerated level.

1.4 Description of Chapters

This thesis is comprised of ten chapters that address the above objectives. First of all, the motivation and objectives of this thesis are discussed in this chapter. After this, a literature review on the topic of this study is presented in Chapter 2. In this chapter, the prevalence of wheelchair users and existence of upper extremity pain and injury among them will be addressed first, and then the classifications of spinal cord injury (SCI) and physiological and biomechanical aspects of SCI are presented. Wheelchair propulsion biomechanics is then discussed along with the important factors influencing wheelchair performance. The existing ergometers and the extent to which they have taken into account the biomechanical factors are discussed next. The VR and its advantages in helping research studies involving wheelchair users is addressed afterward, and the current research involved with VR are listed as well.

Chapter 3 presents a review on validity types, to establish why a validity study is needed for the VR environments developed for this research, and what kind of validity is concerned here.

Chapter 4 shows a practical application of the ergometer in a research study. In this chapter, real wheelchair users' rating of perceived exertion (RPE) when performing activities with everyday life intensities and during exertion is examined, in an attempt to characterize those activities with the RPE ratings, so that it can be used for self regulating activities by wheelchair users.

Chapter 5 discusses some of the underlying biomechanics of the study. In this chapter, the biomechanics of straight-line wheelchair propulsion (SLP) in RW is compared and corresponded to SLP on a roller-based ergometer. Six main biomechanical factors, namely velocity and acceleration, friction and rolling resistance, applied force, trunk swing, energy, and inertia, are examined and correlated between the two statuses. As a result, an equation is obtained that serves as the similarity condition that needs to be satisfied in order to have RW-like biomechanics for the SLP on an ergometer. Based on this, three models are presented for an ergometer that is representative of RW SLP.

Chapter 6 first introduces a more pragmatic model for a representative ergometer. Then, it goes one step further by employing VR to provide visual feedback for an ergometer built based on the forth model. This forms a VR system, VR_sysI, that can be used to simulate wheelchair manoeuvring by benefitting from the clinical values that VR adds. In the same vein, two more systems were also introduced here. VR_sysII uses a pneumatic braking system to change the rolling resistance in real time, proportional to the wheelchair turning, while VR_sysIII takes a perceptual approach and “shows” turning in the visual feedback that are similar to the turning of RW, where turnings are affected by rotational inertia.

Chapter 7 is the first step in testing the VR systems introduced, as the concept of the tests is the Illinois Agility Test (IAT) which is a standard agility test. The reliability of this test for wheelchair users needs to be established first, before further using it to examine VR systems. This chapter, therefore, focuses on assessing the reliability of IAT for wheelchair users when tested in two sessions that are, in average, 10 days apart.

Chapter 8 is concerned with empirically testing the reliability and validity of the three VR systems developed in this study. This was done by recruiting 15 subjects to participate in two sessions of tests. In each session, a series of IATs was performed both in RW and in (at least) one VR system.

Chapter 9 deals with the sense of presence in the VR systems as well as visually induced motion sickness (VIMS) that is often associated with simulators and VR. The 15 subjects recruited underwent up-to four training sessions of a self-selected length, before participating in the aforementioned two main visits. These training sessions were arranged to reduce participants' motion sickness so that the motion sickness does not bother them during the main tests. In this chapter, the effectiveness of these training sessions in reducing motion sickness is assessed as well as investigating participants' level of presence in the VR and their motion sickness during the main visits. Presence and elimination of nausea determines how effective the VR systems developed are in capturing the participants' attention and providing a pleasant and realistic VR experience.

Chapter 10 presents a summary of the results of this study, in addition to drawing the big picture of it. In this chapter, we discuss the relative advantage of the systems and discuss the merits of the most desirable system. Then, this chapter covers the limitation of this study that may have influenced the results. Finally, we suggest new directions for future studies.

1.5 Definition of Terms Used in This Thesis

Instrumented and sophisticated wheelchair ergometer: This term is used to refer to the wheelchair ergometer that we designed and constructed to be used in this study. In order to adjust for participants with different weights, we added a mechanism to both sides of the ergometer. This mechanism allows the inertia of the rollers to be equivalent to the inertia of each participant propelling on the ergometer.

Virtual Reality: In this study, we used virtual reality to simulate RW wheelchair propulsion, more specifically the Illinois agility test court, in the lab environment. It comprises of a cube with three walls and a floor, the images are projected on which. It also includes the software that was developed for this study. Using this software, we interfaced the participants' actions to what happens in the VR cube visual feedback.

Visual feedback: Our VR system comprises of a cube that participants are immersed in. The concept of “virtual reality” was shown to them through a projection they viewed on three walls (in front of the person and at both sides) and also on the floor. This visual feedback was provided by the four projectors on the ceiling of the cube.

Straight-line wheelchair propulsion: Propelling a wheelchair merely in a straight line, or simulation of it in VR environment

Non- straight-line wheelchair propulsion: Propelling a wheelchair on a sloped surface or turning, or simulation of them in VR environment

Wheelchair manoeuvring: Propelling a wheelchair in a curved line, or simulation of it in VR environment

A review of the relevant literature is presented to emphasize the need for conducting the research in this thesis.

2 LITERATURE REVIEW

In this chapter, the rationale for conducting the research reported in this thesis is discussed, based on the research studies completed to date. Firstly, evidence is provided for the effect of different levels of SCI on wheelchair users' biomechanical outcomes. The demography of wheelchair users and prevalence of secondary injuries among them is discussed next. Then the biomechanics of wheelchair propulsion is outlined, making the case for why it is necessary to conduct research studies on this topic. Three modalities for conducting wheelchair biomechanics studies are then introduced: treadmills, wheelchair ergometers, and Virtual Reality (VR). Each is discussed by explaining their differences, advantages, and disadvantages, along with presenting some examples of research studies that have used them. With this information in hand, the rationale for incorporating VR for such studies is then developed.

2.1 Effects of the Level of Injury on Biomechanics of Wheelchair Propulsion

Newsam *et al.* [12] studied the effect of level of injury on the cycle distance (distance travelled in each push) and velocity of propulsion on different surfaces. They recruited 70 wheelchair users and grouped them to 4 classes: low paraplegia (T10 to L3), high paraplegia (T1-T9), C7 quadriplegia, and C6 quadriplegia. According to their results, the low paraplegic group and the high paraplegic group were not significantly different in biomechanical factors (velocity, cycle distance, and cadence). The low paraplegic group had the highest absolute velocity and distance traveled, followed by the high paraplegic group, and then the C7 group. The C6 group was the slowest group with the highest velocity on tile equal to 78.5 m/min. The average velocity of an able-bodied person for community ambulation is 79 m/min (1.32 m/s) [13]. This means that a person with level C6 injury when using a manual wheelchair in the community has to work close to his/her maximum, even if the pathways have minimal resistances (no curb, steep surfaces, etc.). This result was confirmed later by Singh *et al.* [14]. It has been shown, however, that quadriplegic patients who are unable to propel a regular wheelchair for a long distance with functional speed are able to do so using a light-weight wheelchair [14]. Work performance has been also shown to be less in people with SCI, especially those who have higher levels of SCI [14], [15].

Another part of the above experiment that was published by Kulig *et al.* [16] revealed that persons with cervical-level of injury have significantly higher upward force in their shoulders during early push phase of propulsion than participants with high paraplegia. This could be an indication that when quadriplegic persons try to apply the initial upward force to the pushrim, they compensate for the lack of grasp of hand by applying a more upward force that could, in turn, cause a more upward force in the shoulder. This, in turn, increases the risk of shoulder impingement, as explained by Cooper at al. [17]. This study also showed that there is no significant difference in shoulder joint motion for people with different levels of spinal cord injury when adjusted for the speed difference. However, because of the lower speed of the SCI patients with cervical level of injury, their shoulder experiences longer duration of force for every meter traveled, which causes higher pressure on the shoulder of quadriplegic patients relative to paraplegic patients when undertaking the same task [16].

Wylie and Chakera [18] reviewed medical records of 51 patients that had had SCI for more than 20 years and found that higher levels of SCI are associated with higher risks of degenerative changes to hip and sacroiliac joint, as they have the least effective muscle activity that causes impaired circulation and production of synovial fluid which is necessary for normal function of articulation.

2.2 Some Complications in the Life of Paraplegic and Quadriplegic Patients

Among all negative the consequences of SCI, there is one that particularly influences SCI patients' level of activity and that is their ability to propel their wheelchair on rough terrains, to manoeuvre, and to play wheelchair sports. SCI reduces power output and cardiac function and peak oxygen uptake, due to some changes to the bodily function, particularly reduced muscle mass [15], [19]. This causes the upper-body power output of a paraplegic person to be about half of the age-matched able-bodied people, and for a quadriplegic person to be about half to one-third of a matched paraplegic individual's power output [19].

Sometimes electrical stimulation can recruit enough muscle mass to increase the capacity of maximum oxygen uptake [19]. It also can be used to produce purposeful activities such as upright walking, recumbent cycling, and rowing [15]. Upper motor injury produces spastic paralysis and lower motor injury produces flaccid paralysis [19]. Electrical stimulation can be used to produce movement in spastic paralyzed muscles for therapeutic or functional purposes, but the flaccid muscles do not respond to that. Therefore, paraplegic patients with the complete SCI level of T10 [15] or below cannot use the electrical stimulation.

SCI patients and especially quadriplegic persons may experience a number of adverse effects of exercise, including hypertension, hypotension, bradycardia, headache, nasal congestion, and autonomic dysreflexia. Therefore, it is recommended to supervise them and be ready to help during intense activities and exercise, especially for those with quadriplegia [14], [15], [19].

Quadriplegic individuals usually experience bradycardia with a maximum heart rate about 120 [19] to 130 [15] beat/min. Indeed, patients with a complete lesion at T4 or above show dramatically diminished cardiac acceleration which reduces their work capacity [15]. Quadriplegic patients also typically experience thermoregulatory dysfunction that makes it hard for them to adapt to cold, heat, and humidity changes. Hence, they should avoid experiencing sudden changes in these conditions and therefore, this makes it even harder for them to go for outdoor propulsion [19].

SCI patients have a high risk of osteoporosis that causes 25% to 46% of them experiencing fragility fractures, due to torsional stresses on the leg during transfers or due to falling from the wheelchair [20].

What is clear is that SCI not only limits the SCI patient's motor functions and sensory inputs, but also affects their overall health, overall strength, their susceptibility to some diseases, their life expectancy, their exercise and work capacity, their response to treatments, and more. This is in addition to the other aspects of their life affected, such as their job opportunities, psychological status, independency, and so on. This is researchers' responsibility to act to their best to help them diminish pains and enhance their quality of life.

2.3 Prevalence

2.3.1 Population of wheelchair users in the world

About 10% of the world's population live with some kind of disability and among these people, about 10% need to use wheelchairs for ambulation [1], [21]. In other words, wheelchair users comprise about 1% of the global population [1]. In 1999 there were 65 million wheelchair users around the world [1].

In the UK, the population of wheelchair users had a rapid increase in the 1980s and 1990s, with an approximate growth rate of 2 [21], which was identified to be a consequence of increasing social acceptance for wheelchair use, as well as changes in prescription practices and improved medical care [21].

2.3.2 Prevalence of SCI in Canada

Generally, 0.6% of the over-12-years-old population in Canada use wheelchairs for mobility and an additional 2.1% use other assistive devices for mobility [22]. One fifth of individuals in Canada who need assistive devices for mobility suffer from SCI [22].

Every year, about 3700 people with SCI discharge from hospitals and start a new life with spinal cord injury [23]. Based on a study conducted by Rick Hansen Institute, in 2010 there were 86,000 people with SCI in Canada, among those 49% had their SCI resulted from a traumatic SCI [23]. According to this report, among people with traumatic SCI, about 25,000 people had quadriplegia and about 19,000 individuals had paraplegia. Also in the non-traumatic group, about 13,000 persons had quadriplegia and about 29,000 people suffered from paraplegia [23].

Comparing different regions in Canada, Ontario has the highest proportion of the population using assistive devices for mobility whereas Alberta, Northwest Territories, and Quebec have the lowest proportion [22].

2.3.3 SCI in men and women

In 20.6% of the cases, SCI is the cause of the need for assistive devices for mobility. However, in the subgroup of men in the age group of 12-44 years old, the main reason for using assistive devices for mobility, including wheelchair is SCI (55%) [22]. The needs of women under the age of 45 using wheelchairs in Canada for help with Activities of Daily Living (ADL) is twice as men in this age range, as mostly, disease or illness is the cause of using wheelchairs for them, which causes general weakness and poor health [22].

Women in Canada at the age group of over 85 years old were more likely to use mobility support devices than men in this age group; this is likely because women have longer life expectancy [22].

2.4 Upper Extremity Pain and Injury

2.4.1 Prevalence of upper extremity pain in wheelchair users

The majority of wheelchair users have upper extremity pain [24], [25]. Sie *et al.* interviewed 239 wheelchair users for the presence of pain in upper extremity and concluded that 59% of all patients they recruited reported some type of upper extremity pain and 30% reported significant pain that required medication and occurred at least when undertaking two activities of daily living. In this study, the paraplegic group reported a higher prevalence of upper extremity pain (64% of paraplegic group versus 55% of the quadriplegic group), but the percentage of the quadriplegic group who reported having significant pain was greater (41% of paraplegic group versus 59% of the quadriplegic group) [24].

Curtis and colleagues recruited 55 women and 140 men (92 quadriplegic and 103 paraplegic patients) and using a self-recorded survey they found that more than two-thirds of participants had experienced shoulder injury at some point after injury, and 42% of paraplegic patients and 59% of quadriplegic patients had shoulder pain at the time of study [26].

2.4.2 Pain/injury in different regions of upper extremity

To find out the prevalence of pain in different regions of the upper extremity, Sie and colleagues conducted one of the biggest studies on real wheelchair users by interviewing 239 SCI patients (136 quadriplegic and 103 paraplegic patients) [24]. According to their results, 32% of paraplegic participants reported pain in more than one area of the upper extremity. Carpal Tunnel Syndrome (CTS) with 66% was the main complaint of the paraplegic group in this study, followed by shoulder pain (36%), elbow (16%), wrists (13%), and hand (11%) [24]. In the quadriplegic group, 40% reported having pain in more than one place. In this group of participants, shoulder pain was the most reported complaint (46%), and pain in elbow, wrist, and hand was equally reported as 15% of the complaints [24].

In the study conducted by Pentland and Twomey [27], the most complaint of upper extremity pain of subjects recruited, paraplegic women, was shoulder and then wrist and hand. According to Finley and Rogers results [25], 61.5% of participants reported experiencing shoulder pain and 29% reported to have pain at the time of the study. In general, the prevalence of shoulder pain, depending on the group of participants, has ranged between 31 to 73% [17] and the prevalence of wrist, hand, and elbow pain has been reported as 13, 11, and 16%, respectively [17].

CTS prevalence among wheelchair users has been reported to be between 49 to 73% [17]. 65% of wheelchair users with upper extremity pain have tendinitis or rotator cuff tear and 22% have aseptic necrosis [17].

In a study undertaken by Robertson *et al.* [28] participants showed higher joint moments in their shoulder than their elbow or wrist. Also, a large vertical reaction force was seen in the shoulder.

2.4.3 Relation between length of time after injury and prevalence of upper extremity pain

Silfverskiold observed that the percentage of quadriplegic patients who reported shoulder pain during 6 months after injury was 78% and this percentage dropped to 33% in the period of 6 to 18 months post-injury. The prevalence of shoulder pain in paraplegic patients, however, was not different between duration of less than 6 months post-injury and the period of 6-18 months post-injury (35%) [29].

Sie and colleagues associated the prevalence of upper extremity pain to the time passed after injury. According to their results, the prevalence of shoulder pain was highest during five years after injury for SCI patients with quadriplegia [24]. Curtis *et al.* also reported higher scores for Wheelchair User's Shoulder Pain Index (WUSPI) for quadriplegic patients in five years after injury. They believed the lower scores of WUSPI in higher age-groups was due to ceasing the most painful ADL by quadriplegic patients [26].

Cooper *et al.* [17] stated that 20% of wheelchair users experience upper extremity pain five years after injury; after 15-19 years, this percentage rises to 46%. Prevalence of shoulder pain increases until 20 years after injury, then it decreases slightly [17]. Sie *et al.* had similar observation about 20 years post-injury (they observed greater reduction of upper extremity pain (from 46% to 27%)) but concluded that this is unexpected and it could be due to the decreased number of patients 20 years after injury, or due to the reduction of activity level in this group by getting aged [24].

2.4.4 Quadriplegia versus paraplegia

According to Curtis *et al.*, 53% of Americans who suffer from SCI have quadriplegia and the remaining 47% have paraplegia [26]. Sie *et al.* reported that 55 percent of people with quadriplegia and 64 percent of people with paraplegia suffer from upper extremity pain. The most prevalent disorder for the first group was shoulder injuries, while it was the second most prevalent injury for the second group, after CTS [24]. In their study, although the paraplegic group reported a higher prevalence of upper extremity pain, the percentage of the quadriplegic group who reported having significant pain was greater (41% of paraplegic group versus 59% of the quadriplegic group) [24]. This is consistent in part and not consistent in the other part with the study conducted by Curtis *et al.* [26]. According to their results, patients with quadriplegia had both greater prevalence of shoulder pain and more significant pain [26].

Individuals with quadriplegia have fewer complaints about shoulder pain, possibly because they avoid performing painful activities [24], [26]. Curtis and colleagues thought the reason that shoulder pain interfered more with functionality in quadriplegic patients as they generally have weaker strength in elbow flexion and in hand grip [26] while paraplegic group may take the advantage of stronger elbow and hand grip to compensate for the painful shoulder, so that they reduce the load on the shoulder. The quadriplegic group, however, mostly cease those activities of daily living that are more painful or strenuous [26].

2.4.5 Causes of upper extremity pain/injury

Wheelchair propulsion has been shown to be a causative factor for developing shoulder pain/injury [30]. Pentland and Twomey [27] recruited 11 paraplegic women and 11 able-bodied women in a control group that were activity-level matched. They concluded that development of upper extremity pain was clearly and significantly associated with paraplegia. In a study by Mercer *et al.* [31] joint forces and moments were related to shoulder pathology and the authors suggested alternative methods of ambulation, propulsion training, altering wheelchair setup, and reducing the force exerted to the pushrim in order to preserve upper extremity function.

Researchers have introduced a number of causes for the upper extremity pain and injuries in wheelchair users and mainly have attributed those pain and injuries to overuse [26], [32], [33]. Due to the repetitive [30], [32] pushes and high loads on the upper extremity [17], [34] when wheelchair users propel and transfer [17], [35], they are subject to upper extremity pain and injury. Shoulder positioning also has been introduced as a possible factor [30], [36].

Bicipital tendonitis, rotator cuff impingement, and glenohumeral instability are the most common pathologies found in wheelchair users [25]. According to Cooper *et* *al.* [17], there is a critical zone for rotator cuff injury on the top of humeral head, where the *supraspinatus* attaches. This area has small veins and arteries and when transfers are performed, very strong contractions of the muscles around the rotator cuff take place and the pressure in the arteries gets more than twice higher than their typical pressure. In addition, any activity that pushes, or tends to push the humeral head further in the glenohumeral joint is a risk factor for impingement. (The injury caused when the acromion pinches the tendon of supraspinatus, due to activities that need to raise the shoulder and arm, such as swimming, or due to some other trauma around the shoulder.) Some examples of activities that increase the risk of impingement are transfers and when the person pushes downward on the wheel to create more friction while grasping the pushrim. Another important cause of shoulder pain is the muscle imbalance that is caused by overuse, especially the imbalance between the internal and external shoulder rotators [17].

Manual wheelchair users are at high risk of developing CTS [37]. Boninger and colleagues found a significant correlation between the weight of wheelchair user and median nerve function. They believe weight loss could help prevent CTS in wheelchair users [37]. When the median nerve gets compressed, over time carpal tunnel syndrome may develop. Compressing the median nerve could happen due to extreme flexion and extension of the wrist, or when the sheath of the flexor tendons is thickened as a result of repetitive motion, or by being exposed to vibrations [17].

Boninger *et al.* in another study [38] recruited 35 manual wheelchair users and examined their median and ulnar nerve function and biomechanical variables when propelling on a dynamometer with 0.9 and 1.8 m/s speeds. They concluded that wheelchair users who had a greater range of motion of wrist had better median nerve function and, therefore, had a lower risk of developing CTS. They stated that these individuals used long smooth strokes and so they exerted less force, in magnitude and frequency, on the pushrim. Boninger *et al.* believe this method of pushing, hence, is a good strategy for preventing CTS in wheelchair users.

Brose *et al.* [39] recruited 49 athlete wheelchair users who were all men with SCI and used an ultrasound-based rating for shoulder pain along with two shoulder pain questionnaires and found a fair positive correlation between shoulder pain, measured with any of the three measurement tools mentioned, and age, weight, and years of using wheelchairs. Wheelchair type can also contribute to shoulder pain, as Rose [33] found that people that have used manual wheelchairs with folding frames have reported more pain.

2.4.6 Influence of upper extremity pain and injuries on wheelchair users' life

Since wheelchair users rely on their upper extremity for mobility and functionality, there is more intense and more frequent load on their shoulder and, therefore, they are more susceptible to upper extremity pain/injury. On the other hand, the occurrence of upper extremity pain/injury has a more severe effect on the functionality of wheelchair users than for able-bodied individuals and it could practically be equivalent to a much higher level of injury for wheelchair users [24]. Activities that require transfers, reaching overhead, and propulsion, bathing and dressing are mostly associated with the interfering upper extremity pain [25], [26]. 67% of male wheelchair users and 74% of female wheelchair users need help with their basic activities of daily living [22].

Silfverskiold and Waters recruited 60 SCI patients and reported that significant shoulder pain in 84% of patients with quadriplegia leads to functional disability, while in the paraplegic patients the shoulder pain did not lead to problematic functional disability [29]. These findings were supported later by Curtis *et al.* [26]. Individuals with quadriplegia are more likely to be unable to perform their more strenuous activities of daily living by getting aged or by time passing after the onset of their injury [26].

46% of paraplegic individuals and 60% of quadriplegic individuals experience shoulder pain during sleeping [26]. This means that in Alberta alone, about 7,000 [23] people with SCI are experiencing shoulder pain day and night.

2.4.7 Other life complications

According to a report published by Rick Hanson Institute [40], the cost of lifetime care for every paraplegic and quadriplegic individual is 1.6 and 3 million dollars, respectively. In Canada, traumatic SCI costs 3.6 billion dollars annually. Adding this to the aforementioned problems and difficulties that millions of wheelchair users experience in their everyday life shows the significance of the problem; difficulties that force them to be relatively dependent on other people's help in their ADL clearly shows why it is important to advocate sources and research studies in an attempt to slightly ease the life of the wheelchair users.

2.5 Wheelchair Propulsion Biomechanics

2.5.1 Muscle activity

During wheelchair propulsion, the muscles of the trunk, shoulder, arm, and hand cooperate to move the hand during push and recovery phases. The primary muscles that are active during push are shoulder flexors and the primary muscles during recovery phase are shoulder extensors [41]. The transition between recovery and push demands significant force from the shoulder muscles [41]. Here is a summary of muscles function during push and recovery phases:

Push Phase: *Biceps Brachii* is active in early push [42], [43] as an arm and forearm flexor and supinator [43], and *Brachioradialis* as pronator [43]. *Anterior Deltoid* and *Pectoralis Major* are active throughout push [42] as arm flexor [43], *Triceps Brachii* is active mid to late push [42] as forearm extensor [43], and *Flexor Carpi Ulnaris* and also *Extensor Carpi Radialis* are active throughout the push [42]. At late push phase, *Brachioradialis* as arm and forearm flexor, and *Biceps Brachii* as an arm flexor are active.

Recovery Phase: *Posterior Deltoid* is active throughout recovery [42] as arm extensor [43], *Biceps Brachii* as supinator and a forearm flexor working against gravity [43], and *Triceps Brachii* at early recovery [42]. In Bjerkefors' study, a person with SCI (level of injury: T3 complete, as classified clinically) activated all trunk muscles when propelling a wheelchair [44], evidently proving his level of injury was not complete.

Changes in muscle activity associated with fatigue: When fatigued, *Flexor Carpi Ulnaris* is active earlier in push phase and *Biceps Brachii* fires at late recovery phase [42]. *Pectoralis Major* and *Infraspinatus* are more sensitive to fatigue as they have two responsibilities: stability of shoulder joint and handrim power generation [41].

Qi *et al.* (2012) pointed out that there are complex neurological mechanisms in the body that control the recruitment of muscle fibers to manage fatigue. They investigated surface EMG signals of 7 shoulder muscles during fast and slow wheelchair propulsion. The participants of this study propelled the wheelchair on a wheelchair ergometer with the velocities of 0.9 m/s and 1.6 m/s until they became mildly fatigued. According to Qi *et al.*, at the beginning of an activation, the body recruits fast motor units that generate higher frequency EMG signals and can produce higher forces. However, these fast motor units are not very fatigue-resistant. Therefore, when time passes and fast muscle fibers start to fatigue, slower motor units, which have higher endurance, are the recruited. This is the strategy that the body takes to resist fatigue [45].

2.5.2 Important factors in wheelchair performance

Propulsion technique*:* Guthrie identified the angles of the shoulder and elbow that produce peak hub torques as 15° extension to 15° flexion of the shoulder, and 100° to 80° elbow flexion [42]. Yao recognized the arm function as a more important factor than trunk stability and strength in wheelchair propulsion [46].

Boninger at al. conducted a major experiment [47] recruiting 38 wheelchair users and reported four patterns of hand movement during wheelchair propulsion. These four patterns are namely (sorted as the most to the least common propulsion pattern): SLOP (single looping over propulsion), DLOP (Double looping over propulsion), CS (semi-circular), and arc (45%, 25%, 16%, and 14% of the participants, respectively) [47]. Boninger *et al.* stated that the previous results published by Veeger at al. [48], had the statistical type II error due to the small sample size (5) and unlike what Veeger *et al.* suggest, the semicircular pattern of propulsion does not provide higher mechanical efficiency. According to Boninger *et al.* kinematic parameters, including mechanical efficiency, are not influenced by propulsion pattern [47].

Rolling resistance (RR): This is defined as undesired and unhelpful friction when there is a wheel rolling. RR is related to wheel diameter and dimension [49], [50], tire material [49], tire threading [51], caster characteristics [49], rolling surface [42], distribution of load on wheel [49], caster angle and rear wheel camber [42]. Brubaker in his study suggested a methodology for measuring rolling resistance [52].

The kinematics of the shoulder, arm, and wrist does not change significantly with the level of RR (friction) [34], but the kinematics of the trunk does. Trunk flexion and momentum increase significantly when RR increases [34].

Seat position*:* Researchers, especially Brubaker, have conducted several research studies to study the effect of seat position (or axle position) on the biomechanics of wheelchair propulsion [53]–[59]. Based on their efforts we know that seat and rim position have a considerable effect on handrim force. By adjusting the seat at back or a middle position, wheelchair users can exert a higher force to the pushrim, while it also provides better grip at the time of pulling (early push phase) [58]. Joint motion range also varies depending on the changes in seat position [54]. Brubaker found out [60] that propulsion efficiency increases when the seat position moves backward, but on the other hand, this decreases stability which is not considered a good thing. Even though, Brubaker reasons that adjusting the seat position backward could be advantageous, especially for people who have higher levels of injury; the reason he states is that although these people need more stability, the importance of a high efficiency is even greater for them, as they are less-abled. For preventing unwanted tipping, he suggests, wheelchair users could make use of an anti-tipping bar. Another reason that Brubaker states for the advantage of a more forward rear axle position (RAP) is that this way the RR against propulsion would be less [60]. He explains that RR is related to the weight tolerated by the wheels and the size of the wheels, among other factors. A forward RAP will cause a greater portion of the weight to be tolerated by the back wheels of the wheelchair. On the other hand, the RR is inversely related by the size of the wheels, so the RR created in effect of that portion of weight that is shifted to the back wheels will be less. In other words, this mechanism will work out to reduce the overall RR for more forward RAPs. This looks promising, but a study conducted by Hills [59] revealed that the effect of RAP in altering the RR does not significantly change the required propulsive forces.

Richter proposes the idea of a combination of changes to seat position and minimizing the distance between shoulder to the hub, such that the initial shoulder angle (shoulder angle at the beginning of the push) remains between -50 and -75 degrees [56] where 0 degrees is associated to the anatomically natural position of the arm [36].

Lever drive wheelchairs: Brubaker and colleagues have conducted a considerable amount of research studies on the lever drive wheelchairs in the 1980s and early 1990s [53], [54], [57], [61]–[65], as they believed the efficiency of conventional handrim wheelchairs is low [64]. This relatively higher mechanical efficiency of lever wheelchairs has been confirmed again in a study conducted by Lui *et al.* [61]. In addition, Brubaker *et al.* showed in a study, that propulsion efficiency for handrim wheelchairs is significantly dependent on seat position, while there is not such a relationship for lever drive wheelchairs [57]. Brubaker *et al.* stated in a paper [57] that the optimum seat position for each person is dependent on their individual characteristics and analogized it to finding the best shoe size for each person [57]. They tried to substitute the lever drive wheelchairs for handrim wheelchairs, but despite their efforts, the majority of manual wheelchairs are still the conventional handrim wheelchairs. This might be due to some characteristics that Brubaker names for handrim wheelchairs [57]: they are effective interfaces that provide maximum control and feedback over a wheelchair. Also, they are simple and probably the most reliable form of the wheelchair. Nevertheless, Brubaker particularly suggests lever wheelchairs for both extreme ends of wheelchair users [57] or/and people that have special requirements about seating position [54].

Role of trunk swing: Trunk angle has a role in many aspects of wheelchair propulsion. For example, it is inter-related with push length, which is a major factor in pushing strategy [38], or in the start and end point of each cycle of pushing. An incorrect start and end point may result in a shoulder injury. It also prevents tipping and helps with stability in wheelies.

As part of normal usage, trunk swing is used by many wheelchair users. Trunk swing is important, because it enables longer stroke length, adds energy by changing the center of mass, and prevents tipping. Also, if the wheelchair user gets fatigued, in addition to these advantages, trunk flexion helps the wheelchair user overcome fatigue by recruiting some other muscles. However, a fatigued wheelchair user may be reluctant to bend his/her body, because it impairs breathing. In a research study that we have conducted recently, we studied how real wheelchair users behave in terms of trunk flexion when they get fatigued in activities of daily living [66]. According to the results of this study we concluded that there was no general trunk swing behavior when wheelchair users were not fatigued; however, they showed greater trunk flexion, longer strokes, and slower cadence when fatigued.

In a study conducted by Hwang *et al.* [34], participants showed greater trunk flexion when the resistance against propulsion was increased. Hwang stated in his paper, that this could be due to the help trunk flexion does in increasing the efficiency, as it causes higher downward momentum.

Trunk movement could help substantially in wheelchair propulsion, provided that the trunk moves uniformly with the arms [67]. In other words, the trunk should move in the same direction with the arms to help the wheelchair user to save energy or help in the best way to overcome fatigue. We have presented the detailed explanation in biomechanical terms, on how this uniform trunk swing behavior helps in wheelchair propulsion [11]. We made the analogy that the advantage of uniform trunk swing during wheelchair propulsion is like the advantage of trunk motion when swinging on a swing set. As we know, trunk movement during swinging, if done with good timing, creates a "resonance phenomenon". Thus, the person can increase the amplitude of swinging by a little force.

Not all the researchers agree though, that trunk swing is beneficial for wheelchair users. Rodgers *et al.* [32] studied trunk flexion and the influence of it on biomechanics of wheelchair propulsion. They recruited nineteen non-athlete wheelchair users and divided them to "flexed style (FS)" and "non-flexed style (NFS)" groups, based on their performance in the experiment. One truly exciting result that they only mention in passing was that they observed the NFS group experienced about a 50% higher shoulder extension moment. This shows that the group who did not use trunk swing during propulsion experienced much higher stresses on their shoulder. This could be a sign that trunk motion has helped the FS group to reduce the pressure on their shoulder, which is very positive. They also observed that the two groups were not different in their oxygen uptake during the tests, which shows that they had had similar energy consumption during the test and therefore the lower pressure on the shoulder in FS group was not because of less work they were doing. However, they concluded that trunk flexion is debilitating for an upper arm injury, since, they believed, the NFS group were "compensating for peripheral muscle fatigue" [32].

Rogers *et al.* conclusions were led by previous research [32], [68], [69] which considered trunk movement to be detrimental. However, there is an important factor needing to be considered, that they have focused on "paradoxical trunk movement" which they report is greater at patients with higher level of injury [32]. For these wheelchair users, the trunk moves in the opposite direction to the arms. This is a strategy wheelchair users with poor trunk control (mainly due to the injury to the spinal cord) employ to help stabilize the trunk in the absence of strong abdominal muscles [68], [69]. In this case, the Meff (proportion of tangential force to the total force) is less than when trunk swing is not used [69].

2.6 Role of Research Studies

As shown in the aforementioned sections, upper extremity pain and injuries [25], [26] and the aim to reduce pain, boost independence, and enhance the quality of life for wheelchair users has been a motivation for conducting many research studies [23]. Some authors [28], [30], [34], [36] have studied the kinematics and kinetics of shoulder, elbow, and wrist in order to find the causes of upper extremity pain in wheelchair users or in the aim of pain reduction [36]. The impact of wheelchair type [33], mechanical energy and power flow in the wheelchair and wheelchair user [70], different types of wheelchair designs in order to introduce improvements, determining the effect of axle position and wheelchair setup and pushing strategies have all been studied with the intention of achieving better propulsion efficiency [57], [71]–[82].

Although a considerable amount of research has been undertaken to help minimize the pain and difficulties that wheelchair users face in every-day life, the problem still remains unresolved. The causes of developing upper extremity pain/injuries in wheelchair users have not been completely identified yet [30]. Research studies have not yet been able to present complete and comprehensive solutions. Heterogeneity and small sample size and the inconsistency between technologies and methodologies [83] have been limiting factors in the studies undertaken involving wheelchair users. It has been shown that two of the important papers [4], [48] published about wheelchairs that have been cited several times in other studies have had type I and type II [47] statistical errors, due to the small sample sizes they have had. Most of the experiments that have recruited experienced wheelchair users have been completed with low sample sizes; those who have had large sample sizes from real wheelchair users have been mostly the ones using surveys or questionnaires [24], [26].

Sometimes the results of studies are true but the inference and the conclusions drawn have not been done properly. For example, the data presented in one study [32] were supportive for a positive effect of trunk swing during wheelchair propulsion, but they concluded that trunk swing is disadvantageous for wheelchair propulsion.

In some cases, the evidence available is contradictory. Some of these controversial findings are outlined and discussed here.

We already know that wheelchairs with longer wheelbase [84], higher camber angles [72], and the seat positioned forward [85] have higher stability. Also we know that a shorter length of wheelchair helps with manoeuvrability [85], higher camber angles increase propulsion efficiency [1], [84], lower camber angles are more convenient to go through narrow pathways [1], backward seat position increases propulsion efficiency [85], longer push length provides higher efficiency [86] and lower chance of developing CTS [38], and placing hands too back on the wheel is harmful for the shoulder [36]. Albeit Trekinetic [84] is a cutting-edge new design wheelchair that makes the camber angle adjustable while the wheelchair user is going through different pathways, but there is still a lot that needs to be researched. For instance, while many studies have attributed upper extremity pathology to wheelchair overuse [26], [32], [33], Finley and Rodgers [25] reported neither positive nor negative correlation between involvement in sports and experiencing shoulder pain. On the other hand, it has also been shown that being inactive contributes to degeneration of hip and shoulder joints [18]. Studies need to be designed to determine how much activity the wheelchair users need to prevent upper extremity pain or whether special strengthening exercises would be helpful.

Manoeuvrability is considered by many wheelchair users to be a very important functional skill and key performance requirement for their wheelchair. However, it has received little attention in the literature yet. Manoeuvring exerts a large amount of stress on the upper extremities and so could contribute to overuse injuries, but no studies has yet evaluated the level of stress on the shoulder during manoeuvring and strategies that should be taken to avoid overuse injuries. The same could be said about propelling on inclined surfaces and the positive role that trunk swing could have to help in propulsion.

Another area that deserves more attention in wheelchair studies is introducing protocols with simple measures that could be used by the wheelchair user for self-monitoring purposes. It has been reported that many wheelchair users that have started to develop degenerative changes were unaware of it because they are asymptomatic [87]. Certainly, if they learn about the potential for long term conditions earlier they could better avoid them.

New research studies involving the use of wheelchairs should be designed to address the above concerns. This could then be translated to the benefit of wheelchair users, clinicians, and wheelchair manufacturers to introduce the best practices and designs that could lead to a less-painful and more-independent life for wheelchair users.

2.7 Research Tools to Study Wheelchair Biomechanics

Although it is helpful to study the life of wheelchair users where it happens, research studies can benefit considerably if such studies could be performed, validly, in laboratory environments. This is because laboratories provide complex research equipment and controlled conditions that are mostly not achievable as well, outside laboratories. However, to undertake those research studies, the every-day life of the wheelchair users need to be *simulated* inside laboratories, where conditions and measurements are highly under control, using treadmills, wheelchair ergometers, and Virtual Reality techniques.

2.7.1 Treadmills

Treadmills have been used to facilitate wheelchair studies. Usually, motorized single-plate treadmills [88]–[91] are used, but some studies have used double-plates [92], [93], to add control over the simulation [92], or simply because one single-plate treadmill was not wide enough to fit a wheelchair [93] in. Treadmills are favored because they are readily available, help perform wheelchair studies in laboratory environments, provide consistent test conditions [89], can simulate slopes [88], and velocity and power output could be controlled with them [89]. However, they run on a set velocity and do not allow the participants to propel their wheelchair in their comfortable pace [94]. Also, during wheelchair propulsion in the real world, the wheelchair slows down between pushes, but this does not happen on a treadmill. Moreover, different RR and turns are not easy to simulate with treadmills [94]. Sometimes slopes are used to regulate the resistance against propulsion [83], but RR and slopes have different biomechanical implications on wheelchair propulsion. Wheelchairs do not fit [89] on the treadmills sometimes, and above all, the inertia cannot be simulated on treadmills. Examples of studies that have been undertaken using treadmills are briefly mentioned here.

Kwarciak *et al.* [89] compared handrim propulsion biomechanics on the floor (low pile carpet in the lab) and on a special wheelchair accessible treadmill (wide and with a dynamic safety system) (MAX Mobility, LLC). It was a cross-sectional study on 28 experienced wheelchair users. They recorded contact angle, peak force, peak axle moment, power output and cadence in over-ground propulsion first. Then, for the second round of experiments, they set the treadmill speed to be the same as the first round and used a metronome to guide the subjects to maintain the same cadence as their first round. They gradually increased the power output until it reached to the same power output as their first experiment. Then, they recorded the same variables as on the over-ground tests and eventually, checked the correlation between the two rounds' results. The correlation coefficients were all greater than 0.85, and they concluded that this treadmill is a valid substitute for floor wheelchair propulsion. However, they did not consider energy expenditure of subjects, while this is one of the possible sources of differences between the two systems due to the lack of kinetic energy in treadmill wheelchair propulsion. They also did not consider trunk movement.

Gil-Agudo *et al.* [90] considered kinetics of shoulder joint during straight line wheelchair propulsion on a treadmill. They used a SMART^{Wheel} and inverse dynamics technique to find shoulder joint reactions on 16 experienced wheelchair users and concluded that the reaction force in shoulder increases when speed increases.

Brubaker *et al.* [88] investigated the effect of the side slope on wheelchair propulsion on a treadmill using one subject propelling on 0 degrees and 2 degrees slopes. Their results showed that the increase in force was greater than the increase in the net energy when the side slope is compared to the level surface.

Stephens *et al.* [93] compared the overground and treadmill wheelchair propulsion patterns for quadriplegic manual wheelchair users and found a meaningful difference between the two states as well as between the right and left sides. They concluded that caution should be taken when generalizing treadmill tests results to the normal life of wheelchair users and also when assuming lateral symmetry. Mason *et al.* [95] also compared the over-ground wheelchair propulsion with treadmill (0,0.7,1, and 1.3% slopes) and ergometer wheelchair propulsion in 4,6, and 8 km/hr speeds. They concluded that 0.7% slope provides the closest physiological responses to the real-world for slower speeds and 1% slope is the closest to real-world in physiological responses for higher speeds. According to their results, ergometers overestimated the physiological responses and none of the modalities provided a valid representation of forces applied in real world.

Dysterheft at al. [96] investigated the effect of level of physical activity and shoulder pain on propulsion techniques recruiting 14 experienced wheelchair users who propelled their wheelchair on a treadmill with the speed of 2m/s, which is double the normal propulsion speed [97]. They found a significant correlation between high physical activity and high-frequent and short strokes, which are known to contribute to secondary injuries among wheelchair users. They also found a meaningful association between changing the propulsion techniques and higher physical activities, which could be a protective mechanism.

2.7.2 Wheelchair ergometers

Several ergometers have been designed and used in different wheelchair studies, that are different in:

- The measurements they can take
- The biomechanical factors considered in their design
- Whether they are motorized or not
- Are they capable or not of simulating:
- Straight-line propulsion
- Wheelchair manoeuvring
- Up and down hills
- Side slopes, and
- Different rolling resistances.

Ergometers usually incorporate two rollers where wheelchair back wheels stay on top of them and a fixture is used to secure the wheelchair. Many research studies (such as [95], [98]–[102]) have been completed using ergometers, as they allow independent rotation of the wheels, provide some inertial (=mass) against movement, can provide a better fixture for the wheelchair, and more importantly, the wheelchair users are usually in control of their movement at every moment, rather than adjusting to a preset speed, as it happens with treadmills. They also may include a braking system to adjust the RR against propulsion and thus enable the simulation of different power outputs [97]. Another advantage of the rollers is the ability to reproduce testing conditions [83].

Aside from all these advantages, conventional ergometers are only capable of reproducing straight-line wheelchair propulsion, as visual feedback is essential for simulating curvilinear propulsions. We found only one case [103] in the literature that has used a primitive visual feedback, a dial that showed the direction of movement, and hence used the ergometer for curvilinear propulsions too. Virtual Reality can be employed to fill this gap and make use of wheelchair ergometers to simulate wheelchair manoeuvres as well. Ergometers that can simulate rotations are discussed under Virtual Reality, Section 2.7.3., and some examples of the ergometers designed for straight-line wheelchair propulsion are outlined here, before expanding more on the biomechanical factors considered in the design of wheelchair ergometers.

Gonzalez-Quijano *et al.* [51] devised a controllable wheelchair ergometer that can work in four modes of simulation, e. g. constant power. They did not include any verification for their device to show its validity. Dabonneville *et al.* [104] designed an ergometer which they claim can measure biomechanical parameters (force and momentum) and represent real life conditions. They used one able-bodied subject for the verification test. Finley *et al.* [105] used a sample of 10 able-bodied subjects propelling their wheelchair in a straight line and on an ergometer. They did three submaximal fatigue tests and concluded that majority of biomechanical variables obtained from the experiments were test-retest reliable. Devillard *et al.* [106] used a sample of 17 non-wheelchair user subjects for validation of an ergometer (VP100H) through comparing the obtained variables with those obtained from a force transducer which was placed underneath the ergometer. The task was straight-line propulsion with and without trunk movement. They concluded that VP100H is a valid measurement instrument. Wu, *et al.* [102] considered straight-line wheelchair propulsion on two types of carpets with different resistances and then tried to simulate that on an ergometer by controlling the resistance of ergometer's rollers against being rotated. They recruited 25 unimpaired subjects for this study, and concluded that using the formula introduced in their report, ergometer wheelchair propulsion can replicate floor wheelchair propulsion. However, they used a correction factor in their formula which was obtained from the comparison between ergometry tests and floor tests. They indicated that the need to use a correction factor was due to a software lag rather than an arbitrary correction factor. However, using a correction factor that is obtained from one data set is not guaranteed that can work for other data sets as well. Finally, Van der Woude *et al.* [107] have developed a motorized ergometer that uses a mathematical model to simulate inertia, rolling resistance, and up and down hills, independent of speed, using one single roller. It, however, cannot simulate turning.

Biomechanical factors considered usually lack important factors such as energy and inertia. Motorized ergometers [94], [101], [107]–[109] do consider inertia and energy based on a mathematical model. However, if the ergometer is designed with a direct approach, when validated, there will be less chance of errors and if errors happen, there will be less chance of sudden and large errors. This is because an ergometer with a direct approach is designed in such way that participants feel the immediate reaction of their actions without using a PC analysis in between and a motorized device as a medium. Also, these ergometers can be more reliable and there will be no time lags, so it would also work for people who are very sensitive in detecting even the tiny mismatches.

Nonetheless, there are some ergometers with a direct approach that have considered inertia and have tried to design ergometers in such way to replicate it. In one study [110], removable flywheels were used to simulate translational inertia on the ergometer. Also, there is one recent study [99] that has developed a light portable ergometer capable of simulating linear inertia to some extent: it provides an average inertia for three intervals of subject weights. However, it includes several simplifications in the design.

2.7.3 Virtual Reality (VR)

VR is a cutting-edge technology that is currently used widely, with different applications across many different research areas. Rehabilitation of the wheelchair users is not an exception.

Researchers have conducted wheelchair related studies for several years; inventors have reported miscellaneous designs for wheelchairs (first patent: 1937 [49]) and also several wheelchair propulsion simulators such as different types of ergometers and treadmills have been introduced. The introduction of virtual reality to support wheelchair activity simulation is a new and very promising development. One of the limitations of wheelchair laboratory based biomechanics research to date has been that it has been limited to simple propulsion tasks, such as straight line pushing. VR offers an opportunity to conduct laboratory-based studies for much more complex tasks.

There have been several efforts in recent years to take advantage of VR capabilities to facilitate and accelerate rehabilitation [111] or training [112]–[116] of wheelchair users, in addition to broadening the knowledge available on wheelchair propulsion [111]. These efforts usually include simple VR systems, mainly using one or more monitors [101], [111], [112], [114]–[119] along with a joystick [111]–[116], [118]–[120] as the interface. This way, simulation of powered wheelchair propulsion is possible, but not manual wheelchair propulsion. There have been several positive outcomes reported in the literature for using VR in training/rehabilitation of power wheelchair users [111], [113]–[118], [120], but we could find only four cases of a VR environment developed for manual wheelchair users:

In one study [121], a system was developed to use the rotation of the wheelchair wheels to replace a joystick and estimate the location of the wheelchair in the VR using a logarithmic scale for calibration. This study has tried to include inertial navigation by actual movement of the wheelchair. However, there is no one-to-one replication of movement and no immersive VR. In another study [94], a motorized ergometer was designed for replicating curvilinear wheelchair propulsion that uses haptic admittance control to adjust wheelchair wheels velocity in accordance to the moment applied on them. Inertia effects are also applied by the same control system.

AccesSim [101] is a VR-based wheelchair ergometer designed for testing accessibility of urban areas. It provides force feedback by accelerating the wheels or changing the RR based on what is simulated in the VR, e.g. slopes. This ergometer lacks inertial compensation though, and the VR is not immersive.

Toronto Rehabilitation Institute's Challenging Environment Assessment Laboratory (CEAL) is developing a wheelchair simulator [109] in an immersive VR environment to be used for manual wheelchair users. In this wheelchair simulator, the forces applied on the wheels are measured by the sensors embedded in them and sent to the PC real-time where a MATLAB program runs a mathematical model to find the velocity and acceleration of the wheels as well as the heading. The wheelchair is placed on a turning table which rotates based on the heading calculated. This wheelchair simulator is still under development.

2.8 Added Value Using VR

Some benefits and strength points of VR are highlighted here.

2.8.1 Controlled environment

Conducting research experiments requires a controlled environment and different measurement instrumentations. The experiment site should be controlled to eliminate biases and other unwanted factors that could influence the results of the study. Furthermore, quantitative wheelchair research needs measurement devices. Conducting experiments in laboratories where conditions and devices are easily accessible and highly controlled is an advantage, as long as the accuracy of data and generalizability of results to real world situations is not affected. A reliable and validated VR environment could represent the real world and therefore provides the highly controlled environment without losing accuracy and generalizability.

2.8.2 Safe environment

Motor complete SCI patients face a reduction of bone density in their hip, femur, and tibia to 28%-50% during 18 months after injury [20]. This means that if they fall, they may easily break a bone. VR creates a safe environment that could be used both for training new wheelchair users on techniques they need to know to fulfill their activities of daily living and for performing wheelchair experiments, especially those involve techniques like tipping and wheelies. VR has been used for newly injured SCI patients and children to train them to use power wheelchairs [2], [3], [122], but the combination of VR and manual wheelchair propulsion is a new concept that could have a significant impact on the future of manual wheelchair propulsion training and research.

2.8.3 The effect of immersion

Boninger *et al.* [123] suggested in their study that by 2022, VR should be more immersive, fun, less expensive, and easier in responding to biological stimuli such as brain signals. Immersive VR is a VR environment in which user's point of view or at least part of his/her body is within there [124]. Immersion provides many opportunities that cannot be easily obtained using desktop VRs. It also helps with having a greater sense of presence [9]. Albeit this is not a sufficient condition and some other factors such as the frame rate and the lag between input (from the user) to output (the corresponding changes in the visual feedback) is also important and influential [124].

2.8.4 A thermally natural environment

SCI patients and particularly those with quadriplegia usually have impaired thermoregulation and experience difficulties adapting to changes in the temperature and humidity of the environment [15], [19]. This could trigger other problems such as autonomic dysreflexia [19], [125] and fatal fevers [125]. Thus, a thermally controlled environment should be provided [19] for the SCI persons who are participating in experiments, especially for those with high level thoracic or cervical levels of injury. A combination of immersive VR and instrumented ergometer that is reliable and valid in representing real life wheelchair propulsion could be used for experiments recruiting SCI patients in a thermos-natural environment.

2.8.5 VR to motivate recruitment

Wheelchair research studies have always faced the problem of small sample sizes. Although there have been some research studies that have had relatively bigger sample sizes (239 [24], 195 [26], 70 [12], 51 [18], 38 [47], and 28 [89] wheelchair users), they were mainly retrospective experiments and only 3 of these studies could recruit the wheelchair users to participate in a research study that demanded something beyond filling surveys [12], [47], [89]. This is not very surprising, as wheelchair users only comprise about 1% [1] of the population and about 20% [22] of them have spinal cord injury. Difficulties they face in propelling their wheelchair outdoors adds to the participant recruitment difficulties. On the other hand, lack of experience for non-wheelchair-users could be a source of bias [105].

Small sample sizes diminish the power of studies and increase the chance of type I or II statistical errors, some of which occasionally has been detected later [47]. There is a hope that VR could help with increasing the sample sizes to a great deal by providing greater motivation for potential participants.

2.8.6 Winter in summer, summer in winter

Visual feedback along with simulating different surface resistances on the ergometer enables VR to simulate different weather conditions, regardless of the time of year. Other stimuli such as artificial snow and winds could also be added to the VR environment to provide a greater sense of presence.

2.9 VR Helping Wheelchair Studies and Wheelchair Users in General

VR can be used as a good tool to teach power wheelchair driving skills similar to daily-living skills needed for wheelchair users [3]. Virtual reality has been used to train new wheelchair users to use power wheelchairs [3], [5], [6]. In these situations, the investigators usually use a joystick [3], [6] to interact with a software program. They are usually non-immersive and do not require much body function [3], [5], [6]. VR has also been used to check the accessibility of streets and public buildings for wheelchair users and for people with disabilities [101].

Although VR is a relatively new technology, there have been numerous applications reported for it. From applications in industry such as assembling [126], virtual prototyping [127], manufacturing [128], [129], mining [130], and large-scale industrial applications [131], to other applications in architecture, statistics, history, driving training [127], computer graphics [124], and nevertheless, in health sciences such as rehabilitation, therapy of people with disabilities [132], ophthalmology [133], and diagnosis of physical disabilities [132].

Archambault *et al.* [5] claim their power wheelchair simulator can adequately reproduce acceleration, stopping, turning, starting and inclination, using a motion platform. They say they expect their device to do a good job in "visual feedback" and inertial effect. However, they just tested it for inertia by using one adult participant. Archambault *et al.* in another study [6] recruited 29 participants and asked them to propel their wheelchair in real world and VR. The visual feedback was provided through a computer monitor. They used a questionnaire to measure how the participants would rate their sense of presence in VR. According to their results, driving performances, difficulties, and strategies were the same in overground power wheelchair and on the simulator.

