The Ergonomics of Wheelchair Configuration for Optimal Sport Performance

Barry Mason

THE ERGONOMICS OF WHEELCHAIR CONFIGURATION FOR OPTIMAL SPORT PERFORMANCE

Barry Mason

A Doctoral Thesis

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Abstract

Within court sport wheelchair design there are numerous components that can be configured in a variety of ways, yet the effects of different configurations on the ergonomics of sport wheelchair performance remains to be seen. The present thesis investigates the effects of manipulating key areas of the wheelchair-user combination on the mobility performance of highly trained athletes and explores whether optimal configurations can be established.

The first study adopted a qualitative approach to investigate the influence of wheelchair configuration on mobility performance. Disparity amongst participants' perceptions highlighted rear-wheel camber as the primary area of configuration in need of evidence based research. Wheel size, seat position and coupling at the wheelchair-user interface were also perceived to affect mobility, yet methods for identifying optimal settings were limited. The need for customised equipment was established in Chapter 4 when the effect of glove type was investigated. Gloves modified to meet the specific requirements of the user improved maximal linear sprinting and manoeuvrability performance compared to other generic gloves, due to a proposed improvement in coupling between the user and the hand-rims.

During sub-maximal propulsion, external power output (P_O) and physiological demand were both affected by camber (Chapter 5). External P_O, oxygen uptake ($\dot{V}O_2$) and heart rate (HR) were all elevated in 24° camber compared with 15° and 18°, although gross mechanical efficiency (ME) improved with greater camber (20° and 24°). Chapter 6 extended this work during maximal effort, over-ground propulsion. The 18° setting significantly improved linear sprinting performance compared to 24° and manoeuvrability performance to 15° camber.

Different wheel sizes with fixed gear ratios also affected P_O and physiological demand during sub-maximal wheelchair propulsion (Chapter 7). Increased P_O, $\dot{V}O_2$ and HR responses were identified for 24" wheels compared with 26" wheels, yet improved ME. Hand-rim kinetic data established that in order to maintain a constant velocity, smaller wheels required greater force to be applied. Chapter 8 extended this work into a sports environment. Favourable reductions in 20 m sprint times were revealed for 26" wheels compared with 24" wheels, without negatively affecting initial acceleration or manoeuvrability. Optimal settings based on disability were not identified, although the 26" setting enabled favourable responses to sub-maximal and maximal effort propulsion for the highly trained athletes investigated.

This thesis revealed that manipulating areas of the wheelchair-user combination alters the ergonomics of sub-maximal and maximal effort sports wheelchair performance in highly trained athletes. Larger camber (24°) and smaller wheels (24") appeared unfavourable due the greater P_O required, increased physiological demand and decreased maximal linear sprinting performance. Hand-rim kinetic data obtained from the SMART^{Wheel} (Chapter 7) provided a valuable insight into injury risk in different wheel sizes and the inclusion of such data is encouraged in future configuration studies.

Key Words: Wheelchair propulsion, wheelchair athletes, physiology, biomechanics, kinematics, kinetics, sports performance.

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Dedication

This thesis is dedicated to my family, in particular to my Mum and Dad for their continued support and belief in me, even when the inevitable “lows” were reached and I am forever grateful for this.

Preface

The research presented throughout the current thesis has been peer reviewed through the following publications and communications:

Publications

Chapter 3:

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Chapter 4:

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Chapter 5:

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Under Review

Chapter 7:

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Published Abstracts

Chapter 3:

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Chapter 4 (pilot study):

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Chapter 4:

Mason BS, Woude van der LHV, Goosey-Tolfrey VL (2010). The influence of glove type on mobility performance for wheelchair rugby players. In L.H.V van der Woude, F. Hoekstra, S. de Groot, K.E. Bijker, R. Dekker, P.C.T. van Aanholt, F.J. Hettinga, T.W.J. Janssen & J.H.P. Houdijk (Eds.), *Rehabilitation: Mobility, Exercise and Sports. 4th International State-of-the-Art Congress* (pp. 349-351). Amsterdam: IOS Press.

Conference Communications

Chapter 3:

A qualitative examination of wheelchair configuration for optimal sports performance (*Poster*). 4th International State-of-the-Art Congress 2009 – Amsterdam.

Chapter 4 (pilot study):

The effect of glove type on wheelchair rugby performance (*Poster*). British Paralympic Association: Sports Science, Medicine & Coaching Conference 2008 – Loughborough University. Awarded the 'best poster presentation'.

Chapter 4:

The influence of glove type on mobility performance for wheelchair rugby players (*Oral*). 4th International State-of-the-Art Congress 2009 – Amsterdam.

Invited Symposium

Chapters 3-6:

Woude van der LHV, Mason BS, Verellen J – Ergonomics in rehabilitation: wheeled mobility and sports (*Oral*). International Society of Adapted Physical Activity (ISAPA) Annual Conference 2009 – Gavle, Sweden.

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List of Abbreviations

Throughout the current thesis, the following abbreviations and units of measurement were used to signify frequently used terms. All abbreviations were defined the first instance they appear in the text:

°	Degrees
AB	Able-bodied
ANOVA	Analysis of variance
beats·min^{-1}	Beats per minute
CT	Cycle time (s)
EA	End angle (°)
EMG	Electromyography
Fdrag	Drag force (N)
FEF	Fraction of effective force (%)
Fres	Resultant force (N)
Ftan	Tangential force (N)
HCon	Hand contact
HP	High point players
HR	Heart rate (beats·min^{-1})
HRel	Hand release
IWBF	International Wheelchair Basketball Federation
IWRF	International Wheelchair Rugby Federation
L·min^{-1}	Litres per minute
LP	Low point players
ME	Gross mechanical efficiency (%)
N	Newtons
PA	Push angle (°)
P_O	Power output (W)
PT	Push time (s)
RPE	Rating of perceived exertion
RT	Recovery time (s)
RoM	Range of motion (°)
s	Seconds
SA	Start angle (°)
SCI	Spinal cord injury
TDC	Top dead centre
$\dot{V}O_2$	Oxygen uptake (L·min^{-1})
W	Watts
WCB	Wheelchair basketball
WCR	Wheelchair rugby
WCT	Wheelchair tennis
WERG	Wheelchair ergometer

Chapter One

Introduction

Ergonomics describes the scientific study of the relationship between man and his working environment and has been predominantly used in an industrial context. The essence of ergonomics requires an interdisciplinary approach to optimise the interaction between a device and its user so that the efficiency, safety/health and comfort of the resulting task are maximised (Murrell, 1965). It has been recognised more recently that the principles used initially within industrial ergonomics overlap many of the analytical techniques used in the sport sciences (Reilly, 1981; Reilly and Ussher, 1988). In a sporting environment the interfacing between the user and their sports equipment is an important ergonomic consideration, whereby the efficiency, safety/health and comfort of the interaction are again vital (Reilly and Lees, 1984). The additional feature relating to the application of ergonomics in sport, particularly at the elite level, is that performance, by means of increasing the task output is a highly important consideration, even if this occurs at the expense of energy expenditure (Frederick, 1984; Reilly and Ussher, 1988). The demand for continual advancements in athletic performance means that manufacturers of sports equipment are obliged to identify ways in which performance can be augmented. However, manufacturers are also obliged to cater for the efficiency, safety/health and comfort of the user, which can often be overlooked in the endeavour to continually enhance performance (Reilly and Lees, 1984).

1.1 Ergonomics of sports wheelchair performance

The ergonomics of sports wheelchair performance, specific to the wheelchair court sports (basketball [WCB], rugby [WCR] and tennis [WCT]) is dependent upon numerous factors (Figure 1.1). Physical characteristics including age, weight, height, impairment, aerobic fitness, skill and strength of the user can all have a strong bearing on performance in these sports. Factors relating to the design and configurations of the wheelchair are also of importance, especially given the noticeable developments in design and the ways in which sports wheelchairs have been configured over recent years (Ardigo et al., 2005). Subsequently, the interaction between the user and the wheelchair, referred to as the wheelchair-user combination, has a vital impact on the ergonomics of wheelchair performance (Woude et al., 1986; Vanlandewijck et al., 2001).

The major development to the design of the court sport wheelchair has seen a reduction in mass, mainly owing to advancements in technology and materials (LaMere and Labanowich, 1984; Yilla, 2004; Ardigo et al., 2005; Burkett, 2010). In addition to this there are still numerous areas of a court sports wheelchair that can be configured in various ways. For example, the angle at which the main wheels are cambered has increased progressively over the years (Polic, 2000; Ardigo et al., 2005). The number of castor wheels has alternated between a combination of one and two-wheeled designs at both the front and back of the wheelchair (Cooper, 1998; Polic, 2000; Bunting, 2001). The vertical and horizontal positioning of the seat can also vary dramatically between individuals, as can the size of the main wheels. Despite all of these developments and options that are now available to athletes, evidence based research, specific to the wheelchair court sports, is extremely lacking. Consequently, athletes are forced to make decisions about how to configure these areas of their wheelchair based largely on trial and error. Although, subjective opinions must be taken into consideration, a more detailed, scientific understanding of how certain modifications affect the ergonomics of sports wheelchair performance would facilitate athletes' decisions. The conceptual model (Figure 1.1) reiterates the necessity for an interdisciplinary approach by highlighting the physiological and biomechanical factors associated with the wheelchair-user combination that ultimately influence the ergonomics of sports wheelchair performance. A combination of all these factors can have a significant bearing on the efficiency, safety/health, comfort and performance of sports wheelchair propulsion (Woude et al., 1989a).

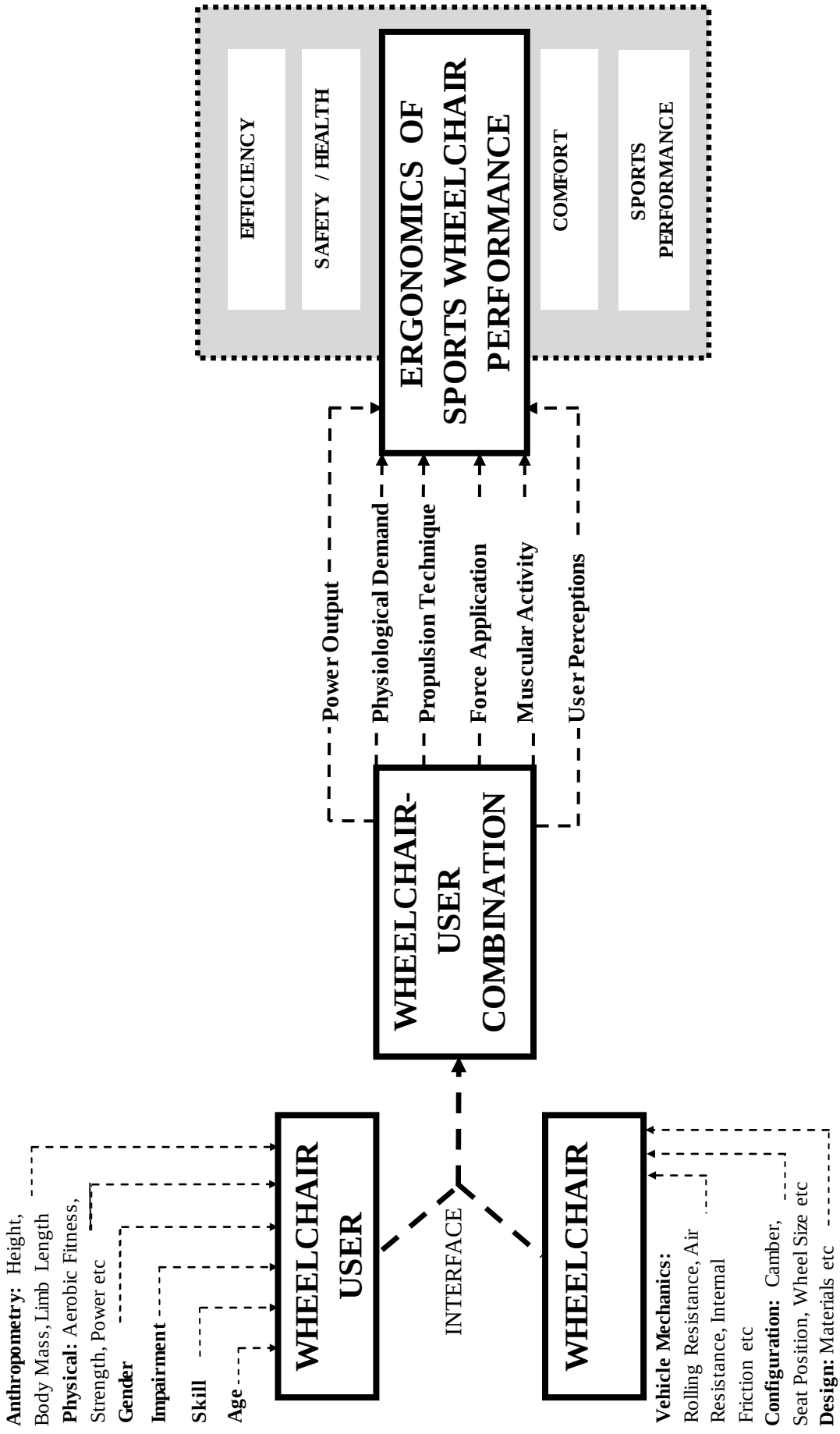

Figure 1.1 – Conceptual model illustrating the factors that influence the ergonomics of sports wheelchair performance. Adapted from Woude et al. (1986).

1.2 Aims and objectives of the thesis

The principal aim of the thesis was to examine how specific areas of wheelchair configuration may optimise the mobility performance of elite wheelchair athletes. In order to achieve this, three main objectives were formulated:

i) To investigate the self-selected configurations of elite wheelchair athletes to subjectively evaluate how modifications to configuration are perceived to affect mobility performance and to establish key areas of sports wheelchair configuration worthy of empirical research.

ii) To analyse the effects of manipulating several key areas of sports wheelchair configuration on the physiological and biomechanical responses of elite wheelchair athletes during sub-maximal wheelchair propulsion.

iii) To analyse the effects of manipulating these key areas of wheelchair configuration on the performance of maximal effort, sport-specific manoeuvres.

1.3 Organisation of the thesis and experimental aims

The current thesis was structured into a total of six experimental chapters (Chapters 3 - 8), with the research questions of each addressed below. Preceding this, a comprehensive literature review (Chapter 2) was conducted. This review focused on the value of an ergonomic approach when modifying sports equipment and critiqued previous research that has adopted this approach when investigating wheelchair configuration.

The first experimental chapter (Chapter 3) evaluated the opinions of experienced, elite wheelchair athletes and detailed their perceptions on areas of wheelchair configuration in relation to mobility performance. It was anticipated that this group of participants would have a strong understanding of the topic area and would demonstrate similar responses to how areas of configuration influence performance. The main aim was to establish the areas of wheelchair configuration in greatest need of future quantitative research, by exploring any gaps or differences between athletes' perceptions in relation to their effects on mobility performance. Key areas identified formed the basis of the following experimental chapters.

Chapter 4 investigated the interaction taking place directly at the wheelchair-user interface in the form of glove type and its relationship with hand-rim configuration. The objective of this study was to determine whether certain glove types and hand-rim configurations used within WCR interacted more favourably than others when performing sport specific field tests.

Chapters 5 and 6 were the first wheelchair manipulation studies whereby rear-wheel camber, which was established as a key area of configuration in Chapter 3, was investigated. The first of these two studies examined the effects of sport-specific degrees of camber upon the physiological and biomechanical responses during sub-maximal wheelchair propulsion. Chapter 6 was an extension of this study, conducted in a field environment to examine the influence of camber upon aspects of maximal effort mobility performance. The aim of these two studies was to investigate the physiological, biomechanical and performance effects of camber adjustments and to determine whether optimal settings could be identified.

Two similar studies then followed, whereby the effects of different wheel sizes with fixed gear ratios on the physiological and biomechanical responses during sub-maximal wheelchair propulsion (Chapter 7) and the performance of maximal effort field tests (Chapter 8) were investigated. The addition to Chapter 7 was the incorporation of a force sensing hand-rim (SMART^{Wheel}) to determine the application of force during wheelchair propulsion with various wheel sizes. As with the previous two chapters, the objective was to examine the physiological, biomechanical and performance responses to wheel size adjustments and to investigate whether optimal settings existed.

Chapter Two

Literature Review

As highlighted in Figure 1.1, numerous factors affect the ergonomics of sports wheelchair performance. The majority of research into this topic has investigated the physiological and biomechanical capabilities and limitations associated with the user. Alternatively, the role of the wheelchair has rarely been considered and when it has, the effects have predominantly been investigated during daily life wheeling tasks (section 2.6). Subsequently, there has been a limited application of an ergonomics approach towards wheelchair configurations reflective of the wheelchair court sports. The previous approaches that have been adopted to address the key ergonomic principles of sports wheelchair propulsion (efficiency, safety/health, comfort and performance) have been discussed below.

2.1 Efficiency in sports wheelchair propulsion

Being able to reliably establish the efficiency of a task often requires the workload to be quantified (Woude et al., 2001). Workload is determined during wheelchair propulsion by the external power output requirements (P_O) of the wheelchair-user combination, which as demonstrated in Figure 1.1 can have a significant bearing on the ergonomics of wheelchair performance (Woude et al., 1988a). As depicted below, P_O can be estimated for a given velocity (v) by calculating the drag force (*Fdrag*) of the wheelchair-user combination, which is the sum of the rolling resistance, air resistance, internal friction and gravitational effects experienced (Woude et al., 1986). Various components of a wheelchair's design and configuration, coupled with certain physical characteristics of the user and the surface used for propulsion can affect the *Fdrag* experienced (Woude et al., 2001).

$$P_O\ (W) = Fdrag \cdot v \qquad \text{(Woude et al., 1986)}$$

Traditionally, physiological analyses have been conducted to assess the efficiency of a task, with a reduction in oxygen uptake ($\dot{V}O_2$), energy expenditure, heart rate (HR) and blood lactate responses all favourable outcomes (Reilly, 1981). Determining the P_O also enables

mechanical efficiency to be calculated. Although numerous indices for the calculation of mechanical efficiency exist including net, work and delta efficiency, gross mechanical efficiency (ME) is the most frequently reported in the wheelchair propulsion and configuration literature (section 2.6). Although ME can underestimate the true efficiency of a task, it is appropriate for within-subject designs as it far simpler to calculate than other indices, where baseline subtractions are required. During sub-maximal, steady-state cyclic exercise such as hand-rim wheelchair propulsion, ME can be defined as the ratio of external work in relation to the total internal energy expended to perform the work (E_n):

$$ME\ (\%) = (P_O / E_n) \cdot 100 \qquad \text{(Woude et al., 1986)}$$

Caution must be exercised when utilising ME to assess the efficiency of wheelchair propulsion, as improved ME often occurs as a result of a greater P_O, which has a curvilinear relationship with $\dot{V}O_2$ (Woude et al., 1988a). Therefore, an improvement in ME may be largely due to increased external P_O, as opposed to a reduction in $\dot{V}O_2$. As a result, pushing economy has also been reported to establish the physiological demand of wheelchair propulsion (Lakomy and Williams, 1996; Goosey and Campbell, 1998; Goosey et al., 2000; Cooper et al., 2003). Economy is calculated independent of the work done, whereby the sub-maximal $\dot{V}O_2$ at a given speed can be used to define ‘efficient’ performance (Cavanagh and Kram, 1985).

Regardless of the methods used, improving propulsion efficiency is of great importance due to the emphasis placed on aerobic conditioning for athletes competing in the wheelchair court sports (Goosey-Tolfrey, 2005; Roy et al., 2006; Bernardi et al., 2010). This would appear to be of further importance based on the inefficient nature of hand-rim wheelchair propulsion as a form of ambulation, with ME values rarely exceeding 11.5%, even in highly trained wheelchair sportsmen (Woude et al., 1986; Woude et al., 1988a; Woude et al., 1988b; Vanlandewijck et al., 1994a). These values are considered low when compared to other cyclic modes of ambulation such as cycling, with ME commonly shown to range between 20 - 23.5% (Sidossis et al., 1992; Horrowitz et al., 1994; Coyle, 2005; Hopker et al., 2010). The low ME values signify that a high loss of energy occurs, partially due to a heavy reliance on the small muscle mass of the arms and difficulties associated with coupling during hand-rim wheelchair propulsion (Woude et al., 1989a; Woude et al., 2001). Combined with being an inefficient form of ambulation, the problem is exacerbated by the fact that little is

known about methods of propulsion technique and wheelchair configurations that can improve efficiency (Cooper et al., 2003).

Physiological measures are often a sufficient means for assessing the effectiveness of a task or wheelchair configuration during steady-state propulsion. However, these parameters do not clarify why a certain task or configuration may be more efficient than another. Due to the necessity for a multidisciplinary approach in ergonomic investigations, biomechanical analyses of wheelchair propulsion in association with physiological analyses are required. Investigating a combination of propulsion technique, temporal parameters, force application and muscle activation will help to underpin the biomechanics of efficient hand-rim wheelchair propulsion (Woude et al., 1989a).

Kinematic investigations have become extremely prevalent in ergonomic research into hand-rim wheelchair propulsion through the inclusion of high-speed video analysis. This has enabled a series of temporal and angular displacement parameters to be ascertained during the propulsion cycle (Figure 2.1). The propulsion cycle has typically been divided into two distinct phases, the push phase and the recovery phase. The push phase refers to the phase of the cycle when the hands are in contact with the hand-rim i.e. from initial hand contact (HCon) to hand release (HRel). The remainder of the propulsion cycle (HRel to HCon of the following push) is referred to as the recovery phase. Temporal parameters such as push times (PT), recovery times (RT) and cycle times (CT) have been used to describe the absolute and relative durations of the push phase, recovery phase and propulsion cycle respectively. In association with this, push angles (PA), which describe the angle over which a push is produced, are often documented. As demonstrated in Figure 2.1, PA can be further subdivided into a start (SA) and end angle (EA), whereby SA refers to the angle created between HCon and top dead centre (TDC) of the hand-rim, with EA defined as the angle between TDC and HRel (Figure 2.1). Investigations that have combined physiological measures with kinematic investigations have identified a number of associations between temporal and angular displacement parameters with efficiency during hand-rim wheelchair propulsion. Push frequency in particular, which is a component of CT, has commonly been linked with efficiency. It has been regularly reported in the wheelchair racing literature that reductions in push frequency lead to improvements in pushing economy (Jones et al., 1992; Goosey and Campbell, 1998; Goosey et al., 2000; Goosey-Tolfrey and Kirk, 2003). Reductions in push frequency have also been observed in conjunction with improved ME during practice periods of wheelchair propulsion with able-bodied (AB) participants (Groot et al., 2002; Groot et al.,

2008). Increases in PT and PA have also been associated with improved ME (Groot et al., 2002; Groot et al., 2008), which are suggested to be favourable due to the greater opportunity that these parameters allow for force to be transmitted to the wheel (Chow et al., 2001).

