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Department of Neurosciences, Biomedicine and Movement Sciences

Development of a human-powered watercraft for people with lower-body disabilities

Ph.D. thesis

by

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Preface

This thesis has been submitted to the Graduate School of Life and Health Sciences at the University of Verona in fulfillments of the requirements for the Ph.D. degree in Neuroscience, Psychological and Psychiatric Sciences.

The project has been supervised by Dr. Luca Paolo Ardigò to whom I express my sincere gratitude for the competent professional guidance and support, which has helped me pursue the scientific goal.

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Abstract

Spinal cord injuries (SCI) or other lower-body disabilities can change the life of a person significantly both physically and psychologically. Chapter 1 describes how physical activity is associated with improved physical fitness, health and psychological well-being for people with lower-body disabilities but also that a number of limitations exist to potentially discourage this population from exercising. One of these limitations is a lack of exercise water activities as traditional aquatic sports such as rowing, kayak and canoe requires a level of trunk muscle activation that is often missing for people with SCI.

The main aim of this dissertation has been to develop a human-powered watercraft that could be maneuvered by people with lower-body disabilities. This process is described in chapter 2 including the results of a study, which showed the metabolic expenditure when using the watercraft to be similar to other physical activities performed by people with lower-body disabilities. Moreover, the mechanical efficiency was found to be comparable to other human-powered watercrafts and could, as a result, be an alternative fitness tool especially for people with lower-body disabilities, who seek water activities.

Chapter 3 describes the development of an improved version of the watercraft, and the results of testing that shows improved hydrodynamic resistance and relationship between mechanical power output and speed.

Chapter 4 investigates how velocity fluctuations may affect the speed of the watercraft and if different propulsion modes have an influence. No significant differences are found but several ideas for further research are given.

Finally, the interface pressure is evaluated in a setting that mimics the one on the watercraft. It is known that high interface pressure for long periods of time increases the risk of pressure ulcer development for people with SCI. Chapter 5 describes a case-study of an Italian handcycling champion arm cranking on an arm ergometer at two difference backrest inclinations while interface pressure

and oxygen uptake was measured. The results showed a difference in pressure between backrest inclinations and from arm cranking and resting. This could help people with SCI to still be able to exercise even if suffering from pressure ulcers.

Contents

Chapter 1

General Introduction

Spinal cord injury (SCI) may impair sensory, motor, and autonomic function below the level of injury. Consequently, many challenges and barriers are created and must be solved to participate in physical activity and exercise. Therefore, it is not a surprise that people with SCI are among the most inactive populations in society (Fernhall *et al*., 2008). A sedentary lifestyle in a wheelchair makes people with SCI especially vulnerable to cardiovascular diseases, pressure ulcers, and musculoskeletal pain and may have a negative effect on the psychological well-being, social participation, and overall quality of life (Boschen *et al*., 2003; Dijkers, 1998; Post and van Leeuwen, 2012). Accordingly, people with SCI are highly encouraged to participate in physical activity.

SCI or other lower-body disabilities often makes the most popular physical activities, walking and running, impossible. Innovation in sports technology is therefore critical in making physical activity easier and more accessible to this population. Innovation in sports technology does not only improve performance in sports but can also improve the daily living for disabled people. For example, many advances in wheelchairs for sports have become integral to the manual wheelchairs that are used nowadays. Materials such as aluminum, titanium, and composite materials were all introduced in the design of sports wheelchairs to improve sports performance by making the frame lighter and stiffer (Cooper and de Luigi, 2014). Manual wheelchairs now use the same materials but instead of improving sports performance, the materials help to increase the ability to independently perform activities of daily living and reduce repetitive strain injuries. Innovation in the frame design of wheelchairs in sports have also impacted the way manual wheelchairs are propelled and made them easier to transport in vehicles (Cooper and de Luigi, 2014).

Up until the 1980´s essentially the same chairs were used for both wheelchair sports and daily mobility (Cooper, 1990). In the last decades however, innovation in sports technology has greatly influenced the way people with disabilities are able to do physical activity. Specialized wheelchairs now exist for basketball, tennis, softball, rugby, and racing events. In addition, adapted sports technology allows people with SCI to participate in cycling, athletics, and skiing. However, as the adaptive sports opportunities have expanded, so have the expectations of people with impairments and one area where adapted sports technology seem to lack is in human-powered watercrafts, which include rowing boats, canoes, and kayaks. It is difficult for especially people with high-to-medium level SCI to navigate with these watercrafts due to the trunk movement impairment that often follows SCI. Accordingly, developing a novel human-powered watercraft, suitable for people with SCI, presents a challenging and relevant task. In the following sub-sections, SCI and the health complications that follows will be described followed by a discussion on which effect exercise, in particular on a novel human-powered watercraft, can have on people with SCI.

Spinal cord injury

The human nervous system transmits signals between different parts of the body and consists of the central nervous system (CNS) and the peripheral nervous system (PNS). The spinal cord and brain make up the CNS and any damage or disruption to the spinal cord that causes changes to its function is called spinal cord injury (SCI). Symptoms often include loss of sensation, motor control, and autonomic function below the level of injury and the severity of the impairment is dependent on the neurological level and the extent of the lesion. In general, the higher the injury on the spinal cord, the more dysfunction. Injury at or above the lumbar and sacral regions of the spinal cord can result in impairment to the legs and hips as well as dysfunction of the bowel and bladder. Injury to the thoracic region can affect the function of the muscles in the trunk in addition to the impairments found in lower-level injuries. SCI at the thoracic, lumbar or sacral level is termed paraplegia. SCI at the cervical region is termed tetraplegia and results in paralysis of the arms in addition to the lower-level impairments (Figure 1.1).

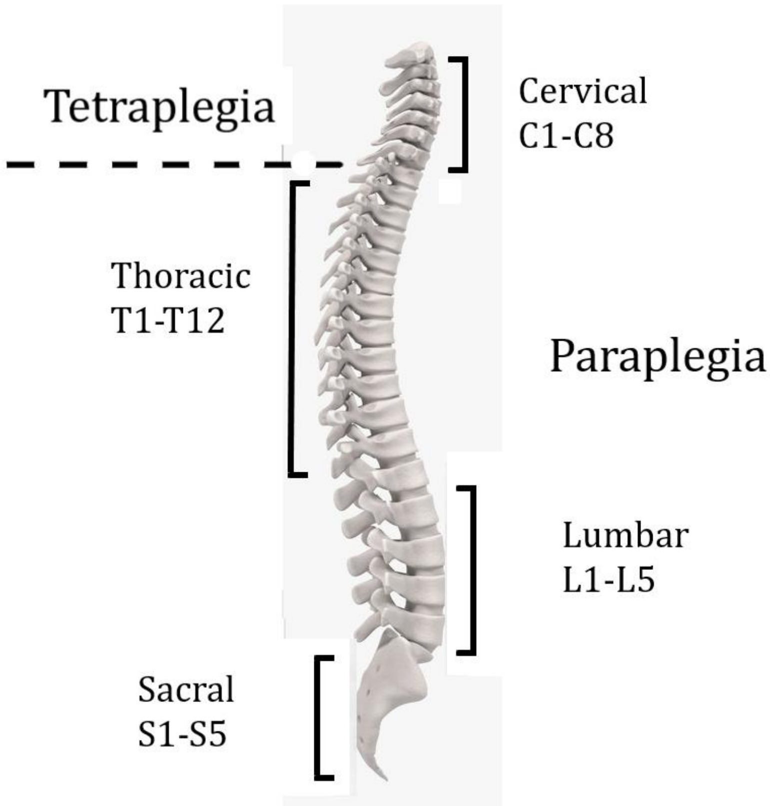

Figure 1.1. Classification of tetraplegia and paraplegia and the accompanying segment levels.

To determine the severity of injury, the American Spinal Injury Association Impairment Scale is often used. This scale divides individuals into five categories from a “Complete injury”, where no sensory or motor function is preserved in the sacral segments S4-S5, to “Normal”, where there are no sensory or motor deficits (Marino *et al*., 2016). The cause of the SCI can be either traumatic (due to external trauma such as falls, violence, or accidence) (Cowan *et al*., 2009) or non-traumatic (can be due to infections, tumors, or ischemia). The incidence of SCI in Italy is estimated to be approximately 20 persons per million (Pagliacci *et al*., 2003). Men are more often affected by traumatic SCI than women and the mean age at a traumatic injury is between 20 and 40 years.

Health complications following SCI

Persons with SCI face a range of health complications immediately as well as many years post injury. These can be categorized into neurological consequences (neurogenic pain or spasticity) and secondary complications (Haisma *et al*., 2007). Many of the secondary complications are often a consequence of a sedentary lifestyle in a wheelchair. After a person experience an SCI, the body will undergo changes in composition and metabolism (Bauman and Spungen, 2008). In general, persons with SCI travel around 2.5 km for a duration of around 50 min per day in the wheelchair (Tolerico *et al*., 2007). This duration and physical strain has shown not to be sufficient to decrease the risk of secondary health conditions (Ginis *et al*., 2012). As physical activity and daily energy expenditure often decline dramatically following an SCI, the risk of obesity and obesity related conditions such as cardiovascular diseases and diabetes is increased. Around 50 % of the SCI population is estimated to be overweight or obese (Rajan *et al*., 2008) and circulatory system diseases are the most common cause of death in people with SCI (Garshick *et al*., 2005). The correlation between lack of physical activity and the risk of developing cardiovascular diseases, diabetes, and obesity is well established in the general population (Warburton *et al*., 2006). Although less investigated, it seems that physical activity is also linked to decreasing the risk of developing chronic diseases for the SCI population (Ginis *et al*., 2012). Accordingly, training exercise programs are often included in the early phase of rehabilitation and the individual is encouraged to maintain an active lifestyle (Devillard *et al*., 2007).

Exercise limitations for people with SCI

Since the body undergoes considerable changes after an SCI, it is important to consider the limitations and problems that arise before prescribing exercise to this population. The physiological impairments that follow SCI typically result in a reduced ability to voluntarily perform large muscle group aerobic exercise and an inability to stimulate the cardiovascular system to support high rates of aerobic metabolism (Figoni, 2003). The limitations are dependent on the neurological lesion level, with cardiovascular dysfunction being greater at higher levels of injury (West *et al*., 2012).

For people with tetraplegia, the major problems that relate to exercise come from sympathetic autonomic impairments. When the sympathetic response is interrupted, it can lead to respiratory problems (maximal ventilation of 25 $Lmin^{-1}$), bradycardia (maximal heart rate of 130 bpm), sweating disturbances, temperature deregulation, insufficient splanchnic outflow or circulation to liver, kidney, and bowel, and autonomic dysreflexia. In addition, people with tetraplegia typically display lower upper body power output, cardiac output, and oxygen consumption compared to people with paraplegia (Devillard *et al*., 2007; Figoni, 2003; Theisen, 2012).

People with high level of paraplegia exhibit respiratory as well as cardiovascular exercise limitations. Injury to the thoracic cord can disrupt function of the intercostal muscles, accessory respiratory muscles, and abdominal muscles and thereby affect the breathing pattern (Bernard *et al*., 2000). Similar to cervical injured people, a high level of thoracic injury can lead to sympathetic autonomic impairments such as autonomic dysreflexia, temperature deregulation, and orthostatic hypotension. In addition, vasomotor disturbances can occur as lack of muscle pump in the legs causes a reduced venous return (Theisen, 2012; West *et al*., 2012).

Other frequent problems that can prevent people from doing exercise include bladder infections, pressure ulcers, and upper body injuries. Bladder infections can be exacerbated by dehydration from training and cause pain, increased muscle spasticity, and autonomic dysreflexia.

Pressure ulcers can significantly affect the quality of life (Dolbow *et al*., 2013) and it is estimated that approximately one third of people with SCI will develop a pressure ulcer at least once in their life (Lim *et al*., 2007) and up to 80% of those who develop a pressure ulcer will have a recurrence (Dolbow *et al*., 2013). People with SCI spend extended periods of time sitting or lying down and are at high risk of developing pressure ulcers (Bartley *et al*., 2016). A common observation is that able bodied people, after a certain duration of sitting, tend to make small movements in the chair, which could be the human body's response to relieve pressure at bony prominences such as the ischial tuberosities and sacrum (Kumar *et al*., 2015). Because of paralysis and a loss of sensory feedback, people with SCI do not experience the same discomfort and urge to relieve pressure and consequently they are at a higher risk of developing pressure ulcers (Giesbrecht *et al*., 2011; Sprigle and Sonenblum, 2011). Furthermore, during exercise local skin temperature can rise and cause perspiration, which leads to skin softening and adhesion and thereby a higher risk of pressure ulcer development (Hanson *et al*., 2010).

Wheelchair dependent persons are forced to rely on their arms for mobility, which can lead to overload of the upper extremities. Elbow, wrist, and especially shoulder pain is very common among people with SCI. To avoid overload, people with SCI are often encouraged to improve physical capacity and muscle strength by regular and appropriate exercise, beginning already in the early rehabilitation (Arnet *et al*., 2012).

The role of physical exercise in improving health for people with SCI

For people with SCI, guidelines state that adults should perform at least 20 minutes of moderate to vigorous intensity aerobic activity twice a week to improve their fitness. They are also recommended to perform strength training, 3 sets of 8-10 repetitions for each major muscle group, twice a week (Ginis *et al*., 2011). Improving the aerobic capacity has been associated with better cardiovascular health, while increasing the muscle strength can improve the functional independence by enhancing the ability to carry out activities of daily living such as propelling the wheelchair and performing transfers (Hicks *et al*., 2011). Muscle mass and strength is typically reduced after an SCI due to denervation and inactivity and consequently, the risk of overuse injuries and pain in the upper extremities is increased. Arm ergometry, wheelchair exercise, strength training, and functional electric stimulation assisted exercise has shown to improve aerobic exercise and power output for people with SCI (Hicks *et al*., 2011). People with SCI, who compete in sports, show greater cardiac hypertrophy, maximal work rate, maximal oxygen consumption, maximal lactate level, and pulmonary function than sedentary people with similar injuries (Ginis *et al*., 2012) and participating in sports might be a particular effective type of physical activity for maximizing cardiorespiratory fitness benefits, as people with SCI tend to participate in sports at higher intensities and longer durations than they participate in exercise (Ginis *et al*., 2010).

After an SCI, the body composition of the individual often changes with an increased fat mass and loss of lean mass. This contributes to the development of secondary health complications and chronic disease and may lead to issues with transfers, activities of daily living, and the need for more assistance and specialized equipment (Blackmer and Marshall, 1997). Sports participation, if performed at sufficiently high intensities and durations, has shown to be associated with lower fat mass and higher lean mass. Following SCI, bone mass density rapidly decreases as well, due to neural, vascular, and hormonal changes, which increases the risk of osteoporosis and fractures (Jiang *et al*., 2006). A review found that especially functional electric stimulation exercise has the potential to increase the bone mass density in persons with chronic SCI (Hicks *et al*., 2008) and participation in sports also seem to improve the bone mass density. Interestingly, a study found no difference in total bone mass density between athletes with SCI and able-bodied controls. This is most likely due to the greater upper body bone mass density in the SCI athletes, which compensates for the reduced lower body bone mass density (Sutton *et al*., 2009).

After an SCI, it is not uncommon that a decline in physical capacity and strength causes insufficient function to perform some activities of daily living. It is estimated that only 25% of young people with paraplegia have the *minimum* required level of fitness needed to maintain independent living (Noreau and Shephard, 1995). Studies have shown that exercise and sport participation can improve some activities of daily living such as performing transfers, self-care, and others (Ginis *et al*., 2012). Athletes have also reported that sport participation has helped them to acquire some of their most essential wheelchair mobility skills (Fliess-Douer *et al*., 2012).

As in the general population, physical activity for people with SCI is linked to reducing the risk for obesity, cardiovascular diseases, and diabetes (Buchholz *et al*., 2009). Studies have also found physical activity to have a positive effect with regard to changes in glucose metabolism and body-weight-supported treadmill training to increase glucose tolerance and insulin sensitivity. Furthermore, exercise and sport participation has shown to improve lipid profiles by increasing the level of the cardioprotective high-density lipoprotein cholesterol (Fernhall *et al*., 2008; Hicks *et al*., 2008; Mojtahedi *et al*., 2008; Nash, 2005; Washburn *et al*., 1999).

There is strong evidence in the literature that physical activity is associated with improvements in a wide range of psychosocial outcomes (Ginis *et al*., 2012). These psychosocial outcomes are often referred to as PSWB (physical and psychosocial well-being), which covers a broad category of phenomena that includes mental health, community integration, social participation, and overall life satisfaction. A metanalysis found that a significant association between lack of physical activity for SCI people and PSWB factors such as stress, mental health (e.g., depression), satisfaction with important domains (e.g., satisfaction with appearance, functioning), and overall life satisfaction (Ginis *et al*., 2010). Another review found that exercise and sports is effective in helping with community integration and social participation for people with SCI (Boschen *et al*., 2008). The authors of the review give examples of sports allowing people with SCI to compete amongst able-bodied athletes, thereby providing opportunities for full integration.

Although there may seem to be some risks associated with physical activity and sport participation (e.g., overuse injuries, autonomic dysreflexia), the benefits (e.g., reducing the risk of cardiovascular diseases, improving aerobic capacity, improving muscle strength, lower fat mass, higher lean mass, improving bone mass density, and improving PSWB) far outweighs the risks (Hicks *et al*., 2011).

Adaptive sports for people with SCI

Up until the 1980´s, people with SCI participated in sports using their daily mobility wheelchairs. This has changed lately as athletes work together with engineers, designers, and manufacturers to create novel sports equipment for adapted sports. Today, the design of sports equipment for adapted sports is tailored towards each specific sport resulting in dramatic improvements in sport performance (Cooper and De Luigi, 2014). The equipment is designed to meet the specific demands and applications of the different sports and to fit the athletes anthropometrics, sport classification need, general preferences, and functional abilities (van der Woude *et al*., 2006). The development of sports technology for adapted sports has allowed people with disabilities to participate more freely in sporting activities. The driving principle in developing sports equipment is to create an efficient human-machine interface. The equipment should almost be looked at as an extension of the body, similar to an orthotic device. There is adapted sports equipment allowing people with SCI to participate in many sports, and some of the more popular ones will be discussed below.

Wheelchair sports

Competitive and recreational wheelchair sports can be categorized into court wheelchair sports, all terrain wheelchair sports, and wheelchair racing. For all sports, the wheelchairs share some basic principles. The wheelchair should fit the user, so they act as one, the weight should be minimized, while keeping high stiffness, the rolling resistance should be at a minimum, and the sports-specific design of the chair should be optimized. The majority of sports wheelchairs are made of *aluminum*, *titanium*, or composite materials. *Aluminum* is used since it is lightweight, easily available, and easy to work with, while composite materials are often used by athletes since they lead to better results, but are more expensive and labor-demanding for manufacture (Cooper and De Luigi, 2014).

Wheelchair basketball is considered one of the oldest sports to be played with wheelchair (Ardigò *et al*., 2005). The game is played on a flat court with specialized wheelchairs constructed with six wheels instead of the four wheels of a typical wheelchair. In addition to the two drive wheels and two front swivel casters, the basketball wheelchair is constructed with two rear swivel casters to allow the drive wheels to be placed near the center of gravity of the basketball player (Figure 1.2). This make the chair more responsive to turning and allows for more efficient propulsion. Camber, the angle the wheels make with respect to vertical, of up to 10° is added to the drive wheels. The camber increases side-to-side stability, speed and turning responsiveness, and prevents opposing players from getting too close to the seat of the wheelchair. Furthermore, the hands are protected when two wheelchairs are next to each other (Cooper and De Luigi, 2014). Straps are used to hold the athlete firmly in the seat and the feet of the player are tucked under the seat to reduce the angular moment of inertia and make it turn quicker. The wheelchair is often adjusted to fit the specific position of the player. Forwards and centers prefer to sit as high as the rules allow to shoot the basketball as close to the basket as possible. Guards however, lower their center of gravity to improve maneuverability (Cooper and De Luigi, 2014).