Researchers in Human Engineering Research Laboratories (HERL) at the University of Pittsburgh made a VR for power wheelchair [2]; it was composed of a ramp and an ergometer using two thin rollers (diameter=~10 cm). The visual feedback was provided using three projectors projecting the view on three walls in front of the participant on an angle of about 15 degrees. It was used for training power wheelchair propulsion. They said, however, that it had a problem: It was hard to perceive the exact difference in driving skills of mediocre and good drivers, and the difference of driving skills of good and great drivers. Therefore, they proceeded to the CAREN system, the Computer Assisted Rehabilitation Environment. This system is a completely immersive VR with a six-degrees-of-freedom hydraulic platform [7]. This system allows participants to have physical interaction with it using some interfaces such as a motion capture system and a split treadmill. CAREN system was developed in the Netherlands by Motek Medical and now is used in many different placed around the world. One local example is Glenrose Rehabilitation Hospital, Edmonton, Alberta, Canada [134]. The CAREN system in the Glenrose Hospital is not currently used for wheelchair users, but the one in the University of Pittsburg is [2].

CEAL environment, the Challenging Environment Assessment Laboratory in the University of Toronto, is one of the world's leading institutions providing VR. The different conditions they are able to simulate in their VR are including winter Lab, driving simulator lab, street lab, and stairs lab [8]. They are currently working on developing a new wheelchair ergometer [109] that will be put in the driving simulator lab to make a VR for manual wheelchair propulsion.

2.10 Summary

In this chapter, we first made the case as to why it is necessary to conduct research studies focusing on wheelchairs and introduced three methods of simulating real world propulsion to support these research studies. We showed how a combination of wheelchair ergometer and VR can greatly help with studies that involve manoeuvring and explained how the literature had shortcomings about an ergometer-based virtual reality system with a direct approach, i.e. no motors used, for manual wheelchair users that encompasses inertial compensation and simulates wheelchair manoeuvring. The following research was conducted to build on this knowledge.

The following chapter enlightens why it is important to assess the reliability and validity of any research tool before employing it in research studies and also explains what kind of validity is relevant for this research.

3 A DESCRIPTION AND CLASSIFICATION OF DIFFERENT TYPES OF VALIDITY IN REHABILITATION

3.1 Abstract

There is a considerable body of literature on research validity across different disciplines. With regard to rehabilitation research, however, this body is narrow and warrants further consideration. Classification of research validity has been considered and developed over the past six decades; however, a literature search returned no comprehensive discussion that has gathered all available classifications under an overarching umbrella. The aim of this chapter is to provide an all-inclusive classification of different types of research validity, focusing on rehabilitation research.

A basic review of the body of literature available was conducted. Different classifications of validity in the literature were recognized, considered, unified, and are presented in this chapter. Moreover, the main threats to each type of validity

and some strategies to minimize them are discussed. Some examples are also provided.

A classification of all types of validity in rehabilitation research is presented in this chapter, a summary of which along with the main threats to each validity is depicted in the file attached to this chapter proposal.

Furthermore, the matter of priority between these research validities is discussed. It is concluded that while all types of research validities are important to be considered, maximizing all of them in one research project is sometimes controversial. Thus, researchers should make a situation-based trade-off between different aspects of validity in order to optimize the overall validity of their research.

Keywords: Research validity, construct validity, classification, threats to validity, rehabilitation research

3.2 Introduction

Getting concerned with the validity of a research is ensuring its empirical integrity [135]. The level of validity of a study is an indicator of cost-effectiveness and accountability of it [135]. A research study is considered valid when: (1) it is able to correctly find the relationship between the variables, (2) it measures what it claims to be measuring, (3) the findings are generalizable, and also (4) it has adequate statistical power to reject a false hypothesis. On the other hand, the power of a study is its ability to find the truth: correctly rejecting a false null hypothesis or supporting a true null hypothesis. Therefore, a valid study is also a powerful study, because its findings are the true results of that study.

All researchers need to ensure the validity of the tests and instruments they use before conducting research studies, but this is of particular importance for rehabilitation researchers: in a review study, 100 data-based studies were reviewed that were focusing on showing the effectiveness of rehabilitation procedures. By a post hoc power calculation, it was shown that there is a high possibility for occurrence of type II error in rehabilitation research studies [135]. Type II error refers to supporting a false null hypothesis which leads to a false negative conclusion. In the rehabilitation studies reviewed in this study, the medians of power for detecting small, medium and large effect sizes have been only equal to 0.08, 0.26 and 0.56, respectively. These low power values clearly show the importance of accounting for power and validity of the study in rehabilitation research, before starting to conduct one. This is because the purpose of all studies is truly discovering the relationship among variables [136], and if a study does not have validity, the results obtained from it will not be trustworthy.

Validity of the research should be considered for all research studies. Research studies concerned with validity have been done for a large variety of different fields of knowledge, including Education [137]–[146], Psychology [147]–[150], Marketing [151], Management [152], Employment [153], Nursing [154], Criminal issues [155], Animal studies [156], Sports [157], Nutrition and food science [158], Health [159]–[161], and Rehabilitation Science [135], [136]. Regarding the number of research studies about validity that are available in the literature for different disciplines, Education, Psychology, and Marketing are the fields that are most prominent. However, the body of literature in this context for health studies is narrow, and even narrower for rehabilitation science. This highlights the necessity of addressing this issue for health sciences in general, and rehabilitation sciences in particular. In this chapter, the validity of rehabilitation studies and different types of it will be discussed. In doing so, the main focus will be on one type of validity that is more complicated aspect of validation: construct validity. We will discourse some threats to each type of validity along with providing some strategies to prevent them, and hence, how to power up the study, as we pass through each validation type.

3.3 What Is Validity?

Validity principles are applicable to all studies, whether they are based on questionnaires, observational studies, or other types of assessments [141]. Research validity helps us know how true are the claims and propositions made in a study. This judgment can be based on the characteristics of a study, such as the research design, adequacy of sample size, the recruitment procedure, instruments and tests used, and the appropriateness of statistical methods used [162].

3.4 Types of Research Validity

Research validity can be categorized into two types: internal validity and external validity [163]. Internal validity refers to the amount of credit that can be attributed to the relationship between variables that is true, and external validity refers to how generalizable are the findings. In another approach [162], research validity has been divided into four types: internal validity, statistical conclusion validity, construct validity and external validity. These four types of validity address four basic questions that practicing researchers face and so it is more practical. Those basic questions are:

1) Is there any relationship between two variables? (Referring to internal validity)

2) If so, is it a causal relationship or it might just have happened by chance and can occur without any treatment? (Referring to statistical conclusion validity)

3) What are the concepts that are involved in this causal relationship? (Referring to construct validity)

4) How generalizable is this relationship to other settings, tools, persons, and time? (Referring to external validity)

The second classification is actually drawn by dividing each of the first classification components in two [162]: the statistical conclusion validity is differentiated from internal validity to distinguish between the relationship that is affected by covariates and the true relationship between the two variables that are of interest. In other words, internal validity takes care of the validity of the relationship obtained between the two main variables of interest, while the statistical validity makes sure that this relationship is not contaminated (and if so, is correctly taken care of) by other variables that may influence the relationship, but are not of the interest in the study.

Also, construct validity is differentiated from external validity to make a distinction between generalization of the constructs of cause and effect to higher order constructs and generalization of the findings to the other settings and the population. In other words, the second classification explicitly states what was implicitly covered in the first classification.

The second classification of validity is now widely accepted and is being used by researchers in different fields [135], [136]. We will briefly introduce each of these types of validity in the following sections of this chapter. Furthermore, some threats to each type of validity as well as some strategies to power up that validity will be discussed.

3.4.1 Statistical conclusion validity

Statistical conclusion validity can be defined as the approximate precision of interpretations concluded about the covariations based on the statistical methods used and the fitness of the research design [136]. In other words, any statistical issue that could influence the results of the study would be a threat to statistical conclusion validity, including small sample size, lack of statistical power for the study [135], and any random error that could happen by chance, despite appropriate use of statistics, e.g. type I and type II errors [164]. These threats to validity happen when the research conditions and statistical process of the study are not rigorous enough [136]. This type of validity is the *most important* type of validity, but in rehabilitation research, it has received little attention [135].

Some of the main threats to statistical conclusion validity are [136], [162]:

Low statistical power: this will reduce the power of the study to reject the null hypothesis. To prevent this threat, researchers should define their experiment characteristics, e.g. sample size and eligibility criteria, in order to provide an acceptable statistical power for the research.

Violated statistical assumptions: each statistical procedure is based on some assumptions, e.g. normal distribution of population, homogeneity of variance, and linearity. If these assumptions are violated, those statistical procedures will not be credible anymore. To prevent this threat, the researcher should ensure that the underlying assumptions are met, e.g. normality of sample data.

Performing multiple statistical tests on a single data set: this increases the likelihood of type I error and is another threat to statistical validity. This is because the Alpha level of the study as a whole is the sum of the Alpha level of all comparisons made. To prevent this threat, researchers can reduce the number of comparisons by a careful pre-planning. Another technique is to determine the Alpha level of each t-test in a way that the cumulative Alpha would be desired.

Lack of inter- and intra-rater reliability of the study: lack of consistency in conditions of implementing experiments is a threat to validity.

Lack of reliability of measures: this would result in both Type I and Type II errors (e.g. they might show no correlation between two variables, when there is a good correlation, and vice versa). To prevent this threat, researchers must only use reliable measures or tools. In fact, reliability is a necessary condition for validity [148].

3.4.2 Internal validity

Internal validity deals with whether the treatment used in the research study has an actual effect on the outcome variable [164]. Thus, internal validity is the extent to which we can be confident that there is a certain type of relationship (e.g. causal) between the dependent and the independent variables of the study [136]. If a study has lack of internal validity, then there are some factors in that study, other than the independent variables, that affect the outcome to some extent, but they have not been accounted [136], [164]. These unaccounted factors are threats to internal validity. There are many threats to internal validity which can be applicable to different types of research. Some of the most common threats are [136], [162], [164]:

Maturation: when experiments get lengthy, the results may be influenced by the participants getting older, wiser, healthier, or stronger. Maturation is considered a threat to validity when it has not been considered in the research design and is not accounted for.

History: When subjects' reaction in the experiment has been influenced by some events that have happened prior to the experiment, e.g. when the participant is studied to observe his/her attitude to people with disabilities after some treatments, but in fact his/her past experiences of encountering a person with disability would affect the results.

Lack of inter and intra-rater reliability will also affect the results and so the causal relationship concluded from those results.

Selection: the researcher should assure that what makes a difference between control and treatment groups is the only factor that is under study. Any other variables that might influence the outcome should not systematically differ between groups. The best protection against this threat is randomly assigning participants to treatment and control groups when possible.

Attrition and mortality: drop-outs that usually happen most for the group that is receiving the harder treatment would influence the outcome, since those who comply with the treatment are generally those who are healthier or are more enthusiastic about the research study they have participated in. Also, mortality can potentially happen for every experiment that needs the attendance of participants for more than one time, but especially it may happen for studies where the participants have severe diseases, such as cancer or heart disease.

Sharing information: If by any means participants have a chance to share their information regarding the experiment, they might influence each other's thoughts and outcomes. This is even more important when the experiment is based on questionnaires and other qualitative methods.

If participants know about other participants in a different group, they may get dissatisfied with the group that they are in and with the treatment type they are receiving and thus they may become disappointed and less motivated to appropriately continue with the treatment (e.g. performing some exercises), which may affect their compliance. Any protection against chances of getting cues about other participants will help to prevent sharing of data. Any strategies that can be taken against hypothesis guessing by participants and, also performing a good concealment for the randomization will also contribute to this aim.

3.4.3 External validity

External validity refers to the population that the study findings can validly be generalized to. In other words, the greater the population the findings can be applied to, both in number and diversity, the higher the external validity of that study. This, however, depends significantly on the sample used in the study and the population that this sample is actually representing. A simple example for this is that if you are aiming for generalizing your findings to all wheelchair users but you only recruit male wheelchair users, you have a problem in external validity of your research. Also, the instruments used in the study and other conditions of the experiments have roles in determining the population the study findings can be generalized to and so in defining the external validity of the study.

This type of validity addresses the generalizability of the findings, which is of particular importance for rehabilitation practitioners, because they need to make inferences from the sample under the study to the treatment provided to a greater population [136]. For making those inferences, we either should perform the same experiment in different occasions of time, settings, participants, etc., or one should perform a systematic review with meta-analysis on the body of literature on a particular issue [164]. However, randomly selecting participants of a study (i.e. random sampling) will provide the best protection against threats to external validity of that study and therefore makes the findings generalizable to the population [136]. Some of the main threats to external validity are [136]:

Sample characteristics (e.g. age, gender, race, education, urban versus rural) restrict the population that the findings can be generalized to. Samples should be a good representative of the desired population. Random sampling helps considerably to this aim, but it does not guarantee that this threat would be eliminated.

Intervention characteristics also restrict the findings to the settings with similar features, e.g. instruments. Some strategies to prevent this threat include: making use of different examiners (when intra-rater reliability is realized) and using multiple measures that are taken from multiple setting.

Context characteristics: There are some conditions that may influence the way subjects react or respond in the experiment, that inhibit generalization of the findings to the situations with different conditions. For instance, some participants try to provide \`correct\` responses, which are responses that they believe examiners like to see, but are not representative of their real state. Moreover, sometimes subjects are receiving multiple treatments at the same time, which may restrict the findings to those people that are on similar treatment regimes.

Sensitization: In research designs that participants receive the same assessment (e.g. questionnaire) pre- and post-test, their knowledge on the way they will be evaluated might affect the outcomes. Since this situation might not be the same as when there is no sensitization on the construct under study, this is a threat to external validity.

3.4.4 Construct validity

For performing powerful studies and also proper clinical accomplishments, we need to make use of robust measures, considering the fact that "science rests on the adequacy of its measurement" [148]. Using proper and robust measures pertains to construct validity. Construct validity deals with whether an instrument or measurement tool or a test is measuring what it claims to be measuring [145]. In other words, construct validity concerns whether the measurement obtained is really representing the underlying construct [148].

The matter of validity is analogous to a study that has a clear hypothesis; researchers should gather as much evidence as they can to prove the hypothesis about validity of the inference [145], [147], [148]. Researchers should continue gathering convincing evidence until they feel that they have a large enough set of evidence to prove the construct validity. There is no best way in validating a study, although, there are several methods in use [145]. Up to four subclasses of construct validity have been defined: face validity, content validity, criterion-related validity, and construct validity. As it is reflected, construct validity is a division of one type of validity that is called construct validity. This is because some researchers [145], [148], [164], [165] believe that all these subclasses should be grouped and gathered under one overarching umbrella which is the construct validity. This is Called the unified view of construct validity. Each of the subclasses of construct validity will be discussed below.

3.4.4.1 Face validity

This is the first judgment about the validity of an instrument by just looking at the appearance of it. It is only guesswork and provides little evidence for validity of the instrument. Besides, some snags can happen when talking about face validity [148]: the fallibility of verdicts that are based on appearance, different interpretations of appearance between developers and users, and some occasions that the judgment based on the appearance of an instrument is contrary to its contents. Therefore, using only face validity is never sufficient.

3.4.4.2 Content validity

This type of validity concerns about the items or elements of a measurement and the extent to which these elements reflect the area they are supposed to be measuring [165], [166]. The adequacy and fitness of each element of the measurement tool in measuring the targeted construct is discussed under content validity. In other words, the targeted construct guides selecting the content of an assessment tool and on the other hand, the content and elements of the assessment tool selected, defines the construct that is actually being investigated [166]. Content validity is particularly considered in assessing the validity of questionnaires.

Using an instrument that is invalid due to content results in erroneous conclusions, because some aspects of the construct are not represented properly, whether underrepresented or overrepresented. Not accounting for content validity in the study could also result in inaccuracy in finding a significant treatment effect [165].

Content validity has a dynamic nature, because the domain and definition of constructs change by time, and accordingly, the elements of an instrument should be changed to be representative of that construct [165]. Content validity of an assessment instrument is dependent on the function of that instrument, population under study, and the situation in which the instrument is used. Therefore, close attention should be taken in order to maintain an acceptable validity for the assessment instrument. Usually, a panel of experts is contacted to judge about content validity of an assessment or instrument [145], [165].

One example of threats to content validity is occasions where the definition of one term in researchers' language is different from the commonly accepted definition of it. In this situation, the readers' interpretations of the contents, results, and reports of the study might be different from the author's and researcher's intent. In these cases, a proper clarification of the constructs is advised [136].

3.4.4.3 Criterion-related validity

In criterion validity, correlation of the instrument or assessment with a `criterion` is examined [145]. Criterion has to be a `superior` measure that is more accurate than the measure being evaluated, otherwise, the failure in validation might be due to a flaw of the criterion, itself [166]. There are two types of criterion validity: [145], [166] concurrent and predictive, which are introduced briefly, here:

Concurrent validity: This pertains to situations that both the assessment tool that is being tested for its validation and the criterion are measured at the same time. For instance, when blood pressure is measured simultaneously using cuff measurements and intra-arterial pressure measurement tools[166].

Predictive validity: When the assessment tool is tested for its validity by checking how well it can predict a criterion that will happen later. For instance, how well the scores of a test obtained by a sample of people predicts their job status in the future [166]. Diagnosis, physiological data, and tests performed in laboratories are examples of instances that predictive validation should be used [166].

3.4.4.4 Construct validity

In construct validity, we experimentally investigate whether a construct is actually measuring what it claims to be measuring [145]. The concept of construct validity emerged when researchers realized there are many occasions that there is not any `superior` criterion to correlate the instrument under validation study with it [166]. Two types of validity can be distinguished within construct validity [166]:

Convergent validity: Convergent validity is used when in validating a method of measurement, the correlation between that measure and a different method of measurement is assessed, while they are used to measure the same construct [167].

Discriminant (divergent) validity: When we experimentally show that our assessment tool being tested for its validation produces results that are different from data produced by another assessment tool that is measuring another construct and thus should produce different results [167].

Lack of construct validity causes two deficits: [164]

Contamination: When the scores obtained by the assessment tool represent features that are not part of the construct being studied.

Deficiency: When there are aspects of the construct being studied that are not included in the assessment tool.

Some threats to construct validity are listed hereafter [136], [145], [147].

Using non-reliable tools: When reliability is threatened, the construct validity of the device can also be threatened.

Narrow stimulus sampling: When the researcher studies a narrow sample or situation while the construct under study is much broader. Case studies are particularly subject to this threat.

Single operations: When the construct is complex, but the measure inspects just one aspect of it. For example, the researcher measures only the 'time spent with friends' as the indication of being happy. To avoid this threat, the researcher should make use of more indicators for the construct.

Single subject design: This design is a threat to construct validity when it is used to implement an intervention or a treatment. This is because the individual's specifications may be responsible for the outcome resulted, not the intervention itself.

Experimenter expectancies: It happens when a researcher has passion and some expectations about the outcome, in a way that it influences his/her interpretation and explanation of the results. This may cause alternative explanation of the relationship between variables, which decreases the construct validity, since it is not declaring the real circumstances of the construct under study.

If by any means **participants get some clues about the study**, it will affect their performance in the experiment. This is because participant might presume the hypothesis or objective of the study, and act, in the sake of the objective, differently from their usual real behavior. This, in fact, changes the construct that has been assessed, since the study is assessing them when they are `motivated`. To prevent or minimize this threat, researchers should attempt to provide fewer cues for participants.

The same situation is established when participants from one group have **the tendency to compete** with the participants of the other group. This also makes them more motivated. To avoid this threat, researchers should minimize the incidental contacts between subjects.

Demoralization: this happens when some individuals from one group are not satisfied with the treatment they are receiving, comparing to the treatment that the other group receives. This makes them less motivated in participating and affects their performance. A solution for this threat is providing another valued treatment for the group that is suspected for demoralization. Of course, this treatment should be a placebo or proved to have no intervening effect on the study.

Mono-method bias: Happens when the researcher uses only one method of measurement. If this measurement has a poor construct or content validity, the study would be flawed. A method for minimizing this threat is using a number of methods at the same time, e.g. questionnaire, self-report, and observation.

Mono-operation: Happens when just one manipulation is used to affect the construct. For instance, to investigate the effect of a special drug as an intervention, participants are divided to one placebo group and one treatment group. A more solid design will establish multi-group and sets different dosage of the drug for each group.

Poor construct definition: This is when the construct is misdefined (e.g. assessing anxiety instead of depression), or has not been defined properly (e.g. assessing job satisfaction to represent overall happiness). Researchers should get advice from experts in the field before starting the study, to prevent this threat.

Figure 3-1 depicts a summary of this paper; the overall classification of validities and main issues that threaten different types of validity.

3.5 Discussion

In this paper, four types of research validity with their subclasses were described. Now, one may ask, among these different types of validity, which one of them is more important and has priority in consideration in validity study? The answer is rather complicated due to different opinions existing in the literature, which will be described here.

Among different types of validity, construct validity is the one that more frequently has been subject to consideration, research, and publication, e.g. [145], [147], [152]. This is because it appears that for some researchers, construct validity (and its subclasses) are the only concepts of research validity. Although this frequency of consideration relative to the other types of validity could be an indication of relatively greater importance for this type of validity, the authors could not find any explicit declaration of that.

However, as it was pointed out in the `statistical conclusion validity` section, Ottenbacher and Barrett [135] believe that this type of validity (statistical conclusion validity) is the most important type of validity, though it has received little attention in rehabilitation research. The importance of this type of validity is that one should make sure that the findings are obtained as a result of a real covariation between variables, rather than chance.

Cook's and Campbell [162] and Mitchell [164], however, do not believe so, since they declare that the internal validity is the most important validity and hence, one should be more concerned about it. This is because this group of researchers believes that as long as a study does not have internal validity, the data achieved from it are not appropriate and trustworthy, and as a result, they are not eligible to be generalized to other situations. Mitchell says that most authors are not concerned with external validity and they just put it under `further research` heading [164]. Bellini and Rumrill gather between these two former opinions for studies that investigate an unknown relationship between variables [136].

Some researchers, however, argue that at the end of the day, we need to generalize our data to other situations, and hence, if they are valid but unable to be generalized, they are useless. They, therefore, believe that external validity deserves higher priority than what it has received in health research so far [168].

Another point that is worth noting here is the point drawn by Shadish *et al.* that construct validity is not a necessary condition for external validity, because "we can generalize across entities known to be confounded albeit less usefully than across accurately labeled entities" [169].

Bellini and Rumrill [136] state that in fact, all four research validities are important in turn, and one should try to have all of them in the higher level possible, but there is a logical order for them: first statistical conclusion validity should be established, to show that two variables covary. The second validity to be established is the internal validity, which focuses on the obtained relationship between variables. Then the researcher should be concerned with the construct validity which speaks about the construct that is involved in the relationship. Eventually, the importance of external validity would arise that is concerned with the generalization of the results to other settings.

To sum up, it should be noted that all validities are of great importance of value for all research. However, one may not be able to maximize all of them at the same time, since maximizing one of them could be dependent on decreasing the other one. For instance, for increasing statistical conclusion validity with a given sample size, one may need to restrict the population under study, in order to decrease the variation between samples and increase the statistical power of the study. This, in turn, obviously leads to a reduction of generalizability of his/her results (lower external validity). Researchers, therefore, need to make trade-offs between different aspects of research validity -considering the conditions they are in; so that they could end up with the optimum validity for their research study.

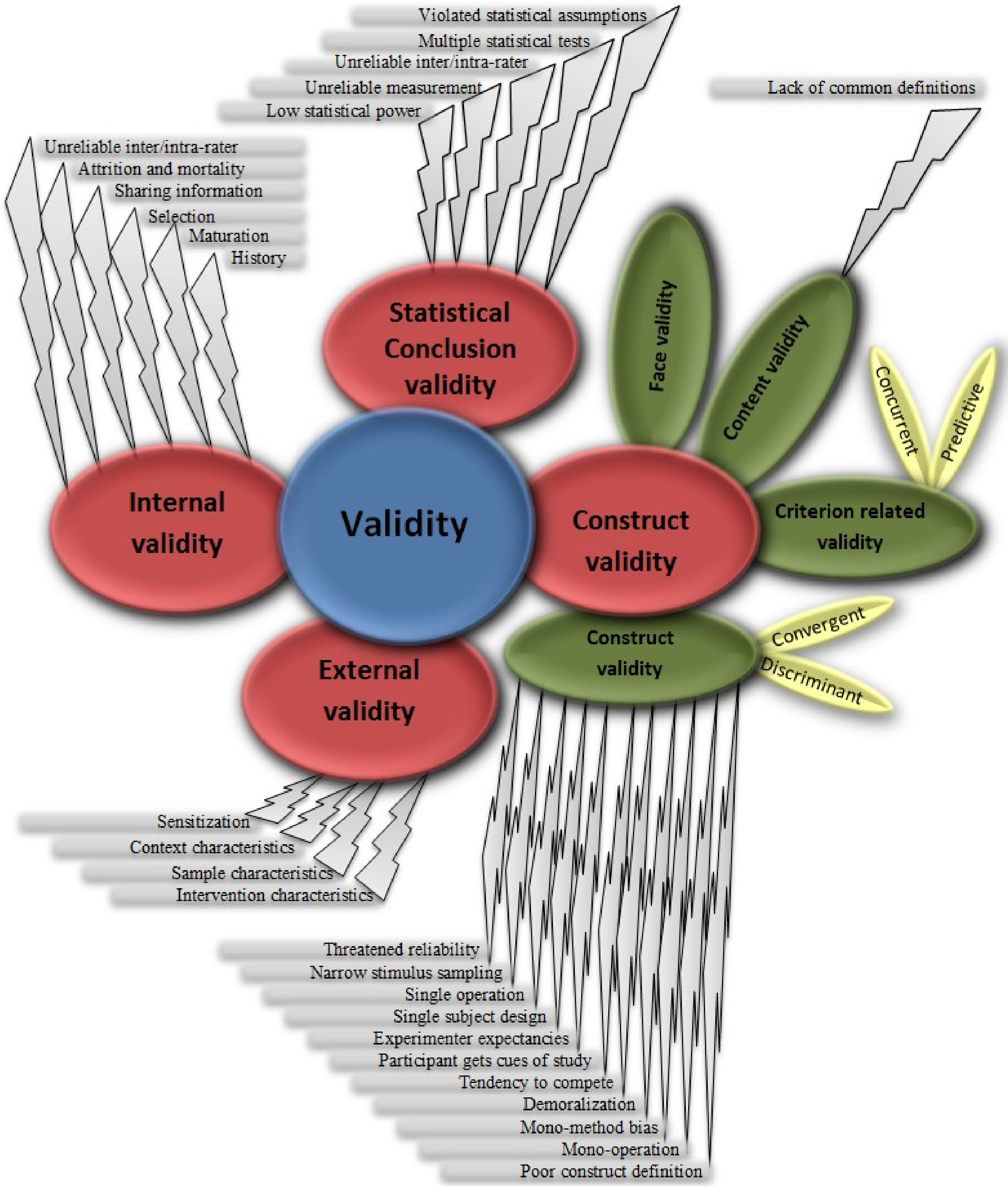


Figure 3-1 Overall classification of validities and main issues that threaten different types of validity (sparks)

The following chapter is included as a practical example for how ergometers can facilitate research studies involving wheelchair propulsion. This chapter focuses on using rating of perceived exertion (RPE) as a self-regulating exercise intensity for wheelchair users. To do so, we assessed the correlation of local and overall RPE readings of the participant with the biomechanical and physiological outcomes of exercises with different intensities: from every-day activity level to exertion.

For this chapter, a separate and stand-alone set of experiments were completed by recruiting 11 experienced wheelchair users. The ergometer used in this set of experiments was a two-roller ergometer with no adjustments for liner or rotational inertia.

4 Wheelchair users' perceived exertion during typical mobility activities

4.1 Abstract

Study design: Each participant performed a series of wheelchair exercises equivalent in intensity to minimal functional speed (1 m s^{-1}), functional walking speed (1.3 m s^{-1}), a relatively challenging speed (1.6 m s^{-1}) and a self-selected speed. Each participant also completed a graded exercise test (GXT) to volitional exhaustion (VO_{2peak}).

Objectives: The purpose of this study was (1) to assess the physical capacity of wheelchair users as they undertake typical mobility activities and (2) to investigate how closely the components of a differentiated model of perceived exertion mirror wheelchair users' own perception of exertion.

Methods: Eleven (eight males and three females) spinal cord-injured or congenitally impaired wheelchair-dependent participants volunteered for the study. Differentiated ratings of perceived exertion ($RPE_{\_arm}$ and $RPE_{\_respiration}$) and oxygen uptake (VO_2) and heart rate were recorded during each exercise.

Results: The mean comfortable speed at which the participants propelled their own wheelchairs on the wheelchair ergometer was 1.1 ± 0.2 m s^{-1}. Speeds of 1 m s^{-1} and 1.3 m s^{-1} are typical of everyday functional propulsion. The corresponding $RPE_{\_respiration}$ and $RPE_{\_arm}$ ranged from 7 to 13 on the Borg scale; the $\%VO_{2peak}$ measured in these trials ranged from 37 to 80% VO_{2peak}. For propulsion intensities used in the present study—low, moderate, high and graded exercise intensity—no difference could be observed between $RPE_{\_respiration}$ and $RPE_{\_arm}$. There were no significant differences between $RPE_{\_arm}$ and $RPE_{\_respiration}$ at the termination of the GXT.

Conclusion: The current study showed potential for the use of RPE to assess and monitor daily wheelchair propulsion intensity in individuals with paraplegia.

4.2 Introduction

People who become wheelchair dependent for mobility because of a spinal cord injury (SCI) must face a double challenge: they have to overcome new obstacles in carrying out activities of daily living and adapt to new ways of maintaining their physical fitness. If they do not keep fit they may become less able to carry out their daily tasks and they may see their general health deteriorate and become subject to such risks as obesity, diabetes and cardiovascular disease [170]–[172].

Although some activities of daily living provide in themselves opportunities for exercise, guided physical training may be included during the initial phase of rehabilitation to initiate a program of routine exercises. Such programs are recommended for maintaining fitness in the long term and a physically active lifestyle with the associated health benefits [15], [173]. It stands to reason that wheelchair users need and want to play a major part in designing and monitoring their individual program, and much of their input and choices will be based on their own perception of what constitutes a reasonable intensity of exercise. Objective measurements in a clinical setting may help assess the level of exercise intensity, but consistent adherence to an exercise program is likely based on the wheelchair user's own perception of exercise intensity.

Borg's rating of perceived exertion (RPE) scale has become an accepted tool in both the assessment and prescription of exercise [174]. There have been a few studies in which wheelchair users' metabolic responses and RPE were recorded during wheelchair exercises and hand-cycling training [79], [175], [176], but they have generally used treadmill or hand cycling at relatively high intensities for elite wheelchair athletes [177], [178]. Daily activities at lower intensities have greater relevance for the majority of wheelchair users. One of the strategies for promoting regular physical activities is assisting people in making close estimates of daily physical activity levels that are conducive for maintaining satisfactory physical capacity.

The purpose of this study was therefore (1) to assess the physical capacity of wheelchair users as they undertake typical mobility activities; and (2) to relate the differentiated perceived effort to physical capacity. As the majority of wheelchair activities do not require maximal effort but rather a repetitive sustained sub-maximal effort, muscle fatigue may be a particularly relevant aspect of wheelchair propulsion. Where upper body exercise is concerned, wheelchair users' arms may fatigue sooner than their cardiorespiratory system [79]. The Differentiated RPE model considers the two perceived signals separately: one peripheral, from the working muscles, the other metabolic, from the cardiorespiratory system. It is thus possible to determine when fatigue is greater in one or the other. Most of all, we are interested in finding a good correlation between wheelchair users' RPE and physical capacity so that they can use this simple scale to self-monitor their daily activity intensities in a way that would benefit their health. Beyond these results, we see an additional value of this study in its pilot nature, intended to show possible directions of future research with more, and less heterogeneous, participants.

4.3 Materials and Methods

4.3.1 Participants

Eleven (eight males and three females) wheelchair-dependent participants with SCI at the T6 level or below volunteered for the study. Table 4-1 shows the injury and physical characteristics of each participant.

Oxygen uptake (VO_2, l min^{-1}), carbon dioxide output (VCO_2, l min^{-1}) and minute ventilation (V_E, l min^{-1}) were continuously measured using a computerized gas analyzing system (Oxycon Mobile, Jaeger, Bunnik, The Netherlands). Respiratory-gas exchange measurements were obtained every 5 s. System calibration was undertaken before each trial. Heart rate (HR, beats per min) was monitored continuously by telemetry (Timex, TIMEX Group Canada, Inc., Markham, ON, Canada).

The participant's own wheelchair was secured to an instrumented roller ergometer, which was connected to a monitor placed in front of the participant to provide visual speed feedback. The ergometer consisted of two independent cylindrical steel rollers with radius 0.158 m and a mass of 26.4 kg, one for each rear wheel, supported by pillow-block bearings (NSK P208, Tokyo, Japan) within a wooden structure to support the wheelchair. Work load was controlled through friction applied to each roller by a fabric strap attached to pneumatic actuators of a digital pressure controller (FESTO, Esslingen am Neckar, Germany), with a proportional valve to regulate the required air pressure. The desired work load through friction was controlled by a computer program (NI LabVIEW 2012, National Instruments Corporation, Austin, TX, USA)

4.3.2 Test procedure and ratings of perceived exertion

RPE were measured using the 15-point Borg scale. Before starting the exercise protocol, all participants received an orientation, including standardized instructions on how to report their feeling of exertion: ease/difficulty of breathing as $RPE_{\_respiration}$, perceived body temperature and the localized exertion in shoulders and arms as $RPE_{\_arm}$.

Table 4-1 Characteristics of participants

Participant code	Sex	Age (year)	Weight (kg)	Type of injury	ASIA impairment scale grade	Time since injury (year)
1	M	35	50.4	SCI	T10 AISA A	2
2	M	45	68.3	SCI	T12/L1 AISA A	2.5
3	M	41	97.2	SCI	T6/T7 AISA A	18
4	M	34	68.4	SCI	T12 AISA A	17
5	M	49	80.5	SCI	T11 AISA A	2
6	M	33	93.1	Spina bifida	T10	18
7	M	47	99.2	SCI	T6 AISA B	12
8	M	44	125.4	SCI	T11 AIS A	10
9	F	55	73.4	SCI	T11 AISA A	3.5
10	F	29	57.8	Spina bifida	L2	12
11	F	51	64.1	SCI	T12 AISA A	18
Mean (s.d.)		42.1 (8.4)	79.8 (22.0)	—	—	10.4 (6.9)

Abbreviations: ASIA, American Spinal Injury Association; F, female; M, male; SCI, spinal cord injury; s.d., standard deviation.

Participants warmed up for about 5 min while getting used to the ergometer and visual propulsion speed feedback. They were then asked to perform a set of 3-min wheelchair propulsion bouts at different speeds. Data were recorded at a self-selected comfortable speed, 1 m s^{-1} (minimal safe speed to cross an intersection with traffic lights), 1.3 m s^{-1}, which is equivalent to typical ablebodied walking speed, and 1.6 m s^{-1} (the upper limit of a self-selected speed among people with paraplegia). The order of exercise bouts was randomized. Perceptual ratings ($RPE_{\_respiration}$, and $RPE_{\_arm}$) were obtained at the end of each exercise bout. A 5-min passive rest period was given between bouts. The rest period also allowed the participants' HR to return approximately to baseline.

After a 15-min rest, participants performed a graded maximal exercise at a constant speed of 1 m s^{-1} to exhaustion. The work load was set at 10 W and then increased by 5 W every minute until exhaustion. Two of the participants were engaged in regular paraplegia sports. For them the work load was increased in steps of 10 W so that volitional exhaustion could occur within 8–14 min. The end point of the test was determined when the participant volitionally stopped because of fatigue or the investigators determined that the participant could not maintain the expected speed after three warnings. Participants were asked every 2 min to give two ratings ($RPE_{\_respiration}$ and $RPE_{\_arm}$) of perceived exertion by nodding to the applicable numbers on the Borg scale. The Borg scale was placed in full view of the participants throughout the exercise trials.

4.3.3 Data analysis

The mean values of the oxygen uptake were calculated over the final 30 s of each constant speed trial. For the graded exercise tests (GXTs), regression analysis was used to identify the time windows equivalent to 40, 60 and 80% VO_{2peak}, and then RPE values, HR and work load were determined in reference to their respective time windows. Metabolic peak values observed at the termination of GXT were used to normalize the values measured during constant speed trials.

4.3.4 Statistics

Statistical analysis was performed using SPSS (SPSS for Windows Version 16.0; SPSS, Inc, Chicago, IL, USA). The normality of the test data was confirmed by the Shapiro–Wilk test (P40.05). The analysis of covariance (ANCOVA) was used to compare two regression lines ($RPE\_{respiration}$ and RPE_{arm}) by controlling $\%VO_{2peak}$ and work load for the graded exercise test. One-way analysis of variance was used to compare $\%VO_{2peak}$, HR, V_E, $RPE\_{respiration}$ and RPE_{arm} between different wheelchair propulsion speeds. The paired t-test was used to compare the difference between $RPE\_{respiration}$ and RPE_{arm} values recorded at different speeds and GXT. All data are reported in the text as mean± s.d. Significance was set at Po0.05 for all statistical procedures.

Statement of ethics. We certify that all applicable institutional and governmental regulations concerning the ethical use of human volunteers were followed during the course of this research.

4.4 Results

4.4.1 Graded exercise tests

Peak values observed at the termination of GXT are reported in Table 4-2. Analysis of covariance showed that there were no significant differences between $RPE\_{respiration}$ and RPE_{arm} when regressed against $\%VO_{2peak}$ and work load. Table 4-3 shows RPE responses, power output, HR and $\%HR_{max}$ at different $\%VO_{2peak}$ levels. The paired t-test showed no significant difference between $RPE\_{respiration}$ and RPE_{arm} at different $\%VO_{2peak}$ levels during the GXT tests.

4.4.2 Constant speed tests

Descriptive statistics of the constant speed wheelchair propulsion tests are reported in Table 4-4. The mean comfortable speed held by the participants was 1.1 ± 0.2 m s−1. Speeds of 1 and 1.3 m s−1 are daily functional propulsion speeds. The RPE reported by the participants ranged from 7 to 13; the mean $\%VO_{2peak}$ corresponding to these trials was 53.8 ± 10.3 to 63.7 ± 15.2$\%VO_{2peak}$. A propulsion speed of 1.6 m s−1 represents a relatively more strenuous intensity; the RPE reported for mean $RPE\_{respiration}$ and RPE_{arm} was 12.4 ± 2.1, and 12.1 ± 2.0, respectively. The HR and $\%VO_{2peak}$ during 1.6 m s−1 propelling were 121 ± 20 beats per min (0.79% HRmax) and 71.6 ± 11.6$\%VO_{2peak}$, respectively. ANOVA shows that 1.6 m s−1 propulsion has significantly higher values on $\%VO_{2peak}$, VE and $RPE\_{respiration}$ than do 1 m s−1 propulsion and self-selected propulsion. There was no significant difference between $RPE\_{respiration}$ and RPE_{arm} in any of the trials.

4.5 Discussion

4.5.1 Daily activity zone for wheelchair users

We expected that propelling a wheelchair at the minimal functional speed (1.0 m s^{-1}), functional walking speed (1.3 m s^{-1}) and comfortable speed (1.1± 0.2 m s^{-1}) is a physical activity of low to moderate intensity. The corresponding RPE ranges from 7 to 13 on the Borg 6–20 scale; the corresponding $\%VO_{2peak}$ ranged from 37 to 80% and the HR from 40 to 60% VO_{2peak}, with the corresponding RPE ranging from 7 to 10 [179]. Compared with able-bodied people, relatively higher RPE and $\%VO_{2peak}$ values reported in the daily activity intensities among wheelchair users can be attributed to the dependency on arm exercise, the extent of paralysis, reduced sympathetic control and relative inactivity, all of which can compromise physical capacity in SCI [180].

We observed that participants chose a speed ~ 1.1 m s^{-1} as a comfortable propelling speed on the ergometer. The corresponding intensity ~ 53% VO_{2peak} and HR averaged 104 beats per min (0.69% HR_{max}). Ratings of perceived exertion during self-selected speed propulsion averaged 8.9 ± 1.9 for overall rating and 9.3 ± 2.5 for peripheral rating. The preferred intensity of exertion selected by the participants is within expected ranges of RPE (7–12 on Borg's 6–20 scale) and relative tolerance for exercise (for example, 36–69% of VO_{2peak}). We think the preferred intensity is safe and health promoting for most community-based wheelchair users. They are more likely to adhere to their own preferred exercise intensity than to adjust to a level based on precise physiological criteria if those criteria conflict with their intensity preference. Further studies are needed on how

Table 4-2 Peak values observed at the termination of the GXT

Test	VO_{2peak} (ml min^{-1} kg^{-1})	HR_{rest} (BPM)	HR_{max} (BPM)	VE_{peak} (l min^{-1})	RER_{peak}	Work load (W)	Time (s)	RPE (respiratory)	RPE (arm)
GXT	16.3±4.2	81±11	151±14	63±23	1.5±0.3	63.2±17.4	620±172	17.1±2.1	17.5±2.2

Abbreviations: BPM, beats per min; GXT, graded exercise test; HRmax, maximal heart rate; HRrest, heart rate at rest; RER, respiratory exchange ratio; RPE, rating of perceived exertion; VE, expired volume per unit time; VO_{2peak}, peak oxygen consumption per unit time.

NOTE: values are mean±s.d.

Table 4-3 RPE values, HR and work load during graded exercise test at 40, 60 and 80% VO_{2peak}.

% VO_2peak	RPE_respiration	RPE_arm	HR(beats per min)	%HR_{max}	Work load (W)
40	9.4±0.7	9.2±0.9	103.4±18.8	0.66±0.01	15.7±1.9
60	11.2±1.8	10.9±1.8	114.0±15.5	0.77±0.06	27.2±10.4
80	13.7±2.1	13.6±2.3	133.8±17.1	0.89±0.04	40.9±17.1

Abbreviations: %VO_{2peak}, percentage peak oxygen consumption per unit time; HR, heart rate; RPE, rating of perceived exertion.

NOTE: values are mean±s.d.

Table 4-4 Descriptive statistics reported during constant speed wheelchair propulsion tests

Variables	Self-selected speed (1.1±0.2m s^{-1})	1m s−1	1.3 ms^{-1}	1.6 ms^{-1}
% VO_2peak	53.2±9.3[a]	53.8±10.3[a]	63.7±15.2	71.6±11.6[a]
HR(beats per min)	104±18	103±16	111±17	121±20
$\%HR_{max}$	0.69±0.09	0.68±0.10	0.73±0.10	0.79±0.11
VE ($lmin^{-1}$)	23±6[a]	24±6[a]	28±6	36±8[a]
RPE_respiration	8.9±1.9[a]	9.8±2.2[a]	10.3±1.7	12.4±2.1[a]
RPE_arm	9.3±2.5	9.6±2.4	10.3±2.6	12.1±2.0

Abbreviations: $\%VO_{2peak}$, percent of peak oxygen consumption per unit time; HR, heart rate; % HR_{max}, percent of maximal heart rate; VE, expired volume per unit time; RPE, rating of perceived exertion. [a]Significant difference between propulsion speeds (Po0.05).

NOTE: Values are mean±s.d.

the preferred exercise intensity, combined with an appropriate duration and frequency, enhances health outcomes and fitness enhancing benefits.

It is encouraging to note that, as shown in Table 4-4, the reported RPE in the self-selected speed (~1.1 m s^{-1}) bout was similar to the RPE reported in the 1 m s^{-1} bout. As the bout order was randomized and none of the participants had any prior exposure to RPE scales, this indicated that RPE results are fairly reproducible at similar exercise intensities.

4.5.2 Differentiated RPE model

Perceptual dominance has been demonstrated in able-bodied subjects performing cycle ergometer and treadmill exercises [181]–[183]. As for wheelchair users, they rely entirely on the upper limbs for both ambulation and weight-bearing tasks. The shoulder is poorly designed for this purpose, and thus becomes exposed to excessive, repeated interarticular pressures in conjunction with a more abnormal distribution of stresses across the subacromial area. The differentiated RPE model suggests that discrete perceptual ratings are linked to specific underlying physiological events [183], [184]. We set out to examine how measured exercise intensity was reflected in differentiated RPE, considering local exertion felt specifically in the upper limbs, as well as overall exertion. It was hypothesized that wheelchair users would report a similar $RPE_{\_arm}$ and $RPE_{\_respiration}$ at low to moderate exercise intensities but that at the relatively higher intensities the participants might report higher $RPE_{\_arm}$ values. In the present study, for wheelchair propulsion at low, moderate and graded exercise intensity our results showed no difference between differentiated $RPE_{\_arm}$ and $RPE_{\_respiration}$. This finding is in agreement with results previously reported, which indicated no significant differences between $RPE_{\_arm}$ and $RPE_{\_respiration}$ during moderate and vigorous exercises among trained SCI people [79]. We also found that there were no significant differences between $RPE_{\_arm}$ and $RPE_{\_respiration}$ at the termination of the GXT. Goosey-Tolfrey and colleagues [79] observed that well-trained wheelchair athletes reported significantly higher peripheral RPE compared with overall RPE at the termination of the GXT and the ramp exercise test.

Our participants are community-based wheelchair users. With a look at the years since injury, some participants with a shorter injury history did report higher $RPE_{\_arm}$ than $RPE_{\_respiration}$. Further research is needed with a larger sample size and more homogeneous participants to test the effect of injury history and strength training of the upper limbs on differentiated RPE model.

4.5.3 Practical applications

Designing exercise programs manageable enough to be adhered to while sufficiently intensive to allow adequate cardiovascular conditioning are essential for individuals seeking to maintain an exercise program. Wheelchair pushing provides both cardiovascular conditioning and improves muscle endurance [185]. The present study provides evidence that wheelchair users' daily mobility activities fall within a training range that would not only result in cardiovascular conditioning, but also feels most comfortable to them. Public health guidelines call for 30 min of moderate intensity exercise 'most days' (5–7 days per week) [186]; the exercise plan is based on doing 'as much as one reliably can' rather than 'as much as one possibly can'. Instructions that guide participants towards a judiciously self-selected exercise intensity may establish a sense of ownership and encourage long-term adherence among a wide range of individuals. Our results also demonstrate that RPE is a valid tool for tracking low to moderate exercise intensity.

4.5.4 Study limitations

Our choice of wheelchair propulsion assessment on the ergometer offered methodological benefits such as simulated graded propulsion. However, preliminary investigations prior to data collection showed that the ergometer had a higher rolling resistance than an indoor tile surface. It has been reported that propulsion velocity decreases with an increase in rolling resistance [187]. That is likely to have been a factor in the relatively low self-selected speed of our participants.

The current study has shown encouraging potential for the use of RPE to monitor daily wheelchair propulsion intensity in persons with paraplegia. However, the findings of this study are limited by a small sample size. To see whether more severely impaired individuals with quadriplegia can benefit from these findings, validation within that additional group would be desirable. The effect of exercise duration and frequency is a potential avenue of enquiry to extend the recent work in this area. More studies are needed to validate the accuracy and repeatability of RPE to monitor exercise intensity, particularly when the wheelchair propulsive exercise bouts involve low to moderate intensity efforts.

4.6 Conclusion

This study assessed the physical capacity of community-based wheelchair users for functional wheelchair speeds that are typically required to complete activities of daily living. We found a good correlation between physical capacity and RPE during low to moderate daily activity intensities. The daily activity intensities among the participants ranges from 37–80% VO_{2peak}, the RPE corresponds to these intensities ranges from 7–13. No differences between differentiated $RPE_{\_respiration}$ and $RPE_{\_arm}$ were found for wheelchair propulsion at low, moderate, high, as well as graded exercise intensities in the present study. RPE could be used as a simple tool to assess and monitor the activities in which wheelchair users engage in their daily tasks, their recreation and exercises.

The following chapter focuses on developing some models for a roller-based ergometer in **order to replicate linear inertia** that is adjustable for everybody. This chapter discusses all the underlying biomechanical factors that need to be met to have a sound replication of wheelchair propulsion on an ergometer.

5 Ergometers Can Now Biomechanically Replicate Straight-Line Floor Wheelchair Propulsion: Three Models Are Presented

5.1 Abstract

Wheelchair ergometers facilitate wheelchair related studies as they allow controlled experiments to be performed inside the laboratory. However, the results obtained from these experiments are of limited value unless we use wheelchair ergometers that biomechanically represent real-world wheelchair propulsion. We could not find any wheelchair ergometers in the literature to date, that have been validated considering all of these important biomechanical criteria: Velocity and acceleration, force and moment, trunk swing, inertial effect, energy consumption and the resistive force against propulsion.

In this paper we have considered wheelchair propulsion on an ergometer and have compared it to straight-line floor wheelchair propulsion. From equating these two situations, we have found the necessary conditions to meet the above criteria. Finally, we propose three models for a wheelchair ergometer that satisfies these conditions and will be able to biomechanically represent straight-line floor wheelchair propulsion.

Keywords: Wheelchair ergometer, straight-line propulsion, biomechanical representation

5.2 Introduction

The need for wheelchair studies is undeniable; one percent of the population are wheelchair users, each with specific needs and circumstances. Researchers have conducted wheelchair related studies for several years now. Since the invention of the first wheelchair, (first patent: 1937 [1]), numerous modifications to wheelchair design have been introduced, as well as few wheelchair propulsion simulators such as ergometers and treadmills. Results from these ergometers are of limited value, because existing ergometers have not been shown to correctly represent Floor Wheelchair Propulsion (FWP) considering all relevant mechanical aspects. Here, we will briefly review the literature to date.

Gonzalez-Quijano *et al.* [2] devised a controllable wheelchair ergometer which can work in four modes of simulation, e.g. constant power. They did not, however, include any verification or validation for their device. Dabonneville *et al.* [3] designed an ergometer which they claim could measure biomechanical parameters (force and moment) in real life conditions. They used just one able-bodied subject for the verification test who performed straight line wheelchair propulsion. Gil-Agudo *et al.* [4] considered kinetics of shoulder joint during straight line wheelchair propulsion on a treadmill. They used a SMARTWheel and inverse dynamics technique to find shoulder joint reactions for 16 experienced wheelchair users and concluded that reaction force in shoulder increases while speed increases.

Wu [5] considered straight line wheelchair propulsion on two types of carpets with different resistance and tried to simulate that on an ergometer by controlling the resistance of ergometer's rollers against being rotated. She had her special focus on the transition between these two surfaces. She recruited 25 unimpaired subjects for this study, and concluded that using the formula introduced in her report, ergometer wheelchair propulsion can replicate floor wheelchair propulsion. She unfortunately experienced technical challenges with the control system requiring the use of an arbitrary correction factor.

Finley, *et al.* [6] used a sample of 10 able-bodied subjects propelling their wheelchair in a straight line and on an ergometer. They did three submaximal fatigue tests and concluded that majority of biomechanical variables obtained from the experiments were test-retest reliable. Devillard *et al.* [7] used a sample of 17 non-wheelchair user subjects for validation of an ergometer (VP100H) through comparing the obtained variables with those obtained from a force transducer which was placed underneath of ergometer. The task was straight line propulsion with and without trunk movement. They concluded that VP100H is a valid measurement instrument.

Kwarciak *et al.* [8] compared handrim biomechanics on floor (low pile carpet in the lab) and on a special wheelchair accessible treadmill (MAX Mobility, LLC). They did a cross-sectional study on 28 experienced wheelchair users and recorded contact angle, peak force, peak axle moment, power output and cadence in over ground propulsion and then for the second round of experiments, they set the treadmill speed to be the same as first round and used a metronome to guide the subject to maintain the same cadence as the first round. They gradually increased the power output until it reached to the same power output of their first experiment. Other variables were recorded to check correlation between two rounds' results. The correlation coefficients were all greater than 0.85, and they concluded that this treadmill is a valid substitute for floor wheelchair propulsion. They did not, however, consider energy expenditure of subjects, while this is definitely one of the possible sources of differences between two systems due to the lack of kinetic energy ($1/2\ mv^2$) in treadmill wheelchair propulsion. They also did not consider trunk movement.

DiGiovine [108] *et al.* used Laplace transformation and introduced three models for a dynamic test to be used for calibration of a dynamometer, in order to provide all the necessary information for the results to be extendable to real-world situations. They specifically focused on resistance and inertia for the rollers and the wheels of a wheelchair on a dynamometer. Their equation, however, was developed for a dynamometer which includes motors and cannot be used for non-motorized ergometers. While all these studies reported valuable information regarding introducing new wheelchair ergometers or validity of wheelchair ergometers or treadmills in wheelchair studies, there is still a shortage of studies presenting or introducing a biomechanical model for a wheelchair ergometer that has been shown to be able to comprehensively represent the biomechanics of floor wheelchair propulsion. In this paper, we have considered Ergometer Wheelchair Propulsion (EWP) from all basic mechanical aspects, and compared it to straight-line FWP, and eventually, we have introduced our model for EWP to be representative of straight-line FWP.