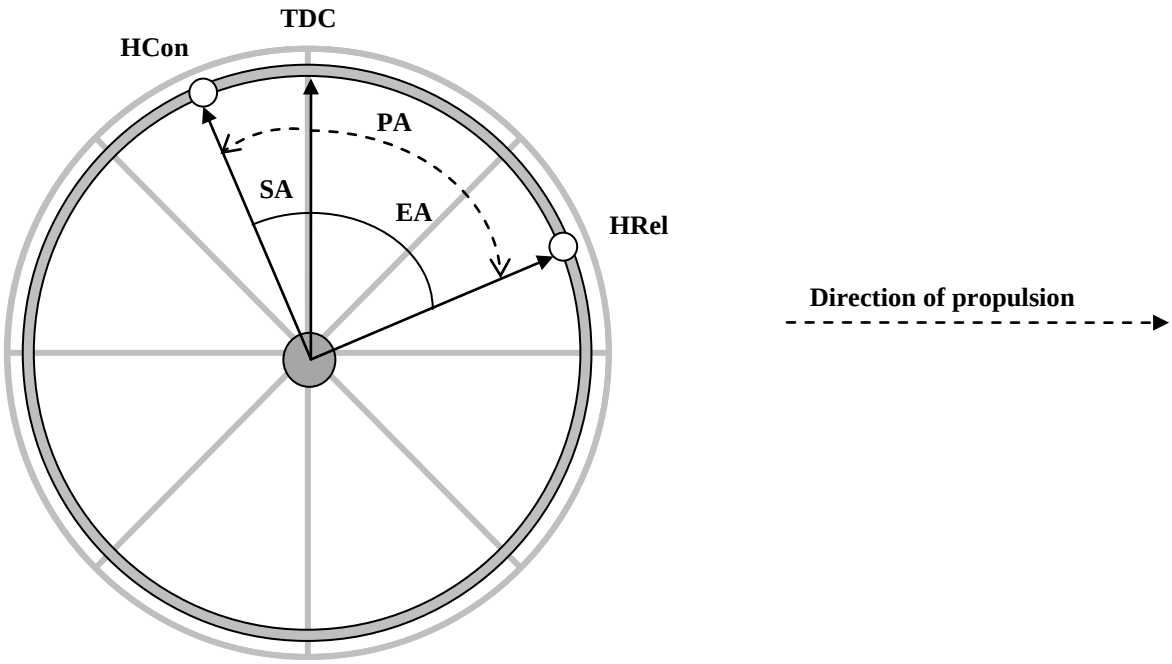

Figure 2.1 – Temporal and angular displacement parameters displayed during the wheelchair propulsion cycle. Adapted from Spaepen et al. (1996) and Chow et al. (2000).

The upper body joint kinematics have also been frequently investigated during hand-rim wheelchair propulsion, and when done so in conjunction with physiological measures, inferences towards efficiency have been presented. During constant velocity wheelchair racing propulsion, O'Connor et al. (1998) revealed that increased wrist velocity at HCon and elbow velocity during the push phase were positively correlated with $\dot{V}O_2$. However, since hand-rim wheelchair propulsion is a guided movement during the push phase, the movement trajectories of the upper body are somewhat limited (Sanderson and Sommer, 1985). Larger variations in these movements are permitted during the recovery phase and therefore actions during this phase may have more of a bearing upon the efficiency of propulsion (Vanlandewijck et al., 1994a; Shimada et al., 1998). Subsequently, the style of recovery has been investigated, with four styles commonly referenced (Figure 2.2). Veeger et al. (1989a) identified that a circular technique elicited a significantly higher ME in wheelchair sportsmen compared to a pumping technique. Furthermore, Groot et al. (2008) identified a higher ME in

able-bodied (AB) participants who adopted a double-looping technique compared to those with a single-looping technique. Despite the observed changes in upper body joint kinematics that have been associated with a reduced physiological demand, the exact mechanisms responsible for improvements in efficiency remains unclear. In order to reliably ascertain these mechanisms, an integrated approach is required whereby the kinematics of hand-rim wheelchair propulsion is investigated alongside kinetic and physiological measures (Woude et al., 1989a).

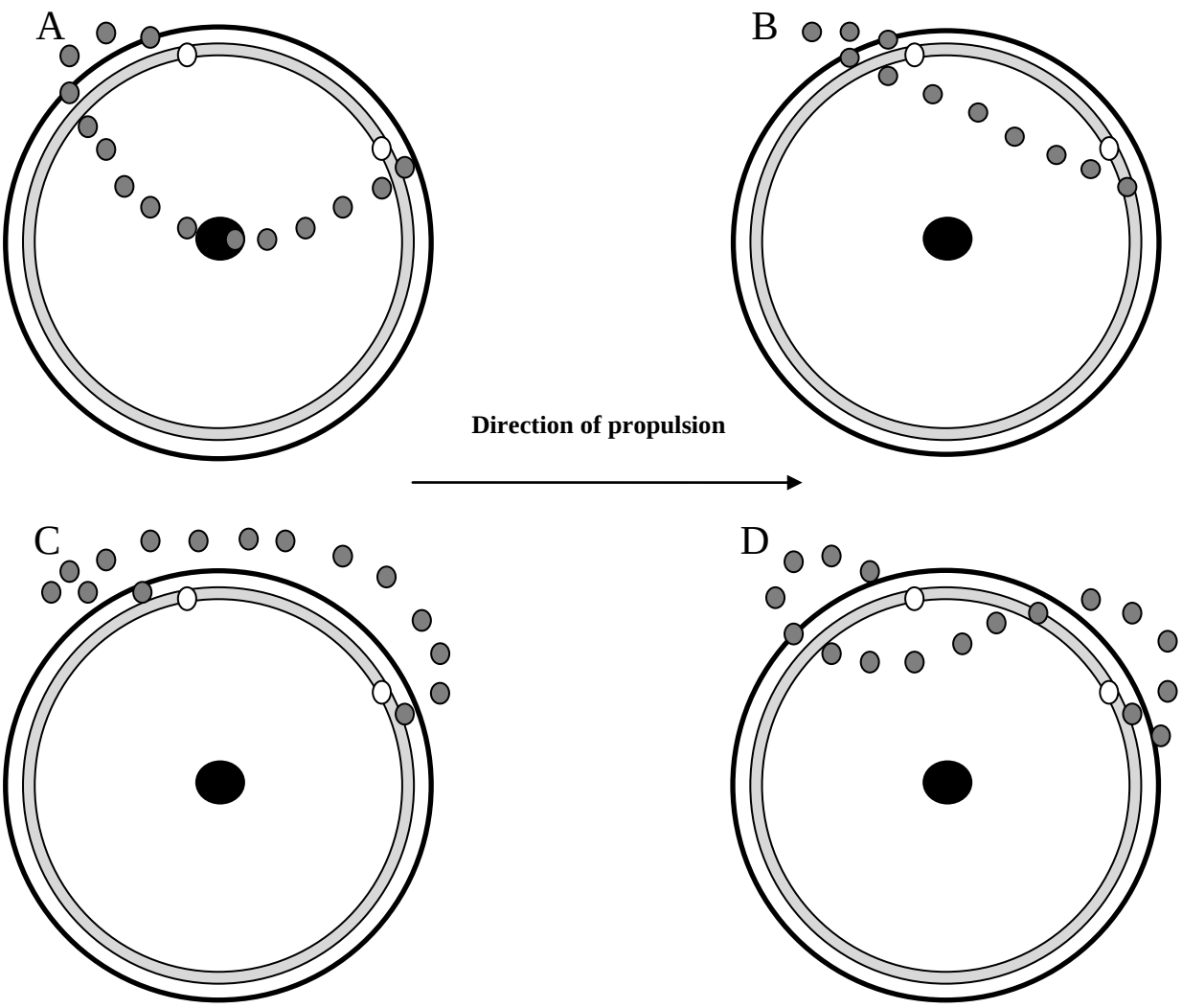

Figure 2.2 – Trajectory of the hand during the recovery phase used to distinguish between four styles of daily hand-rim propulsion: (a) circular; (b) pumping; (c) single-looping; (d) double-looping. Adapted from Groot et al. (2008).

2.2 Safety/health in sports wheelchair propulsion

User safety/health is a major concern to manual wheelchair users, who, as a population, are thought to be at an extremely high risk of injury to the upper extremities (Gellman et al., 1988; Curtis and Black, 1999; Curtis et al., 1999; Finley and Rodgers, 2004). The wrist in particular has a high incidence of injury, with 49 - 73% of wheelchair users shown to demonstrate signs and symptoms of carpal tunnel syndrome (Aljure et al., 1985; Gellman et al., 1988; Tun and Upton, 1988; Davidoff et al., 1991). In addition to this, between 31 – 75% of wheelchair users have reported some form of shoulder pain as a result of wheelchair propulsion (Bayley et al., 1987; Wylie and Chakera, 1988; Pentland and Twomey, 1991; Curtis and Black, 1999; Curtis et al., 1999; Finley and Rodgers, 2004). Although participation in sports has not been shown to exacerbate the risk of injury (Taylor and Williams, 1995; Fullerton et al., 2003; Finley and Rodgers, 2004; Ustunkaya et al., 2007) further research is required in this area. Despite a reasonable amount of literature focusing on the causes of the high injury prevalence, little has been afforded to prevention strategies for which wheelchair configuration could be crucial (Kulig et al., 2001; Mercer et al., 2006).

Over the last two decades tremendous technological advancements have taken place in the evaluation of wheelchair propulsion, particularly in force sensing hand-rim equipment, which has enabled kinetic investigations during wheelchair propulsion. A computer-controlled wheelchair simulator was developed in the Netherlands, as described by Niesing et al. (1990), which allowed the forces and moments applied by the user to be determined. More recently instrumented wheels have been developed, including the SMART^{Wheel}, as described by Asato et al. (1993) and used extensively by Cooper and his research group in Pittsburgh. Various other instrumented wheels have since been manufactured and used by Richter and colleagues in Nashville (Richter et al., 2007), Drongelen and colleagues in Amsterdam (Drongelen et al., 2005) and Wu and colleagues in Taiwan (Wu et al., 1998). Similarly to the computer-controlled wheelchair ergometer, the instrumented wheels can calculate the three-dimensional forces and moments that are applied to the hand-rims. The advantage of these wheels is that they can be attached to various wheelchairs with different configurations and can be utilised during over-ground propulsion.

Investigations into hand-rim kinetics have enabled injury risk to be examined in further detail. As previously mentioned, carpal tunnel syndrome is a serious risk factor for wheelchair users (Aljure et al., 1985; Gellman et al., 1988; Tun and Upton, 1988; Davidoff et

al., 1991). Boninger et al. (2002) investigated median nerve dysfunction, which predisposes to carpal tunnel syndrome, and revealed that greater magnitudes of peak resultant forces applied to the hand-rims and a higher rate of force development were positively correlated with increased dysfunction. The inclusion of anthropometric and kinematic measures to hand-rim kinetic investigations enables the moments and forces exerted around the joints of the upper extremity to be calculated through inverse dynamics, which can add to the depth of information derived relating to injury risk. It has been established that greater magnitudes of force applied to the hand-rims have coincided with greater joint moments and forces (Robertson et al., 1996). In support of this, greater joint forces and moments have also been associated with increased shoulder pathology (Mercer et al., 2006).

Investigating kinematic variables during wheelchair propulsion can also be of value when establishing the risk of injury to a performer. A larger joint range of motion (RoM) and more extreme maximum and minimum joint displacements have been proposed as potential risk factors for wheelchair users (Shimada et al., 1998; Veeger et al., 1998). Although Boninger et al. (2004) revealed that a greater flexion/extension RoM at the wrist improved median nerve amplitudes, implying a reduced risk of injury at this joint. Other assumptions have also been made relating to injury risk from kinematic investigations into hand-rim wheelchair propulsion. The circular propulsion technique, which had previously been stated to be a more efficient style of propulsion (Veeger et al., 1989a), may also predispose towards a reduced risk of injury. This was suggested based on a reduction in joint accelerations demonstrated at the elbow and shoulder compared to the single and double-looping techniques (Shimada et al., 1998). Temporal parameters have also been stated to have a positive bearing upon user safety/health. Due to the repetitive nature of wheelchair propulsion, overuse injuries are likely to occur and as a result, a reduction in push frequency may also promote a reduced risk of injury (Boninger et al., 1999; Boninger et al., 2004). Unfortunately, these statements are predominantly assumptions, which reiterate that relatively little can be derived about the ergonomics of wheelchair performance through kinematic investigations alone and further emphasises the need for multidisciplinary approaches.

2.3 Comfort in sports wheelchair propulsion

Comfort is another key ergonomic consideration, which is difficult to assess objectively. Subsequently, the most appropriate means for assessing comfort can be through

the use of subjective measures in order to obtain user's perceptions. Nevertheless, this approach has been scarcely adopted by previous ergonomic investigations into wheelchair propulsion. User comfort and preferences have been considered in response to various wheelchair designs and configurations through the incorporation of questionnaires (Gaines and La, 1986; Woude et al., 2003; Richter and Axelson, 2005; Koontz et al., 2006; Perdios et al., 2007; Dieruf et al., 2008). However, all of these studies were conducted in response to daily life conditions, indicating that user comfort of wheelchair athletes has never been considered in sports wheelchair configurations. This may seem surprising, given that researchers and manufacturers may be expected to experience difficulties in empathising with the needs of a wheelchair user. Therefore the input of the athlete could be extremely valuable when attempting to optimise the ergonomics of wheelchair configuration based on the substantial amount of time these users spend in their wheelchairs.

2.4 Performance in sports wheelchair propulsion

When evaluating the ergonomics of equipment used within a sporting context, a final element that is vital for manufacturers to consider is the performance of the task (Fredericks, 1984; Reilly and Ussher, 1988). Batteries of field tests have been developed to assess sport-specific performance in the wheelchair court sports (Brasile and Hedrick, 1996; Yilla and Sherrill, 1998; Vanlandewijck et al., 1999a; Lutgendorf et al., 2009). However, with the exception of Lutgendorf et al. (2009), the majority of these batteries have focused on the performance of tasks involving the ball, such as shooting, passing etc. Although these are important aspects of wheelchair court sport performance, the current thesis was primarily concerned with the evaluation of mobility performance.

Other than aerobic fitness, it has been suggested that the ability to accelerate from a standstill, turn, brake and sprint are all highly important indicators of successful mobility performance for the court sports (Vanlandewijck et al., 2001). The batteries of field tests developed by Brasile and Hedrick (1996), Yilla and Sherrill (1998) and Vanlandewijck et al. (1999a) have all incorporated 20 m sprints. However, only the times taken to perform the sprints were assessed. This may provide a reasonable insight into linear sprinting performance, but fails to objectively account for initial acceleration performance. Also, manoeuvrability performance has only been considered in a 'figure of eight' drill (Vanlandewijck et al., 1999a). In addition to the limited assessment of sport-specific mobility

in previous field tests, the aforementioned batteries have predominantly been utilised to compare performance between different classification levels (Brasile and Hedrick, 1996) or to assess the validity and reliability of such tests (Yilla and Sherrill, 1998; Vanlandewijck et al., 1999a), as opposed to any intervention studies. Field tests have only been utilised on limited occasions for the assessment of modifications to the wheelchair-user combination (Faupin et al., 2002; Goosey-Tolfrey and Moss, 2005; Lutgendorf et al., 2009). Faupin et al. (2002) used a 15 m sprint from a standstill and a 'figure of eight' drill to assess the effects of varying rear-wheel camber on both linear and manoeuvrability performance. Alternatively, Goosey-Tolfrey and Moss (2005) used a 20 m linear sprint to investigate the effects of pushing with and without a tennis racket. The interesting inclusion in the latter study was the use of a velocometer. The velocometer, as used by Moss et al. (2003) and Goosey-Tolfrey and Moss (2005) attaches to individuals' wheelchairs and allows the velocity of each push to be analysed with regards to time. Subsequently, acceleration, deceleration and velocity values can be deduced to improve the depth of information that can be fed back about performance. More recently, Lutgendorf et al. (2009) investigated the influence of glove type on maximal effort performance. The incorporation of the velocometer enabled initial acceleration and braking performance to be quantified, which had never previously been attempted, in addition to linear sprinting and manoeuvrability.

The P_O requirements of the user have already been discussed with regards to steady state wheelchair propulsion and how these can affect the sub-maximal efficiency of wheelchair propulsion. However, P_O is also an important consideration from a sports performance perspective, as the ability to generate a high peak P_O is favourable due to the positive relationship it shares with sprinting performance (Coutts and Stogryn, 1987; Coutts, 1994; Dallmeijer et al., 1994; Woude et al., 1998). If the drag forces experienced by the wheelchair in a certain configuration can be minimised for a given P_O the potential to reach higher velocities exists (Woude et al., 1986). This further demonstrates the difficulties faced when optimising the ergonomics of sports wheelchair performance, as a higher P_O may be favourable for maximal effort performance, yet may impair aspects of efficiency.

2.5 Methodological considerations

Previous studies investigating the ergonomics of wheelchair configuration have employed a number of different methodologies. This has introduced a number of potential

confounding factors, which has often prevented direct comparisons from being made between investigations. The majority of these investigations have also adopted methodological protocols which assess areas of propulsion reflective of daily life activities, as opposed to sporting activities. This has been demonstrated by a combination of the wheelchairs, sub-maximal protocols and the participants used during testing, as discussed throughout section 2.6 and illustrated in Tables 2.1, 2.2, 2.3 and 2.4.

2.5.1 Wheelchair design

A large majority of previous investigations that have manipulated areas of wheelchair configuration have done so in a daily life wheelchair or in a wheelchair racing set-up. However, it is clear that the designs and configurations of these wheelchairs (Figure 2.3) differ dramatically to the configurations adopted in the wheelchair court sports (Figure 2.4). The main difference with the configuration of a court sports wheelchair is the increased camber angle and size of the main wheels. Both daily life and racing wheelchairs tend to have far less camber, with the former often having smaller wheels and the latter demonstrating larger wheels compared to court sport wheelchairs. The other distinct addition to a court sports wheelchair is the inclusion of a rear anti-tip castor wheel/s. All of the different features demonstrate how these wheelchairs are being designed to meet the functional, task specific demands of each activity. However, due to the considerable differences between each wheelchair's general design, translating findings from studies that have manipulated daily life or racing wheelchair configurations to the wheelchair court sports is not recommended.

Figure 2.3 – Active daily living wheelchair (left) and a racing wheelchair (right).

Figure 2.4 – Typical wheelchair configurations observed in the wheelchair court sports: (a) wheelchair basketball, (b) wheelchair rugby, (c) wheelchair tennis.

2.5.2 Protocols

Comparisons between previous wheelchair configuration studies and inferences towards the wheelchair court sports have also been hindered by the vast range of experimental speeds and workloads employed. It has been projected that during competitive WCB match play, the mean velocities are approximately 2 $m \cdot s^{-1}$ with intermittent bursts of high intensity activity reaching speeds in excess of 4 $m \cdot s^{-1}$ (Coutts, 1992). As demonstrated by the manipulation studies in Tables 2.1, 2.2, 2.3 and 2.4, the speeds adopted have been somewhat sub-maximal by comparison, which reiterates the predominantly daily life focus of previous research.

Methods pertaining to workload, as quantified by P_O, have also varied considerably between previous wheelchair configuration investigations, which can have a substantial bearing upon the ergonomics of sports wheelchair performance. This is of further importance in the context of the current thesis based on the fact that wheelchair configuration can influence vehicle mechanics such as rolling resistance, air resistance and internal friction, which ultimately affect P_O (Woude et al., 2001). Unfortunately, not all previous ergonomic investigations into wheelchair configuration have reported P_O. This is a slight concern as it would facilitate comparisons between studies if an indicator of workload is provided. Also on one instance, P_O has been controlled between different configurations through the use of a pulley system, whereby load is added to increase the resistance experienced (Veeger et al., 1989b). The intention of this approach was to purely investigate the role of wheelchair configuration on the ergonomics of daily life wheelchair propulsion, independent of any

effects it may have on P_O. However, the ecological validity of this approach is questionable, particularly in a sporting context, whereby P_O cannot be controlled by athletes during on court propulsion. If a certain wheelchair configuration influences the P_O requirement from its user and consequently the ergonomics of wheelchair propulsion, then this should not be excluded when interpreting the data. Alternatively, P_O should be calculated and reported whenever possible to facilitate interpretations, but not controlled as this prevents reliable findings relating to the effects of wheelchair configuration from being established.

2.5.3 Participant groups

Previous investigations into the ergonomics of wheelchair configuration have used AB and/or wheelchair dependent individuals. However, differences exist between these participant groups, where physiologically AB participants have been shown to possess a reduced ME during wheelchair propulsion tasks (Brown et al., 1990; Patterson and Draper, 1997). Variations in temporal parameters (Patterson and Draper, 1997), upper body joint kinematics (Brown et al., 1990; Veeger et al., 1992a) and force application (Veeger et al., 1992a; Robertson et al., 1996) have also been reported between these groups. Both groups can obviously display within group differences if factors such as age, experience and level of physical activity are not carefully controlled. However, previous studies have utilised AB participants as they are thought to be a more homogenous sample, due to the intact nature of their musculoskeletal system. Alternatively, large inter-individual differences exist amongst wheelchair dependent individuals, owing mainly to the severity of their physical impairment (Woude et al., 2001). The use of AB participants may be acceptable as a starting point for research into an area of wheelchair configuration, to overcome any potential heterogeneity within impaired individuals (Woude et al., 2001). Although, testing under the most ecologically valid conditions are recommended (Vanlandewijck et al., 2001). Therefore, when attempting to optimise the ergonomics of mobility performance in elite wheelchair athletes, these are the population group that need to be examined.

2.5.4 Simulation of wheelchair propulsion

The environment in which manual wheelchair propulsion is investigated is another important methodological consideration and potential confounding factor. Researchers have

often favoured the simulation of wheelchair propulsion in laboratories, as conditions can be better controlled and more accurately assessed (Vanlandewijck et al., 2001). This has resulted in a large number of physiological and biomechanical investigations being conducted on either wheelchair ergometers (WERG) or motor driven treadmills.

Two distinct types of WERG have been previously used to investigate manual wheelchair propulsion; a roller WERG or a computer-controlled simulated WERG. Both types of WERG allow the collection of reliable data due to the highly reproducible conditions created (Woude et al., 2001). A roller WERG, which uses either mechanical or electric brakes and allows for the inertia of the system to be quantified are thought to simulate wheelchair propulsion most reliably (Theisen et al., 1996; DiGiovine et al., 2001; Woude et al., 2001). These systems also allow for a reliable calculation of P_O through the incorporation of a deceleration test (Theisen et al., 1996). Wheelchair simulators can also determine P_O if momentary torque and velocity can be calculated (Niesing et al., 1990; Pare et al., 1993; Keyser et al., 1999; Rodgers et al., 2000). Certain wheelchair simulators also enable the calculation of 3D forces applied by the user to the wheel (Niesing et al., 1990; Keyser et al., 1999; Rodgers et al., 2000). This has already been highlighted as a valuable consideration when investigating the ergonomics of wheelchair propulsion, particularly injury risk. An advantage of a roller WERG is that participants' individual wheelchairs can be attached to replicate a more realistic form of propulsion. Alternatively, the wheelchair simulator prevents this, but does allow for adjustments to the simulator's configuration to be introduced, which is rarely possible in participant's own wheelchairs. The major limitation associated with both types of WERG is that they fail to account for any shifts in the centre of gravity during propulsion, which affects the rolling resistance of the wheelchair-user combination (Vanlandewijck et al., 2001). Also, the inertial forces exerted on the wheelchair, which are created by the accelerations and decelerations of the trunk are neglected due to the static attachment of the wheelchair to the device (Vanlandewijck et al., 2001). Therefore, although testing in a laboratory setting has proven popular due to the controlled environment it creates, the use of a WERG as a mode of propulsion can influence the reliability of the physiological and biomechanical data collected. For instance, the prevention of backward tilting in wheelchair ergometry has been suggested to affect the upper body joint kinematics and the forces applied compared to over-ground propulsion (Vanlandewijck et al., 2001).

Motor driven treadmills have also been frequently employed by previous investigations and may be viewed as a more realistic mode for simulating wheelchair

propulsion, since oscillations of the trunk and arms and steering issues are accounted for (Woude et al., 2001). External P_O can also be easily determined via a drag test, which detects the *Fdrag* acting on the wheelchair-user combination. Workload can be manipulated by altering the speed and/or incline of the treadmill or the mass attached to the wheelchair (Woude et al., 1986; Veeger et al., 1989b). Subsequently, the motor driven treadmill would seem to be a more ecologically valid method for assessing the physiological and biomechanical responses during manual wheelchair propulsion. However, the most realistic conditions for wheelchair athletes to be tested under are during over-ground propulsion to accurately simulate the conditions experienced on court. Admittedly, the collection of physiological and biomechanical data during over-ground manual wheelchair propulsion can be challenging. However, an assessment of performance during maximal effort sport-specific movements in a realistic environment is a necessity (Vanlandewijck et al., 2001). Such movements cannot be accurately simulated and assessed under controlled laboratory conditions.