Figure 1.2. Basketball wheelchair (Ardigò et al., 2005).

Tennis wheelchairs are very similar to the wheelchairs in basketball as quickness and maneuverability are important elements (Sindall *et al*., 2013). However, as there is no contact between the players in tennis, the chair can be more streamlined than the basketball wheelchairs. Some tennis chairs have handles mounted on the front of the seat to help the athlete keep the balance as they lean or extend to strike the ball with the racquet (Figure 1.3; Cooper and De Luigi, 2014).

Figure 1.3. Tennis wheelchair (Cooper and De Luigi, 2014).

In contrast to able-bodied rugby, wheelchair rugby is played indoors. Wheelchair rugby differentiates itself from most other wheelchair sports as direct contact between the chairs occur often and with high impacts. To cope with the frequent impacts with other chairs, solid wheel covers made of *aluminum* are used to protect the spokes and let other chairs glance off upon contact. The frame of the rugby chair is very similar to the wheelchair in basketball with six wheels and the drive wheel located near the center of mass. The seat is often tilted to form a “V” shape between the legs and torso, which enables the player to carry the ball in a secure position (Cooper and De Luigi, 2014). Gloves are used primarily for hand protection but also to improve ball handling and maximize the application of propulsive forces. A firm grip is important in most wheelchair sports as the majority of athletes have reduced hand function (Figure 1.4).

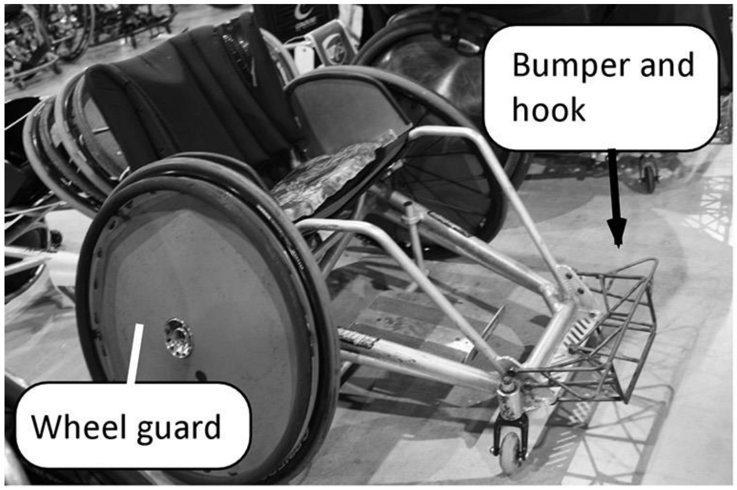

Figure 1.4. Rugby wheelchair (Cooper and De Luigi, 2014).

Throwing-chairs are used when competing in shot put, discus, club throw, and javelin. All throwing chairs use a footrest for stability and support and can also include side rests and a holding bar of either metal or fiberglass. The chairs have to be rigid with no rotation and a *maximum* height of 75 cm is allowed. Recently advances have been made regarding adjustable features for throwing chairs. Footrest height and angle, backrest height and depth, pole angle and height, and seat height to accommodate different cushion heights are now quicker and easier to adjust. This has several benefits as more throwers can use the same chairs at sport clubs and spend less time adjusting the chair to each thrower (Figure 1.5; Laferrier *et al*., 2012).

Figure 1.5. Throwing wheelchair (Laferrier et al., 2012).

Wheelchair racing dates back to World War I, with stories of recovering veterans racing each other around the halls of hospitals (Linker, 2011). It wasn't until the 1980s that the wheelchairs were designed specifically for racing, but rule restrictions still limited the design options. As the overall length restriction was removed and steering gear was permitted, a 3-wheeled racing chair design was made possible. Larger front wheels and simpler, lighter, and stiffer frame designs have recently made racing much faster. The athlete leans forward in a kneeling position with the chest close to the knees. This position allows *maximum* power to be transferred from the arm and trunk muscles to the pushrims. Straps are used to hold the athlete locked into the seat and the pushrims should fit so the arms can be nearly fully extended during the propulsion cycle (Cooper and De Luigi, 2014). Racing gloves are used to allow skilled athletes to maintain control of the wheel and deliver propulsive forces without slipping (Figure 1.6).

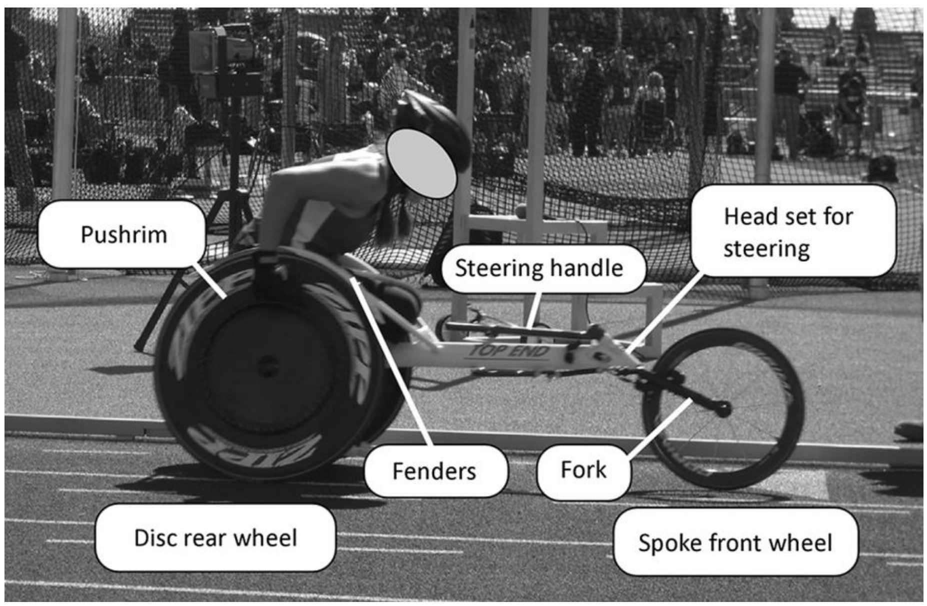

Figure 1.6. Racing wheelchair (Cooper and De Luigi, 2014).

Handcycling

The sport of handcycling started spreading in the 1980s, but handbikes were already being developed after World War I for veterans to travel longer distances over rough terrain (Linker, 2011). The frames were large and heavy, while the seat was in an upright position much like a wheelchair seat (Zipfel *et al*., 2009) (Figure 1.7).

Figure 1.7. Historic handcycle (Zipfel et al., 2009).

Handbikes can be either attach-unit systems or fixed frame systems. The attach-unit handbikes are constructed by adding a crank system to the front of a wheelchair and can be practical for daily outdoor propulsion (Dallmeijer *et al*., 2004). The fixed frame system can be much faster mainly because of improved drivetrain efficiency and aerodynamics. For the sport of handcycling, primarily recumbent or knee-seat handbikes are used. On the recumbent handbike, the athlete is in a supine position with the crank placed above the chest. This position is mostly used by people with a high-level SCI, as the power for the push phase is mainly generated from the *triceps brachii* and *pectoralis major* muscles, whereas the *biceps brachii*, *posterior deltoid*, and *trapezius* muscles generate the power in the pull phase (Faupin *et al*., 2010). On the knee-seat handbikes, the athlete is in a kneeling position, which allows for forward rotation of the *pelvis*. In this position, the trunk muscles are also used to generate power, but it also requires excellent trunk balance and coordination (Figure 1.8; Verellen *et al*., 2012).

Figure 1.8. Recumbent handbike (on the left) and knee-seat handbike (on the right) (http://www.sports.org.au/cycling/).

Most handbikes have multiple gears allowing for racing in different terrains. Lean steering or fork steering is used to turn the handbike, with the latter being the more frequently used option by elite athletes as fine adjustments are made easier and it provides greater lateral stability at high speeds. Most frames are made from high-strength *aluminum*, which tends to fatigue with use, and consequently elite athletes will replace the frame relatively often. For more recreational athletes, a handbike can last several years. The handbike frame is designed as either a single-strut frame with a single beam running under the seat, or a parallel strut design, in which the frame is used to form the seat (Laferrier *et al.*, 2012).

Sit skiing

Sit skiing has been practiced since around the 1980s and allows wheelchair bound people to participate in downhill and cross-country skiing (Pringle, 1987). People with low level paraplegia and good upper body strength can use a mono-ski, which is a frame with a seat mounted on top of one ski. The frame is connected to the ski by means of a shock absorber system, which has the same role as the legs of abled bodied skiers. The skier is strapped to the seat and creates a three-point stance for stability with the use of an outrigger ski in each hand. Turning the mono-ski is done by leaning to the side while using the outriggers to keep the balance. A bi-ski is available for people with high level paraplegia or even tetraplegia. The bi-ski has the frame and seat mounted on top of two skis and is most often used for assisted skiing, as another skier is in control of the frame with straps or a handle and can assist with braking and turning. The frames are made of a resistant metal alloy and the height and width of the seat, as well as the foot support, can be adjusted to fit the user. The binding plate at the base of the sit-ski is similar to the binding of a ski-boot, which makes it easy to switch between different skis. To provide thermal comfort, an external fairing is often mounted in front of the legs to protect the user from wind and snow while improving the aerodynamics at the same time. Cross country sit skiing is mostly performed by low level paraplegics or amputees, as it requires the use of the trunk muscles to create forward propulsion with the poles. The cross country sit-ski is similar to the bi-ski, as the frame and seat is connected to two skis. The poles have straps, which are beneficial to athletes with a weak grip (Figure 1.9).

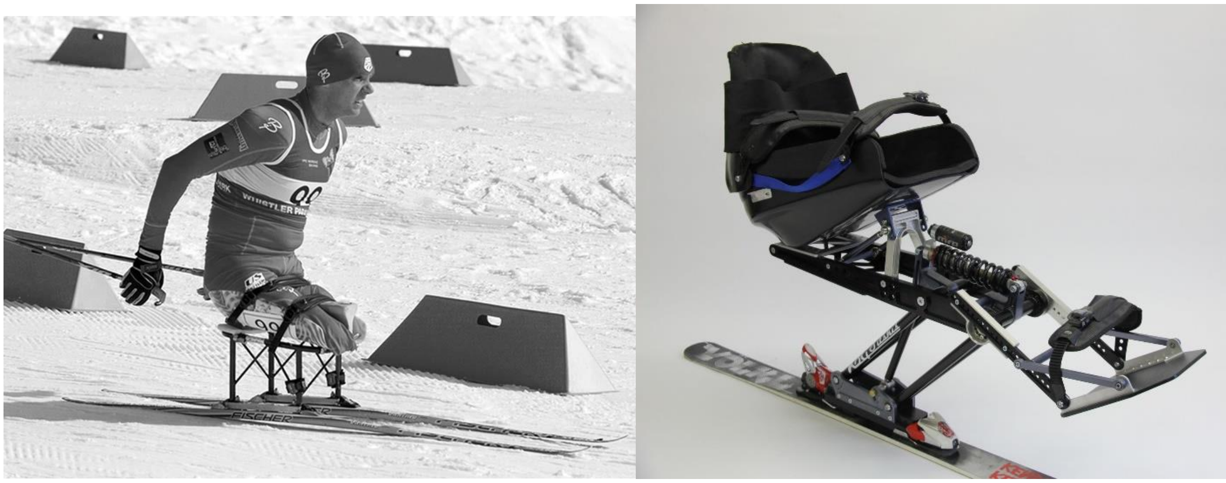

Figure 1.9. Cross country sit-ski (on the left) and downhill mono-ski (on the right) (https://enablingtech.com/products/kbg-lynx).

As seen in this overview, people with SCI have several options for physical activity and can participate in exercise in many different adapted sports due to the advancements in sports technology. An obvious absence in adapted sports technology is human-powered watercrafts, which include rowing boats, canoes, and kayaks. It is difficult for especially people with high-to-medium level SCI to navigate these watercrafts due to the trunk movement impairment that often follows SCI. The following chapter will describe the development and testing of a novel human-powered watercraft intended for people with SCI.

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Chapter 2

Development and Testing of a Novel Arm Cranking-Powered Watercraft

Abstract

There is a lack of human-powered watercrafts for people with lower-body disabilities. The purpose of this study was therefore to develop a watercraft for disabled people and investigate the metabolic cost and efficiency when pedaling. The watercraft was designed by combining parts of a waterbike and a handbike. Nine able-bodied subjects pedaled the watercraft at different speeds on a lake to provide steady-state metabolic measurements, and a deceleration test was performed to measure the hydrodynamic resistance of the watercraft. The results showed a linear correlation between metabolic power and mechanical power (r^2=0.93). Metabolic expenditure when pedaling the watercraft was similar to other physical activities performed by people with lower-body disabilities. Moreover, the efficiency of the watercraft showed to be comparable to other human-powered watercraft and could, as a result, be an alternative fitness tool especially for people with lower-body disabilities, who seek water activities. A number of suggestions are proposed however, to improve the efficiency and ergonomics of the watercraft.

Introduction

Spinal cord injury (SCI) may impair sensory, motor, and autonomic function below the level of the injury. As a result, people suffering from SCI often develop secondary impairments such as cardiovascular diseases, pressure ulcers, and musculoskeletal pain. These impairments are often the consequence of a sedentary lifestyle in a wheelchair, a device that 80-90% of persons with SCI rely on in everyday life (Biering-Sørensen *et al.*, 2005). People with SCI are among the most inactive ones in our society (Fernhall *et al*., 2008), as 50% of people with SCI reported less than 30 min of mild intensity activity *per* day in a study, which was considered insufficient to maintain or improve physical capacity by the authors (Gibbons *et al.*, 2014). Accordingly, additional physical activity is recommended for wheelchair users as physical activity is related to a reduced risk of cardiovascular diseases (Phillips *et al*., 1998), and obesity (Buchholz and Pencharz, 2004), and leads to less pain and fatigue in everyday life (Tawashy *et al.*, 2009).

A popular form of physical activity for SCI persons is handcycling. Handbikes for handcycling are driven by an arm crank system, similarly to what is known in cycling, but they are usually equipped with synchronous hand-pedals, three wheels, and a seat. Handbikes are more mechanically efficient than everyday wheelchairs, and handcycling is highly recommended to maintain the level of physical fitness and to prevent atherosclerosis, as it is characterized by a relatively high-energy consumption at moderate training intensities (Abel *et al.*, 2006). Furthermore, the closed-chain motion of handcycling enables propulsion force to be applied throughout the whole 360° of crank rotation, which is suggested to cause less musculoskeletal strain compared to everyday wheelchair use, and thereby decrease the risk of overuse injury (Dallmeijer *et al.*, 2004b).

People suffering from SCI also have the option of practicing both alpine and cross-country skiing (using sit skis), which assist the para-skiers with balancing, turning, and controlling the speed. Equipment also exists that enables SCI people to participate in activities such as basketball, rugby, tennis, and throwing events. At present, however, it is difficult especially for people with high-to-medium level (HML) SCI to navigate with human-powered watercrafts such as rowing boats, canoes, and kayaks, mainly due to the trunk movement impairment that often follows SCI. Accordingly, a new human-powered watercraft that is suitable for people with HML SCI would represent a highly valuable option for them for fitness and leisure activities as well. The aim of this study was to develop such a watercraft and investigate the metabolic cost and efficiency when pedaling the watercraft, using correlational analysis.

Material and Methods

Subjects

The experiments were carried out on nine (eight males and one female) able-bodied subjects (34.2±8.3 years; 75.3±9.5 kg; 1.75±0.07 m). The subjects had no experience in handcycling or injuries of the upper extremities. All subjects gave their written informed consent before testing, and were thoroughly informed about the purpose, benefits, and potential risks of the study, in conformity with the Code of Ethics of the World Medical Association (Declaration of Helsinki). The protocol and the methods applied in the study were approved by the Ethical Committee of the Department of Neurosciences, Biomedicine and Movement Sciences, University of Verona.

Watercraft

Figure 2.1*. The Handwaterbike.*

The watercraft, named the Handwaterbike, is a catamaran consisting of two carbon hulls to provide buoyancy, whereas a recumbent handbike seat (Maddiline Cycle, Sant´Ambrogio di Valpolicella, Italy), and footrests are fixed between the two hulls by means of a custom-built aluminum frame (Figure 2.1). The frame is placed on top of three aluminum pipes, which are tightened to the hulls using screws. The seat is attached to a wooden plate, which is attached to the frame using angular fittings. A synchronous arm-crank system in front of the seat is connected *via* a roller chain to the transmission system, which drives a flexible shaft and the propeller. The ratio of the chain ring to the sprocket is 52:15 or 3.47. The propeller has a diameter of 330 mm and a pitch of 450 mm. When pedaling is paused, the propeller folds together, removing potential seaweed from the blades. The rudder is positioned rear of the seat, and is connected to the gear-shifters on the left and right handle (Maddiline Cycle, Sant´Ambrogio di Valpolicella, Italy) *via* a wire. Because the wire is in tension, the rudder turns either left or right when one of the gear-shifters is pushed, depending on which gearshifter is pushed. The more it is pushed, the more the rudder turns. This system allows the user to keep pedaling even when maneuvering the boat. The main dimensions of the Handwaterbike are: length over all: 4.89 m; length of water line: 4.77 m; weight without subjects: 69.17 kg; maximal beam: 1.08 m; draught (the vertical distance between the waterline and the bottom of the hull): 0.10 m.