5.3 Methods

Before considering FWP and its representative model in EWP, we will give a brief introduction about friction in wheelchair propulsion. This will be covered first, because for representing FWP, it is critical in the understanding of friction in wheelchair propulsion.

Secondly, we will provide complete equations of motion governing FWP and EWP. Again, before starting to solve the problem of wheelchair propulsion, we have to have a clear idea of Free Body Diagram (FBD) of the system and also the related equations of motion. These equations will be referred to later in the paper.

Thirdly, all basic mechanical criteria for making the representative model of FWP on EWP will be mentioned and discussed in turn. Considering all these criteria we will obtain necessary conditions for the representative model and based on these conditions, we will offer two representative models of FWP, later in this paper.

5.3.1 Friction in wheelchair propulsion

Sliding friction: As we all know, when we apply a force (F) to an object in order to move it, a friction force (f) would accordingly start to appear and grow. As we are increasing the driving force, the friction would grow accordingly, until a point called f_{max} (Figure 5-1Figure 5-1 Schematic diagram of sliding friction). Yet, there is no movement in this epoch and the friction is called static friction. By slightly increasing the F beyond this amount, movement will happen and friction force will (usually) be decreased to f_{dy}. Friction in this epoch is known as dynamic friction.

Thus, friction type in wheelchair propulsion is static friction (because we need no relative movement between the wheel and the ground) and therefore there is no specific magnitude for friction, because it changes according to our propulsive force. In both EWP and FWP, this friction is a useful force which makes propulsion possible for us. There is another set of friction forces that resist wheelchair propulsion and increases when surface roughness is increased. This 'unhelpful' friction is a different type of friction that is present where something is rolling on a surface, and is called: rolling resistance. Both frictions are present in wheelchair propulsion.

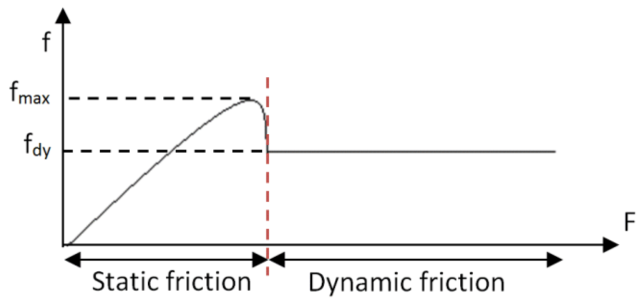


Figure 5-1 Schematic diagram of sliding friction

Rolling friction: This friction appears when an object is rolling on a surface. Although there are some experimental equations reported which are applicable only in some situations with limited conditions, no general equation has been introduced for it yet. This is the force resisting rolling a wheel, i.e. in cars, bicycles, and wheelchair driving. It is dependent on tire material [1], wheel diameter and dimension [3, 1], caster characteristics [1], load on wheel and distribution of it [1], tire tread design [2], and some other factors. This is the resisting force that we feel it when we are driving a wheelchair. This friction will be shown as “f” in this paper.

5.3.2 Governing equations of motion

In this section, Newton equations of motion for the wheelchair ergometer will be derived. It should be said that in this paper, we assume a wheelchair ergometer to consist of two rollers that are placed on a platform (see Figure 5-2). The wheels of wheelchair place on these rollers, so they can rotate, while the seat is fixed to the platform.

Equations of motion for the wheel, roller, caster (front wheel) of one side (assuming symmetrical conditions), as well as equations of motion for the subject and the seat as a package will be derived here; first for EWP and then for the FWP situation.

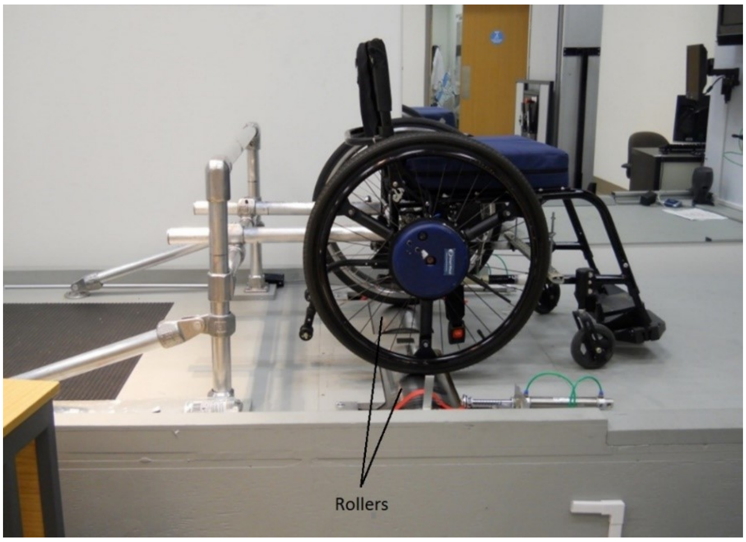


Figure 5-2 Wheelchair ergometer considered in this paper [188]

5.3.2.1 EWP, Wheel

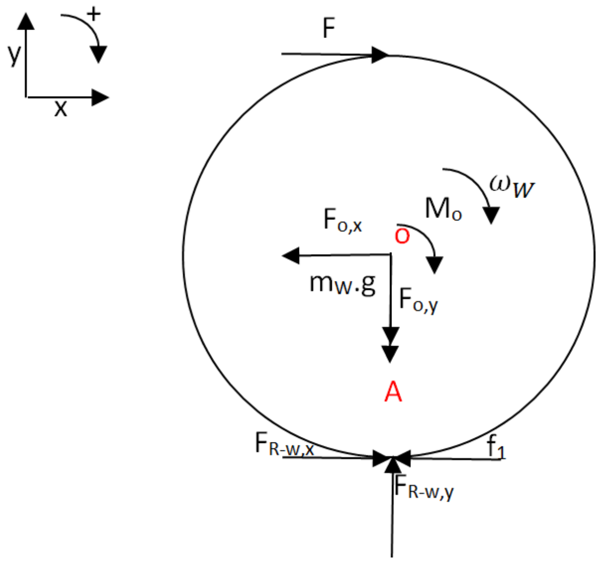


Figure 5-3 Free body Diagram (FBD) for governing Newton equations of motion for wheel on EWP

$$\sum \boldsymbol{F_x} = \boldsymbol{0} \quad =\gg \quad \boldsymbol{F_{o,x}} + \boldsymbol{f_1} = \boldsymbol{F} + \boldsymbol{F_{R-w,x}} \tag{1}$$

$$\sum F_y = 0 \quad =\gg \quad F_{R-w,y} = m_w.g + F_{o,y} \tag{2}$$

$$\sum M_o = I_{w,o}.\alpha_w \tag{3}$$

$$=\gg \quad M_O + F.r_w + (f_1 - F_{R-w,x}).r_w = I_{w,o}.\alpha_w$$

5.3.2.2 EWP, Roller

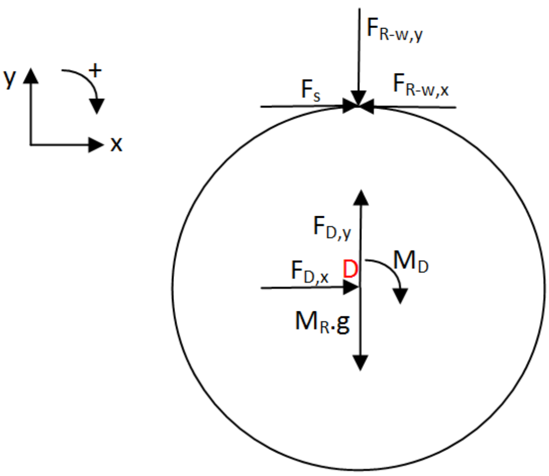


Figure 5-4. FBD of roller for governing Newton equations of motion on EWP

$$\sum \boldsymbol{F_x} = \boldsymbol{0} \quad =\gg \quad \boldsymbol{F_s} = \boldsymbol{F_{R-w,x}} - \boldsymbol{F_{D,x}} \tag{4}$$

$$\sum F_y = 0 \quad =\gg \quad F_{D,y} = m_R . g + F_{R-w,y} \tag{5}$$

$$\sum M_D = I_{R,D} . \alpha_R \tag{6}$$

$$=\gg \quad M_D + (F_s - F_{R-w,x}) . r_R = I_{R,D} . \alpha_R$$

5.3.2.3 EWP, Subject + seat

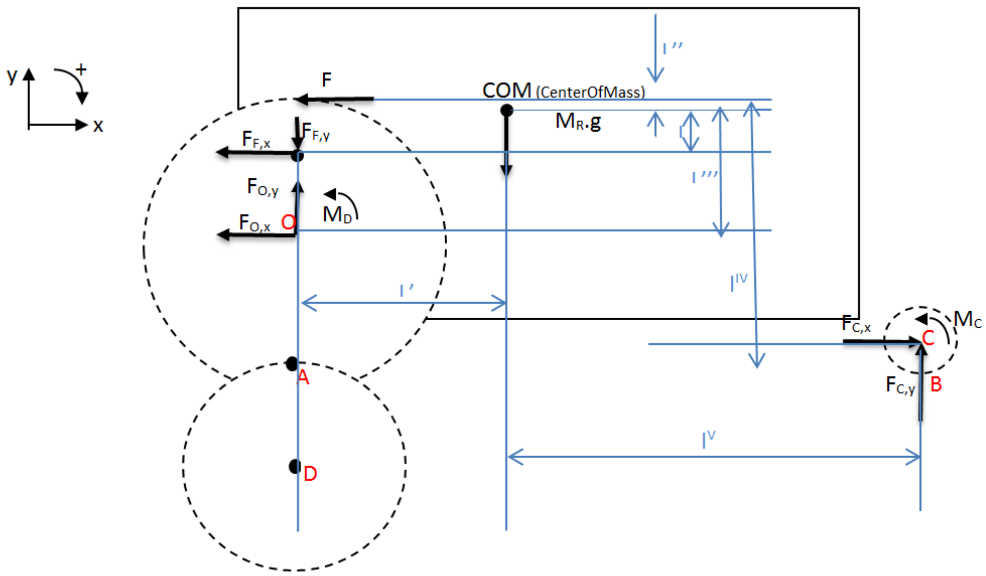


Figure 5-5 FBD of subject+ seat for governing Newton equations of motion on EWP

$$\sum \boldsymbol{F_x} = \boldsymbol{0} \quad =\gg \quad \boldsymbol{F_{o,x}} + \boldsymbol{F_{c,x}} = \boldsymbol{F} + \boldsymbol{F_F} \tag{7}$$

$$\sum F_y = 0 \quad =\gg \quad \frac{m_{tot}}{2} . g = F_{O,y} + F_{C,y} - F_{F,y} \tag{8}$$

$$\sum M_{COM} = 0 \tag{9}$$

$$=\gg \quad F_F . l + F_{O,y} . l' = F . l'' + F_{O,x} . l''' + M_O + M_C + F_{C,x} . l^{IV} + F_{C,y} . l^{V} + F_{F,y} . l'$$

5.3.2.4 EWP, Caster

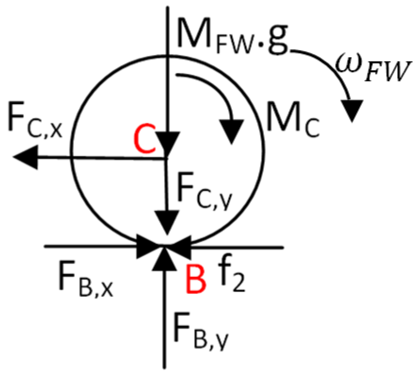


Figure 5-6. FBD of caster for governing Newton equations of motion on EWP

$$\sum \boldsymbol{F_x} = \boldsymbol{0} \quad =\gg \quad \boldsymbol{F_{C,x}} = \boldsymbol{F_{B,x}} - \boldsymbol{f_2} \tag{10}$$

$$\sum F_y = 0 \quad =\gg \quad F_{B,y} = m_{FW} . g + F_{C,y} \tag{11}$$

There is no rotation for casters, so:

$$\sum M_C = 0 \quad =\gg \quad M_C = (F_{B,x} - f_2) . r_{FW} \tag{12}$$

5.3.2.5 FWP, Wheel

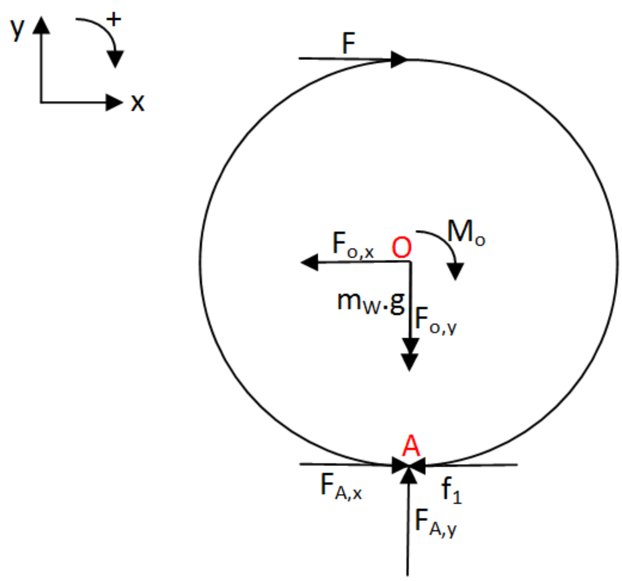


Figure 5-7. FBD of wheel for governing Newton equations of motion on FWP

$$\sum \boldsymbol{F_x} = \boldsymbol{m_W.a} \quad =\gg \quad \boldsymbol{F + F_{A,x} - f_1 - F_{o,x} = m_W.a} \tag{13}$$

$$\sum F_y = 0 \quad =\gg \quad F_{A,y} = m_w.g + F_{o,y} \tag{14}$$

$$\sum M_o = I_{w,o}.\alpha_w$$

$$=\gg \quad M_O + F.r_w + (f_1 - F_{A,x}).r_w = I_{w,o}.\alpha_w \tag{15}$$

A is IC (instant center of rotation) , so:

$$\begin{cases} a_O = \propto_W.r_w = a \\ \sum M_A = I_{W,A} \propto_W =\gg M_O + F \times 2r_W = I_{W,A}.\propto_W + F_{x,o}r_w \end{cases} \tag{16}$$

5.3.2.6 FWP, Subject+ seat

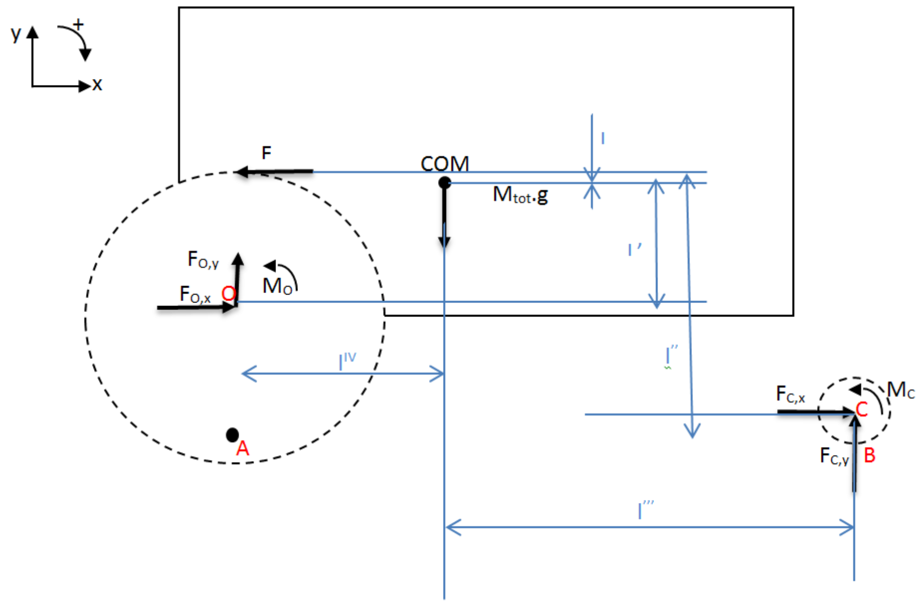


Figure 5-8 FBD of subject+seat for governing Newton equations of motion on FWP

$$\sum \boldsymbol{F_x} = \boldsymbol{m_{ss}.a} \quad =\gg \quad \boldsymbol{F_{o,x} + F_{c,x} - F = \frac{m_{tot}}{2}.a} \tag{17}$$

$$\sum F_y = 0 \quad =\gg \quad \frac{m_{tot}}{2}.g = F_{O,y} + F_{C,y} \tag{18}$$

$$\sum M_{COM} = 0 \quad =\gg$$

$$F_{O,y}.l^{IV} = F.l + F_{O,x}.l' + M_O + M_C + F_{C,x}.l'' + F_{C,y}.l''' \tag{19}$$

5.3.2.7 FWP, Caster

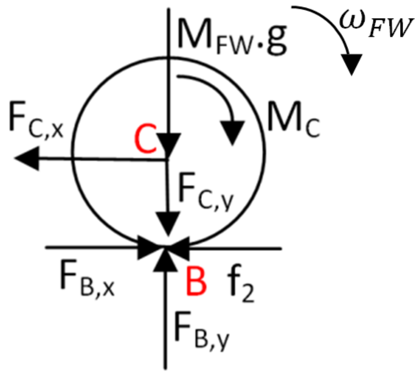


Figure 5-9 FBD of caster for governing Newton equations of motion on FWP

$$\sum \boldsymbol{F_x} = \boldsymbol{m_{FW}}.\boldsymbol{a} \quad =\gg \quad \boldsymbol{F_{C,x}} + \boldsymbol{m_{FW}}.\boldsymbol{a} + \boldsymbol{f_2} = \boldsymbol{F_{B,x}} \tag{20}$$

$$\sum F_y = 0 \quad =\gg \quad F_{B,y} = m_{FW}.g + F_{C,y} \tag{21}$$

$$\text{B is IC} \quad =\gg \quad \sum M_B = I_{FW,B}.\alpha_{FW}$$

$$=\gg \quad M_C = F_{C,x}.r_{FW} + I_{FW,\mathrm{B}}.\alpha_{FW} \tag{22}$$

$$\text{Also:} \quad \sum M_C = I_{FW,C}.\alpha_{FW}$$

$$=\gg \quad M_C = (F_{B,x} - f_{2m}).r_{FW} + I_{FW,C}.\alpha_{FW} \tag{23}$$

5.3.3 Biomechanical factors and criteria needed in modelling FWP

There are five basic biomechanical criteria need to be the same in FWP and EWP for good representation. These criteria are:

- Resistive force/ torque against propulsion
- Applied force
- Velocity and acceleration
- Inertial force
- Energy consumption

Another important criterion that should be considered is the role of trunk movement in wheelchair propulsion. This case, along with the other criteria mentioned has been investigated in here.

Resistive force/ torque against propulsion: In order to find a relation for the rolling friction (f), we need to use Equation 1 to Equation 23 (equations of motion in EWP and FWP). It should be said that f_l in EWP is portion of f_l of FWP which is related to hysteresis, because roughness of floor surface is not existent in roller's surface. The portion of surface roughness in the f_l of FWP is compensated in EWP by Fs.

Although we tried many ways to find "f" in terms of factors such as F, ∝, ω and geometrical parameters, there are always at least two unknown forces acting in a same line of affection, and subsequently they appear in equations in added-up format. Therefore, we always just ended up with an equation for "CF-f" instead of "f" (CF is driving forces in contact points of wheels with floor/ergometer; so simply any force in these points other than f is CF). Equations 24 to 26 show the approach for finding CF-f in EWP, and Equations 27 and 28 show the approach for finding CF-f in FWP.

From Equation 1 and 10:

$$\boldsymbol{F_{R-w,x} - f_1 + F_{B,x} - f_2 = F_{o,x} - F + F_{C,x_1}} \tag{24}$$

Using Equation 7:

$$\boldsymbol{F_{R-w,x} - f_1 + F_{B,x} - f_2 = F_F} \tag{25}$$

Considering $f_1 + f_2 = f$ and $F_{R-w,x} + F_{B,x} = CF$ we have (for only one side):

$$\boldsymbol{CF - f = F_F} \tag{26}$$

For FWP, considering Equations 13, 17, and 20 we have:

$$\boldsymbol{F_{A,x} + F_{B,x} - f_1 - f_2 = \frac{m_{tot}}{2}.a} \tag{27}$$

Considering $f_1 + f_2 = f$ and $F_{A,x} + F_{B,x} = CF$ we have (for only one side):

$$\boldsymbol{CF - f = \frac{m_{tot}}{2}.a} \tag{28}$$

Considering "CF-f" as the force exerted at point A, we can simulate this force in both cases. These (CF-f)s can be equal, because F_s (surface roughness portion of resistance) has been already affected $F_{w,x}$ in EWP. In FWP, CF (contact point friction) is always available, but not fixed. It is changing from an initial value to a maximum (associated to yielding point in Figure 5-1). It is the thing that is preventing wheel from slipping. So both in static and dynamic situation it is available and acting. However, during dynamic situations, while there is no applied force, there is a minimum CF-f, nonetheless, it is a negative number, showing that f is greater than CF. CF may also be zero, but due the fact that CF and f are co-lined, we cannot say it for sure.

On the other hand, the same situation will go for EWP. Again, the force acting in the wheels' contact points with floor/roller is CF-f. We can experimentally find this force through a free roll-down test both in EWP and FWP, and then equate them in our experiments. In a free roll-down test, the only acting force is friction and therefore it can be obtained by measuring time and velocity. By providing the same CF-f in both FWP and EWP, we can claim that participant is experiencing the same resistive force during propulsion.

The way to do this is firstly, we will need to do a series of roll down tests for FWP on some surfaces with different roughness and friction. A roll down test is defined as a test in which the participant accelerates the wheels and then releases it. Velocity of the chair at the time of release and, also the time it takes for the wheel takes to stop should be recorded. Using the following equation, wheel's resistive angular deceleration would be obtained.

$$\boldsymbol{\alpha} = \frac{\boldsymbol{\omega}}{\boldsymbol{t}} \tag{29}$$

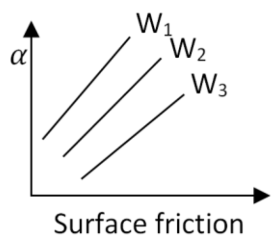


Figure 5-10 A typical curve resulted from a roll down analysis for FWP. W1, W2 and W3 are different weights

Considering $T = I\alpha$, we have T_surface friction curve, and from it we will have CF-f_surface friction curve in FWP ($\mathrm{T} = (\mathrm{CF} - \mathrm{f}) \times \mathrm{r_w}$) where CF-f belongs to the situation of recovery phase (rather than push phase in one push cycle).

Then we need to perform some roll down tests for EWP with different piston pressures to build a $\alpha - \mathrm{Pp}$ curve. In the same way, we will have T-Pp curve and then CF-f_Pp curve. This all was just to show how these roll down tests will be related together, but in practice, we just need to find α −surface friction curve in FWP, and by using α as an input, find Pp from $\alpha - \mathrm{Pp}$ curve in EWP.

It should be noticed that because friction is affected by the weight of 'participant plus the wheelchair', some separated test should be done for some different weights (see Figure 5-10). It is also reasonable to do a series of calibration roll down tests for different weights, so for other weights, the right curve would be achieved by conducting an interpolation/extrapolation.

Applied force: Simply consider the participant is applying the same force. We will show that after meeting all other conditions, if the participant applies the same propulsion force in both situations, the ergometer would acceptably represent the FWP. This is the aim of the study; however, if he/she applies a different propulsion force, the situation would be different relative to the F that has happened in FWP, but similar to an imaginary F in corresponding conditions in floor wheelchair propulsion (FWP).

Velocity and acceleration: In order for the subject to have the same experience of wheelchair pushing both in EWP and FWP, he/she needs to have the same velocity and acceleration. Now the question is: how a wheelchair can have velocity and acceleration when it does not have displacement? Well, we will consider relative speed and acceleration instead. As we know:

$$\boldsymbol{V} = \frac{\boldsymbol{dx}}{\boldsymbol{dt}} \tag{30}$$

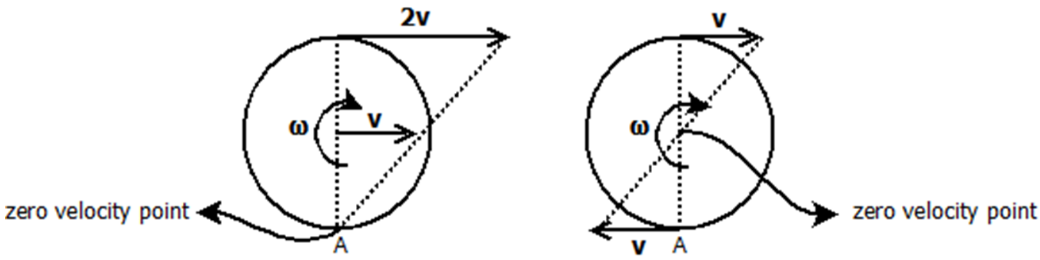
In FWP, overall velocity of wheelchair is V. This V causes dx meter displacement in dt second. On the other hand, wheeling on EWP causes the point A on the roller to rotate. The same V in dt seconds would cause dx meter displacement. So in the same epoch of time, virtual displacement in EWP is the same as the one in FWP.

So basically they have the same velocity (absolute velocity in FWP and relative velocity in EWP). As it is shown in Figure 5-11, both systems also have the same angular velocity (Equations 31 and 32). The same goes for acceleration, too.

$$FWP:\ \boldsymbol{\omega} = \frac{V_O}{\boldsymbol{r_W}} = \frac{V}{\boldsymbol{r_W}} V = \frac{dx}{dt} \tag{31}$$

$$EWP: \omega = \frac{V_A}{r_W} = \frac{V}{r_W} \tag{32}$$

Where r_W is radius of wheel, V_O is velocity of the center of wheel and V_A is velocity of point A in Figure 5-11.



(a) (b)

Figure 5-11 Velocities of three points of wheel in (a) FWP (b) EWP

Trunk movement: Generally, in straight-line motion (or curved-lines with sufficiently large radiuses) we could represent a rigid body with a point in its center of mass (COM). So we just consider the whole body as a point and solve kinetic and kinematic equations for it.

Consider a FWP situation when subject A (who is sitting on the wheelchair) is using trunk movement technique to increase his/her displacement, velocity and acceleration. As it is shown in Figure 5-12 , when the wheelchair user is not using his/her trunk in FWP, the COM will be almost in the same place relative to the person (it changes a bit though). However, when he/she takes advantage of trunk movement in FWP, he/she is basically increasing the displacement of COM during push phase of propulsion; so the hands are doing the same thing (L (meter) displacement), but trunk is also participating in “moving COM ahead” and makes the “l (meter)” difference in COM displacement during push phase. If the velocity and acceleration of system in the end of push phase is the same, the COM would have the same path and displacement during recovery phase (as it is shown in Figure 5-12); but the final velocity at the end of push phase in upper trace is greater than the ones in lower trace, due to the fact that during push phase (basically the same time in both trace) COM has a greater displacement in upper trace, which leads it to have greater velocity. As a result, the displacement length in the recovery phase would be greater in the upper trace rather than lower trace (Equation 32), which doubles the cause of having a greater overall displacement of upper trace in a complete cycle.

There might be some other reasons for trunk swing helping wheelchair propulsion. One of them is the push length that will be increased as a consequence of leaning backward when grabbing the wheel (hand could go further back on wheel), and leaning forward at the end of push phase (hand could go further ahead on wheel). Another possible advantage of this technique is that it affects the recruitment pattern of muscle contractions for doing the task.

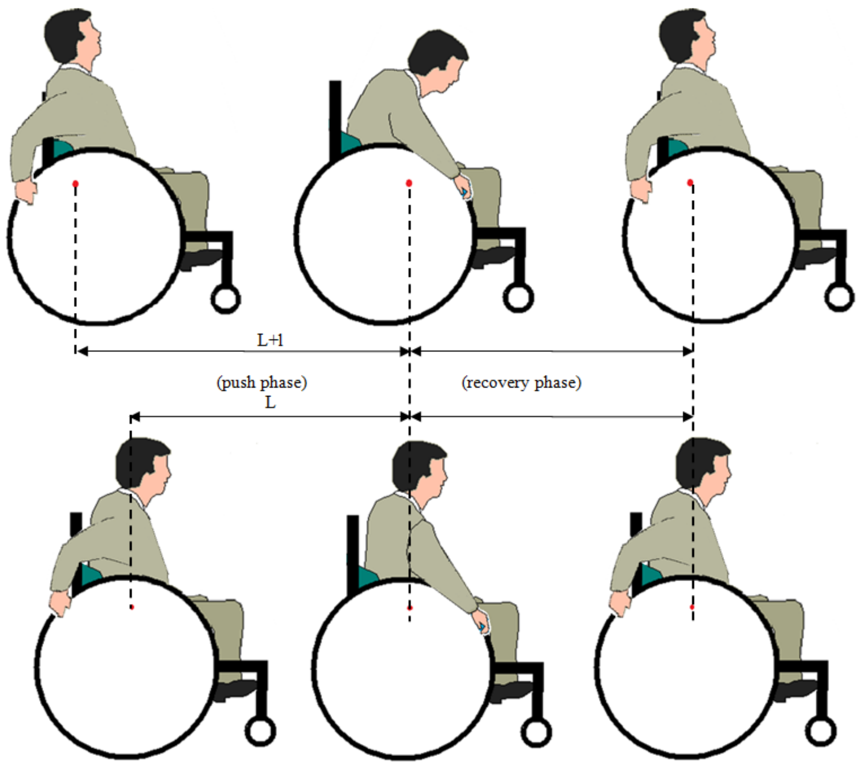


Figure 5-12 The role of trunk swing in FWP/EWP. Red spots are representing system COMs. Upper trace: using trunk movement in propulsion, Lower trace: FWP/EWP without swing trunk

We can see that in both FWP and EWP, the body can take advantage of different muscle contraction patterns as well as enlarged push length during trunk swing.

Regarding the added displacement of wheelchair during one cycle of propulsion due to trunk movement (distance l in 'L+l', Figure 5-12), an explanation is needed: considering the second explanation for this added displacement, we can see that the effect of trunk movement in displacement of our reference point (center of wheel) is seen in recovery phase, where there is no propulsion force. Therefore, the only effective force in this period is rolling friction, and COM will move in its path according this basic kinematic equation:

$$\boldsymbol{x} = -\frac{\mathbf{1}}{\mathbf{2}}\boldsymbol{a}\,\boldsymbol{t}^{\mathbf{2}} + \boldsymbol{v}_{\mathbf{0}}\boldsymbol{t} \tag{33}$$

$$a = \frac{f}{m} \tag{34}$$

Where f is rolling friction and v_0 is velocity of the COM at the end of push phase. Due to moving trunk from forward to backward, our reference point will go ahead for l (m) extra displacement. This displacement could not be achieved without increasing the velocity of the wheels; the velocity of COM will not be changed though. So we can see that during recovery phase, velocity of wheels has been increased relative to the one in propulsion without trunk swing. The same goes for EWP: the velocity of the wheels will be increased due to trunk movement in the recovery phase, and this enhancement of wheels velocity will lead to enhancement of the rollers' velocity which represents virtual displacement in ergometer. So, from this perspective, EWP could represent FWP too. We just need to act the same in both situations: using trunk swing or not.

For these conditions, the effect of trunk swing is the same for both EWP and FWP, but there is still another point of view remained, to check if these two situations are technically the same in using trunk movement. As it was stated before, basically COM gets a higher velocity at the end of push phase in the trunk swing situation due to passing a greater path in the same period of time. Also, x (displacement of COM in recovery phase) is dependent on this velocity (Equation 33). So in FWP, x is greater using swing trunk technique. However, in EWP, trunk movement in push phase just increases the fixture force, and consequently, reaction force in points D and B (Figure 5-5). Unfortunately, none of these forces have an effect on rotation of wheels or rollers (Equations 4 and 10). So in EWP, this possible additional effect of trunk swing is absent, and for this reason, EWP may not thoroughly represent FWP.

Inertial effect: Here we will give an example to give a better and more approachable understanding of inertial acceleration and inertial force which is absent in EWP. So, if we attach an Inertial Measurement Unit (IMU) to chest of our subject and ask him/her to do EWP and FWP, we will observe that in EWP there is no significant magnitude for linear acceleration in frontal axis (just some noise), while in FWP, this acceleration is available and observable (IMU's results). Well, this is the missing inertial acceleration in EWP, which would become inertial force when multiplied by mass. This inertial acceleration is created due to displacement of the COM, which is equal to zero in EWP. However, there is still something moving in EWP which is absent in FWP that we count on it for compensating this missing inertial force: the roller. When we start wheeling in EWP, we are not overcoming wheelchair's+our inertia, but we have to overcome the rollers' inertia to make them rotate. Here, we equate these inertial effects in these two cases:

EWP:

$$\boldsymbol{T_{EWP} = 2 \times I_R . \propto} \tag{35}$$

Where T_{EWP} is inertial torque of the rollers in EWP and I is moment of inertia of one of the rollers around its central axis. We assume that both rollers are the same and have the same moment of inertia; so the number 2 in this equation is because there are two rollers which the subject is making them rotate. Now we need an equation for *I*:

$$\boldsymbol{I_R = \int r^2 . dm = \frac{1}{2} m_R {r_R}^2} \tag{36}$$

$$T_{EWP} = 2 \times \frac{1}{2} m_R {r_R}^2 \times \propto = m_R \, {r_R}^2 \propto_R \tag{37}$$

And:

$$\propto_R = \frac{a_A}{r_R} \tag{38}$$

Considering condition of no slip: $\propto_R = \frac{a}{r_R}$

Where a_A is acceleration of point A in Figure 5-5 Now:

$$T_{EWP} = m_R \, {r_R}^2 \frac{a}{r_R} = m_R . r_R . a \tag{39}$$

To overcome to this torque, subject's hand has to apply a force which leads to a CF-f which creates T_{EWP}. So inertial torque is:

$$T_{EWP} = 2(CF - f).r_R \tag{40}$$

So:

$$2(CF - f) = m_R . a \tag{41}$$

FWP: We know the inertial force in this system is:

$$F_{FWP} = m_{tot} . a \tag{42}$$

And from Equation 28:

$$CF - f = \frac{m_{tot}}{2} a \tag{43}$$

Where m_W is mass of the subject along with wheelchair. In order for equating inertial effect in both systems, from Equations 41 and 43 we write:

$$m_R = m_{tot} \tag{44}$$

Therefore, in order to have the same inertial effect in both EWP and FWP, we need to have the same mass for each roller as wheelchair+subject's mass.

Energy consumption: In FWP, wheelchair has displacement and absolute velocity, and therefore it has kinetic energy of:

$$E_{FWP} = \frac{1}{2} m_{tot} V^2 + 2 \times \frac{1}{2} I_W \, \omega_W^2 \tag{45}$$

The second term is due to rotation of the wheels. In EWP, displacement of wheelchair has been omitted and rotation of the rollers has been added. So the kinetic energy is:

$$E_{EWP} = 2 \times \frac{1}{2} I_R \omega_R^2 + 2 \times \frac{1}{2} I_W\ \omega_W^2 \tag{46}$$

Again, the first number 2 in this equation is because we have two rollers. For the first term:

$$2 \times \frac{1}{2} I_R \omega_R^2 = 2 \times \frac{1}{2} \times \frac{1}{2} m_R {r_R}^2 \times \omega_R^2 = \frac{1}{2} m_R V^2 \tag{47}$$

So:

$$E_{EWP} = \frac{1}{2} m_R V^2 + 2 \times \frac{1}{2} I_W\ \omega_W^2 \tag{48}$$

From Equations 45 and 48:

$$m_R = m_{tot} \tag{49}$$

Therefore, in order to have the same energy consumption in both EWP and FWP, we need to have the same mass for each roller as wheelchair+subject's mass. Fortunately, the condition of representation of inertial effect and energy consumption is the same.

5.4 Developing the Models

In this section, we are offering three approaches for making the representative model. These models were the only solutions we could find for this problem.

5.4.1 Equal-mass model

According Equations 44 and 49, for representing FWP in EWP, we need to have the same mass for each roller as wheelchair+subject's mass. The problem is that not all subjects and wheelchairs are of the same weight, so that we could make a roller with this given mass. What we can do about it, is considering a maximum weight of subject+wheelchair in the study, say 120 kg, and preparing an ergometer with the rollers of this weight. In this situation, all floor propulsions should be done with a compromising added mass in order to always have the maximum mass: M_{max}. Having set up this equipment along with asking the participant to try not to use swing trunk technique for propulsion, we have made a representative model of FWP on EWP.

5.4.2 Changeable-inner-diameter model

The former model has the limitation of "always having M_{Max} in all tests. Considering equations of "inertial effect" and "energy consumption" section of this paper, this M_{max} is inevitable, unless we could have a roller with changeable moment of inertia, and considering Equation 36, the only option we can count on for changing moment of inertia is the inner radius f rollers. With this introduction, the second proposed model is described in here.

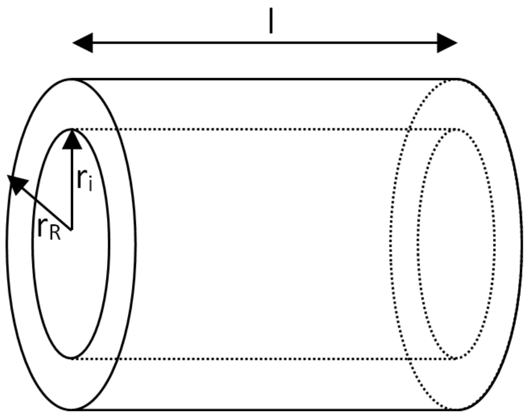


Figure 5-13 A hollow cylinder which is a schematic of a roller

Moment of inertia for a hollow cylinder is:

$$\boldsymbol{I = \int r^2 dm = \frac{\rho l \pi}{2}\left(r_R^4 - r_i^4\right)} \tag{50}$$

For having the same energy consumption:

$$\boldsymbol{E_{FWP} = E_{EWP}} \tag{51}$$

By equating Equations 45 and 46 :

$$\boldsymbol{\frac{1}{2} m_{tot} = \frac{I_R}{{r_R}^2}} \tag{52}$$

By substituting Equation 50 in Equation 52:

$$\boldsymbol{\frac{r_i^4}{r_R^4} = 1 - \frac{m_{tot}}{m_R}} \tag{53}$$

Condition of having the same energy consumption:

$$\frac{r_i^4}{r_R^4} + \frac{m_{tot}}{m_R} = 1 \tag{54}$$

For having the same Inertial effect:

$$\frac{T_{EWP}}{r_R} = F_{FWP} \tag{55}$$

From Equations 35 and 42:

$$\frac{2 \times I_R . \propto}{r_R} = m_{tot}\ a \tag{56}$$

From Equations 50 and 56:

$$\frac{r_i^4}{r_R^4} = 1 - \frac{m_{tot}}{m_R} \tag{57}$$

Condition of having the same inertial effect:

$$\frac{r_i^4}{r_R^4} + \frac{m_{tot}}{m_R} = 1 \tag{58}$$

Again, both having the same energy consumption and having the same inertial effect lead to a same equation (Equations 54 and 58). This equation could be used for building a roller of hollow cylinder type with changeable internal radius. This is an alternative to our first model, which implies doing all FWP propulsions with M_{max}. As this equation implies, we could have a semi-solid cylinder (r_i=0) for $m_{tot}=m_{max}=m_R$. However, for $m_{tot}< m_{max}$, the internal radius of hollow cylinder would be increased. A method for this type of roller is depicted at Figure 5-14. So, the roller could be built of accordion with two adjusting handles for adjusting the interior radius of roller.

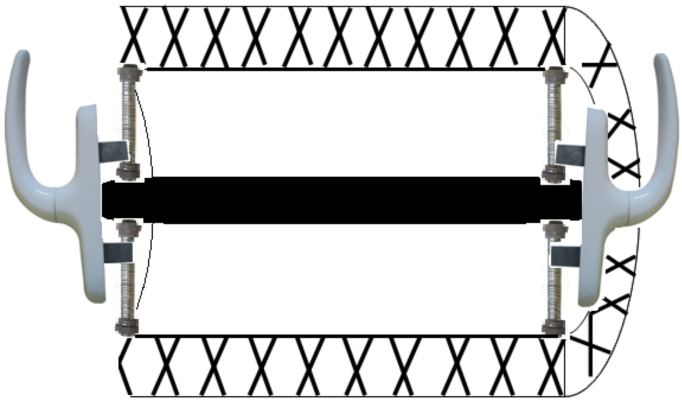


Figure 5-14. Proposed roller structure for Changeable-inner-diameter approach

5.4.3 Additive-disks model

In this model, we make use of some additive disks to adjust the rollers' moment of inertia corresponding to the weight of subject. By either equating Equations 45 and 46 (equating energy), or Equations 35 and 42 (equating inertial effect) we have:

$$\boldsymbol{m_{tot}} = \frac{\boldsymbol{2I_R}}{\boldsymbol{r_R}^2} \tag{59}$$

Where m_{tot} is mass of the participant and the wheelchair taken together. By having a given amount for the radius of the rollers, rollers' moment of inertia is directly related to mass of the subject. As Figure 5-15 shows, a number of disks that are all of the same shape and moment of inertia mount to the roller, in accordance to the mass of participant and the wheelchair together. Then the disks will be locked to the rollers by means of a clamp.

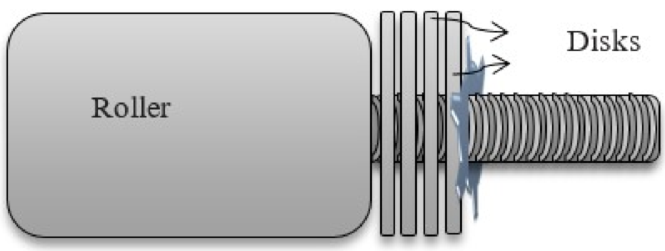


Figure 5-15 Proposed roller structure for additive-disks model

For example, an ergometer of this model, with steel rollers that are 20 cm in diameter and 22 cm in length, and have a screw shaft attached to lateral side of the rollers will be able to represent wheelchair propulsion of a 40-kg person sitting on a 15 kg wheelchair. However, for every 3.7 kg increment in the mass of participant and/or wheelchair, a steel disk of 20 cm diameter and 1.5 cm thickness should be mounted on each roller. The final length of each roller for a 120-kg participant would be 55 cm.

5.5 Conclusion

In this study, straight-line floor wheelchair propulsion was considered from basic biomechanical principles in order to suggest a model for a wheelchair ergometer that would be able to biomechanically represent straight-line floor wheelchair propulsion. On this basis, three models for rollers that should be used in the representative wheelchair ergometer were proposed. By using each one of these models for the rollers of a wheelchair ergometer that has the same structure as the ergometer structure assumed in this paper, we therefore put forth that wheelchair ergometer propulsion can represent straight-line floor wheelchair propulsion, only if the subject uses the same strategy of trunk swing in both conditions: using trunk swing or not.

With regard to ranking these models, first model -equal-mass approach- is the most straight forward, but less flexible. Also, in this model, people of lighter weights have to overcome the same weight of a wheelchair/rollers that heavier people do in the experiment, which may be a weakness of this model. Instead, the second approach is more flexible, but more difficult to be built. On the other hand, we claim that our third model is a model that can be accurate enough, flexible, and not difficult to build. In conclusion, we pose that the third model presented here is the best model for a wheelchair ergometer that can represent floor wheelchair propulsion on wheelchair ergometer.

The following chapter includes technical notes about development of three ergometer-based VR systems that **simulate wheelchair manoeuvring**. In addition to linear inertia, this chapter presents two methods for simulating **rotational inertia** that are employed in the development of two of the systems (VR_sysII and VR_sysIII).

The VR systems developed for this study are ergometer-based, developed for manual wheelchair propulsion, and do not include any motors in their design.

6 Development of Three Versions of a Wheelchair Ergometer for Curvilinear Manual Wheelchair Propulsion Using Virtual Reality

6.1 Abstract

Virtual Reality is an emerging technology that is being applied to everything from industry to education to entertainment and to research. Although very helpful in providing a safe and controlled environment for studying or training wheelchair users, until recently, wheelchair ergometers have a major disadvantage in only being capable of simulating straight line wheelchair propulsion. In order to simulate rotation, visual feedback is required for steering and avoiding obstacles. Virtual Reality has helped overcome this problem and broaden the usability of wheelchair ergometers. However, for a wheelchair ergometer to be validly used in

research studies, it needs to be able to simulate the biomechanics of real world wheelchair propulsion.

In this study, three versions of a wheelchair simulator were developed, to provide a sophisticated wheelchair ergometer in an immersive virtual reality environment. The virtual reality simulators were developed for manual wheelchair propulsion and all were able to simulate simple translational inertia. Each of the systems reported used a different approach to simulate wheelchair rotation and accommodate rotational inertial effects. The first system did not provide extra resistance against rotation and relies on merely linear inertia, hypothesizing that it can provide acceptable replication of biomechanics of wheelchair manoeuvres. The second and third systems, however, are designed to simulate rotational inertia: system 2 uses mechanical compensation and system 3 uses visual corrections based on the influence of rotational inertia on wheelchair movement to induce the perception of rotation. Details of designing and manufacturing of the three Virtual Reality systems are presented in this paper.

Keywords: Wheelchair ergometer, biomechanical replication, inertial compensation in transformation and rotation, virtual reality

6.2 Introduction

Wheelchair ergometers (since 1970 [189], [190]) are instruments that have the functionality of treadmills for wheelchair users. They usually comprise two rollers, although single roller ergometers are available too [95]. The rear wheels of the wheelchair are placed on the rollers and the wheelchair frame is secured to the ergometer. In several cases [95], [98]–[101] wheelchair ergometers have been used to facilitate wheelchair-related studies. They enable use of complex non-wearable physiological sensor systems and are intended to make these experiments controllable and repeatable.

Ergometers reported in the literature, however, are usually only capable of simulating straight-line wheelchair propulsion (SLP) [95], [98]–[100]. Available SLP ergometers lack robust replication of the different biomechanical factors involved in real-world wheelchair propulsion; usually, these factors are only partially included in their analyses [98], [99], [105], [106] and other factors such as energy and inertia are often ignored [51], [90], [106].

Nonetheless, there are some studies that have considered inertia and have tried to design ergometers in such way as to replicate it. In one study [110], removable flywheels were used to simulate translational inertia on the ergometer. In another study [190], a robot wheelchair propeller was designed that is able to measure biomechanical factors of propulsion, in addition to having the advantage of performing a given task in a measured and calculated way. The authors believe this will help in the future to find the effects of inertia and rolling resistance (RR), passively. Also, there is one recent study [99] that has developed a light portable ergometer capable of simulating linear inertia to some extent: it provides an average inertia for three intervals of subject weights. However, it includes several simplifications in the design.

Virtual Reality (VR) is a cutting-edge technology that is currently used widely, with different applications across many different areas of research. Rehabilitation of the wheelchair users is not an exception. There have been several efforts in recent years to take advantage of VR's capabilities to facilitate and accelerate rehabilitation [111] or training [112]–[116] of wheelchair users, in addition to broadening the knowledge available on wheelchair propulsion [111]. These efforts usually include simple VR systems, mainly using one or more monitors [101], [111], [112], [114]–[119] along with a joystick [111]–[116], [118]–[120] as the interface. This way, simulation of powered wheelchair propulsion is possible, but not manual wheelchair propulsion. There have been several positive outcomes reported in the literature for using VR in training/rehabilitation of power wheelchair users [111], [113]–[118], [120], but we could find only four cases of a VR environment developed for manual wheelchair users:

In one study [121], a system was developed to use the rotation of the wheelchair wheels to replace a joystick and estimate the location of the wheelchair in the VR space using a logarithmic scale for calibration. This study tried to include inertial navigation by actual movement of the wheelchair, however, there was no one-to-one replication of movement and no immersive VR. In another study [94], a motorized ergometer was designed for replicating curvilinear wheelchair propulsion that uses haptic admittance control to adjust wheelchair wheels velocity in accordance to the moment applied on them. Inertia effects are also applied by that control system.

AccesSim [101] is a VR-based wheelchair ergometer designed for testing accessibility of urban areas. It provides force feedback by accelerating the wheels or changing the RR based on what is simulated in the VR, e.g. slopes. This ergometer lacks inertial compensation though, and the VR is not immersive.

Toronto Rehabilitation Institute‘s Challenging Environment Assessment Laboratory (CEAL) is developing a wheelchair simulator [109] in an immersive VR environment to be used for manual wheelchair users. In this wheelchair simulator, the forces applied on the wheels are measured by the sensors embedded on the wheels and sent to the PC real-time where a MATLAB program runs a mathematical model to find the velocity and acceleration of the wheels as well as the heading. The wheelchair is placed on a turning table which rotates based on the heading calculated. This wheelchair simulator is still under development.

In this study, we have developed three versions of an ergometer-VR system that can simulate both SLP and wheelchair manoeuvring, considering all underlying biomechanical factors [11]: VR_sysI, VR_sysII, and VR_sysIII. These ergometer-VR systems are designed for manual wheelchair propulsion and are interfaced with an immersive VR environment. All three systems use a carefully designed translational inertia system, but take different approaches to simulate rotations: VR_sysI provides rotations with no rotational inertia compensation, VR_sysII provides mechanical compensation for the rotational inertia, and VR_sysIII perceptually compensates rotational inertia via the simulation software. These VR-ergometer systems provide a unique state of the technology setup for studying wheelchair propulsion, as well as having the potential for training Paralympics athletes and preparing newly injured wheelchair users for real-life wheelchair propulsion scenarios, before discharging them from the hospital.

6.3 Methods

In order to have a good biomechanical replication of Real World (RW) wheelchair propulsion on a wheelchair ergometer and in the virtual environment, we propose that there are at least 6 biomechanical factors that need to be checked to be similar in the two environments: RW and virtual world [11]. These factors are namely: velocity and acceleration, applied force, resistive force, trunk swing, inertial effects, and energy consumption. Based on these factors, we have developed three versions of a virtual environment for wheelchair users that are outlined below.

6.3.1 System I

This system is designed to replicate the biomechanics of SLP, considering the aforementioned six factors. The underlying calculations are explained with details elsewhere [11]. In that paper, we have proposed three models for a wheelchair ergometer based on the main underlying formula that we have shown is the necessary condition for having similar inertial effects and energy consumption in the RW and on the ergometer. Later, with the same factors in mind, we designed and built a fourth, better model for the ergometer. In this model, there are two solid cylinders as the base of the rollers. These rollers alone replicate the inertia of a person weighing about 38 Kg (more or less, depending on the weight of their wheelchair), which is the average weight of a 12 years old child [191]. To replicate weights greater than this, we designed four disk-shape masses for each roller, that are mounted on threaded bars (Figure 1). The threaded bars allow for adjusting the distance of the disks from their center of rotation, which in turn, makes the system adjustable for different participant plus wheelchair weights. Figure 6-1 shows the inertia system, adjusted for two different weights.