2.5.5 Controlled standardisation of wheelchair configuration

Adjusting one area of wheelchair configuration can directly impact on another area (Cooper, 1998). In order to reliably determine the effects of manipulating an area of wheelchair configuration on mobility performance, standardisation methods are required on other areas of configuration. Failure to standardise other areas of wheelchair configuration has been a limitation associated with numerous previous manipulation studies, as demonstrated throughout section 2.6. In association with this, a number of previous studies investigating wheelchair configuration have intentionally manipulated more than one area of configuration at a time. This also prevents any direct cause and effect relationships between the influences of each individual area of configuration on mobility performance being established.

Standardisation methods should also be extended to account for anthropometric differences between participants. For instance a given seat height will affect the propulsion technique, muscle mechanics and physiological demand of a smaller individual differently to a taller individual, primarily due to the differences in distance between the shoulder and TDC. Woude et al. (1989b) were the first researchers to control for anthropometrical differences by accounting for upper extremity limb length. Woude et al. (1989b) used the degree of elbow extension exhibited when the hands were placed on TDC of the hand-rims as a method for

standardising seat height (Figure 2.5). Not only would this facilitate comparisons between studies, it would also improve the likelihood of identifying optimal wheelchair configurations, since the specific characteristics of each individual are catered for.

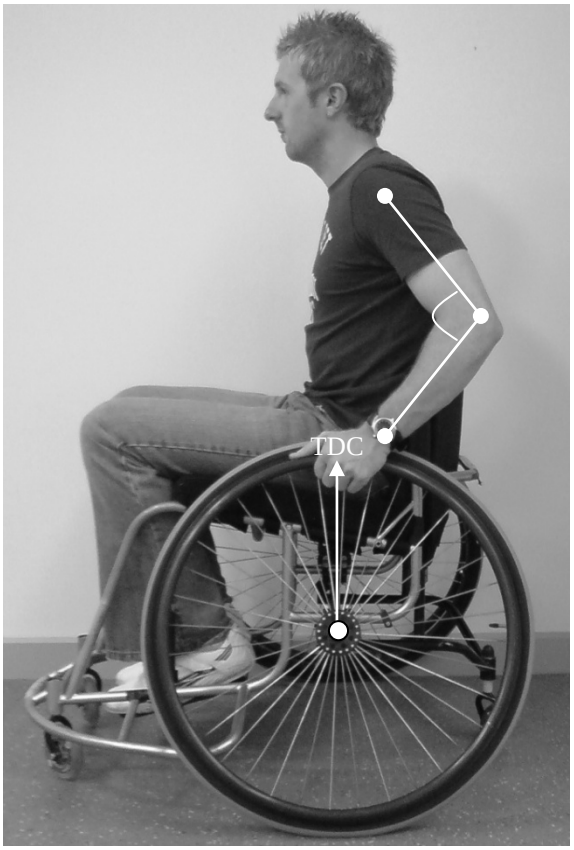

Figure 2.5 – Standardised method of seat height adjustment with hand positioned at TDC and subsequent joint angle induced at the elbow. Adapted from Woude et al. (1989b).

2.6 Wheelchair configuration

As previously mentioned, a court sports wheelchair is comprised of numerous individual components that can be configured in a variety of different ways (Figure 2.6). Due to the enormity of individual areas of configuration that may influence performance, it is not possible within the context of the current thesis to evaluate each of these in isolation. The areas of wheelchair configuration that have received the greatest amount of empirical research have focused on areas of the seat and the main wheels, which are reviewed throughout the current section. Unfortunately, these investigations have been conducted from a predominantly daily life focus and have introduced numerous confounding factors, as

discussed during section 2.5. This has often prevented translations being made to the wheelchair court sports, yet the findings and methodologies still warrant discussion to benefit future investigations.

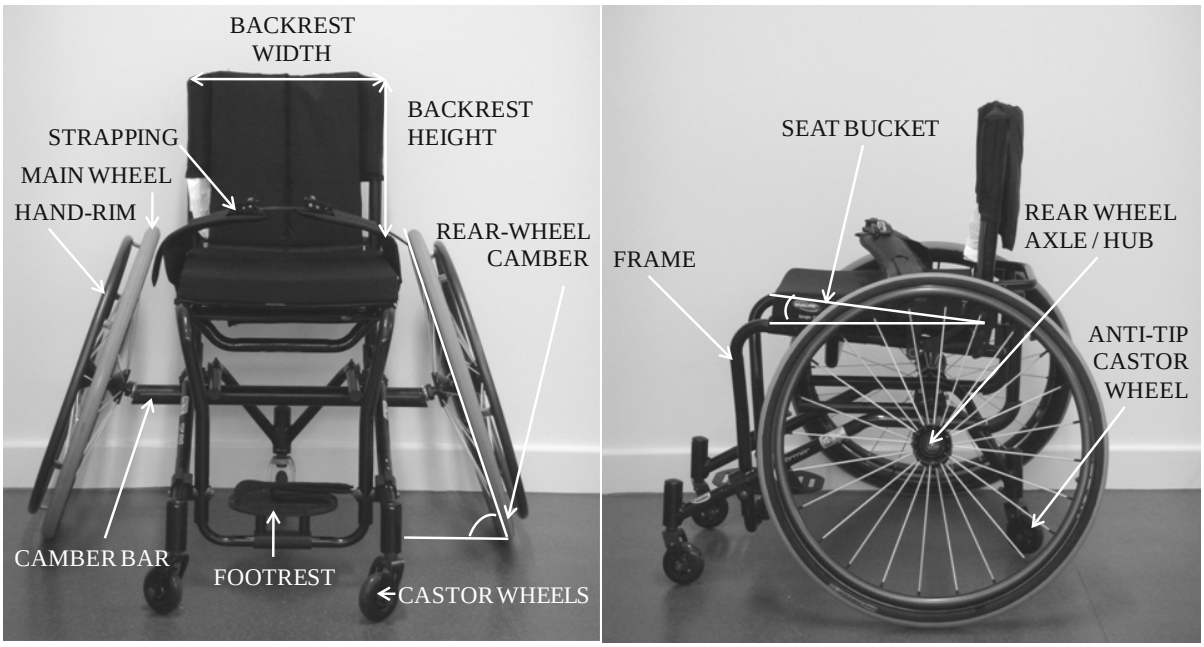

Figure 2.6 – Frontal (left) and sagittal (right) view of a typical court sports wheelchair and a number of the adjustable areas of configuration.

2.6.1 Seat positioning

The positioning of the seat in a vertical and horizontal direction (referred to as seat height and fore-aft position respectively) can be a critical decision for wheelchair users and sports performers alike (Higgs, 1983; Cooper, 1998). The position of the seat influences the centre of gravity, and subsequently the stability of the wheelchair-user combination (Masse et al., 1992; Majaess et al., 1993; Vanlandewijck et al., 1999b). Adjustments to seat position can also affect vehicle mechanics, such as the drag forces experienced (Vanlandewijck et al., 1999b). However, the resultant effects that such changes can have on the ergonomics of sports wheelchair performance remains somewhat limited due to the variety of methodological approaches adopted and the subsequent confounding factors introduced. As highlighted in Table 2.1, the prominent confounding factor between previous seat position investigations has been the frequency with which vertical and horizontal adjustments have been analysed simultaneously. This makes it difficult to underpin whether any adaptations to the ergonomics of wheelchair performance are the direct result of height adjustments, fore-aft adjustments or a combination of both.

Table 2.1 – Testing procedures of previous studies investigating the effects of vertical and/or horizontal seat positions on the ergonomics of wheelchair performance.

Reference	Height / Fore-aft	Wheelchair	Participants	Mode	Speeds ($m \cdot s^{-1}$)	Power Output	Standardisation	Measures
Woude et al. (1989b)	H	WCB	AB – 9	MDT	0.55 – 1.39	0.17 – 0.44 W/kg	S - elbow angle (°)	Physiology, temporal, kinematics
Woude et al. (1990)	H	n/a	AB – 3 WD – 2	WERG [s]	1.11 – 1.39	Reported, not controlled	S - elbow angle (°)	Physiology, temporal
Woude et al. (2009)	H	n/a	WD – 12	WERG [s]	0.42 – 0.83	Reported, not controlled	S - elbow angle (°)	Physiology, temporal, kinetics
Brubaker et al. (1980)	H & F	ADL	AB – 4	WERG [r]	Not reported	0.25 W/kg	U	Physiology, temporal
Brubaker et al. (1984)	H & F	ADL	WD – 9	WERG [r]	1.1	20 W	U	Physiology
Walsh et al. (1986)	H & F	RAC	WD – 9	WERG [r]	3.45 – 6.18	Not controlled or measured	U	Performance
Hughes et al. (1992)	H & F	n/a	AB – 9 WD – 6	WERG [s]	0.83	15 W	S - elbow angle (°) & arm length (%)	Temporal, kinematics
Masse et al. (1992)	H & F	RAC	WD – 5	WERG [r]	3.22 – 3.81	Not controlled or measured	S - elbow angle (°) & U (fore-aft)	Temporal, kinetics
Boninger et al. (2000)	H & F	ADL	WD – 40	WERG [r]	0.9 & 1.8	Not controlled or measured	U	Temporal, kinetics
Wei et al. (2003)	H & F	ADL	WD – 11	WERG [r]	Self selected (not reported)	Not controlled or measured	S - elbow angle (°) & arm length (%)	Temporal, kinematics
Kotajarvi et al. (2004)	H & F	WCB	AB – 20 WD – 13	OG	Self selected (1.48 ± 0.16)	Not controlled or measured	U	Temporal, kinetics
Samuelsson et al. (2004)	H & F	ADL	WD – 12	MDT	1.0	Reported, not controlled	U	Physiology, temporal
Gutierrez et al. (2005)	F	ADL	WD – 13	WERG [r]	Self selected (1.07 – 2.22)	Not controlled or measured	S - elbow angle (°) U (fore-aft)	Kinetics
Mulroy et al. (2005)	F	ADL	WD – 13	WERG [r]	Self selected (1.07 – 2.22)	Not controlled or measured	As Gutierrez et al. (2005).	Kinetics

Key: **Wheelchair** – (WCB) Basketball, (ADL) Daily Life, (RAC) Wheelchair Racing; **Participants** – (AB) Able-bodied, (WD) Wheelchair User; **Mode** – (WERG) Wheelchair Ergometer, [r] roller, [s] simulator, (MDT) Motor Driven Treadmill, (OG) Over-ground. **Height / Fore-aft** – (H) Height, (F) Fore-aft; **Standardisation** – (S) Standardised, (U) Un-standardised.

2.6.1.1 Seat height

From a physiological perspective the majority of investigations into seat positioning have focused on its relationship with ME, with varying results reported (Brubaker et al., 1980; Woude et al., 1989b; Woude et al., 1990; Samuelsson et al., 2004; Woude et al., 2009). Brubaker et al. (1980) and Samuelsson et al. (2004) revealed no differences in ME between different seat heights, whereas Woude and colleagues (1989b; 2009) identified a significant effect of seat height on ME. The investigations of Brubaker et al. (1980) and Samuelsson et al. (2004) manipulated a combination of seat positions in a vertical and horizontal direction (Table 2.1), which as previously mentioned is an unfavourable approach. However, a further common denominator that existed within the methodologies of Woude and colleagues (1989b; 2009), which differ to those of Brubaker et al. (1980) and Samuelsson et al. (2004) relates to the standardisation methods applied during seat height manipulations. Both Brubaker et al. (1980) and Samuelsson et al. (2004) made absolute changes to the height of the seat that were not standardised to the anthropometric characteristics of the user. These changes only evoked a difference of 10.1 cm (Brubaker et al., 1980) and 5.5 cm (Samuelsson et al., 2004) between the two extreme seat height settings, which may not have been sufficient to induce any physiological adaptations during sub-maximal propulsion. Alternatively, Woude and colleagues (1989b; 1990; 2009) investigated seat height in isolation and standardised the adjustments to the anthropometrics of the user (Figure 2.5). The range of seat heights investigated across these investigations spanned from 70° to 160° elbow extension (whereby 180° represent full extension), which may have been more substantial than those previously examined and increased the likelihood of observing physiological adaptations. Investigating seat heights ranging from 100° to 160° elbow extension, Woude et al. (1989b) revealed that the higher seat heights (140° & 160°) increased $\dot{V}O_2$ and reduced ME in comparison to the lower seat heights (100° & 120°). When evaluating a lower range of seat heights (70° to 90°), Woude et al. (1990) revealed that seat height had no significant effect on P_O and also revealed that the lowest two settings (70° & 80°) increased $\dot{V}O_2$ in relation to the highest setting (90°). These two investigations suggested that a physiologically optimal seat height existed, since physiological demand was elevated in the extreme high (Woude et al., 1989b) and low (Woude et al., 1990) settings. Unfortunately, optimal seat heights could not be proposed from these two investigations, primarily due to the combination of both AB and wheelchair dependent participants, which should be avoided (Patterson and Draper, 1997).

A more recent study, which investigated wheelchair users across a larger range of seat heights (70° to 140°), was conducted to reliably establish optimal ranges of seat height settings (Woude et al., 2009). It was again established that seat height manipulations had no significant effect on the P_O requirements from the user. Significant effects of seat height were though established for physiological responses, with favourable effects on $\dot{V}O_2$ and ME revealed between seat heights inducing 100° to 130° elbow extension. The authors proposed that this range of seat heights was optimal for daily life wheelchair propulsion. Although this information could not be translated to a sporting population performing high intensity tasks, it demonstrated the importance of seat height optimisation. Woude et al. (2009) revealed that an absolute change of 1.5% in ME was achievable through seat height manipulations, which reflected a potential relative improvement of 25 %.

Given the absence of any incidental effects on P_O from seat height adjustments, changes in physiological demand were likely to be due to the biomechanical adaptations that these adjustments also cause. Manipulating the height of the seat ultimately dictates how accessible the wheels are for the users and can subsequently affect propulsion technique and temporal parameters during the propulsion cycle. A number of studies have identified that standardised adjustments in sitting height caused modifications to upper body joint kinematics during manual wheelchair propulsion (Woude et al., 1989b; Hughes et al., 1992; Masse et al., 1992; Richter, 2001; Wei et al., 2003). An increased trunk RoM has been observed with increasing seat height, which is likely to compensate for the decreased reach of the users in relation to the wheels (Woude et al., 1989b). A qualitative assessment of muscular activity was also incorporated by Woude et al. (1989b) and revealed that the rectus abdominis and erector spinae were active for longer periods at higher seat positions. These muscles are responsible for trunk flexion and extension respectively and the increase in muscular activity of these relatively large muscle groups, would not only explain the greater trunk RoM but may account for the decreased ME observed at the higher sitting heights (Woude et al., 1989b). The actions of the elbow (Woude et al., 1989b; Richter, 2001) and the wrist (Wei et al., 2003) have also been shown to be influenced by seat height. Woude et al. (1989b) identified that increased seat heights led to greater degrees of elbow extension, which could explain the increase in elbow torque predicted by Richter (2001). This was reinforced by the qualitative electromyography (EMG) data, whereby triceps activation commenced earlier during the propulsion cycle, over a prolonged period of time (Woude et al., 1989b) and at a greater magnitude (Masse et al., 1992) in higher seat positions. Similarly to the trunk, the

cause of this could be related to the increased necessity to reach in order to access a sufficient portion of the wheels at a higher seat height. Wei et al. (2003) revealed that a decreased RoM occurred at the wrist in higher seat positions primarily owing to the fact that the wrist remained in a more extended position throughout the propulsion cycle.

Like the wrist, shoulder motion has been shown to become restricted in higher seat positions. Using mathematical modelling, Richter (2001) predicted that a decrease in shoulder torque would occur when the distance between the shoulder and the hub is increased. This appeared to support the findings of Woude et al. (1989b), who revealed that abduction/adduction and flexion/extension RoM at the shoulder decreased with increasing seat height. The fact that shoulder RoM decreased in the sagittal plane at higher seat heights could be further explained by the observation that the anterior deltoid and pectoralis major were shown to be active for shorter periods (Woude et al., 1989b). During wheelchair racing propulsion, Masse et al. (1992) also established that the magnitude of both these muscles' activity was in fact greater at higher seat positions. Based on the findings of Woude et al. (1989b) relating to muscular activity and physiological demand it could be proposed that propulsion became less efficient in the higher seat positions investigated by Masse et al. (1992). Despite all of the resultant alterations to propulsion technique caused by manipulating seat height, Woude et al. (1989b) revealed that peak angular velocities of upper body joints still occurred in sequence. A proximal to distal sequencing pattern was established from the trunk to the shoulder to the elbow and was evident for all seat heights investigated (Woude et al., 1989b). Subsequently, all that appears to be influenced is the RoM permitted at each joint during propulsion, whereby higher seat positions increased the involvement of the trunk and elbow, whereas the shoulder and wrist seem to be inhibited.

Lower seat heights have also been associated with increases in PA and PT, potentially due to the larger portion of the hand-rim that is available (Woude et al., 1989b; Masse et al., 1992; Vanlandewijck et al., 1999b; Boninger et al., 2000; Wei et al., 2003; Kotarjarvi et al., 2004; Samuelsson et al., 2004) and reductions in push frequency (Masse et al., 1992; Samuelsson et al., 2004). Given the increased push frequency associated with increasing seat height and the smaller RoM of the shoulder, a greater force may be necessitated in order to maintain a given wheelchair velocity. The only previous force application investigation into standardised seat height manipulations supported this assumption, whereby Woude et al. (2009) established increases in the total force applied to the hand-rims with increasing seat height. Subsequently, this may have implications on user safety/health at higher seat positions

given the relationship between force magnitude, push frequency and injury risk (Boninger et al., 1999; Boninger et al., 2002; Boninger et al., 2004).

Only Walsh et al. (1986) have investigated different seat positions whilst performing maximal effort bouts of propulsion so it is clear that further research into the effects on sport performance is required. Walsh et al. (1986) revealed that seat height had no effect on the maximal velocities reached. Unfortunately, as demonstrated in Table 2.1 this study was conducted in a racing wheelchair configuration on a roller WERG and manipulated fore-aft positions in addition to seat height adjustments. Therefore, the results were unlikely to have been applicable to maximal effort, over-ground propulsion in the wheelchair court sports. Subsequently, more attention needs to be dedicated to the influence that standardised seat height adjustments can have on sport specific movements performed at speeds that are reflective of match conditions.

2.6.1.2 Fore-aft position

As with seat height, adjustments in a fore-aft direction also directly influence the centre of gravity of the wheelchair-user combination and the stability of the user (Masse et al., 1992; Majaess et al., 1993). However, the influence of this type of manipulation on the ergonomics of sports wheelchair performance again remains unclear. This is largely due to the variety of methodological approaches employed between studies (Table 2.1). In particular, only Gutierrez et al. (2005) and Mulroy et al. (2005) have investigated fore-aft seat positions in isolation. All other ergonomic investigations into this area of wheelchair configuration have done so in conjunction with seat height manipulations (Table 2.1). In addition to this, standardisation methods have also been somewhat lacking in previous fore-aft investigations (Hughes et al., 1992; Wei et al., 2003). Both these studies quantified changes in fore-aft position to the anthropometrics of the user by using arm length percentiles as a means for adjusting the seat base/backrest intersect position in relation to the axle.

Brubaker et al. (1980; 1984) and Samuelsson et al. (2004) are the only authors to have given any consideration to the physiological responses to manipulations in fore-aft seat position. Samuelsson et al. (2004) revealed that neither $\dot{V}O_2$, HR nor ME was affected by two different seating positions, which may seem unsurprising given that these positions were un-standardised and differed by only 1.2 cm horizontally. Although, no statistical results were

reported by Brubaker et al. (1980), trends existed for the posterior position at all three heights to result in increased $\dot{V}O_2$ and reduced ME. In contrast, a follow up study showed a trend for the posterior position at all three seat heights to elicit a higher ME (Brubaker et al., 1984). Unfortunately, no distinctions can be made between these two investigations due to the lack of standardisation in relation to the anthropometrics of the users and the fact that differences between the absolute positions of the seat in relation to the axle existed. For instance, the posterior position examined by Brubaker et al. (1984) was only approximately 20 cm behind the axle, whereas this position was 40 cm behind the axle in the earlier investigation (Brubaker et al., 1980). As previously established with seat height, comparisons need to be made between the anthropometrics of the user and the wheelchair in order to investigate optimal seat positions. Only then can more reliable findings relating the physiological demands of making adjustments to the fore-aft seat position be established.

The fore-aft positioning of the seat has also been suggested to influence the biomechanics of wheelchair propulsion. More posterior seat positions have been suggested to permit a greater PA (Masse et al., 1992; Kotajarvi et al., 2004; Samuelsson et al., 2004). However, each of these studies investigated a combination of vertical and horizontal seat positions, so it cannot be reliably ascertained whether the increases in PA were the result of seat height changes, fore-aft changes or a combination of both. Boninger et al. (2000) investigated 40 manual wheelchair users in their own daily life wheelchairs and documented the horizontal positioning of the shoulder in relation to the axle of the main wheel. A more anterior seat position was significantly correlated with a decreased PA and push frequency. Therefore, being positioned slightly behind the main wheels seemed to increase the portion of hand-rim available for propulsion. It also decreased the need for such a frequent stroke rate, which could offer support to the efficiency trends suggested by Brubaker et al. (1984) based on the association between push frequency and physiological demand (Jones et al., 1992; Goosey and Campbell, 1998; Goosey et al., 2000; Goosey-Tolfrey and Kirk, 2003).

Upper body joint kinematics during manual wheelchair propulsion have also been affected by adjustments in the fore-aft positioning of the seat. However, unlike adjustments to seat height, it has been revealed that changes in the fore-aft direction have no bearing on the motion of the trunk (Hughes et al., 1992; Masse et al., 1992) or the wrist (Wei et al., 2003). Elbow RoM has been revealed to be affected by changes in fore-aft position with a reduced RoM demonstrated in the standardised posterior settings examined by Hughes et al. (1992). In

the posterior settings investigated by Masse et al. (1992), EMG data revealed that triceps activity was reduced, which was likely to be due to a greater contribution of the biceps during a pulling motion in this position. Alternatively, in a more anterior setting the biceps are less likely to contribute, placing a greater demand on the triceps to drive the wheels.

Shoulder motion seemed to be the most significantly affected joint in different fore-aft seat positions. Hughes et al. (1992) identified an increase in shoulder RoM in the sagittal plane during the posterior fore-aft positions. It seemed as though this increase in RoM reduced the muscular activity of the muscles surrounding the shoulder joint, as EMG activity in the anterior deltoid and the pectoralis major have been reduced in posterior settings (Masse et al., 1992; Gutierrez et al., 2005). Therefore, posterior seat positions appear to allow the shoulder to act over a greater range and due to the associated decrease in push frequency, may also have a bearing on minimising injury risk. This was supported to a certain extent by Mulroy et al. (2005) who through the use of an instrumented hand-rim and an inverse dynamics algorithm revealed that the resultant forces acting on the shoulder were reduced in their posterior setting. Boninger et al. (2000) also revealed significant correlations between a reduced rate of force development and posterior settings, further implicating this type of setting for improved user safety/health (Boninger et al., 2002).

Although it seemed probable that posterior settings may be more favourable from an efficiency and safety/health perspective, the lack of standardised settings in the majority of previous investigations has prevented optimal settings being established. In order to optimise this position for wheelchair sportsmen, standardisation methods similar to those employed by Hughes et al. (1992) and Wei et al. (2003) are vital. These settings then need to be examined under more sport specific conditions, as again only Walsh et al. (1986) assessed maximal effort performance during a combination of different vertical and horizontal seat positions. It was also apparent that user acceptance and subsequently comfort has not been considered by previous ergonomic investigations into seat positioning

2.6.2 Main wheels

The main wheels are responsible for driving and manoeuvring the wheelchair, yet there are still numerous components concerning this area of wheelchair configuration that can be manipulated to influence the ergonomics of sports wheelchair performance. Obviously

factors such as the design of the tyres and the inflation pressure may influence vehicle mechanics and mobility performance (Woude et al., 2001). However, for the purpose of this review the areas of rear-wheel camber, wheel size and hand-rim configuration were all investigated based on the significant amount of empirical research received, albeit from a predominantly daily life perspective.