Measurements

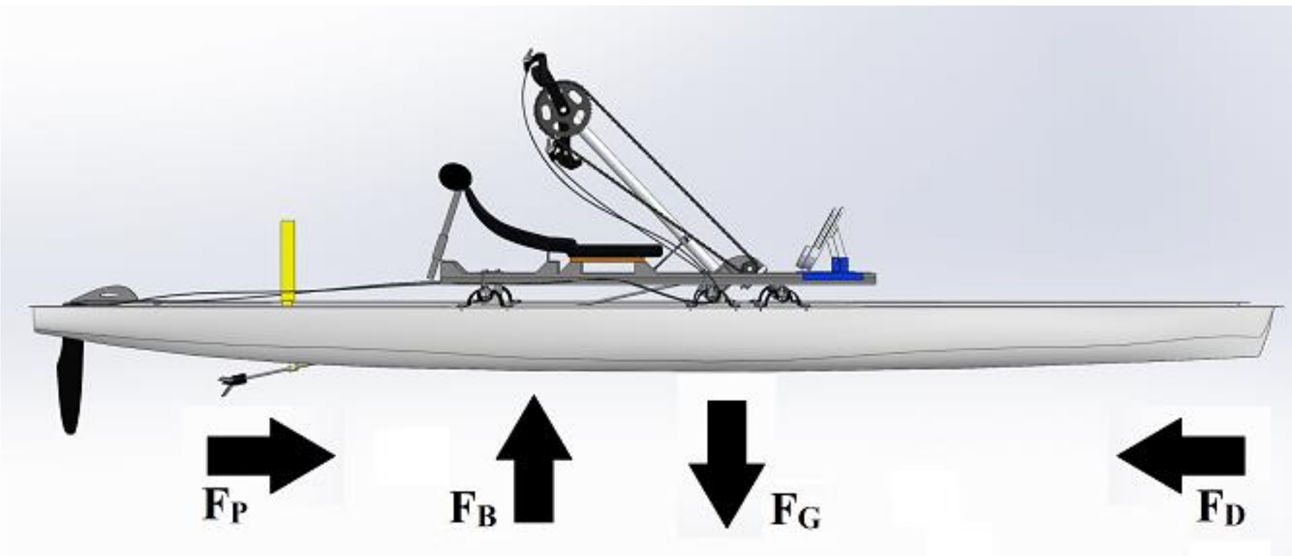

Figure 2.2. *3D CAD model of the handwaterbike and a free body diagram of the forces acting on the system. FP=propulsive force, FB=buoyant force, FG=gravitational force, and FD=drag force.*

Oxygen consumption ($\dot{V}O2, \mathrm{L} \cdot \mathrm{min}^{-1}$), carbon dioxide production ($\dot{V}\mathrm{CO}_2, \mathrm{L} \cdot \mathrm{min}^{-1}$), and heart rate (HR, bpm) were assessed breath-by-breath using a portable metabolic system (K5, COSMED, Rome, Italy). Before each test, the system was calibrated according to the manufacturer's instructions. The Handwaterbike was instrumented with a power-meter crankset (Quarq RIKEN R, SRAM, Spearfish, SD, USA) with a crank length of 170 mm allowing the measurement of pedaling frequency ($\mathrm{rev} \cdot \mathrm{min}^{-1}$) and of external mechanical power ($\dot{W}$) at a sampling rate of 60 Hz. Boat speed was measured at a sampling rate of 1 Hz by means of a GPS receiver (Rider 20, Bryton Inc., Taipei City, Taiwan) fastened to the arm-crank system. A 3D computer-aided design model of the Handwaterbike (Figure 2.2) was created using a CAD software (SolidWorks, Dassault Systèmes SolidWorks Corporation, Waltham, MA, USA) in order to measure the frontal surface area of the submerged part. The submerged area was corrected by accounting for the additional volume of water displaced when a subject was on board.

Experimental Protocol

The experiments were performed along the shore of the Garda Lake (Italy) in basically calm water and with wind speed always less than 2 $\mathrm{m} \cdot \mathrm{s}^{-1}$. The subjects were asked to pedal in a linear direction for at least 5 min, to allow steady-state metabolic measurements, and at three different constant speeds: (1) a speed the subject would be able to maintain for approximately 1 hour, (2) a “little bit slower than that”, and (3) a “little bit faster than that”. Before the trials, the subjects were asked to sit on a chair for 5 min. for measurement of metabolic variables at rest. Before the subjects could begin a new trial, they had to rest for at least 5 min, have a heart rate below 100 bpm, and communicate their availability to re-start.

A deceleration test was also performed to measure the hydrodynamic resistance of the Handwaterbike. A subject was instructed to pedal the boat at a constant speed of 2.2-2.8 $\mathrm{m} \cdot \mathrm{s}^{-1}$ before letting go of the handles. This procedure was repeated along the same segment six times, three times *per* direction to account for the influence of water stream, and overall average result was used. The speed of the Handwaterbike was measured by the GPS receiver during deceleration and using the method described by Bilo and Nachtigall (1980), and Capelli *et al.* (2009). The boat drag was calculated by analyzing the time course of the decreasing speed as a function of time. Data recorded for the first ~10 seconds during the deceleration tests were used for calculation of the hydrodynamic resistance.

Finally, the relationship between mechanical power output and boat speed was investigated by increasing the power by 10 W every 10 seconds until reaching 120 W. This was done three times in each direction to account for the influence of water stream, and overall average result was used.

Data Analysis

Only data collected from the last minute of each trial was used for analysis. Coefficient of variation (CV) of oxygen consumption data for the last minute of each trial was compared to the preceding minute of each trial to ensure steady-state was reached. All mean CV was less than 10%, which indicates that steady-state was reached. Mean oxygen consumption was converted into metabolic power using the empirical formula $\dot{\mathrm{E}} = ([4.94 \times \mathrm{RER} + 16.04] \times \dot{V}\mathrm{O2}\ /\ 60)$ (Garby and Alstrup, 1987), where RER is respiratory exchange ratio, and $\dot{V}\mathrm{O2}$ is the net oxygen uptake (above that measured at rest). The metabolic power was then divided by the average speed during the corresponding trial to calculate the metabolic cost of locomotion, C. The water drag (D) was calculated using the formula: $D = C_D \times A \times \rho \times v^2$ / 2 (Capelli *et al.*, 2009).

Metabolic equivalent (MET) was calculated by dividing the oxygen consumption by 3.5 $\mathrm{mL^{-1} \cdot kg^{-}1 \cdot min^{-1}}$ (Ainsworth *et al.*, 2011). The methods of Zamparo *et al.* (2008) were used to calculate power to overcome hydrodynamic resistance $\dot{\mathrm{W}}_d$, net mechanical efficiency η_0, propelling efficiency η_p, and drag efficiency η_d.

As descriptive statistics, mean values and standard deviations were used. Data from the metabolic system were low-pass filtered (2nd order Butterworth with a cutoff frequency of 0.1). The Pearson correlation coefficient (r) was used for correlation analysis. For inferential statistics purpose, level of significance was set at $P<0.05$. Coefficient of determination (r^2) was calculated to assess the strength of r. Statistical analysis was carried out using Microsoft Excel (Microsoft, Redmond, Washington, USA).

Results

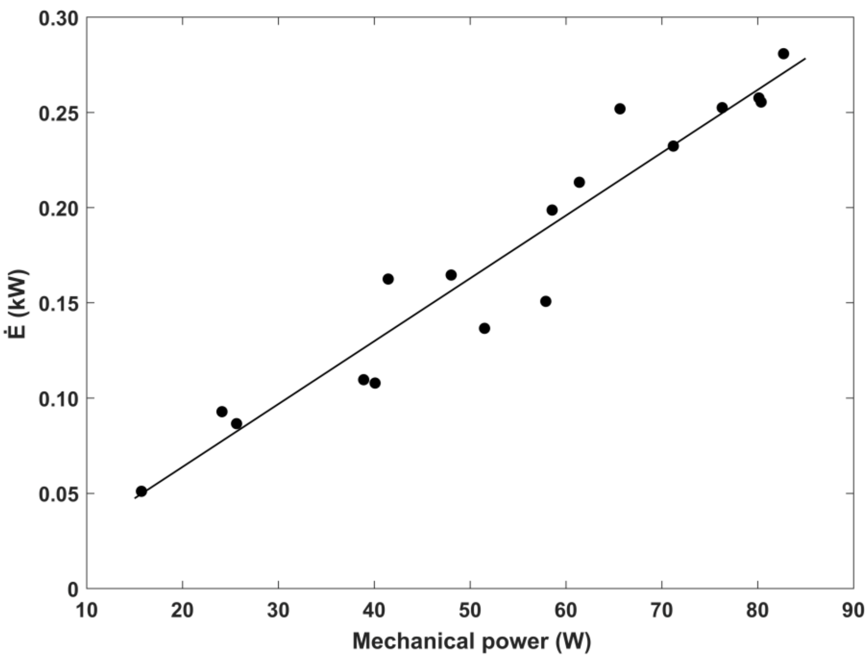

Figure 2.3. Metabolic power (Ė) at steady state plotted as a function of the mechanical power (Ẇ) while pedaling the Handwaterbike.

In Figure 2.3, the metabolic power at steady state Ė (kW) during pedaling the Handwaterbike is plotted as a function of mechanical power ($\dot{W}$). Data were fitted by a linear function $\dot{\mathrm{E}} = 0.0033 \cdot \dot{W} - 0.0022$; $r^2 = 0.93$; $SEE = 0.02$; $P < 0.005$. The metabolic cost of locomotion C (kJ · m^{-1}) for pedaling the Handwaterbike as a function of the speed v $(m \cdot s^{-1})$ is shown in Figure 2.4 and fitted by a (even if not significant) power function $C = 0.0901 \cdot v^{0.84}$; $r^2 = 0.19$; $SEE = 0.04$; $P > 0.05$.

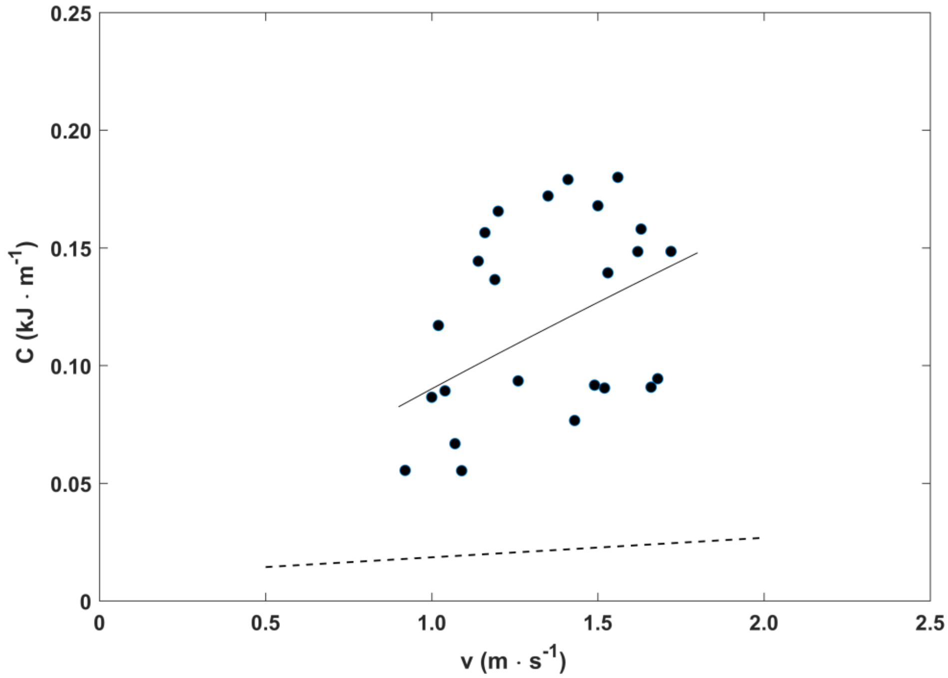

Figure 2.4. The metabolic cost needed to cover one unit distance (C) plotted as a function of the speed (v) for the Handwaterbike (continuous line) and for handbiking (dotted line) (Capelli et al., 2008).

In Figure 2.5, the metabolic equivalent (MET) during pedaling the Handwaterbike is plotted as a function of mechanical power ($\dot{W}$). Data were fitted by a linear function $\mathrm{MET} = 0.0485 \cdot \dot{W} + 0.5105$; $r^2 = 0.92; SEE = 0.31$; $P < 0.005$.

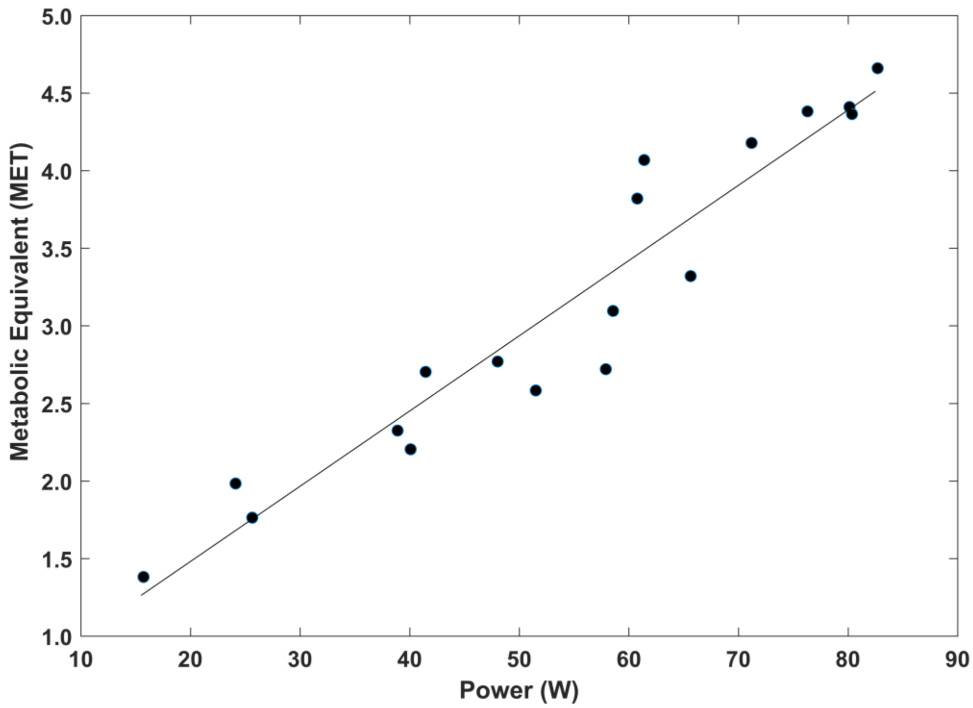

Figure 2.5. The metabolic equivalent (MET, where 1 MET is defined as the oxygen cost of sitting calmly, equivalent to 3.5 ml/kg/min) plotted as a function of the mechanical power (W).

In Figure 2.6, the reciprocal of the decreasing speed, recorded during one of the deceleration trials, is shown as a function of time. The average slope of the linear regressions calculated from the six trials was 0.082±0.015 $s \cdot m^{-1}$. With this value and the overall mass of the boat, the maximal frontal submerged area and the water density, the dimensionless C_D was calculated as 0.427 using the method described by Capelli *et al.* (2009). Knowing this, water drag of the Handwaterbike could be described by the following equation: $D = 5.64 \cdot v^2$.

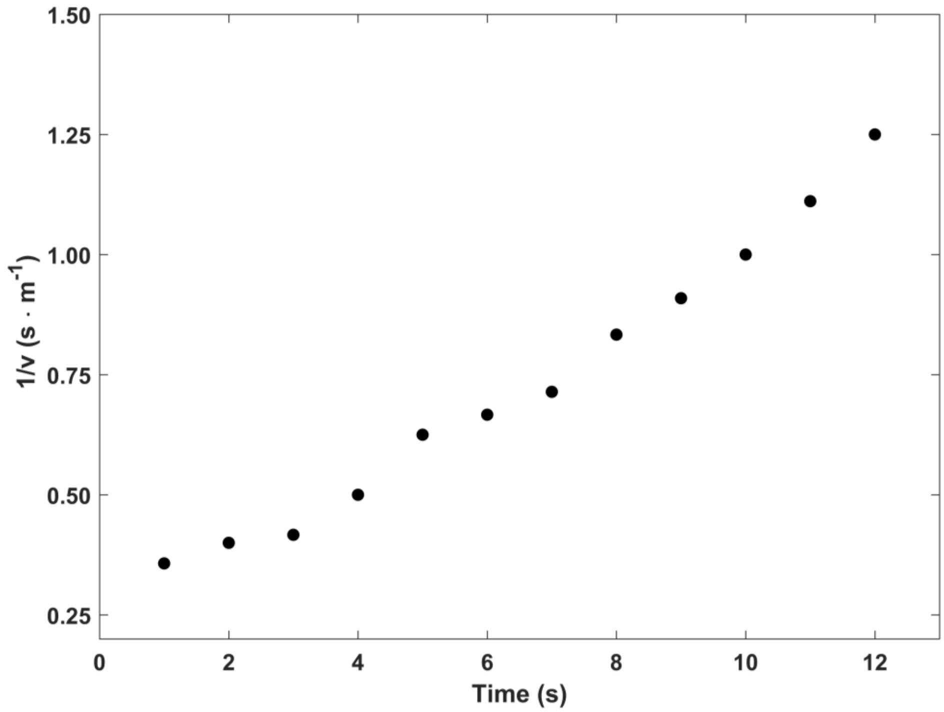

Figure 2.6. The reciprocal of the decreasing speed (v) obtained during a typical experiment of spontaneous deceleration plotted as a function of the time.

Net mechanical efficiency (η_0) when pedaling the Handwaterbike was 0.29±0.03 and was independent of both mechanical power output: $\eta_0 = 0.0001 \cdot W + 0.2906; r^2 = 0.0079;\ P > 0.05$, and cadence (RPM): $\eta_0 = 0.00007 \cdot RPM + 0.3033; r^2 = 0.00009;\ P > 0.05$.

Table 2.1. Comparison of five boats at a metabolic power ($\dot{E}$) of 0.5 kW (adapted from Zamparo et al., 2008).

	Paddle-wheel boat	Slalom kayak	Water bike	Rowing shell	Handwaterbike
υ (m · s^{-1})	1.3	1.8	2.3	2.4	2.4
$\dot{W}_d$ (W)	44	85	73	99	74
$\dot{W}_{tot}$ (W)	127	122	128	141	152
η_p	0.39	0.70	0.57	0.70	0.49
η_0	0.27	0.24	0.27	0.27	0.29
η_d	0.09	0.17	0.14	0.19	0.15

υ speed, $\dot{W}_d$ power to overcome hydrodynamic resistance, $\dot{W}_{tot}$ total power output, η_p propelling efficiency, η_0 overall efficiency, η_d drag efficiency.

Discussion

The aim of this study was to investigate efficiency and metabolic cost of pedaling a novel human-powered watercraft for people with lower-body disabilities. The data shows a strong linear correlation between the metabolic power and the mechanical power. The subjects could reach mechanical power outputs above 80 W. Arm cranking at this power has been shown to be sufficient for maintaining or improving cardiovascular health and fitness, and likely help prevent cardiovascular diseases for people with spinal cord injury (Abel *et al.*, 2003a). The Handwaterbike has a single gear ratio and the testing demonstrated that this was sufficient at power outputs from 20-80 W. Experienced handcyclists, however, may be able to produce a significantly higher power output, resulting in a higher cadence, which at some point will be impossible to increase. This could limit the utilization of the Handwaterbike as a training tool for this population. Another potential problem relates to propeller ventilation, which occurs when air is drawn into the water flowing to the propeller (Kozlowska *et al.*, 2009). The most common cause of propeller ventilation is when waves cause the propeller blades to break the water surface and become exposed to the air. Propeller ventilation can also occur during rapid accelerations. Then, if the spinning of the propeller is sufficiently fast, it causes a vortex from the surface to blades. Ventilation leads to sudden large losses of propeller thrust and torque, and could damage the propeller.

Several studies have demonstrated that additional energy consumption due to physical activity decreases the risk of mortality and morbidity (Blair and Brodney, 1999; Blair and Connelly, 1996). Figure 2.5 shows a linear relationship between metabolic equivalent (MET) and mechanical power output. At a moderate mechanical intensity of 50 W, the corresponding metabolic power is 2.9 MET. This is similar to activities such as wheeling outside, weight training and circuit training (Collins *et al*., 2009). Tweedy *et al.* (2017) recommend people with SCI to do ≥30 min of moderate exercise, defined as 3-6 METs, at least 5 days *per* week.

The metabolic cost of locomotion and speed showed a very low correlation with each other. This is likely due to the multiple testing days. Even though the wind speed was lower than 2 $\mathrm{m} \cdot \mathrm{s}^{-1}$ during all trials, it is plausible to think that some light water streams may have influenced the boat speed. If the testing had been performed during the same day or in a pool, the correlation would presumably have been higher at the cost of a lower ecological setting.