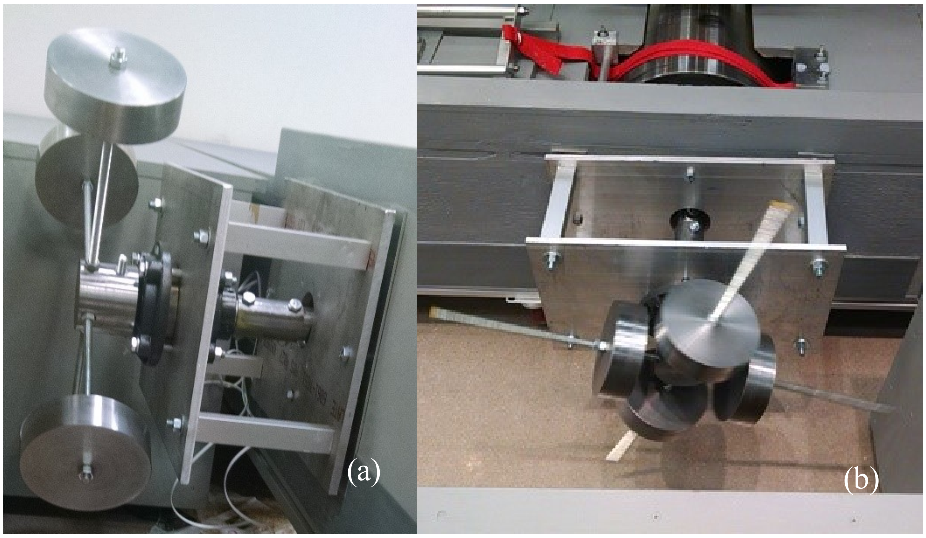


Figure 6-1 The inertia system developed for SLP. (a) and (b) show the inertia system adjusted to two different weights

Each disk is 5” in diameter, 1.5” thick, weighs 3.77kg (8.3 lb), and is made of AISI 1018 mild low-carbon steel. To design the inertia system and to find the specifications of the structure supporting these masses, a thorough analysis was completed, including bending stress, normal stress, shear stress, bearing stress, torsion, deflection, fatigue (the effect of dynamic loading), and vibration resonance. The Safety Factor (SF) used in each analysis was at least 14, where the minimum needed SF was 2.746, according to the following formula [192]:

$$\begin{aligned} Needed\ SF = {} & Material\ SF \times Stress\ SF \times Geometry\ SF \\ & \times Failure\ analysis\ SF \times Reliability\ SF \\ & = 1.1 \times 1.3 \times 1 \times 1.2 \times 1.6 = 2.746 \end{aligned} \tag{60}$$

Values for each factor is derived based on the tables and instructions noted in reference [192]:

$Material\ SF = 1.1$ as we used the manufacturer values for material properties,

$Stress\ SF = 1.3$ as the method used may result in errors less than 50%,

$Geometry\ SF = 1$ as manufacturing tolerances are average,

$Failure\ analysis\ SF = 1.2$ as we used slight simplifications in the theoretical principals, and

$Reliability\ SF = 1.6$ as we need the parts to be very reliable.

One important factor in simulating SLP is providing the same RR as real world's in the simulator. To do so, the first step was to ensure the friction of the system was as low as propulsion on linoleum. For simulating higher rolling resistance surfaces, a pneumatic belt system was used on each roller.

To reduce the system's friction as low as linoleum's, after trying different brands and models, we finally used Timken, model number: 1108KLLB (wide inner ring ball bearings, with a single row in a deep groove). Using these bearings, we were able to simulate RR similar to that when propelling on linoleum (Table 6-1).

The method used to find the RR deceleration in the RW and on the ergometer was based on a coast down test and using the following formula:

$$a_{RR} = Vmax/\Delta t \tag{61}$$

Where a_{RR} is deceleration caused by the RR, $Vmax$ is the velocity of the wheelchair when starting to coast down, and Δt is the time needed to reach to full stop.

To test the RR, six subjects were recruited weighing 53 Kg to 138 Kg, and each performed the coast down test five times in the RW and five times on the ergometer. Participants were instructed to keep their trunk still at all times. Table 6-1 shows the average RR deceleration results from this test.

Table 6-1 The results of the rolling resistance tests

		Rolling resistance deceleration (m/s^2)
Ergometer	Left	-0.20
	Right	-0.17
Floor	Left	-0.25
	Right	-0.12

The two inertia systems were mounted on two shaft extensions that were connected to the rollers using universal joints and bedding on the aforementioned bearings. Some extra technical figures regarding VR_sysI are provided in Appendix A.

For a proper simulation of wheelchair propulsion, having visual feedback is essential; it is very beneficial for simulating SLP and crucial when wheelchair manoeuvring is being studied. The reason lays in a key difference between walking and propelling a wheelchair: the relative strength of the legs does not significantly affect the ability of the person to walk in a straight line without visual feedback. However, for wheelchair propulsion the relative strength of the arms substantially affects any attempts to achieve SLP without visual feedback. Since the dominant hand of the wheelchair users is stronger, there will be a difference in the acceleration of the two wheels which results in rotation of the wheelchair. If the users do not have visual feedback, they will not be aware of this change in the direction to control it. On the other hand, as walking is more of a mechanical work and people have a feeling about the size of their strides, going straight without visual feedback is not much affected by the relative strength of the legs.

There were different approaches to provide the visual feedback for the ergometer. It could be provided through monitors, VR goggles, augmented reality, or VR cube. Monitors were not used, as they do not provide immersive VR, while immersion is shown [9] to help with feeling “present” in the VR. Augmented reality was also not a good option as participants were going to perform an intense manoeuvring task on an stationary ergometer. Seeing a large amount of stationary background while seeing oneself manoeuvring around the cones is extremely nauseating, as it sends mismatching signals to the brain about the movement of the body. On the other hand, both VR goggles and VR cubes provide immersive VR. Between these two, a VR goggle has the advantage of being much cheaper and requiring much less administrative work. However, a VR cube provides advantages that well illustrates the superiority of it for this study. First of all, they cause less motion sickness than VR goggles. Secondly, the person can see their body and their wheelchair in the VR cube and thus it can further help with better sense of presence. Also, if the participant prefers or to alleviate motion sickness, a VR cube has the option of disabling the track of head movements, or even providing-two dimensional images. Therefore, to provide visual feedback on the wheelchair ergometer, the ergometer was placed in the EON Icube™ Mobile which is an immersive VR system and is well-suited to perform different tasks, including wheelchair propulsion.

By adding the visual feedback, this system is able to replicate SLP; however, it can also be used for simulating manoeuvres by the simplification that rotational inertia is negligible. This is the basic idea of the first system (VR_sysI) in this paper.

To associate the visual feedback with the way the wheelchair user is propelling the wheelchair in VR_sysI, the formulas mentioned in our previous work [11] were used to find the position and rotation of the wheelchair in the VR environment based on the velocities of the wheels. Since the wheelchair has a planar movement and is a rigid body, describing two points determines the motion of the whole system. So, position or velocity of the rear wheels determine the actual rotation of the wheelchair. The ergometer wheels are connected to digital rotary encoders (quadrature encoders. Bourns Inc., California, USA. Model Number: EMS22Q51) which are read and processed by a Teensy (32 Bit Arduino-compatible microcontroller. PJRC LLC, Oregon, USA. Model: Teensy 3.2) which sends the data to a server over a USB connection. The interface program (written in C# 2015, see Attachment D) running on the server then takes that data, further processes it, logs it, sends it to LabView for further processing, and sends it to the EON server for visualization (Figure 6-2).

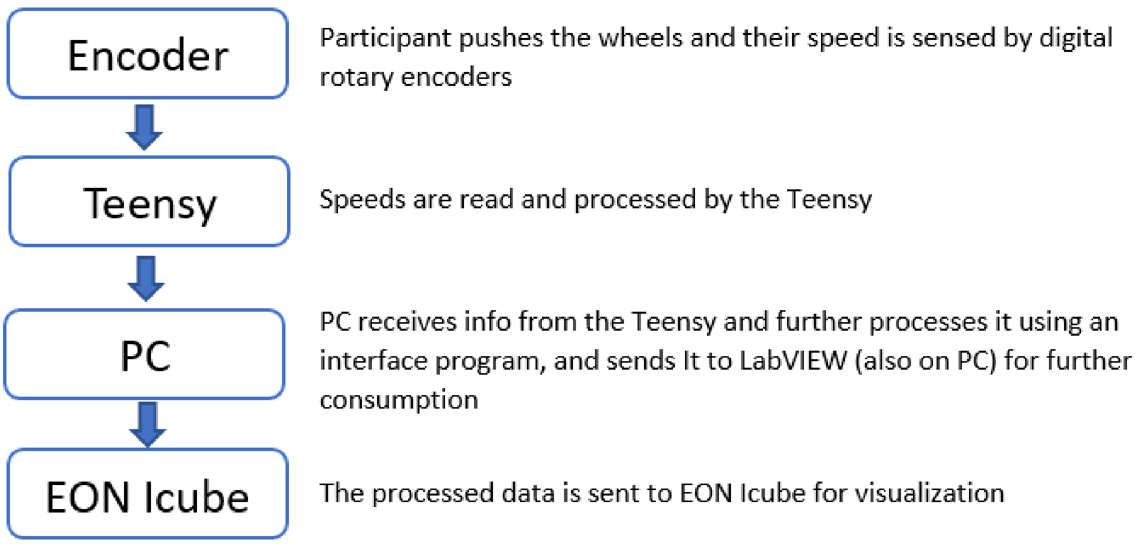


Figure 6-2 Block diagram of the VR systems

After finishing the setup, a thorough calibration of the system was performed, including the calibration of:

- The Icube screens,
- Icube grid alignments,
- Ergometer speed encoders,
- Calibration of the system for people with different weights,
- SMART^{Wheel}s,
- The Icube motion analysis system (Vicon)

6.3.2 System II

The ease or difficulty of turning is dependent on the moment of inertia of the wheelchair-user system, and is experienced as the resistance to turning. The design of the second system was based on finding the effect of inertia in the rotation and designing a mechanical system that compensates that in the VR_sysII.

From an inertia standpoint, VR_sysI is suitable for SLP: both hands are supposed to apply the same force for SLP, so both wheels would have the same velocities. However, when the right and left forces differ, the wheelchair will turn. Although the visual feedback will show the person is turning, since there is no rotational resistance, the wheelchair will appear to turn more easily than what would happen with the same amount of left and right forces in the RW. The inertia system implemented for SLP does not help with the (whole) inertial force in turning due to a difference in the forces on the two wheels and the independent rotation of the wheels on the wheelchair ergometer.

In the RW the wheels are connected through the seat. If only one wheel rotates, it will make the other wheel turn too, which is not the case on the wheelchair ergometer. On the latter, the two rollers are completely independent, which makes the rotation of the left and the right wheels of the wheelchair totally independent. Thus, the rotation of one wheel does not affect the other.

On the ergometer, the inertia in SLP is compensated, but not when the forces of the left and right wheels differ, which in turn makes the velocity of the two wheels different and the turning occurs (θ). This means that there will be a center of rotation of the wheelchair user system and a radius of rotation (R). The following relationship formulates this:

$$\begin{aligned} M_{applied} - M_{resistive} &= 2 \times \bar{F} \times R + \Delta F \times \frac{W}{2} = I\ddot{\theta} \\ &= (m \times R^2 + \bar{I}) \times \ddot{\theta} = m \times R^2 \times \ddot{\theta} + \bar{I} \times \ddot{\theta} \end{aligned} \tag{62}$$

Where:

$M_{applied}$= The moment applied by the wheelchair user on the wheelchair, around wheels' center

$M_{resistive}$ = Moment of the sum of all the resistances against rotation of the wheelchair

$\bar{F}$= The average of left and right forces

ΔF= Difference between left and right force

R= Radius of rotation

I= Moment of inertia of the wheelchair and participant (together), around the center of rotation

$\bar{I}$= Moment of inertia of the wheelchair and participant (together), around their center of mass

$\ddot{\theta}$= Rotational acceleration of the wheelchair

W= Wheelchair width (wheel span)

m= Wheelchair and participant's mass, together

The linear inertial force multiplied by the radius of rotation gives the rotational inertia (moment) which is equal to $2 \times \bar{F} \times R$. The following formula can be obtained from dividing the former formula into two main parts.

$$2 \times \bar{F} \times R - M_{resistive_1} = m \times R^2 \times \ddot{\theta} = ma \times R \tag{63}$$

Therefore:

$$F_L + F_R - F_{resistive} = ma \tag{64}$$

Where a is the linear acceleration and $F_{resistive}$ is the sum of all the resistive forces against translation of the wheelchair. So far this is accounted for using our linear inertia system. Now let us assume that the left and the right forces are different but $\bar{F} = 0$, which means $F_R = -F_L$. In the RW this makes a force couple on the wheelchair and therefore it will start spinning in the spot. There is, meanwhile, a resistance that the wheelchair should overcome: the rotational inertia. This can be obtained from the following formula, which is derived from Formula 62 .

$$(F_R - F_L) \times \frac{W}{2} - M_{resistive_2} = \bar{I}\ddot{\theta} \tag{65}$$

In other words, we can break down any wheelchair propulsion to two parts: the SLP part and the non-straight-line propulsion part. $2 \times \bar{F}$ produces the SLP and ΔF produces the non-straight-line part by determining the heading (Figure 6-3).

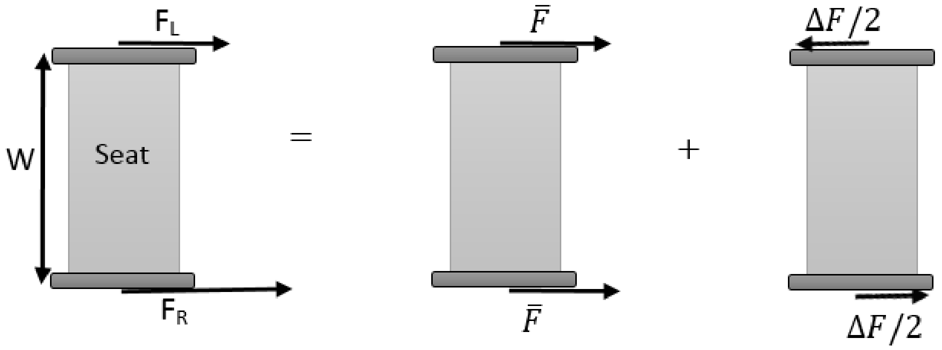


Figure 6-3 Break down of forces applied on the wheelchair wheels to the part that provides SLP (the first term) and the part that determines the heading (the second term)

Due to absence of rotational inertia, turning in VR is as $\bar{I}\ddot{\theta}$ easier in VR_sysI. In other words, $M_{resistive}$ should be as $\bar{I}\ddot{\theta}$ greater when turning, to compensate the absence of the rotational inertia. As shown in Figure 6-4, the missing $\bar{I}\ddot{\theta}$ is around the z-axis, and it is a resistance against the moment created by $\Delta F = F_R - F_L$. These forces also (and in the first place) create moments around y-axis on the wheels. To resemble the rotational inertia which is a moment in z-axis we can add/decrease the RR against rotation of the wheels, in real time. In other words, since no way could be found to adjust and build a real inertia system to replicate the RW rotational inertia without influencing the already developed linear inertia ergometer system, the extra force increment/decrement should be spent on some extra resistance against rotation of the wheels that can be controlled real time. In System II the extra RR that is calculated real time, will be applied on one of the wheels at a time, which will be determined using Labview code to control feedback.

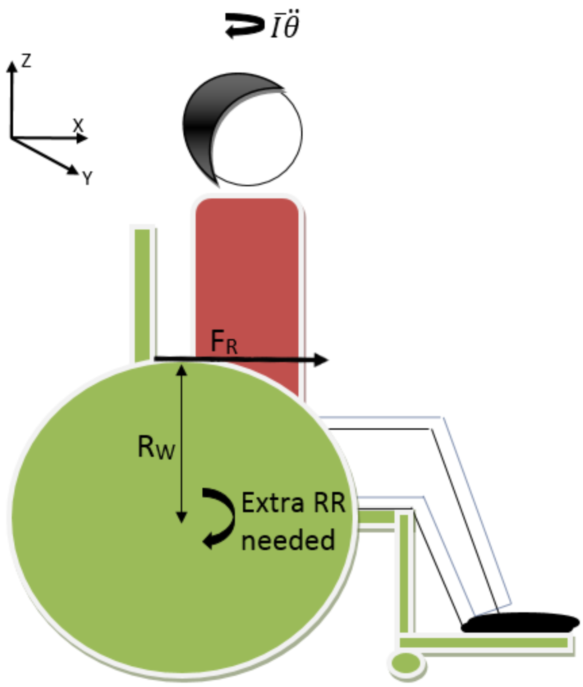


Figure 6-4 Schematic of some of the forces and moments on a wheelchair when turning

The force applied on the wheels should be $\Delta F = \frac{2\bar{I}\ddot{\theta}}{W}$ bigger to replicate the RW where rotational inertia is present. The amount of the extra RR (resistive moment) needed can be calculated from the following formula:

$$RR_{extra} = \Delta F \times R_W = \frac{2\bar{I}\ddot{\theta}}{W} \times R_W \tag{66}$$

Where

$\bar{I}$= Moment of inertia of the wheelchair and participant (together), around their center of mass ($Kg.m^2$)

$\ddot{\theta}$= Rotational acceleration of the wheelchair (rad/s^2)

W= Wheelchair width (meter)

R_W= Wheels' radius (meter)

Also, it can be shown that:

$$\ddot{\theta} = \frac{a_R - a_L}{W} \tag{67}$$

Where a_R and a_L are the linear acceleration of the right and the left wheel, respectively. Therefore, the Formula 66 can be written as:

$$RR_{extra} = \frac{2\bar{I}R_W}{W^2} \times (a_R - a_L) = \frac{2 \times \bar{I} \times R_W^2}{W^2} \times (\alpha_R - \alpha_L) \tag{68}$$

Where α_R and α_L are the rotational acceleration of the right and the left wheel, respectively. In this formula R_W and W are predetermined and acceleration of the wheels are obtained from the speed encoders that are in contact with the rollers. However, we need to find the moment of inertia of the wheelchair and the wheelchair user.

Finding the wheelchair and wheelchair user's moment of inertia

A novel approach for calculating the moment of inertia was developed for this study. 10 subjects were recruited (weights ranged from 53 to 137 Kg) and were asked to perform 6 bouts of a simple spin-on-the-spot test, 3 to the right side and 3 to the left side, by pushing one wheel forward and pulling the other one backward. The wheelchair was equipped with two SMART$^{\text{Wheel}}$s which gave us the moment applied on each wheel as well as the velocities. Then, using the following formulation, the moment of inertia of each subject on the wheelchair was obtained.

$$M - RR = \bar{I}\ddot{\theta}_1 \tag{69}$$

$$0 - RR = \bar{I}\ddot{\theta}_2 \tag{70}$$

$$M = \bar{I}(\ddot{\theta}_1 - \ddot{\theta}_2) \tag{71}$$

Where θ can be obtained from Formula 67 and:

$$M = \frac{Mz_L + Mz_R}{Rw} \times \frac{W}{2} \tag{72}$$

M is the equivalent moment applied on the wheelchair, around the z-axis,

RR is the rolling resistance of spinning the wheelchair on the floor (linoleum),

$\bar{I}$ is the moment of inertia of the wheelchair and the wheelchair user about their center of mass,

$\ddot{\theta}_1$ is the angular acceleration of the wheelchair when turning in the push/pull phase,

$\ddot{\theta}_2$ is the angular deceleration of the wheelchair when turning in the coast-down phase,

W= Wheelchair width

R_W= Radius of wheels

Mz= Moment applied on the wheels by the subject in the push/pull phase,

We also found the moment of inertia of the wheelchair solely (I=1.76 Kg.m^2), by running the same test, without having anybody sitting on the wheelchair. The research collaborator applied the moments on the wheels and then the wheelchair was left to spin on its own. The Table 6-2 shows the results. For each participant, the height, weight, and shoulder width was recorded. The linear regression of few different functions of these factors was then found (Figure 6-5) which showed the highest R^2 when person's weight was multiplied by their height. Therefore, Formula 73 obtained using curve-fitting was used to find the moment of inertia of each person for the interface used to control System II:

$$I = 0.375 \times m \times h \tag{73}$$

Where m is the participant's mass and h is the participant's weight.

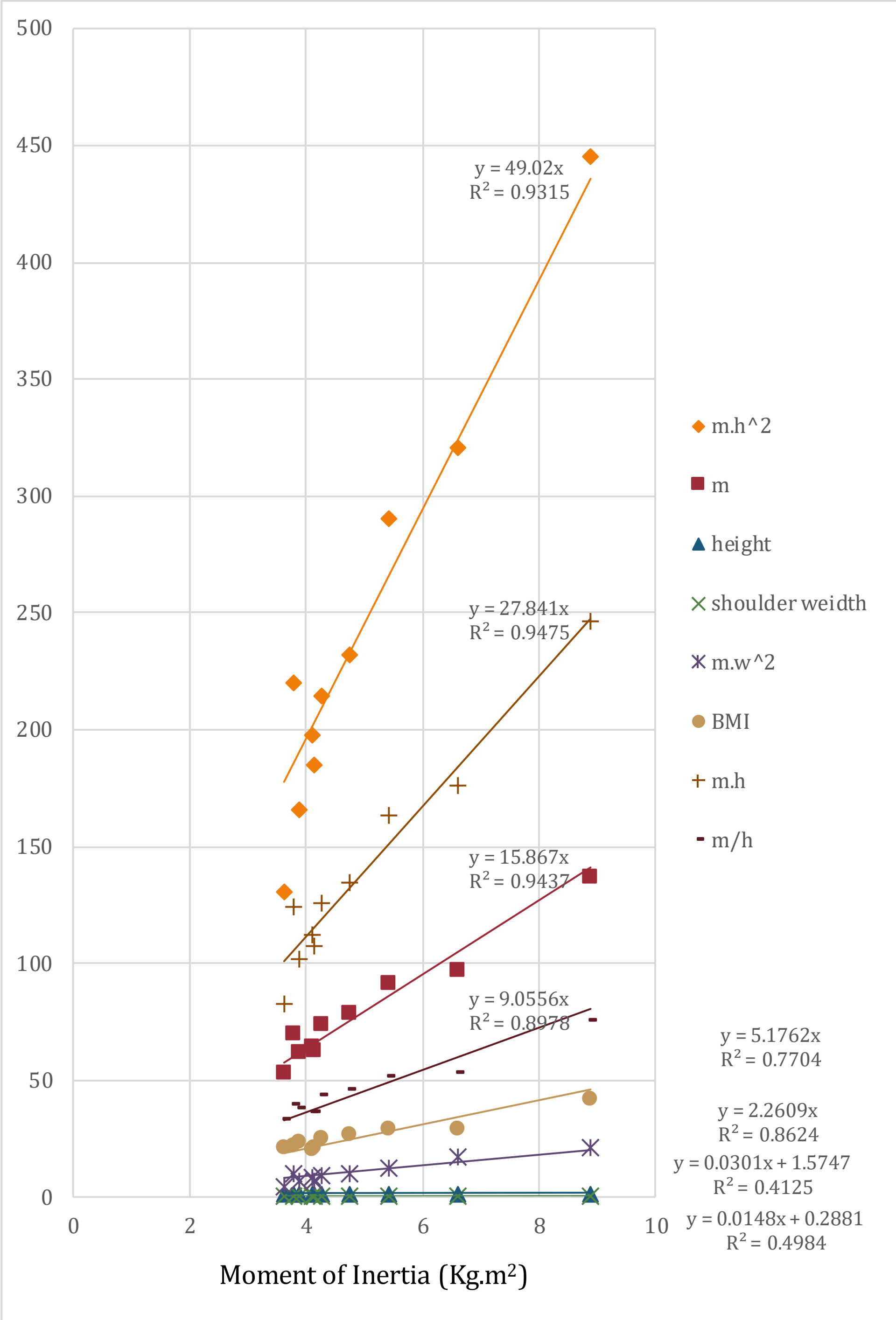


Figure 6-5 Finding the best predicting function for moment of inertia of the wheelchair user while seated in a wheelchair

Table 6-2 Data and results of tests of moment of inertia

Participants' weight (Kg)	Participants' height (m)	Participants' shoulder width (m)	Participants' moment of inertia ($Kg.m^2$) - averaged	Moment of inertia of participant and wheelchair ($Kg.m^2$)- averaged
53	1.57	0.308	3.62	5.38
62.5	1.63	0.328	3.90	5.66
62.9	1.71	0.332	4.13	6.16
64.4	1.75	0.367	4.11	5.87
70.2	1.77	0.375	3.77	5.53
74.2	1.7	0.35	4.28	6.04
79	1.71	0.36	4.74	6.50
92	1.778	0.37	5.43	7.19
97	1.82	0.428	6.61	8.37
137	1.8	0.395	8.89	10.65

Development of system II

Here, we will first explain the main procedure. The devices and the other details are discussed afterward.

As discussed before, the VR_sysII was designed to compensate the rotational inertia in real time. Since we were unable to simulate a pure inertia system, we decided to add or remove rolling resistance independently for each wheel when turning. This was done using the pneumatic braking system. Here is how this system works: First, the velocities of the wheels and also a force feedback from the pistons of the braking system is sent to the PC through the Teensy microcontroller and a Data Acquisition (DAQ) Module, respectively (See Figure 6-6). Then, the RR needed to be added corresponding to the participant's turning behaviour is calculated using velocity and force feedback and the Formula 68, through a Labview program written by the authors. The voltage for the required pressures, corresponding to the calculated RR, is then sent to the pistons of the brake system through compact Rio (cRio). By pulling the pistons (as calculated), the strap attached on the pistons is also pulled. This creates friction on one of the rollers in real time, in response to the turning that happens in the VR by the wheelchair user. The wheel on which the pressure is applied is also determined using the Labview program, based on the side the wheelchair user is turning to, and also the relative acceleration of the wheels.

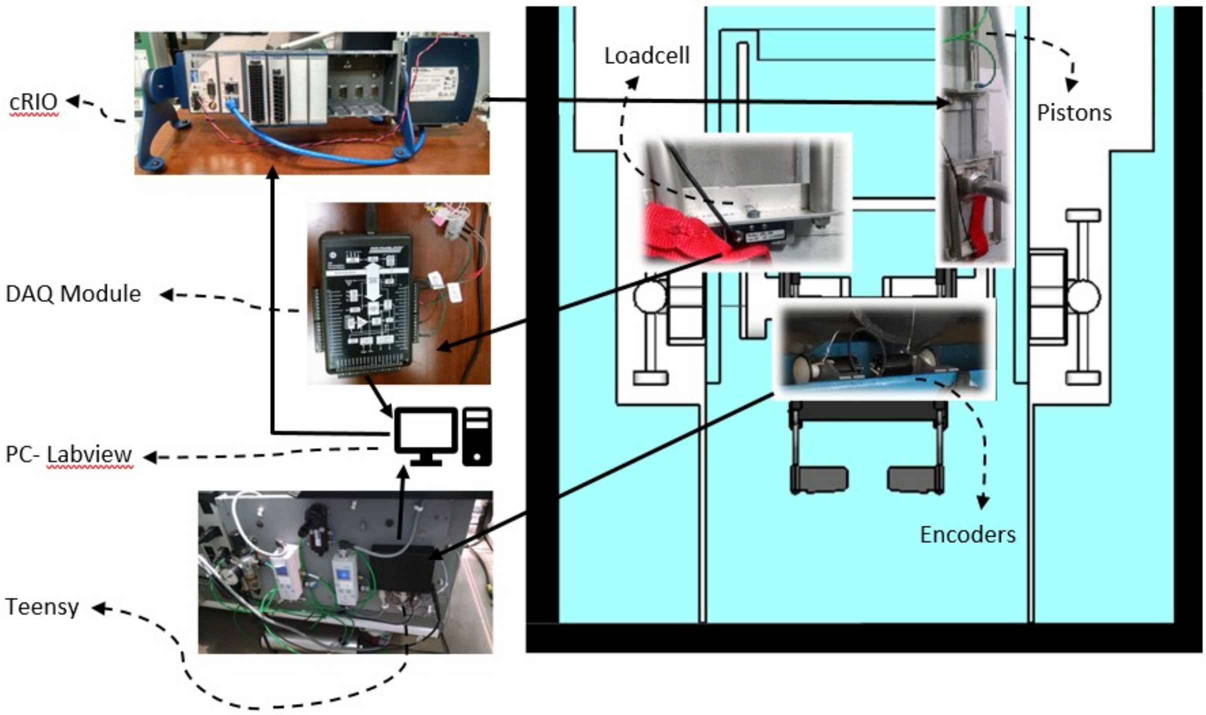


Figure 6-6 Schematic of VR_sysII

To figure out the relationship between the required RR, the force sensed by the load cells, and the voltage to be sent to the pistons (i.e. the level of activation of the pistons), a series of calibration tests were undertaken recruiting 6 participants (weighing 53 Kg to 138 Kg). Each participant performed 20 coast-down tests on the ergometer, each with a different percentage of activation of the pistons: from 0 to 100% activation with 5% increments. Each of these activation levels was associated with a different voltage and dictated a different RR and force reading. Then, the relationship between the piston activation levels and RR was obtained for each weight. Finally, in the Labview program, the required voltage for producing the required RR was found by interpolating between those relations.

The details for the components in VR_sysII are shown in Figure 6-6and are:

DAQ Module: Data translation inc., Massachusetts, USA. USB data acquisition function module, Model Number: DT9802

Pistons: SMC, China. Model Number: CD85N25-125-B, 1.0 MPa.

Labview: NI, Texas, USA. Labview 2012, Version 12.0f3n (32-Bit).

cRIO: NI, Texas, USA. Model NI PS-15. Made in Czech Republic.

Loadcells: Interface inc. Arizona, USA. Model No: SML-50, Capacity: 50 lb, Serial Number: 230805. Loadcells were calibrated for being used in this system using a graded bottle, filled with water from 0 ml to 1000ml, with 50 ml increments.

Teensy: 32 Bit Arduino-compatible microcontroller. PJRC LLC, Oregon, USA. Model: Teensy 3.2

Encoders: Quadrature encoder. Bourns Inc., California, USA. Model Number: EMS22Q51

One main problem that we faced when developing VR_sysII was in regard to the built-up hysteresis in the pistons when trying to change the RR real-time. Many different ways were tried to find a relationship for this built-up hysteresis with no luck. The only way that the pressure-force graph behaved consistently was when the pistons were activated from a 0 to different target pressures (Figure 6-7).

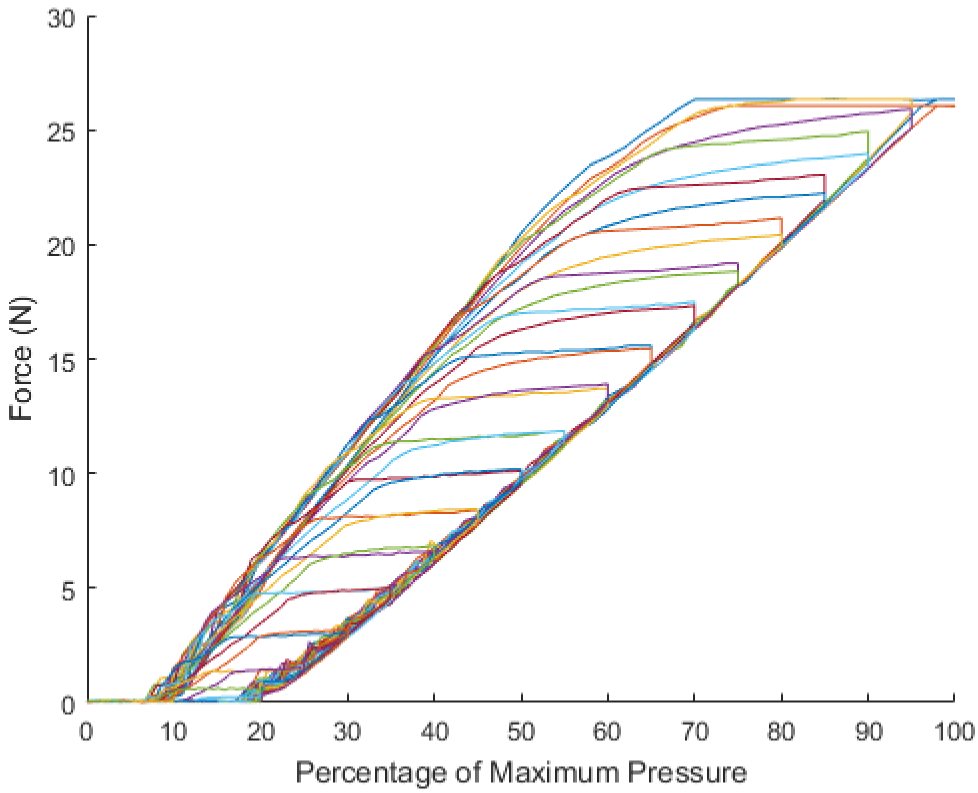


Figure 6-7 Graph of force to pistons pressures

Thus, dealing with hysteresis necessitated resetting the pneumatic system, before applying a new target pressure level to the pistons. In other words, for each calculated RR, the piston's pressure needed to go to 0 first and then increase to the target pressure. Since the pneumatic system is a mechanical system and requires a relatively large amount of time, it limited the response of the system to about 10 Hz and prevented us from having a continuous smooth system.

6.3.3 System III

VR_sysIII is based on perceptually compensating the moment of inertia, in the way that the visual feedback (the simulation) slows down at turns, proportional to RW turns. To do so, first, we need to look at the kinetics of the system. For each wheel we can write:

$$F_R \times R_W - RR_R = \bar{I}_W \times \alpha_R + \bar{I}_R \times \alpha_R \tag{74}$$

$$F_L \times R_W - RR_L = \bar{I}_W \times \alpha_L + \bar{I}_R \times \alpha_L \tag{75}$$

Where $\bar{I}_R$ is the moment of inertia of the rollers around their center line. Therefore:

$$\Delta F \times R_W - RR_R + RR_L = (\bar{I}_W + \bar{I}_R) \times (\alpha_R - \alpha_L) \tag{76}$$

By assuming $RR_R = RR_L$

$$\Delta F \times R_W = (\bar{I}_W + \bar{I}_R) \times (\alpha_R - \alpha_L) \tag{77}$$

The idea for the VR_sysIII was to find how much rotation (degrees of angle) would happen in RW if the same forces were applied on the wheels in there. Afterwards, the angle of rotation in VR would be corrected, accordingly. Formula 66 should be used to calculate the extra RR needed for compensating rotational inertia, in order to have the same ΔF as in RW. Let us assign prime mark ($'$) to variables belonging to the situation where rotational inertia is taken into account. According to Formula 66:

$$\Delta F \times R_W = RR_{extra} = \frac{2 \times \bar{I} \times R_W^2}{W^2} \times (\alpha_R' - \alpha_L') \tag{78}$$

From Formula 77 and 78:

$$\frac{2 \times \bar{I} \times R_W^2}{W^2} \times \Delta\alpha' = (\bar{I}_W + \bar{I}_R) \times \Delta\alpha \tag{79}$$

Now let us define:

$$\mathrm{C} = \frac{\Delta\alpha'}{\Delta\alpha} \tag{80}$$

Therefore:

$$C = \frac{I_{Wheel} + I_{Roller}}{I_{\mathrm{Wheelchair+Person}}} \times \frac{\left(\frac{Wheel\ Span}{Wheel\ Radius}\right)^2}{2} \tag{81}$$

The rotational inertia is defined by Δa while rotation is defined by ΔV (rotation will happen if ΔV is detected). Thus, we need to find the corrected velocities too (V_L' and V_R'). The following formula shows that the same C can be defined for $\frac{\Delta V'}{\Delta \mathrm{V}}$:

$$\Delta V' = \int \Delta a' = \int \mathrm{C} \times \Delta a = C \int \Delta a = C \times \Delta V \tag{82}$$

It can also be shown that:

$$\frac{V'_L}{V_L} = \frac{V'_R}{V_R} = C \tag{83}$$

Now, the final formulation for controlling System III can be obtained as:

$$V'_{(L|R)} = V_{(L|R)} \times (C + (1 - C) \times C') \tag{84}$$

$$\text{If} \quad V_L \times V_R > 0 \quad \text{Then} \quad C' = \frac{1}{2}\left(1 + \frac{Min(|V_L|,|V_R|)}{Max(|V_L|,|V_R|)}\right) \tag{85}$$

$$\text{If} \quad V_L \times V_R < 0 \quad \text{Then} \quad C' = \frac{1}{2}\left(1 - \frac{Min(|V_L|,|V_R|)}{Max(|V_L|,|V_R|)}\right) \tag{86}$$

For providing both corrected translation (proceeding) and corrected rotation (heading), we need the corrected velocity of the wheels, accordingly. The above relations are developed in a way to apply no changes to the velocities at pure translation and apply the maximum correction to the velocities at pure rotation; for the remaining mixed situations, only a fraction of the C’ is applied which causes the corrected velocities to be generated in a linear slope from $C.V$ to V.

Pure rotation: $C' = 0 \quad V' = C.V$ (87)

Pure translation: $C' = 1 \quad V' = V$

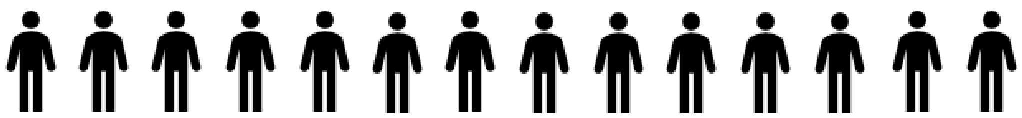
6.4 Conclusion

In this study, three versions of a wheelchair simulator (sophisticated wheelchair ergometers in the VR) were developed, each taking a different approach in simulating rotation. VR_sysI only accounted for linear inertia, VR_sysI provided a mechanical compensation for rotational inertia and VR_sysIII provide a perceptual simulation of rotational inertia. The VR_sysI has the advantage of providing a smooth and one to one visual feedback, but the disadvantage of turning too easily. VR_sysII has the advantage of providing close-to-real-life difficulty of turning and a one to one visual feedback, but a disadvantage of non-smooth visual feedback. Finally, VR_sysIII provides a smooth visual feedback and a close-to-real-life difficulty of turning, but not a one to one visual feedback. Experiments are warranted to realize which system is eventually the most popular by the users or which system is the best in providing closest to real world conditions.

After developing the VR systems, the reliability and validity of them needed to be determined. But first the reliability of the test which is the basis of these experiments need to be shown. The following chapter focuses on assessing the reliability of the Illinois agility test when used for wheelchair users.

In total, 14 abled-bodied participants were recruited for the studies presented in Chapters 7, 8, and 9. Participants underwent a maximum of 4 preconditioning sessions to get used to navigating in the VR. Then they participated in up to 3 main sessions of the experiment. The first row in the below table shows the session numbers; the main sessions are named as 1 to 3 and considering that participants had 1 to 4 preconditioning sessions, theses sessions were named in a way to keep the successive sequence of the numbers, from -3 to 0. The Roman numerals in the table show the system number. It is worth noting that participants completed the Illinois Agility Test both in VR and RW for every session.

Since the research presented in this thesis took a developmental approach, participant recruitment was started after development of system I, and systems II and III were developed and considered the feedback received from the participants. Therefore, participants underwent slightly different experimental protocols. In order to maintain a sound statistical approach, a portion of data was used for every study. The following table summarizes the experiments and highlights the part that was used in the following chapter.



-3	I	I			II				II			II		
-2	I	I		II	II			II	II			II	II	II
-1	I	I		II	II	II		II	II	II	II	II	II	II
0	I	I	I	II	II	II	II	II	II	II	II	II	II	II
1	I	I	I	II	II	II	II	II	II	I,II	I,II	I,II	III	III
2				II	II	II	II	I,II	I,II	I,II	I,II	I,II	III	III
3				III	III	III	III	III	III	III	III			

7 Investigating the test-retest reliability of Illinois Agility Test for wheelchair users

7.1 Abstract

The Illinois Agility Test (IAT) is a standard agility course used to assess and train able-bodied athletes [193], [194] as well as wheelchair-sport athletes [195]. It has been shown to be a reliable and valid tool to assess able-bodied population [193], [196], but the reliability of this test for assessing wheelchair propulsion has never been shown. The purpose of this study is to investigate the test-retest reliability of IAT to assess wheelchair propulsion.

In this paper, the test-retest reliability of using IAT for wheelchair users is found for peak and average velocity, acceleration, tangential and total force of the push, each for the left and the right wheel. Each of these variables was found using a custom program code written in MATLAB (MATLAB Student TAH Campus License, University of Alberta) for thirty-two decisive points throughout the IAT path. The Intra-class Correlation Coefficient (ICC) was found to be very strong for all these variables except for the average total force for the right side. The average ICC of variables was 89%. Also, the average 95% confidence interval was [44% 96%].

In addition, thirty-seven other significant propulsion parameters were found for IAT for each person which are clinically important, such as the number of pushes participants take to go around cones on the right relative to turning around the cones on the left. The average values of these parameters are reported in this paper; Also, all thirty-seven variables were compared between the two sessions using four separate MANOVAs; the results showed no significant difference between IAT performed in the two sessions which were at least one week apart. This, in turn, confirms the reliability of IAT for wheelchair users.

These results are sufficient evidence to show that IAT is a reliable tool to test wheelchair agility for fifteen variables tested for non-wheelchair users. Since experienced wheelchair users are much more consistent in wheelchair propulsion compared to non-wheelchair-users, the results of this study show that IAT can be used as a reliable tool to assess and train wheelchair users, both for clinical and athletic applications.

Keywords: Wheelchair, Reliability, Illinois agility test, ICC, SEM, MANOVA, wheelchair propulsion parameters

7.2 Introduction

Wheelchair propulsion requires intense effort that wheelchair users have to undergo continuously every day. This is firstly because the efficiency of wheelchair propulsion is almost half the efficiency of walking (~14% [197] vs 26 to 27% [198]), and secondly because wheelchair users use their arms instead of their legs to ambulate which are not evolved for that purpose. This is one of the reasons that makes the shoulder joint vulnerable when one uses wheelchairs as a primary method of ambulation. In fact, about 20% [17] of wheelchair users develop secondary injuries in their upper extremity around five years after starting to use a wheelchair. This percentage increases to 46% [17], [24] about 20 years after the injury. This mainly happens because of the excessive forces that wheelchair propulsion exerts on structures around the shoulder while propelling their wheelchair over rough terrains and also manoeuvring.

Manoeuvring is performed extensively for indoor wheelchair use [199] and is responsible for a greater portion of the excessive loading on upper extremities [200] and yet it has seldom been studied in the literature. Having standard and reliable tests for measuring wheelchair manoeuvres will greatly help to develop a more sophisticated understanding of the biomechanics of manoeuvring.

The Illinois Agility Test (IAT) is a standard agility test that has been used for both training and assessment of able-bodied athletes for many years [193], [194]. It was introduced for measuring multidirectional agility for different sports [193], [196]. For instance, IAT has been used to test the effect of strengthening exercises and also to test athletes' agility in soccer [201]. However, some researchers have used IAT for wheelchair athletes as well. William [195] has tried to find physiologic determinants to assess the training of wheelchair basketball players [202] and Usma-Alvarez, *et al.* have considered developing some outcomes for IAT that are suitable in assessing wheelchair rugby.

We could not find any publications that have shown the reliability of an agility test for wheelchair users, although there are a few papers that have investigated the reliability of different agility tests for the able-bodied population. Namely, Within-day reliability of IAT for 66 able-bodied semi-professional rugby players has been shown to be ICC=86% [203]. In another study, the reliability of IAT was found on a sample of 97 able-bodied young military service men: ICC=99% inter-rater reliability and ICC=68% test-retest reliability [193]. Also, an ICC of 96% was reported elsewhere for IAT for a sample of 89 able-bodied sportsmen from football, handball, and rugby players when repeating the test on different days [196].

Williams [202] has investigated the construct validity of three different agility tests, including IAT, for athlete wheelchair basketball players. They compared the timed performance of a group of elite to a group of competitive players, and concluded the test was valid as there was a significant difference between time to finish the test. They also found excellent test-retest reliability with correlations above 0.9 between 2 trials performed in one session. However, they only measured time of finishing the test and not other important biomechanical factors. This is important because when using an assessment tool for measuring some outcome we need to be confident that this tool will create similar results if used some time later [204].

One important factor when testing agility is how fast the participant could finish the agility path. This is the reason the primary output of all agility tests is the time needed to finish the task. However, a good agility test should be designed in a way to measure the acceleration, as the changes in the direction which are the main parts of agility tests are more correlated with acceleration than speed [205]. Although completion time is not a great concern when studying wheelchair manoeuvres, the latter characteristic makes agility tests suitable for the purpose of this study, as it is closely related to manoeuvrability for wheelchair users. The purpose of this study was to investigate the test-retest reliability of IAT in order to introduce a reliable standard agility test that can be used not only for training athlete wheelchair users but also for studying wheelchair manoeuvring.

7.3 Methods

7.3.1 Subjects

Eleven healthy able-bodied subjects volunteered to participate in the study (6 women and 5 men: (mean ± SD) average age= 27.9 ± 4.74 years, average weight= 63.5 ± 10.96 Kg, and average height= 1.69 ± 0.1 m). They signed a written informed consent form, a video release form, as well as a “PAR_Q and You” (Physical Activity Readiness Questionnaire) prior to participating. participants were excluded if they put “yes” for any of the questions in “PAR_Q and You” or if they had a musculoskeletal injury that affects normal wheelchair use, exercise-induced asthma, or heart disease.

7.3.2 Experiment content

Subject participation was according to an experimental protocol that was approved by the Human Research Ethics Board of the University of Alberta (ID: Pro00003315_AME6). Each subject had two sessions of testing that were at least one week apart (mean [SD]=10 [4.1] days); In the first session, they performed four trials and at the second session they performed eight trials of IAT (See Figure 7-1). Subjects were given some time between the trials (at least three minutes) in both sessions to rest. The next trial did not start until the participants declared they had had enough rest and their heart rate returned to their resting heart rate.

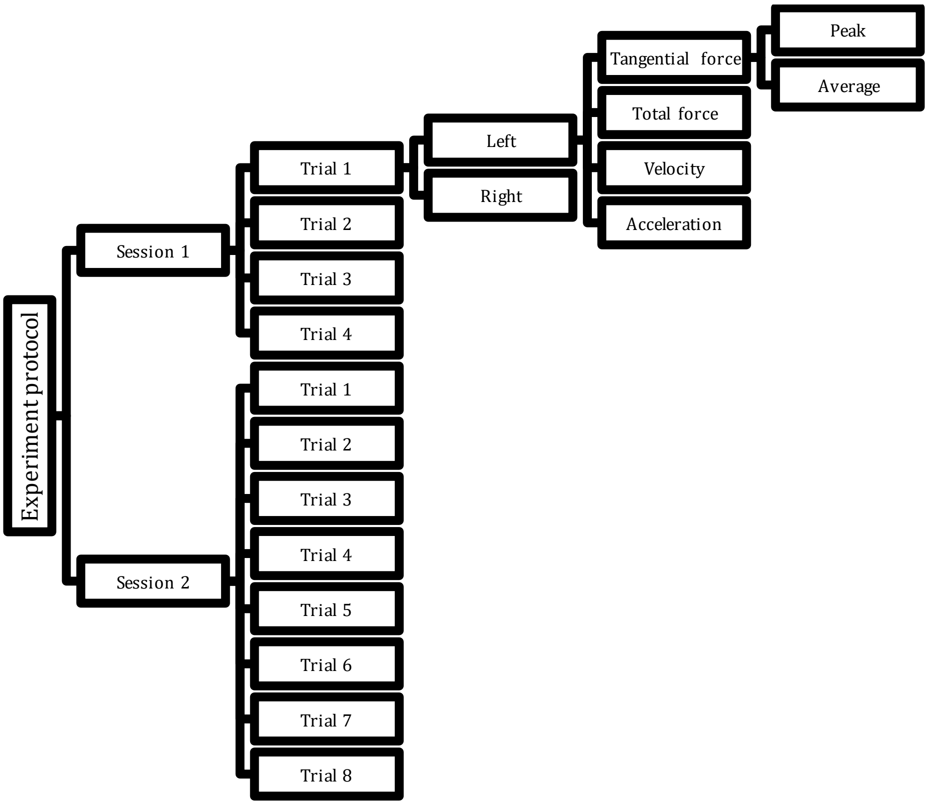


Figure 7-1 Flow chart of the experiment including the measured variables

IAT is an obstacle course consisting of 8 cones that are positioned as shown in Figure 2 and the participants should run the path (shown in Figure 7-2) as quickly as they can. IAT was originally designed to be started from the prone position; the person performing the test was to lay down in the prone position and at the beginning of IAT he/she would quickly stand up and start running [193]. However, IAT sometimes has been used without the prone start [203].

For this study, a standardized version of IAT was used [193], which is overall 10 by 5 meters court; the cones on the horizontal lines are placed 2.5 meters in between and on the vertical line they are positioned 3.3 meters apart (See Figure 7-2). Subjects were instructed to propel their wheelchair through the IAT path at their comfortable speed. They were asked to read their heart rate just before and just after finishing the test, using a Fitbit Charge HR (wireless heart rate and activity wristband). The wheelchair that was used for the experiments was Quickle GP and was equipped with two 24-inch SMART^{Wheel}s, the validity of which has been shown by Asato *et al.* [206]. The subjects were video-captured while performing the test.

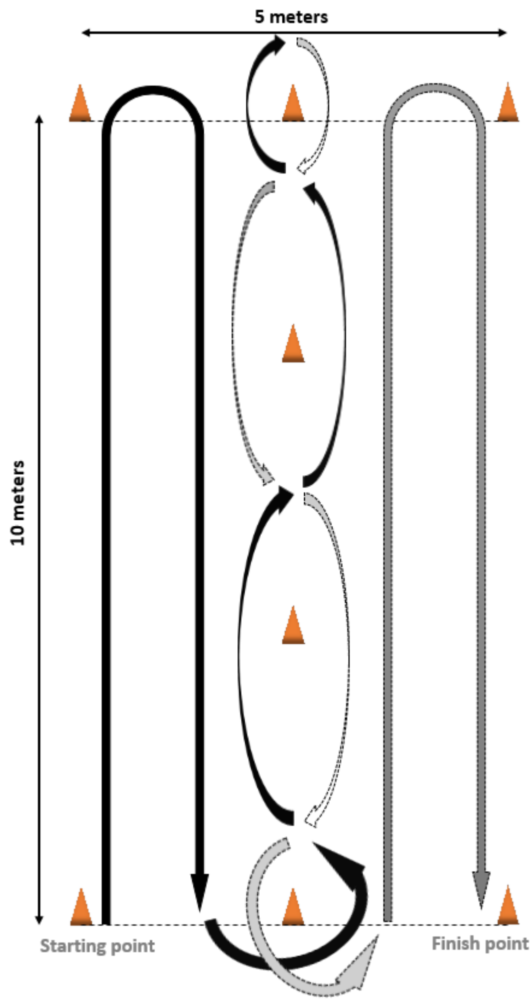


Figure 7-2 Illinois Agility Test [193]

7.4 Data Analysis

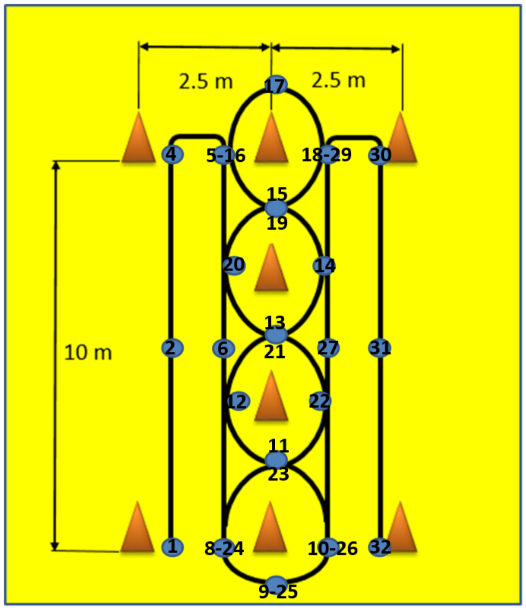
At each trial, velocity and force information were recorded using SMART^{Wheel}. Then, using a custom code the authors wrote in MATLAB, two categories of parameters were derived for each trial: data series and single-value parameters.

7.4.1 Data-series parameters:

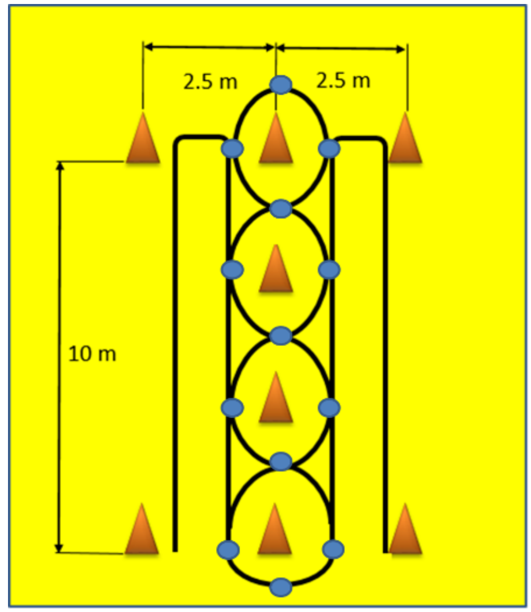
Sixteen variables were derived at thirty-two measurement points (see Figure 7-3) throughout the IAT path: peak and average magnitude of tangential and total force, velocity, and acceleration, both for the left and the right side (See Figure 7-1). Twenty-nine measurement points are shown in Figure 7-3(a). Three other points (points number 3,7, and 28) pertained to the last Straight-Line Propulsion (SLP) push in the SLP part of IAT, just before turning pushes were started Figure 7-3(c) and (d)). The guidelines used to find the measurement points are presented in Table 7-1.