2.6.2.1 Rear-wheel camber

Rear-wheel camber has been defined by Higgs (1983) as the angle of the main wheel in relation to the vertical (Figure 2.6). Camber is now a particularly common feature of court sports wheelchairs, with increasing degrees being selected (Cooper, 1998, Woude et al., 2001; Ardigo et al., 2005). Polic (2000) has stated that the degree of camber selected in wheelchair court sports can now be as extreme as 24°. A rationale for selecting increased camber relates to the increased wheelbase that it generates (Trudel et al., 1995). The main advantage resulting from the increased wheelbase has been through the subsequent improvements in the lateral stability of the wheelchair-user combination (Trudel et al., 1997). This wider wheelbase can also provide a greater deal of protection to the hands and fingers, which is of relevance to WCB and WCR, whereby contact with other wheelchairs is common (Veeger et al., 1989b; Cooper, 1998; Woude et al., 2001). Alternatively, a disadvantage of a wider wheelbase is the increased difficulty users can experience when negotiating smaller gaps (Ball, 1994; Trudel et al., 1995; Faupin et al., 2004; Perdios et al., 2007). This would again appear to have implications for WCB and WCR players, who dependent upon their role, often need to ‘pick’ or avoid opposing players and the width of their wheelbase can assist or hinder the performance of these tasks (Faupin et al., 2004).

Rear-wheel camber is a particularly complex area of wheelchair configuration, as manipulations can have consequential effects on vehicle mechanics and other areas of wheelchair configuration. Rolling resistance is thought to increase with increasing camber due to a proportionate increase in the contact area between the tyres and surface, as a result of greater tyre deformation (Cooper, 1998; Faupin et al., 2004). Veeger et al. (1989b) did not experience this relationship between camber and rolling resistance, which was likely to be due to the absence of any standardisation procedures to control for the alignment of the wheels in the transverse plane (Figure 2.7), often referred to as toe-in toe-out (Frank and Abel, 1993; Ball, 1994; Cooper, 1998). Toe-in toe-out was identified as a key confounding factor that

needed to be controlled between camber settings, as O'Reagan et al. (1981) reported that as little as 2° of 'toe' can double the rolling resistance experienced.

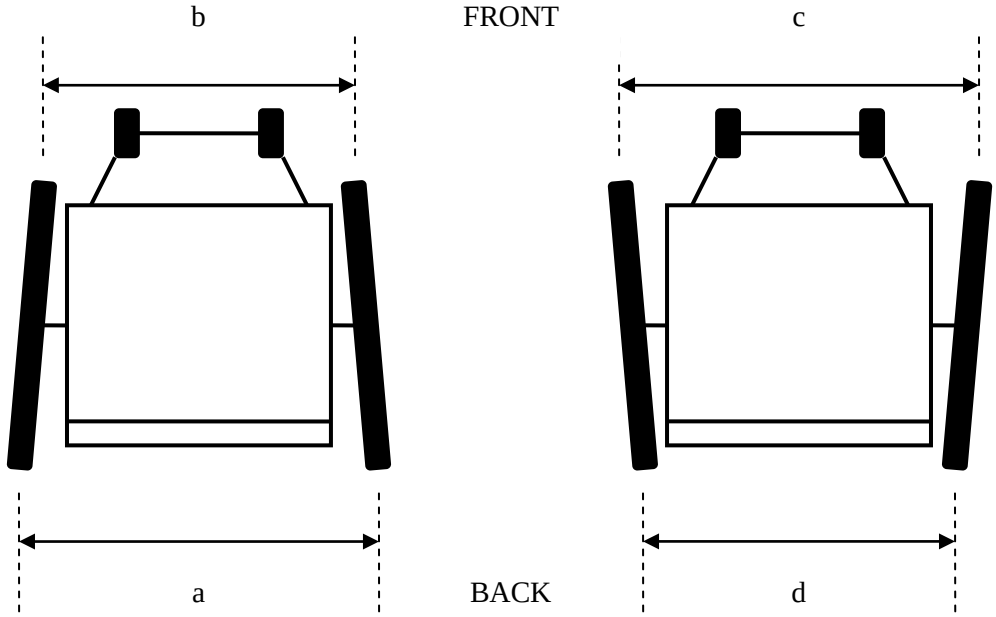

Figure 2.7 – View of a wheelchair in the transverse plane demonstrating toe-in (left) and toe-out (right). Toe-in is depicted by a greater distance at the back of the main wheels compared to the front (a > b). Toe-out is represented by a greater distance at the front compared to the back (c > d).

Adjusting the degree of rear-wheel camber has also been suggested to have a direct impact upon other areas of a wheelchair's configuration, particularly the seat height and as a result the centre of gravity of the wheelchair-user combination (Trudel et al., 1997). Manipulating rear-wheel camber can also affect the distance between TDC of both main wheels. If both these by-products of camber adjustments are not controlled for, the propulsion kinematics will be affected and a false interpretation of the ergonomic effects of camber can be deduced. Veeger et al. (1989b) overcame the effect of seat height changes evoked by camber adjustments by making minor modifications to the seat in order to maintain a constant elbow angle between camber conditions. Attempts have also been made to control the distance between TDC of both main wheels through the use of different length camber bars (Buckley and Bhambhani, 1998; Faupin et al., 2004). However, as demonstrated in Table 2.2, standardisation methods are not always reported and appear to have never been applied to all subsidiary areas of wheelchair configuration associated with camber adjustments during the same investigation.

Table 2.2 – Summary of previous investigations into the ergonomics of rear-wheel camber manipulations, highlighting the predominant focus on aspects of daily life wheelchair propulsion and documenting the methodologies employed.

Reference	Wheelchair	Camber (°)	Standardisation			Participants	Mode	Speeds ($m \cdot s^{-1}$)	Power Output	Measures
			Seat Height	'Toe'	TDC to TDC					
Buckley & Bhambhani (1998)	ADL	0, 4 & 8	U	S	S	AB – 19	WERG [r]	0.56	Not controlled or measured	Physiology
Perdios et al. (2007)	ADL	0, 3 & 6	U	U	U	AB – 21 WD – 13	OG & Q	Self selected (not reported)	Not controlled or measured	Physiology, subjective
Veeger et al. (1989b)	WCB	0, 3, 6 & 9	S	U	U	AB – 8	MDT	0.56 – 1.39	0.17 – 0.44 W/kg	Physiology, temporal, kinematics
Faupin et al. (2002)	WCB	9, 12 & 15	?	?	?	WD – 9	OG	Maximal effort	Not controlled or measured	Performance
Faupin et al. (2004)	WCB	9, 12 & 15	U	S	S	WD – 8	WERG [r]	Maximal effort	Reported, not controlled	Temporal, performance
Lim et al. (2004)	WCB	16 & 20	?	?	?	AB – 9	OG & MDT	Maximal effort & 1.11 -1.39	?	Physiology, performance
Wang et al. (2004).	WCB	8, 16 & 20	?	?	?	WD – 3	MDT	1.2	?	Kinetics

Key: **Wheelchair** – (WCB) Basketball, (ADL) Daily Life; **Participants** – (AB) Able-bodied, (WD) Wheelchair User; **Mode** – (WERG) Wheelchair Ergometer, [r] roller, (MDT) Motor Driven Treadmill, (OG) Over-ground, (Q) Questionnaire; **Standardisation** – (S) Standardised, (U) Un-standardised.

As a result of differing methodologies, the predominant focus on daily life ambulation and absence of stringent standardisation methods (Table 2.2), the effects of camber on the ergonomics of sports wheelchair performance are not well understood. The physiological adaptations to rear-wheel camber have received a rather limited amount of research attention and have yielded varying results. Buckley and Bhambhani (1998) revealed an increase in physiological stress during wheelchair propulsion with increasing camber, as demonstrated by increased $\dot{V}O_2$ and HR values. However, studies by Veeger et al. (1989b) and Perdios et al. (2007) identified no significant effect of camber on these variables, with Veeger et al. (1989b) and Lim et al. (2004) also reporting that ME was unaffected. Although Veeger et al. (1989b) measured the effects of camber on rolling resistance and reported P_O, they failed to control for toe-in toe-out between conditions (Table 2.2). This was likely to have produced unreliable findings relating to the resistances experienced, and the subsequent P_O and physiological responses observed. In addition to this, Veeger et al. (1989b) also corrected for any changes in rolling resistance to maintain a constant P_O between camber conditions, which as previously alluded to is not a favourable approach from a sporting perspective due to the lack of control players have over P_O on court (section 2.5.2). Although Buckley and Bhambhani (1998) experienced a significant effect of camber on physiological demand when toe-in toe-out was standardised, seat height was not controlled for. Unfortunately limited translations can be made to a sporting environment due to the daily life focus of these physiological investigations, as evidenced in particular by the daily life wheelchairs and the low camber settings investigated (Table 2.2).

Only minimal research has investigated the influence of camber on the biomechanics of wheelchair propulsion. Veeger et al. (1989b) revealed that adjusting camber between 0° and 9° had a significant impact upon certain technique parameters during the push phase. It was shown that 6° camber significantly increased the PA, PT and shoulder abduction exhibited during the push phase in relation to other settings. The authors proposed that these curvilinear results may imply that the 6° was optimal for the AB participants sampled, as it implied that force can be applied over a greater range, without negatively influencing ME. However, the failure to standardise certain additional areas of the wheelchair's configuration between different camber settings severely compromised the validity of the kinematic findings. In particular, its failure to control the distance between TDC of both main wheels had a major influence on the results, especially those relating to shoulder abduction.

As a result of the limited biomechanical data attributed to the effects of camber, little can be derived relating to the role of this area of configuration upon user safety/health. The force application patterns exhibited during extreme camber ranges, more reflective of those used within the wheelchair court sports have been briefly investigated by Wang et al. (2004). It was established that peak total forces increased significantly with camber. This could be construed that greater camber settings may place the user at an increased risk of injury. However, interpreting the findings of Wang et al. (2004) may be troublesome as the results were only published as conference proceedings, therefore specific details concerning the standardisation methods and precautions taken were somewhat lacking. User acceptance and comfort associated with various camber settings has only been considered by Perdios et al. (2007), who revealed that acceptance improved with increased camber. Although perceptions were based on over-ground propulsion, which is favourable, this research again focused on daily life wheelchair propulsion and subsequently only investigated limited camber angles.

Compared to other areas of wheelchair configuration, rear-wheel camber has actually received a reasonable amount of research from a sports performance perspective. Studies by Faupin et al. (2002; 2004) and Lim et al. (2004) have investigated the effects of manipulating camber upon the movements and intensities that are reflective of those performed in the wheelchair court sports. Faupin et al. (2004) investigated the maximal linear sprinting performance of WCB players and established that increments in camber were accompanied by significant increases in PT and decreases in mean velocities during 8-second sprints on a roller WERG. This suggested that players were less effective when pushing with higher degrees of camber, as sprinting performance was negatively affected, even though the time over which they were applying force to the wheels increased. An earlier field based study by Faupin et al. (2002) did not support the latter study's findings with regards to straight line performance. Alternatively, using the same three camber settings, no significant effect of camber was evident for the mean or peak velocities reached during a 15 m over-ground sprint. In addition to the advantage of being investigated during over-ground propulsion, a further benefit of the field based study (Faupin et al., 2002) was that it considered the effect of camber upon manoeuvrability performance. During a 'figure of eight' drill it was revealed that camber had a significant effect on turning performance, with increasing degrees of camber shown to reduce the times taken to perform turns. Therefore, the results of this study suggested that 15° may be the optimal camber setting for the group of WCB players investigated given the improvements in manoeuvrability, without any significantly negative

consequences on linear performance. Unfortunately standardisation methods were not reported preventing the validity of these findings from being presented.

Although Faupin et al. (2002; 2004) both have far greater relevance to wheelchair court sports performance, there were still a number of limitations associated with these studies. Both investigations revealed linear relationships with regards to aspects of linear and manoeuvrability performance up to 15° camber. However, more extreme ranges of camber, reflective of those currently used in the wheelchair court sports need to be investigated in order to establish truly optimal settings. Lim et al. (2004) identified that turn velocities actually increased between 16° and 20° camber, suggesting that an optimal in terms of manoeuvrability may lie in this range of camber settings. Unfortunately, as with Wang et al. (2004), Lim et al. (2004) was also only presented as conference proceedings, therefore the findings derived were again somewhat limited due to the absence of detailed methodologies. Also, as highlighted in Table 2.2, Faupin et al. (2002; 2004) failed to control the influence that camber can have on sitting position. As previously mentioned, increasing camber can reduce the overall height of the frame (Trudel et al., 1995), which if not compensated for, can increase the distance between TDC and the shoulder and alter propulsion kinematics. Therefore, some of the findings from Faupin et al. (2002; 2004) could be attributed to adjustments to the sitting position of the user in relation to the wheel. Although, these are only likely to be minor modifications, it prevents a direct cause and effect between camber and performance from being established. Subsequently, seat height standardisation methods, similar to those employed by Veeger et al. (1989b) are necessary for future investigations in order to eliminate the influence that other areas of wheelchair configuration can have. Finally, although linear performance has been investigated, no consideration has been afforded to the effect that camber can have on initial acceleration, which is one of the key determinants of mobility performance in the wheelchair court sports (Vanlandewijck et al., 2001).

2.6.2.2 Wheel and hand-rim diameter

The size of the main wheels is another aspect of wheelchair configuration that can vary quite dramatically between athletes within the wheelchair court sports (Hutzler et al., 1995). The size of the main wheels affects the rolling resistance of the wheelchair-user combination, with a smaller wheel known to experience a greater resistance at a given velocity (Kauzlarich and Thacker, 1985). The implications of the mechanical by-products

associated with wheel size have rarely been investigated from an ergonomics perspective and to the author's knowledge have only been considered in an investigation published as conference proceedings (Lim et al., 2004). The investigation by Lim et al. (2004), which was previously referred to in relation to rear-wheel camber, also investigated the effects of two different wheel size conditions relevant to the wheelchair court sports. This investigation revealed that 24" wheels increased the peak velocities that could be reached within a 10 m over-ground sprint in comparison to 26" wheels, whereas, 26" wheels enabled greater velocities to be reached during a 20 m sprint. It was also suggested that wheel size had no effect on ME during steady state wheelchair propulsion. However, it must be reiterated that little can be interpreted from this study since major details relating to standardisation methods in particular have been omitted. Subsequently, the practical effects of wheel size manipulations on the ergonomics of daily life or sports wheelchair performance remains relatively unknown.

The majority of ergonomic investigations relating to the size of the main wheels have been attributed to the diameter of the hand-rims (Woude et al., 1988b; Gayle et al., 1990; Guo et al., 2006; Costa et al., 2009). All of these investigations have manipulated the diameter of the hand-rims in relation to a fixed wheel size. Subsequently, no effects on rolling resistance at a given velocity are likely. Alternatively, any changes in the ergonomics of wheelchair performance relating from hand-rim diameter manipulations are the result of changes in the gear ratios used for propulsion (Veeger et al., 1992b). However, as documented in Table 2.3 there are again numerous confounding factors introduced in previous investigations resulting from the methodological protocols and the standardisation methods imposed. As documented in previous sections, the focus of wheelchair configuration investigations has frequently been attributed towards aspects of daily life wheelchair propulsion. However, it was clear that the majority of hand-rim diameter investigations were conducted specific to wheelchair racing, which as mentioned previously cannot be translated to the wheelchair court sports due to the differences in the movements performed and the general designs of the two wheelchairs.

Table 2.3 – Testing procedures adopted by previous investigations into the effects of wheel and or hand-rim diameter on the ergonomics of wheelchair performance.

Reference	Wheelchair	Participants	Mode	Speeds ($m \cdot s^{-1}$)	Power Output	Hand-rim Diameters (m)	Seat Position	Measures
Wheel size:								
Lim et al. (2004)	WCB	AB – 9	OG & MDT	1.11 -1.39 & maximal effort	?	?	?	Physiology, performance
Hand-rim diameter:								
Veeger et al. (1992b)	n/a	AB – 9	WERG [s]	0.83 – 1.67	0.25 & 0.5 W/kg	0.26	S	Physiology, kinetics, temporal, kinematics
Guo et al. (2006)	ADL	AB – 12	?	Self selected	Reported, not controlled	0.32, 0.43 & 0.54	U	Kinetics, kinematics
Woude et al. (1988b)	RAC	WD – 8	MDT	0.83 - 4.17	Measured, not reported	0.30, 0.35, 0.38, 0.47 & 0.56	U	Physiology, temporal, kinematics
Gayle et al. (1990)	RAC	WD – 15	WERG [r] & OG	1.1, 2.2 & maximal effort	Not controlled or measured	0.25 & 0.41	U	Physiology, performance
Costa et al. (2009)	RAC	WD – 1	OG	3.33 – 6.67	Not controlled or measured	0.34, 0.36 & 0.37	U	Physiology, temporal

Key: **Wheelchair** – (WCB) Basketball, (ADL) Daily Life, (RAC) Wheelchair Racing; **Participants** – (AB) Able-bodied, (WD) Wheelchair User; **Mode** – (WERG) Wheelchair Ergometer, [r] roller, [s] simulator, (MDT) Motor Driven Treadmill, (OG) Over-ground; **Seat Position** – (S) Standardised by elbow angle, (U) Un-standardised.

The physiological responses to hand-rim diameter adjustments in relation to fixed wheel sizes at incremental speeds have been investigated using wheelchair athletes (Woude et al., 1988b; Gayle et al., 1990; Costa et al., 2009). Investigating a total of five different diameter hand-rims, Woude et al. (1988b) revealed that the largest two settings significantly increased the physiological demand, through reductions in ME and increases in both $\dot{V}O_2$ and HR responses. Although HR was shown to be unaffected, Gayle et al. (1990) also reported a reduction in physiological demand with smaller hand-rim diameters. Lower $\dot{V}O_{2,}$ rating of perceived exertion (RPE) and blood lactate responses were reported during sub-maximal wheelchair propulsion in the smaller of two hand-rim diameters (Gayle et al., 1990). More recently, Costa et al. (2009) investigated the influence of three different hand-rim diameters on the physiological responses to wheelchair propulsion at incremental speeds. In contrast to Woude et al. (1988b) and Gayle et al. (1990), HR and blood lactate responses were elevated in the smallest hand-rim diameter setting at the lowest speed condition. However, this pattern was reversed at the higher speeds, with increases in these physiological measures observed with larger hand-rim diameters (Costa et al., 2009). The fact that different physiological trends were observed suggests that optimal settings are likely to exist but differ dependent on the speed of propulsion and the physical characteristics of the user.

A major drawback with these investigations again concerned the standardisation methods employed (Table 2.3). The use of fixed wheel sizes and seat heights created an increased distance between the shoulder and TDC of the hand-rim when hand-rim diameter is decreased. Therefore, body positioning in relation to the hand-rims may have had more of a bearing upon the physiological responses than the hand-rim diameter manipulations themselves, since muscle mechanics are likely to be affected. Biomechanical investigations into hand-rim diameter manipulations appeared to support the physiological findings of Woude et al. (1988b) and Gayle et al. (1990), whereby increased hand-rim diameters increased the metabolic demand, since Guo et al. (2006) revealed that larger diameter hand-rims increased the linear velocities of the hand, forearm and upper arm during propulsion. Woude et al. (1988b) also attributed the differences in physiological demand observed with hand-rim diameter to be due to the greater upper body segmental excursions demonstrated. Greater shoulder flexion/extension RoM, maximum abduction and elbow flexion values resulted from increasing hand-rim diameters (Woude et al., 1988b).

Veeger et al. (1992b) investigated the effects of different hand-rim velocities on the ergonomics of wheelchair performance at a constant simulated speed. This replicated the effects that different diameter hand-rims would have in relation to fixed wheel sizes by creating different gear ratios. A smaller hand-rim in relation to a fixed wheel size creates a reduced gear ratio. At a constant simulated wheelchair velocity a smaller hand-rim displays a lower hand-rim velocity. Veeger et al. (1992b) revealed that the lower gear ratios evoked from lower hand-rim velocities resulted in reduced $\dot{V}O_2$ and HR responses, thus reducing the physiological demand. This was in line with what has previously been associated with smaller hand-rim diameters (Woude et al., 1988b; Gayle et al., 1990; Costa et al., 2009). Although, similar theoretical principles were employed by Veeger et al. (1992b), their results cannot be directly compared to hand-rim diameter investigations, which have physically manipulated the gear ratio owing to differences in the standardisation of seat positions (Table 2.3). Seat height was kept constant between different gear ratio settings imposed by Veeger et al. (1992b), whereas previously mentioned, seat height was not standardised in any of the hand-rim diameter investigations (Woude et al., 1988b; Gayle et al., 1990; Costa et al., 2009). Therefore, large differences in the biomechanics of wheelchair propulsion would be expected between these studies. This proved to be the case, as Woude et al. (1988b) established no differences in temporal parameters between hand-rim conditions. Whereas Veeger et al. (1992b) revealed that PT increased and RT decreased in lower gear ratios.

Numerous differences have also been identified between previous biomechanical investigations into the effects of various hand-rim diameters. With regards to temporal parameters, Costa et al. (2009) revealed that the largest diameter hand-rim decreased the PT, yet increased push frequency, whereas Woude et al. (1988b) witnessed no temporal adaptations in response to different hand-rim diameters. In addition to this, Woude et al. (1988b) revealed no changes in the amount of work per cycle between hand-rim diameters (Woude et al., 1988b). Whereas Guo et al. (2006) identified that a significantly greater amount of work per cycle was performed with increasing hand-rim diameter. These discrepancies are likely to be due to the differences in methodological approaches employed and further emphasise the need for standardisation procedures to be developed. For instance, differences in the temporal parameters observed between Woude et al. (1988b) and Costa et al. (2009) could potentially be due to the fact that the latter investigation was purely a one participant case study. Therefore, the results obtained are unlikely to be representative of the population of wheelchair athletes. Alternatively, the differences observed between Woude et

al. (1988b) and Guo et al. (2006) relating to the amount of work per cycle may stem from the differences in ability and experience of the participants investigated, given that the range of hand-rim diameters were similar (Table 2.3).

Currently no research into the effects of differing hand-rim diameters on the performance of maximal effort tasks, specific to the wheelchair court sports, exists. However, both Woude et al. (1988b) and Costa et al. (2009) noted that the majority of participants could not maintain the highest test velocity in the largest hand-rim condition, suggesting that this type of configuration may be ineffective for maximal sprinting performance. Although wheel or hand-rim diameter was not directly investigated, Coutts (1990) revealed that WCB players accelerated their wheelchairs quicker over the first push, yet failed to reach such high peak velocities as their wheelchair racing counterparts. It was purported that the smaller wheel sizes and larger hand-rim diameters selected by the WCB players largely accounted for the observed results. Coutts (1990) stated that for a given force applied, a smaller wheel would reach a set velocity quicker due to the lower rotational inertia experienced. This was useful information, yet as referred to in previous sections, when attempting to maximise performance by establishing optimal wheelchair configurations more needs to be known about the exact effects on various areas of performance.