As indicated in Table 2.1, where all data refer to a metabolic power ($\dot{E}$) of 0.5 kW, the Handwaterbike reaches similar speeds to other reported boats in the literature, such as a waterbike and a rowing shell, but is faster than a paddle-wheel boat and a slalom kayak. The power to overcome drag ($\dot{W}_d$) for the Handwaterbike is similar to the waterbike but lower than for the slalom kayak and rowing shell. Consequently, the same is true for the drag efficiency (η_d) since it is calculated as $\eta_d = \frac{\dot{W}_d}{\dot{E}}$. η_d is defined as the efficiency with which the metabolic power is transformed into useful mechanical power (Zamparo *et al.*, 2008). This is to be expected since the waterbike and the Handwaterbike are catamarans with two symmetric hulls, whereas the kayak and the rowing shell are monohull boats and therefore likely with a smaller submerged frontal area. Lowering the weight of the relatively heavy aluminum frame could reduce the submerged frontal area of the Handwaterbike. It has to be noted though that the Handwaterbike is mainly intended for use by people with lower-body disabilities. Consequently, a high level of stability is needed, even if it negatively influences the performance of the watercraft. Similar to how a handbike worsens its performance by having three wheels instead of two (Fischer *et al.*, 2015; 2014). Some boats are able to maintain a low drag and high stability by having a long keel that provides lateral resistance to prevent the wind or waves from pushing the boat sideways. This solution however, can be problematic as the long keel makes it difficult for the boat to access smaller ports and to travel along the beach. Instead, a catamaran is able to achieve a good level of stability, low draught, and relatively low drag by using two hulls but thus increasing the width of the watercraft.

Table 2.1 shows that at an $\dot{E}$ of 0.5 kW the corresponding mechanical power output would be 152 W for the Handwaterbike. This is comparable, even if slightly higher, than a paddle-wheel boat (127 W), slalom kayak (122 W), water bike (128 W), and a rowing shell (141 W). This is likely due to the different propulsion styles of the boats. Propulsion is generated for the paddle-wheel boat and the water bike by leg pedaling in a semi-recumbent position, whereas propulsion for the rowing shell is generated by leg drive and arm pull of the oars. Propulsion of the slalom kayak is generated by an upper hand push and lower hand pull of a paddle, supported by a concomitant rotation of the torso. It should be noted though that Table 2.1 refers to a comparison between all the watercrafts at an $\dot{E}$ of 0.5 kW and even though Figure 2.3 shows a linear relationship between Handwaterbike $\dot{E}$ and W, we only have data up to 0.3 kW in the present study. It cannot be excluded that the relationship will look different at higher power outputs.

The Handwaterbike has a net mechanical efficiency (η_0) similar to the other boats in Table 2.1. η_0 is also similar to that measured with cycle ergometers (Zamparo *et al.*, 2008). This is somewhat surprising since η_0 of arm exercise has been reported to be lower than that of leg exercise (Marais *et al.*, 2002), probably due to the smaller and different (in terms of fiber types content) muscle mass involved in arm cranking versus cycling. An explanation could be that the subjects in this study were able-bodied and able (or freely self-compelled) to use all of their upper body to create propulsion and not just their arms. Especially people with high-level spinal cord injury are likely to have a lower η_0 when pedaling the Handwaterbike. It was expected that able-bodied subjects would respond relatively homogenously to pedaling the Handwaterbike, since they were equally inexperienced in arm cranking and had no restriction due to disability. It should also be noted that energy use is likely lower in persons with SCI compared to abled-bodied persons due to decreased fat-free mass and sympathetic nervous system activity (Buchholz and Pencharz, 2004). As a higher level of spinal cord injury has been shown to correlate with lower energy use (Abel *et al*., 2008; Collins *et al.*, 2009), it is likely that the energy use would be lower if the Handwaterbike were propelled by SCI people as well. Net mechanical efficiency (η_0) showed to be independent of mechanical power output and cadence. This is similar to Goosey *et al*. (2008), who found no significant difference in mechanical efficiency between arm cranking at 70 RPM and 85 RPM. This is in contrast to cycling, where the mechanical efficiency tends to decrease when cadence is increased (Sacchetti *et al*., 2010). Reservations should to be made, however, towards analyzing the cadence in our study since the Handwaterbike was limited to a single gear ratio. As a result, cadence will have been greatly influenced by especially stream direction and magnitude. Propelling efficiency (η_p), defined as the ratio of useful work to total work production, was lower for the Handwaterbike than all other boats in Table 2.1 except for the paddle-wheel boat, which was expected to be less efficient, since energy losses for paddle-wheel system has shown to be quite large (Zamparo *et al.*, 2008). The efficiency of the customized propeller mounted on the Handwaterbike varies according to the loading condition, but should be around 78%, when used on the Handwaterbike, according to the manufacturer. The remaining loss of efficiency is due to friction in the transmission chain, especially in the bearings and the roller chain.

As seen in Figure 2.2, the watercraft is affected by four forces: gravity, buoyancy, propulsion and drag. At (floating) constant speeds, the gravity and buoyancy forces are equal, and the propulsion and drag forces are equal as well. The drag forces consist of air and hydrodynamic drag. The hydrodynamic drag is dominant for most watercrafts as air density is much lower than water density, and air drag only contributes to 10% of the total drag for a rowing system (Baudouin and Hawkins, 2002). The hydrodynamic drag consists of form drag, skin drag, and wave drag. Form drag depends on the shape of the watercraft and hulls such as the one used in this study already approach the optimal shape for drag considerations (Baudouin and Hawkins, 2002). Skin drag depends on the friction that occurs between the water and the hull, and is responsible for over 80% of the hydrodynamic drag on a racing hull (Baudouin and Hawkins, 2002). A boundary layer is created when a thin layer of water is accelerated to the speed of the hull (Buckmann and Harris, 2014). The Reynolds number, i.e. the ratio of inertial forces to viscous forces, determine whether the boundary layer is laminar or turbulent. Laminar flow is a smooth and steady flow, whereas turbulent flow is a chaotic and unsteady flow. At the hull´s bow, the boundary layer is laminar but it quickly starts turning turbulent. In fact, approximately 80% of the hull is in a transitional flow from laminar to turbulent (Pendergast *et al.*, 2005). Skin drag can be decreased if such a transition is delayed and the laminar area is increased, since laminar flow produces less skin friction than turbulent flow (Day *et al.*, 2011). It has been suggested that skin drag could be reduced by applying a hydrophobic coating on the surface of the hull (Baudouin and Hawkins, 2002). Another skin drag possibility could be to alter the texture of the wetted hull surface. A smooth surface does not always result in minimal skin drag, as researches, inspired by the riblet surface of shark skin, have shown (Dean and Bhushan, 2010). Another skin drag reduction example is how dimples reduce the drag on a golf ball. Separation bubbles with small distinctive dimensions are generated inside the dimples resulting in low Reynolds numbers (Choi *et al.*, 2006). Due to the streamlined shape of the catamaran hull, dimples are unlikely to reduce the drag however. A further and obvious way to reduce skin drag is to reduce the weight of the system, thereby reducing the wetted area. As mentioned earlier however, this could negatively impact the stability of the Handwaterbike. The last form of hydrodynamic drag, wave drag, occurs as a result of pushing the water away from the hull. The hulls in the present study have bows with sharp angles, thereby pushing the water away gradually. As a result, wave drag is not a significant source of drag for long thin hulls (Baudouin and Hawkins, 2002).

As previously mentioned, the arm-crank system for the Handwaterbike is synchronous. Whereas this system has shown to be more efficient than an asynchronous arm-crank system in handbiking (Dallmeijer *et al.*, 2004a; Bafghi *et al.*, 2008; Abel *et al.*, 2003b), the applied tangential force is not constant through the entire crank revolution, but peaks during the pull down and push up phase (Arnet *et al.*, 2012). On a watercraft, this could result in fluctuations of boat speed. Water resistance is increased approximately four times when speed is doubled (Hill and Fahrig, 2009). Therefore, it would be most effective to keep the speed of the watercraft as constant as possible. When pedaling the Handwaterbike, an asynchronous arm-crank system should provide a more constant power output and might lead to the result of a higher efficiency. This should be tested in the future by comparing the metabolic cost of synchronous and asynchronous arm-cranking on the Handwaterbike at similar mechanical power outputs.

Testing highlighted some ergonomic problems with the Handwaterbike, mainly regarding handles, seat, and backrest and footrests adjustability. Every element is adjustable to a certain degree, but it requires specific tools and know-how. A future version of the Handwaterbike should look to make the possibilities for adjusting the watercraft to fit different users easier and faster. As mentioned previously, the subjects in this study were able-bodied. This is a limitation of the study since the Handwaterbike is intended for people with lower-body disabilities, a population likely to have a different metabolic response to propelling the watercraft. Outside of this study, several people with lower-body disabilities have tried the Handwaterbike and a paraplegic man participated in the 12 km long Gardalonga boat race on the Garda Lake in Italy (http://www.gardalonga.it/en), and experienced no problems. In the future however, a similar study should be performed with lower-body disabled subjects. Another limitation was that accuracy and precision of the GPS-receiver were not assessed, but there is already literature supporting its use for boat navigation research (e.g. with 1 Hz units achieving <1% error over >100 s durations (Smith and Hopkins, 2012).

Conclusions

The Handwaterbike is a novel human-powered watercraft designed for people with lower-body disabilities. This study showed that pedaling the Handwaterbike could be a promising fitness tool for people with lower-body disabilities looking for physical activities on the water. Pedaling the Handwaterbike demonstrated metabolic expenditure and power outputs similar to those thought to be sufficient to maintain or improve cardiovascular health and fitness, and reduce the risk of cardiovascular diseases. Several improvements are proposed however, mainly regarding ergonomics and efficiency.

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Chapter 3

Development of the HWB v. 2

Design considerations for the HWB v. 2

In chapter 2 it was described how the first version of the Handwaterbike (HWB) showed to be a promising fitness tool for people with lower-body disabilities looking for options to participate in physical exercise on water. A number of potential design enhancements were identified. When designing the HWB v. 2, the focus was on improving the performance, comfort and accessibility, while maintaining a safe to use product.

The two most obvious ways to improve the performance of a human-powered watercraft are to decrease the water drag and improve the drivetrain efficiency. It is possible to reduce the water drag by altering the shape or surface of the submerged parts of the watercraft. The hulls used for the HWB v. 1 are shaped like racing shells and already approach optimal shape for drag considerations (Baodouin et al., 2002). A coating applied to the surface of the hull could decrease the skin friction drag coefficient, leading to improved speeds (Baodouin et al., 2002). The skin friction develops in the boundary layer, which is the thin region close to the hull surface where the viscosity of the fluid is not negligible and the fluid velocity gradient is significant. The boundary layer can either be in laminar flow (a smooth and steady flow) or turbulent flow (a chaotic and unsteady flow). The laminar flow produces less skin friction than the turbulent flow and consequently drag can be reduced if the transition from laminar to turbulent flow is delayed. A coating such as hydrophobic polymers can be effective in delaying the transition but at the expense of polluting the water (Brooks *et al*., 1986). Another way of decreasing water drag is to reduce the weight of the watercraft, thereby reducing the submerged area. Reducing the weight of the watercraft can be done either by using less of the materials or using different and lighter materials.

As the watercraft is intended primarily for people with lower-body disabilities, including people with SCI, it is important that the watercraft has a high stability as some users will have limited trunk activation. Some boats use a long keel under the boat to minimize movement from side to side. This design makes it difficult for the watercraft to approach beaches or smaller harbors however, as the keel would hit the ground. Instead it was decided to continue the catamaran-design used for the HWB v. 1.

To ensure the watercraft is comfortable to use, accessible and safe to use, it was decided to design the frame based on the frame of modern handbikes. Handbikes have been used by lower-body disabled for years and have proven to be a good fitness tool with a low risk of injury (Arnet *et al.*, 2012). Based on user feedback from the HWB v. 1, the primary focus was on making the seating position easier and quicker to adjust to fit the individual user.

This project has made extensive use of computer-aided design (CAD) during the design cycle. CAD is a general term referring to the method of drawing and designing objects using a computer system. The purpose of CAD is to create and analyse technical drawings that represent an idea or concept. The drawings are often used in the architectural and engineering fields and can be done on a variety of CAD software applications. The drawings and analyses in this project have been made using SOLIDWORKS Student Edition 2016.

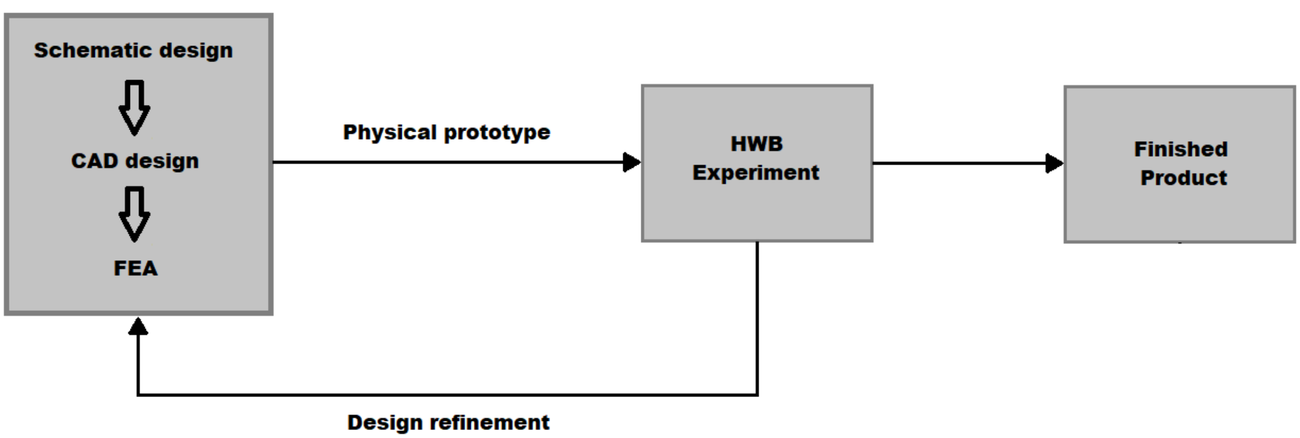

Figure 3.1. The design cycle used in the project.

The first step in the design cycle (Figure 3.1) is the schematic design. In this phase, the project goals and requirements are determined and a rough drawing, often by hand, is made to illustrate the basic concepts of the design.

CAD model and FEA of the HWB v. 2

Based on the schematic design, a more complete and precise CAD model is then developed. The CAD model provides a detailed 3D view of the product which is the Handwaterbike in this case. Some CAD applications are then able to do a Finite Element Analysis (FEA), which is the simulation of any given physical phenomenon, such as structural or fluid behavior, using the numerical technique called Finite Element Method. These phenomena can be described using partial differential equations. Using FEA, it is possible to solve these partial differential equations and compute relevant quantities of a structure such as stresses and strains. This analysis is performed to assess the design and predict how it will perform under a given condition. By using CAD and FEA it is possible to reduce the number of physical prototypes and consequently significantly reduce the time and costs of the research and development phase of a product (Bathe, 2008).

When the CAD model and FEA produce the desired results, a physical prototype can be manufactured. The physical prototype can then be used in an experiment where the potential benefits and safety of the product can be proved. Based on the observations and analyses of the experiment, it can be determined if the prototype is the finished product or if further design refinements are needed. Typically, it requires several iterations until the finished product is chosen, but a well-made CAD design and FEA are helpful in reducing the number of iterations.

For the HWB v. 1, the project goal was to design a human-powered watercraft that was safe to use for people with SCI. A CAD model and physical prototype was constructed (Figure 3.2), described in chapter 2, and an experiment was conducted to investigate the metabolic expenditure when pedaling the watercraft. In addition, the performance was assessed by performing a decelerating test and investigating the relationship between mechanical power output and boat speed. Based on this experiment, it was decided to perform further design refinements and optimize the HWB.

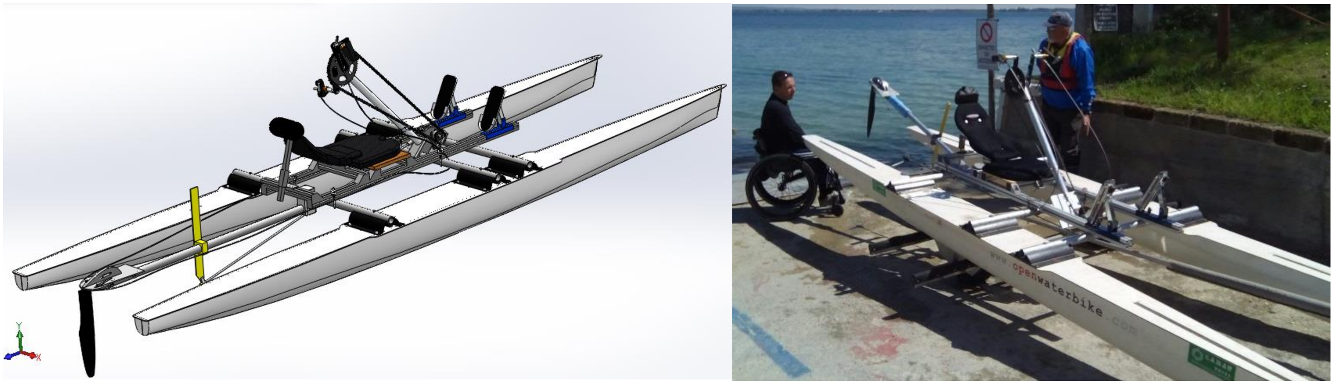

Figure 3.2. CAD design of the Handwaterbike version 1 (on the left) and the physical prototype of the Handwaterbike version 1 (on the right).

To improve the performance, it was decided to either reduce the frontal submerged area or the total submerged area. If the watercraft was changed from a catamaran to a monohull it could greatly reduce the frontal submerged area. A CAD model of a monohull was developed (Figure 3.3).

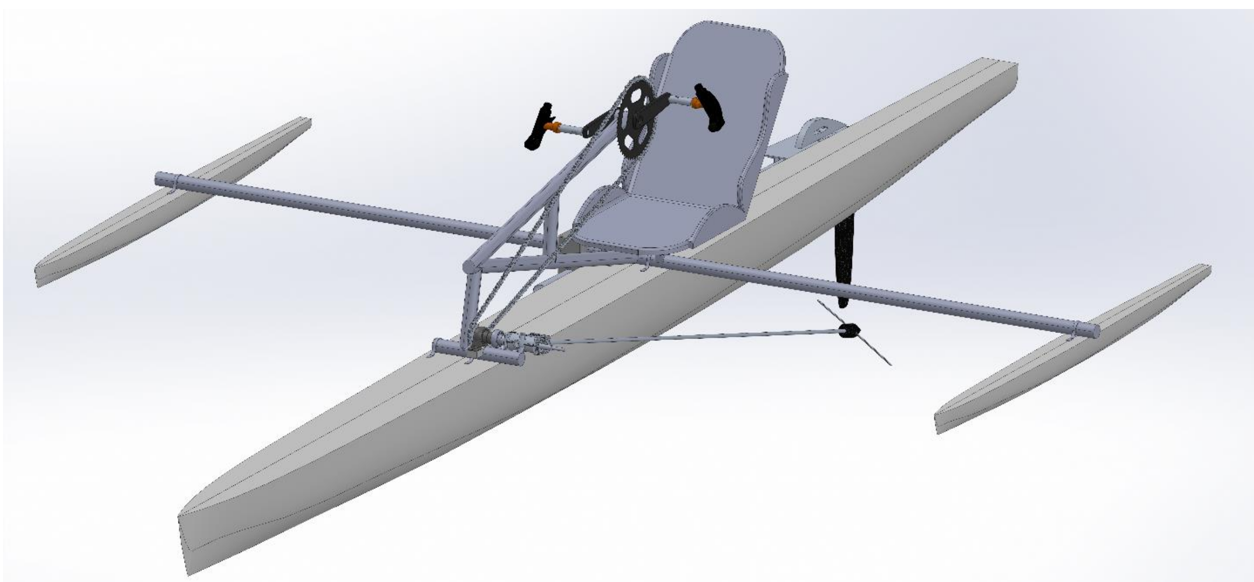

Figure 3.3. CAD design of a possible monohull.