To find the measurement points, we were able to use Optitrack (motion analysis) or regular cameras to visually check the video recordings. Optitrack (NaturalPoint, Inc., USA, ARENA Motion Capture- Software: Motive_Tracker- Hardware: Prime 17W and S250e cameras) is a very high precision device and provides very accurate data but due to relatively prolonged nature of the tests and the number of repetitions of the tests, using Optitrack would require tremendous amount of time to track each tests frame by frame and run custom algorithms to find the measurement points. On the other hand, using video recordings, it was possible to reduce the analysis time by a factor of 5 and also prevent possible misinterpretations of participant's movement., e.g. when the participant makes a substantial turn but is still on SLP and has not yet started the real turn.

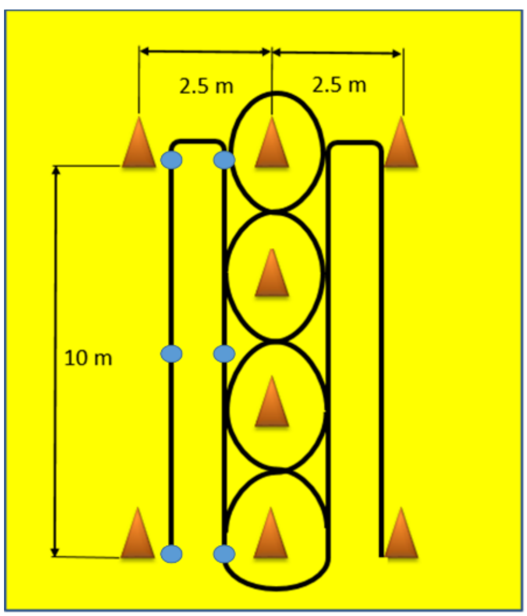
To better decide which method should be used for finding measurement points, for only one subject, the test was recorded using both Optitrack and regular cameras. Then, the measurement points were found using the trajectory obtained from the Optitrack data and the guidelines presented in Table 7-1. The measurement points were also found visually by reviewing the video record of the test frame by frame (frame rate of 30 fps) using Avidemux 2.6.13. The points found using each method were then analyzed using SPSS (IBM® SPSS® Statistics Premium



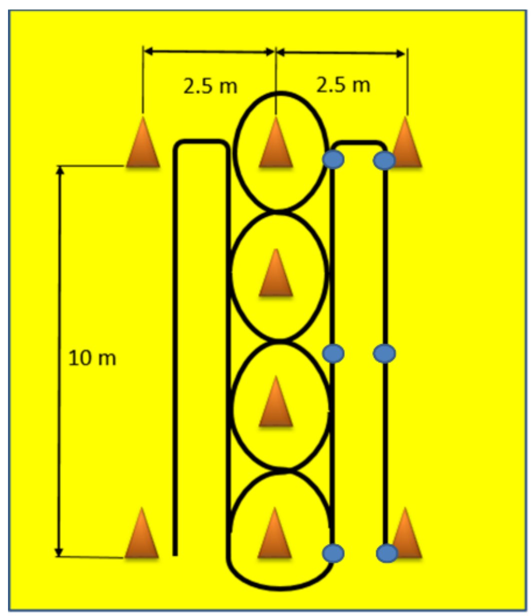
(a)



(b)



(c)



(d)

Figure 7-3 The measurement points considered throughout the IAT path; (a) shows the full path, (b) shows the manoeuvring part of the path, and (c) and (d) specify the straight-line propulsion part

Table 7-1 Method of finding the location of measurement points. Only information for half of the points are provided here; the other half of the points use similar methods, due to symmetry. For point numbers please refer to Figure 7-3(a)

Point	Method of finding the coordinates
1	The first data set
4	At the time the yaw angle of body passes -7 deg*
2	When y equals to average of y coordinates of point 1 and 4
5	At the time the yaw angle of body passes -173 deg
8	At the time the yaw angle of body returns to -173 deg and passes it
6	When y equals to average of y coordinates of point 5 and 8
9,11,13,15,17	When x equals to 0**
10,12,14,16	At the extrema of y
* (-7) degrees is the inverse tangent of ((2.5m/2)/10m) and is the threshold after that we are sure turning has been started. ** The origin of the coordinate system is placed at the middle cone at the starting and finish line.	

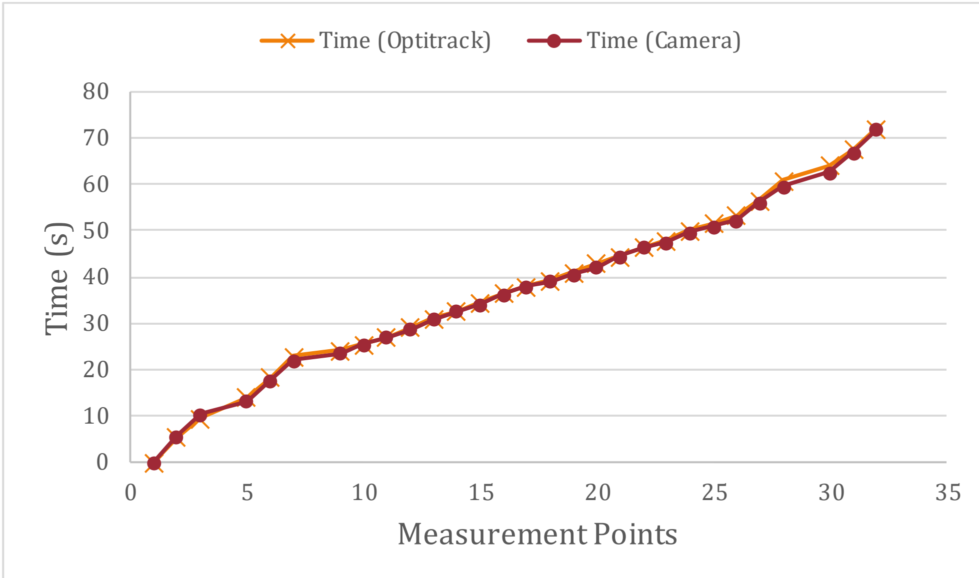


Figure 7-4 The times one subject passed the measurement points obtained from Optitrack and camera records

GradPack 24 for Windows) to find the agreement ICC and Pearson Correlation Coefficient (PCC).

The data obtained using the two techniques are shown in Figure 7-4. This graph shows the number of times the person passes the measurement points. They look very close and the PCC and ICC were found to be 100% between points obtained from Optitrack and camera (Confidence Interval (CI)= 99.8%-100%). This is compelling evidence in favor of using the camera data which helps reduce the analysis time and prevent errors after finding all the measurement points for all variables and for all participants-trials, variables pertaining to each trial were averaged over all subjects. Then, the 32 measurement data for each variable were averaged over all trials in each session. As a result, 32 averaged measurement points were achieved for each session-variable.

Data were checked for normality. Data analysis was performed using SPSS. In case any variables had a non-normal distribution, the Two-Step [207] method was used to transform the data of that variable for both sessions. Then, the test-retest reliability of the 16 variables was found using absolute agreement Intra-Class Correlation coefficient (ICC model (2,4)) between the scores of the first and second session. Also, the Standard Error of Measurements (SEM) was determined for each variable, using this formula [208]:

$$SEM = SD \times \sqrt{(1 - ICC)} \tag{88}$$

Where SD is the pooled standard deviation of all data for each variable and ICC is the intra-class correlation coefficient obtained for that variable.

7.4.2 Single-value parameters

Thirty-seven variables were derived as single-values for each trial-subject, which are significant parameters in biomechanics of wheelchair propulsion, and decisive indices of similarity of the tasks repeated over different sessions. These variables are classified into four groups:

Group I: time used to finish:

- The whole trial (t_Tot (s))
- The straight-line part (t_SLP (s))
- The manoeuvring part (t_Mnv (s))

Group II: Participant's average cadence (frequency of pushing) when performing one IAT trial, for the right and the left wheel (Cadence (1/s).

Group III: Push parameters for the right and the left side:

- Push starting angle (Starting Angle (Deg))
- Push length (Push Length (Deg))
- Number of pushes used to perform the first U-turn in IAT (U-TURNS1)
- Number of pushes used to perform the second U-turn in IAT (U-TURNS2)
- Number of pushes used to perform the SLP part of IAT (ALL-SLP)
- Number of pushes used to go around the cones on the left (GOING-AROUND-CONES-ON-LEFT)
- Number of pushes used to go around the cones on the right (GOING-AROUND-CONES-ON-RIGHT)

Group IV: Sum of forces applied by the wheelchair user on the right and the left side:

- Sum of tangential forces applied on the rim during the whole path (SigmaFt_(TOT) (N))
- Sum of tangential forces applied on the rim during the SLP part (SigmaFt_(SLP) (N))
- Sum of tangential forces applied on the rim during the manoeuvring part (SigmaFt_(MNV) (N))
- Sum of absolute tangential forces applied on the rim during the whole path (SigmaFt_ABS(TOT) (N))
- Sum of absolute tangential forces applied on the rim during the SLP part (SigmaFt_ABS(SLP) (N))
- Sum of absolute tangential forces applied on the rim during the manoeuvring part (SigmaFt_ABS(MNV) (N))
- Sum of total forces applied on the rim during the whole path (SigmaFtot_(TOT) (N))
- Sum of total forces applied on the rim during the SLP part (SigmaFtot_(SLP) (N))
- Sum of total forces applied on the rim during the manoeuvring part (SigmaFtot_(MNV) (N))

Data analysis for the single-value parameters were also performed using SPSS. Firstly, for each subject all variables were averaged over all trials in each session. Secondly, these averaged data were checked for having normal distribution using SPSS. Thirdly, variables that did not have normal distribution were transformed using the Two-Step method [207]. Fourthly, one one-way MANOVA was performed for variables belonging to each group (four MANOVAs in total). To check the assumption of homogeneity of variances (a necessary condition for performing MANOVA [209]), when computing Box's Test of equality of covariance matrices was not possible because of non-singular cell covariance matrices in the data (being caused by occasional missing data), the inter-item covariance matrix of all variables belonging to that group was developed and variances were visually checked for having close values.

7.5 Results

7.5.1 Data series parameters:

10 out of 32 data series variables (16 variables for each session) needed a transformation to comply with normal distribution. Since some of these non-normal variables belonged to only one session and some of them were repeated in both session, 14 variables were transformed in total. To check the agreement of data, both data series were compared and were based on the same construct: either the real variable or the transformed function of it.

Table 7-2 details the results achieved for the data series parameters. The average ICC of these variables was 89%. Also, the average 95% confidence interval was (44%-96%) (See table 2 for details). Considering ICC=0.6 as the minimum acceptable ICC [210], all the data series variables studied were shown to be reliable when performing IAT by wheelchair users, except for the average total force of the right side.

7.5.2 Single-value parameters:

The results of single-value parameters of eleven subjects are presented in this part. After checking for compliance with normal distribution, 15 out of 37 variables (38%) were shown to need variable transformation, which was done using the Two-Step method [207]. The results of each group of single-value parameters are presented under the respective headings.

7.5.2.1 Group I: Time

The mean and standard deviation of time taken for subjects to finish the whole IAT path and the SLP and manoeuvring part of it is presented in Table 7-3.

Table 7-2 ICC, lower and upper band of confidence interval, as well as standard error of measurement (SEM) for the sixteen variables of data series parameters: Average and peak values of velocity, acceleration, tangential and total force, each for the left and right

	Mean	*SD*	*ICC*	*CI_LB*	*CI_UB*	*SEM*
Avg Vel_L (m/s)	0.77	0.025	0.88	-0.07	0.97	0.01 m/s
Peak Vel_L (m/s)	0.86	0.02	0.83	-0.07	0.96	0.01 m/s
Avg Ft_L (N)	6.16	32.09	0.96	0.82	0.99	6.42 (N)
Peak Ft_L (N)	29.09	146.14	0.96	0.71	0.99	32.68 (N)
Avg Ftot_L (N)	16.07	8.28	0.89	0.69	0.96	2.75 (N)
Peak Ftot_L (N)	39.82	56.83	0.94	0.54	0.98	15.04 (N)
Avg Acc_L (m/s^2)	-0.006	0.009	0.98	0.97	0.99	0.001 m/s^2
Peak Acc_L (m/s^2)	0.295	0.016	0.93	0.04	0.98	0.006 m/s^2
Avg Vel_R (m/s)	0.724	0.026	0.96	0.19	0.99	0.01 m/s
Peak Vel_R (m/s)	0.82	0.019	0.91	-0.06	0.98	0.01 m/s
Avg Ft_R (N)	2.43	35.18	0.98	0.95	0.99	6.09 (N)
Peak Ft_R (N)	22.45	121.32	0.95	0.77	0.98	29.72 (N)
Avg Ftot_R (N)	14.93	5.81	0.42	-0.22	0.76	4.43 (N)
Peak Ftot_R (N)	37.02	71.01	0.73	-0.15	0.93	37.58 (N)
Avg Acc_R (m/s^2)	-0.019	0.013	0.98	0.97	0.99	0.002 m/s^2
Peak Acc_R (m/s^2)	0.26	0.019	0.97	0.93	0.98	0.004 m/s^2

Table 7-3 The mean and standard deviation of time taken to finish different tasks (s)

	Session 1		**Session 2**	
	Mean	SD	Mean	SD
Whole path	100.37	12.41	90.14	21.57
Manoeuvring part	46.54	7.07	41.57	9.72
SLP part	46.58	16.66	39.75	10.12

The significance of Box's test of equality of covariance matrices for these variables was obtained as 0.307 and this does not provide enough evidence to reject the null hypothesis of Box's test which says the covariance of matrices are equal; so this MANOVA assumption is also met.

The significance of MANOVA results was 0.441 (Pillai's Trace method) which cannot reject the null hypothesis that says the time taken for the subjects to finish different parts of IAT is statistically the same at the two sessions.

7.5.2.2 Group II: Cadence

The mean and standard deviation of the left and right cadence is presented in Table 7-4. Box's test of equality of covariance matrices was not significant (sig=0.564) which means the assumption of homogeneity of variances and covariances is met. Pillai's Trace significance was 0.428 which again shows the MANOVA results for cadence was not significant (P-value of 0.05) and therefore the push frequencies of the left and right side were statistically the same between session 1 and session 2.

Table 7-4 The mean and standard deviation of push frequency (cadence) for the left and right wheel (1/s)

	Session 1		**Session 2**	
	Mean	SD	Mean	SD
Left cadence	0.86	0.040	0.87	0.041
Right cadence	0.85	0.035	0.84	0.038

7.5.2.3 Group III: Push parameters

To find the main results obtained for push parameters during IAT task performed by eleven able-bodied wheelchair users at two different days see **Error! Not a valid bookmark self-reference.**. For this group of parameters, the Box's test of equality of covariance matrices returned no results due to non-singular cell covariance matrices in the data which are caused by missing data of one person in the first session. Thus, the inter-item covariance matrix of all variables belonging to that group was visually checked for having close values. It was observed that in 85% of all cases the data were fairly close and therefore it was concluded that the assumption of homogeneity of variances and covariances is met.

Regarding the MANOVA results for this group, Pillai's Trace significance obtained as 0.378 which means, here again, there is no statistically significant difference between push parameters in session one compared to session 2.

Table 7-5 The mean and standard deviation of push parameters for the left and right wheel

	Session 1		**Session 2**	
	Mean	SD	Mean	SD
Left Starting Angle* (Deg)	-9.72	0.23	-9.29	0.8
Right Starting Angle* (Deg)	-9.6	0.26	-9.5	0.39
Left Push Length (Deg)	18.63	3.74	21.12	4.66
Right Push Length (Deg)	17.59	3.96	20.87	5.11
Left U-turn 1	4.9	0.76	4.2	0.91
Right U-turn 1	4.2	1.29	3.7	0.96
Left U-turn 2	5.1	1.53	4.6	1.06
Right U-turn 2	4.9	1.84	3.8	1.54
Left ALL-SLP	37	9.3	31.5	8.85
Right ALL-SLP	36	8.5	31.6	8.89
Left wheel-going around cones on left side	4.6	1.08	3.8	0.92
Right wheel-going around cones on left side	4.4	0.97	3.9	0.96
Left wheel-going around cones on right side	4.7	1.12	4.2	1.08
Right wheel-going around cones on right side	4.5	0.86	3.9	0.98

***-0 degree corresponds to the person's longitudinal axis and negative angles represent the participants extending their hands backward.**

7.5.2.4 Group IV: Sum of forces

Table 7-6 details the main outcomes obtained regarding the sum of forces applies on the right and left wheel when performing IAT by non-experienced wheelchair users.

Similar to parameters of group 3, the Box's test of equality of covariance matrices returned no results and checking the inter-item covariance matrix of all variables showed that 93.2% of them had relatively close values. Therefore, it was concluded that the assumption of homogeneity of variances and covariances was met.

Also, in the MANOVA results for this group, Pillai's Trace significance was 0.356 which means the two sessions were not statistically different in terms of sum of forces applied on the wheels.

Table 7-6 The mean and standard deviation of push parameters for the left and right wheel

	Session 1		Session 2	
	Mean	SD	Mean	SD
Left SigmaFt_Tot (KN)	170.84	101.54	111.39	39.21
Right SigmaFt_Tot (KN)	55.86	14.05	51.77	17.28
Left SigmaFt_Mnv (KN)	67.51	46.59	39.12	21.45
Right SigmaFt_Mnv (KN)	36.07	5.87	34.13	7.53
Left SigmaFt_SLP (KN)	77.23	42.3	55.95	15.3
Right SigmaFt_SLP (KN)	30.71	5.2	30.73	11.65
Left SigmaFt_ABS(Tot) (KN)	252.92	69.18	224.67	28.16
Right SigmaFt_ABS(Tot) (KN)	192.56	26.4	196.86	33.39
Left SigmaFt_ABS(Mnv) (KN)	124.76	27.41	115.6	17.03
Right SigmaFt_ABS(Mnv) (KN)	93.26	15.61	91.5	14.48
Left SigmaFt_ABS(SLP) (KN)	93.18	35.15	79.51	14.54
Right SigmaFt_ABS(SLP) (KN)	62.82	6.81	66.15	10.54
Left SigmaFtot_Tot (KN)	419.33	200.95	323.1	199.26
Right SigmaFtot_Tot (KN)	350.81	205.76	393.4	108.01
Left SigmaFtot_Mnv (KN)	206.21	86.46	164.16	92.51
Right SigmaFtot_Mnv (KN)	164.79	93.91	181.55	49.65
Left SigmaFtot_SLP (KN)	161.26	96.53	118.77	80.89
Right SigmaFtot_SLP (KN)	141.18	83.56	155.15	50.64

7.6 Conclusions

In this study, we recruited eleven able-bodied subjects to participate in two sessions where they performed wheelchair manoeuvring with the aim of studying the reliability of a standard agility test for wheelchair users. A comprehensive assessment was performed considering fifty-three parameters in total, out of which, sixteen were obtained as a data series throughout the IAT. 94% of these variables were shown to have a good to excellent reliability. The average total force on the right wheel was the one with lowest reliability (42%), while the same variable for the left wheel had a very good repeatability (89%) and other force variables also had good to excellent repeatability. Therefore, we suspect that there may have been fundamental differences in the force application methods when participants repeated the test over two visits. Since the devices were calibrated before starting the experiments we do not believe that there was an instrumentation issue. Furthermore, performing MANOVA on each group of single-value parameters showed that there was no statistically meaningful difference between all of these thirty-seven variables between session one and two, which, in turn, confirms the reliability of IAT for wheelchair users.

It is worth mentioning that although the number of variables tested here are large, there is no need to use the Bonferroni correction, as this is used when we are worried that the "changes (differences) detected" are not real and could be occurred by chance. Where as in this study, we were interested to show "no differences" among the two sessions. In fact, *not considering* the Bonferroni correction adds to the power of study here, as the null hypotheses are *maintained* by Alpha level of 0.05. In other words, without a Bonferroni correction, a P-value of 0.04 would reject our null hypotheses, and show an inconsistency between the two sessions (not favorable for this study), while if Bonferroni Corrections are used, the same P-value of 0.04 would maintain the null hypotheses (favorable for this study). Therefore, as the focus of this study is to detect similarities rather than differences, the Bonferroni correction should not be used.

In this study, we recruited able-bodied subjects instead of experienced wheelchair users; this helped with the recruitment, but more importantly with the main purpose of this study which was studying the reliability of IAT rather than the wheelchair users themselves. Since wheelchair users are much more consistent in the way they propel wheelchairs compared to non-wheelchair-users due to the greater experience they have, if we could show that non-experienced wheelchair users were very consistent in performing a wheelchair manoeuvring task when repeating it 10 days later (on average), this would be a stronger evidence that this test (IAT) can be used for both experienced and naive wheelchair users, as a reliable assessment tool.

Although differences in wheelchair manoeuvres associated with pathology of wheelchair users could be thought of as a possible source of propulsion variance among wheelchair users, it does not violate our assumption of consistency of wheelchair users, as the consistency that works to the reliability benefit is within-subject which is higher among this population. While it is likely that between subject variance is more substantial in wheelchair users due to pathology, the within subject difference is expected to be much less among this group, due to their experience in wheelchair propulsion. This means that although the between subject variance could be higher in wheelchair users, it does not harm our assumption that recruiting non-wheelchair-users provide stronger evidence for reliability of IAT for wheelchair users.

The parameters that were measured in this study are useful not only to research the reliability of IAT but also because they are clinically meaningful and beneficial. For example, we observed that the comfortable speed when manoeuvring with a wheelchair was 0.85 m/s which is considerably lower than 1.1 m/s [97], [197] or 1.4 to 1.5 m/s (equal to 83.4 to 90.7 m/min) [16] which is the comfortable speed that has been reported before for performing straight-line wheelchair propulsion. Table 7-4 is also particularly useful as it displays valuable push parameters when performing IAT. One interesting observation in this table, for instance, is that when performing a rather complex manoeuvring task, participants did not show any indication of having more pushes on their dominant hand. However, although statistically the number of pushes in completing different parts of IAT was not different in session one compared to session two, we can see a consistent decrease in the number of pushes in session two relative to session one which could be clinically meaningful and a sign that participants used fewer pushes as they began to master the task.

Furthermore,

Table 7-2 shows the average velocity, acceleration, and forces participants used in one average push. However, we have also reported the actual and absolute sum of forces applied by the participants for finishing different parts of IAT. Note that this is the summation of forces at all frames in each part that is being studied (240 Hz). The clinical importance of these parameters is that eventually we are interested in the amount of force a wheelchair user needs to apply when performing different tasks.

In summary, here we were able to introduce reliability of a standard agility test, IAT, so that now it can be used in research studies focusing on wheelchair manoeuvring and also in training assessments of wheelchair-sport athletes.

After developing the VR systems, the reliability and validity of them need to be researched. This is the focus of the following chapter.

8 INVESTIGATING THE RELIABILITY AND VALIDITY OF THREE VIRTUAL REALITY ENVIRONMENTS WITH DIFFERENT APPROACHES TO SIMULATE WHEELCHAIR MANOEUVRES

8.1 Abstract

Wheelchair manoeuvring has received little attention in the literature despite its importance in mobility and performing activities of daily living and its role in developing secondary injuries for wheelchair users. The focus in this study was

technology development with iterative and proof of concept testing: three versions of a wheelchair simulator that were designed and developed for simulating curvilinear wheelchair propulsion in Virtual Reality were tested for their validity and reliability. The wheelchair simulators comprise of a sophisticated wheelchair ergometer in an immersive Virtual Reality environment and are developed for manual wheelchair propulsion. These simulators all replicate inertia in translation, in addition to taking three approaches for simulating turning.

The three systems were then tested in comparison to the real world to see how reliable and valid they are. Fifteen healthy participants were recruited to perform the Illinois Agility Test in two sessions that were at least one week apart. The ICC and Pearson correlation coefficient were found for sixteen variables to find the test-retest reliability and convergent construct validity of the systems, respectively. Overall, the three systems showed good validity and moderate reliability, with the VR_system 2 (mechanical compensation for rotational inertia) having the best scores and the VR_system 3 (software compensation for rotational inertia) having the lowest scores. Also, it was observed that performing IAT in the real world needed fewer pushes and often accompanied more negative pushes. Participants also used longer strokes in the real world compared to Virtual Reality environment.

Keywords: Wheelchair ergometer, biomechanical replication, inertial compensation in transformation and rotation, virtual reality, reliability, validity

8.2 Introduction

A significant portion of wheelchair users develop secondary injuries in the form of injuries to the shoulders and upper extremity structures just a few years after starting to use wheelchairs as their primary method of ambulation [17], [24], [26], [29]. This adds to their level of disability and reduces their quality of life. Wheelchair manoeuvring is prevalent [199] for indoor wheelchair propulsion while doing activities of daily living and has been shown [200] to have a much more significant role in causing secondary injuries in upper extremities of wheelchair users and therefore needs to be studied in more detail. However, the majority of wheelchair-related studies are focused on Straight-Line wheelchair Propulsion(SLP) [43], [51], [89], [90], [104]–[106], [211]; the number of studies that have considered non-straight line propulsion are limited and they usually only consider very simple tasks [200] or introducing devices or methods for measuring general wheelchair activities in the community [212]–[218], or using robots to simulate wheelchair manoeuvres [219].

While these contribute to our understanding of the general aspects of everyday life of wheelchair users, more in-depth and detailed research targeting real-life activities of wheelchair users are needed to establish a better understanding of risk factors and causes of the secondary injuries, along with finding strategies and solutions to minimize or eliminate those. Laboratories are equipped to conduct such research studies, as long as they provide valid and reliable instruments that can replicate biomechanics of Real-World (RW) wheelchair propulsion.

Wheelchair ergometers have been used by a number of researchers [95], [98]–[101] to help conduct such studies in the lab environment. Except for a few, the wheelchair ergometers they have used are only able to simulate SLP [95], [98]–[100]. Adding Virtual Reality (VR) helps provide wheelchair ergometers that can also simulate turning. Some researchers have attempted to develop VR environments for wheelchair users. They have been able to show promising results for training [112]–[116] and rehabilitation [111] of power wheelchair users. VR environments created for manual wheelchair users, however, have been very limited: one [109] is still under development with no validation study reported for it yet, one [101] lacks inertial compensation and does not use an immersive VR, one [121] uses a logarithmic scale and thus does not provide a one to one simulation, and finally another one [94] is very promising, except for it uses a motorized ergometer to passively simulate the biomechanics of wheelchair propulsion, rather than letting the person feel the immediate reaction of what they do and how they perform on the wheelchair ergometer.

We have recently developed [220] three versions of a wheelchair ergometer that are equipped with an immersive VR cube, developed for manual wheelchair users. They all use the natural way of pushing the wheels to navigate in the VR, plus letting the participants feel the immediate reaction of their own actions in the VR environment. More importantly, they all make use of an adjustable linear inertia system that is designed to replicate the biomechanical factors in straight-line propulsion and are discussed elsewhere [11]. All three VR_ergometer systems can simulate wheelchair manoeuvring, but they differ in the approach they take to do so. The first system (VR_sysI) represents turning by providing the corresponding visual feedback to participants' actions and relies only on replicating the linear inertia, assuming that rotational inertia can be neglected, and assumes the system will provide an experience of turning that is close enough to RW. The second system (VR_sysII) uses a mechanical system to compensate for the forces associated with rotational inertia, and finally, the third system (VR_sysIII) takes a perceptual approach for simulating rotational inertia by slowing down the simulation of rotations according to RW rotations. The objective of this study was to test and compare the reliability and validity of these three systems, standing alone, and compared to each other, to determine which system can better replicate the biomechanics of RW wheelchair propulsion. Having said that, since this study had a developmental approach and systems II and III were developed after participant recruitment had been started, the sample size of each analysis was smaller than the total sample size of 15, which is a limitation to this study.

8.3 Methods

15 able-bodied healthy subjects participated in the study, all of whom gave their informed consent prior to participating. For more information about the participants of this study please see our previous work [221]. The experiment protocol was approved by human research ethics board of University of Alberta.

After acclimatizing to VR and wheelchair propulsion in VR and RW, subjects participated in two sessions of the experiment. In the first session, they performed two rounds of Illinois Agility Test (IAT), which is a valid [202] and reliable [202], [221] agility test for wheelchair users: one round in RW and one round in VR, in a random order. In the second session, they performed one round of IAT in VR. In each round, they did four trials of the test. The two sessions were at least one week apart.

We started the experiments with only VR_sysI (besides the RW). However, from the feedback we received from the participants, we realized that participants liked SLP in the VR_sysI, but not manoeuvring. This prompted us to design VR_sysII: a system that mechanically compensates for rotational inertia.

We continued the experiments with both VR_sysI and VR_sysII: in each round of the VR tests, out of four trials, three were performed using VR_sysII and one trial using VR_sysI. By including the feedback from participants, we realized that they were not yet satisfied with manoeuvring experience in the VR_sysII. The force feedback system was not fast enough for some of the participants and this made controlling the VR_sysII bothersome. Unfortunately, we were at the technical limits of the system in terms of further decreasing its response time (10Hz).

This encouraged the design for VR_sysIII: a software-based compensation for rotational inertia, to make the perception of turning compatible with the RW experience.

After undertaking tests and reviewing the feedback from participants, we understood that VR_sysIII provided the most favorable VR experience for the participants. This encouraged us to call back the participants for another session to perform one trial of VR_sysIII and one trial of RW. Details on the development of each system is provided in our previous work [220].

Due to developmental and iterative approach used for these experiments, the sample size of data available for testing each objective of this study was different (see Table 8-1). Also, Figure 8-1 shows the flowchart of the experiments and subject recruitments. The Roman numerals indicate the systems and the first column shows the session numbers, the main sessions being numbered as 1 to 3 and the preconditioning sessions being numbered successively before them, ending by 0. In all of the main sessions, the agility test was also performed in the real world.

Table 8-1 Sample size of each objective of this study

System	VR_sysI		VR_sysII		VR_sysIII	
Objective	Reliability	Validity	Reliability	Validity	Reliability	Validity
Sample size	4	11	8	8	2	10

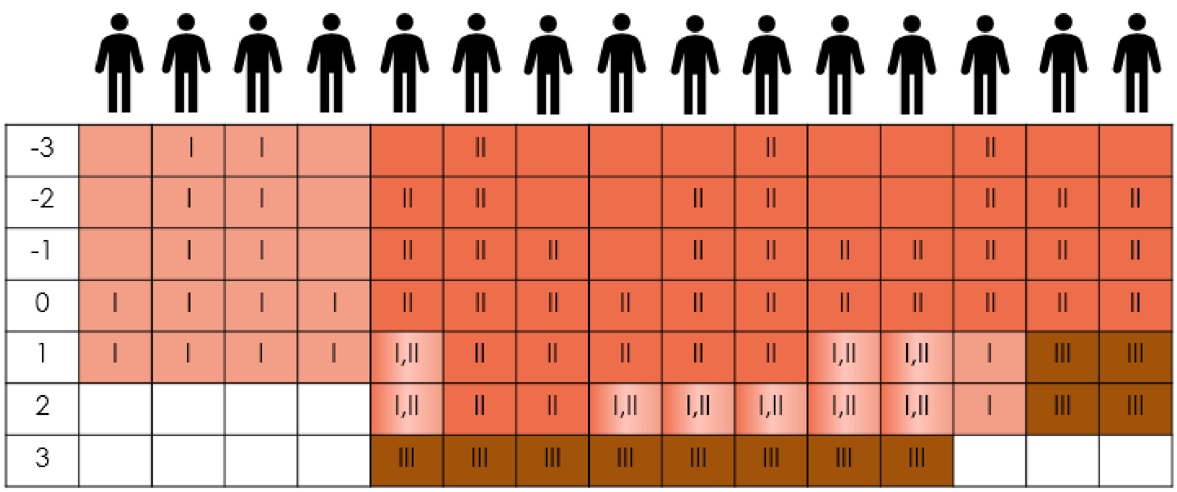


Figure 8-1 Flowchart of the experiments

8.4 Results

53 variables were measured at the experiments, out of which 16 were data series variables (measured at 32 points throughout the IAT path) and 37 variables were single-value parameters (only one reading for each trial of IAT). The results of each group are presented separately.

All data analyses were conducted using SPSS (IBM® SPSS® Statistics Premium GradPack 24 for Windows) and the α-level for all tests was 0.05.

8.4.1 Data series parameters

Variables measured/obtained in this group were: peak and the average magnitude of velocity, acceleration, the total and tangential force applied to the wheels; each measured for both the right and the left side. A MATLAB program developed by the authors of this paper was used to find the measurement variables for each test. For details on the measured variables please see our earlier publication [221].

Test-retest reliability of each system developed was found using Intra-Class Correlation coefficient (ICC) model (2,4) of each data-series variable. Also, Pearson correlation was found to assess the construct validity of each system-variable. Table 8-2 summarizes the reliability and correlation results of each variable-system. Although some of the sample sizes in this table are very small, we decided to report the results, firstly because this study is mainly designed to identify the characteristics of a VR system that is truly replicative of real world wheelchair manoeuvring, and secondly to show the promise of reliability of VR_sysI and VR_sysyIII. To clarify, correlation tests are sensitive to sample sizes, in a way that the smaller the sample size the much higher chance of getting lower correlation coefficients. When using sample sizes as low as 4 and 2, respectively 7 and 5 variables showed good to excellent correlations, using higher sample sizes will likely result in very good results for reliability of these systems.

To better understand Table 8-2, this table is color coded where dark green is for 100% correlation and dark red is for no correlation at all. Also, coefficients greater than 60% which we set as the lowest acceptable correlation are underlined. Furthermore, the number of acceptable coefficients is noted at the bottom of each column. However, it is still not easy to decide which system performed better overall. Therefore, for relative comparison of the reliability and validity of the three systems, we used ANOVA to better evaluate all coefficients obtained for each system and each correlation (reliability/validity). Results of this ANOVA test is summarized here:

Levene's test which tests the assumption of homogeneity of variances was not significant for either reliability or validity (p-value of 0.188 and 0.879 respectively). Thus, the assumption of homogeneity of variances is met.

The ANOVA results were significant for both reliability and validity (p-value of 0.02 and 0.026, respectively); therefore, the three systems were statistically different. The mean and standard deviation of each system/correlation is presented in

System I/II/ or III-Reliability/Validity		I_Rel	I_Val	II_Rel	II_Val	III_Rel	III_Val
Sample Size		4	11	8	8	2*	10
# tests averaged for each measurement point		7	26	30	30	8	16
Left	Avg Vel	0.49	0.835	0.96	0.867	0.5	0.67
	Peak Vel	0.24	0.832	0.97	0.882	0.43	0.718
	Avg Ft	0.72	0.716	0.91	0.666	0.29	0.404
	Peak Ft	0.68	0.696	0.91	0.701	0.77	0.535
	Avg Ftot	0.11	0.457	0.44	0.382	0.13	0.2
	Peak Ftot	0.75	0.568	0.86	0.571	0.33	0.352
	Avg Acc	0.9	0.801	0.58	0.671	0.6	0.632
	Peak Acc	0.14	0.664	0.9	0.823	0.51	0.645
Right	Avg Vel	0.58	0.796	0.38	0.804	0.07	0.654
	Peak Vel	0.69	0.838	0.91	0.774	0.66	0.604
	Avg Ft	0.9	0.603	0.9	0.60	0.38	0.61
	Peak Ft	0.05	0.791	0.81	0.743	0.31	0.634
	Avg Ftot	0	0.14	0.14	0.374	0.56	0.18
	Peak Ftot	0.41	0.65	0.37	0.572	0.77	0.471
	Avg Acc	0.93	0.916	0.79	0.757	0.47	0.69
	Peak Acc	0.46	0.796	0.83	0.776	0.82	0.656
# Acceptable coefficients		6	13	11	12	5	10
- Poor correlation (0-0.2): red, fair correlation (0.2-0.4): orange, moderate correlation (0.4-0.6): Yellow, good correlation (0.6-0.8): light green, excellent correlation (0.8-1): dark green							

Table 8-3. The means are also depicted in Figure 8-2.

Table 8-2 Reliability and validity results (Pearson correlation coefficient and ICC, respectively) of the three VR systems developed (Systems I, II, and III)

System I/II/ or III-	I_Rel	I_Val	II_Rel	II_Val	III_Rel	III_Val

Reliability/Validity							
Sample Size		4	11	8	8	2*	10
# tests averaged for each measurement point		7	26	30	30	8	16
Left	Avg Vel	0.49	0.835	0.96	0.867	0.5	0.67
	Peak Vel	0.24	0.832	0.97	0.882	0.43	0.718
	Avg Ft	0.72	0.716	0.91	0.666	0.29	0.404
	Peak Ft	0.68	0.696	0.91	0.701	0.77	0.535
	Avg Ftot	0.11	0.457	0.44	0.382	0.13	0.2
	Peak Ftot	0.75	0.568	0.86	0.571	0.33	0.352
	Avg Acc	0.9	0.801	0.58	0.671	0.6	0.632
	Peak Acc	0.14	0.664	0.9	0.823	0.51	0.645
Right	Avg Vel	0.58	0.796	0.38	0.804	0.07	0.654
	Peak Vel	0.69	0.838	0.91	0.774	0.66	0.604
	Avg Ft	0.9	0.603	0.9	0.60	0.38	0.61
	Peak Ft	0.05	0.791	0.81	0.743	0.31	0.634
	Avg Ftot	0	0.14	0.14	0.374	0.56	0.18
	Peak Ftot	0.41	0.65	0.37	0.572	0.77	0.471
	Avg Acc	0.93	0.916	0.79	0.757	0.47	0.69
	Peak Acc	0.46	0.796	0.83	0.776	0.82	0.656
# Acceptable coefficients		6	13	11	12	5	10
- Poor correlation (0-0.2): red, fair correlation (0.2-0.4): orange, moderate correlation (0.4-0.6): Yellow, good correlation (0.6-0.8): light green, excellent correlation (0.8-1): dark green							

Table 8-3 The mean and standard deviation of each system/correlation

Correlation	System	Mean	Standard deviation
Reliability	I	0.50	0.32
	II	0.73	0.26
	III	0.48	0.22
	Total	0.57	0.29
Validity	I	0.69	0.19
	II	0.69	0.15
	III	0.54	0.17
	Total	0.64	0.18

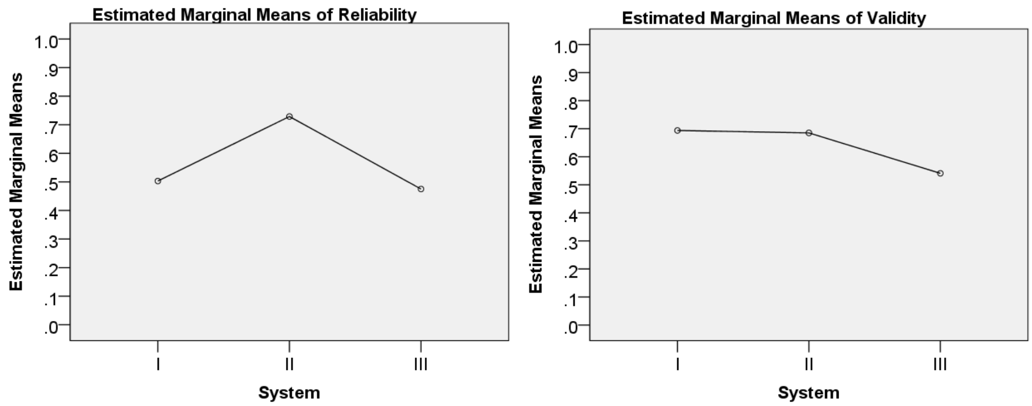


Figure 8-2 Means of reliability and validity coefficients for the VR systems

Since the number of participants in each group were equal (16), Tukey method was used for the post-hoc analysis. Table 8-4 shows the results of the post-hoc analysis.

Table 8-4 The results of the post hoc analysis

Sys. / Cor.	I, II	I, III	II, III
Reliability	different	not different	Different
Validity	not different	different	Different

Table 8-5 SEM of each system/variable

System / variable		I	II	III
Left	Avg Vel (m/s)	0.000302	0.002046	0.016002
	Peak Vel (m/s)	0.000246	0.001308	0.012777
	Avg Ft (N)	6.134281	2.787983	9.164773

	Peak Ft (N)	29.50913	13.79747	46.71399
	Avg Ftot (N)	0.127778	3.708301	0.119076
	Peak Ftot (N)	18.19551	11.69879	46.32416
	Avg Acc (m/s^2)	0.001667	5.41E-05	0.00013
	Peak Acc (m/s^2)	0.000421	0.004306	0.000556
Right	Avg Vel (m/s)	0.000265	0.000284	0.001087
	Peak Vel (m/s)	0.000168	0.002859	0.018184
	Avg Ft (N)	0.102853	0.10497	0.594043
	Peak Ft (N)	0.021929	0.558748	3.715344
	Avg Ftot (N)	0.136292	4.052248	4.750181
	Peak Ftot (N)	25.05839	23.15162	67.43278
	Avg Acc (m/s^2)	4.18E-05	6.44E-05	0.000259
	Peak Acc (m/s^2)	0.000245	0.000191	0.01636

The Standard Error of Measurements (SEM) was also found for each variable (shown in Table 8-5) using this formula [208]:

$$\boldsymbol{SEM} = \boldsymbol{SD} \times \sqrt{(\mathbf{1} - \boldsymbol{ICC})} \quad \textbf{(89)}$$

Where SD is the pooled standard deviation of all data for each variable and ICC is the intra-class correlation coefficient obtained for that variable.

8.4.2 Single-value parameters

The single-value parameters obtained in this study were grouped into four groups:

1. Time (total time to finish IAT, time taken for the manoeuvring part, and time taken for the SLP part)

2. Cadence (for the right and left wheel)
3. Push parameters (average starting angle of pushes, average push length, number of pushes needed to perform the first and second U-turn, the SLP part, turning around the cones on the left, and turning around the cones on the right, each for both the left and the right side)
4. Forces (sum of tangential and total forces applied on the right and left wheels for the whole IAT, the manoeuvring part, and the SLP part; also, the sum of the absolute tangential forces for each part)

MANOVA was found for each group to see if the difference between the two data series involved were meaningful. Table 8-6 to Table 8-11 show the details of MANOVA results for single-value parameters for reliability and validity of each system.

Table 8-6 MANOVA results for single-value parameters for reliability of VR_systemI

	Group 1	Group 2	Group 3	Group 4
# variables having non-normal distribution	0	3	3	12
Name variables having non-normal distribution	-	Tot (s) Mnv (s) SLP (s)	L_Starting Angle (Deg) L_Uturn1 L_turning around cones on right	L_Sigma Ft_SLP (N) L_Sigma Ftot_Tot (N) L_Sigma Ftot_Mnv (N) L_Sigma Ftot_SLP (N) R_Starting Angle (Deg) R_Sigma Ft_Tot (N) R_Sigma Ft_Mnv (N) R_Sigma Ft_SLP (N) R_Sigma Ft_ABS(SLP) (N) R_Sigma Ftot_Tot (N) R_Sigma Ftot_Mnv (N) R_Sigma Ftot_SLP (N)
All normal after 2-step variable transformation?	-	Yes	No- L_Uturn1 was normalized using ARTAN function.	Yes
Sig of Box's Test of Equality of Covariance	0.125	Cannot be computed	Cannot be computed	Cannot be computed
Percentage of homogeneity assumption based on checking inter-item covariance matrices	-	100	97.1	89.2
Pillai's trace sig	0.169	0.041	0.873	0.251
VR1 different from VR2?	No	Yes	No	No
Which variables?	-	Total time	-	-

Table 8-7 MANOVA results for single-value parameters for reliability of system II

	Group 1	**Group 2**	**Group 3**	**Group 4**
# variables having non-normal distribution	1	0	3	2
Name variables having non-normal distribution	R_Cadence	-	L_Starting Angle (Deg) L_Uturn2 R_Starting Angle (Deg)	L_Sigma Ftot_Tot (N) L_Sigma Ftot_Mnv (N)
All normal after 2-step variable transformation?	Yes	-	Yes	Yes
Sig of Box's Test of Equality of Covariance	0.85	0.664	Cannot be computed	Cannot be computed
Percentage of homogeneity assumption based on checking inter-item covariance matrices	-	-	94.3	91.8
Pillai's trace sig	0.55	0.885	0.745	0.813
VR1 different from VR2?	No	No	No	No
Which variables?	NA	NA	NA	NA

Table 8-8 MANOVA results for single-value parameters for reliability of system III

	Group 1	Group 2	Group 3	Group 4
# variables having non-normal distribution	0	1	4	1
Name variables having non-normal distribution	-	Mnv (s)	L_turning around cones on left R_Starting Angle (Deg) R_Uturn1 R_Uturn2	L_Sigma Ft_SLP (N)
All normal after 2-step variable transformation?		Couldn't be successfully transformed due to small sample size		
Sig of Box's Test of Equality of Covariance	Cannot be computed	Cannot be computed	Cannot be computed	Cannot be computed
Percentage of homogeneity assumption based on checking inter-item covariance matrices	100	100	97	81.6
Pillai's trace sig	0.029	0.704	0.738	0.131
VR1 different from VR2?	Yes	No	No	No
Which variables?	R_Cadence	-	-	-
Sig of Tests of Between-Subjects Effects	0.047	-	-	-

Table 8-9 MANOVA results for single-value parameters for validity of system I

	Group 1	Group 2	Group 3	Group 4
# variables having non-normal distribution	0	0	5	4
Name variables having non-normal distribution	-	-	L_Starting Angle (Deg) L_Uturn1 L_Uturn2 R_Starting Angle (Deg) R_Push Length (Deg)	L_Sigma Ft_Abs_Mnv (N) R_Sigma Ftot_Tot (N) R_Sigma Ftot_Mnv (N) R_Sigma Ftot_SLP (N)
All normal after 2-step variable transformation?	-	-	No- Log10 was used for L_Uturn1	Yes
Sig of Box's Test of Equality of Covariance	0.090	0.03	Cannot be computed	Cannot be computed
Percentage of homogeneity assumption based on checking inter-item covariance matrices	NA	NA	94.8	80.7
Pillai's trace sig	0.034	0.176	0.091	0.222
RW different from VR?	Yes	No	No	No
Which variables?	-	NA	NA	NA

Table 8-10 MANOVA results for single-value parameters for validity of system II

	Group 1	Group 2	Group 3	Group 4
# variables having non-normal distribution	NA	NA	R_Starting Angle (Deg)	L_Sigma Ft_Tot (N) L_Sigma Ft_Mnv (N) L_Sigma Ft_SLP (N) L_Sigma Ft_ABS(Tot) (N) L_Sigma Ft_ABS(SLP) (N) L_Sigma Ftot_Tot (N) L_Sigma Ftot_Mnv (N) L_Sigma Ftot_SLP (N) R_Sigma Ft_Tot (N) R_Sigma Ft_Mnv (N) R_Sigma Ft_SLP (N) R_Sigma Ft_ABS(SLP) (N)
All normal after 2-step variable transformation?	NA	NA	Yes	Yes
Sig of Box's Test of Equality of Covariance	0.296	0.069	Cannot be computed	Cannot be computed
Percentage of homogeneity assumption based on checking inter-item covariance matrices	NA	NA	93.8	90.2
Pillai's trace sig	0.086	0.003	0.362	0.277
RW different from VR?	No	Yes	No	No
Which variables?	NA	t-Tot (s) t-Mnv (s) t-SLP (s)	NA	NA

Table 8-11 MANOVA results for single-value parameters for validity of system III

	Group 1	Group 2	Group 3	Group 4
# variables having non-normal distribution	0/2	0/3	0/14	3/18
All normal after 2-step variable transformation?	NA	NA	NA	Yes
Sig of Box's Test of Equality of Covariance	0.854	0.941	Cannot be computed	Cannot be computed
Percentage of homogeneity assumption based on checking inter-item covariance matrices	NA	NA	89.5	86.6
Pillai's trace sig	0.458	0.004	0.010	0.165
RW different from VR?	No	Yes	Yes	No
Which variables?	NA	Tot (s) Mnv (s) SLP (s)	L_Starting Angle (Deg) L_Uturn1 R_Uturn1 L_Uturn2 L_All_SLP R_All_SLP L_turning around cones on left R_turning around cones on left L_turning around cones on right R_turning around cones on right	NA

Table 8-12 RW and VRs' main single-value parameters averaged over all first-session IATs

side	Environment	Cadence (1/s)	Push Length (deg)	Time to finish IAT (s)	# pushes	$\sum \boldsymbol{F_t}$ of the whole path (KN)	$\sum \lvert\boldsymbol{F_t}\rvert$ of the whole path (KN)	$\sum \boldsymbol{F_{tot}}$ of the whole path (KN)
L	RW	0.86	18.42	107.53	57.44	173.70	263.79	399.18
	VR_sysI	0.86	14.07	166.86	87.57	268.05	323.80	408.70
	VR_sysII	0.80	13.66	190.69	94.15	371.62	435.33	736.95
	VR_sysIII	0.84	14.76	170.59	88.78	562.21	586.80	595.35
R	RW	0.84	17.26	107.53	42.29	84.51	220.96	398.17
	VR_sysI	0.86	8.63	166.86	86.77	171.87	252.15	391.53
	VR_sysII	0.79	12.52	190.69	93.11	265.64	370.43	426.64
	VR_sysIII	0.83	15.65	170.59	88.85	243.31	313.02	664.73

Table 8-13 Results of ANOVA tests for main single-value parameters

Parameter	Side	RW-VRI	RW-VRII	RW-VRIII	VRI-VRII	VRI-VRIII	VRII-VRIII
Cadence	Left	=	=	=	=	=	=
	Right	=	=	=	=	=	=
Time		RW<VR	RW<VR	RW<VR	=	=	=
Push Length (Deg)	Left	=	RW>VR	=	=	=	=
	Right	RW>VR	RW>VR	=	=	=	=
# Pushes	Left	RW<VR	RW<VR	RW<VR	=	=	=
	Right	RW<VR	RW<VR	RW<VR	=	=	=
Total tangential force throughout whole IAT (N)	Left	=	RW<VR	RW<VR	=	I<III	=
	Right	RW<VR	RW<VR	RW<VR	=	=	=
Total absolute tangential force throughout whole IAT (N)	Left	=	RW<VR	RW<VR	=	I<III	=
	Right	=	RW<VR	RW<VR	I<II	=	=
Total force throughout whole IAT (N)	Left	=	RW<VR	RW<VR	=	=	=
	Right	=	=	RW<VR	=	=	=

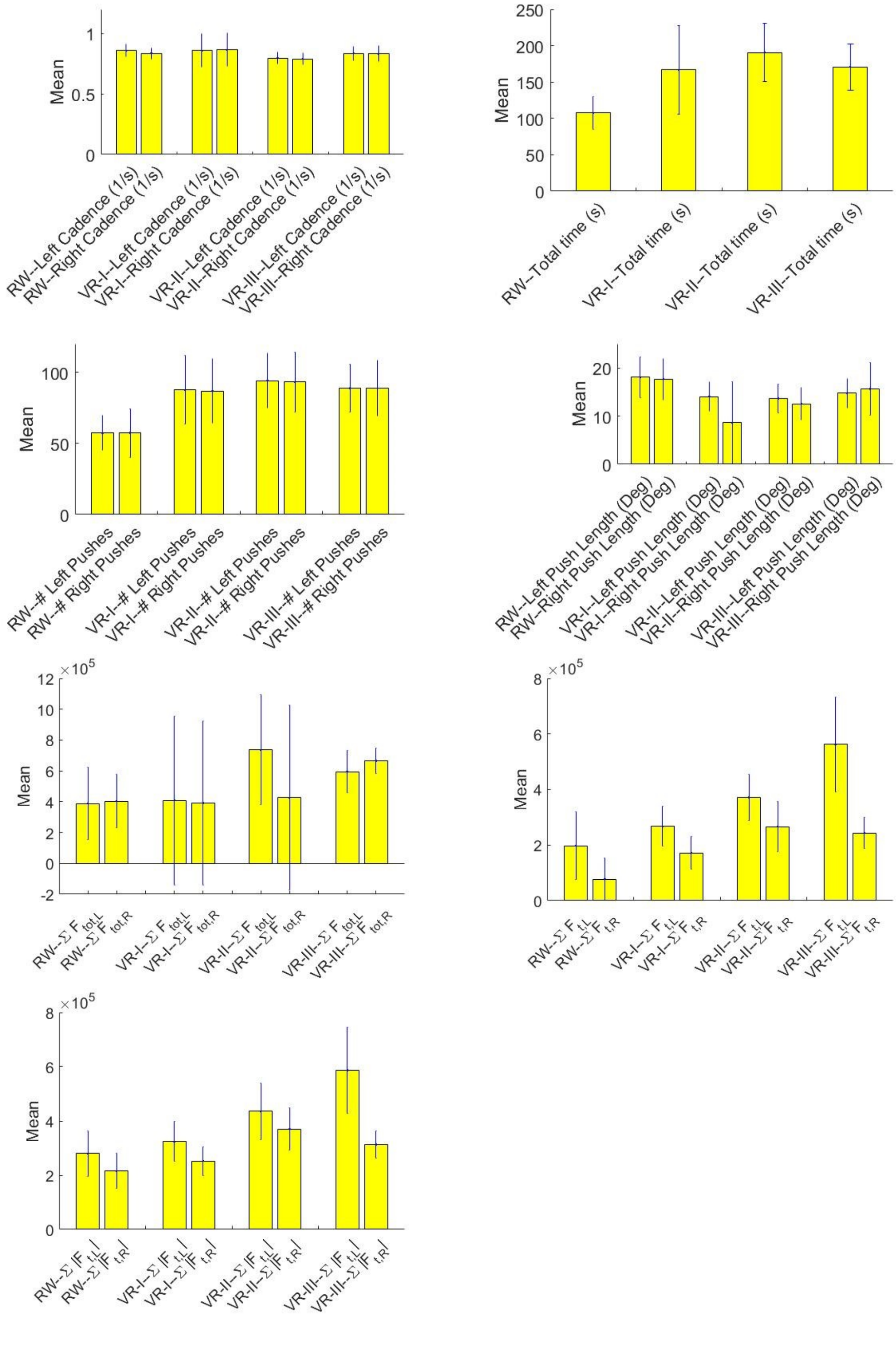


Figure 8-3 Mean and standard deviation of RW and VRs' main single-value parameters

Out of all 37 single-value variables, 13 main ones were selected to examine more fully. The average results of these main single-value parameters for RW and VR environments are shown in Table 8-12. Also, Figure 8-3 depicts the mean and standard deviation of them to make it easier to compare between the environments.