2.6.2.3 Hand-rim configuration

Areas of hand-rim configuration, including the diameter, shape, material and flexibility of the tube have also been investigated from an ergonomic perspective (Table 2.4). All of these are vital considerations when configuring a new wheelchair since the hand-rim is the immediate interface between the wheelchair and the user and the site of force transmission between the two (Gaines and La, 1986). Given the importance of optimising the wheelchair-user interface, adaptive equipment for the user's hands and wrists have also been afforded a reasonable amount of research attention (Table 2.4). Unlike previous areas of wheelchair configuration, a reduced emphasis is placed on the standardisation of other areas of a wheelchair's configuration when manipulating hand-rim configuration or adaptive equipment for the hands. This is due to the fact that altering either of these two variables has no incidental effects on vehicle mechanics or propulsion biomechanics. Alternatively, one confounding factor specific to these areas of the wheelchair-user combination is the number of variables manipulated during the same investigation. Previous investigations have often

varied more than one area of a hand-rim's configuration at once, preventing direct cause and effect relationships from being established. Although a large number of these studies have considered the comfort of the user by incorporating questionnaires to obtain subjective feedback about different interactions at the wheelchair-user interface. However, in association with other areas of wheelchair configuration, the key confounding factor when attempting to translate findings to the wheelchair court sports was the fact that the majority of studies all focused on daily life propulsion (Table 2.4).

Linden et al. (1996) investigated the effects of two different hand-rim tube diameters on various physiological and biomechanical parameters and revealed that the hand-rim with a larger tube diameter reduced $\dot{V}O_2$ and improved ME. However, the changes in physiological demand observed with different tube diameters could not be explained by any biomechanical adaptations, as temporal parameters and force application remained consistent between conditions. Woude et al. (2003) also examined the physiological and biomechanical responses to various hand-rim configurations, which differed in material, shape, and tube diameter. No significant differences were observed for any of the physiological or biomechanical parameters investigated between the hand-rim configurations. However, the subjective analyses revealed that cylindrical, rubber coated hand-rims were most favourable in terms of user acceptance/comfort. Although, user acceptance was the only measure to be affected by the manipulations, it was unclear whether the material, shape, tube diameter or a combination of each were what led to the favourable performance of the cylindrical, rubber coated hand-rim.

More recently, innovative hand-rim designs have been developed in an attempt to improve wheelchair performance. A variable compliance (Richter and Axelson, 2005; Richter et al., 2006) and a 'natural-fit' hand-rim (Koontz et al., 2006; Dieruf et al., 2008), which differ in terms of material, shape and flexibility in comparison to standard hand-rims have been investigated. A reduction in finger and wrist flexor activity has been established via an EMG analysis in a flexible, high friction hand-rim (Richter et al., 2006). This may have implications on efficiency, as Linden et al. (1996) proposed that the improvements they observed in ME may have been the result of a reduced amount of muscular activity being required to grip the hand-rim. User safety/health has also been considered between hand-rim configurations, with increases in compliance having been shown to increase the rate of rise of force development, thus potentially exacerbating the risk of injury (Richter and Axelson,

2005). Subjectively, comparisons between the performances of these hand-rims in relation to standard hand-rims have also been sought, with reductions in hand and wrist pain and improvements in the ease of propulsion reported in the 'natural-fit' hand-rims (Koontz et al., 2006; Dieruf et al., 2008). Unfortunately, all the aforementioned investigations have focused on improving the ergonomics of propulsion for the daily life user and the application of such findings to sporting populations is once again prevented.

It may also be of relevance for wheelchair sports men/women to consider adaptive equipment that can accompany the hand and wrists, because, as previously mentioned, this can also directly influence the interaction between the user and the wheelchair. Glove use may improve efficiency through an enhanced coupling with the hand-rims, but may also have implications for injury risk as abrasions to the hands are common during braking manoeuvres in particular (Richter et al., 2006). The high impact nature of wheelchair propulsion and the subsequent stress that users place through their hands and wrists led Burnham et al. (1994) to investigate whether glove use could minimise this risk. However, the use of a padded glove did not significantly reduce the conduction velocity in the median nerve, which was thought to strongly predispose towards carpal tunnel syndrome. Malone et al. (1998) also identified hyperextension of the wrist as a condition predisposing to injury and investigated whether the use of a wrist splint or glove/wrist splint combination influenced this movement during propulsion. It was revealed that both the splint and glove/splint conditions reduced the degree of hyperextension, demonstrating potentially favourable effects for user safety/health. Shimada et al. (1997) established that a fingerless glove they investigated increased the magnitude of tangentially directed force and torque around the wheel axle in comparison to trials without the use of a glove. These authors proposed that glove use may subsequently also improve the efficiency and performance of propulsion in addition to offering hand protection based on the greater forces observed that contribute directly towards forward motion. Malone et al. (1998) also considered mobility performance during their investigation and revealed that the use of both the wrist splint and glove/wrist splint combination did not negatively affect the maximal wheeling velocity that was attainable. This is particularly vital from an ergonomics perspective as one component, in this instance user safety/health, can be maximised by the manipulation to the wheelchair-user interface, however it has not occurred to the detriment of another, i.e. performance. The only aspects lacking from Malone et al. (1998) would be an assessment of efficiency and comfort to satisfy all the requirements of an ergonomics approach.

Table 2.4 – Summary of protocols adopted by previous ergonomic investigations into the role of hand-rim configurations and adaptive equipment acting directly at the wheelchair-user interface.

Reference	Wheelchair	Participants	Mode	Speeds ($m \cdot s^{-1}$)	Power Output	Manipulation	Measures
Hand-rim configuration:							
Gaines & La (1986)	ADL	WD – 29	Q	n/a	n/a	Shape	Subjective
Linden et al. (1996)	n/a	AB – 6	WERG [s]	1.11 – 1.67	18 – 25 W	Tube diameter, shape	Physiology, kinetics, temporal
Woude et al. (2003)	n/a	AB – 10	WERG [s] Q	1.11	0.15 – 0.40 W/kg	Material, tube diameter, shape	Physiology, kinetics, temporal subjective
Richter & Axelson (2005)	ADL	WD – 24	OG & MDT	0.22 – 0.94	Reported, not controlled	Flexibility	Physiology, kinetics, temporal, subjective
Koontz et al. (2006)	ADL	i) WD – 10 ii) WD – 46 iii) WD – 82	WERG [r] Q Q	0.9 & 1.8	Not controlled or measured	Material, shape	Kinetics, temporal, subjective
Richter et al. (2006)	ADL	WD – 24	MDT	Self selected	Not controlled or measured	Material, flexibility	Kinetics
Dieruf et al. (2008)	n/a	WD – 87	Q	n/a	n/a	Material, shape	Subjective
Hand protection:							
Shimada et al. (1997)	ADL	WD – 7	WERG [r]	1.8	Not controlled or measured	Glove	Kinetics
Burnham et al. (1994)	WCB	AB – 16 WD – 19	WERG [r]	Self selected	Not controlled or measured	Glove	Kinematics
Malone et al. (1998)	WCB	AB – 13	WERG [r]	Maximal effort & self selected	Not controlled or measured	Glove & wrist splint	Kinematics, performance

Key: **Wheelchair** – (WCB) Basketball, (ADL) Daily Life; **Participants** – (AB) Able-bodied, (WD) Wheelchair User; **Mode** – (WERG) Wheelchair Ergometer, [r] roller, [s] simulator, (MDT) Motor Driven Treadmill, (OG) Over-ground, (Q) Questionnaire.

2.7 Summary

Having reviewed the previous literature investigating the ergonomics of wheelchair configuration, it was clear that numerous procedures must be considered to further improve understanding and to make findings applicable to the wheelchair court sports. Previous ergonomic investigations into wheelchair configuration have adopted an extremely biased focus on aspects of daily life wheelchair propulsion. This has obviously limited translations that can be applied to a sporting context. However, there are still a number of procedures that can be implemented and precautions that must be taken that have been derived from these studies. Firstly, it was evident that making adjustments to one area of configuration can directly impact another. Therefore, in order to detect reliable cause and effect relationships, more stringent standardisation methods are an absolute necessity. In addition to this, it has also been made clear that certain manipulations can potentially improve one area of wheelchair ergonomics, but may do so at the expense of another. Subsequently, research needs to focus on identifying the points whereby performance can be maximised without negatively influencing other areas of performance or by placing the athlete at a greater risk of injury. Further key considerations are also needed in order to optimise the ergonomics of sports wheelchair performance, based on the review of literature. For instance, investigations must examine ranges of configurations that are representative of those used within the wheelchair court sports. To provide an ecologically valid examination of these configurations, wheelchair sportsmen/women also need to be investigated performing movements (at speeds) that are specific to these sports. All of these considerations demonstrate the complexity of the task and may partly explain why this area has received so little evidence based research in the past.

Chapter Three

A Qualitative Examination of Wheelchair Configuration for Optimal Mobility Performance in Wheelchair Sports

Mason BS, Porcellato L, Woude van der LHV & Goosey-Tolfrey VL (2010). A qualitative examination of wheelchair configuration for optimal mobility performance in wheelchair sports: a pilot study. *Journal of Rehabilitation Medicine*, 42(2), 141-149.

3.1 ABSTRACT

***Purpose*:** To examine how experienced wheelchair athletes perceived manipulations in wheelchair configuration to influence aspects of mobility performance. ***Methods*:** Nine elite wheelchair athletes from WCB (N = 3), WCR (N = 3) and WCT (N = 3) were interviewed using a semi-structured format. Interview transcripts were analysed using an Interpretative Phenomenological Analysis, whereby emergent themes with common connections were identified and clustered into three superordinate themes: (i) performance indicators, (ii) principal areas of wheelchair configuration, and (iii) supplementary areas of wheelchair configuration. ***Results*:** Participants identified stability as the most important contributor towards successful performance. It was also apparent that participants considered aspects of ball handling and match play factors ahead of mobility related factors when configuring a new sports wheelchair. Conformity was demonstrated amongst participants regarding the general performance effects resulting from adjustments to the majority of areas of wheelchair configuration. However, disparity existed between participants' perceptions regarding the effect of rear-wheel camber on linear mobility performance. ***Conclusions*:** Whilst experienced athletes displayed a good understanding of how modifying wheelchair configurations can affect wheelchair sports performance, methods for identifying optimal positions were often extremely vague. Therefore future quantitative research into specific areas of configuration is imperative to identify optimal settings. This should serve to inform athletes about the decisions they make when configuring a new sports wheelchair.

3.2 INTRODUCTION

Wheelchairs used within the court sports (WCB, WCR and WCT) have undergone major developments over recent years in terms of their design (LaMere and Labanowich, 1984; Yilla, 2004; Ardigo et al., 2005). In addition to the improved physical conditioning of wheelchair athletes, modifications to these chairs have been suggested to be largely responsible for improvements in performance (LaMere and Labanowich, 1984; Cooper, 1991; Yilla, 2004).

Extensive research has been conducted on the physiological and biomechanical responses to adjusting wheelchair configurations under conditions of daily life propulsion (section 2.6). However, the effects of manipulating selected areas of wheelchair configuration on dynamic aspects of mobility performance specific to the wheelchair court sports has been a topic of limited research. Subsequently, very little is known about the contribution that specific areas of wheelchair configuration has had towards these performance improvements. The need for future research into wheelchair configuration is of further importance when it is considered how large a phenomenon this is due to the number of areas of wheelchair configuration that can potentially contribute towards performance.

Wheelchair users have often been the subject of quantitative studies that have investigated wheelchair configuration. Yet to the author's knowledge, no previous studies have investigated the opinions of the wheelchair users to help further understanding on this area. Kratz et al. (1997) emphasised that adapted equipment was essential for athletes with impairment and demonstrated the value of gaining users' experiences in this process. Gauging the opinions of experienced athletes who have been through the wheelchair configuration process on numerous occasions may yield a valuable insight into the ergonomics of sports wheelchair propulsion and the optimisation of performance. Adopting a qualitative approach can often enable a better understanding to be developed about certain questions and phenomena due to the more holistic appraisal it can provide. This approach is particularly ideal in such circumstances where little is known about the area under investigation (Thomas and Nelson, 2001; Reid et al., 2005; Ryan et al., 2007).

The purpose of the current study was to investigate how elite wheelchair athletes perceived certain areas of wheelchair configuration to impact upon areas of sports performance, with a major focus on wheelchair mobility. It was anticipated that this study would challenge some of the existing literature and help to identify areas of wheelchair configuration that could benefit from future evidence based research. Consequently, the

information derived should help to inform athletes about the selections they make when configuring a new sports wheelchair to improve the ergonomics of wheelchair performance.

3.3 METHODS

3.3.1 Participants

Nine male wheelchair athletes (39 ± 5 years) participating in WCB (*N* = 3), WCR (*N* = 3) and WCT (*N* = 3) were interviewed as part of the current investigation after providing their written, informed consent. Purposive sampling was employed to recruit participants, to ensure that all individuals had a strong understanding of the phenomena in question (Ryan et al., 2007). To this extent, inclusion criteria required participants to have > 10 years playing experience at an international level, as these athletes were more likely to possess a profound practical understanding of wheelchair configurations for their sports. Since wheelchair configurations are affected by factors such as impairment and role on court within each sport (Hutzler et al., 1995; Yilla et al., 1998), it was also imperative that a range of athletes with differing classifications and impairment levels were included to obtain a more representative sample (Table 3.1). To account for this, participants were categorised as either high point (least impaired) or low point (most impaired) players. For WCB, participants with a classification of ≥ 3.0 were categorised as high point players (HP) and in WCR participants with a classification of ≥ 2.0 were classed as HP. Finally for WCT, participants who competed in the ‘open division’ were classed as HP and those who competed in the ‘quad division’ were classed as low point players (LP).

Table 3.1 – Details of participants and their current sports wheelchair configurations.

Sport	Age	Level / world ranking	Classification / Group		Wheel size (inches)	Camber (degrees)
WCB	36	International	4.0	HP	27	15
WCB	42	International	1.5	LP	25	18
WCB	44	International	1.0	LP	25	18
WCR	31	International	3.5	HP	25	18
WCR	34	International	2.5	HP	25	18
WCR	36	International	1.5	LP	24	18
WCT	41	Top 10	Amputee	HP	26	24
WCT	46	Top 25	T6 SCI	HP	25	20
WCT	46	Top 10	T4 SCI (C7/8 Hemiplegia)	LP	26	20

Key: WCB classified by International Wheelchair Basketball Federation (IWBF); WCR classified by International Wheelchair Rugby Federation (IWRF); SCI = Spinal Cord Injury.

3.3.2 Procedures

A semi-structured interview format was adopted for the current investigation due to the greater flexibility it allows for probing specific areas in more detail than structured interviews. It also places a reduced emphasis on previous interviewing experience required in relation to unstructured interviews (Robson, 2002; Stanton et al., 2005; Tenenbaum and Driscoll, 2005). Following a successful pilot interview to test the validity of the questioning, all participants were interviewed at a time and location that was convenient for them. Anonymity and confidentiality was ensured to participants, who also possessed the right to terminate the interview at any stage without further questioning. All interviews were recorded using a Sony ICD-SX57 digital voice recorder (San Diego, CA).

3.3.3 Data analysis

Dialogue from the interview recordings was transcribed into word processing format. Transcripts were then analysed using an Interpretative Phenomenological Analysis (IPA). This method of analysis was selected to accommodate the small sample size and because of the subsequent detail that IPA can construct about a phenomenon (Hycner, 2000; Smith and Osborn, 2003; Reid et al., 2005). The fact that this analysis was predominantly inductive was

also a contributing factor to the use of IPA, as no predetermined framework had been created prior to interviewing (Fade, 2004).

All transcripts were read through several times in order to get an overall sense of the content (Fade, 2004). Dialogue then underwent a coding process, whereby interpretations were made on any themes present in the interviews and coded with headings and annotations (Hycner, 2000; Robson, 2002). Member feedback was then sought from two randomly selected participants in order to improve validity (Hycner, 2000; Stanton et al., 2005). This involved a copy of the transcripts, complete with the interviewer's codes, being sent to the participants. This ensured that interpretations gave an accurate reflection of what had been said and gave participants the opportunity to alter or add any other information (Amis, 2005). Two further investigators were also involved in the analysis process. One investigator possessed a vast amount of qualitative research experience, whilst the other investigator had a substantial knowledge of wheelchair sports. All coded transcripts and interpretations were verified by these investigators to further enhance validity and to guard against researcher bias (Burnard, 1991).

The initial list of themes that emerged from the interviews were then further analysed and clustered into a smaller number of themes with common connections (Smith and Osborn, 2003). Each cluster was then titled with a superordinate theme based on the nature and content of the subordinate themes present.

3.4 RESULTS

Data from the current investigation was grouped into a total of three superordinate themes: (i) performance indicators, (ii) 'principal' areas of wheelchair configuration, and (iii) 'supplementary' areas of wheelchair configuration. A series of quotes from the transcripts were included to support any interpretations and were documented by the sport and classification level of the participants.

3.4.1 Performance indicators

Prior to discussing how areas of wheelchair configuration were perceived to affect performance, it was vital to establish which aspects of mobility performance were important, specific to the wheelchair court sports. Participants across all three sports repeatedly identified four important areas that they felt were paramount to successful sports performance:

3.4.1.1 Stability

When attempting to establish the most important areas of mobility performance within each sport, the majority of participants acknowledged the need for stability in their wheelchairs as the principal element. It was frequently stated that without stability, all of the other areas of performance could become negatively affected.

3.4.1.2 Initial acceleration

In terms of actual wheelchair mobility, participants of all classifications from WCB and WCR collectively identified that acceleration over the first couple of pushes was the most important facet of performance in their sports:

> *"…..you are going from a starting position a lot because you are getting stopped, so you have got to be able to start again quickly….." (WCB - LP)*

Low point players within WCB and WCR seemed to place further emphasis on the need for quick initial acceleration in order for them to gain dominant positions on the court and to compete with HP:

> *"…..you are not particularly as quick as high point players over longer distances, but if you can remain as competitive as possible over the first two pushes it gives you a defensive advantage, as people cannot get round as easily." (WCB - LP)*

Although initial acceleration was also revealed to be important in WCT, it seemed to be slightly less vital from a standstill for these individuals, as it was for WCB and WCR players. Alternatively, it was suggested that acceleration was more important over the first two pushes from a rolling start as a reaction to an opponents shot:

> *"…..it is trying to keep away from that big first push by keeping the chair moving….." (WCT - HP)*

3.4.1.3 Manoeuvrability

Participants from WCT alternatively appeared to value turning ability as the most important area of mobility performance in their sport, regardless of impairment, due to the frequency with which this movement is performed. However, there did seem to be a slight discrepancy surrounding the need for manoeuvrability amongst classification groups within WCB and WCR. HP seemed to rate manoeuvrability far higher than LP due to their differing roles on court:

"…..as a low pointer, twisting and turning is not that important, because it is rare that I am going to have the ball and that I am going to have break press." (WCR - LP)

3.4.1.4 Sprinting

The ability to reach high top-end speeds was a desirable aspect of performance for all sports, but one that was not viewed as such a high priority. It was suggested that the distances covered in a straight line are often not sufficient to reach top speeds. Linear propulsion was also thought to be limited, particularly in WCB and WCR due to the multidirectional nature of the movements involved:

"…..once you are going in your chair you are never really going in a straight line, you know what I mean, there are other people to go around….." (WCB - LP)

3.4.2 Principal areas of wheelchair configuration

Participants' responses to the 'principal' areas of wheelchair configuration were clustered into two higher order themes that related to 'seating' and 'main wheels' (Table 3.2). These principal areas were labelled and clustered as such, based on the fact that they were areas that have received previous quantitative research attention from an ergonomic perspective, albeit with a large focus on daily life propulsion conditions.

Table 3.2 – List of subordinate themes and clusters surrounding the 'principal' areas of wheelchair configuration.

Higher Order Themes	1st Order Themes	Subordinate Themes
Seating	Height	- Game related benefits of sitting high - Relationship with stability - Influence on manoeuvrability - Association to propulsion technique - Methods of optimisation
	Fore-aft position	- Association with manoeuvrability - Alterations to sprinting capabilities - Relevance to propulsion technique - Game related drawbacks of posterior seat positions
	Bucket	- Improvements in stability - Limitations associated with mobility - Consequences for match play activities
	Backrest	- Influence of height on stability - Heights association with mobility - Relevance of inclination angle - Role of tension in stability provision - Effect of tension on propulsion
Main Wheels	Size	- Consequences for initial acceleration - Impact on turning capabilities - Relationship with sprinting performance - Association with pushing economy - Dependence on physical strength and impairment - Relationship to sitting height
	Camber	- Relationship with manoeuvrability performance - Influence on straight line performance - Impact of wheelchair maintenance - Contribution towards stability - Reliance on sitting position - Dictated by sport and individual roles on court
	Hand - rims	- Distance to wheels and relationship to technique - Role of anthropometrics in proximity selection - Proximity settings relevance to match play - Influence of material on pushing economy - Associations between materials and grip - Importance of material selection on match play - Contribution of tube diameter towards comfort - Ratio to wheel size as a gear mechanism

3.4.2.1 Seat height

Participants' initial responses to the issue of sitting height predominantly centred on match play related factors, as opposed to wheelchair mobility factors. For example, players from all three sports commented that sitting in a higher position was ideal from a ball handling perspective in WCB and WCR and for a better view of the court in WCT. Despite being viewed as the ideal position, players acknowledged its potentially negative impact on stability and that the ability to sit high was subsequently governed by each player's specific level of impairment. In addition to improving stability, sitting lower was also thought to benefit aspects of manoeuvrability performance:

> *"…..lower point players are far better off to be sat lower down in order to be more manoeuvrable and illusive, because they just do not have the function to sit high….." (WCR - HP)*

Another area, relating to wheelchair mobility that participants deemed important when selecting seat height was determined through how accessible the wheels are. A number of participants actually identified this as being the most important consideration of wheelchair configuration. It was thought that by having more of the wheel available to push on, quicker acceleration and sprinting was achievable due to the longer pushing stroke it permitted.