The monohull would consist of a seat and propulsion system mounted on top of a single hull with a rudder and propeller running along the sides of the hull. To each side a small hull would be connected by a crosswise attached pipe. If the main hull is properly balanced, the small hulls would be clear of the water. Only when the boat is rolling to one of the sides, would one of the small hulls be submerged. The seat and backrest could be constructed with supportive sides to hold the user in place when the hull is rolling. While this design could improve the performance it would most certainly reduce the comfort and potentially safety. As this project intends to design human-powered watercrafts for people with lower-body disabilities, even including tetraplegics, it was decided to keep the catamaran design. In the future, it could be interesting to develop a monohull human-powered watercraft for people with lower-body disabilities but with a sufficient amount of trunk stabilization.

It was decided to further develop the catamaran design. To reduce the weight of the relatively heavy frame on the HWB v. 1, a frame of a modern handbike was modified to fit the transmission system and to be mounted on the hulls (Figure 3.4).

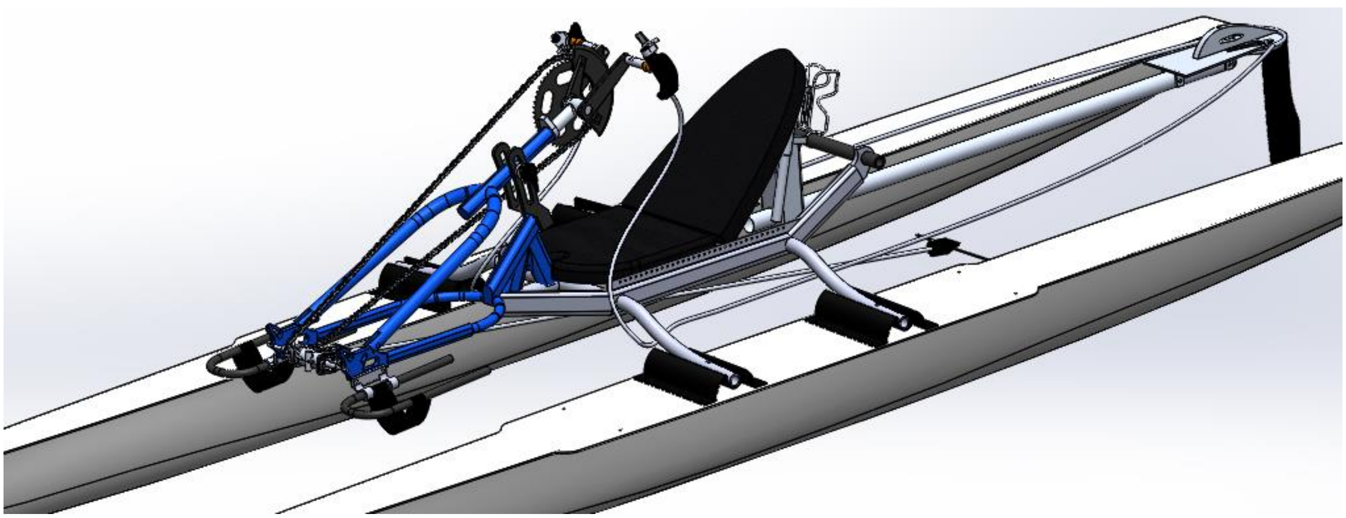

Figure 3.4. CAD design of the HWB V. 2.

The frame is lifted 10 cm by bending the pipes that are connected to the hulls to reduce the risk of water getting in contact with the user. The front fork, which connects the chain to the drive wheel on a handbike, was instead modified to fit the customized transmission system that drives the propeller. Most handbikes turn by rotating the front fork. As the turning mechanism on the HWB works by pushing a switch on the handles that turns the rudder, the front fork was locked to prevent unwanted rotation.

One of the major goals in the design of the HWB v. 2 was to improve the ease and speed of adjusting the seating position to fit the user. On the previous version specific tools and know-how were needed to adjust the seating position, whereas on the HWB v. 2 the backrest and front fork angle is easily adjusted by unlocking a handle. It is also possible to move the seat and footrests forward and backwards by use of a common hex key. In addition, a screen protector is attached at the end of the front fork steering tube to protect the chainring from getting in contact with the user. Finally, a water bottle cage was attached behind the seat.

The frame was designed to make it possible to attach an electric motor if needed. The motor could be connected to the transmission system and assist in providing propulsion. This could especially be useful for high level SCI with insufficient muscle force to generate propulsion on their own. The motor could also assist in situations where the HWB malfunctions or the user gets injured or fatigued.

Center of mass (COM) for the frame, transmission system and rudder was computed in SOLIDWORKS to determine the location of the hull-connecting pipes (Figure 3.5). If the COM is not located between the pipes it could cause a pitch motion of the watercraft in the water.

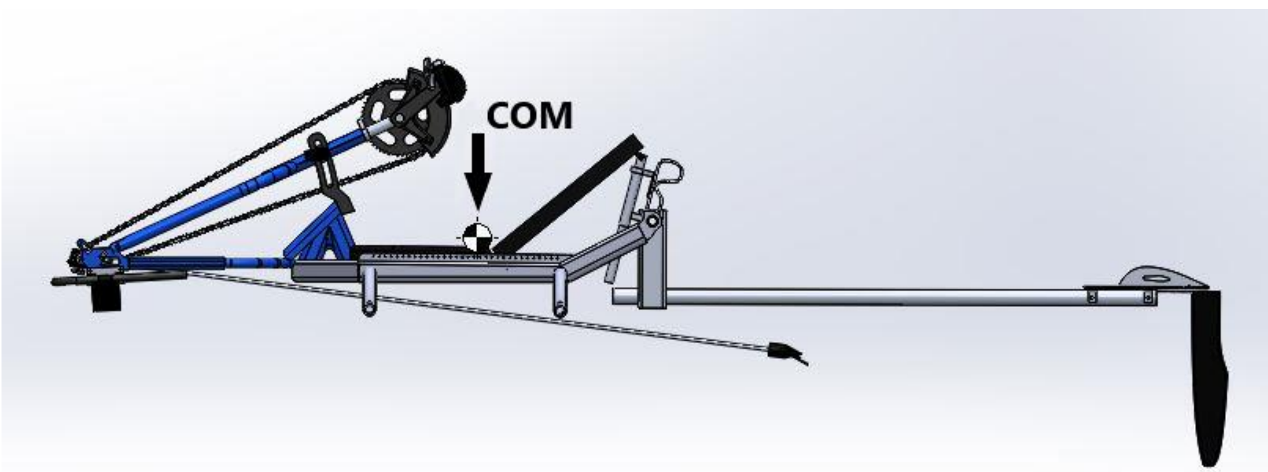

Figure 3.5. Center of mass computation in SOLIDWORKS.

The next step in the design cycle was to perform a finite element analysis (FEA) of the frame. The first step in a FEA is to convert the continuous mechanical system into a discrete system called a mesh. The mesh is a mathematical model of the CAD model, which consists of a large number of building blocks called finite elements. A denser mesh results in smaller and less distorted elements and a more accurate analysis. A parabolic (2nd order) tetrahedral curvature-based mesh was applied to the CAD model to create more elements in higher-curvature areas where the maximum stresses were expected to be located (Figure 3.6). The maximum and minimum element size was set to 23.5 mm and 4.7 mm respectively.

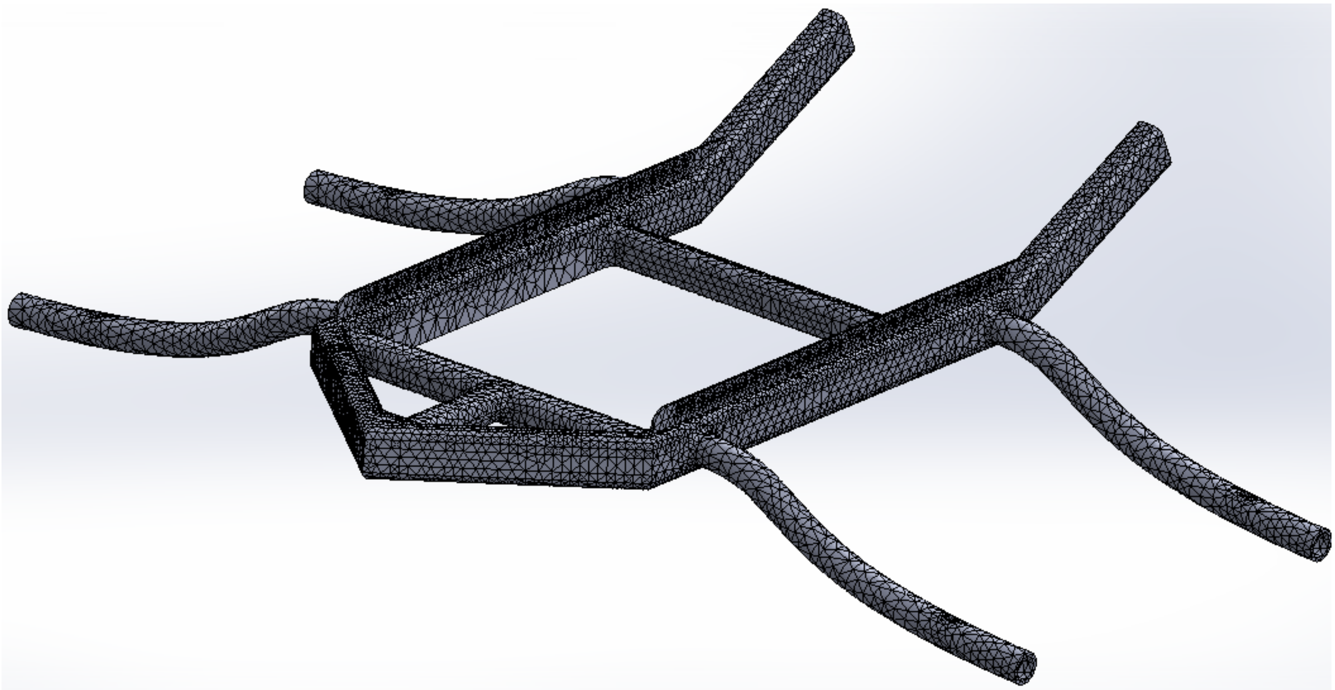

Figure 3.6. Finite element mesh of the frame designed for the HWB V. 2.

To investigate the structural behavior of the frame, a static loading condition was then simulated (Figure 3.7). A homogeneously distributed loading of 1000 N was applied over the surface of the central part of the frame. The material chosen for the frame was aluminum 7020 with a typical yield strength of 280 MPa. This specific alloy was chosen for its relatively high yield strength combined with a low cost. Less material is needed for the frame if the yield strength is high. A fixed geometry was applied at the end of the pipes to simulate the connection with the hulls.

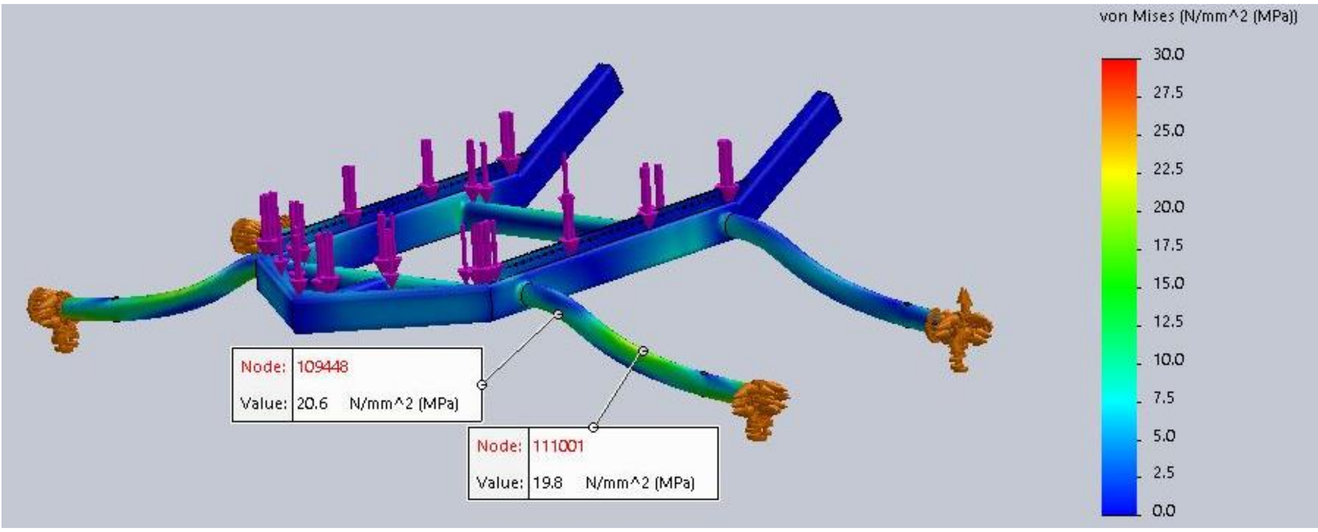

Figure 3.7. Results of the FEA showing the load, fixtures and von Mises stresses on a heat map.

For the analysis, von Mises stresses and maximum displacement were computed. In situations involving combined tensile and shear stresses on the same point, it is often convenient to define a stress that that can be used to represent the stress combination. von Mises effective stress does exactly that for ductile materials such as aluminum and is defined as "the uniaxial tensile stress that would create the same distortion energy as is created by the actual combination of applied stresses" (Norton, 2011). In SOLIDWORKS, von Mises stress was computed for each element or node. On the heat map in Figure 3.7 the von Mises stresses throughout the frame is shown. Most of the frame displays von Mises stress in the region 2.5-7.5 MPa. The two front hull-connecting pipes show larger von Mises stresses as two nodes or elements are computed to around 20 MPa. This is expected since the pipes are designed to bend and consequently will be affected by both normal and shear stresses, which is exactly what von Mises stress combines. Maximum displacement was computed to 0.63 mm.

It is critical that the structure can fully support the applied load to ensure safety. Factor of safety (FoS) is the ratio of the failure load to the allowable load:

$$FoS = \frac{Failure\ of\ stress}{Actual\ stress} = \frac{Strength\ of\ material}{Maximal\ computed\ stress}$$

In Figure 3.7, the maximal computed stress on the frame is shown to be 20.6 MPa. Aluminum 7020 has a typical yield strength of 280 MPa, which is the load where the material goes from elastic to plastic deformation. Based on this information, the dimensionless FoS was computed to asses the safety of the frame:

$$FoS = \frac{280\ MPa}{20.6\ MPa} = 13.6$$

If FoS > 1 the structure avoids failure. In most cases a FoS > 2 or even significantly higher is desirable due to different reasons (Hibbeler, 2017):

- Errors can occur in the fabrication or assembly of its component parts.
- Unknown and unexpected impacts, vibrations or accidental loadings can occur that was not accounted for in the design of the structure.
- Atmospheric corrosion, decay, or weathering tend to cause materials to deteriorate during service.

The safety of the frame with regards to stress failure was therefore considered satisfactory. Consequently, based on technical drawings, a physical prototype of the frame for the HWB v. 2 was manufactured in a workshop (Figure 3.8).

Figure 3.8. The physical prototype of the frame for the HWB v. 2.

Performance testing of the HWB v. 2

To compare the performance of the HWB v. 2 with the HWB v. 1, the former was exposed to the same experimental protocol as the HWB v. 1 in chapter 2, i.e. a deceleration test and an investigation of the relationship between mechanical power output and boat speed. The relationship between mechanical power output and boat speed was investigated by increasing the power from 30 W to 120 W by increments of 10 W every 10 seconds. This was done three times in each direction to account for the influence of water stream, and overall average result was used. External mechanical power was measured at a sampling rate of 60 Hz using a power-meter crankset (Quarg RIKEN R, SRAM, Spearfish, SD, USA) with a crank length of 170 mm.

The deceleration test was performed to measure and compare the hydrodynamic resistance of the watercrafts. A subject was instructed to pedal the boat at a constant speed of between 2.2-2.8 $m \cdot s^{-1}$ before letting go of the handles. This procedure was repeated along the same segment six times, three times per direction to account for the influence of water stream, and overall average result was used. The speed of the HWB was measured at a sampling rate of 1 Hz by means of a GPS receiver (Rider 20, Bryton Inc., Taipei City, Taiwan) fastened to the arm-crank system. Using the methods described by Bilo and Nachtigall (1980) and Capelli *et al*. (2009), the boat drag was calculated by analyzing the time course of the decreasing speed as a function of time. Data recorded for the first approximately 10 seconds during the deceleration tests were used for calculation of the hydrodynamic resistance.

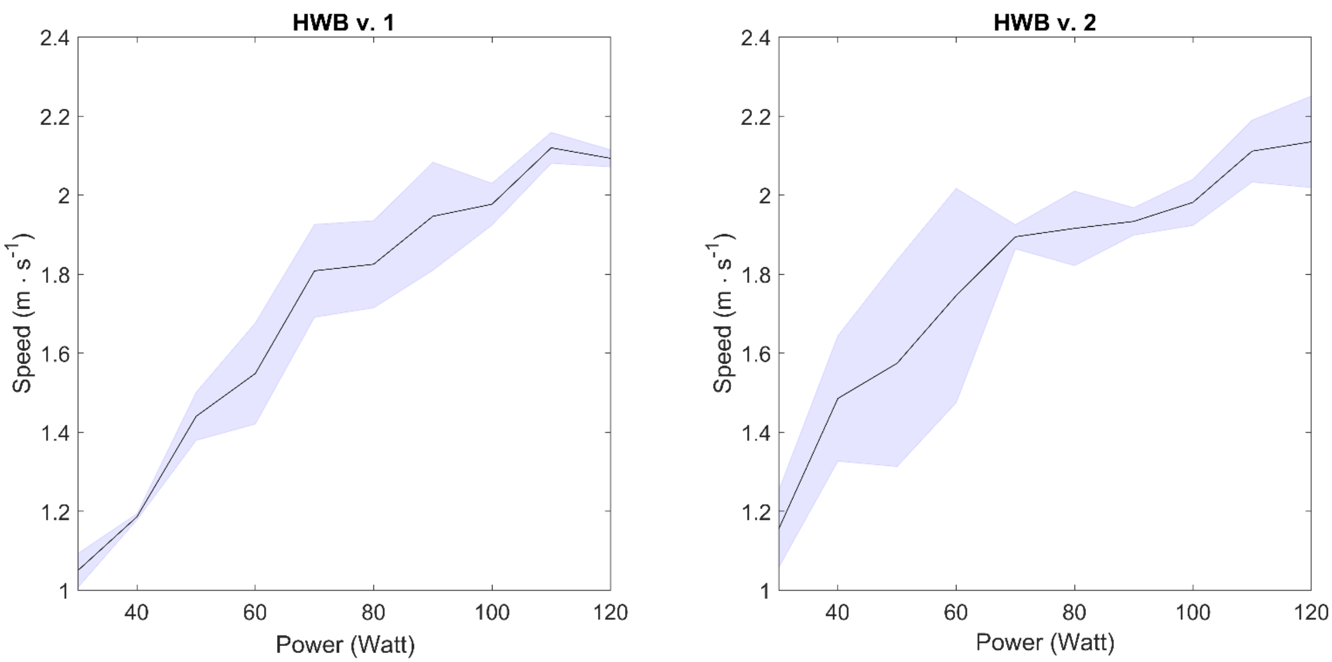

Figure 3.9. The relationship between mechanical power and speed for the HWB v. 1 (on the left) and HWB v. 2 (on the right). Data is the average of the six trials with standard deviation as the shaded area.