To determine if the differences that we observe between the environments in Figure 2 is statistically significant or not, ANOVA was used for each main single-value variable and Tukey method was used for the posthoc analysis. Meaningful differences between the environments for main single-value parameters are summarized in Table 8-13.

8.5 Discussion

8.5.1 Data series variables

All the three systems showed good correlations with the real world for many of the variables tested; therefore, we can say that the VR systems tested here have acceptable validity, in general. However, the reliability of systems I and III is not as good for most of the variables. Overall and based on Table 8-2, the VR systems developed in this study look more valid than reliable.

According to Table 8-4 and Figure 8-2 reliability of VR systems I and III are both low and not meaningfully different from each other but VR_sysII has statistically better reliability than both. Also, the validity of VR systems I and II are both good with no meaningful differences between them, and VR_sysIII has statistically lower validity than the other two.

Based on Table 8-2, VR_sysII has the highest number of reliable variables (11 out of 16) and VR_sysI has the highest number of valid variables (13 out of 16) with VR_sysII very close to it with 12 valid variables throughout the IAT test, out of 16 variables. Out of the three VR systems, VR_sysII which mechanically compensates rotational inertia achieved the highest reliability and validity factors, based on the 16 quantitative variables measured in this study. Table 2 is color-coded with the greener colors showing stronger correlations. Visually checking this table shows that VR_sysII looks greener in general, both for reliability and validity. Thus, we propose that the approach taken for VR_sysII can be considered to be the best VR system, based on the results of this study. Also, based on the quantitative measurements of this study and ICC and Pearson correlations, VR_sysIII that perceptually compensates rotational inertia, obtained lowest reliability and validity coefficients.

Deliberating on Table 8-2 reveals some other interesting results:

- The average total force (Avg Ftot) of both right and left sides had the lowest validity and reliability coefficients (always lower than 0.6) and has had the lowest consistency among all other variables tested. This is while peak total force (Peak Ftot) of both sides are above 0.6 in 4 out of 6 coefficients and has been much more consistent.
- The right peak velocity had the highest validity and reliability coefficients (always greater than 0.6), followed by the left peak tangential force and the left average acceleration, in general.

Based on the quantitative measurements of this study, the VR_sysIII is the least biomechanically-suitable system among the three VR systems studied, however, participants seemed to be happier with the system III than systems II or I. The authors believe that these results might be influenced by the smaller sample size and tests performed for systems I and III. In fact, a Pearson correlation between sample size used for each test/system and the number of variables with acceptable coefficients returns 92% and a Pearson correlation between the number of tests averaged for each test/system and the number of variables with acceptable coefficients returns 91%! Therefore, examining the three systems with some qualitative measurements based on participants' feedback is recommended.

8.5.2 Single-value parameters

- For VR_sysI, only 1 out of 37 variables (total time) was statistically different between the two VR sessions (reliability test), and none of the variables were statistically different between the VR and RW (validity test).
- For VR_sysII, none of the variables were statistically different between the two VR sessions (reliability test), and only 3 out of 37 variables

(time variables) were statistically different between the VR and RW (validity test).

- For VR_sysIII, only 1 out of 37 variables (right cadence) was statistically different between the two VR sessions (reliability test), and time and push parameters (13 out of 37 total variables) were statistically different between the VR and RW (validity test).

The above points provide evidence in favor of the reliability of all three VR systems and the validity of VR_sysI and VR_sysII, and indicate lower validity for VR_sysIII. Among all the variables, the time participants took to finish IAT seemed to be less consistent between RW and VR environments.

According to Table 8-13, some very interesting differences between RW and VR systems can be observed that can be interpreted as the main differences between the two environments:

1. Participants took longer to finish the tests in VR environments (I, II, and III). This can be the consequence of not feeling totally comfortable in the VR, not having proprioception feeling of movement in the VR, which is inevitable, and also having some degree of difficulties in controlling the wheelchair movement, especially for system II.
2. Participants tended to have shorter pushes in the VR, more so in system II, relative to the RW. This was observable to the researchers while performing the experiments and was confirmed by the data obtained. VR push lengths are consistently shorter in the VR (Figure 8-3), and even when the differences are not statistically significant, they can still be clinically important. Authors believe that this could be caused by slightly higher rolling resistance in the VR and some degree of discomfort with the visual feedback when moving in VR.
3. Push counts have always been greater in the VR, which is also a consequence of shorter push lengths.
4. Forces applied on the wheelchair have always been greater or almost equal to the forces applied in RW. VR_sysI has been the closest to the RW in force magnitude and VR_sysIII has been the most different. These results are a little

surprising, as participants were describing it differently: “system I needed less force than RW and system II needed the greatest forces”.

5. Forces are not usually statistically different among the VR systems, and when they are, VR_sysI has needed less force.
6. The tangential force which is the effective force in wheelchair propulsion is always higher than RW in VR_sysII and VR_sysIII. Forces in VR_sysI were also higher than RW, but not statistically significant. The total force applied in RW looks (and statistically is shown to be) more similar to VR than the tangential forces.
7. The error bars in the total force are very large in VR_sysI and VR_sysII.

8.6 Conclusion

In this study, three versions of a wheelchair simulator (sophisticated wheelchair ergometer in the VR) developed by the authors were tested against the same real-world task (Illinois Agility Test) to see how reliable and valid they are. Overall, they have good validity and moderate reliability, with VR_sysII (mechanical compensation for rotational inertia) having the best scores and VR_sysIII (perceptual compensation for rotational inertia) having the lowest scores.

Performing the task (IAT) in the RW needed fewer pushes and often accompanied more negative pushes. Participants also used longer strokes, which has been previously shown to be a better strategy in wheelchair propulsion [222]. It seems that these differences are at least partly due to the fact that proprioception does not function in the VR environment. In fact, since the feeling of the “body force” of inertia is absent in the VR, no matter how well the “boundary conditions” are simulated, this absence causes misunderstandings or missing cues for the correct movement in the VR and therefore, participants tend to act slightly different in the VR.

The results of this study are in concordance with the other studies comparing RW and VR, as in those studies also the VR task took longer to perform and accompanied poor kinematics with more errors [111], [120].

Exposing susceptible subjects to virtual reality causes them to experience visually-induced motion sickness. This could be a significant obstacle in using VR. Some participants of this study also suffered from motion sickness at their first exposure to VR, and this needed to be managed before testing the main hypotheses of this study. The following chapter shows that by exposing participants to a maximum of four preconditioning sessions, held on different days, reduced the level of motion sickness to an easily-tolerated level, so much so that they declared that they were ready to participate in a complex manoeuvring task in VR.

The other focus of this chapter is to assess the sense of presence of the participants while performing the agility test in the VR systems.

9 MOTION SICKNESS AND SENSE OF PRESENCE IN VIRTUAL REALITY ENVIRONMENTS DEVELOPED FOR MANUAL WHEELCHAIR USERS

9.1 Abstract

Visually Induced Motion Sickness (VIMS) is a bothersome and sometimes unsafe experience and is frequently experienced in Virtual Reality (VR) environments. In this study, the effect of up to four training sessions in decreasing VIMS in VR environment to a minimal level was tested by recruiting 14 healthy subjects. It was observed that all of the subjects declared readiness to undergo a series of challenging (motion sickness wise) Illinois Agility Tests (IAT) in a VR environment, after a maximum of four training sessions. Motion Sickness Assessment Questionnaire (MSAQ) was used at the end of each training session to measure different aspects of VIMS. Total, gastrointestinal, and central motion

sickness were shown to decrease significantly by the last training session, compared to the first ones.

After acclimatizing to motion sickness, participants' sense of presence and the level of their motion sickness in three novel and sophisticated VR environments were assessed. The VR environments were developed for manual wheelchair users and the difference among them is the approach they take in simulating rotational inertia: VR_sysI only replicates linear inertia, assuming rotational inertia can be neglected, and VR_sysII and VR_sysIII take a mechanical and perceptual approach in simulating rotational inertia, respectively, in addition to providing linear inertia compensations.

Participants tried each system for a maximum of two sessions. They performed up to four trials of IAT in the VR systems and in Real World (RW), and filled in MSAQ and Igroup Presence Questionnaire (IPQ) at the end of each session.

In general, the three VR systems studied here generated relatively little motion sickness and high virtual presence scores which are indicative of the good general quality of them.

With regards to the difference among the VR systems, no statistically meaningful difference was detected for either MSAQ or IPQ. As a result of high between-subject variances, which are due to natural differences between people, bigger sample sizes would be needed for statistically significant results. Nevertheless, based on the participants' comments, the VR_sysIII was the best tolerated motion-sickness-wise, and the most realistic one. Questionnaires' results also, although not statistically significant, show higher IPQ scores for VR_sysIII and lower IPQ scores for VR_sysII. Thus, we can conclude that VR_sysIII and VR_sysII gain the most and the least user satisfaction, respectively. Also, it was shown that presence has a significant negative correlation with VIMS.

Keywords: visually induced motion sickness, presence, virtual reality, linear and rotational inertia, manual wheelchair propulsion

9.2 Introduction

The VR industry is rapidly growing and finding applications in widely different areas. Many believe that VR will have a prominent role in many aspects of life, including business, education, entertainment, medicine, and research facilities, but there are others who disagree. One of the main reasons is the VIMS that is associated with VR [223]–[225], particularly the immersive ones [9]. VIMS is the unpleasant and nauseating experience that if experienced, can have a strong role in throwing VR users' sense of presence off and adversely affect how they behave and perform in VR [9]; so much so that the VR users may get totally reluctant to make use of VRs again. VIMS is an accumulative [226] construct during a VR session that when triggered can rapidly escalate and the consequences such as disorientation and vertigo may remain for hours [227] or even days [228]. This, in addition to potentially shrinking the pool of potential VR users, could be unsafe, e.g. when the users return to normal activities, as it is reported in the literature [229].

There are a number of factors that are understood to have roles in triggering VIMS, such as eye separation, Field Of View (FOV), frame rate, latency [223], interactivity [124], quality of the images and projection, and calibration of devices. However, even after providing the best quality and state of those factors, VIMS still happens, as a consequence of a mismatch between vestibular and visual stimuli's [230]. There have been ways identified to deal with this issue, but clearly, in many of VR applications, there will always be an inconsistency between those two. Fortunately, VIMS and motion sickness in general, tend to diminish when subjects are exposed to them during multiple sessions repeated over days [225], [230]–[233]. Also, other techniques could be used such as Puma exercises [234] and vestibular training [233].

A wide FOV is shown to cause higher VIMS [224], as the peripheral vision which is responsible for detecting movements, is also more sensitive to fake movements and can trigger motion sickness [235]. However, having a wide FOV helps greatly in the feeling of presence [224] and immersion, which in turn helps VR users perform better in VR [9]. Immersion affects presence [9] and a VR with a holistic design would provide a higher sense of presence.[237] We have recently developed an immersive VR environment for manual wheelchair users that is comprised of a wheelchair ergometer in a VR cube. The interaction with the VR is through the natural and realistic process of pushing the wheels. Being able to see oneself when propelling the wheelchair adds to the feeling of immersion and is thought to intensify the feeling of presence.

The VR environments used in this study took three approaches in simulating wheelchair manoeuvres: VR_sysI only replicates linear inertia, assuming rotational inertia can be neglected, and VR_sysII and VR_sysIII take a mechanical and perceptual approach in simulating rotational inertia, respectively, in addition to providing linear inertia compensations.[221] The objective of this study was to assess motion sickness and virtual presence of participants when performing wheelchair manoeuvres in each of those three systems, in addition to assessing the effect of up to four training sessions in acclimatizing to motion sickness. We hypothesized that up to 4 preconditioning sessions will supress VIMS to an easily-tolerable level, and also the VR systems tested in this study would receive relatively high presence score.

9.3 Methods

14 able-bodied subjects -8 female and 6 male- participated in the experiments approved by Human Research Ethics Board of University of Alberta. Subjects signed a consent form and a ParQ & You physical readiness form prior to participating. They were 27.9 ± 4.74 years old with no significant prior VR experience.

On their first visit, participants were instructed on propelling a wheelchair in both VR and RW. Based on how they reacted in that session in regard to motion sickness, some of them were asked to complete up to 3 more training sessions to be prepared for participating in the main sessions, without motion sick being a problem. The number of additional training sessions were determined based on their motion sickness score at the end of each training sessions and also from asking then directly if they feel ready to participate in the main experiments. This protocol was designed based on research studies that have reported having participants try VR in 4 [232], 5 [233], [237], and 6 [225] sessions has helped them acclimatized to motion sickness. These training sessions should be held on different days [229], as sleeping between the sessions help the brain “learn” it better (neuro-plasticity).

At the end of each training session, participants filled in MSAQ, a questionnaire that has shown to be valid and reliable [238] in assessing different aspects of motion sickness. Subjects’ readiness for participation in the main tests was decided based on their MSAQ score in the last training session *and* their written declaration that they feel ready and confident. Subjects then participated in two main sessions that each consisted of performing a standard and reliable [221] and valid [202] agility test, IAT, with a wheelchair, both in RW and in VR environment; they completed four trials of VR IATs and four trials of RW IATs. At the end of each main session, participants completed the MSAQ and also IPQ, both can be found in Appendices B and C. Most studies have tested the sense of presence in the VR using post-test questionnaires [239]. In this study, therefore, to measure the sense of presence in the VR systems we used the IPQ questionnaire, The validity and reliability of which has been shown using a large sample size (n=296) [240].

The VR environment consisted of a sophisticated wheelchair ergometer placed inside an EON Icube^{TM} Mobile (Figure 9-1). This VR environment allows a real-life-like experience of ambulation in VR by using the *wheelchair* as the interface to move around in VR. The wheelchair ergometer is equipped with an inertia system that replicates biomechanics of straight-line wheelchair propulsion [11]. This first VR system, VR_sysI, is a system that replicates straight-line biomechanically, but does not provide inertial compensation for simulating turns and relies on the participant’s perception of turning. In other words, the participant will use the wheels to turn as they do in RW, and the VR scene will show the turn, but there will be no inertia resisting against turning. Subjects recruitment started using this system. It is worth stating that all the participants recruited were included in data analyses of the VR systems they had tried during the main sessions.

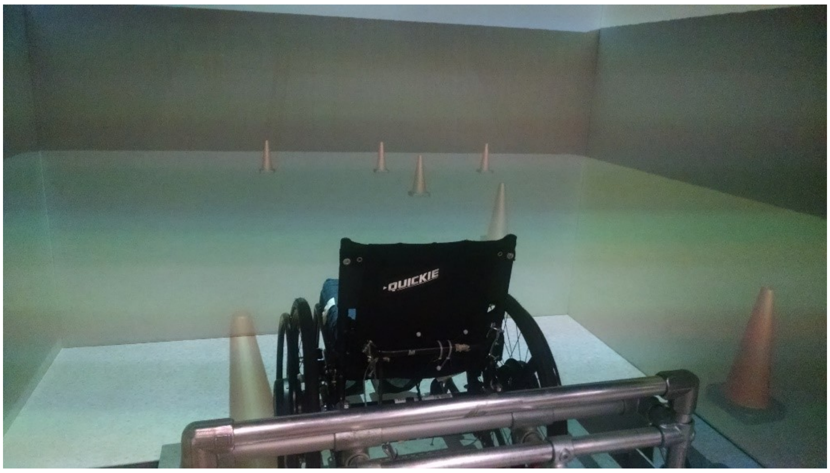
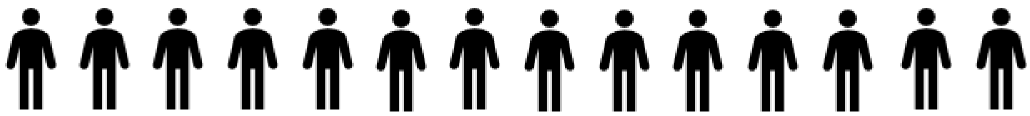


Figure 9-1 EON IcubeTM Mobile

After recruiting a few subjects, screening the comments they made during their visit led us to believe that lacking the inertial compensation for wheelchair turning triggers motion sickness and is hard to tolerate and get used to, and also, it did not *feel like the real world*. Therefore, VR_sysII was designed and built: a system that uses a pneumatic braking system to simulate rotational inertia by adding friction to one of the wheels when turning, as needed. However, as this system uses a pneumatic system, alternating between the set braking pressures builds up hysteresis and makes it impossible to formulate and control the set pressures. To solve this problem, the pneumatic pressure was reset back to 0 between every two set points; this, inevitably, limited response time in the system, which could not be bigger than ~10 Hz, due to mechanical constraints. This way, when the participant wanted to turn fast and pushed harder on one of the wheels, sometimes they could not see the corresponding turning on the visual feedback right away, due to the slower response time. Therefore, they pushed harder, and only then saw that they were turning more than intended, because they had pushed harder than needed. This made the control of the VR_sysII to be rather hard.

Participant recruitment was restarted using VR_sysII; After recruiting a few more participants and based on the feedback received from them, it was realized that although helpful at the beginning, participants started showing dissatisfaction for VR_sysII after acclimatizing to VR. This was because VR_sysII was harder to control, due to the slow response time of the pistons. Thus, to compare the two systems, 1 VR trial and 1 RW trial of each session were completed using VR_sysI and the rest using VR_sysII (see Figure 9-2). In addition, a new system was designed and built: VR_sysIII. Instead of mechanically compensating the rotational inertia, this system was based on inducing the perception of inertially rotating by slowing down the rotation in the VR scene. In other words, we calculated how much turning would happen in RW if the participant pushed the wheels the way they did, and slowed down the turning in the VR, accordingly. Details on the VR systems can be found in our former publication [221]. When this system was ready, 2 participants used it in their main sessions. Based on their feedback and as they seemed very satisfied with their experience, former participants were called back to participate in one more main session to try VR_sysIII. Figure 9-2 shows a summary of the experiments. The first column shows the session numbers (-3 to 0 indicate preconditioning sessions and 1 to 3 indicate the main sessions), and the other columns are indicative of the VR systems tried by the participant. In all of the main sessions, real-world tests were also performed by the participants.



-3	I	I			II				II			II		
-2	I	I		II	II			II	II			II	II	II
-1	I	I		II	II	II		II	II	II	II	II	II	II
0	I	I	I	II	II	II	II	II	II	II	II	II	II	II
1	I	I	I	II	II	II	II	II	II	I,II	I,II	I,II	III	III
2				II	II	II	II	I,II	I,II	I,II	I,II	I,II	III	III
3				III	III	III	III	III	III	III	III			

Figure 9-2 Flowchart of the experiments

Table 9-1 shows the technical descriptions of the VR systems that relate to VIMS.

The data then was classified into 7 sessions: from session -3 to session 3, session -3 to 0 being training sessions and session 1 to 3 being the main sessions. This was designed based on the fact that participants had any number of training sessions from 1 to 4 and so the sessions were defined as how far they were from the first main session. This is also rationally the right choice, as people who needed fewer training sessions were those who were less susceptible to motion sickness and so generally scored less for MSAQ in their first training session.

Table 9-1 Some technical descriptions of the VR systems

Factor	Our VR systems	Known thresholds
Time lag	<10 ms	<10 ms [228]
Frame rate	60 Hz	>10 Hz [124]
Haptics	Inertia is felt immediately by hand when pushing the wheels, as the systems are not motorized	-
Response time or interactivity (for VR_sysII)	Up to 0.1 s	<0.1 s* [124]
Control system	Proportional	-
Inter ocular distance	Measured for each participant and accordingly adjusted in the simulation	-

*- May not be enough if the participant loses the control at times [124]

The MSAQ [238] consists of 16 questions in 4 subcategories (5 scores): total (T), gastrointestinal (GI), central (C), peripheral (P), and sopite-related (S). In the original MSAQ each question has a score from 1 to 9; so the total score is the sum of all scores that will be a number between 11 and 144. The four subcategories also follow a similar rule. This scale was not intuitive in the context of this study, so we slightly modified the MSAQ to make the scores more understandable: we changed the scale of each question to 0 to 9 and at the end, scaled the total score and all subcategory scores to a maximum of 100. This way, all scores are from 0 to 100, which helps to make judgments and comparisons much easier. At the end of MSAQ and for the training sessions one question was added to directly ask if they felt ready to participate in the main experiments.

The IPQ contains 4 subcategories, hidden in 14 questions: INVolvement (INV), Experienced Realism (ER), Spatial Presence (SP), and General (G). In addition to the original 14 questions, 2 more questions were added to the IPQ to specifically ask them about their idea of similarity/difference between RW and VR, as below:

How do you rate your trials in real world relative to your trials in the virtual world?

VR was much easier the same VR was much harder

0 --5--- 10

How much similar were the forces you needed to apply to turn a given angle, i.e. 45 deg?

VR needed much less force the same VR needed much more force

0 --5--- 10

However, it was later realized that question 15 was faulty and not very clear. Question 15 was supposed to ask about the general rating of VR relative to RW, in terms of how challenging it was, and question 16 was to specifically ask about rating the force they applied on the wheels, relative to RW. However, using the words ‘harder’ and ‘easier’ was misleading. Therefore, question 15 was omitted from the analyses.

9.4 Results

A number of research questions are addressed here, by analyzing the data obtained from this study. SPSS (IBM® SPSS® Statistics Premium GradPack 24 for Windows) was used for statistical analyses. At first, MSAQ results are presented and then, the IPQ data.

Small sample size and large standard deviations of most of the measurements have limited the statistical power in this study. Therefore, the statistical procedures are selected in a way to maximize the power. That is wherever possible, the parametric methods are used that have higher statistical power than non-parametric methods. However, if the assumptions were not met, the non-parametric methods have been utilized.

For each question, data were checked for normality first, using Shapiro-Wilk test which is highly recommended for testing normal distribution in SPSS [242]; If not normal, an attempt was made to make them follow normal distribution using 2-step [207] method (or other transformation functions). Then, for any subcategories that followed a normal distribution, MANOVA was used to test the null hypothesis. In the case of non-normally distributed subcategories, we used non-parametric methods to test the null hypotheses. Data of groups involved were checked, firstly for having similar distributions, and secondly for satisfying the Assumption of Homogeneity of Variances (AHV). If assumptions were met, Kruskal-Wallis method was used to find whether there were any statistically significant results.

Since we improved the VR environment based on the feedback received from the participants as the experiments were going forward, the data obtained from each participant and for each session-system were different. Table 9-2 and Table 9-3 show data available for each participant. These tables were used to extract the statistically right data for testing each question, noted under each.

Table 9-2 MSAQ data available for each participant; numbers under each VR system indicates the sessions participants filled in the MSAQ

Subject	1	2	3	4	5	6	7	8	9	10	11	12	13	14
VR_sysI	-1,0,1	-1,0,1	0,1					2	-3,2	1,2	1,2	1,2	-1	0
VR_sysII				-2,-1,0,1,2	-3,-2,-1,0,1,2	-1,0,1,2	0,1,2	-2,-1,0,1,2	-2,-1,0,1,2	-1,0,1,2	-1,0,1,2	-3,-2,-1,0,1,2	-2,0	-2,-1
VR_sysIII				3	3	3	3	3	3	3	3		1,2	1,2

Table 9-3 IPQ data available for each participant; numbers under each VR system indicates the sessions participants filled in the IPQ

Subject	1	2	3	4	5	6	7	8	9	10	11	12	13	14
VR_sysI	1	1	1				2	2	2	1,2	1,2	1,2		
VR_sysII				1,2	1,2	1	1,2	1,2	1,2	1,2	1,2	1		
VR_sysIII				3	3	3	3	3	3	3	3		1,2	1,2

Repeated measures MANOVA was not used here as the data in this study could not be paired between groups, e. g. a group of same people did not try different systems in equal time differences (see Table 9-2 and Table 9-3).

9.4.1 MSAQ

Participants took 1 to 4 training sessions based on their need. Table 9-4 shows the number of participants taking different number of training sessions. Also, Figure 9-3 shows the average MSAQ score of all data (14 subjects). MSAQ scores are from 0 to 100.

Table 9-4 Number of participants taking each number of training sessions

# training sessions	**1**	**2**	**3**	**4**
How many participants?	2	3	4	5

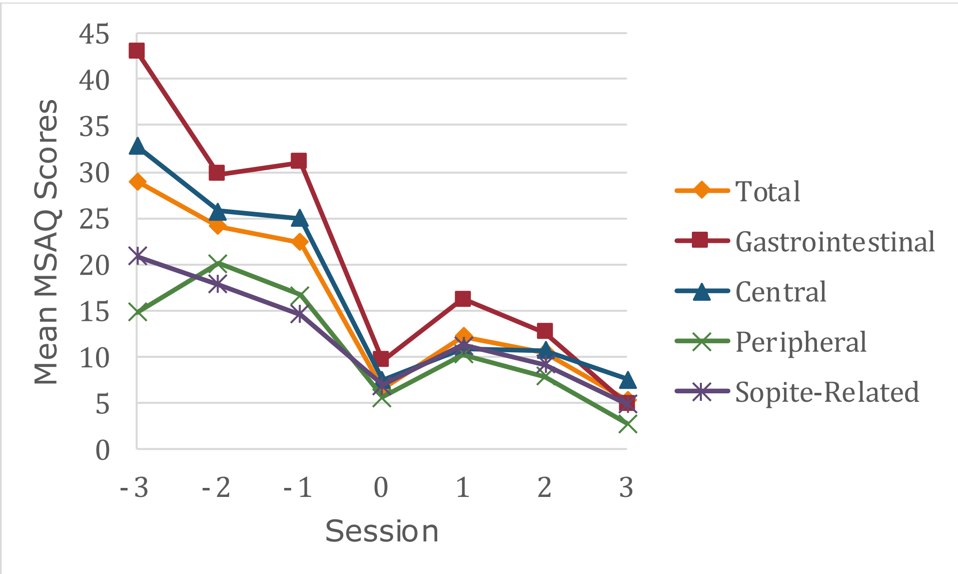


Figure 9-3 Average MSAQ score of all data for each subcategory

9.4.1.1 **Analysis 1:** The influence of training sessions in decreasing motion sickness.

Data: First training sessions' MSAQ_sysII (Sample Size (SS)= 9) versus last training sessions' MSAQ_sysII (SS=10).

Out of five subcategories of MSAQ, GI and P did not initially have a normal distribution but could be transformed using 2-step [207] method to do so. Also, Box's M test was significant (significance of 0.035 [243]) which means AHV is met. Therefore, MANOVA was used which turned Pillai's Trace significance of 0.008, with a total observed power of 0.91 and a total effect size of 0.72. A post hoc analysis revealed that the subcategories of T, GI, and C showed significant results; therefore, the training sessions significantly helped participants by decreasing those three factors of motion sickness (total, gastrointestinal, and central). Although P and S did not show statistically significant results, but as it is depicted in Figure 9-4, there is a visible reduction in these motion sickness categories that may have clinical impact.

The observed power for T, GI, C, P, and S was, respectively: 0.98,0.64, 0.95, 0.25, and 0.16, which clearly shows the great probability of type II error (false negative) for P and S ($\beta = 75\% \ and \ 84\%$). This is a consequence of the high standard deviation of the data, especially for P and S, which is a result of the big between-subject variations for motion sickness (see the overlapping standard devotions).

Nevertheless, one conclusive outcome for the effect of the training sessions in reducing VIMS to a minimal tolerable level is the question they were asked whether they felt ready to take part in the complex and rather long sessions of a complex manoeuvring task to which all participants replied "yes" after a maximum of four sessions.

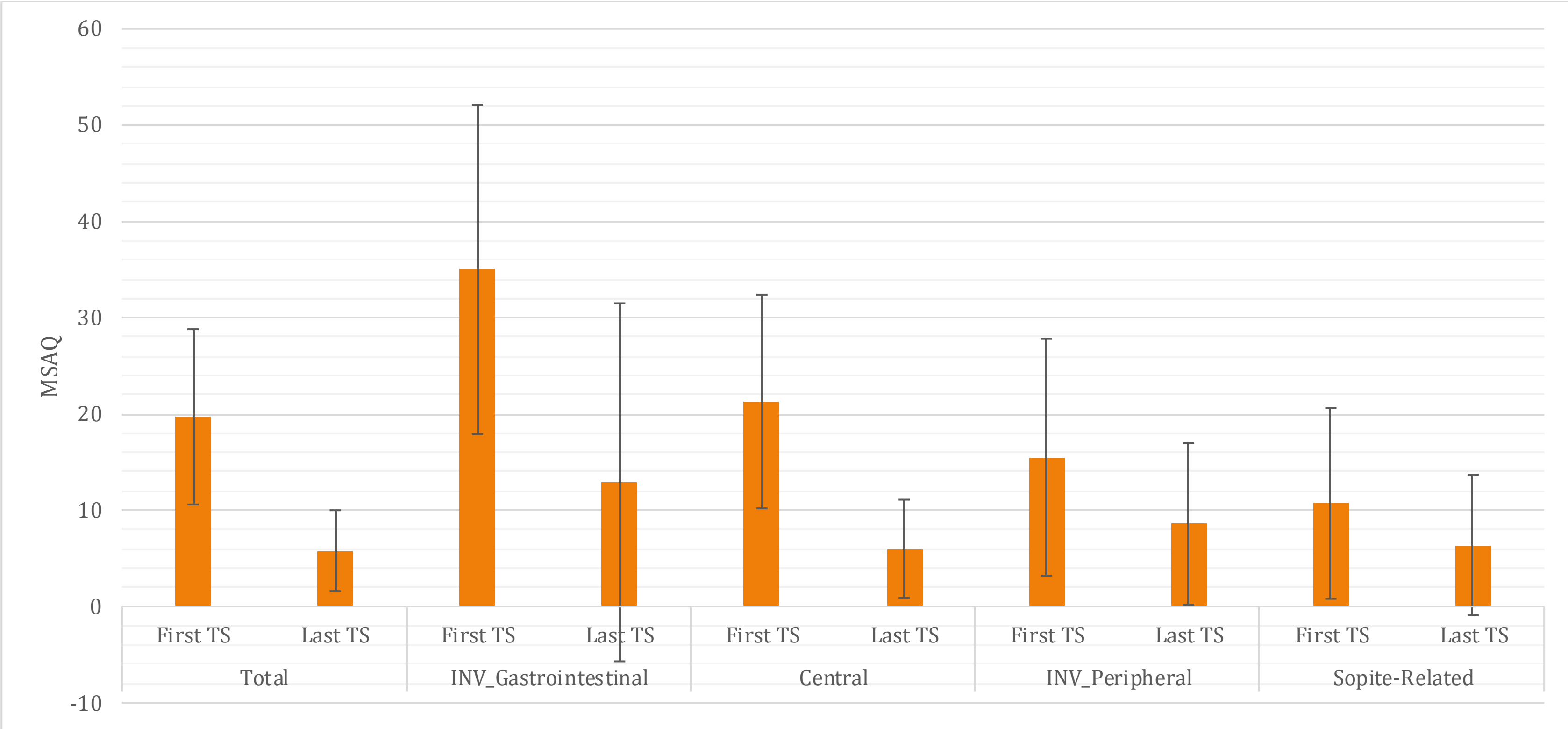


Figure 9-4 Mean and SD of subcategories of MSAQ for the first and the last training sessions

9.4.1.2 **Analysis 2:** VR_sysII vs. VR_sysI during the training stage.

Data: MSAQ scores of the training sessions: VR_sysI (SS= 6) versus VR_sysII (SS=11). Data were averaged where there was more than one data per system per participant.

All subcategories (T, GI, C, P, and S) had a normal distribution for each group (mean and standard deviations are presented in Table 9-5). MANOVA results were not significant (Pillai's Trace significance of 0.285) and the mean data also do not show any clinical impact (see the below table). Therefore, there was no statistically meaningful difference between the VR systems I and II during the training sessions in terms of the measured motion sickness.

Table 9-5 Mean and SD of MSAQ subcategories for the training sessions

MSAQ subcategory	Total		Gastrointestinal		Central		Peripheral		Sopite-Related	
VR system	I	II	I	II	I	II	I	II	I	II
Mean	24.9	23.4	31.0	32.6	32.6	26.5	19.8	16.2	13.0	15.9
Std. Deviation	18.3	16.0	28.9	24.1	26.4	19.1	23.0	14.0	10.5	16.2

9.4.1.3 **Analysis 3:** Compering the VR systems during the main sessions, for MSAQ scores.

Data: MSAQ scores of the main sessions: VR_sysI (SS= 8) versus VR_sysII (SS=9) versus VR_sysII (SS=10). Data were averaged where there was more than one data per system per participant.

Out of the five subcategories, G, C, and P did not have normal distributions at least for one system. An attempt was made to make them follow a normal distribution using transformations which only helped with the C data (using 2-step method [207]). Therefore, a non-parametric method was used for subcategories G and P.

To test the underlying assumption of Kruskal-Wallis, AHV, we ranked all data of the three systems, then found the average of those ranked data for each system and found the absolute difference of each ranked data from the mean of the respected group, and finally, found out if there were real differences between those absolute difference data, using MANOVA. This is the non-parametric equivalent of Levene's test. It was shown that the AHV is met. Then, Kruskal-Wallis test was used that returned an exact significance of 0.388. Therefore, there is no meaningful difference between different VR systems during the main experiments (stabilized motion sickness).

For subcategories total, central, and Sopite-Related that had/could get a normal distribution, MANOVA test was used. The Box's M test was significant (sig=0.076) meaning that the AHV is met. The Pillai's trace significance was 0.414 with the observed power of only 36.5%. This means that the difference between the systems during the main sessions was not statistically meaningful. However, looking at the mean data (Figure 9-5) we see a considerable difference between MSAQ scores of VR_sysII and the other systems which certainly has clinical impact. Here again, the large between-subject variability in propensity for motion sickness has caused high standard deviations and high chances of type II error.

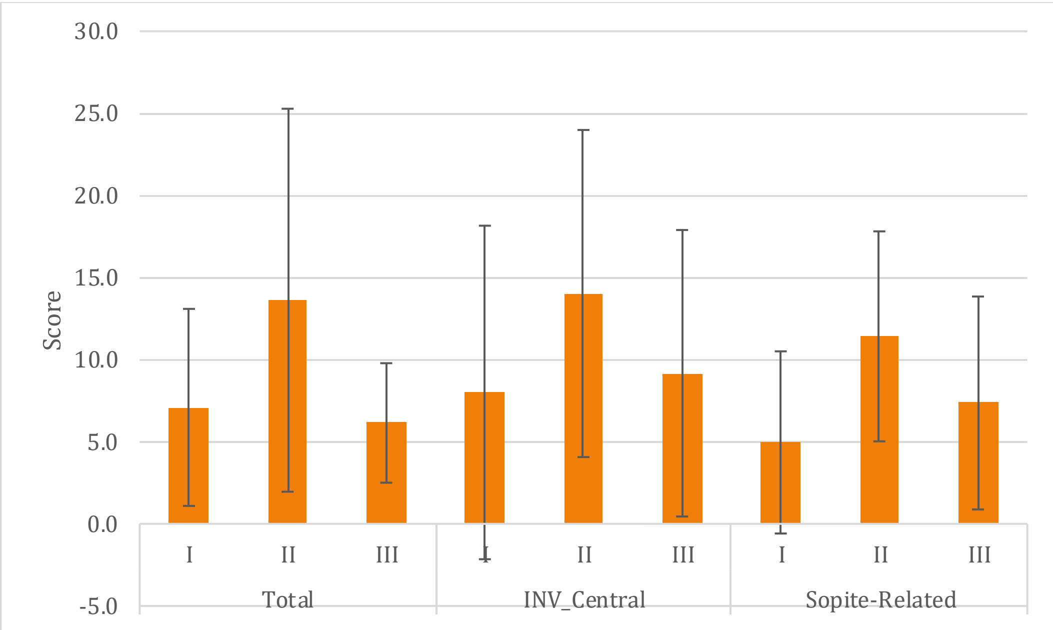


Figure 9-5 Mean and SD of MSAQ subcategories for the main sessions

9.4.2 IPQ

9.4.2.1 **Analysis. 4:** Comparing VR systems regarding the IPQ scores.

Data: IPQ scores of the main sessions: VR_sysI (SS= 9) versus VR_sysII (SS=9) versus VR_sysII (SS=10). Data were averaged where there was more than one data per system per participant.

The four subcategories were tested for normal distribution using SPSS and by means of Shapiro-Wilk test. It was shown that INV and ER were normally distributed for each system; however, G and SP were not normally distributed (medians presented in Table 9-6) neither did they respond to using transformation functions. Therefore, we used a non-parametric method (Kruskal-Wallis) for analyzing the results of G and SP, and for the others (INV and ER) we used MANOVA which is more powerful.

Kruskal-Wallis: Performing the non-parametric equivalent of Levene's test, the significance of 0.43 showed that the AHV is met.

The P-value of 0.19 and 0.23 for Kruskal-Wallis test for G and SP suggested that there is no meaningful difference between the VR systems with regard to G and SP. Also, the effect sizes of 0.13 and 0.11 for G showed that about 10 percent of the variability in rank scores of G and SP are accounted for by the VR systems.

The median of G and SP scores for each VR systems and for all VR systems as a whole are presented in the below table.

Table 9-6 Median of general and spatial presence (scores are out of 6)

VR system	Median			
	I	II	III	Total
Spatial Presence	4.4	4.2	5.1	4.4
General	4.3	4	5	4.5

MANOVA: The significance of 0.734 for Box's test of equality of covariance matrices assured the homogeneity of variances across groups, and Pillai's trace significance of 0.128 for MANOVA showed that the three systems were not statistically different. Table 9-7 presents the mean and standard deviation of normally distributed presence factors for each system and for all data.

Although no meaningful differences were detected between the VR systems regarding the presence factors, looking at means (Figure 9-6) we can see an almost linear trend from VR_sysII receiving the lowest scores to VR_sysIII receiving the highest scores. Therefore, linear contrast analysis was performed on INV and ER that revealed a *significant linear trend from VR_sysII to VR_sysI to VR_sysIII* (sig=0.033 and 0.012, respectively) with good observed powers (0.58 and 0.74) and considerable effect sizes (0.17 and 0.23). In other words, about 20% of the variability in INV and ER scores is accounted for by the VR system.

Table 9-7 Mean, median, and standard deviation of presence factors (out of 6)

VR system	Mean				Std. Deviation			
	I	II	III	Total	I	II	III	Total
Involvement	4.2	3.7	4.5	4.1	0.9	0.8	0.6	0.8
Experienced Realism	3.5	2.7	3.9	3.4	1.2	0.9	0.7	1

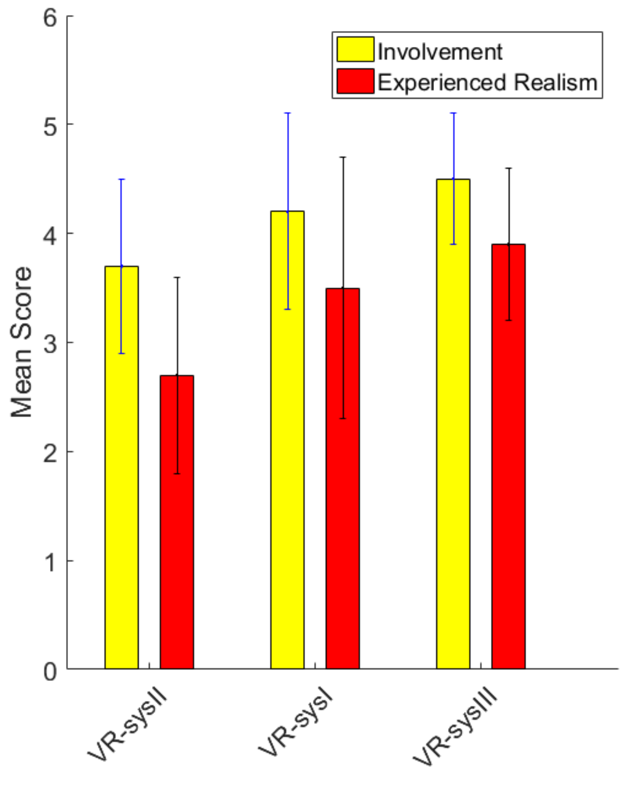


Figure 9-6 Linear trend for INV and ER among the three systems

9.4.2.2 **Analysis 5:** Comparing the IPQ scores among sessions.

Data: IPQ scores of the main sessions wherever IPQ of at least 2 sessions are available: Session 1 (SS= 11) versus session 2 (SS=9) versus session 3 (SS=8). Data were averaged where there was more than one system tried per session per participant.

All data were shown to have normal distribution across groups, using Shapiro-Wilk test. AHV is met using Box's M test with the significance of 0.828. MANOVA results, however, showed no significant difference among sessions for IPQ, with the significance of 0.066 and observed power of 75.4%.

9.4.2.3 **Analysis 6:** Correlation of IPQ and MSAQ scores.

Data: All IPQ and MSAQ (SS=14) scores of all sessions, averaged for each participant (according to the former results of analyses, it is permittable to average for each participant).

Before performing Pearson correlation test, two assumptions of normality and linearity should be met. The Shapiro-Wilk test of normality for total motion sickness and general presence and also the deviation from linearity were not significant (P-values, respectively=0.279, 0.066, and 0.93), which mean the two assumptions are met. Then, Pearson coefficient of correlation was obtained as -0.533 which is significant for a sample size of 14. This is an interesting result since it shows the more the person suffers from motion sickness, the less they grade their presence in VR.

9.4.3 Direct questions

Question number 16 (Q16) had a normal distribution. ANOVA was used to find possible statistical differences between the scores given to each system for Q16, which was rejected by 23.5% observed power. Table 9-8 shows the descriptive statistics for each system and Figure 9-7 illustrates how far from similar are the mean scores of each VR system. According to these results, with the system III, participants generated the forces that are the closest to the RW conditions.

Table 9-8 Descriptive statistics for Q16

	Mean	SD
VR_sysI	3.6	2.6
VR_sysII	6.5	4.4
VR_sysIII	5.4	1.3

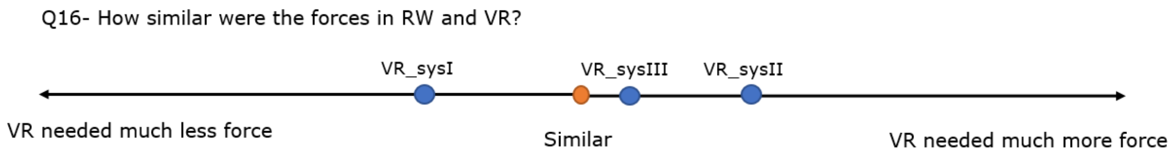


Figure 9-7 Distances each system score has from the ideal situation (absolute similarity)

9.4.4 Free text

At the end of each session, participants were asked to write down any comments that they had about their experience. Word clouds were then made out of the comments gathered for each system, that are shown in figures below. Note that, for each word cloud frequent but non-informative words were excluded, namely: VR, RW, real, virtual, world, some, more and time.

Figure 9-8 Word cloud for VR_sysI. Word cloud was made using [244]

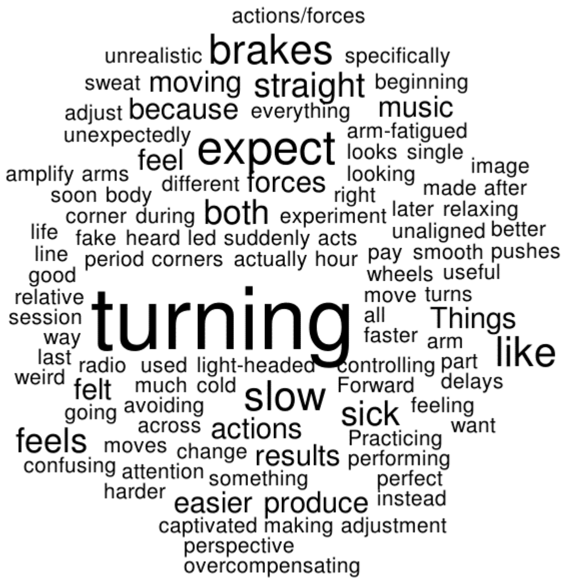


Figure 9-9 Word cloud for VR_sysII. Word cloud was made using [244]

Figure 9-10 Word cloud for VR_sysIII. Word cloud was made using [244]

As the word clouds clearly show, participants were mainly displeased with their experience of VR_sysI, complaining about it being too *sensitive* to *turning*, so much so it would make them *sick* and feel like "the world is *spinning*". For VR_sysII, however, they did not talk about sensitivity and spinning anymore, but they were not satisfied with the system yet, either. Scenes were sometimes *slower* and not how they would *expect* for *turning*. For the VR_sysIII though, they generally wrote about how this time it was *better*, *smoother*, the *easiest* for their motion sickness, and *more realistic*. These comments nicely show how the VR systems gradually improved based on the participants' feedback.

9.5 Conclusion

In this study, motion sickness and sense of presence in three VR systems developed for wheelchair manoeuvring were assessed. Also, the effect of providing up to four training sessions in reducing motion sickness to VR to a tolerable level was assessed. According to the results of this study, the training sessions significantly reduced the gastrointestinal and central motion sickness, as well as the total motion sickness level, with a very high effect size (0.72). This is a very positive and encouraging result, as one of the main issues of usability of VR is still the motion sickness that it causes in many of its users [223], [235], [245]. Motion sickness, in addition to all the negative consequences and unfavorable feelings accompanying it, could be unsafe, as what was experienced by of one of the participants of this study. When he tried VR for the first time, he suddenly became motion sick. He was helped out of the VR and recovered. However, he reported later that when he went home on his bike, he still was disoriented and had vertigo, and did not feel safe riding. This clearly illustrates the importance of looking after, predicting, and controlling VIMS for the safety of VR users.

Using questionnaires is a good way to measure presence as it is cheap and easy, does not interrupt the experiment, and has high face validity [239]. However, the recency effect is a main disadvantage of questionnaires [239] which means the scores participants give to the questions about presence are usually the way they were feeling at final parts of the test. On the other hand, motion sickness is a cumulative construct and it builds up as the time passes [226] during the experiment and gradually tends to throw the participant's concentration away. This means participants who, after receiving some training, still have some susceptibility to motion sickness have probably given scores that show more sickness than how they have felt during the whole experiment. Despite this, in general, the three VR systems studied here received relatively low motion sickness and high virtual presence scores during the main sessions, which is indicative of the good general quality of them.

It was shown in this paper that direct questions provided stronger evidence about user preference regarding sense of presence and also the overall feeling of motion sickness. Although great care was taken when selecting validated questionnaires to characterize motion sickness and virtual presence well, this seemed to be inadequate to capture users' idea about our VR systems. Therefore, it was felt that there was a need for some objective questions. This indicates a serious shortcoming in evaluating VR using currently available validated tools.

With regards to the difference among the VR systems, no statistically meaningful difference was detected for either MSAQ or IPQ. As a result of high between-subject variances, which are due to natural differences between people rather than a consequence of the study methods, much bigger sample sizes are needed for significant results. However, future work should focus on creating a VR user experience that is a significant improvement on the designs used in this study rather than recruiting a larger sample of participants, simply to demonstrate the differences between these systems.

Based on the participants' comments (word clouds), the VR_sysIII was the easiest motion-sickness-wise, and the most realistic one. Questionnaires' results also, although not statistically significant, show higher IPQ scores for VR_sysIII and lower IPQ scores for VR_sysII. Thus, we can conclude that VR_sysIII and VR_sysII gained the most and the least user satisfaction, respectively.

Interactivity constraint is an important factor that influences the sense of presence in the VR [124]. It means the time it takes from when the participant provides an input to the VR system to when he/she feels/observes the effect of it. In our case, it is the time from when the participant applies a force to the wheels to when he/she observes the corresponding displacement/rotation in the visual feedback. Interactivity constraint was a considerable issue in the VR_sysII, as it was based on a mechanical system which encompasses some delay in its application.

As already mentioned, the statistical analyses failed to detect any differences among the systems, although participants' comments caused us to anticipate some.

Also, 2 of the 5 subcategories of motion sickness (SR and P) were not shown to be significantly decreased from the first training session to the last one, despite the large decrease in their score (percent of reduction for each subcategory were: T: 70.6, G: 63.1, C: 71.8, P: 44.5, and SR:40.2). The large standard deviations in most data in this study led to a low study power and therefore a high chance of type II error, and thus, some results of this study could not be confirmed statistically. Despite this, we believe that the results of this study have important clinical and practical value.

The sample size of 10 that was used for testing most of the hypotheses here, is quite respectable when the research involves expensive, complex, and time-consuming study protocols. Nevertheless, large between-subject variations indicate that large sample sizes are needed to detect meaningful effects from natural variations. A retrospective power analysis based on the effect sizes obtained in this study, for example, shows that for testing the effects of training sessions in eliminating peripheral and sopite-related motion sickness (Analysis 5), we need to recruit 628 and 1751 subjects, respectively [246] (power=0.8 and $\alpha = 0.05$). The same observation was made by another study on simulator sickness that was unable to find significant results and stated that a much bigger sample size is needed due to great inter-individual differences of the study participants [229].

The third session took place about 4-5 months after the second session and we believe this has considerably affected the IPQ scores of the third session, as the VR was not so exciting and a new experience anymore and thus received lower scores.

According to the results of this study, there is a meaningful inverse relationship between the level of motion sickness in the VR and the level of presence the VR users experience, which is consistent with similar results of another study [245]. In other words, for the VR users to have a realistic experience, it is important to make sure that the VR is carefully designed and calibrated to minimize issues that throw the users off and trigger motion sickness. Additionally, it is necessary to ensure that the users take enough training sessions, as per needed, to eliminate/minimize the nausea and therefore, increase the usability of the VR by enhancing the realism of the experience.

The following chapter summarizes and integrates the findings of the other chapters. Limitations of this study and some suggestions for future research are also summarized.

10 CONCLUSION AND FUTURE WORK

10.1 Conclusion

The focus of this study was to develop a state of the art and sophisticated wheelchair simulator and then demonstrate its validity. Before all else, a practical example for performing a wheelchair study on an ergometer recruiting real wheelchair users is presented. It was shown that RPE can be used as a simple self-monitoring tool in regulating every-day physical activity of wheelchair users with the aim of maintaining physical fitness while avoiding over exertion. No difference was detected between local (upper arm) and overall RPE readings, both in low to moderate physical activity and exertion. Participants' comfortable SLP speed was 1.1 m/s, and they had relatively higher RPE and $\%Vo_{2Peak}$ than able-bodied people in performing low to moderate physical activities.

To make the ergometer replicate the biomechanics of real-life wheelchair propulsion, a comprehensive analysis of the biomechanics of straight-line wheelchair propulsion was performed, based on which a rigorous inertia system was constructed for the roller wheelchair ergometer. The inertia system consisted of 8 disk-shape masses mounted on threaded bars on both rollers. By altering the distance of the disks from their center of rotation, the inertia system adjusts to inertia of participants weighing approximately from 38 kg to 143 kg (more or less, depending on their wheelchair's weight). This inertia system equips the wheelchair ergometer to precisely replicate SLP.