> *"…..if you have not got enough wheel to push on and the pushing stroke is not long enough, then you spend too much time recovering for the next stroke than more importantly pushing the wheel." (WCB – LP)*

Methods for determining how high to sit and how much 'enough' of the wheel actually was, involved a number of different subjective approaches:

> *"…..it is relative to the hub of the wheel, your arm to the hub. I feel comfortable pushing when my hand can comfortably hang down and reach the hub." (WCT - LP)*
>
> *"…..when I am sat in my chair and I fold my arms in a relaxed position, my elbows just touch the top of the wheel….." (WCB - HP)*

3.4.2.2 Fore-aft seat position

The fore-aft positioning of the seat was frequently referred to as the horizontal positioning of the camber bar and was viewed by some participants as the most important area when setting up a new sports wheelchair. Participants felt strongly that the positioning of the camber bar influences the manoeuvrability of a wheelchair:

"The further forward the camber bar is, you are more manoeuvrable…..The further back the camber bar is, a lot of the weight of the chair is behind you, which makes it harder to turn." (WCR - LP)

Alternatively, contrasting views emerged regarding the influence of the camber bars positioning on straight line performance, with a combination of negative and positive remarks associated with having the camber bar positioned further forward:

"…..you have to find that point where you are not on the back wheel of the chair otherwise you are almost wheelying all the time and you have to get your body weight forwards to get the power through to the wheels." (WCB - LP)

This was supported to some extent by a participant from WCT, who warned against the negative consequences of spending too much time tipping on to your back wheels:

"If you tip in the chair you are going to have a wheel off the ground, which means you are going to lose speed…..you need your wheels on the ground to give you grip." (WCT - LP)

Conversely, some participants were of the opinion that you may be able to accelerate and sprint faster in a straight line if your camber bar is in a more anterior position. It was suggested that having the wheels slightly in front of you potentially allows a greater portion of the wheel to push on. Having enough of the wheel to push on was again an area that generated different opinions between players of different impairment levels. A few HP participants highlighted the desire to have a push that lasted from between "12 o'clock to 3 o'clock" (0° to 90°) on the wheel in order to drive the wheels down effectively (Figure 3.1). Therefore, they advocated a camber bar that was positioned slightly further back, so they could sit directly above TDC. However, a participant from WCR actually commented that this may not be a suitable approach for those with a more severe level of impairment who often lack triceps function:

"…..as we cannot all really extend our arms properly there is no point trying to sit on top of the wheel and trying to push down…..I think some should sit behind the wheel and pull and try and use that more." (WCR - LP)

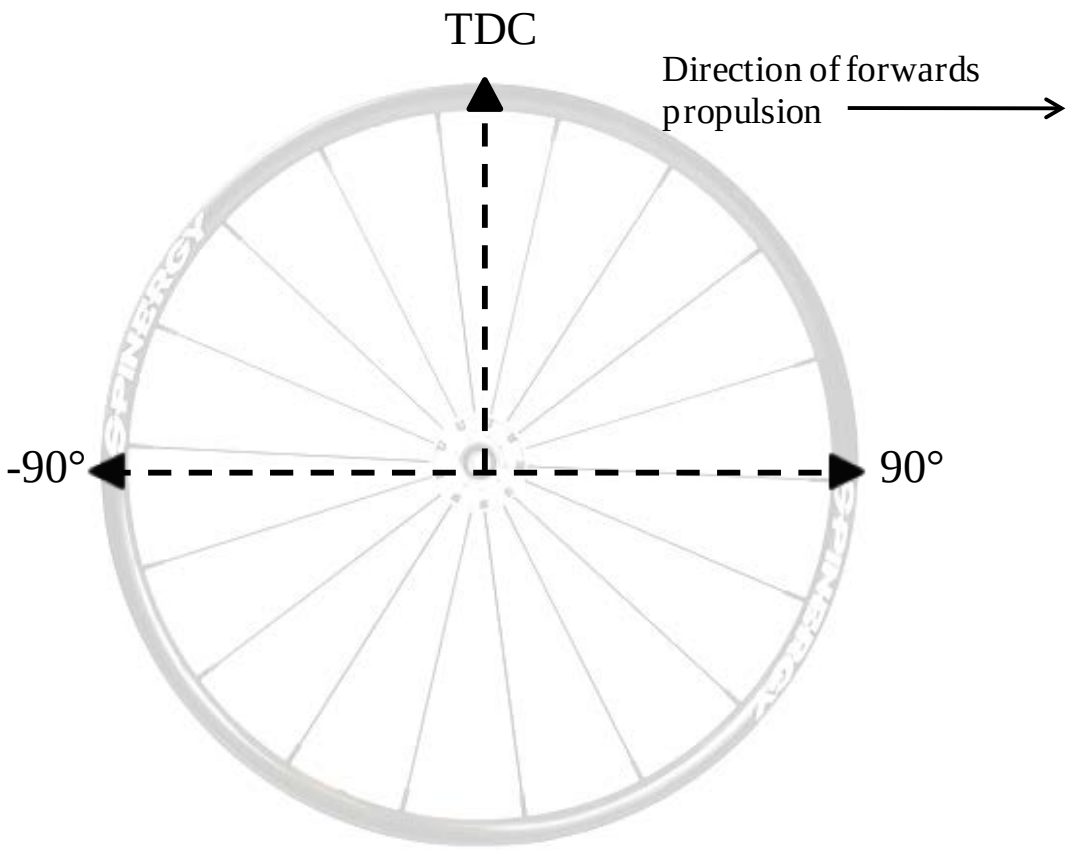

Figure 3.1 – Anatomy of a wheel accompanied by the terminology used to explain angular displacements and temporal parameters during wheelchair propulsion. TDC = top dead centre.

Although LP seemed to favour a more posterior seating position by having the camber bar positioned more towards the front of the wheelchair, some potential drawbacks during match play were associated with this, as demonstrated in WCT:

"….you want to be hitting the ball out in front of you…..whereas if you are laid back in your chair, you are almost hitting the ball back from behind you….." (WCT - LP)

3.4.2.3 Seat bucket

Having the front of the seat higher up than the back of the seat creates what is known as a 'bucket'. Low point players suggested that this bucket was particularly useful for them as it provided them with a greater degree of stability in their wheelchairs and could also be of value to game related skills in WCR:

"…..the way we carry the ball in rugby is that we put it on our laps and you can almost use your knees to hide the ball…..When you have flat legs, the ball is almost sat up high and makes it easier for opponents to steal." (WCR - HP)

High point players agreed that having a bucket improved stability, yet felt that it could be more of a hindrance to their propulsion technique, mobility and game related activities:

"With too much bucket you start to restrict your pushing because your knees are above your hips and you are pushing almost like reclined, it's hard to explain really, but basically you cannot utilise your trunk as much." (WCT - HP)

3.4.2.4 Seat backrest

Similarly to the bucket of the seat, the configuration of the backrest was thought to play a major role in a player's stability. The height of the backrest appeared to be influential in this, with a higher backrest perceived to provide a higher degree of stability, which appeared to be of particular value to LP:

"If it is too low, because we don't all have the stomach function, you end up falling out of the back and have to grab the wheels to pull you back up….." (WCR - LP)

In accordance with what was previously mentioned with respect to the seat bucket, it was unsurprising to discover that HP favoured as lower backrest as possible, so that it did not restrict their movements.

The inclination angle of the backrest was also commented on by participants. Having an upright backrest was suggested to be beneficial by WCT players, as it pushed them further forward, which was said to assist ball striking. However, LP from WCB and WCR felt that being thrown too far forward in their seat negatively affected their stability. Subsequently, these players mentioned how the tension of the backrest can also be an important factor to remedy this:

"…..to have the backrest upholstery quite loose, so you sit back into a kind of pocket and then get support from the uprights of the backrests is ideal….." (WCB - LP)

Alternatively, HP from WCB and WCT seemed to favour a tighter backrest to keep them in a better position to receive or hit a ball, respectively. A looser backrest was thought to have negative consequences on their mobility too, as they felt like they were losing "thrust" and "energy" during propulsion.

3.4.2.5 Wheel size

A number of participants strongly believed that smaller diameter main wheels contributed towards greater initial acceleration and may enable better manoeuvrability:

"…..he went from having 24 inch wheels up to 27 inch wheels…..but he could not get the chair going that quickly and his turning speed became pretty awful….." (WCT - HP)

These smaller wheels may be more appropriate for players with a higher degree of impairment due to the subsequent magnitude of force that can be required to accelerate a chair with larger wheels. It was mentioned by LP that they often do not possess the physical power to get the chair moving from a standstill. Despite this, larger diameter wheels were proposed to be advantageous for other areas of performance. Sprinting over longer distances was thought to be more effective using larger wheels, as was the economy of propulsion:

"…..with a big wheel you would have to put less effort in once you are going because of the fewer rotations needed….." (WCT - HP)

Another area of wheel size given considerable attention was its relationship to seat height. As previously mentioned, correct seat positioning is vital to allow sufficient access to the wheels. One reason for selecting bigger wheels was to allow players to sit quite high, but to still be able to access enough of their wheels:

"…..if you have got bigger wheels when you are sat higher you are in a better position…...but if you keep the height of the chair and go to a smaller wheel, then you have got far less wheel to push on so you may be slower." (WCT - LP)

3.4.2.6 Rear-wheel camber

Rear-wheel camber was commonly thought to have a favourable influence on manoeuvrability by all participants, regardless of impairment. However, the effect of camber on areas of straight line performance seemed less conclusive, with conflicting views expressed. Some participants felt that increased degrees of camber had negative effects on straight line performance:

"…..from increasing camber, my speed went down and it seemed like I was sucking the floor in as I was going along….." (WCR - LP)

Other participants did not believe that such a negative impact existed, especially if the wheels were well maintained, as one participant emphasised:

"I don't actually believe there is a great deal in your acceleration (with increased camber) if your wheels are true and toed properly." (WCR - HP)

Rear-wheel camber was also thought to directly influence the lateral stability of the wheelchair-user combination:

"You need camber to give you the stability…..and bigger camber gives you a wider base, making you less likely to tip in your chair." (WCT - LP)

However, LP warned against selecting excessively cambered wheels to avoid compromising the stability of the user during turning:

"…..in a severely cambered chair their (low pointers) head would be going one way and their chair would be going the other and it would take a few seconds to recover and that might be too late." (WCT - HP)

The vertical positioning of the seat was also thought to strongly influence camber selection. As previously mentioned with regards to seat height, the lower you sit, the more stability you have. As a result, some participants commented that players who sit lower may not require as much camber to aid with stability and alternatively players who sit higher may rely on greater camber. Camber selection also seemed dependent upon the sport and role of the participant. For example, LP from WCB and WCR tended to favour slightly higher degrees of camber to HP, due to their defensive responsibilities:

"…..lower pointers want to be as long and wide as possible so they can take up a lot of court space to make it a long way to travel round them." (WCR - HP)

Alternatively, HP frequently opt for lower degrees of camber to assist them with their more offensive roles on court:

"…..I think that for hitting those smaller gaps, less (camber) is definitely an advantage….." (WCR - HP)

3.4.2.7 Hand-rims

Numerous areas of hand-rim configuration were thought to impact upon mobility performance in wheelchair sports, including the proximity of the rims in relation to the wheels. Participants who pushed with a combination of both the hand-rim and tyre seemed to favour having the rims set in closer to the tyre, as did those with smaller hands. However, players who felt more comfortable pushing solely on the hand-rims tended to favour a slightly wider hand-rim setting. Participants also commented on how the proximity of the rim to the tyre can influence game related activities and selection can be influenced by players' roles, particularly in WCR:

"…..if they (hand-rims) were further out it would give you better picking potential for stopping other players….." (WCR - LP)

"…..from an offensive point of view, obviously the less rim that is visible the better." (WCR - HP)

Material was also cited as a vital area of hand-rim configuration that contributed significantly to performance. Like with the proximity of the hand-rim, the material also seemed to have a relevance to match play related activities in WCR. Hand-rims of different materials appeared to be suited to players of differing roles:

"From a game perspective, if you are alongside a high pointer, all you have to do is touch their rims and you know you are going to pick them…..if I had metal rims, I would just slide off, but as soon as they touch that red (rubber) rim, they turn in and then you get that pick." (WCR - LP)

Hand-rim material appeared to be of greater significance for individuals with a more severe level of impairment. LP from WCR felt that a rubber or foam coated rim used in conjunction with gloves served to improve their movement capabilities:

"…..I have not got as much strength or as much grip as the 2's (classification), so my stopping capabilities were not too good…..whereas with the red (rubber) rims I have definitely got more purchase….." (WCR - LP)

These players emphasised the importance of the interaction between the user and the hand-rim. Two participants from WCR highlighted the vital role that glove type can have on maximising the amount of grip that can be obtained at the wheelchair-user interface for these players, in addition to protecting the hands. However, the gloves they used to achieve this were not task specific and were frequently modified by players until they felt that the optimal amount of friction could be generated between their gloves and the material of their hand-rims.

The diameter of the hand-rim in relation to the diameter of the wheel was also highlighted by two participants to contribute towards performance. It was commented that the hand-rims can act as a gear ratio for users by having different diameters in relation to the size of the wheels. One participant mentioned that this was an area that players were starting to experiment with in WCT, although neither could offer any insight into the potential effects that these changes had on performance.

3.4.3 Supplementary areas of wheelchair configuration

Areas of wheelchair configuration that have been relatively unexplored by previous research were clustered into the superordinate theme 'supplementary' areas of wheelchair configuration. These included areas of wheelchair configuration that had previously not been directly manipulated and investigated from a physiological, biomechanical or sports performance perspective. These areas were frequently proposed, unprompted, by the participants as areas which can influence performance and were grouped into a total of six higher order themes (Table 3.3).

Table 3.3 - List of higher order and subordinate themes concerning the 'supplementary' areas of wheelchair configuration.

Higher Order Themes	Subordinate Themes
Frames	- Relationship between weight of material and mobility - Importance on strength of material
Chair length	- Association to specific sports and roles on court - Impact on manoeuvrability - Incidental effects on stability
Footrest	- Positioning in the sagittal plane for manoeuvrability - Positioning of the feet for enhanced stability - Use of foot positioning for assisted propulsion
Strapping	- Additional stability - Positive influence on manoeuvrability - Potential hindrances on mobility
Castor wheels	- Anti-tip wheels involvement towards improved manoeuvrability - Positioning of anti-tip for match play and propulsion - Significance of number of front castor wheels
Tyres	- Effect of pressure on straight line speed and manoeuvrability - Relationship of tyre pressure with playing surface

3.4.3.1 Frames

The main consideration given to the frames of wheelchairs, seemed to centre on the weight of the material used. All participants favoured the lightest wheelchair possible, due to the positive impact upon mobility and the efficiency of propulsion. Adjustability was another consideration that was given to the frames. Participants frequently referred to the use of an 'adjustable' or a 'fixed' frame and commented on the advantages and disadvantages of both. A proposed advantage of an adjustable frame was the flexibility in terms of configurations it allows:

"…..I would recommend that nobody really gets a fixed frame chair until you know exactly what you want and that can take many years. So I still think that having that bit of adjustability, especially as you are growing up, is important." (WCT - HP)

However, participants interviewed in the current investigation were experienced, elite athletes and subsequently favoured fixed frame chairs, due to the limitations experienced with adjustable chairs. Adjustable wheelchairs were described as being less rigid than fixed frame chairs, which was said to impact on the chairs durability and longevity:

"…..there were too many moving parts (in adjustable chairs), so after a battering there were too many things that could go wrong with it because it was not as fixed or as stable as it should be." (WCR - LP)

3.4.3.2 Chair length

The length of a sports wheelchair was also considered by some participants as a contributor towards mobility performance, particularly manoeuvrability:

"…..by having a shorter wheelbase, it can be more manoeuvrable because the wheels are closer together….." (WCR - HP)

However, a HP from WCR was of the opinion that stability could be compromised if the length of the chair was too short, particularly if this was combined with a high sitting position.

3.4.3.3 Foot-rest position

Positioning of the feet seemed to be an area that was worthy of consideration for participants when setting up a new sports wheelchair. This was thought to influence both the manoeuvrability and stability of a performer respectively:

"If you have your feet in front of your knees, your chair is not going to turn as well as if you had them further back, because your weight is more forward." (WCT - HP)

"…..if you have got your feet tucked right underneath you, your tendency is that when you lean forward, you feel like you will almost fall on your nose." (WCB - LP)

Having your feet too far back was also considered to have implications on aspects of propulsion as well:

"If your feet are tucked right behind you, you cannot put any force through them. So if you have got your legs at 90 degrees for example, you can utilise that and you can get a much stronger push because you can use them to help you come back up." (WCT - LP)

3.4.3.4 Strapping

Although, not directly related to wheelchair configuration, all participants alluded to the vital contribution that strapping has had on aspects of performance. The main benefit was clearly the positive effect that it had on players' stability, as one participant from WCB commented how it can benefit players of all classifications:

> *"…..you have seen at the lower end (of the classification system), far more function from the lower point players because you are kind of using another device as a substitute for missing muscle groups…..it gives you this core stability by fixing yourself down to the chair. Then at the high end of the classification, you are seeing tilting moves of the chair from the 4, 4.5 point players simply because they are fixed to it and are able to make their chair move as if it is part of their body….." (WCB - LP)*

Moreover, strapping may allow the chair to be configured in a way that participants' impairment may not have previously allowed:

> *"…..there are certain shots when you are reaching out wide, where you need to have the confidence that if you reach out, you are not going to fall out, so it (strapping) is absolutely essential, especially sitting at the height that I do now." (WCT - HP)*

However, there appeared to be a risk that some players can strive for so much stability through strapping, that they do so at the expense of their mobility:

> *"It can be negative for some people if you try and wear too much to become very stable, but are more or less just becoming mummified in their chair and can't really move as effectively." (WCB - LP)*

3.4.3.5 Castor wheels

The smaller wheels at the front and rear of sports wheelchairs, referred to as castor wheels, were proposed by a large number of participants to have contributed towards improved performance. Front castors wheels have varied between one and two wheeled designs in WCT. One participant perceived that having one castor wheel at the front allowed for greater straight line speed through a reduced feeling of drag. However, the two wheeled design was actually favoured by this participant due to the greater stability that was permitted during turning at high speeds.

The rear 'anti-tip' castor wheel was given the majority of attention by participants and was described as being one of the major developments in wheelchair sports, due to its positive relationship with stability and manoeuvrability performance:

> *"…..they have made a massive difference because you can have your main wheels further forward, which gives you more turning capabilities in your chair." (WCB - LP)*

Factors relating to the anti-tip wheel, such as the distance it is set off the ground and how far forwards or backwards it is positioned were also discussed:

> *"…..if you fix your anti-tip more forward, it makes you more tippy and sometimes this is useful when you are shooting because you have that bit of give to allow you to lean back that bit further…..The problem if it is too low to the ground is that you get too much of a drag on it." (WCB - LP)*

3.4.3.6 Tyres

The tyre pressure of the main wheels was another area that two participants gave consideration to in relation to optimising their mobility performance. One participant commented that the higher the pressure, the less drag and resistance they experienced during propulsion. However, too greater pressure was suggested by the same participant to lead to reduced grip during turning, as a result of the smaller surface area of the wheel in contact with the surface. Therefore, this participant indicated that he had decided on his optimal tyre pressure through trial and error. Interestingly, a participant from WCT mentioned how tyre pressures need to be adjusted to suit the hardness of the surface they are competing on. Obviously, WCT players compete on a variety of different surfaces, unlike WCB and WCR players. This participant felt that it was beneficial to have a lower tyre pressure than normal on softer playing surfaces and a higher pressure on harder surfaces.

3.5 DISCUSSION

It was evident from the current investigation that participants gave a great deal of consideration to game related activities when configuring a new sports wheelchair. Being in a position to handle the ball in WCB and WCR and hit the ball effectively in WCT was an extremely high priority for players when selecting areas of their wheelchair configuration. Mobility performance, the main focus of previous wheelchair configuration studies, as well as the current investigation, was given a fair amount of consideration by participants too, although to a slightly lesser extent.

3.5.1 Performance indicators

The ability to accelerate, sprint, brake and turn have previously been identified as the key determinants of mobility performance in wheelchair court sports (Vanlandewijck et al., 2001). However, the current investigation revealed that stability of the user was viewed as the most important performance indicator for participants from these sports, as it enabled the performance of all other movements to be improved. In accordance with Vanlandewijck et al. (2001), the ability to accelerate from a standstill was still viewed by the current participants as a vital determinant of successful mobility performance. Yet, it appeared as if this element of performance was more crucial to WCB and WCR than WCT. Alternatively, participants from WCT expressed a desire to keep their wheelchairs moving at all times to avoid having to accelerate from a standstill, as this requires greater force to overcome inertia and to get the chair moving. The ability to accelerate was still thought to be important in WCT, but more so from a rolling start than from a standstill. An explanation for this could be that in WCB and WCR players have obstacles, such as other team-mates and opponents on the same court. These obstacles can directly influence a player's movement and was said to cause them to frequently stop and start again, which is why acceleration from a standstill was so vital. With WCT, no obstacles exist and players have a greater control over their movements, which could explain why turning ability was seen as slightly more important to these participants. Although, WCB, WCR and WCT have often been discussed collectively in terms of their movements, there appeared to be subtle differences between the importance placed on each performance indicator. This seemed to stem from the team nature of WCB and WCR and the individual nature of WCT. Despite subtle differences, the ability to sprint was not prioritised as highly as other movements by participants from all three sports.

3.5.2 Principal areas of wheelchair configuration

Determining how important participants' perceived certain areas of performance was a valuable step when attempting to explore athletes' experiences of wheelchair configuration. This was due to the fact that it became evident that making even minor adjustments significantly influenced these performance indicators.

3.5.2.1 Seating

A major area for concern that emerged from the current investigation would appear to be how participants determine their optimal configurations. A fine line appeared to exist between adjusting one area to the benefit of one aspect of performance, without it being

detrimental to another area. This was best demonstrated when exploring the effects of sitting position in relation to wheel size, as players frequently commented on how important it was to access 'enough' of their wheels during propulsion. Some participants felt that having more of the wheel to push on allowed for quicker propulsion due to the longer stroke it permitted. However, explanations concerning 'how much was enough' and when 'more' became 'too much' seemed to be slightly ambiguous. In addition to this, participants' methods for determining their seat height in order to access enough of the wheels were extremely subjective. Participants commented on methods whereby their hands should comfortably be able to reach the hub of the wheel when sat in a relaxed position. One concern about this approach is that it does not take the size of the wheel into consideration. For example, if the height of the seat was maintained, but the wheel size was reduced, the part of the wheel that is used for propulsion would be further away, subsequently altering the temporal and kinematic parameters of propulsion. These trial and error methods that players are employing may seem suitable for the experienced participants interviewed for the current investigation. However, for younger, less experienced players who have not been through the process of configuring sports wheelchairs before, this could be a particularly demanding task.

Woude et al. (1989b) identified a more standardised method for adjusting seat height, whereby seat height was referred to as the angle of elbow extension induced, when the hands were placed on TDC of the wheel. It was subsequently revealed that increasing seat height had a significant impact on the amount of the wheel that could be accessed. Mean PA was significantly greater at a seat height inducing 100° of elbow extension compared to a seat height that induced 160° of elbow extension. Further research along these lines is required in order to establish optimal seat heights for sports wheelchair performance, since recommendations can then be made relative to the specific anthropometrics of individuals.

Consideration was given to sitting position in relation to the part of the wheel they push on when configuring a new sports wheelchair, as it was apparent that a number of participants used increments in wheel size to accommodate any increases to their sitting height. It was evident that participants felt sitting high was advantageous to performance in all sports if they had the trunk function to remain stable and access enough of their wheels. Several participants mentioned that they would accompany this change with an increase in wheel size to make sure the wheels were not too far away from them. However, it was clear from comments participants made relating to wheel size that this can have consequences on other areas of performance too. For instance, going to larger wheels was seen as advantageous for sprinting and economy of propulsion during competition, yet it was also thought to

negatively influence acceleration from a standstill. Therefore, although it seemed clear that participants can make adjustments to other areas of a wheelchair in order to compensate for some potential drawbacks of the original adjustment, other areas of performance can be directly influenced as a result. Consideration to the movements that were most important to each individual, given their specific impairment level and role on court seemed to be needed when configuring a sports wheelchair, as previously mentioned by Yilla et al. (1998).

A similar problem appeared to exist for establishing the optimal position of the seat in the fore-aft direction in order to access enough of the wheels. Only one participant provided a remotely quantifiable method as to how they determined their optimal seat position in the fore-aft direction. This participant felt that as long as the seat was in a position whereby he could get his shoulders forward to a point directly above the hub of the wheel, then it should be adequate for allowing a sufficient stroke length. Previous quantitative methods have utilised percentiles of arm length as an objective measure for adjusting the seat in relation to the hub of the wheel in the fore-aft direction (Hughes et al., 1992; Wei et al., 2003). Further research is required to assess this, as both the studies of Hughes et al. (1992) and Wei et al. (2003) were conducted at sub-maximal speeds and subsequently their findings may have little relevance to the more dynamic mobility involved in wheelchair court sports. However, the standardisation methods employed for fore-aft seat positions are required in order to optimise seat positions specific to the anthropometrics and ability level of the users.

The current study also identified some areas of wheelchair configuration that revealed conflicting beliefs from participants as to their impact on mobility performance. Firstly, looking at the fore-aft positioning of the seat, it was clear that participants felt that the further forward the camber was positioned, the more manoeuvrable the wheelchair became. However, participants' perceptions concerning the effect that this modification had on linear performance was less obvious. Some noted that by having the camber bar further forward (posterior sitting position) more of the wheel was accessible to allow for a longer pushing stroke, which was thought to allow for greater acceleration and sprinting capabilities. However, Woude et al. (2001) suggested that having the centre of mass of the wheelchair-user combination positioned over the axle of the main wheels in the fore-aft direction would reduce the rolling friction experienced. This implied that this position would in fact have the most favourable effect on mobility performance. Other participants identified additional drawbacks from sitting too far back, as they felt they ended up rocking back and forth on to their anti-tip castor wheels all the time. This was thought to not only be a waste of energy, but

to also reduce manoeuvrability, as too much tipping can cause a loss of grip with the playing surface.