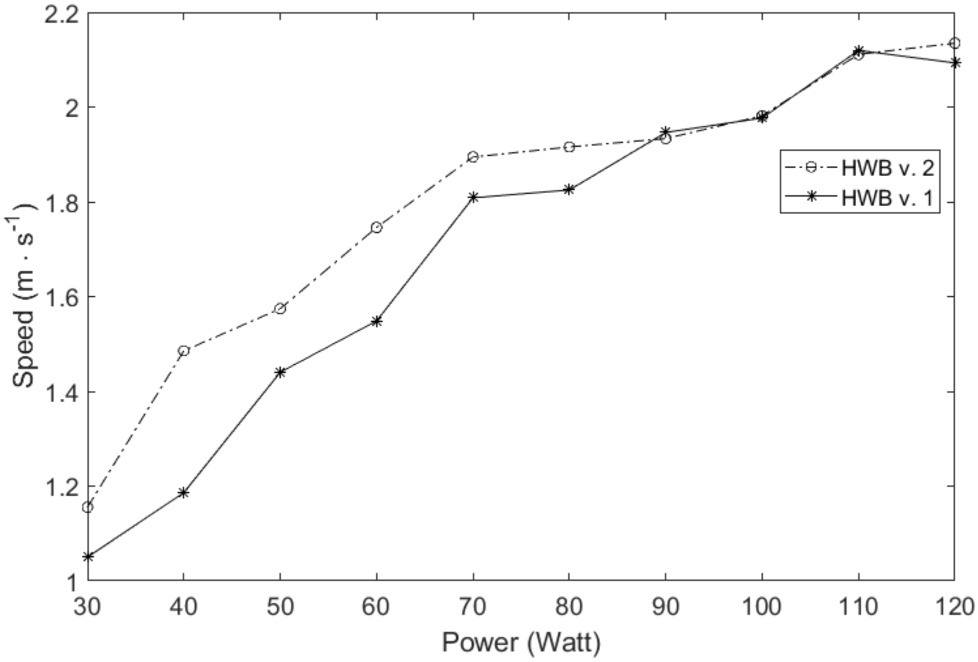

Figure 3.10. The relationship between mechanical power and speed of the HWB v. 1 and HWB v. 2 on the same graph.

An ANOVA with a 2x10 design (HWB type: HWB 1, HWB 2, and power: 30, 40, 50, 60, 70, 80, 90, 100, 110 and 120W) was applied to determine the effect of HWB type and power on speed. To determine (post-hoc) significant differences between HWB type at each power output, paired student t-tests were applied. All analyses were performed using the software IBM® SPSS® Statistics version 25.0 (IBM Corporation, Somers, NY, USA). The level set for significance was $p \leq 0.05$.

Table 3.1. Results of the ANOVA test.

	HWB type		Power		Interaction (HWB type x Power)	
	F	p-value	F	p-value	F	p-value
Speed	30.9	0.000	176.9	0.000	4.05	0.001

The analysis, shown in table 3.1, revealed a significant effect of both HWB type and power on the speed, and a significant interaction between HWB type and power was found as well. Post-hoc analysis showed that the effect of HWB type (i.e. the difference between HWB v.1 and HWB v. 2) was significant only at power outputs of 30, 40, 60 and 70 watts as seen in table 3.2. Since the p-value at

50 watts is close to significance (p=0.065) and between other significant values, it seems reasonable to hypothesize that a significant value would have been calculated in a greater or different sample for 50 watts as well. The analysis indicates that the HWB v. 2 performs faster at low power outputs but not at higher where it seems both watercrafts reaches a maximum speed of approximately 2.2 m·s^{-1}. This could be due to the propellers inability to increase the pedaling frequency further than what is reached at this speed. The relationship between power and speed on the watercrafts is not linear but rather logarithmic since water resistance is increased approximately four times when speed is doubled (Hill & Fahrig, 2009).

*Table 3.2. Results of the post-hoc analysis of ANOVA for repeated measures. Significant differences are marked *.*

Power	30 vs. 30	40 vs. 40	50 vs. 50	60 vs. 60	70 vs. 70	80 vs. 80	90 vs. 90	100 vs. 100	110 vs. 110	120 vs. 120
p-value	0.023*	0.015*	0.065	0.001*	0.042*	0.216	0.903	0.685	0.877	0.535

The better performance of the HWB v. 2 was assumed to be due to a lower hydrodynamic resistance from the lighter frame. The water drag (D) was calculated using the formula from Capelli *et al*. (2009):

$$D = C_D \times A \times \rho \times v^2/2$$

Where C_D is the drag coefficient, A is the submerged frontal area, ρ is water density and v is boat speed. To obtain C_D the average slope of the reciprocal of the decreasing speed recorded during the deceleration trials is needed (Capelli *et al.*, 2009). The time course of a typical trial is shown in Figure 3.11. The average slope of the linear regressions calculated from the six trials was $0.039 \pm 0.008\ s \cdot m^{-1}$. Together with this value, the overall mass of the boat, the maximal frontal submerged area and the water density, the dimensionless C_D was calculated as 0.208 using the method of described by Capelli *et al*. (2009). Water drag of the HWB v. 2 could then be described by the following equation:

$$D = 2.49v^2$$

The proportionality coefficient between water drag and the square of the speed turns out to be 2.49 for the HWB v. 2 compared to 5.64 for the HWV v. 1.

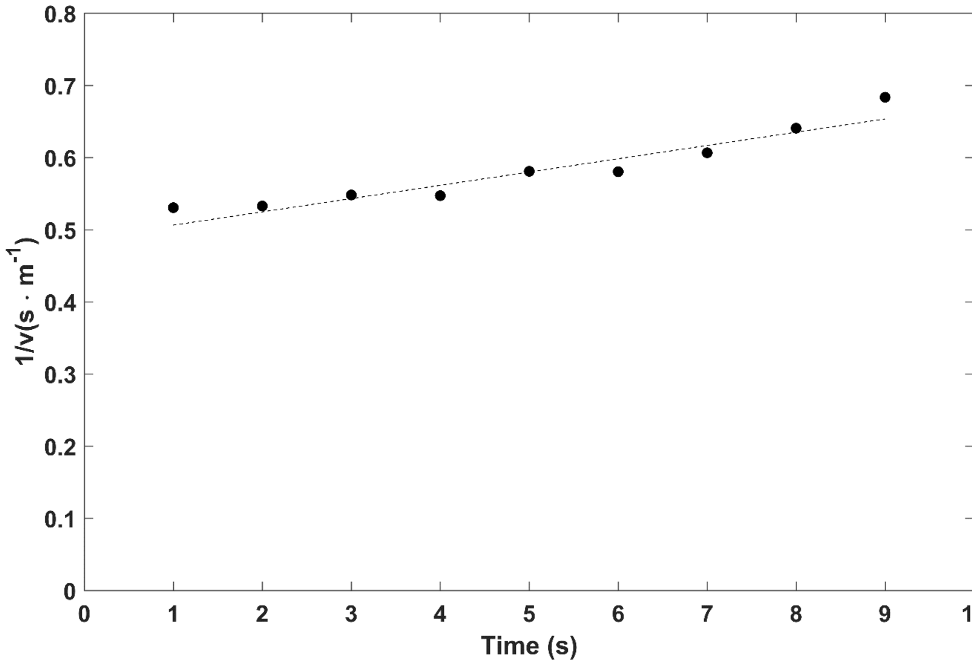

Figure 3.11. The reciprocal of the decreasing speed (v) obtained during a typical experiment of spontaneous deceleration plotted as a function of time.

In this chapter it was described how the Handwaterbike v. 2 was designed and analysed using finite element method to investigate the structural behavior. A physical prototype was built and exposed to a number of tests to measure the hydrodynamic resistance, and to determine the relationship between mechanical power output and speed. It was shown that the Handwaterbike v. 2, in addition to adding a number of extra features, performed better than the previous model in the performance tests.

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Chapter 4

A comparison of synchronous and asynchronous arm cranking on the Handwaterbike

Introduction

For people with lower body disabilities, arm crank exercise can be an important and effective tool to prevent cardiovascular diseases (Abel *et al*., 2003), and a recent study even suggested arm cranking to improve walking ability in chronic stroke patients (Kaupp *et al*., 2018). The most common arm crank exercise is handbiking, but recently a new watercraft, the so called Handwaterbike was developed (Fuglsang *et al*., 2017). The Handwaterbike is a human-powered watercraft, that permits people with lower body disabilities to exercise on water, using arm cranking mechanics similar to a handbike's one.

In modern handbiking, due to better steering ability, the crank arms typically turn in the same direction simultaneously, resulting in a synchronous motion. Most arm crank ergometry studies, however, have used an asynchronous arm cranking set up, where the cranks are positioned 180° from each other (Abel *et al*., 2003; Dallmeijer *et al*., 2004; Goosey-Tolfrey and Sindall, 2007; van der Woude *et al*., 2008). Studies have shown that the crank mode can influence the physiological responses and exercise performance during arm cranking. Several recent studies have found a higher mechanical efficiency and a lower oxygen uptake during synchronous arm cranking compared to asynchronous arm cranking (Abel *et al*., 2003; Dallmeijer *et al*., 2004; van der Woude *et al*., 2008; van der Woude *et al*., 2000). Other studies did not find any differences in physiological responses between the two crank modes (Hopmann *et al*., 1995; Mossberg *et al*., 1999), whereas Goosey-Tolfrey *et al*. (2007) found a higher mechanical efficiency and a lower oxygen uptake during asynchronous arm cranking compared to synchronous arm cranking. It has been suggested that the contradictory findings are a result of different test protocols and subjects (Goosey-Tolfrey *et al*., 2007). Experienced handbikers for example would be expected to prefer synchronous arm cranking since they are more accustomed to this crank mode.

The pedaling cycle of an arm crank revolution is similar to that of cycling and can divided into a pulling and a pushing phase. Arnet el al. (2013) and Faupin *et al*. (2010) found that most of the tangential force, i.e., the force applied to the handles of the handbike that is perpendicular to the crank and creates propulsion, was produced in the pulling phase. An irregular force application to the handles could create velocity fluctuations, which are known from other water activities such as rowing (Hill and Fahrig, 2009; Hofmijster *et al*., 2007; Michael *et al*., 2009). Velocity fluctuations increase the water resistance and occur due to the drive and recovery phases during the rowing cycle (Hill and Fahrig, 2009). When velocity is doubled, water resistance is increased four times since it is approximately proportional to v^2 (Brearley *et al*., 1998). The increase in water resistance, when the boat velocity is larger than the mean velocity of a rowing cycle, is therefore greater than the decrease in water resistance when the boat velocity is below the mean velocity. Theoretically, reducing the velocity fluctuations should consequently result in higher average velocity at a given power output. Synchronous pedaling might also cause greater periodical pitching resulting in lost mechanical work and increased water drag. Contrary, asynchronous pedaling could cause greater yaw motion of the watercraft if irregular force is applied to the handles, which also results in wasted mechanical work.
The aim of this study was therefore to compare the physiological responses and boat speed of synchronous and asynchronous arm cranking on the Handwaterbike.

Methods and Materials

Subjects

The experiments were carried out on seven able-bodied subjects (28.5 ± 6.4 years; 77.4 ± 8.3 kg; 1.78 ± 0.04 m). The subjects had no experience with handbiking or injuries in the upper extremities. All subjects gave their written informed consent before testing, and were thoroughly informed about the purpose, benefits, and potential risks of the study, in conformity with the Code of Ethics of the World Medical Association (Declaration of Helsinki). The protocol and the methods applied in the study were approved by the Ethical Committee of the Department of Neurosciences, Biomedicine and Movement Sciences, University of Verona.

Watercraft

The watercraft, named Handwaterbike, was a modified version of the watercraft previously described by Fuglsang *et al* (2017). It is a catamaran intended for use by people with lower body disabilities. The arm cranking mechanics is similar to handbiking, as a synchronous (or asynchronous) arm-crank system in front of the seat is connected *via* a roller chain to the transmission system, which drives a flexible shaft and the propeller. The crank height was set to ensure that the scapula-humeral joint and the crank pedal axle were on the same level. The main dimensions of the Handwaterbike without subjects were: length overall: 4.89 m; length of water line: 4.75 m; weight: 63.89 kg; maximal beam: 1.08 m; draught: 0.09 m.

Experimental protocol

The experiments were performed along the shore of the Garda Lake (Italy) in calm water and wind speeds below 2 $m \cdot s^{-1}$. After 5 min of familiarization with both synchronous and asynchronous crank mode, the subjects were asked to pedal in a linear direction and perform three submaximal exercise trials of 5 min duration each at power outputs of 30, 50 and 70 W. A maximum of 70 W was chosen to ensure mainly aerobic metabolism occurred. In random order, each trial was performed once with synchronous arm cranking and once with asynchronous arm cranking. Before the test, the subjects were asked to sit on a chair for 5 min for measurement of metabolic variables at rest. The subjects were allowed to begin a new trial after 5 min of rest, when the heart rate was below 100 bpm, and when they communicated their availability to re-start.

Measurements

Oxygen consumption ($VO2, \mathrm{L} \cdot \mathrm{min}^{-1}$), carbon dioxide production ($VCO2, \mathrm{L} \cdot \mathrm{min}^{-1}$), and heart rate (HR, bpm) were assessed breath-by-breath using a portable metabographic system (K5, COSMED, ROME, Italy). The system was calibrated according to the manufacturer´s instructions previous to each test. The Handwaterbike was instrumented with a power-meter crankset (Quarq RIKEN R, SRMA, Spearfish, SD, USA) with a crank length of 170 mm in order to measure pedaling frequency ($rev \cdot min^{-1}$) and external mechanical power (W) at a sampling rate of 60 Hz. Boat speed was measured at a sampling rate of 1 Hz by means of a GPS receiver (Rider 20, Bryton Inc., Taipei City, Taiwan) fastened to the arm-crank system.

Energy expenditure was calculated from $VO2$ and converted to joules · min^{-1} by using the associated respiratory exchange ratios in accordance with Peronnet and Massicotte (1991).

Net efficiency (NE) was calculated according to the method of Hintzy and Tordi (2004):

$$Net\ efficiency = \frac{Power\ Output}{Energy\ expenditure - energy\ expenditure\ at\ rest} \times 100$$

Data analysis

Only data collected from the last minute of each trial were used for analysis. Coefficient of variation (CV) of oxygen consumption data for the last minute of each trial was compared to the preceding minute of each trial to ensure steady-state was reached. All mean CV was < 10%, which indicates that steady-state was reached. All descriptive data are presented as means and standard deviation. All analyses were performed using the software IBM® SPSS® Statistics version 25.0 (IBM Corporation, Somers, NY, USA). An ANOVA for repeated measures with a 2 × 3 design (crank mode: synchronous, asynchronous and Power: 30, 50 and 70 W) was applied to determine the effect of crank mode and power output on the physiological parameters and boat speed. Assumption of normality was assumed using the Shapiro-Wilk test. The level set for significance was $p \leq 0.05$.

Results

The mean values and standard deviations for $VO2$, HR and net efficiency are presented in table 4.1. For the net oxygen uptake (above that measured at rest) there were no differences between modes of cranking although slightly higher net oxygen uptake was observed during synchronous cranking at 50 W (0.58±0.09 vs. 0.55±0.09 l·min^{-1}). No significant differences were found for boat speed, heart rate (HR), net efficiency (NE) and pedaling frequency either (table 4.1). As anticipated, net oxygen uptake, heart rate, and boat speed increased with rising power output (Figure 4.1, 4.2 and 4.4) ($p \leq 0.05$). For net efficiency there was no main effect for power output (Figure 4.3).

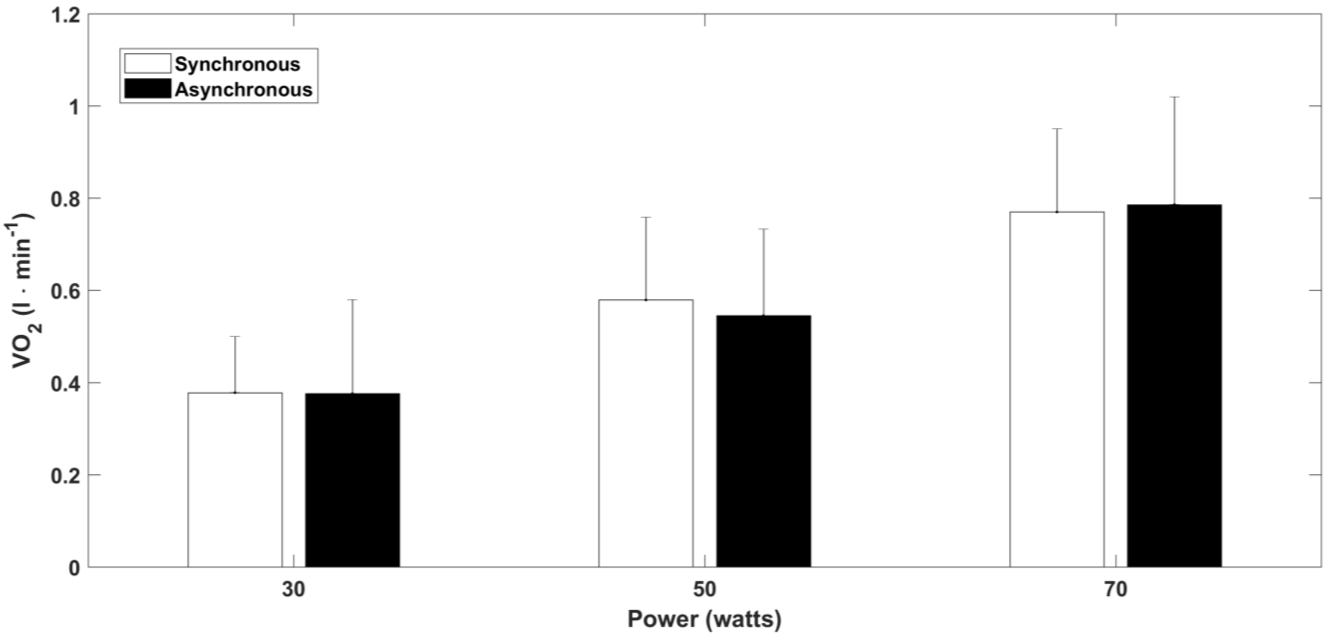

Figure 4.1. Relationship between net oxygen uptake and power output during synchronous and asynchronous arm cranking at 30, 50 and 70 W.