Then, the wheelchair ergometer was placed in an immersive VR cube to prepare it for simulating wheelchair manoeuvring by providing visual feedback. Having compensation for rotational inertia is crucial not only to provide a real-life like experience and to prevent motion sickness, but also to replicate the biomechanics of wheelchair manoeuvres, in order for the results to be research-worthy. Three approaches were implemented in regard to simulating wheelchair turning, forming three VR systems. The first system only accommodates linear inertia, while the second and third systems compensate for rotational inertia mechanically and perceptually, respectively. The VR_sysI has the advantage of providing a smooth and one-to-one visual feedback. In other words, turning angles correspond to those associated with difference in the velocity of the wheels so that hand movement and pushes are proportional to the turning angle and feel real. However, this system has the disadvantage of turning too easily. VR_sysII has the advantage of providing close-to-real-life difficulty of turning and one-to-one visual feedback, but the disadvantage of non-smooth visual feedback. Finally, VR_sysIII provides a smooth visual feedback and a close-to-real-life difficulty of turning, but not a one-to-one visual feedback.

The validity and reliability of the VR systems developed was tested using the Illinois Agility Test to provide a standardized manoeuvrability task. However, before using IAT for this purpose, the reliability of IAT needed to be established. A comprehensive assessment was performed considering fifty-three wheelchair propulsion parameters in total, out of which, sixteen were obtained as data series throughout the IAT. 94% of these variables were shown to have a good to excellent reliability (ICCs ranging from 0.73 to 0.98). The average total force on the right wheel was the one with lowest reliability (42%), while the same variable for the left wheel had a very good repeatability (89%) and other force variables also had good to excellent repeatability. Therefore, we suspect that there have been any fundamental differences in the force application methods when participants repeated the test over two visits. Furthermore, there was no statistically meaningful difference between all the thirty-seven single-value variables between the two visits, which, in turn, confirms the reliability of IAT for wheelchair users.

Subsequently, the validity and reliability of the three VR systems was assessed. All the three systems showed good correlations with the real world for many of the variables tested; therefore, we can say that the VR systems tested here have decent validity, in general. However, the reliability of systems I and III is not as good for most of the variables. Reliability of VR systems I and III are both low and not meaningfully different from each other. VR_sysII on the other hand had statistically better reliability than the other two systems. The validity of VR systems I and II are both good with no meaningful differences between them, and VR_sysIII has statistically lower validity than the others.

VR_sysII has the highest number of reliable variables (11 out of 16) and VR_sysI has the highest number of valid variables (13 out of 16) with VR_sysII very close to it with 12 valid variables out of 16 for the IAT test. Out of the three VR systems, VR_sysII which mechanically compensates for rotational inertia acquired (almost) the highest reliability and validity factors, based on the 16 quantitative variables measured in this study. Thus, VR_sysII should be selected as the best VR system, based on the results of the biomechanical factors. Also, based on these measurements and ICC and Pearson correlations, VR_sysIII that perceptually compensates rotational inertia, obtained lowest reliability and validity coefficients.

The above points sum up the evidence supporting the moderate to good reliability of all three VR systems and the acceptable validity of VR_sysI and VR_sysII. The validity of VR_sysIII is less compelling than the other systems but it is still good validity. Among all the variables, the time participants took to finish IAT seemed to be less consistent between RW and VR environments.

In addition to the biomechanical parameters of the three VR systems, and as it is the case with almost all motion simulators, VIMS was a factor also to consider. The motion sickness and sense of presence of the participants in the three VR systems were assessed. The effect of providing up to four training sessions to precondition participants to VR was also assessed. We found that the training sessions significantly reduced the gastrointestinal and central motion sickness, as well as the total motion sickness level, with a very high effect size (0.72). This is a very positive and encouraging result, as one of the main issues for the usability of VR is that motion sickness causes many of its users [223], [235], [245] to abandon the VR experience, or at least feel very uncomfortable using it.

In general, the three VR systems studied here resulted in relatively low motion sickness and high virtual presence scores during the main sessions, which is indicative of the good general technical and experiential quality of them.

With regards to the difference among the VR systems, no statistically meaningful difference was detected for either MSAQ or IPQ. Nevertheless, based on the participants' comments, VR_sysIII was the most comfortably tolerated and the most realistic one. Questionnaires' results showed that although not statistically significant, there were higher IPQ scores for VR_sysIII and lower IPQ scores for VR_sysII. Thus, we can conclude that based on surveys and qualitative data, VR_sysIII and VR_sysII gained the most and the least user preference, respectively.

Interestingly, the results of the biomechanical measurements are completely different from the results obtained from the questionnaires and participants' comments. Based on the former, VR_sysII is the best and VR_sysIII is the worst system of the three, while the latter indicates that VR_sysIII and VR_sysII were the most and the least popular systems, respectively. System II was clearly an improvement in the biomechanics of system I. The main question we were interested in at first was to see if system II shows a closer to RW biomechanics than system I. While the results presented in Chapter 8 confirmed this, we realized that participants disliked the experience they had in system II. Therefore, we designed and built the third system to have both the biomechanical advantage of simulating rotational inertia and the smooth visual experience. The results showed that while participants were most satisfied with the system III, the best biomechanical results were achieved from system II.

One key observation in this study was that when simulating wheelchair manoeuvres, the technical and perceptual challenges of simulating turning is the main issue, especially when the rotational inertia is totally absent (VR_sysI). For having both a biomechanically-sound wheelchair simulation and user satisfaction, it is important to compensate for both linear and rotational inertia while providing a smooth visual feedback; something that complies with user expectations.

To sum up, unfortunately none of the systems showed both biomechanical strength and participant satisfaction. The reason was that the systems were not either providing a smooth visual feedback or RW_like inertial displacement and rotation. A more sophisticated ergometer system with improved simulation of rotational inertia appears to be the key. A better system, therefore, should provide both of these features together. As the end goal in this research was to find a recommendation for an ergometer-based VR system that well replicates wheelchair manoeuvring of real world, we suggest that such system can be built using a cantilever braking system that is controlled by Teensy and is based on the formulation provided in chapter 6. Alternatively, as wireless VR display systems become available and there have been improvements in their design to cause less motion sickness [247], an approach that enables the wheelchair user to physically rotate on an ergometer, within the VR environment would appear to be the most promising next step.

Some other interesting findings were also obtained as results of this study that are briefly listed here.

The parameters that were measured in Chapter 7 are useful not only to research the reliability of IAT but they are also clinically meaningful and beneficial. For example, we observed that the comfortable speed when manoeuvring with a wheelchair was 0.85 m/s which is considerably lower than 1.1 m/s which is the comfortable speed for performing straight-line wheelchair propulsion, obtained earlier in this thesis (Chapter 4). Also, when performing a rather complex manoeuvring task, participants did not show any indication of having more pushes on their dominant hand.

Although statistically the number of pushes in completing different parts of IAT was not different in session one compared to session two, we can see a consistent decrease in the number of pushes in session two relative to session one which could be a sign that participants used fewer pushes as they began to master the task.

It was also observed that participants took longer to finish the tests in the VR environments (I, II, and III). This is likely the consequence of not feeling totally comfortable in the VR environment, not having the proprioceptive feedback of movement in the VR, which is inevitable, and also having some degree of difficulty controlling the wheelchair movement, especially for system II.

Participants tended to have shorter pushes in the VR, more so in system II, relative to the RW. This was observable to the researchers while participants performed the experiments and was confirmed by the data obtained. VR push lengths are consistently shorter in the VR, and even when the differences are not statistically significant, they can still be clinically important. Authors believe that this could be caused by slightly higher rolling resistance in the VR and some degree of discomfort by the visual feedback of moving in the VR. Also, push counts have always been greater in the VR which, in turn, is a consequence of shorter push lengths.

Forces applied on the wheelchair in the VR environment were greater or almost equal to the forces applied in RW. VR_sysI was the closest to the RW in force magnitude and VR_sysIII was the most different. However, forces were not typically statistically different among the VR systems, and when they were, VR_sysI required less force.

Performing the task (IAT) in the RW needed less pushes and was often accompanied by more negative pushes. Participants also used longer strokes, which has been previously shown to be a better strategy in wheelchair propulsion [222]. It seems that these differences are at least partly due to the fact that proprioception does not function in the VR. In fact, since the feeling of the "body force" of inertia is absent in the VR, no matter how well the "boundary conditions" are simulated, this absence causes misinterpretation of the correct movement in the VR and therefore, participants tend to act more tentatively in the VR.

According to the results of Chapter 9, there is a meaningful inverse relationship between the level of motion sickness in the VR and the level of presence the VR users experience. In other words, for the VR users to have a realistic experience, it is important to make sure that the VR is carefully designed and calibrated to minimize issues that throw the users off and trigger motion sickness. Additionally, it is necessary to ensure that the users take enough training sessions, to eliminate/minimize nausea and therefore, increase the usability of the VR by enhancing the realism and comfort of the experience.

10.2 Limitations of the Study

- Since this study took a developmental approach based on the feedback received form the subjects, a major limitation was the small sample size. Although 15 subjects participated in the experiments the actual sample size used to test different hypotheses was usually around 10. This sample size is quite respectable when the research involves this level of expensive, complex, time-consuming study protocols and extensive analysis. Large between-subject variations call for large sample sizes to detect statistically meaningful effects that are also influenced by large natural variations in the participants. A retrospective power analysis based on the effect sizes obtained in this study shows that for powerfully testing the effect of training sessions in eliminating peripheral and sopite-related motion sickness, for example, we would need to recruit over 600 participants [246] (power=0.8 and α = 0.05). This is largely driven by the very large differences in the response of individual participants. The effect size between the systems is large enough, but the natural variation between participants overwhelms it.

- Large standard deviations in most of the data of this study leads to a low study power and therefore a high chance of type II error and therefore some results in this study could not be confirmed statistically. A less heterogeneous sample would greatly help with increasing the power of the study by decreasing the between-subject variations. This, on the other hand, will compromise generalizability (external validity) of the results.

- The results of qualitative and quantitative parameters of this study in finding the best VR system contradicted each other. This shows the importance of using mixed methods for these sort of research, where relying on biomechanical factors or qualitative data is not enough.
- The third session took place about 4-5 months after the second session and we believe this has considerably affected the IPQ scores of the third session, as the VR was not so exciting and a new experience anymore and thus received lower scores.
- Interactivity constraint was a considerable issue in the VR_sysII, as it was based on a mechanical system which had technical limitations due to a rather slow response time.
- Participants who, after receiving some training, still had some susceptibility to motion sickness probably gave scores that show more sickness than how they felt during the whole experiment, due to recency effect.

10.3 Directions for Future Studies

- Our most immediate suggestion for future research is to use a cantilever braking system for compensating rotational inertia based on the formulation presented in this thesis. This will further improve the VR experience by providing a good representation of biomechanics of RW wheelchair propulsion as well as a smooth visual feedback.
- There is an inherent difference between rotational and linear inertia, as the human body can adapt much easier to a not-perfect linear inertia or even absence of it than inexact rotational inertia. Hence, and since a simulation of rotational inertia, as it is a simulation, is always different from the real world to some degree, it may be better to refrain from simulating rotational inertia and only be concerned about linear inertia; instead of simulating rotational inertia, actual rotations can take place using turning tables. This could contribute greatly to the realistic feeling in the wheelchair simulators, especially if wireless VR goggles with

very precise design in terms of preventing motion sickness are used. This approach, then, could be considered for the next step. The benefit we identified of using the cube environment where the participant can see their body and the wheelchair might be overcome using augmented glasses (e.g. Hololens) reality rather than VR goggles but the drawback is that they are very likely to create greater nausea than the VR cube environment.

- One big challenge for wheelchair users is, however, the effect of slopes (up-down and side), which is not simulated yet in this VR environment and this presents challenges in VR at present. The next step after ensuring about a good simulation of rotational inertia should be to address this limitation.
- Although both the tangential (the useful force) and total forces in VR are greater than RW (significantly or insignificantly), the ratio of tangential force : total force is greater in VR. This could have happened for a number of reasons:

- As discussed in Chapter 5, trunk swing helps in propulsion. It is possible that in the VR, since the person do not feel the inertia in their body, their trunk swing was less in the VR and therefore they needed relatively more tangential force.
- It is possible that participants used less trunk swing as they did not feel that necessary to prevent tipping.
- Although the RR of the rollers was calibrated to simulate linoleum RR before starting the experiments, it could have increased gradually. Possibly, the increased RR was responsible for the greater increase in the tangential force.
- Or, it might have physiological roots, rather than biomechanical causes. The required force in the VR was greater than RW. Perhaps when the body is applying greater forces, the portion of unuseful forces relative to the total force applied are lower.
- Alternatively, there could be a link between the shorter stroke length and the higher tangential force in VR compared to RW.

- These or other causes could be responsible for higher tangential force : total force ratio. A study can be designed to investigate reasons for this observation.
- The work undertaken for this thesis was intended to provide a foundation to conduct valid research studies involving wheelchair propulsion for simulations ranging from everyday propulsion to elite athlete wheelchair propulsion performance. The next sequence of research questions that could be conducted in VR systems include:
- How rear axle position affects biomechanical and physiological variables of wheelchair manoeuvring on elderly wheelchair users?
- How effectively assistive devices and interfaces could reduce the biomechanical and physiological demands of wheelchair propulsion?
- How haptic based interfaces could help quadriplegic wheelchair users in navigating through narrow pathways?
- How much effect could the weight of wheelchairs have on wheelchair performance from physiological and biomechanical standpoint?
- What are the strategies used by wheelchair users during maneuovring with different radius and angle? Which ones have higher propulsion efficiencies and cause less loads on the shoulder and wrist?
- What kind of exercise and activities should be recommended to SCI patients with different levels of injury and to what extent?
- How do wheelchair users react to sudden disturbances effectively? What are the best strategies for them when facing sudden disturbances to prevent falls?
- How different are EMG activities of muscles around the shoulder when manoeuvring compared to straight-line propulsion?
- Can group wheelchair sports such as wheelchair rugby be studied to draw strategies to better make use of all members of the team?

- How can we optimize the energy expenditure when performing intense wheelchair manoeuvring?

In the following, we develop several of these research questions in more detail:

The real effect of rear axle position: Brubaker [60] has shown that placing rear axle position forward will reduce the rolling resistance on the front casters as it would reduce the portion of weight carried by front casters. Therefore, he recommends that rear axle should be positioned as forward as practically possible, especially for people with quadriplegia, to gain the benefits associated with that: enhancing wheelchair propulsion performance, reducing rolling resistance, reduction in downhill turning tendency, and reducing the force needed for turning. He accepts that this decreases the stability of the wheelchair and so increases the risks of fall. Falling could be very dangerous as many of wheelchair users suffer from osteoporosis below their level of injury and even a low velocity fall could cause bone fractures, which ironically implies more difficulties to cure and hill for wheelchair users [20]. Brubaker suggests using anti-tipping mechanisms to prevent this [60].

One point that a biomechanical researcher could raise here is about the effect of the reduction of weight the casters are bearing. Positioning the rear axle backward will reduce the amount of weight on casters only by transferring that to the rear wheels. This means that reduction of rolling resistance on casters accompanies an increase in the rolling resistance on the rear wheels. Hills, *et al.* [59], [248] conducted a series of experiments recruiting paraplegic patients to compare biomechanical factors when rear axle is positioned the furthest forward (tippy) versus when rear axle is positioned the furthest backward (stable). They ran the test for propelling on linoleum and Astro, ascending slop (1:12), and ascending a 3” curb. They concluded that although the rear axle position affected the wheelchair user’s capacity to perform, it did not directly influence the propulsion force, except for the curb. This study questioned the extent to which the “efficiency enhancement” suggested by Brubaker was *clinically significant*.

The Virtual Reality (VR) environment we are developing has the potential to be used to conduct further experiments to shed more light on this topic and its clinical relevance and significance. The safe environment that immersive VR provides allows representation of different terrains encountered by wheelchair users on a daily basis, including curbs, snowy or slippery surfaces, narrow pathways, ramps, turfs, and also turns.

Manoeuvring was an aspect not included in the Hills study [59], [248]. Our VR environments provide an ideal setting to perform an experiment on the effect of rear axle position on biomechanics of wheelchair manoeuvring. This could also be undertaken on some other target populations, such as quadriplegic patients and elderly wheelchair users, who mainly have not been included in the studies on stability, but need that even more.

To test new technologies: In 2012, Boninger in collaboration of ten other SCI researchers, including Dr. Ferguson-Pell, published a paper [123] to introduce the advances in technology in 10 years that are needed in regard to mobility of SCI patients. In that paper, they have identified machine vision (to help with eye-controlled wheelchair propulsion) and function-prediction features as helpful interfaces that could reduce the magnitude of action needed from the wheelchair user to control the wheelchair, and also to make the controlling task faster. Also, they have suggested advances needed in manipulators or robotic arms. These devices have been used to help quadriplegic patients in wheelchair control and also in performing their activities of daily living, but the downside is that they have been slow. Boninger *et al.* suggested advances in this area to make these devices faster and less-strength demanding. They also suggested adding new features such as sensory feedback to them.

Dr. Jacqueline Hebert [249], [250] and Dr. Patrick Pilarsky [250]–[252], both at University of Alberta, have been working on artificial limbs and robotic arms, sensory feedback, intelligence and reinforcement learning for years. Also, Dr. Kim Adams in collaboration with Dr. Mahdi Tavakoli is now working on developing haptic interfaces for people with disabilities. Haptic interfaces could help wheelchair users with high levels of injury that have weaker hand grasp and less hand movements to avoid obstacles much easier. There is a potential for the VR to be used in order to conduct experiments using these new interfaces in a safe controlled environment to test their application for wheelchair users.

Dr. Martin Ferguson-Pell and Adam Pinkoski, both at University of Alberta, has invented *Dema Dash* which is a real-time predictor of muscle fatigue based on real-time processing of EMG muscle activity. This device can be used in the VR systems developed in this study to study how this device could help improve the performance in wheelchair sports.

The effect of the weight of the wheelchair: Work performance of SCI patients, especially those with higher levels of injury has been shown to be less than able-bodied people [14], [15]. On the other hand, it has been shown that using light-weight wheelchairs could reduce the energy cost for 17% [14]. The instrumented ergometer developed for this thesis is designed to account for the weight of wheelchair and wheelchair user together for inertia representation. While keeping other factors the same, the new ergometer will allow us to test the effect of lighter wheelchairs on the wheelchair users' performance. This could be performed by specifying the desired weights of the wheelchair in the ergometer/VR setting.

Manoeuvrability: Manoeuvrability has received little attention in the area of wheelchair research, although it is a necessary functional skill that is encountered widely when performing activities of daily living indoors [199], and although it needs more strength to perform and generates more pressure on the shoulder. Little is known about strategies that could enhance propulsion biomechanical factors, such as increasing propulsion efficiency and reducing the amount of load on shoulders, wrists, and elbows. We still do not have clear and all-inclusive answers to questions like: what are pushing strategies that generate less stress on upper extremity? How trunk swing contributes to energy flow while turning? How characteristics of the wheelchair, i.e. its length, affect wheelchair manoeuvrability? Finding the answers to these questions and implementing them should help in the reduction of upper extremity pain and injuries in manual wheelchair users. The VR systems introduced in this thesis could be used to help to this end.

Effect of exercise: Finley and Rodgers [25] has shown that involvement in wheelchair sports does not have positive or negative effects on overuse injuries. It has also been shown that moderate activity is needed to protect the shoulder joint from degeneration, but high levels of activity could cause more degeneration [18]. There is not enough evidence to guide the exercise recommendations for wheelchair users [19] and so research is needed to expand our knowledge on the amount and types of activities that could help wheelchair users to enhance their strength and prevent joint degenerations, but avoid overuse injuries. VR environment could represent real world scenarios and yet be a thermally controlled and safe environment. Therefore, it could be used to conduct research studies on the effect of exercise on wheelchair users, especially those with higher levels of injury who need such environments to avoid adverse effects of exercise [14], [15], [19].

Sudden disturbance: One important difference between wheelchair propulsion and walking is the presence of the rolling resistance. Rolling resistance, as its name also shows, is present where there is an object rolling. It is dependent on different factors, including the type of surface. While the presence of sliding friction is helpful and necessary for both walking and wheeling, the person who is wheeling must overcome the drag caused by the rolling resistance. This could be clearly seen on the difference in the level of difficulty/ease a walking person feels when walking on linoleum and turf versus the difference that a wheelchair user feels on those two surfaces. Wheelchair users, including SCI patients, have to face this added burden for ambulation, even though they have less body strength and are more susceptible to fractures.

Grainy and rough terrains or other occasions where there is a sudden obstacle in the pathway could cause the person to become unstable. The safe VR environment would provide a perfect setting to study the effect of sudden disturbance on maintaining the balance by wheelchair users.

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Appendix A SUPPORTING TECHNICAL FIGURES

This appendix includes some technical figures and tables regarding development of the VR systems.

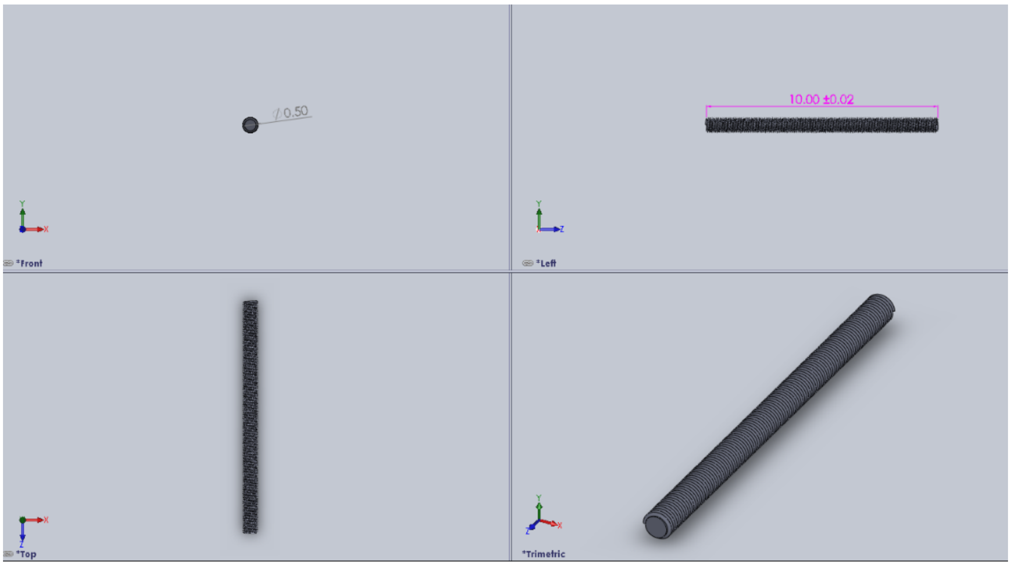
Table 0-1 Bill of materials

Part number	BOM level	Part name	Description	Quantity	Procurement Type
01	1	Encoder	quadrature encoders. Bourns Inc., California, USA. Model Number: EMS22Q51	2	off-the-shelf
02	3	Teensy	32 Bit Arduino-compatible microcontroller. PJRC LLC, Oregon, USA. Model: Teensy 3.2	1	off-the-shelf
03	2	Threaded bars		8	made-to-specification
04	2	Disks	Each disk is 5" in diameter, 1.5" thick, weighs 3.77kg (8.3 lb), and is made of AISI 1018 mild low-carbon steel	8	made-to-specification
05	2	Hubs	AISI 1018 mild low-carbon steel	2	made-to-specification
06	2	Threaded pins		2	made-to-specification
07	2	Nuts		48	off-the-shelf
08	4	Compact Rio (cRio)	NI, Texas, USA. Model NI PS-15. Made in Czech Republic	1	off-the-shelf
09	4	Pneumatic pistons	SMC, China. Model Number: CD85N25-125-B, 1.0 MPa	2	off-the-shelf
10	4	Pneumatic board		1	off-the-shelf
11	4	Pneumatic	proportional valve	2	off-the-shelf

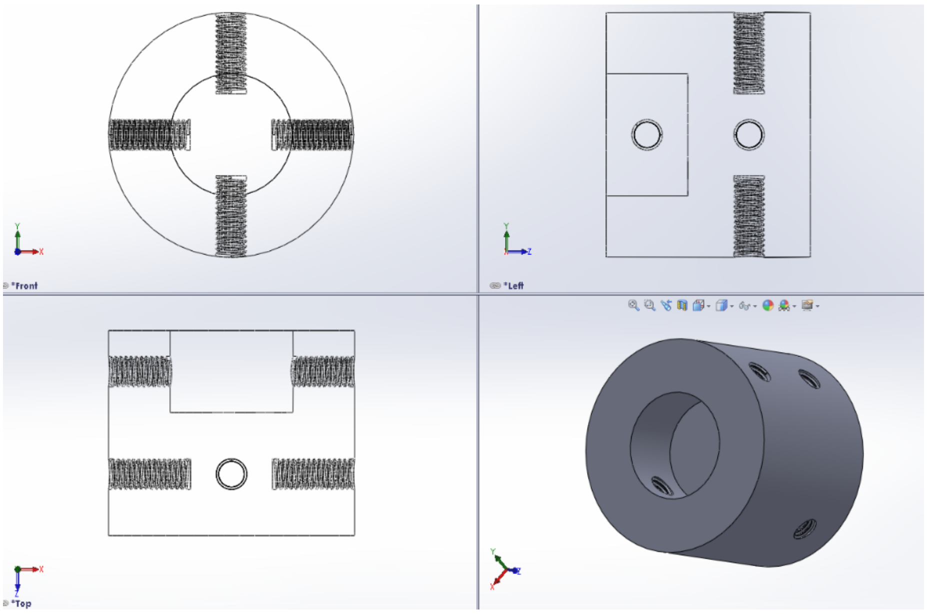
		regulators	to regulate the required air pressure		
12	4	Pneumatic controllers	digital pressure controller (FESTO, Esslingen am Neckar, Germany)	1	off-the-shelf
13	4	Belts	a fabric strap attached to pneumatic actuators	2	made-to-specification
14	1	SMARTWheels	An instrumented wheel that gives biomechanical factors of wheelchair propulsion	2	off-the-shelf
15	1	Fixture bars		11	off-the-shelf
16	1	Fixture connectors		12	off-the-shelf
17	1	Bearings	Timken, model number: 1108KLLB (wide inner ring ball bearings, with a single row in a deep groove)	6	off-the-shelf
18	2	Universal joints		2	off-the-shelf
19		Aluminium bars	1" in diameter	2	made-to-specification
20	2	Aluminium plates	1 cm thickness, with custom holes drilled into them	4	made-to-specification
21	1	Drums	radius 0.158 m and a mass of 26.4 kg	2	made-to-specification
22	1	Platform	wooden structure to support the wheelchair	1	made-to-specification
23	2	Guards	mass guards placed on the linear inertia systems and are built for users' safety	2	made-to-specification
24	4	Loadcells (transducers)	Interface inc. Arizona, USA. Model No: SML-	2	off-the-shelf

			50, Capacity: 50 lb, Serial Number: 230805. Loadcells were calibrated for being used in this system using a graded bottle, filled with water from 0 ml to 1000ml, with 50 ml increments.		
25	3	EON Icube mobile	EON Icube^{TM} Mobile which is an immersive VR system	1	off-the-shelf
26	3	Central computer		1	off-the-shelf
27	3	Cameras		2	off-the-shelf
28	3	Microphones		2	off-the-shelf
29	4	Labview	NI, Texas, USA. Labview 2012, Version 12.0f3n (32-Bit)	1	off-the-shelf
30	3	Data Acquisition (DAQ) Module	Data translation DT9800	1	off-the-shelf

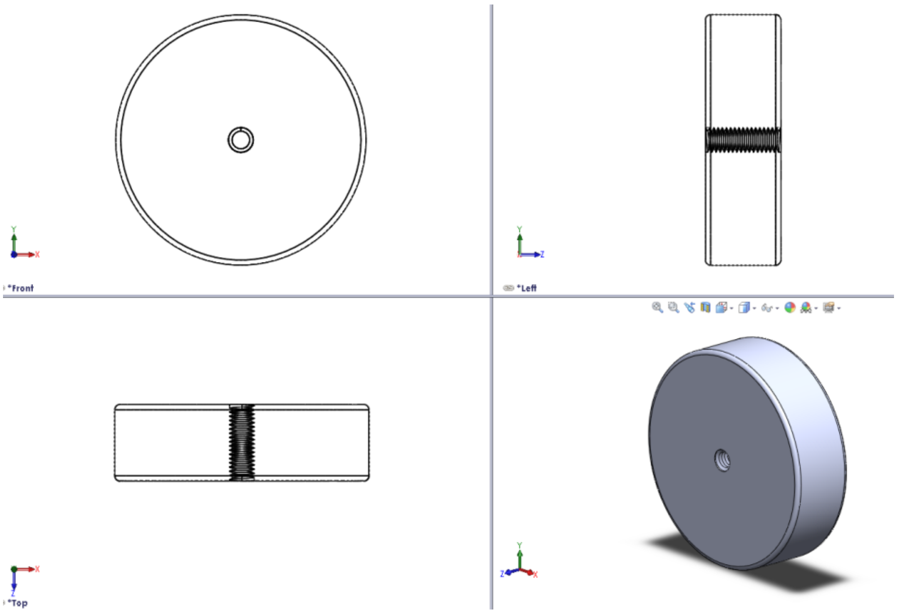
The following figure depicts the sketch of parts used in the linear inertia system.



(a)



(b)



(c)

Figure A-1 The sketch of different main parts of the linear inertia system: (a) the threaded bars, (b) the hubs that the threaded bars screw into them, (c) the disks that screw on the threaded bars

Figure **A-**0-1 shows the base wheelchair ergometer (including the hollow cylinder rollers) that the inertia system mounts on it. The structure depicted on the left side of this figure shows the fixture system (more details on Figure A-0-2) that is used to be easily removable to let the wheelchair/wheelchair user in, then fixes the wheelchair on the rollers, and is designed to provide a stable hold of the wheelchair both in straight-line propulsion and in manoeuvring.

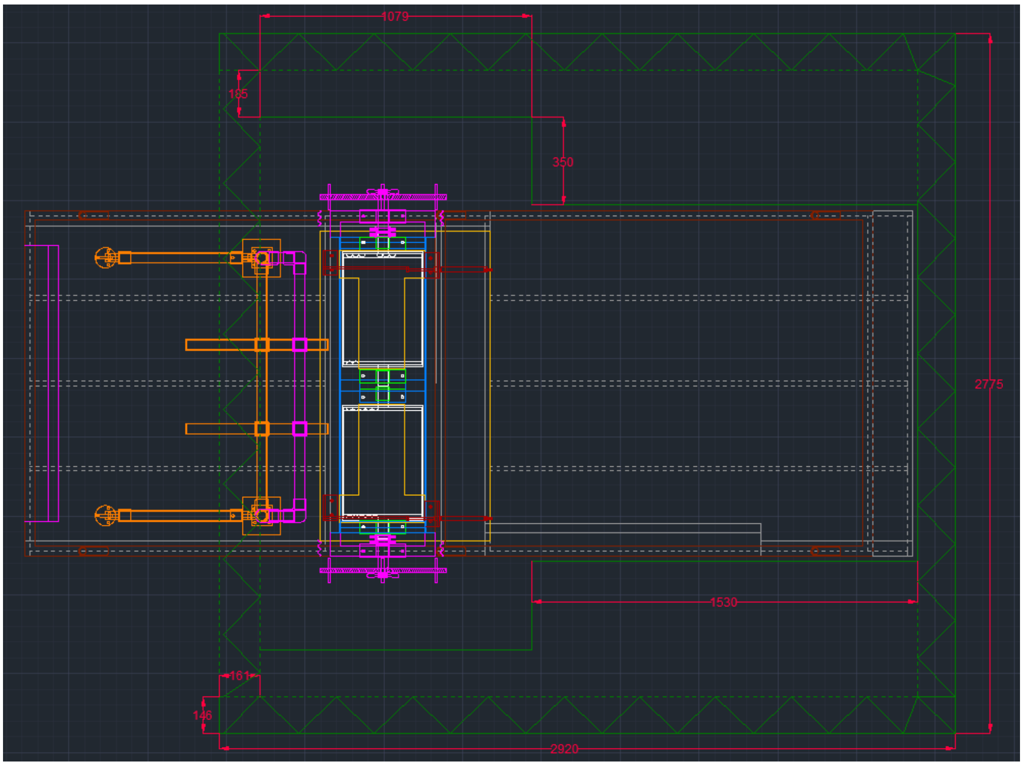


Figure A-0-1 The base wheelchair ergometer that the inertia systems mount on it

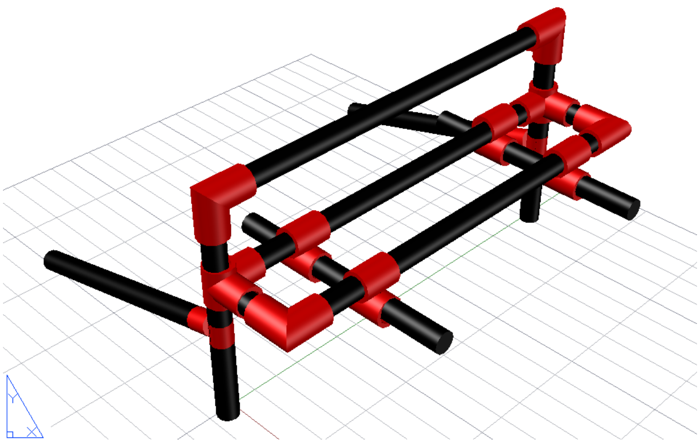
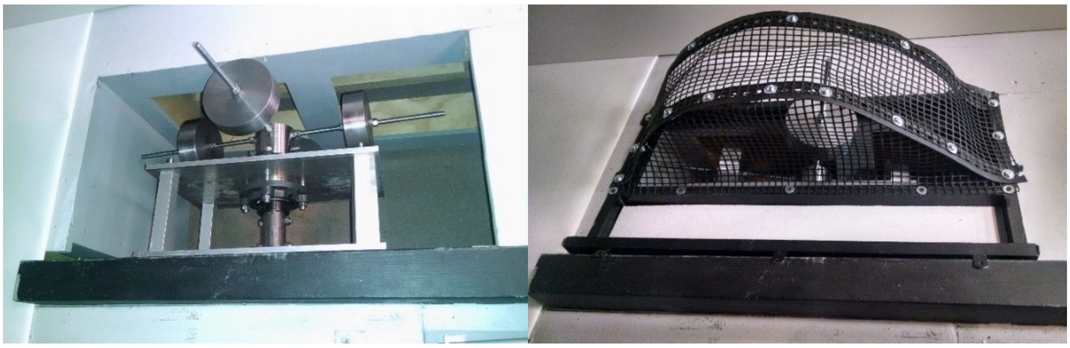


Figure A-0-2 The fixture system that fixes the wheelchair on the rollers.

The mass guard shown in Figure **0-3** was built for users' safety. The two inertia systems were mounted on two shaft extensions that were connected to the rollers using universal joints and bedding on the bearings (see Figure 0-3(a) for more details).



(a) (b)

Figure 0-3 (a) The linear inertia system, exposed and without protection, (b) the mass guard placed on the linear inertia system

Figure A-00-4 depicts a schematic of the wheelchair ergometer and the footprint of the Icube truss on it.

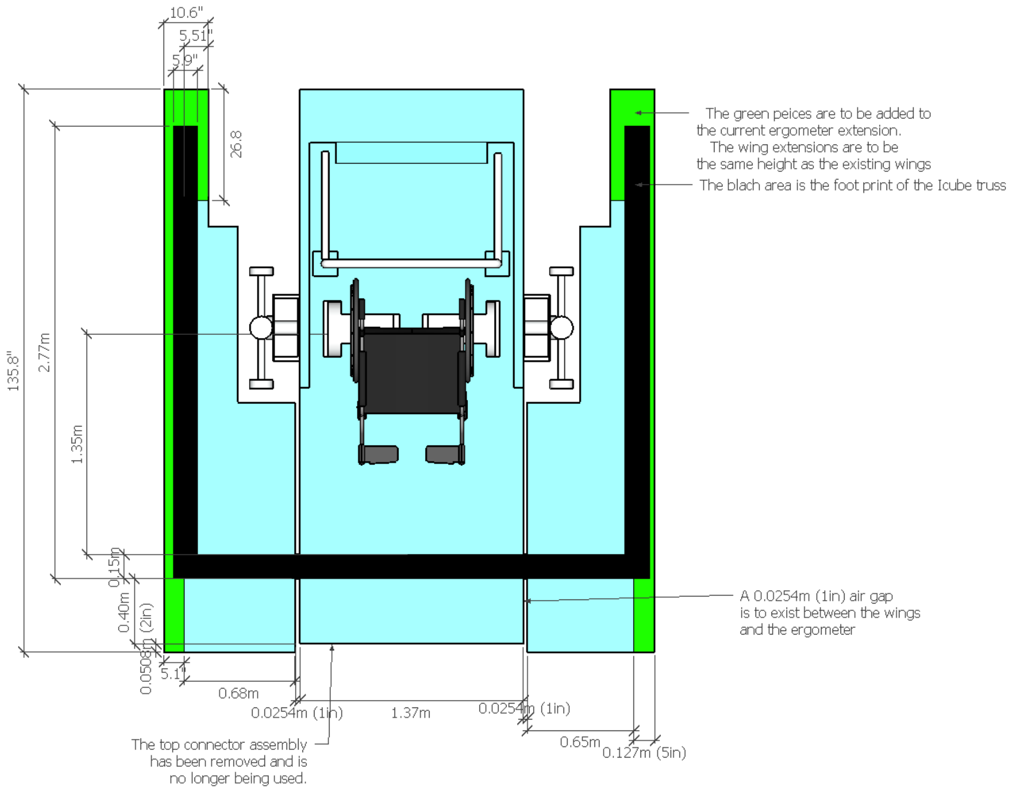


Figure A-00-4 Schematic of the wheelchair ergometer and the footprint of the Icube truss on it

The Table A-2 determines the distance the disk weights in the inertia systems need to have from their center of rotation in order to be adjusted for different 'participant+wheelchair' weights.

Table A-2 Table used to adjust the inertia systems for each participant

Mass of subject and participant, together (Kg)	Distance of the disks from the center (m)
56.16673377	0
56.28987552	0.01
56.65930078	0.02
57.27500954	0.03
58.1370018	0.04
59.24527757	0.05
60.59983685	0.06
62.20067963	0.07
64.04780591	0.08
66.1412157	0.09
68.48090899	0.1
71.06688578	0.11
73.89914608	0.12
76.97768989	0.13
80.3025172	0.14
83.87362801	0.15
87.69102233	0.16
91.75470015	0.17
96.06466148	0.18
100.6209063	0.19
105.4234346	0.2
110.4722465	0.21
115.7673418	0.22
121.3087207	0.23
127.096383	0.24
133.1303289	0.25
139.4105582	0.26
145.9370711	0.27
152.7098675	0.28
159.7289474	0.29
166.9943107	0.3
174.5059576	0.31
182.263888	0.32

Appendix B MOTION SICKNESS ASSESSMENT QUESTIONNAIRE (MSAQ)

This questionnaire is produced by [239]

Motion Sickness Assessment Questionnaire (MSAQ), Used for Training Sessions.

Instructions. Using the scale below, please rate how accurately the following statements describe your experience

Not at all Severely

0——1——2——3——4——5——6——7——8——9

1. I felt sick to my stomach ( )
2. I felt faint-like ( )
3. I felt annoyed/irritated ( )
4. I felt sweaty ( )
5. I felt queasy ( )
6. I felt lightheaded ( )
7. I felt drowsy ( )
8. I felt clammy/cold sweat ( )
9. I felt disoriented ( )
10. I felt tired/fatigued

- Overall fatigue ( ) - Arm fatigue ( )

11. I felt nauseated ( )
12. I felt hot/warm ( )
13. I felt dizzy ( )
14. I felt like I was spinning ( )
15. I felt as if I may vomit ( )
16. I felt uneasy ( )

17. If this is a training session, in general, how do you feel about your motion sickness relative to last session?

Very much worse Very much better

0 ---10

18. If this is a training session, in general, do you think you are ready to take the main experiments? ◯ Yes ◯ No

Motion Sickness Assessment Questionnaire (MSAQ), Used for Main Sessions.

Instructions. Using the scale below, please rate how accurately the following statements describe your experience

Not at all Severely

0——1——2——3——4——5——6——7——8——9

1. I felt sick to my stomach ( )
2. I felt faint-like ( )
3. I felt annoyed/irritated ( )
4. I felt sweaty ( )
5. I felt queasy ( )
6. I felt lightheaded ( )
7. I felt drowsy ( )
8. I felt clammy/cold sweat ( )
9. I felt disoriented ( )
10. I felt tired/fatigued ( )

- Overall fatigue ( ) - Arm fatigue ( )

11. I felt nauseated ( )
12. I felt hot/warm ( )
13. I felt dizzy ( )

14. I felt like I was spinning ( )

15. I felt as if I may vomit ( )

16. I felt uneasy ( )

Appendix C IGROUP PRESENCE QUESTIONNAIRE (IPQ)

This questionnaire is produced by Igroup, available from: http://www.igroup.org/pq/ipq/index.php. The wording of this questionnaire is minimally edited.

Please answer the questions by choosing the number that best describes your experience.

1. How aware were you of the real world surrounding you while navigating in the virtual world? (i.e. sounds, room temperature, other people, etc.)?

Extremely aware | moderately aware | Not aware at all

0	1	2	3	4	5	6

2. How real did the virtual world seem to you?

About as real as an imagined world | Indistinguishable from real world

0	1	2	3	4	5	6

3. I had a sense of acting in the virtual space, rather than operating something from outside.

Fully disagree | Fully agree

0	1	2	3	4	5	6

4. How much did your experience in the virtual environment seem consistent with your real world experience?

Not consistent | Moderately consistent | Very consistent

0	1	2	3	4	5	6

5. How real did the virtual world seem to you?

Not real at all | Completely real

0	1	2	3	4	5	6

6. I did not feel present in the virtual space.

Did not feel present | Felt present

0	1	2	3	4	5	6

7. I was not aware of the real environment.

Extremely aware Moderately aware Not aware at all

0	**1**	**2**	**3**	**4**	**5**	**6**

8. In the computer generated world I had a sense of "being there".

Fully disagree Fully agree

0	**1**	**2**	**3**	**4**	**5**	**6**

9. Somehow I felt that the virtual world surrounded me.

Fully disagree Fully agree

0	**1**	**2**	**3**	**4**	**5**	**6**

10. I felt present in the virtual space.

Fully disagree Fully agree

0	**1**	**2**	**3**	**4**	**5**	**6**

11. I still paid attention to the real environment.

Fully agree Fully disagree

0	**1**	**2**	**3**	**4**	**5**	**6**

12. The virtual world seemed more realistic than the real world.

Fully disagree Fully agree

0	**1**	**2**	**3**	**4**	**5**	**6**

13. I felt like I was just seeing pictures.

Fully agree Fully disagree

0	**1**	**2**	**3**	**4**	**5**	**6**

14. I was completely captivated by the virtual world.

Fully disagree Fully agree

0	**1**	**2**	**3**	**4**	**5**	**6**

15. How do you rate your trials in real world relative to your trials in virtual world?

Virtual reality was much easier the same Virtual reality was much harder

0 --5-- 10

16. How much similar were the forces you needed to apply to turn a given angle, i.e. 45 deg?

Virtual reality needed much less force the same Virtual reality needed much more force

0 --5--- 10

Appendix D C# CODE

This appendix includes the C# codes developed for interfacing VR to the other components.

MainWindow.xaml

```
<Window x:Class="ErgoInterface2.MainWindow"
        xmlns="http://schemas.microsoft.com/winfx/2006/xaml/presentation"
        xmlns:x="http://schemas.microsoft.com/winfx/2006/xaml"
        xmlns:d="http://schemas.microsoft.com/expression/blend/2008"
        xmlns:mc="http://schemas.openxmlformats.org/markup-compatibility/2006"
        xmlns:local="clr-namespace:ErgoInterface2.ViewModels"
        xmlns:sparrow="http://sparrowtoolkit.codeplex.com/wpf"
        xmlns:sharpGL="clr-namespace:SharpGL.WPF;assembly=SharpGL.WPF"
        xmlns:xctk="http://schemas.xceed.com/wpf/xaml/toolkit"
        xmlns:conv="clr-namespace:ErgoInterface2.ValueConverters"
        xmlns:winformchart="clr-
namespace:System.Windows.Forms.DataVisualization.Charting;assembly=System
.Windows.Forms.DataVisualization"
        mc:Ignorable="d"
        Title="MFP Lab Ergometer Interface (v2)" Height="540" Width="960"
        Background="{DynamicResource {x:Static
SystemColors.WindowBrushKey}}">
    <Window.DataContext>
        <local:Main></local:Main>
    </Window.DataContext>
```

```
<Window.Resources>
    <conv:Checkmark x:Key="CheckmarkConverter"/>
    <conv:ConnectDisconnect x:Key="ConnectDisconnectConverter"/>
</Window.Resources>
<Grid Margin="5">
    <Grid.ColumnDefinitions>
        <ColumnDefinition Width="auto"/>
        <ColumnDefinition Width="*"/>
    </Grid.ColumnDefinitions>
    <Grid.RowDefinitions>
        <RowDefinition Height="*"/>
        <RowDefinition Height="*"/>
    </Grid.RowDefinitions>
    <StackPanel HorizontalAlignment="Left" Grid.Row="0" Grid.Column="0"
Grid.RowSpan="2" Margin="0,0,5,5">
        <GroupBox Header="Configuration">
            <Grid>
                <Grid.ColumnDefinitions>
                    <ColumnDefinition/>
                    <ColumnDefinition Width="80"/>
                </Grid.ColumnDefinitions>
                <Grid.RowDefinitions>
                    <RowDefinition Height="*"/>
                    <RowDefinition Height="*"/>
                    <RowDefinition Height="*"/>
```

```
<RowDefinition Height="*"/>
<RowDefinition Height="*"/>
<RowDefinition Height="*"/>
</Grid.RowDefinitions>
<Label Content="Wheelspan (m)" Grid.Row="0" Grid.Column="0"
VerticalAlignment="Center"></Label>
<xctk:DoubleUpDown Value="{Binding Config.WheelSpan}"
Increment=".01" Grid.Row="0" Grid.Column="1" Margin="0,5" />
<Label Content="Left Calibration" Grid.Row="1" Grid.Column="0"
VerticalAlignment="Center"></Label>
<xctk:DoubleUpDown Value="{Binding Config.LeftCalibration}"
Increment=".001" Grid.Row="1" Grid.Column="1" Margin="0,5" />
<Label Content="Right Calibration" Grid.Row="2" Grid.Column="0"
VerticalAlignment="Center"></Label>
<xctk:DoubleUpDown Value="{Binding Config.RightCalibration}"
Increment=".001" Grid.Row="2" Grid.Column="1" Margin="0,5" />


<Label Content="Participant's Weight(kg)" Grid.Row="3"
Grid.Column="0" VerticalAlignment="Center"></Label>
<xctk:DoubleUpDown Value="{Binding Config.WeightOfPerson}"
Increment=".001" Grid.Row="3" Grid.Column="1" Margin="0,5" />


<Label Content="I (Wheelchair + Person)" Grid.Row="4"
Grid.Column="0" VerticalAlignment="Center"></Label>
<xctk:DoubleUpDown Value="{Binding Config.MOIWheelchair}"
Increment=".001" Grid.Row="4" Grid.Column="1" Margin="0,5" />
```

```
<Label Content="Wheel Radius" Grid.Row="5" Grid.Column="0" VerticalAlignment="Center"></Label>
<xctk:DoubleUpDown Value="{Binding Config.WheelRadius}" Increment=".001" Grid.Row="5" Grid.Column="1" Margin="0,5" />
</Grid>
</GroupBox>
<GroupBox Header="Ergometer">
<StackPanel>
<Button Content="{Binding Path=ErgometerConnected, Converter={StaticResource ConnectDisconnectConverter}}" Margin="0,5" Command="{Binding ConnectToErgometer}"></Button>
<Button Content="Reset" Margin="0,0,0,5" IsEnabled="{Binding ErgometerConnected}" Command="{Binding ResetWheelchair}"></Button>
<Button Content="Apply Brakes" Margin="0,0,0,5" IsEnabled="{Binding ErgometerConnected}" Command="{Binding ApplyBrakes}"></Button>
<Grid>
<Grid.ColumnDefinitions>
<ColumnDefinition Width="auto"/>
<ColumnDefinition Width="*"/>
</Grid.ColumnDefinitions>
<Label Content="Filter:" Grid.Row="0" Grid.Column="0"/>
<ComboBox Grid.Row="0" Grid.Column="1" ItemsSource="{Binding Config.FilterSelectors}" SelectedValue="{Binding Config.SelectedFilter}"/>
</Grid>
</StackPanel>
```

```
        </GroupBox>
        <GroupBox Header="Net">
          <Grid>
            <Grid.ColumnDefinitions>
              <ColumnDefinition Width="*"/>
              <ColumnDefinition Width="*"/>
              <ColumnDefinition Width="*"/>
              <ColumnDefinition Width="*"/>
            </Grid.ColumnDefinitions>
            <Grid.RowDefinitions>
              <RowDefinition Height="*"/>
              <RowDefinition Height="*"/>
              <RowDefinition Height="*"/>
            </Grid.RowDefinitions>
            <Label Content="EON" Grid.Row="0" Grid.Column="0"
VerticalAlignment="Center"></Label>
            <TextBox Text="{Binding Config.EONIPPort.IPString}"
Grid.Row="0" Grid.Column="1" Margin="5,5"/>
            <xctk:IntegerUpDown Value="{Binding Config.EONIPPort.Port}"
Increment="1" Maximum="65535" Minimum="1" Grid.Row="0"
Grid.Column="2" Margin="0,5"/>
            <Label Content="{Binding Path=Config.EONIPPort.IPOK,
Converter={StaticResource CheckmarkConverter}}" Grid.Row="0"
Grid.Column="3" VerticalAlignment="Center"></Label>
            <Label Content="LabVIEW" Grid.Row="1" Grid.Column="0"
VerticalAlignment="Center"></Label>
```

```
                <TextBox Text="{Binding Config.LabVIEWIPPort.IPString}" Grid.Row="1" Grid.Column="1" Margin="5,5"/>
                <xctk:IntegerUpDown Value="{Binding Config.LabVIEWIPPort.Port}" Increment="1" Maximum="65535" Minimum="1" Grid.Row="1" Grid.Column="2" Margin="0,5"/>
                <Label Content="{Binding Path=Config.LabVIEWIPPort.IPOK, Converter={StaticResource CheckmarkConverter}}" Grid.Row="1" Grid.Column="3" VerticalAlignment="Center"></Label>
                <Label Content="Pressure Port" Grid.Row="2" Grid.Column="0" Grid.ColumnSpan="2"></Label>
                <xctk:IntegerUpDown Value="{Binding Config.PressureControlPort}" Increment="1" Maximum="65535" Minimum="1000" Grid.Row="2" Grid.Column="2" Margin="0,5"/>
            </Grid>
        </GroupBox>
    </StackPanel>
    <Grid Grid.Row="0" Grid.Column="1" Margin="0,0,0,5">
        <Grid.ColumnDefinitions>
            <ColumnDefinition Width="*"/>
            <ColumnDefinition Width="200"/>
        </Grid.ColumnDefinitions>
        <Grid.RowDefinitions>
            <RowDefinition Height="*"/>
        </Grid.RowDefinitions>
        <sharpGL:OpenGLControl x:Name="CoursePreview" OpenGLVersion="OpenGL4_4" Grid.Row="0" Grid.Column="0" RenderContextType="FBO" OpenGLDraw="CoursePreview_OpenGLDraw" Resized="CoursePreview_Resized"></sharpGL:OpenGLControl>
```

```
<GroupBox Grid.Row="0" Grid.Column="1" Header="Telemetry">
  <Grid>
    <Grid.ColumnDefinitions>
      <ColumnDefinition Width="auto"/>
      <ColumnDefinition Width="*"/>
    </Grid.ColumnDefinitions>
    <Grid.RowDefinitions>
      <RowDefinition Height="auto"/>
      <RowDefinition Height="auto"/>
      <RowDefinition Height="auto"/>
      <RowDefinition Height="auto"/>
      <RowDefinition Height="auto"/>
      <RowDefinition Height="auto"/>
      <RowDefinition Height="auto"/>
      <RowDefinition Height="auto"/>
      <RowDefinition Height="*"/>
    </Grid.RowDefinitions>

    <TextBlock Text="Force (Left):" Grid.Row="0" Grid.Column="0"
Margin="0,0,5,5" />
    <ProgressBar Grid.Row="1" Grid.Column="1" Minimum="0"
Maximum="5" Margin="0,0,0,5" Value="{Binding LeftForce}" />

    <TextBlock Text="Force (Right):" Grid.Row="1" Grid.Column="0"
Margin="0,0,5,5" />
```

```
<ProgressBar Grid.Row="0" Grid.Column="1" Minimum="0"
Maximum="5" Margin="0,0,0,5" Value="{Binding RightForce}" />

<TextBlock Text="Pressure (Left):" Grid.Row="2" Grid.Column="0"
Margin="0,0,5,5" />
<ProgressBar Grid.Row="2" Grid.Column="1" Minimum="0"
Maximum="1" Margin="0,0,0,5" Value="{Binding LeftPressure}" />

<TextBlock Text="Pressure (Right):" Grid.Row="3"
Grid.Column="0" Margin="0,0,5,5" />
<ProgressBar Grid.Row="3" Grid.Column="1" Minimum="0"
Maximum="1" Margin="0,0,0,5" Value="{Binding RightPressure}" />

<TextBlock Text="Orientation (Q0):" Grid.Row="4"
Grid.Column="0" Margin="0,0,5,5" />
<TextBlock Text="{Binding Orientation[0]}" Grid.Row="4"
Grid.Column="1" Margin="0,0,0,5" />

<TextBlock Text="Orientation (Q1):" Grid.Row="5"
Grid.Column="0" Margin="0,0,5,5" />
<TextBlock Text="{Binding Orientation[1]}" Grid.Row="5"
Grid.Column="1" Margin="0,0,0,5" />

<TextBlock Text="Orientation (Q2):" Grid.Row="6"
Grid.Column="0" Margin="0,0,5,5" />
<TextBlock Text="{Binding Orientation[2]}" Grid.Row="6"
Grid.Column="1" Margin="0,0,0,5" />
```

```
            <TextBlock Text="Orientation (Q3):" Grid.Row="7" Grid.Column="0" Margin="0,0,5,5" />
            <TextBlock Text="{Binding Orientation[3]}" Grid.Row="7" Grid.Column="1" Margin="0,0,0,5" />

            <!--TextBlock Text="Time (s):" Grid.Row="8" Grid.Column="0" Margin="0,0,5,5" />
            <ProgressBar Grid.Row="8" Grid.Column="1" Minimum="0" Maximum="10000" Margin="0,0,0,5" Value="{Binding Time}" /-->

          </Grid>
        </GroupBox>
      </Grid>
      <WindowsFormsHost x:Name="host" Grid.Row="1" Grid.Column="1">
        <winformchart:Chart x:Name="VelocityChart" Dock="Fill">
          <winformchart:Chart.ChartAreas>
            <winformchart:ChartArea/>
          </winformchart:Chart.ChartAreas>
        </winformchart:Chart>
      </WindowsFormsHost>
    </Grid>
</Window>
```