3.5.2.2 Wheels

Another area of wheelchair configuration that was clearly deemed to have a positive influence on manoeuvrability, but again caused uncertainty amongst participants with respect to its influence on linear performance, was the degree of rear-wheel camber. Increasing the degree of camber was unanimously linked to improved turning performance by all participants. This appeared to support the findings of Faupin et al. (2002) who discovered that the speed of turning increased with increasing camber during field testing with WCB players. However, it must be acknowledged that the camber angles investigated by Faupin et al. (2002) only ranged between 9° to 15°, whereas the camber angles used by athletes from wheelchair court sports were slightly more extreme (Table 3.1).

There was a distinct lack of congruence amongst participants about the impact rear-wheel camber selection had on aspects of linear performance. Some believed that an increase in camber was associated with reduced straight line performance, with an increased feeling of drag and resistance cited under these conditions. However, other participants felt that increasing camber had no or little effect on linear performance.

This disparity also exists within the scientific literature. Veeger et al. (1989b) identified small but significant decreases in rolling resistance when increasing rear-wheel camber from 0° to 9°, whereas Buckley and Bhambhani (1998) believed that the influence of camber on rolling resistance was negligible. Differences in the standardisation methods imposed were evident between these studies, particularly relating to toe-in toe-out. One participant from the current investigation felt that camber had negligible effects on straight line performance, as long as the main wheels were toed correctly. This appeared to correspond with the findings Faupin et al. (2004) who controlled toe-in toe-out across camber conditions. Faupin et al. (2004) subsequently revealed that rolling resistance increased and mean velocities decreased significantly when performing an 8-second sprint in camber angles which increased from 9° to 15°. Therefore, further research is clearly required to investigate the effects of rear-wheel camber on linear wheelchair propulsion given the discrepancies evident amongst the subjective beliefs of experienced wheelchair sportsmen and in previous quantitative literature. Yet based on the current findings it was interpreted that such research must stringently control the effects of toe-in toe-out between camber conditions.

An advantage of the current study was its role in identifying novel areas of wheelchair design that have been integrated into some sports, which may benefit from future quantitative research. An example of this was hand-rim diameter in relation to the diameter of the main wheel (gear ratio). Traditionally, hand-rims for the wheelchair court sports are one inch smaller in diameter than the wheel size. However, it emerged that varying gear ratio was an area of wheelchair configuration that was being manipulated by some competitors in WCT. Physiologically, favourable effects of reducing gear ratio have been identified during sub-maximal propulsion in racing wheelchairs (Woude et al., 1988b; Gayle et al., 1990). However, it remains to be seen whether a similar trend would exist within court sport wheelchairs and what the effects of gear ratio would be on maximal effort mobility performance.

The current investigation also demonstrated how important coupling between the hand and hand-rims was, especially for the more severely impaired individuals. This has previously been highlighted by Gaines and La (1986) and Woude et al. (2003) who described the value of coupling between the hand and the hand-rim as crucial, as this is the site where force is transferred from the user to the wheelchair. An area that may contribute significantly towards improved coupling at the wheelchair-user interface also emerged. The use of gloves was considered to form a highly important part of WCR players' equipment, yet no research has been attributed to this area. Even though gloves are not a direct area of wheelchair configuration, they do act directly at the wheelchair-user interface and were clearly thought to influence performance. Also given that no gloves specific to the demands of WCR exist, future investigations into the effects of different glove types on the ergonomics of wheelchair performance would seem beneficial.

3.5.3 Supplementary areas of wheelchair configuration

Some of the smaller areas of wheelchair configuration, which have not previously been considered from a quantitative research perspective, were still thought to have a significant bearing on performance. For instance, participants were of the opinion that selecting a longer wheelchair with a more posterior footrest position, front and rear castor wheels and the use of strapping all contributed towards improved stability. This seemed particularly valuable to LP, as the additional stability that they attributed to these areas has allowed for more advanced configurations to be selected, which their impairment level would previously never have allowed. Yet, participants acknowledged that adjusting these areas of wheelchair configuration 'too much' compromised mobility and manoeuvrability. However,

the point at which 'too much' occurs was again, frequently decided by trial and error and what subjectively felt right. Once again this demonstrates how a fine line exists with optimising wheelchair configuration, even with some of the potentially smaller areas. In order to assist athletes with the selection process when configuring a new sports wheelchair, future quantitative research is required to determine where the optimal positions for each of these settings occur in relation to the user.

Although, future research into the exact effects of manipulating some of these supplementary areas on performance would be beneficial, it is perhaps not the most pressing issue, as it has become clear that a great deal of sport specific research is still required into some of the principal areas of wheelchair configuration. However, given the obvious contribution that the supplementary areas of configuration were perceived to have, it is imperative that future studies into the principal areas of wheelchair configuration acknowledge and at least control these supplementary areas.

3.5.4 Limitations

It may be considered that the small sample size included in the current investigation was insufficient to achieve data saturation. This may hold some truth, however, given the phenomenological nature of the study, a sample size in the range of three to ten participants is suitable for this type of analysis (Creswell, 2007). In addition to this, the fact that participants were recruited through purposive sampling meant the participant group were particularly homogenous. This served to ensure that the information provided by this group of participants should be detailed enough not to warrant a larger sample size.

The homogenous nature of the participants could also be viewed as a limitation. In order to gain greater detail into how wheelchair athletes perceive areas of configuration to impact on performance and to establish which areas are in need of future research, a more heterogeneous sample may be advantageous. For example, establishing the opinions and beliefs of less experienced athletes may have provided further insights into the phenomenon. However, given that this was the first study of its kind into wheelchair configuration, it should serve as an extremely useful foundation for any future research to explore and build on.

3.6 CONCLUSION

The results of the current investigation have demonstrated that experienced wheelchair athletes possess a strong and relatively united understanding of how making 'general' modifications to areas of wheelchair configuration affected their performance. However, it

was noticeable that establishing optimal settings was a very complex process that athletes found difficult to isolate. Therefore, it is essential that future quantitative research attention is undertaken to help optimise areas of players' wheelchair configuration, specific to the anthropometrics and impairment of the individual. This should make players more aware of the consequences of some of the decisions they are making when configuring a new sports wheelchair, as it was extremely apparent that decisions are currently based predominantly on trial and error. Not only may these decisions and subsequent selections be limiting their performance, but they could also be placing them at an increased risk of injury.

The current investigation has also played a valuable role in identifying which areas are in need of the most urgent research attention. The effects of rear-wheel camber in particular, warrants further research as a result of the disparity amongst participants' subjective views as well as within scientific literature. Investigations into materials and configurations that affect the coupling between the user and the hand-rims were highlighted as a vital contributing factor towards mobility performance. Seat positioning and the size of the main wheels were also identified as key determinants of mobility performance. General adjustments to both these areas of configuration appeared to be somewhat understood, yet the detection of optimal settings was not possible and is in need of future evidence based research.

The aim of the next study (Chapter 4) was to establish the effect of glove type on the coupling between the user and the hand-rims and the resultant effects that such manipulations to the wheelchair-user interface had on maximal effort mobility performance. This area was selected as the first quantitative experimental study, not only because of the emphasis placed on coupling during the current study, but also due to the proximity of the Beijing 2008 Paralympic Games. It was deemed inappropriate and potentially disruptive to investigate manipulations to areas of wheelchair configuration so close to a major sporting competition. Subsequently, modifications to an area of adaptive equipment which were thought to be less invasive, yet equally meaningful from a performance perspective formed the focus of the first experimental study. To this extent, investigations into the effects of key areas of wheelchair configuration (rear-wheel camber – Chapters 5 & 6; wheel size – Chapters 7 & 8), were investigated post Beijing.

Chapter Four

The Influence of Glove Type on Mobility Performance for Wheelchair Rugby Players

Mason BS, Woude van der LHV & Goosey-Tolfrey VL (2009). The influence of glove type on mobility performance for wheelchair rugby players. *American Journal of Physical Medicine & Rehabilitation*, 88(7), 559-570.

4.1 ABSTRACT

Purpose: To determine the effectiveness of different glove types on mobility performance in a series of sport-specific field tests. ***Methods:*** A group of international WCR players (N = 10) performed three drills in four glove conditions: (i) players' current glove selection (CON), (ii) American football glove (NFL), (iii) building glove (BLD) and (iv) new prototype glove (HYB). Performance was measured through a velocometer sampling at 100 Hz to obtain times taken, velocities and accelerations within each drill. Subjective data were also collected on glove performance via a short likert scale questionnaire. ***Results:*** A two-way analysis of variance (ANOVA) with repeated measures revealed that participants performed statistically better for measures of acceleration and sprinting in CON compared to HYB ($P < 0.05$). Subjective data identified that players also favoured CON in comparison to other glove types. Slight discrepancies were evident amongst classification levels concerning BLD, which appeared to be less suited to LP due to the reduced grip and protection they provided. ***Conclusions:*** Participants' current gloves that had been modified for the specific demands of WCR and the individual user were more effective for aspects of mobility performance than other glove types.

4.2 INTRODUCTION

Experienced wheelchair sportsmen interviewed in Chapter 3 signified the importance of coupling between the hand and hand-rim in wheelchair propulsion, because it is the site of force transfer from the user to the wheels. Hand-rim wheelchair propulsion has been identified as an inefficient form of ambulation, as even in highly trained and experienced athletes ME rarely exceeds 11.5% (Woude et al., 1986; Woude et al., 1988a; Woude et al., 1988b; Vanlandewijck et al., 1994a). Poor coupling between the user and the hand-rim contributes to and often exacerbates the inefficiencies observed. The more grip and friction that can be generated between the user and the hand-rims, the greater the potential for effective force transmission (Gaines and La, 1986). Obtaining sufficient grip at the wheelchair-user interface can be particularly problematic for WCR players due to the severity of their impairment and the limited hand function they often possess. Therefore, the ability to generate sufficient torque around the hub of the wheels in order to maintain or develop speeds can be difficult (Woude et al., 2003). Players often seek to optimise their efficiency and force generation through making modifications to areas of wheelchair design. One additional method used to improve wheelchair mobility is through the use of adaptive equipment. Subsequently, the majority of WCR players use some form of gloves, taping and/or hand-rim modifications during competition to compensate for this lack of grip (Chapter 3).

Kratz et al. (1997) have highlighted the need for adaptive equipment for wheelchair athletes, including WCR players. However, a glove developed specifically for the demands of WCR does not currently exist, causing players to wear a range of gloves that have been designed for other purposes (Lutgendorf et al., 2009). What is also noticeable is that players frequently modify these gloves, through the incorporation of additional materials and substances, such as handball wax or glue, in an attempt to accommodate the performance requirements of their sport, as observed in wheelchair racing (Woude et al., 2003).

Gloves also act as a form of protection for WCR players, since wheelchair propulsion is an extremely demanding action that places both the hands and wrists under severe stress (Burnham et al., 1994; Malone et al., 1998; Woude et al., 2003; Richter et al., 2006). This is predominantly evident during braking manoeuvres, which is a common feature of WCR (Vanlandewijck et al., 2001). The large coefficient of friction developed whilst braking can cause burning and abrasion to the hands if unprotected (Richter et al., 2006). Wrist injuries are also common, as exemplified by Burnham and Steadward (1994) who revealed that in a group of highly trained wheelchair athletes, 46% were diagnosed with some form of medial nerve dysfunction in the carpal tunnel. Malone et al. (1998) identified that a combination of a glove

and wrist splint significantly reduced the degree of extension at the wrist, which is thought to predispose to carpal tunnel syndrome, without negatively affecting maximal wheelchair velocity. Gellman et al. (1988) indicated that in a group of paraplegic individuals who used wheelchairs for daily life ambulation, 49% also experienced symptoms of carpal tunnel syndrome. Therefore, although the focus of the present investigation is on mobility performance in WCR, the use of gloves may also have implications on aspects of daily life wheelchair propulsion. Minimising the impact forces may lead to a lower grip force being required, which could potentially lead to a lower energy cost and more efficient form of propulsion for all wheelchair users.

The absence of a glove specific to WCR may seem unsurprising, given the limited research conducted in the area. To the author's knowledge, only Lutgendorf et al. (2009) have analysed the performance effects of different types of gloves used for WCR on sport-specific parameters. Using AB participants this study revealed that NFL and BLD performed consistently better than multipurpose and no gloves for aspects of mobility. However, differences in functional capabilities and propulsion kinematics that have been observed between AB and wheelchair dependent individuals, means the transfer of this data to WCR players with tetraplegia may be questionable (Veeger et al., 1992a). Kinematic differences (Dallmeijer et al., 1994) and reduced effectiveness of force application (Dallmeijer et al., 1998) have also been observed between wheelchair users with tetraplegia in relation to those with paraplegia. In addition to this, Dallmeijer and Woude (2001) identified that differences in physiological responses existed within individuals with tetraplegia. It was shown that participants with an incomplete cervical lesion had a higher peak $\dot{V}O_2$ than those with a complete lesion. The inter-individual differences that have been observed in tetraplegic wheelchair users is of clear relevance to WCR players and glove selection could subsequently be influenced by both lesion and level of completeness.

In contrast to glove material, consideration has been given to different hand-rim configurations and their effects on physiological and kinetic variables, with varying results reported (Linden et al., 1996; Woude et al., 2003; Richter and Axelson, 2005; Koontz et al., 2006; Richter et al., 2006). However, it was evident that all of these studies have been conducted to improve wheelchair propulsion efficiency during daily life activities, as exemplified by the sub-maximal nature of their testing protocols. No research has been conducted on the influence of the interaction between the hands and hand-rims during the dynamic bouts of wheelchair propulsion required for WCR.

An investigation into game efficiency in WCR by Molik et al. (2006) identified differences in the match analysis patterns of WCR players of differing impairments for numerous aspects of match play. Using the IWRF classification system it was shown that players with a classification of ≤ 1.5 (most impaired) differed significantly in their game related activities in comparison to players with a classification of ≥ 2.0 (least impaired). Molik et al. (2006) suggested that these differences in activities were the result of players being assigned with different roles on court dependent upon the severity of their impairment. More severely impaired individuals were often found to occupy defensive roles on court and were not predominantly ball handlers. Alternatively, this role was occupied by the more able players who often fulfilled more offensive roles. Subsequently, it was suggested that differences in equipment between these two groups could contribute to the differences in the efficiency of their match play activities.

The aim of the current investigation was to compare the performance of four types of gloves, including three that are currently used by WCR players and a newly developed prototype glove, during a series of sports specific WCR field tests and through their subjective ratings. It was hypothesised that players would perform more effectively in the new prototype glove, which had been designed to accommodate the demands of the sport and the functional capabilities of users. A secondary purpose was to identify whether any relationships existed between glove type and performance for players of varying impairment levels.

4.3 METHODS

4.3.1 Participants

An international WCR squad volunteered for the current study. To be included, players' current choice of glove (CON) could not match any of the other three gloves sampled. Subsequently, ten highly trained members of the squad (male, N = 9; female, N = 1; age: 30 ± 5 years; mass: 66.2 ± 6.9 kg) were able to participate. Using a similar approach to Molik et al. (2006) participants were categorised into two classification levels based on their current IWRF classification, as adopted in Chapter 3. Participants with a classification of ≥ 2.0 (N = 5) were categorised as HP (least impaired) and players with a classification of ≤ 1.5 (N = 5) were categorised as LP (most impaired). All participants provided their written informed consent prior to any involvement in the study. Ethical approval was obtained through the University Ethics Advisory Committee.

4.3.2 Equipment

The range of materials and substances that were used in the CON condition are documented in Table 4.1. Participants were prevented from applying any substance immediately before testing and whatever substance was on their wheels was also present during other glove conditions. All participants completed testing in their own rugby wheelchairs with details of each player's wheel size and wheelchair mass reported in Table 4.1. Table 4.1 also illustrated that all of HP used a standard anodised metal rim on their wheelchairs, whereas LP all used a rubber coated hand-rim. In addition to this, hand-rim tube diameters and the distance they were set away from the main wheel differed between classification groups. All HP used a hand-rim that was smaller in diameter and was located closer to the main wheel than LP, who adopted a larger tube diameter hand-rim that was set further away from the wheel.

Table 4.1 – Participant characteristics and descriptions of materials in CON.

Impairment & classification		Age (years)	Mass (kg)	Description of current gloves and materials used at the wheelchair-user interface (CON)
C7/8 Com	2.5	29	60.7	NFL with suede patch on palm & diving glue. Metal hand-rim
C6/7 Inc	2.5	26	67.9	Cycling glove with leather palm & handball wax. Metal hand-rim
C7 Com	2.5	34	61.9	NFL with rubber patch on thumb/palm & glue. Metal hand-rim
C6/7 Com	2.0	27	66.3	NFL with rubber patch on thumb/palm & glue. Metal hand-rim
C6/7 Inc	2.0	35	78.0	Leather glove with suede patch on palm & diving glue. Metal hand-rim
C7 Com	1.5	34	76.1	NFL with leather patch on palm & diving glue. Rubber hand-rims
C6/7 Com	1.5	21	55.4	Leather glove with suede patch on palm and rear knuckles. Rubber hand-rims
C5/6 Com	1.0	35	65.8	BLD underneath fingerless gardening gloves & handball wax. Rubber hand-rims
C5/6 Com	1.0	36	68.3	Rubber glove with added rubber cuff around palm and wrist. Rubber hand-rims
C6 Com	0.5	24	61.7	Rubber glove with tape around fingers. Rubber hand-rims

Key: Classifications as governed by IWRF; Com = Complete SCI; Inc = Incomplete SCI.

In addition to CON, participants were tested in three other types of gloves: (i) NFL, (ii) BLD and (iii) HYB. The NFL and BLD gloves (Figure 4.1) were sampled in the current

investigation to extend the work of Lutgendorf et al. (2009). The NFL was a synthetic glove manufactured by Neumann's (Cookeville, TN) and is reinforced with a tackified cowhide leather material on the palmar side of the hands. The BLD was a knitted cotton glove manufactured by Reflex, which incorporated a latex coated palm. Finally, HYB was a new prototype glove that had been developed at Loughborough University (UK) to specifically target LP in WCR. It was aimed at providing these players, who are predominantly not ball handlers, with added protection around the dorsal side of the hand, wrist and distal section of the forearm (Figure 4.2).

Figure 4.1 - American football glove (NFL) and building glove (BLD).

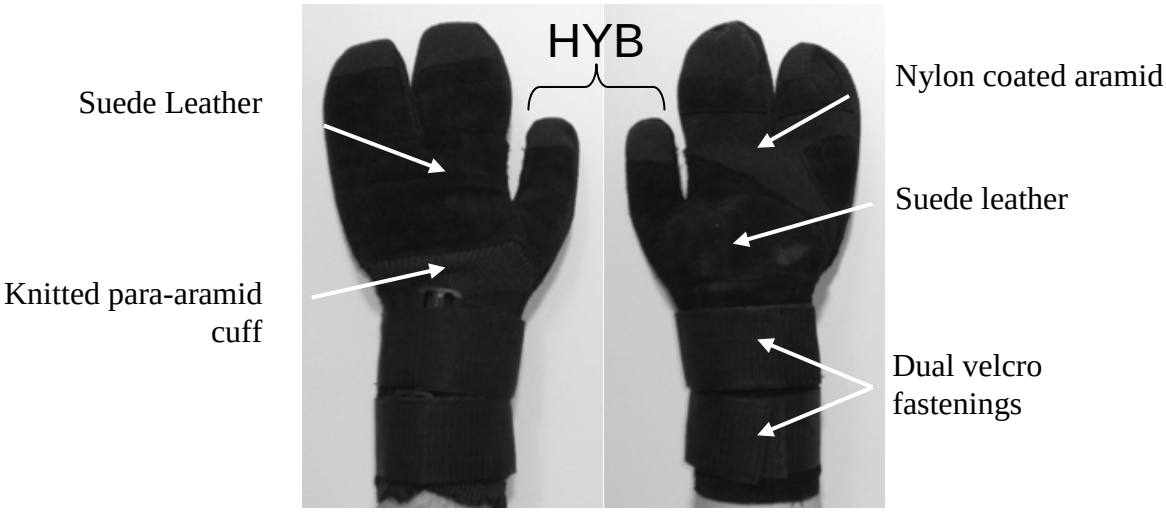

Figure 4.2 - Dorsal (left) and palmar (right) view of HYB and its properties.

To assess the effects of glove type on aspects of mobility performance, a velocometer, used by Moss et al. (2003) was fitted to each participant's wheelchair. The velocometer developed at Manchester Metropolitan University (UK), uses an optical encoder to transmit

pulses per revolution of the wheel to an analogue-to-digital converter. Sampling at 100 Hz, this produced velocity traces with respect to each push. Further calculations allowed for accelerations and decelerations to be deduced. Brower wireless timing gates (Draper, UT) were also utilised to assess the times taken to perform the drills.

4.3.3 Wheelchair handling skills

Three drills, incorporating movements specific to WCR were used to assess the effectiveness of each glove. All drills were performed on a sprung indoor sports hall surface that was frequently used by WCR squads for training and competition. A series of performance measures were taken and grouped into three categories based on the aspects of performance they assessed: ‘acceleration’, ‘braking’ and ‘sprinting’. All drills were familiar to the participants, having been performed on a regular basis as part of a sport science monitoring programme.

4.3.3.1 Drill 1 (acceleration drill)

Drill 1 incorporated acceleration, braking and backwards pulling movements. Participants were required to repeatedly accelerate from a standstill over increasing distances of 2.5 m, 5 m and 10 m with a series of braking and backward pulling manoeuvres in between (Figure 4.3). Acceleration performance was assessed in this drill by the peak velocities reached within the 2.5 m and 5 m sprints. Braking performance was measured through the decelerations that occurred at the end of both of these sprints, as derived from the velocometer. Decelerations were determined from the highest point where braking commenced to the point immediately before backwards pulling was initiated.

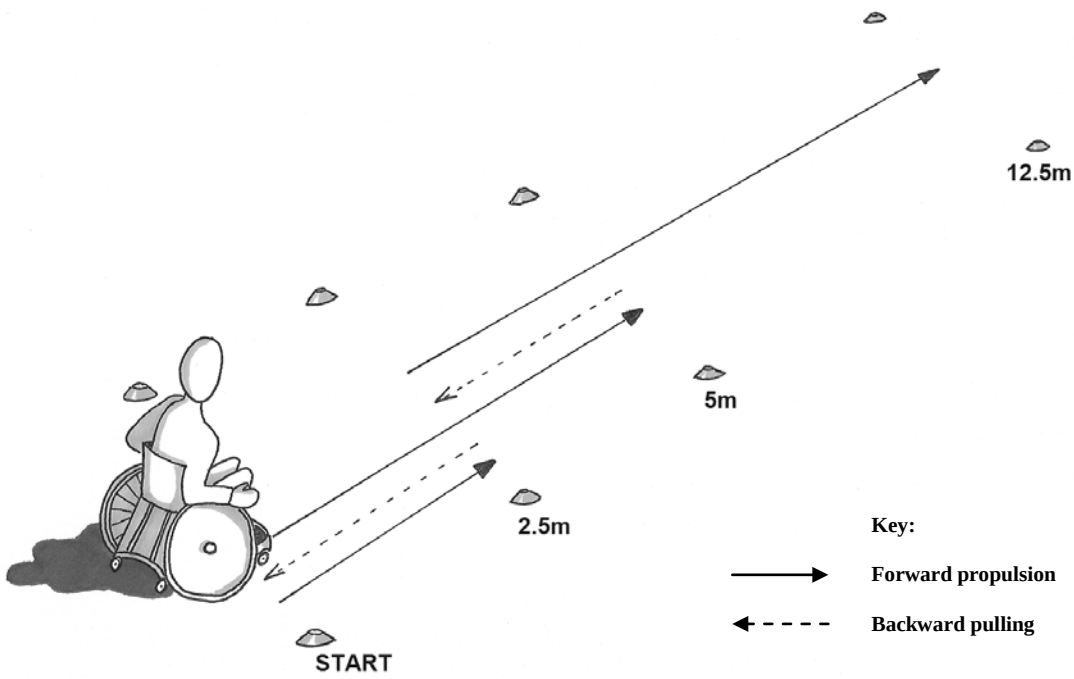

Figure 4.3 - The acceleration drill.