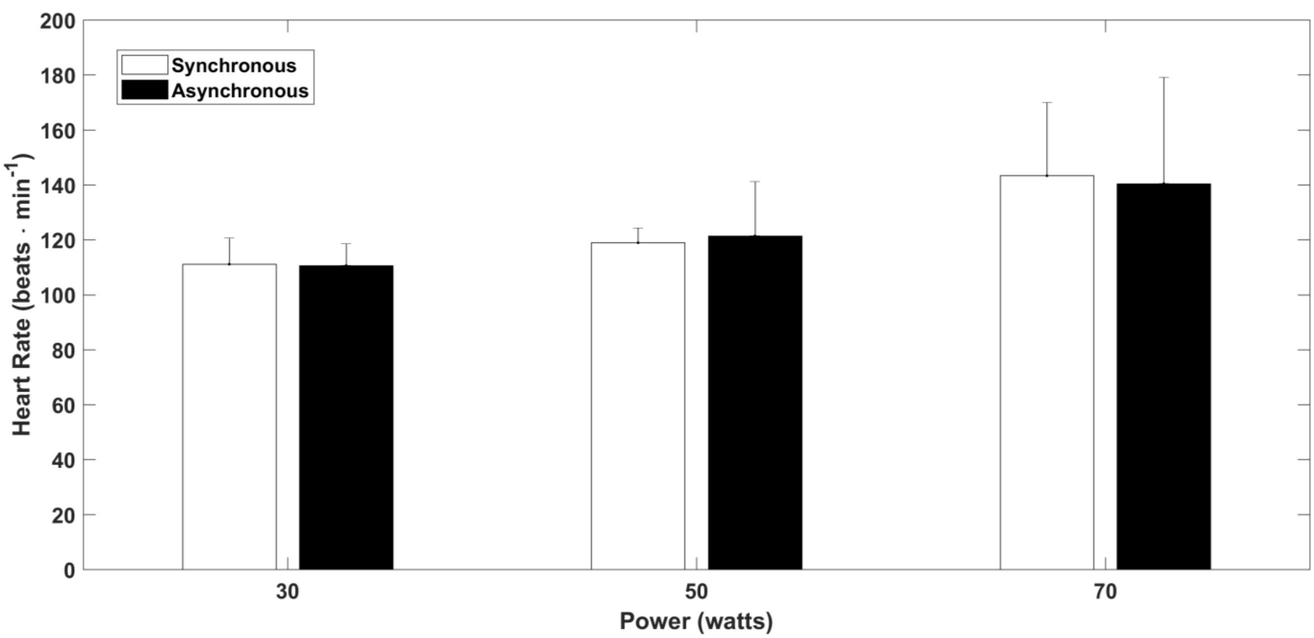

Figure 4.2. Relationship between heart rate and power output during synchronous and asynchronous arm cranking at 30, 50 and 70 W.

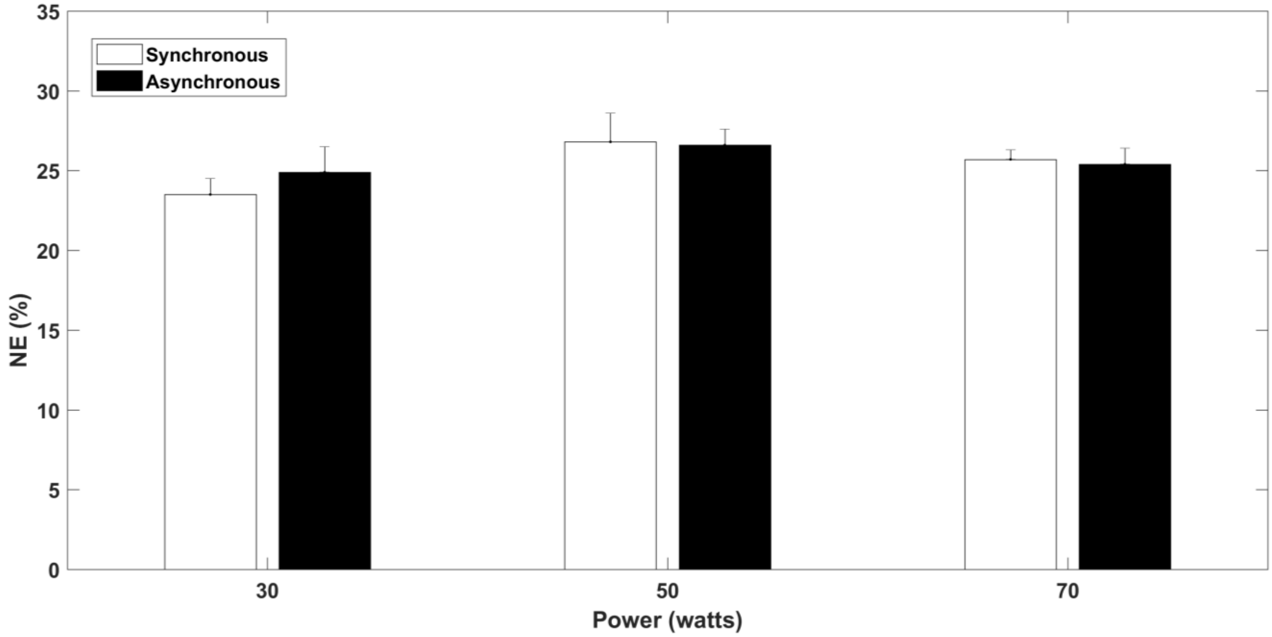

Figure 4.3. Relationship between net efficiency and power output during synchronous and asynchronous arm cranking at 30, 50 and 70 W.

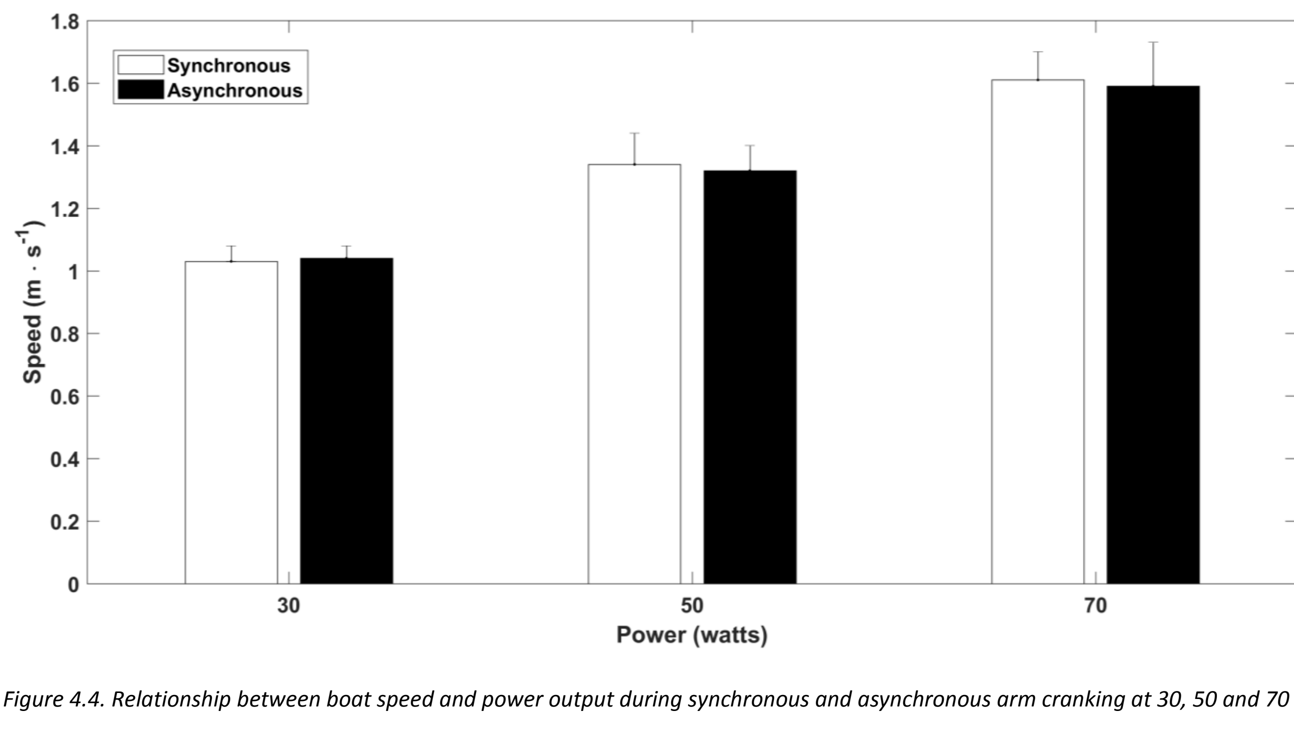

Figure 4.4. Relationship between boat speed and power output during synchronous and asynchronous arm cranking at 30, 50 and 70 W.

Table 4.1. Physiological responses to synchronous and asynchronous arm cranking on the HWB. VO2 = Net oxygen uptake; HR = Heart rate; NE = Net efficiency; Freq.= Pedaling frequency. Values are mean ± SD.

Variable	Power Output (w)	Synchronous	Asynchronous
VO2 ($l\cdot min^{-1}$)	30	0.38 ± 0.06	0.38 ± 0.10
	50	0.58 ± 0.09	0.55 ± 0.09
	70	0.77 ± 0.09	0.77 ± 0.11
HR ($beats\cdot min^{-1}$)	30	111.13 ± 4.77	110.63 ± 3.95
	50	118.97 ± 2.69	121.40 ± 9.87
	70	143.34 ± 13.33	140.44 ± 19.36
NE (%)	30	23.5 ± 0.5	24.9 ± 0.8
	50	26.8 ± 0.9	26.6 ± 0.5
	70	25.7 ± 0.3	25.4 ± 0.5
Speed ($m\cdot s^{-1}$)	30	1.03 ± 0.05	1.04 ± 0.04
	50	1.34 ± 0.10	1.32 ± 0.08
	70	1.61 ± 0.09	1.59 ± 0.14
Freq. ($rev\cdot min^{-1}$)	30	23.25 ± 5.91	24.12 ± 5.29
	50	32.74 ± 4.57	31.57 ± 5.20
	70	41.05 ± 3.50	40.82 ± 4.26

Discussion

The purpose of the study was to compare physiological responses and boat speed of the Handwaterbike during synchronous and asynchronous submaximal aerobic arm cranking tests. In agreement with Hopman *et al*. (1995) and Mossberg *et al*. (1999), we found no significant differences in the physiological responses between the two modes of cranking. Smith *et al*. (2008) has previously found no difference in electromyogram activity of selected muscles of the arm, shoulder, and legs during synchronous and asynchronous arm cranking, which supports our findings. Differently, Abbasi-Bafghi *et al*. (2008) found increased muscle activity in the *m. obliquus externus abdominis* during asynchronous arm cranking. It should be noted that studies have shown that differences between synchronous and asynchronous performance largely depends on the exercise intensity, exercise protocol, and choice of subjects (Abbasi-Bafghi *et al*., 2008; Smith *et al*., 2008; van der Woude *et al*., 2008). In the present study, non-disabled subjects were tested. Non-disabled subjects are more likely to have an equal level of inexperience in arm cranking and a lack of a pre-existing preference for a specific arm crank configuration. Wheelchair dependent people are already used to produce propulsion by a synchronous push on the wheelchair handrims and may even be trained and experienced in handcycling which nearly always deploys a synchronous cranking system. Consequently, it is reasonable to expect differences in the physiological responses between disabled and non-disabled subjects. Even within disabled subjects, differences could be expected depending on level of injury. It is likely that tetraplegics would encounter difficulties in asynchronous arm cranking due to the required trunk muscle activation (Abbasi-Bafghi *et al*., 2008), which would lead to differences in physiological responses when comparing the two arm cranking modes. In synchronous arm cranking, the trunk muscles are activated to help with propulsion whereas, as indicated by Abbasi-Bafghi *et al*. (2008), the trunk muscles also act to the stabilize the trunk along the longitudinal axis during asynchronous arm cranking. In the present study, however, the power output may not have been high enough to require a significant trunk stabilization effort. It has been suggested that trunk stabilization is especially needed when using non-fixed ergometers such as handbikes on a treadmill as steering is an added component (Dallmeijer *et al*., 2004). The studies by Hopman *et al*. (1995) and Mossberg *et al*. (1999), both using fixed arm ergometers, also did not find any differences between the two crank modes. When using the Handwaterbike the subjects were required to steer the watercraft by pushing the gear-shifters rather than the fork steering on a handbike. This steering mechanism might not require the same trunk muscle activation as when using handbikes and could help explain the lack of differences in physiological responses to synchronous or asynchronous arm cranking in this study.

Contrary to our expectations, we did not find any differences in boat speed when comparing synchronous and asynchronous arm cranking on the Handwaterbike. Torque analysis has shown that torque pattern during the crank cycle is different for synchronous and asynchronous arm cranking (Smith *et al*., 2008). Intuitively, an irregular force application to the handles of the Handwaterbike could create additional boat movements. Linear longitudinal translational motion and rotational pitch could be expected during synchronous arm cranking, while rotational roll and yaw could be expected during asynchronous arm cranking, and lead to lower efficiency due to higher increased hydrodynamic drag. Our results suggest that no additional boat movements were created from irregular force application in one crank configuration compared to the other, however we cannot be certain about that. It is possible that inertial measurement units would have detected additional boat movements, but that the movements were not sufficient to create differences in boat speed. We did not however, find any differences in pedaling frequencies and since power output is calculated by multiplying torque and angular velocity, it can be assumed that the applied torque was similar between the two cranking modes as well since power output was fixed. This does not exclude irregular force applications occurring during the crank cycle, but indicates a similar net torque for the two cranking configurations.

In the study by Smith *et al*. (2008), they found different torque patterns at 100 W arm cranking. In the present study the maximum power output was 70 W. It is possible that a higher output would have created differences in additional boat movements resulting in differences in boat speed. Another possibility is that the, in theory, added pitch motion from synchronous arm cranking on the Handwaterbike corresponds in some way to the added yaw motion from asynchronous arm cranking. This could be investigated by mounting gyroscope sensors on the watercraft during testing.

Conclusion

In conclusion there were no differences, under the current test conditions, in physiological responses or boat speed when comparing synchronous and asynchronous arm cranking on the Handwaterbike. When navigating the watercraft, it should therefore be up the individual user which crank configuration he or she uses. If the user is already trained and experienced in handcycling, synchronous arm cranking is suggested, however.

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Chapter 5

The influence of backrest inclination on interface seat pressure during arm cranking – a case study

Introduction

A pressure ulcer (PU) is a general term for ulceration of the skin or underlying tissue, usually over a bony prominence (Black et al., 2007), and can be divided into four categories from redness of skin to missing tissue and exposed bone (Dolbow et al., 2013). PUs can affect the quality of life significantly (Dolbow et al., 2013) and it is estimated that approximately one third of people with spinal cord injury (SCI) will develop at least a PU over their lives (Lim et al., 2007) and up to 80% of those who develop a PU will have a recurrence (Dolbow et al., 2013). In addition, PUs are a major cost for the healthcare system as treatment is responsible for approximately 25% of the cost associated with SCI in the United States (Dolbow et al., 2013). Since PUs develop quickly but heal slowly, it is important to focus on prevention rather than treatment (Chenu et al., 2013).

Typically, PUs develop when mechanical loading is applied to a small body area, which leads to necrosis, either by ischemia or deformation of cells (Olesen et al., 2010). The formation of PUs however, is complex since many types of mechanical loading during activities of daily living, such as sitting in office chairs or riding a bike, does not lead to formation of PUs. Even though the underlying mechanism behind PU development is unclear, PUs do not develop without pressure on the tissue. As a result, people who spend extended periods of time sitting or lying down are at risk of developing PUs (Bartley & Stephens, 2016). A common observation however, is that people, after a certain duration of sitting, tend to make small movements in the chair, which could be the human body's response to relieve pressure at bony prominences such as the ischial tuberosities and *sacrum* (Kumar et al., 2015). Because of paralysis and a loss of sensory feedback, people with SCI do not experience the same discomfort and urge to relieve pressure, and consequently they are at a higher risk of developing PUs (Giesbrecht et al., 2011; Sprigle & Sonenblum, 2011).

It is recommended that people with SCI change position every two hours and perform pressure relief procedures for 30 seconds every 15 to 20 minutes when seated (Ho & Bogie, 2007), by pushing down on the seat or armrests to lift the buttocks of the supporting surface, or by leaning forward or to the side. In certain situations, such a regime can be difficult, for example when doing exercise such as handcycling, sit-skiing, or water activities (Fuglsang et al., 2017). Especially handcycling is a popular recreational and Paralympic sport. The athlete creates propulsion of the handbike by turning an arm-crank system, while either kneeling or sitting on a reclined seat. Currently, handcycling is essentially performance driven as engineers and biomechanists aim to develop designs that optimize the performance while satisfying the specific needs and capabilities of the athlete (Rice, 2016). Little consideration is put on comfort however, and even though pain and discomfort may not be felt, PUs can still develop from excessive loading or repeated frictional movement. The semi-reclined position is often used in handcycling as it is more aerodynamic than the upright position. Several studies have come to the conclusion that the semi-reclined position increases the risk of PU development in hospital beds (Hanson et al., 2010; Sprigle & Sonenblum, 2011), chairs (Defloor & Grypdonck, 1999) and wheelchairs (Kobara et al., 2014). The risk of PU in this position is mainly elevated, because the seated person will have a tendency to slide down and out of the bed or chair, which increases the shear force loaded onto the buttocks (Hanson et al., 2010). Furthermore, during exercise local skin temperature can rise and cause perspiration, which leads to skin softening and adhesion, and thereby a higher risk of PU development (Hanson et al., 2010). To the authors´ knowledge, no studies have been made on the measurement of interface pressure when exercising on a handbike seat.

The aim of this study is therefore to investigate the interface seat pressure of a handcyclist at different backrest inclinations when arm cranking.

Material and Methods

Subjects

One female (age: 45 years, height: 165 cm, mass: 45 kg) gave her informed consent and participated in the study. The subject was an Italian handcycling champion in the H1 classification due to a spinal cord injury in the C6-C7 *vertebrae*.

Materials

Interface pressure measurement is the most used method to predict the risk of PU development, to analyze and understand different sitting postures, and to compare different supporting surfaces (Reenalda et al., 2009). This method measures the pressure between the skin and the supporting surface, and is often used to compare different wheelchair cushions (Dolbow et al., 2013). In this study it was used to identify the interface pressure of a handbike seat at different inclinations while the subject was arm cranking.

To measure the interface pressure, a force sensing system similar to the one described and validated by the authors of a seat adaptation system was used (Bertolotti et al., 2011). Twenty Force Sensing Resistors (FSR) by Interlink Electronics Inc. (Camarillo, CA, USA) have been arranged in a 4 x 5 matrix. A FSR is a polymer thick film device, which exhibits a decrease in resistance with an increase in the force applied to the active surface. The FSRs used have a square-shaped active area of 16 cm^2 and the centre-to-centre horizontal and vertical distances are approximately 8 cm. FSR matrix, interconnections and resistors are covered with two sheets (one per side) of imitation leather (Figure 5.1).

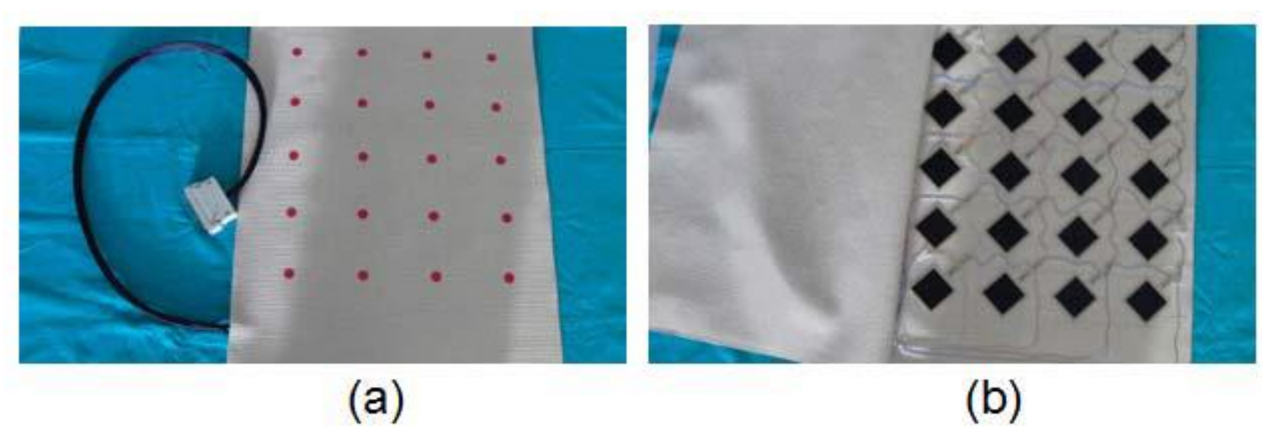

Figure 5.1: Force sensing system: a) external view; b) inside view.