Main.cs

```
using ErgoInterface2.Hardware;
```

```
using ErgoInterface2.Filter;
using ErgoInterface2.Models;
using System;
using System.Collections.Generic;
using System.Collections.ObjectModel;
using System.Collections.Specialized;
using System.ComponentModel;
using System.Linq;
using System.Net;
using System.Text;
using System.Threading.Tasks;
using SharpGL;
using System.Net.Sockets;
using System.Windows.Input;
using System.Windows;
using System.Windows.Threading;
using System.Windows.Forms.DataVisualization.Charting;

namespace ErgoInterface2.ViewModels
{
    class Main : ViewModelBase
    {
        public ObservableQueue<Velocity> _VelocityData { get; set; }
        public ObservableQueue<Velocity> VelocityData
        {
```

```
    get { return _VelocityData; }
    set
    {
        if (value != _VelocityData)
        {
            _VelocityData = value;
            NotifyPropertyChanged("VelocityData");
        }
    }
}

private Wheelchair Wheelchair;

private Config _Config;
public Config Config
{
    get { return _Config; }
    set
    {
        if(value != _Config)
        {
            _Config = value;
            NotifyPropertyChanged("Config");
        }
    }
```

```
}

private bool _ErgometerConnected = false;
public bool ErgometerConnected
{
    get { return _ErgometerConnected; }
    set
    {
        if (value != _ErgometerConnected)
        {
            _ErgometerConnected = value;
            NotifyPropertyChanged("ErgometerConnected");
        }
    }
}

private double _LeftForce = 0;
public double LeftForce
{
    get { return _LeftForce; }
    set
    {
        if (value != _LeftForce)
        {
            _LeftForce = value;
```

```
            NotifyPropertyChanged("LeftForce");
        }
    }
}

private double _RightForce = 0;
public double RightForce
{
    get { return _RightForce; }
    set
    {
        if (value != _RightForce)
        {
            _RightForce = value;
            NotifyPropertyChanged("RightForce");
        }
    }
}

private double _LeftPressure = 0;
public double LeftPressure
{
    get { return _LeftPressure; }
    set
    {
```

```
        if (value != _LeftPressure)
        {
            _LeftPressure = value;
            NotifyPropertyChanged("LeftPressure");
        }
    }
}

private double _RightPressure = 0;
public double RightPressure
{
    get { return _RightPressure; }
    set
    {
        if (value != _RightPressure)
        {
            _RightPressure = value;
            NotifyPropertyChanged("RightPressure");
        }
    }
}

private double[] _Orientation = new double[4];
public double[] Orientation
{
```

```
get { return _Orientation; }
set
{
    if (value != _Orientation)
    {
        _Orientation = value;
        NotifyPropertyChanged("Orientation");
    }
}



/*        private double _Time =0;
       public double Time
   // { get; set; }
       {
           get { return _Time; }
           set
            {
               if (value != Time)
               {
                   _Time = value;
                   NotifyPropertyChanged("Time");
               }
           }
```

```
        }*/
    }

    private bool _brakesOn = false;




    private Ergometer Ergometer;
    private IMU IMU;
    private UdpClient EonClient;
    private UdpClient LabVIEWClient;
    private UdpClient PressureControlHost;
    private IPEndPoint PressureControlEndPoint;

    private Dispatcher uiDispatcher;

    private GLRenderer Renderer;

    public Main()
    {
        uiDispatcher = Dispatcher.CurrentDispatcher;

        Ergometer = new Ergometer();
        Ergometer.Connected += Ergometer_Connected;
```

```
Ergometer.Disconnected += Ergometer_Disconnected;
Ergometer.NewData += Ergometer_NewData;

IMU = new IMU();
IMU.NewData += IMU_NewData;

VelocityData = new ObservableQueue<Velocity>();

Config = new Config();
Config.EONIPPort.Updated += EONIPPort_Updated;
Config.LabVIEWIPPort.Updated += LabVIEWIPPort_Updated;
Config.PressureControlUpdated += Config_PressureControlUpdated;
Config.SelectedFilterUpdated += Config_SelectedFilterUpdated;
Config.WheelSpanUpdated += Config_WheelchairSettingUpdated;
Config.LeftCalibrationUpdated += Config_WheelchairSettingUpdated;
Config.RightCalibrationUpdated += Config_WheelchairSettingUpdated;
Config.WeightOfPersonUpdated += Config_WheelchairSettingUpdated;
Config.MOIWheelchairUpdated += Config_WheelchairSettingUpdated;
Config.WheelRadiusUpdated += Config_WheelchairSettingUpdated;
Config.Initialize();

Wheelchair = new Wheelchair();



//double DeltaT = Wheelchair.dt;
```

```
            Wheelchair.Reset(Config.WheelSpan);
        }

        private void IMU_NewData(IMUDataPacket dp)
        {
            uiDispatcher.Invoke((Action)delegate ()
            {
                Orientation = dp.q;
            });
        }

        private void Config_WheelchairSettingUpdated(double data)
        {
            Properties.Settings.Default.Save();
        }

        public void Initialize(MainWindow window)
        {
            InitializeChart(window);
            Renderer = new GLRenderer(window);
            Renderer.SetWheelchairLocation(Wheelchair.centerPosition.x,
Wheelchair.centerPosition.y, Wheelchair.theta);
        }
```

```
// use WinForms charts because WPF doesn't have a good, free charting library :/
private Chart chart;
private void InitializeChart(Window window)
{
    chart = window.FindName("VelocityChart") as Chart;
    chart.Palette = ChartColorPalette.Bright;

    // add a legend
    Legend legend = new Legend();
    chart.Legends.Add(legend);

    // add axis labels
    chart.ChartAreas[0].AxisX.Title = "Time, t (s)";
    chart.ChartAreas[0].AxisX.LabelStyle.Format = "F1";
    chart.ChartAreas[0].AxisY.Title = "Velocity, v (m/s)";
    chart.ChartAreas[0].AxisY.LabelStyle.Format = "F2";

    // add the data series
    chart.Series.Clear();

    Series leftSeries = new Series("Left");
    leftSeries.ChartType = SeriesChartType.FastLine;
    chart.Series.Add(leftSeries);
```

```
        Series rightSeries = new Series("Right");
        rightSeries.ChartType = SeriesChartType.FastLine;
        chart.Series.Add(rightSeries);
    }

    private void UpdateEON()
    {
        if(EonClient != null)
        {
            // correct EON's angle
            double degTheta = -180 * Wheelchair.theta / Math.PI;
            while (degTheta >= 360)
                degTheta -= 360;
            while (degTheta < 0)
                degTheta += 360;

            // build the packet
            string packetStr = (Wheelchair.centerPosition.x).ToString() + "#" +
(Wheelchair.centerPosition.y).ToString() + "#" + degTheta.ToString();
            byte[] packet = System.Text.Encoding.ASCII.GetBytes(packetStr);

            // and send it
            EonClient.Send(packet, packet.Length);
        }
```

```
        }

        private void UpdateLabVIEW(ErgoDataPacket dp)
        {
            if(LabVIEWClient != null)
            {
                // TODO: detect pylon hits!
                int pylonHits = 0;
//              double DeltaT = 0;
                // convert the velocities to bytes
                List<Byte[]> parts = new List<byte[]>();
                parts.Add(BitConverter.GetBytes(dp.t));
                parts.Add(BitConverter.GetBytes(dp.velocityLeft));
                parts.Add(BitConverter.GetBytes(dp.velocityRight));
                parts.Add(BitConverter.GetBytes(Orientation[0]));
                parts.Add(BitConverter.GetBytes(Orientation[1]));
                parts.Add(BitConverter.GetBytes(Orientation[2]));
                parts.Add(BitConverter.GetBytes(Orientation[3]));
                parts.Add(BitConverter.GetBytes(LeftForce));
                parts.Add(BitConverter.GetBytes(RightForce));
                parts.Add(BitConverter.GetBytes(pylonHits));
                parts.Add(BitConverter.GetBytes(Wheelchair.centerPosition.x));
                parts.Add(BitConverter.GetBytes(Wheelchair.centerPosition.y));
                parts.Add(BitConverter.GetBytes(Wheelchair.theta));
                // fix their endianness
```

```
if (BitConverter.IsLittleEndian)
{
    for(int i = 0; i < parts.Count; i++)
    {
        Array.Reverse(parts[i]);
    }
}

int numBytes = 0;
for(int i = 0; i < parts.Count; i++)
{
    numBytes += parts[i].Length;
}

// pack it into one array
Byte[] sendBytes = new Byte[numBytes];
int accumLength = 0;
for(int i = 0; i < parts.Count; i++)
{
    Array.Copy(parts[i], 0, sendBytes, accumLength, parts[i].Length);
    accumLength += parts[i].Length;
}

// and send!
int sent = LabVIEWClient.Send(sendBytes, sendBytes.Length);
```

```
            //System.Diagnostics.Debug.WriteLine(String.Format("[{0}]  Sent  {1}
bytes to labview!", DateTime.Now.ToLongTimeString(), sent));
        }
    }

    private void Ergometer_NewData(ErgoDataPacket dp)
    {
        try
        {
            // process the wheelchair
            Wheelchair.Update(ref dp, Config);

            // fire the packets off
            UpdateEON();
            UpdateLabVIEW(dp);

            // update the UI
            uiDispatcher.Invoke((Action)delegate ()
            {
                // update the display
                Renderer.SetWheelchairLocation(Wheelchair.centerPosition.x,
Wheelchair.centerPosition.y, Wheelchair.theta);

                // update the chart
                if (chart != null && chart.Series != null && chart.Series.Count == 2)
                {
```

```
// if we get too many points, just clear the chart
/*if (chart.Series[0].Points.Count >= 1000)
{
    chart.Series[0].Points.Clear();
    chart.Series[1].Points.Clear();
}*/
// if we get too many points, start trimming the start
if(chart.Series[0].Points.Count >= 750)
{
    chart.Series[0].Points.RemoveAt(0);
    chart.Series[1].Points.RemoveAt(0);
}

// add the points
chart.Series[0].Points.AddXY(dp.t, dp.velocityLeft);
chart.Series[1].Points.AddXY(dp.t, dp.velocityRight);

// only look at the last 10 seconds
chart.ChartAreas[0].AxisX.Maximum = dp.t;
chart.ChartAreas[0].AxisX.Minimum = dp.t - 10;

// scale the y-axis appropriately
double minLeft =
chart.Series[0].Points.FindMinByValue("Y1").YValues[0];
double minRight =
chart.Series[1].Points.FindMinByValue("Y1").YValues[0];
```

```
                double min = Math.Min(minLeft, minRight);
                if (min > 0) min = 0;

                double maxLeft =
chart.Series[0].Points.FindMaxByValue("Y1").YValues[0];
                double maxRight =
chart.Series[1].Points.FindMaxByValue("Y1").YValues[0];
                double max = Math.Max(maxLeft, maxRight);
                if (max < 0) max = 0;

                if(min == 0 && max == 0)
                {
                    min = -2;
                    max = 2;
                }

                chart.ChartAreas[0].AxisY.Minimum = min;
                chart.ChartAreas[0].AxisY.Maximum = max;
            }

            // now update the force readings
            LeftForce = dp.forceLeft;
            RightForce = dp.forceRight;
        });
    }
    catch (Exception)
```

```
    { }
}

private void Ergometer_Connected()
{
    ErgometerConnected = true;
}

private void Ergometer_Disconnected()
{
    ErgometerConnected = false;
}

private void EONIPPort_Updated(bool ok, IPEndPoint endPoint)
{
    if (!ok) return;
    EonClient = new UdpClient();
    EonClient.Connect(endPoint);
}

private void LabVIEWIPPort_Updated(bool ok, IPEndPoint endPoint)
{
    if (!ok) return;
    LabVIEWClient = new UdpClient();
    LabVIEWClient.Connect(endPoint);
```

```
        }

        private void Config_PressureControlUpdated(int port)
        {
            try
            {
                PressureControlEndPoint = new IPEndPoint(IPAddress.Any, port);
                PressureControlHost = new UdpClient(PressureControlEndPoint);
                PressureControlHost.BeginReceive(new
AsyncCallback(PressureReceiveCallback), null);
            }
            catch(Exception e)
            {
                MessageBox.Show(e.Message, "Error", MessageBoxButton.OK,
MessageBoxImage.Error);
            }
        }

        private void PressureReceiveCallback(IAsyncResult ar)
        {
            byte[] receivedBytes = PressureControlHost.EndReceive(ar, ref
PressureControlEndPoint);

            if (receivedBytes.Length == 16)
            {
                if(BitConverter.IsLittleEndian)
```

```
{
    Array.Reverse(receivedBytes);
}

double leftPressure = BitConverter.ToDouble(receivedBytes, 0);
double rightPressure = BitConverter.ToDouble(receivedBytes,
sizeof(double));

            leftPressure = Math.Abs(leftPressure);
            rightPressure = Math.Abs(rightPressure);

leftPressure = Math.Max(0, Math.Min(1, leftPressure));
rightPressure = Math.Max(0, Math.Min(1, rightPressure));

// update the ergometer
Ergometer.SetPressure((float)leftPressure, (float)rightPressure);

// update the UI
try
{
    uiDispatcher.Invoke((Action)delegate ()
    {
        LeftPressure = leftPressure;
        RightPressure = rightPressure;
    });
```

```
            }
            catch { }
        }

        PressureControlHost.BeginReceive(new
AsyncCallback(PressureReceiveCallback), null);
    }

    private void Config_SelectedFilterUpdated(string filterName)
    {
        switch (filterName)
        {
            case "None":
                Ergometer.SetFilters(new PassThruDataFilter(), new
PassThruDataFilter());
                break;
            case "MWA-5":
                Ergometer.SetFilters(new MWADataFilter(5, 0), new
MWADataFilter(5, 0));
                break;
        }
    }

    public ICommand ConnectToErgometer
    {
        get
```

```
{
    return new DelegateCommand<object>(context =>
    {
        try
        {
            if(ErgometerConnected)
            {
                // disconnect
                IMU.Disconnect();
                Ergometer.StopCollection();
                Ergometer.Disconnect();
            }
            else
            {
                // connect
                Ergometer.Connect();
                Ergometer.StartCollection();

                IMU.Connect();

                EONIPPort_Updated(true, Config.EONIPPort.IPPort);
                LabVIEWIPPort_Updated(true, Config.LabVIEWIPPort.IPPort);
            }
        }
        catch (Exception e)
```

```
                {
                    MessageBox.Show(e.Message, "Error", MessageBoxButton.OK,
MessageBoxImage.Error);
                }
            });
        }
    }

    public ICommand ResetWheelchair
    {
        get
        {
            return new DelegateCommand<object>(context =>
            {
                Wheelchair.Reset(Config.WheelSpan);
                Renderer.ResetTrail();
                Renderer.SetWheelchairLocation(Wheelchair.centerPosition.x,
Wheelchair.centerPosition.y, Wheelchair.theta);
            });
        }
    }

    public ICommand ApplyBrakes
    {
        get
        {
```

```
            return new DelegateCommand<object>(context =>
            {
                    if(_brakesOn) Ergometer.SetPressure(0, 0);
                    else Ergometer.SetPressure(1, 1);
                    _brakesOn = !_brakesOn;
            });
        }
    }
  }
}
```

Wheelchair.cs

```
using ErgoInterface2.Hardware;
using ErgoInterface2.ViewModels;
using System;
using System.Collections.Generic;
using System.Linq;
using System.Text;
using System.Threading.Tasks;
using SharpGL.SceneGraph;

namespace ErgoInterface2.Models
{
  class Wheelchair
  {
```

```
private ErgoDataPacket lastPoint = new ErgoDataPacket();
private Vector2 forward = new Vector2(0, 1);

public Vector2 leftPosition = new Vector2(0, 0);
public Vector2 rightPosition = new Vector2(0, 0);
public Vector2 centerPosition = new Vector2(0, 0);
public double theta = 0;

public double lastVelocityLeft = 0, lastVelocityRight = 0;
double coefficient = 0;
public void Update(ref ErgoDataPacket dp, Config config)
{
    // process the data
    double dt = dp.t - lastPoint.t;

    // calibrate
    dp.velocityLeft *= config.LeftCalibration;
    dp.velocityRight *= config.RightCalibration;

    // integrate the position
    double forwardVelocity = (dp.velocityLeft + dp.velocityRight) / 2;

    double MOIRoller = 0.66;
    double mass = config.WeightOfPerson;
    double moiwheel = 0.115;
```

```
double inertiaFraction = (moiwheel+ MOIRoller)/config.MOIWheelchair ;
double radiusFraction = Math.Pow(config.WheelSpan/config.WheelRadius, 2)/2;
if (dp.velocityLeft != 0 || dp.velocityRight != 0)
{
   if (dp.velocityLeft * dp.velocityRight > 0)
   {
      coefficient = 0.5 + 0.5 * Math.Min(Math.Abs(dp.velocityLeft), Math.Abs(dp.velocityRight)) / Math.Max(Math.Abs(dp.velocityLeft), Math.Abs(dp.velocityRight));
   }
   else
   {
      coefficient =0.5- 0.5 * Math.Min(Math.Abs(dp.velocityLeft), Math.Abs(dp.velocityRight)) / Math.Max(Math.Abs(dp.velocityLeft), Math.Abs(dp.velocityRight));

   }

   dp.velocityLeft *= (inertiaFraction * radiusFraction + (1 - (inertiaFraction * radiusFraction)) * coefficient);
   dp.velocityRight *= (inertiaFraction * radiusFraction + (1 - (inertiaFraction * radiusFraction)) * coefficient);
}

double accelerationLeft = (dp.velocityLeft - lastVelocityLeft) / dt;
double accelerationRight = (dp.velocityRight - lastVelocityRight) / dt;
```

```
double accelerationTerm = accelerationRight - accelerationLeft;


// update the rotation
double deltaTheta = ((dp.velocityRight - dp.velocityLeft) / config.WheelSpan) * dt;
theta += deltaTheta;

// rotate it!
Matrix2x2 rotationMatrix = new Matrix2x2(Math.Cos(theta), -1 * Math.Sin(theta), Math.Sin(theta), Math.Cos(theta));
Vector2 forwards = rotationMatrix * forward;

// move it!
 leftPosition += (forwards * dp.velocityLeft) * dt;
 rightPosition += (forwards * dp.velocityRight) * dt;
 centerPosition = (leftPosition + rightPosition) / 2;

// store the old point
lastPoint = dp;
lastVelocityLeft = dp.velocityLeft;
lastVelocityRight = dp.velocityRight;
}

public void Reset(double wheelSpan)
```

```
        {
            lastPoint.velocityLeft = 0;
            lastPoint.velocityRight = 0;
            lastPoint.forceLeft = 0;
            lastPoint.forceRight = 0;

            centerPosition = new Vector2(1.7f, 5f);

            leftPosition = centerPosition + new Vector2(-wheelSpan / 2, 0);
            rightPosition = centerPosition + new Vector2(wheelSpan / 2, 0);
            theta = Math.PI;
        }
    }
}
```

Ergometer.cs

```
using ErgoInterface2.Filter;
using HidLibrary;
using OpenLayers.Base;
using System;
using System.Collections.Generic;
using System.Linq;
using System.Text;
using System.Threading.Tasks;
```

```
namespace ErgoInterface2.Hardware
{
    class Ergometer
    {
        public delegate void DataConnectedHandler();
        public delegate void DataNewDataHandler(ErgoDataPacket dp);
        public delegate void DataDisconnectedHandler();
        public event DataConnectedHandler Connected;
        public event DataNewDataHandler NewData;
        public event DataDisconnectedHandler Disconnected;

        public IDataFilter vlFilter = new PassThruDataFilter(), vrFilter = new
PassThruDataFilter();

        private readonly int VendorID = 0x16C0;
        private readonly int ProductID = 0x0486;
        public HidDevice hidDevice;

        public Device analogDevice;
        public AnalogInputSubsystem ain;
        public DeviceMgr analogManager;

        private double t = 0;
        private bool collecting = false;
```

```
public Ergometer()
{

}

~Ergometer()
{
    Disconnect();
}

public void SetFilters(IDataFilter vlFilter, IDataFilter vrFilter)
{
    this.vlFilter = vlFilter;
    this.vrFilter = vrFilter;
}

public void Connect()
{
    hidDevice = HidDevices.Enumerate(VendorID,
ProductID).FirstOrDefault();

    if (hidDevice != null)
    {
        // attach events
        hidDevice.Inserted += device_Inserted;
```

```
            hidDevice.Removed += device_Removed;
            hidDevice.MonitorDeviceEvents = true;

            // open the device
            if (!hidDevice.IsOpen) hidDevice.OpenDevice();
        }
        else
        {
            throw new Exception("Couldn't find the ergometer (through USB)! Are you sure it's plugged in?");
        }

        // connect to the analog in system
        analogManager = DeviceMgr.Get();
         string[] deviceNames = analogManager.GetDeviceNames();
         if(deviceNames.Length < 1) {
                throw new Exception("No analog input device found, are you sure the datatranslation box is plugged in?");
         }
        analogDevice = analogManager.GetDevice(deviceNames[0]);
        ain = analogDevice.AnalogInputSubsystem(0);
        ain.DataFlow = DataFlow.SingleValue;
        ain.Config();
    }

    void device_Inserted()
```

```
{
    if(Connected != null) Connected();
}

void device_Removed()
{
    StopCollection();
    if (Disconnected != null) Disconnected();
}

public void Disconnect()
{
    // stop the analog connection
    if(ain != null) ain.Dispose();
    if(analogDevice != null) analogDevice.Dispose();

    // stop the USB connection
    StopCollection();
    if (hidDevice != null) hidDevice.Dispose();
    if (Disconnected != null) Disconnected();
}

public void StartCollection()
{
    collecting = true;
```

```
            hidDevice.ReadReport(OnReport);
        }

        private void OnReport(HidReport report)
        {
            if (!collecting) return;

            byte[] data = report.GetBytes();

            // grab the time in seconds between reports
            // should be accurate to microseconds
            uint dt = (uint)(data[1] << 24) | (uint)(data[2] << 16) | (uint)(data[3] << 8) |
(uint)data[4];
            double ddt = (double)dt * 0.000001;
            t += ddt;

            // grab the velocity
            double rightVelocity = (double)BitConverter.ToSingle(data, 5);
            double leftVelocity = (double)BitConverter.ToSingle(data, 9);

            // sample the forces
            double leftForce = ain.GetSingleValueAsVolts(0, 1);
            double rightForce = ain.GetSingleValueAsVolts(1, 1);

            // update the program
```

```
    if(NewData != null)
    {
        NewData(new ErgoDataPacket(t,
        vlFilter.filterPoint(leftVelocity),
        vrFilter.filterPoint(rightVelocity),
        leftForce, rightForce));
    }

    // continue collecting data
    hidDevice.ReadReport(OnReport);
}

public void StopCollection()
{
    collecting = false;
}

public void SetPressure(float left, float right)
{
    if (hidDevice != null)
    {
        byte[] packet = new byte[sizeof(float) * 2 + 1];
        BitConverter.GetBytes(left).CopyTo(packet, 1);
        BitConverter.GetBytes(right).CopyTo(packet, sizeof(float) + 1);
        hidDevice.Write(packet);
```

```
            //hidDevice.WriteAsync(packet);
        }
    }
  }
}
```

ErgoDataPackets.cs

```
using System;
using System.Collections.Generic;
using System.Linq;
using System.Text;
using System.Threading.Tasks;

namespace ErgoInterface2.Hardware
{
  class ErgoDataPacket
  {
    public double t = 0;
    public double velocityLeft = 0;
    public double velocityRight = 0;
    public double forceLeft = 0;
    public double forceRight = 0;

    public ErgoDataPacket()
    {
```

```
        }

        public ErgoDataPacket(double t, double vl, double vr, double fl, double fr)
        {
            this.t = t;
            velocityLeft = vl;
            velocityRight = vr;
            forceLeft = fl;
            forceRight = fr;
        }
    }
}
```

Matrix2x2.cs

```
using System;
using System.Collections.Generic;
using System.Linq;
using System.Text;
using System.Threading.Tasks;

namespace ErgoInterface2.Models
{
    class Matrix2x2
    {
```

```
        public double a, b, c, d;

        public Matrix2x2(double a, double b, double c, double d)
        {
            this.a = a;
            this.b = b;
            this.c = c;
            this.d = d;
        }

        public static Vector2 operator *(Matrix2x2 m, Vector2 v)
        {
            return new Vector2(m.a * v.x + m.b * v.y, m.c * v.x + m.d * v.y);
        }
    }
}
```

Vector2.cs

```
using System;
using System.Collections.Generic;
using System.Linq;
using System.Text;
using System.Threading.Tasks;

namespace ErgoInterface2.Models
```

```
{
    class Vector2
    {
        public double x, y;

        public Vector2(double x, double y)
        {
            this.x = x;
            this.y = y;
        }

        public static Vector2 zero
        {
            get { return new Vector2(0, 0); }
        }

        public static Vector2 operator +(Vector2 v1, Vector2 v2)
        {
            return new Vector2(v1.x + v2.x, v1.y + v2.y);
        }

        public static Vector2 operator +(Vector2 v, double d)
        {
            return new Vector2(v.x + d, v.y + d);
        }
```

```
        public static Vector2 operator *(Vector2 v, double s)
        {
            return new Vector2(v.x * s, v.y * s);
        }

        public static Vector2 operator /(Vector2 v, double d)
        {
            return new Vector2(v.x / d, v.y / d);
        }

        public override string ToString()
        {
            return String.Format("[{0:0.0}, {0:0.0}]", x, y);
        }
    }
}
```

IMUDataPackets.cs

```
using System;
using System.Collections.Generic;
using System.Linq;
using System.Text;
using System.Threading.Tasks;
```

```
namespace ErgoInterface2.Hardware
{
    class IMUDataPacket
    {
        public double[] q = new double[4];

        public IMUDataPacket()
        {

        }

        public IMUDataPacket(double[] q)
        {
            this.q = q;
        }
    }
}
```

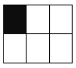
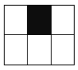
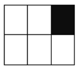
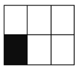
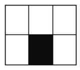
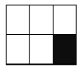
PassThruDataFilter.cs

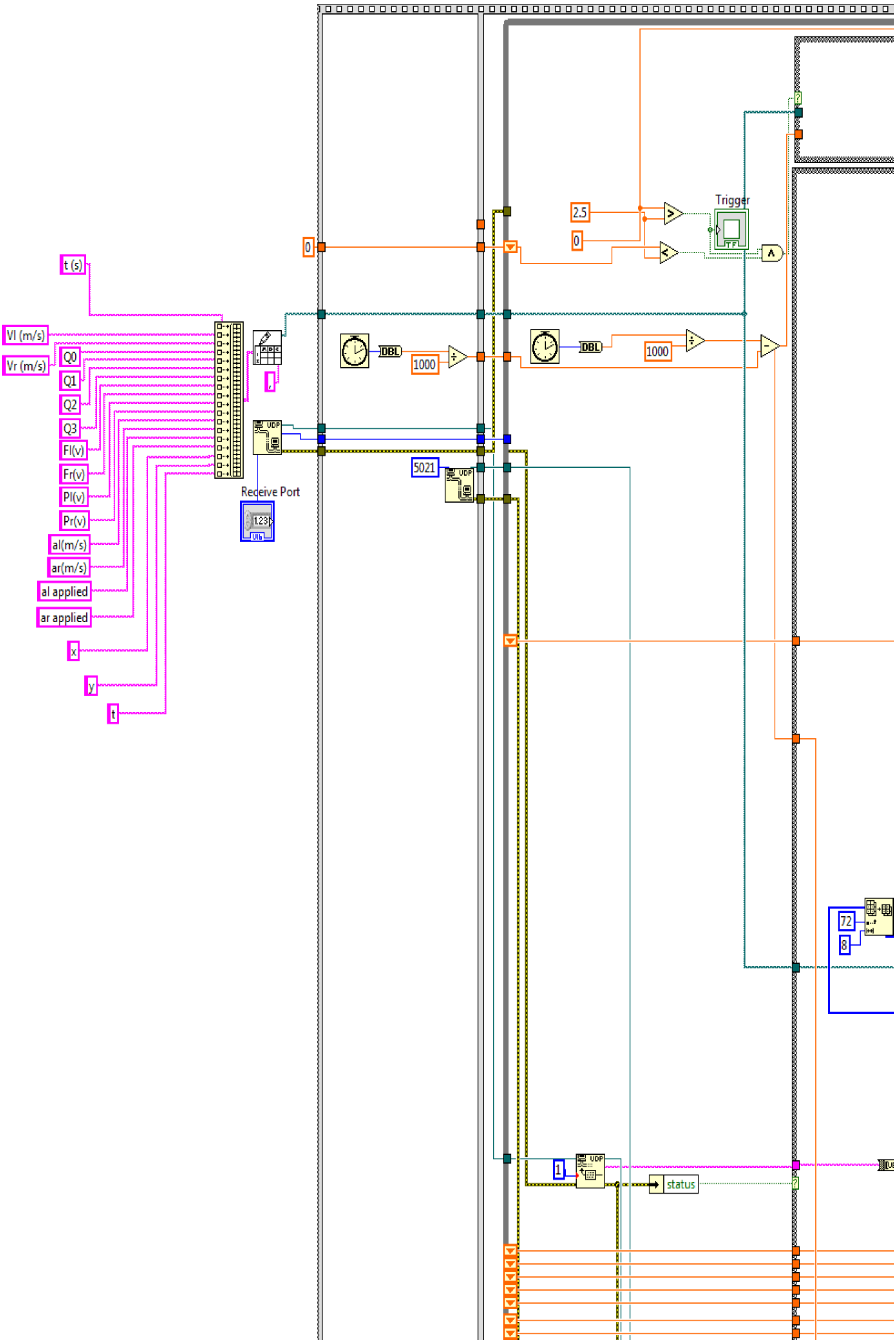
```
using System;
using System.Collections.Generic;
using System.Linq;
using System.Text;
using System.Threading.Tasks;
```

```
namespace ErgoInterface2.Filter
{
    class PassThruDataFilter : IDataFilter
    {
        public double filterPoint(double p)
        {
            return p;
        }
    }
}
```

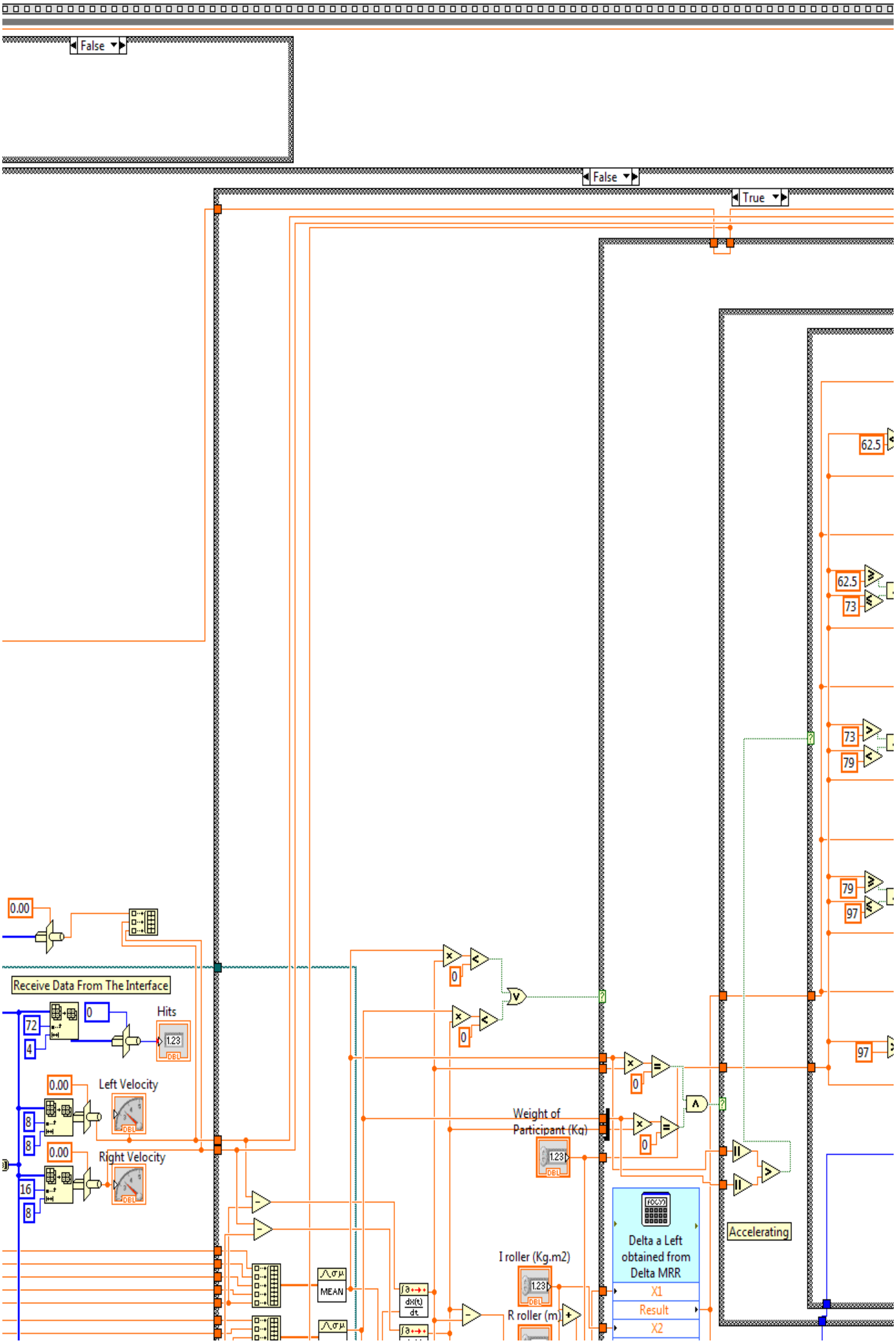
Appendix E LabVIEW Block Diagram

This chapter includes the LabVIEW block diagram used in VR_sysII. For visibility purposes, it has been divided to 6 subfigures as the following:

a: , b: , c: , d: , e:  , and f: 

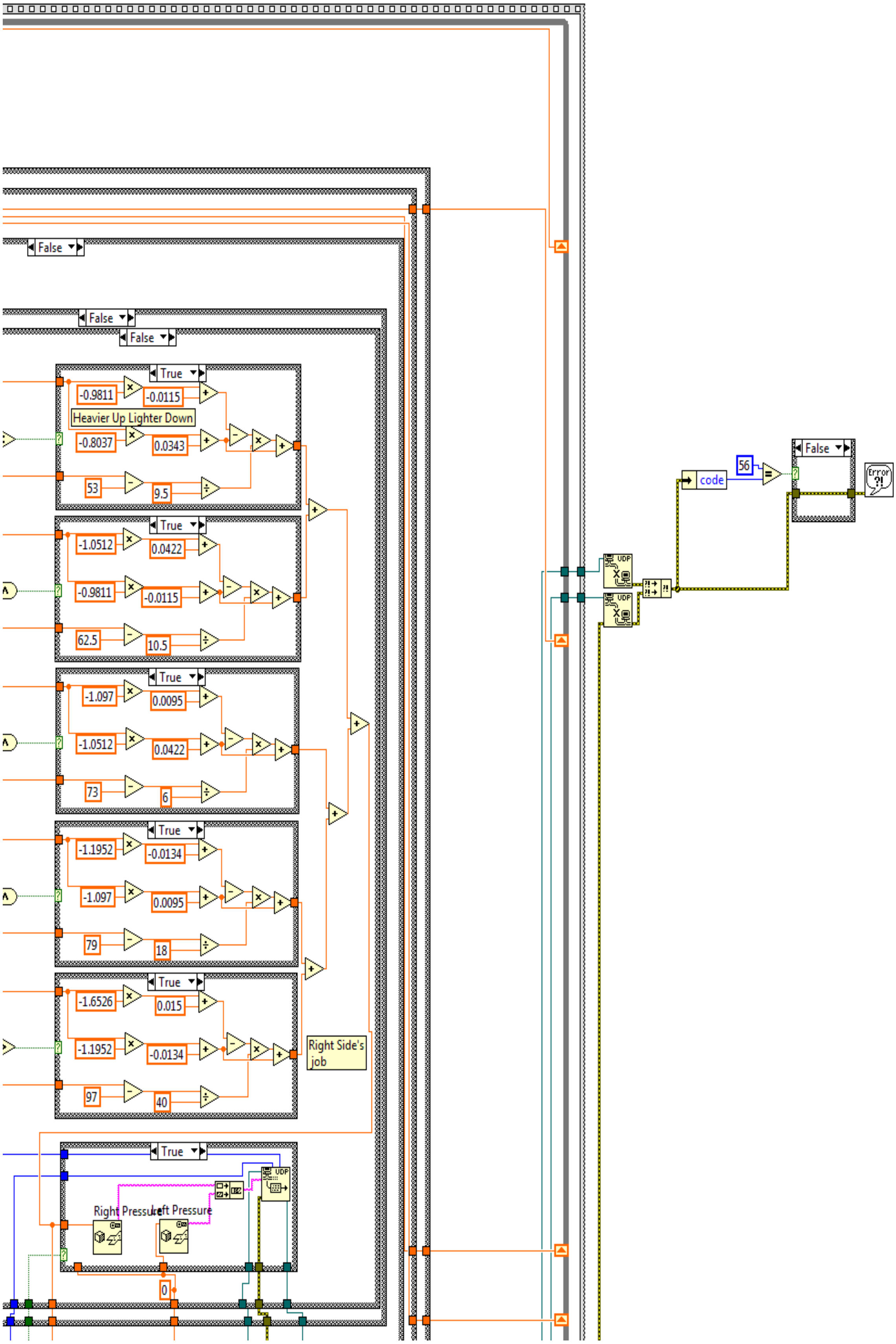

t (s)
Vl (m/s)
Vr (m/s)
Q0
Q1
Q2
Q3
Fl(v)
Fr(v)
Pl(v)
Pr(v)
al(m/s)
ar(m/s)
al applied
ar applied
x
y
t
Receive Port
0
2.5
0
Trigger
DBL
1000
DBL
1000
5021
1
status
72
8

a)

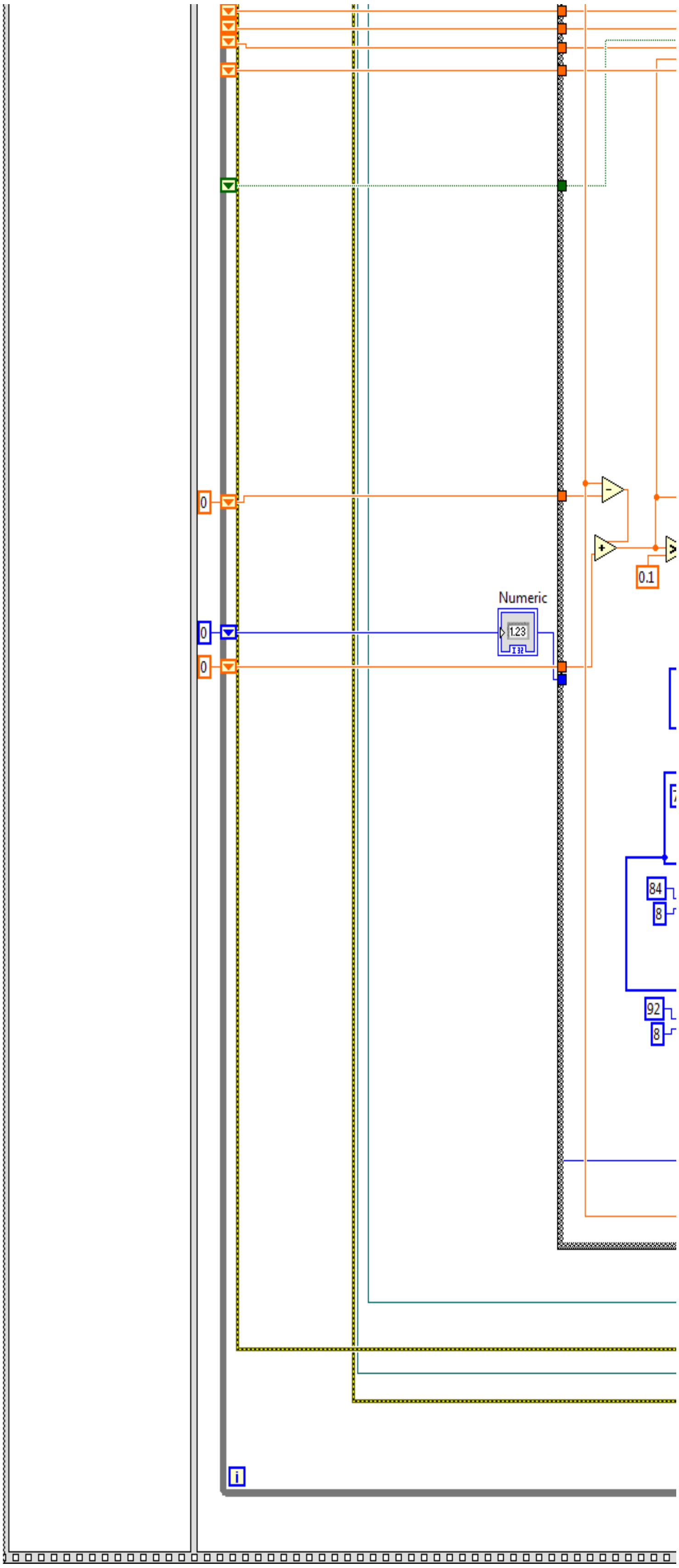


False
False
True
62.5
62.5
73
73
79
79
97
97
0.00
Receive Data From The Interface
72
4
0
Hits
0.00
Left Velocity
8
8
0.00
Right Velocity
16
8
MEAN
Weight of Participant (Kg)
I roller (Kg.m2)
R roller (m)
Delta a Left obtained from Delta MRR
X1
Result
X2
Accelerating

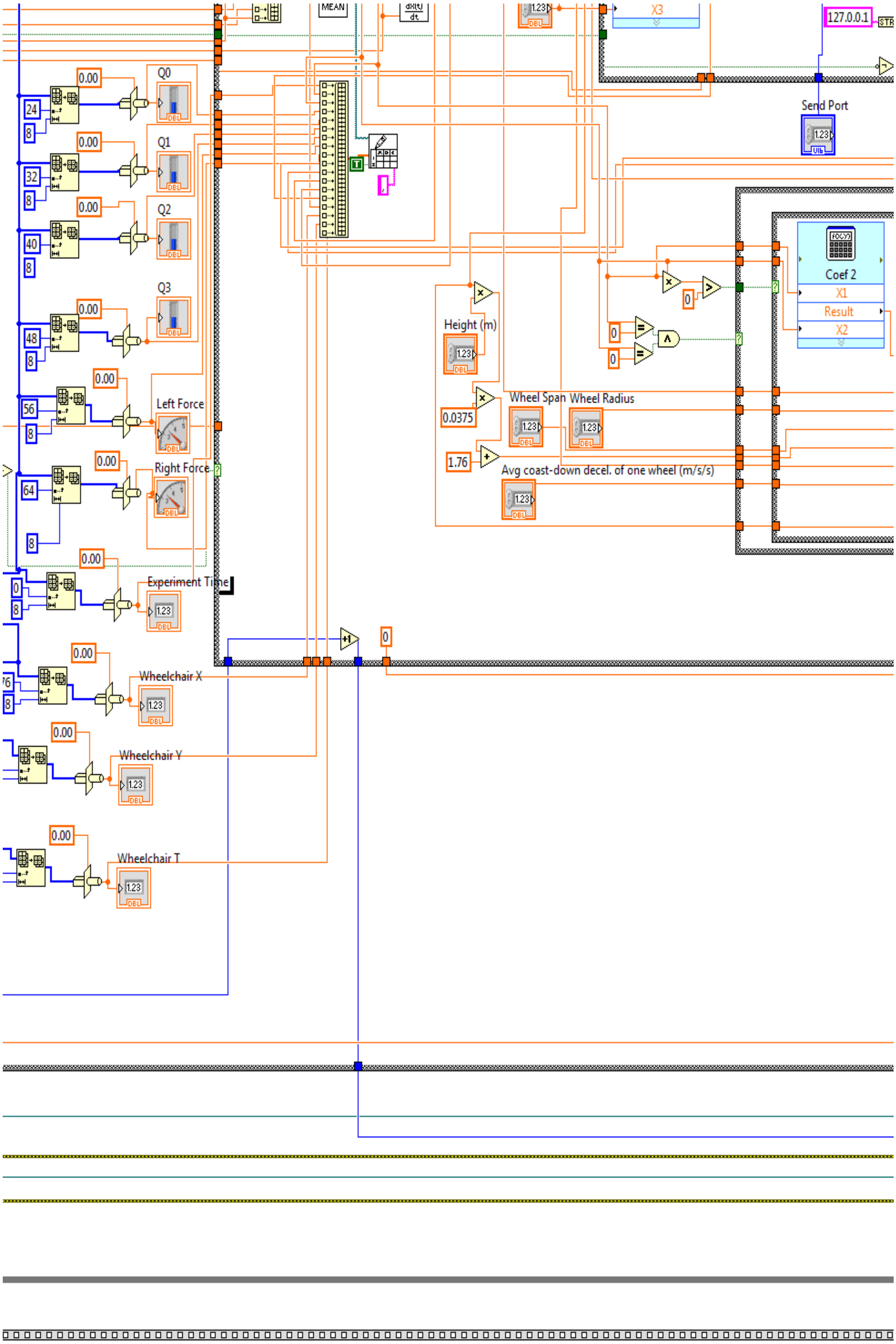
b)


False
False
False
True
-0.9811
-0.0115
Heavier Up Lighter Down
-0.8037
0.0343
53
9.5
True
-1.0512
0.0422
-0.9811
-0.0115
62.5
10.5
True
-1.097
0.0095
-1.0512
0.0422
73
6
True
-1.1952
-0.0134
-1.097
0.0095
79
18
True
-1.6526
0.015
-1.1952
-0.0134
97
40
Right Side's job
True
Right Pressure
Left Pressure
UDP
0
code
56
False
UDP
UDP

c)

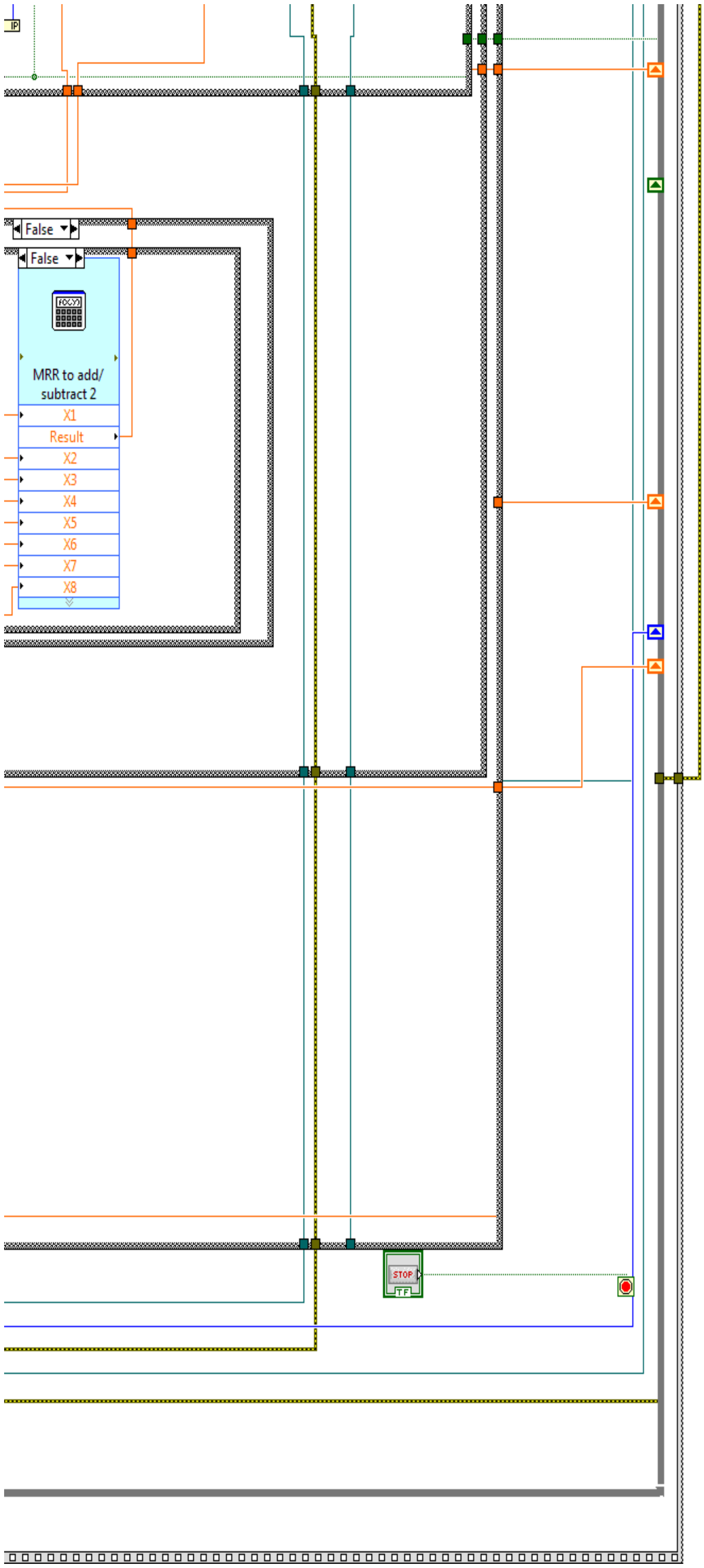


d)



Q0
Q1
Q2
Q3
Left Force
Right Force
Experiment Time
Wheelchair X
Wheelchair Y
Wheelchair T
MEAN
Height (m)
0.0375
1.76
Wheel Span
Wheel Radius
Avg coast-down decel. of one wheel (m/s/s)
Send Port
127.0.0.1
Coef 2
X1
Result
X2
X3

e)



f) Figure E-1 LabVIEW block diagram of VR_sysII