The force sensing resistors are connected to the microcontroller-based acquisition unit by means of a proper conditioning circuit (a series of voltage dividers and amplifiers). All 20 force values were collected with a sample frequency of 25 Hz and the instrument was calibrated before use.

Oxygen consumption (VO2) and respiratory exchange ratio (RER) were also assessed breath-by-breath using a portable metabographic system (K5, COSMED, ROME, Italy). The system was calibrated according to the manufacturer´s instructions prior to use.

Experimental protocol

A modified arm crank ergometer (Figure 5.2) was equipped with the force sensing system on the handbike seat. The subject was then able to complete two trials with different ergometer setups, during which pressure data were recorded. For the first trial the backrest was set to a 60° angle with respect to the horizontal plane and the crank height was set to ensure that the scapula-humeral joint and the crank arm axle were on the same level. After a 5-minute self-chosen warm-up, the subject was asked to rest in the seated position with the hands holding the handles for a static measurement of the interface pressure. The duration of the static measurement was 5 minutes, as it takes up to 4 minutes before static interface pressure measurements stabilize (Crawford et al., 2005). Following the static measurement, the subject was then asked to arm crank at 40 W for 5 minutes, during which pressure data and oxygen consumption were recorded. After this trial, the subject had a 5-minute rest period and the backrest angle was adjusted to 30° with respect to the horizontal plane and the crank height was again set to ensure that the scapula-humeral joint and the crank arm axle were on the same level. Another 5-minute static measurement was then recorded followed by a 5-minute measurement of arm cranking at 40 W.

Figure 5.2: A modified arm crank ergometer equipped with a handbike seat and backrest.

Data analysis

All data processing was performed using MATLAB R2017a. (MATLAB, MathWorks, Natick, Massachusetts, USA). Only data collected from the last 30 seconds of each trial were used for analysis. Coefficient of variation (CV) of interface pressure data for the last 30 seconds of each rest trial was compared with the preceding 30 seconds of each trial to ensure stabilized interface pressure measurements were reached. All mean last-30 second CVs were < 5%, which indicates that the interface pressure measurements were stabilized during the rest trials. Peak pressure was defined as the peak value that occurred from any single sensor during the rest and arm cranking trials. The peak pressure was located in the region of the *sacrum* in all trials. Average pressure was defined as the average value of all sensors reading greater than or equal to 1 mmHg to exclude all sensors reading 0 mmHg. Contact area was calculated from the total number of sensors under load.

In order to have a better definition of the pressure maps, a bilinear interpolation was adopted to transform the original 4 x 5 matrix into a 7 x 9 grid according to the method of Marenzi et al. (2013).

The interpolation estimates the pressure values placed exactly halfway between two sensors, making a smoother distribution.

Energy expenditure was calculated from VO2 and RER, and converted to joules · min^{-1} by using the associated respiratory exchange ratios in accordance with Peronnet and Massicotte (1991).

Gross mechanical efficiency (ME) was calculated according to the method of Hintzy and Tordi (2004):

$$Gross\ Mechanical\ efficiency = \frac{Power\ Output}{Energy\ expenditure} \times 100$$

Results

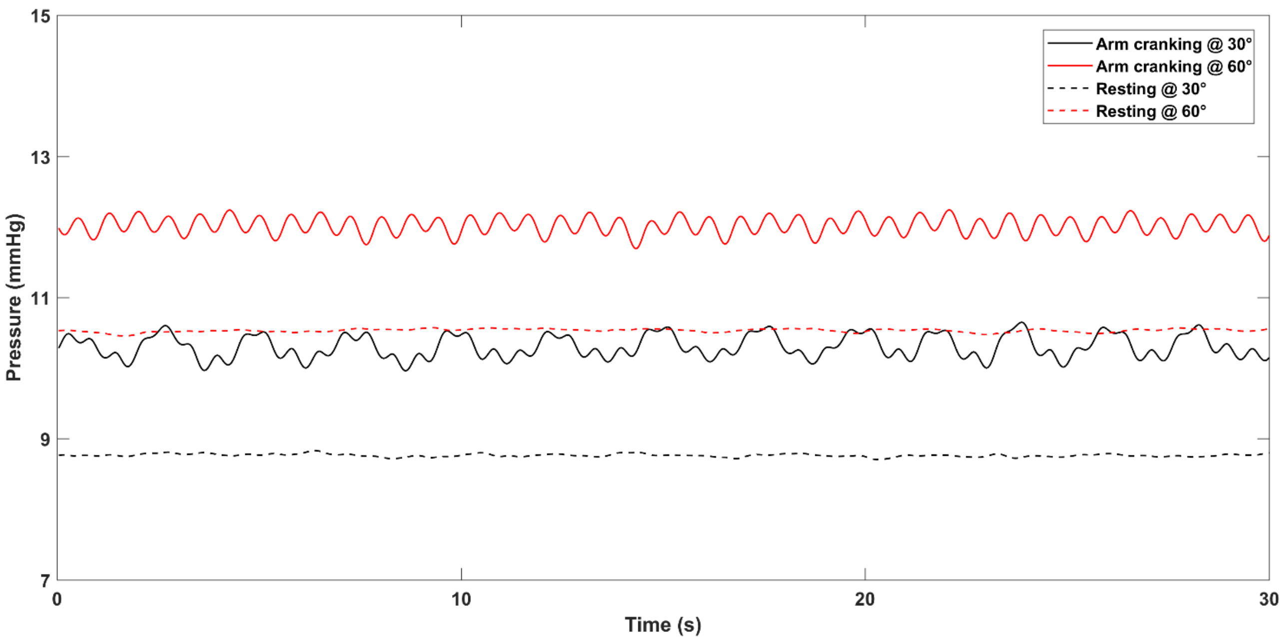

Figure 5.3: Comparison of peak pressure during arm cranking and resting at backrest angles of 60° and 30°.

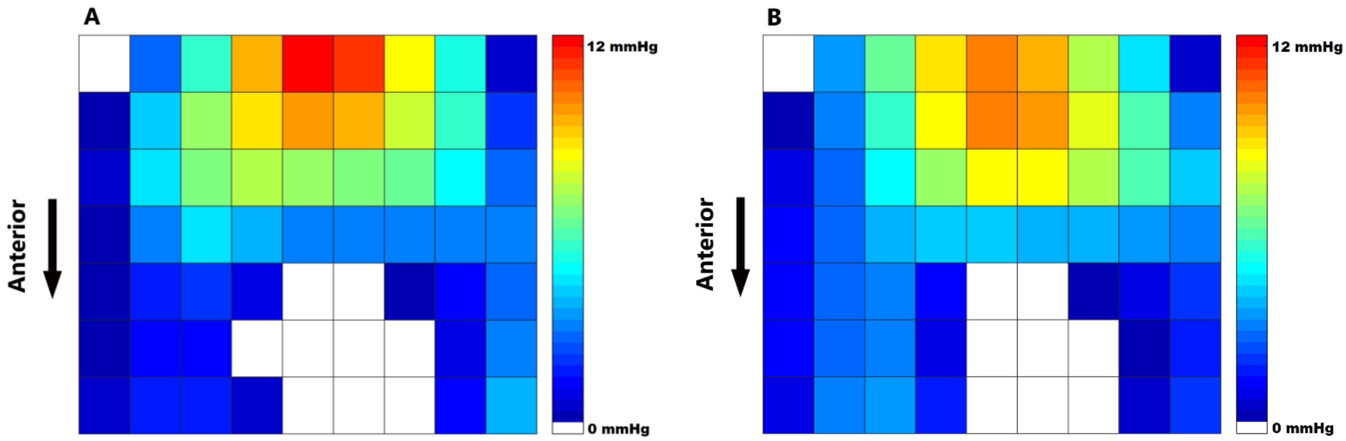

Figure 5.4: Pressure maps of the seat during rest at backrest angles of 60° (A) and 30° (B).

In Figure 5.3 it can be seen that changing backrest angle from 30° to 60° changes the peak pressure as average peak pressure increased by 1.72 mmHg during the arm cranking trial and by 1.77 mmHg during the rest trial (table 5.1). On the pressure maps it can further be seen that the peak pressure for the subject was located at the bony prominence of the *sacrum* for both backrest angles (Figure 5.4).

Table 5.1: Average peak pressure during arm cranking and rest, and the average of the peaks and valleys during arm cranking.

	Backrest angle	
	60°	30°
Average peak pressure during arm cranking (mmHg)	12.02	10.30
Average peak pressure during rest (mmHg)	10.53	8.76
Average of peaks during arm cranking (mmHg)	12.17	10.56
Average of valleys during arm cranking (mmHg)	11.86	10.04

The average pressure for all sensors under load was higher for the backrest angle at 60° compared with 30° (4.53 *vs.* 4.26 mmHg, +6%). Contact area, however, did not change whereas the oxygen consumption was 0.07 l/min higher (+9%) during the arm cranking at 60° compared with 30° (table 5.2). Gross mechanical efficiency was 7.9% lower at 30° backrest angle.

Table 5.2: Average pressure for all sensors under load and contact area during the resting trials, and oxygen consumption (VO2) and gross mechanical efficiency (ME) for the subject during the arm cranking trials.

	Backrest angle	
	60°	30°
Average pressure (mmHg)	4.53	4.26
Contact area (m^2)	0.019	0.019
VO2 (l/min)	0.82	0.75
ME (%)	14.0	15.2

It was also noticeable that average peak pressure was higher during the arm cranking trials than during rest. Going from resting to arm cranking, the average peak pressure increased by 12.4% when the backrest was at 60° and 15% at a 30° backrest angle. During the arm cranking trials, a *sine* wave pattern was visible for the peak pressure (Figure 5.3) and the peak pressure varied by an average of 3% from the *minimum* value to the *maximum* value during each cycle for the 60° backrest angle and 5% for the 30° backrest angle.

Discussion

This case study is the first to measure the seat interface pressure during arm cranking and compare different backrest angles. It was shown that changing the backrest angle of a handbike seat changes the interface pressure as the average pressure for all loaded sensor was higher during the 60° backrest angle compared with 30°. A more upright posture will shift part of the weight from the backrest to the seat and since the contact area was found to stay the same, the average pressure will increase.

On the pressure maps it can be seen that the peak pressure is located at the *sacrum* for both backrest positions. Studies looking at the interface pressure on wheelchair seats find that the peak pressure location varies between the *sacrum* and the ischial tuberosities from subject to subject depending on body posture, anatomy, and positioning (Dolbow et al., 2013; Maurer & Sprigle, 2004; Park & Jang, 2011; Pellow, 1999; Sprigle & Sonenblum, 2011). A handbike seat differs from a wheelchair seat by the curved transition from seat to backrest (Figure 5.2). This transition may have created a kyphotic posture typified by a posterior pelvic tilt for the subject. More research is required to determine if the *sacrum* is always the location of peak pressure values on a handbike seat however.

The average peak pressure for the subject during arm cranking was lower than reported in the literature on wheelchair studies (Dolbow et al., 2013; Park & Yang, 2011; Pellow, 1999). This may be explained by the large variance in interface pressure from subject to subject. Giesbrecht et al. (2011) had standard deviations up to 57% of the average pressure at the *sacrum* in subjects with *tetraplegia* and the subject in this study had a low weight (45 kg). The absolute pressure value also depends on anatomy and posture, which is different from wheelchair seat to handbike seat as mentioned earlier.

Tissue damage has previously been reported to occur after 7.5 mmHg loading, which is lower than the peak pressure values found in this study (Kernozek & Lewin, 1998). It is difficult, however, to determine a specific threshold at which loads are classified as harmful or safe between different people and body sites. The tissue´s tolerance to load varies according to the condition of the tissue, location, age, hydration, anatomy, and loading duration and magnitude (Sprigle & Sonenblum, 2011). A handcyclist can be exposed to low hydration, increased sweating, and long periods of time without being able to perform pressure relieving maneuvers during exercise. Handcyclists´ with SCI, who suffer from PU developments, should accordingly be attentive to the possibility of handcycling worsening the PU and take the proper precaution. If PUs develop, all load relieving or pressure redistribution interventions could be helpful. Since the lowest backrest inclination shows the lowest peak pressure in this study, the subject would therefore be advised to use a backrest angle of 30° to reduce the risk of further worsening a potential PU.

Most wheelchair studies have measured the static seating pressure, which can provide a baseline, but it is likely that the seat interface is loaded differently throughout the day with different activities of daily living or activities. The data in this study show an increase of average peak pressure from rest to arm cranking. A *sine* wave pattern is also noticeable in the peak pressure during arm cranking (Figure 5.3), likely due to pushing and pulling the handles. Kernozek & Lewin (1998) theorized that fluctuations in peak pressure during wheelchair locomotion could facilitate a “pumping mechanism” to stimulate blood and lymphatic activity and potentially be beneficial. Since the valleys of the fluctuations are always higher than the corresponding peaks during rest in this study, it is not likely to be the case during handcycling.

In addition to the average pressure and average peak pressure, the oxygen consumption was also lower at 30° backrest angle compared to 60° backrest angle. Consequently, lowering the backrest inclination to potentially reduce the risk of PU development did not increase the oxygen uptake, but rather resulted in a higher mechanical efficiency. This could, however, be the result of the subject being accustomed to a more supine posture during handcycling and should be investigated further.

A potential application for the future could be to use a pressure sensor mat, such as the one used in this study, to continuously monitor the interface pressure during handcycling. If certain pressure thresholds for specific periods of time are exceeded, an alarm could warn the user and suggest a pressure relieve movement. Another option could be to implement a loop control system, in which the sensors are used to continuously measure the interface pressure and posture of the user. When pressure thresholds are exceeded, inflatable air cushions in the seat could be activated to relieve the pressure or change the posture of the user.

Conclusion

A number of general conclusions can be made based on the results in this study. The average pressure appears to increase when going from a supine to a semi-supine position as the contact area stays the same. Peak pressure was located in the region of the *sacrum* for both backrest inclinations and found to increase from rest to arm cranking while exhibiting a *sine* wave pattern during the exercise trials. Oxygen consumption was lowest and mechanical efficiency highest at the backrest angle of 30°, indicating this position to be most beneficial for the subject to improve performance and reduce the risk of developing pressure ulcers.

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Chapter 6

General discussion and implication

Physical activity has been associated with improved physical fitness, health, and psychological well-being for people with lower-body disabilities, and recent technological developments have allowed this population to exercise in a number of different ways. Still, limitations exist that potentially discourage physical activity such as a lack of exercise water activities as traditional aquatic sports (e.g. rowing, kayak and canoe) since a certain level of trunk muscle activation is required to maintain balance. The general problem in *focus* was therefore to develop a human-powered watercraft to be used by people with lower-body disabilities, and to measure the physiological responses when maneuvering the watercraft.

The human-powered watercraft was developed as a catamaran consisting of two hulls to provide buoyancy, and a recumbent handbike seat and footrests were fixed between the two hulls by means of a custom-built *aluminum* frame. An arm crank system in front of the seat was connected *via* a roller chain to the transmission system in order to drive a flexible shaft and the propeller. The metabolic expenditure when arm cranking on the watercraft was shown to be similar to other physical activities performed by people with lower-body disabilities and the mechanical efficiency was found to be comparable to other human-powered watercrafts (Zamparo et al., 2008) and, thus, revealed to be an alternative fitness tool especially for people with lower-body disabilities, who aim at water activities. The development of the watercraft will hopefully lead to more people with lower-body disabilities engaging in physical activity.

The watercraft was further developed in order to improve the user experience and performance. A new custom-built frame made the watercraft lighter and thereby improved the performance while it was made much easier to fit the seating position to the user. Several further developments could be made on the watercraft to improve the user experience such as a new steering system. Besides using the watercraft for leisure-time physical activity, another application could be to use the watercraft for sports in racing. For this application it would be interesting to build a monohull human-powered watercraft for people with lower-body disabilities.

It was also investigated how velocity fluctuations may affect the speed of the watercraft and if different propulsion modes have an influence. No significant differences were found between asynchronous and synchronous arm cranking on either speed or physiological responses. There is still no *consensus* in the literature on which crank mode is the most efficient and it is likely dependent on factors such as subjects, exercise intensity, exercise protocol, and the ergometer used in the study (Abbasi-Bafghi *et al*., 2008; Smith *et al*., 2008; van der Woude *et al*., 2008). We found no effect on speed or physiological responses between the two crank modes, but for future research, mounting inertial measurement units on the watercraft could help further explain the relevance of velocity fluctuations.

Finally, the interface pressure was evaluated in a setting that mimics the one on the watercraft. It is known that high interface pressure for long periods of time increases the risk of pressure ulcer development for people with spinal cord injury. A case-study was made on an Italian handcycling champion arm cranking on an arm ergometer at two difference backrest inclinations, while interface pressure and oxygen uptake were measured. The results showed a difference in pressure between backrest inclinations and from arm cranking and resting. This could help people with SCI to still be able to exercise even if suffering from pressure ulcers. More research in this area is needed, but it could lead to the use of a pressure sensor mat to continuously monitor the interface pressure during arm cranking such as on the watercraft or a handbike. If certain pressure thresholds for specific periods of time are exceeded, an alarm could warn the user and suggest a pressure relieve movement. Another option could be to implement a loop control system, in which the sensors are used to continuously measure the interface pressure and posture of the user. When pressure thresholds are exceeded, inflatable air cushions in the seat could be activated to relieve the pressure or change the posture of the user, such as the system made by Bertolotti *et al.* (2013).

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Doctorate outline 2015-2018

Conference abstracts:

- **Fuglsang, T**., and Ardigò, L. (2017). The effect of backrest inclination on the risk of pressure ulcers during handcycling. *Lavori in Corso,* 17 Feb. 2017, Universitá degli Studi di Milano, Milano, Italy.
- **Fuglsang, T**., Padulo, J., Spoladore, M., Dalla Piazza, M., and Ardigò, L. (2017). Development and Testing of a Novel Human-Powered Watercraft for People with Lower-body Disabilities. *35th International Conference in Biomechanics in Sports,* 14-18 June 2017, German Sport University, Cologne, Germany.

Publications:

- **Fuglsang, T**., Padulo, J., Spoladore, M., Dalla Piazza, M., and Ardigò, L. (2017). Development and Testing of a Novel Arm Cranking-Powered Watercraft. *Front. Physiol.* 8(635).
- Padulo, J., Iuliano, E., Brisola, G., Iacono, A., Zagatto, A., Ardigò, L., Lupo, C., **Fuglsang, T.,** and Cular, D. (2018). Validity and reliability of a standalone low-end 50-Hz GNSS receiver during running. *Biol Sport.* Accepted on 10/2018.

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- **Fuglsang, T**., Padulo, J., and Ardigò, L. (2018). A Comparison of Synchronous and Asynchronous Arm Cranking on the Handwaterbike. *Submitted for publication in Journal of Sports Sciences.*
- **Fuglsang, T**., Rottenbacher, C., Cristiani, A, Gualea, M., and Ardigò, L. (2018). The Influence of Backrest Inclination on Interface Seat Pressure during Arm Cranking – a Case Study. *Submitted for publication in Sports Engineering.*