4.3.3.2 Drill 2 (sprint drill)

Drill 2 assessed both initial acceleration and maximal linear sprinting performance during a 15 m sprint from a standstill. Acceleration was calculated over the first two pushes via the velocometer, using the point at which acceleration commenced to the peak velocity of the second push. Sprinting performance was assessed by the overall time taken to complete the drill and the peak velocities reached.

4.3.3.3 Drill 3 (agility drill)

Drill 3 was an agility drill that measured multidirectional sprinting performance. Participants had a 5 m rolling start before timing commenced and accelerated for a further 9 m before performing a sharp turn (approximately 230˚). Participants then manoeuvred themselves back through three more cones in a slalom fashion to complete the course (Figure 4.4). If a cone was hit, the trial was void and had to be repeated. Performance was purely assessed by the time taken to complete this drill.

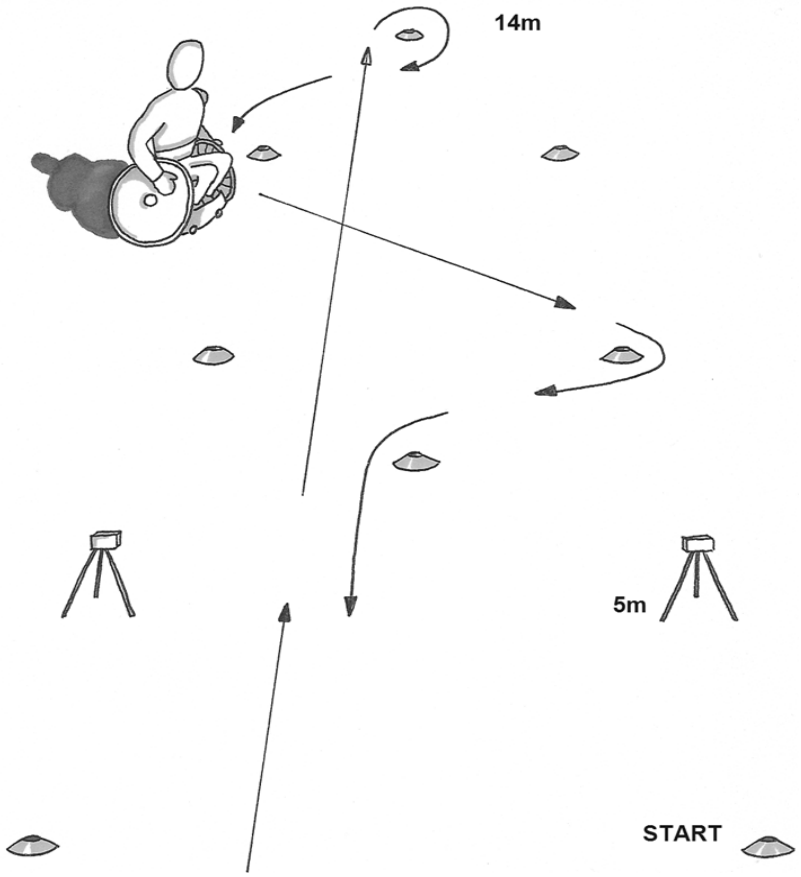

Figure 4.4 - The agility drill.

Participants performed drills 1 and 2 once for each glove condition and repeated drill 3 twice (once with a right turn at the top and once with a left turn). The order of the drills remained the same throughout testing. All participants completed the first circuit of drills in CON to aid with familiarity, however, the order for the three remaining gloves was randomised. To ensure that the condition of the gloves had no bearing upon the results, all gloves (excluding CON) were only worn twice (from brand new and after one use). Each drill demonstrated high levels of within-day reliability, with coefficient of variation values ranging from 1.0 % (manoeuvrability) to 2.1 % (linear mobility).

Participants were instructed to perform all trials at maximal effort and as quickly as possible. Sufficient recovery between trials was permitted using a rating of perceived exertion scale (RPE), whereby participants were not allowed to commence the next trial until they were at or below 10 on the scale (Borg, 1970). These were chosen to predict exertion levels instead of monitoring HR, as HR values are often unreliable in tetraplegic individuals due to the reduced activation of their sympathetic nervous system (Coutts et al., 1985).

4.3.4 Subjective appreciation

On completion of the wheelchair handling skills, participants completed a brief likert scale questionnaire in order to provide subjective feedback on the gloves. This questionnaire investigated each glove's comfort, grip, protection and ease of donning and offing (preparation) and was ranked on a scale of 1 – 5 (1 = very poor; 5 = excellent).

4.3.5 Statistical analyses

All data were analysed using the Statistical Package for the Social Sciences (SPSS, Chicago, IL). A two-way ANOVA with repeated measures was conducted on all objective and subjective measures of performance to determine any main effects for glove type and interactions between glove type and classification level. Planned simple contrasts were incorporated since HYB was hypothesised as being the best performing glove type. Pairwise comparisons with a Sidak adjustment were utilised to explore any additional significant main effects between other glove conditions (CON, NFL & BLD). All data was accepted as statistically significant at $P < 0.05$. Effect sizes were reported to signify the meaningfulness of the findings, whereby an $r > 0.5$ represented a large effect (Cohen, 1992).

4.4 RESULTS

4.4.1 Wheelchair handling skills

It was revealed that participants achieved significantly higher peak velocities in CON than in NFL, BLD and HYB over both 2.5 m ($P \leq 0.036$, $r = 0.70$ to 0.81) and 5 m ($P \leq 0.041$, $r = 0.73$ to 0.88). However, acceleration performance as assessed over the first two pushes of the 'sprint drill' did not reveal any effect of glove type on this aspect of performance (Table 4.2).

Table 4.2 also demonstrates that braking performance did not differ significantly between glove types, as shown by the decelerations produced at the end of the 2.5 m and 5 m bouts of forwards propulsion in the 'acceleration drill'. On inspection of the mean values, a slight trend did appear to exist, with CON seemingly being able to stop at a greater rate than NFL ($r = 0.75$) and HYB ($r = 0.90$) after 5 m of forwards propulsion, although these trends were not shown to be statistically significant.

Aspects of sprinting performance were also influenced by glove type, as differences were evident for the times taken to complete the 'sprint drill' and the 'agility drill'. Significantly quicker times were achieved in CON than HYB in the 15 m sprint (P 0.012, r =

0.76) and for all other gloves in the 'agility drill' ($P \leq 0.013$, r = 0.75 to 0.77). Pairwise comparisons also showed that times were significantly quicker for NFL in relation to HYB during the 'agility drill' ($P = 0.008$, $r = 0.85$).

Table 4.2 – Glove performance during the wheelchair handling skills. All values are means (± SD).

	***P* value**	**CON**	**NFL**	**BLD**	**HYB**
Acceleration:					
Peak velocity – 2.5 m ($m \cdot s^{-1}$)	0.005	2.12 (0.22)	2.05[a] (0.17)	2.00[a] (0.21)	1.97[a] (0.16)
Peak velocity – 5 m ($m \cdot s^{-1}$)	0.010	2.56 (0.24)	2.46[a] (0.25)	2.49[a] (0.26)	2.48[a] (0.20)
Over first two pushes ($m \cdot s^{2}$)	0.395	1.81 (0.32)	1.76 (0.45)	1.73 (0.44)	1.69 (0.34)
Braking:					
After 2.5 m of acceleration ($m \cdot s^{2}$)	0.190	5.80 (2.91)	5.57 (2.28)	5.28 (1.32)	4.19 (1.63)
After 5 m of acceleration ($m \cdot s^{2}$)	0.096	7.52 (2.70)	5.93 (2.99)	6.23 (2.30)	5.64 (3.32)
Sprinting:					
15 m sprint times (s)	0.041	5.85 (0.51)	5.86 (0.52)	5.91 (0.58)	5.98[a] (0.51)
Peak velocity – 15 m ($m \cdot s^{-1}$)	0.540	3.69 (0.35)	3.61 (0.36)	3.58 (0.38)	3.56 (0.31)
Agility drill - times (s)	0.001	12.38 (0.91)	12.52[a] (1.03)	12.57[a] (1.11)	12.84[ab] (0.93)

Key: [a] represents a significant difference to CON.
[b] represents a significant difference to NFL.

Classification level appeared to have no bearing on glove performance. No interactions existed between glove type and classification level with respect to performance for any of the wheelchair handling skills. Yet it was clear from the between-subject effects that performance in all aspects of wheelchair handling performance was significantly greater in HP than in LP ($P < 0.01$). Therefore, although performance clearly differed between classification groups, it appeared that glove type did not significantly affect this.

4.4.2 Subjective appreciation

Results from the questionnaire demonstrated that participants' subjective ratings of glove performance differed considerably, as identified by significant main effects of glove

type for the comfort, grip, protection and preparation they were perceived to provide (P < 0.0005).

4.4.2.1 Comfort

Participants perceived that CON were significantly more comfortable than all other glove types (CON = 4.6 ± 0.52; NFL = 3.3 ± 1.64; BLD = 2.65 ± 0.75; HYB = 1.9 ± 0.88; $P \leq 0.01$, r = 0.76 to 0.95). An interaction between classification groups, CON and NFL existed for perceived comfort (P = 0.041). As illustrated in Figure 4.5a, HP rated CON and NFL to be equally comfortable, whereas NFL were considered less comfortable than CON by LP.

4.4.2.2 Grip

Improved grip was perceived to be afforded in CON compared to other types of glove (CON = 4.4 ± 0.52; NFL = 3.5 ± 0.85; BLD = 3.6 ± 1.07; HYB = 2.4 ± 0.84; P < 0.029, r = 0.69 to 0.88). Pairwise comparisons also revealed that participants favoured the grip of NFL (P = 0.001, r = 0.93) over HYB. Participants from different classification groups also appeared to favour different gloves for the amount of grip they provided (P = 0.012) with HP demonstrating comparable grip ratings between CON and BLD. However, LP implied that BLD offered slightly reduced grip in comparison to CON (Figure 4.5b).

4.4.2.3 Protection

A significantly greater degree of protection was thought to be provided by CON and HYB in relation to NFL and BLD (P < 0.0005, r = 0.93 to 0.95). A significant interaction was also evident, which identified that LP felt that BLD offered far less protection than CON. However, HP did not seem to experience such larger differences in protection between these two glove types (Figure 4.5c).

4.4.2.4 Preparation

It was revealed that participants found some gloves easier to take on and off than others, with CON believed to be easier than NFL (P = 0.013, r = 0.75) and HYB (P < 0.0005, r = 0.92). Pairwise comparisons also established that participants felt preparation was facilitated in both NFL (P = 0.006, r = 0.87) and BLD (P = 0.005, r = 0.88) compared to HYB (Figure 4.5d).

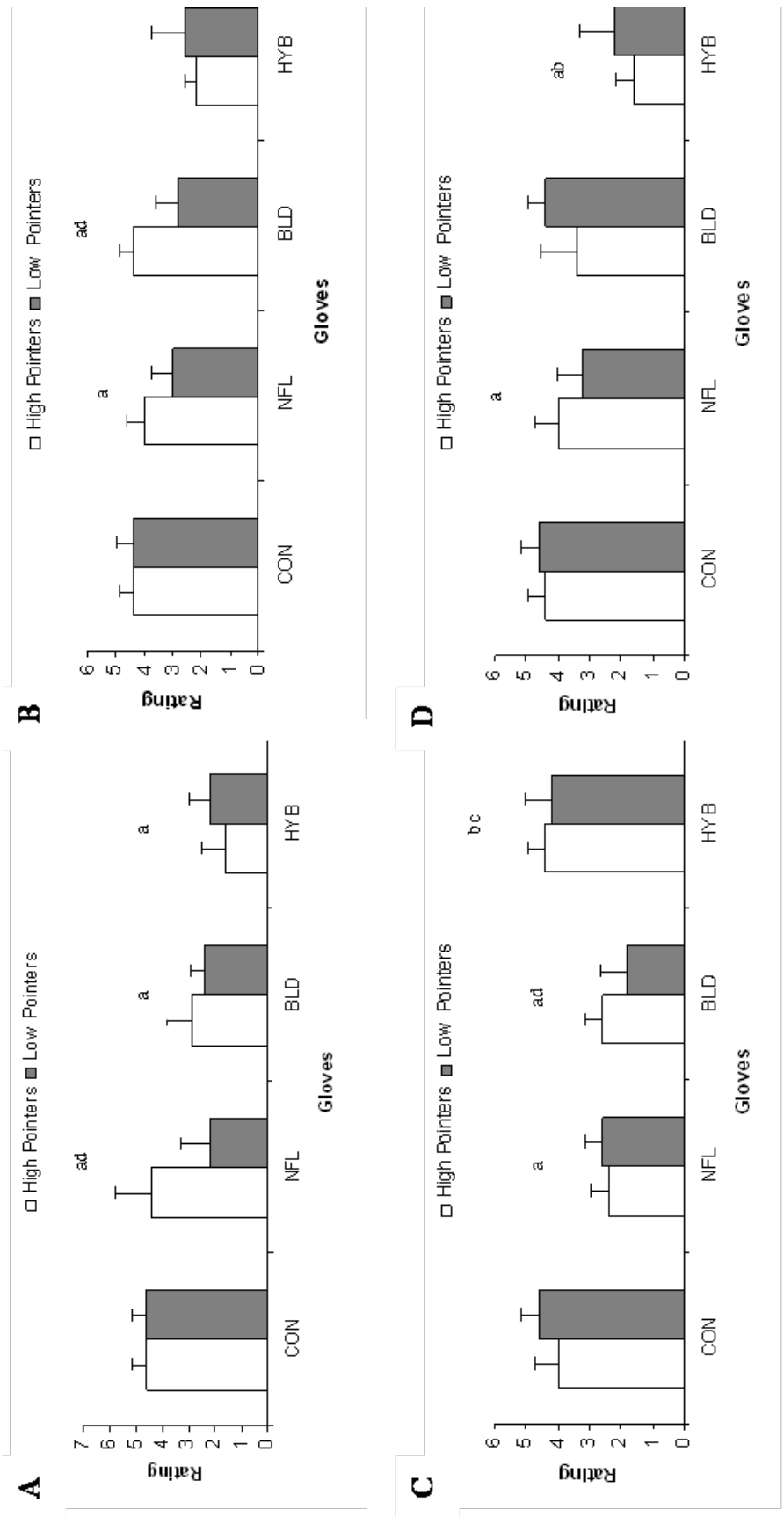

Figure 4.5 - Subjective assessment of glove performance: (a) comfort, (b) grip, (c) protection, (d) preparation. [a] denotes a significant difference with CON, [b] with NFL and [c] with BLD. [d] denotes a significant interaction between classification level and CON.

4.5 DISCUSSION

The results of the current study disproved the hypothesis that a new glove developed specifically for the demands of WCR (HYB) would outperform other gloves. Alternatively, gloves which have been specifically modified and customised to the requirements of each individual (CON) proved to perform most effectively. Acceleration performance, which is a vital performance indicator in the wheelchair court sports (Vanlandewijck et al., 2001), was shown to be superior in CON, as demonstrated by the significantly higher velocities reached within 2.5 m and 5 m sprints from a standstill. Measures of sprinting performance were also shown to be improved in CON, with faster times exhibited during the linear 15 m sprint and the 'agility drill'.

The greater performance that CON enabled could be attributed to a number of potential contributing factors. It may have simply been due the improved familiarity players would have had with this glove, as they did subjectively feel more comfortable when performing with them. Familiarity has previously been shown to play a significant role within wheelchair performance. It has been reported that when manipulating propulsion frequency, participants' self-selected frequencies have been revealed to be the most effective in terms of $\dot{V}O_2$, ME (Woude et al., 1989c) and pushing economy (Goosey et al., 2000). However, Koontz et al. (2006) identified an improvement in grip moments when using a 'new' ergonomic hand-rim in comparison to participants 'current' hand-rims. Yet the reason for this was likely to be due to the fact that Koontz et al. (2006) allowed participants to familiarise themselves with the new piece of equipment for at least two weeks prior to testing. Therefore, familiarity with new equipment could play a vital role in performance and would warrant consideration for future studies looking at the influence of glove type.

It was also likely that the improved performance in CON resulted from an improved amount of grip generated at the wheelchair-user interface, as suggested by participants' perceptions. Although this suggestion could not be supported quantitatively within the current study, it is quite possible that a greater amount of torque could be generated around the wheels when using CON. This potential improvement in grip could have also resulted from the glue or handball wax that was used on some players' gloves. Often WCR players, use an adhesive substance to gain more purchase on their wheels and to aid with their ball handling skills. Therefore, a possible limitation of the current study was its failure to completely control the use of these substances and even though use was restricted prior to data collection, it could not be prevented altogether.

Another possible explanation for the superior performance of CON was through the physical condition of the gloves. Although all other gloves were tested in either a 'brand new' or 'worn once' condition, the condition of CON could not be controlled, as they were the player's own gloves and subsequently were not as new as the other glove types tested. Therefore, it may be that a breaking in period exists for gloves and that CON were more worn in and prepared for wheelchair propulsion than other, newer glove types. In addition to this a slight order effect may have been evident, since participants performed all trials in CON first. The reason for this order was because testing was conducted during a national training camp on a day where the squad were doing some low intensity scrimmaging. Therefore, participants were already wearing CON and given the time and effort it takes these players to take gloves off and on prior to performance, it was deemed more feasible to conduct the first circuit of all testing in these gloves. It is unlikely that this order had a substantial effect on the results, as players were familiar with all the drills. Also, fatigue was unlikely to be a contributing factor given the participants high level of conditioning enabled by their full time status and the fact that drills were short in duration with recovery controlled by RPE ratings. To ensure that the results were not influenced by an order effect, a one-way ANOVA was conducted for the results of one drill, regardless of glove type and no significant differences were reported (P = 0.971).

The NFL and BLD gloves, which were previously shown to be the best performing gloves when sampled with AB participants (Lutgendorf et al., 2009), did not perform as well for WCR players when compared to CON. The only area of wheelchair handling skills whereby either of these gloves performed favourably was during the 'agility drill', where it was shown that NFL enabled significantly quicker times than HYB. This may seem slightly surprising, as it can be seen in Table 4.1 that both NFL and BLD formed the basis of many participants' modified gloves, so it would appear that they were somewhat valued by participants. However, from the subjective appreciation it was clear that NFL and BLD were perceived as being significantly poorer than CON for the level of comfort, grip and protection they afforded.

The prototype glove (HYB) that had been designed specifically for WCR, was shown to perform relatively poorly in relation to other glove types. This glove was predominantly composed of a suede leather material around the area of the hands that players used for propulsion. This material was frequently seen to be selected by players to modify their own gloves (Table 4.1), which as previously mentioned, performed well for all wheelchair handling skills. Therefore, it was likely that areas other than material accounted for the

inferior performance exhibited by HYB. It maybe that this glove was too bulky, rigid and inflexible to allow for efficient speed and grip to be generated around the wheels, as many players alluded to. As a result, the only area that HYB seemed to excel at was for their protective properties, as subjectively rated by the participants. It was also possible that HYB could prevent some of the overuse traumas experienced at the hand and wrist owing to its bulky, rigid design. This is likely to have minimised the amount of forced wrist extension, although this would obviously require further kinematic investigations.

Lutgendorf et al. (2009) revealed that poor braking performance was observed for AB participants when using no gloves and commented that this was the result of participants' anxieties about sustaining injuries to the hands when attempting to stop the wheels at high speeds. However, the fact that HYB scored highly for protection, yet not for braking again suggested that they were too bulky to brake effectively, as it could be expected that the grip would be similar to CON given the similarities in materials. It seemed likely that the more strategic location of added material in CON provides grip and protection to the specific areas of the hand used for wheelchair propulsion. Whereas HYB would appear to offer a more overall protection to the hands, even the parts that are not directly used for wheelchair propulsion, which could impinge hand dexterity and account for the reduced performance of wheelchair handling skills in these gloves. This demonstrates the fine line that appears to exist between ensuring enough protection to the hands to prevent injuries and providing too much protection, whereby players' movements and subsequent performance can become inhibited.

The current study did not identify any interactions between glove types and classification levels for the performance of any of the wheelchair handling skills. However, results from the questionnaire revealed that subjective ratings for each gloves performance did vary dependent on classification level. It could be suggested that no interactions between glove type and classification level for performance in the wheelchair handling skills were evident due to the small sample size within each group. However, heterogeneity within groups was not thought to account for the lack of any differences, given the highly significant between-subject effects for all measures of performance and the degree of consistency in performance within classification levels. Significant interactions were established for perceived comfort between CON and NFL, whereby HP found CON and NFL to be equally comfortable. Alternatively, LP felt that NFL was considerably less comfortable in relation to CON. Other interactions existed between classification groups for the perceived amount of grip and protection that was afforded by CON and BLD, with little difference between the grip and protection provided by both gloves expressed in HP. However, LP identified that

BLD offered reduced grip and protection in relation to CON. It was possible that these subjective interactions between classification groups were the result of the different pushing techniques employed. All HP pushed their wheelchair with the palmar side of their hand, whereas LP frequently switched to a backhanded technique and contacted the hand-rims with the dorsal side of their hand. Dallmeijer et al. (1994) have observed kinematical differences at the hand between individuals with differing levels of SCI, although small sample sizes prevented any statistically significant differences from being identified. A 'para-backhand' technique has previously been observed within wheelchair racing, whereby individuals contact the hand-rim with the dorsal side of the fingers (Chow et al., 2001). Therefore, it could be that significant kinematic differences exist between classification levels within WCR and that this could explain the differences in subjective appreciation of glove types. The knitted cotton material on the dorsal side of BLD may not provide the necessary grip or protection required for this style of pushing adopted by LP.

The presence of variability between classification groups for gloves' subjective performances, yet the absence of any such interaction during the objective wheelchair handling skills could be attributed to the potential role of the hand-rims. As previously mentioned the configuration of the hand-rims differed between HP and LP. It was clear that there was no variation in performance between CON, NFL, BLD and HYB as a result of these differing hand-rims, due to the absence of any significant interactions between classification groups. Given some of the previous research findings concerning hand-rim configuration, further research would be warranted to identify the direct impact that both types of hand-rims used for WCR can have on performance. Linden et al. (1996) found that larger diameter hand-rim tubes equated to improvements in ME. Richter et al. (2006) also revealed that more flexible hand-rims may be more efficient due to the lower muscular activity they produced during sub-maximal wheelchair propulsion. Both these findings may have implications, particularly on the foam coated hand-rims used by low point WCR players, which are larger in diameter and are also likely to offer more flex in relation to metal hand-rims.

The use of field testing in the current investigation allowed for a thorough assessment of each gloves performance under sport-specific conditions. However, future research of a similar nature may benefit from a more comprehensive set of measures to assess each area of performance. Although, the current study identified differences in some aspects of acceleration and sprinting performance between gloves, these were not seen for all measures within these categories. Also, no significant differences were observed for the decelerations used to assess braking performance. Although the mean values did suggest a trend for CON to

again be more efficient at this aspect of performance, this was not statistically significant, which appeared to be due to the large standard deviations present (Table 4.2). Therefore, it may be that the velocometer could not sample at a sufficiently high frequency in order to accurately assess the sharp decelerations that were occurring from relatively high speeds. Given that braking forms an important part of WCR match play, further research may be advised into methods for assessing this area of performance in a field environment (Vanlandewijck et al., 2001).

The influence of glove type upon ball handling skills also requires future consideration in order to provide a more holistic evaluation of glove performance specific to WCR. Lutgendorf et al. (2009) demonstrated that glove type had no significant effect on ball handling accuracy in AB participants. However, it cannot be assumed that this finding would also apply to tetraplegic individuals with limited hand function and hence ball handling skills need to be investigated in these individuals whilst wearing different gloves.

In order to determine the exact causes for some of the performance effects observed under different glove conditions, further investigations under controlled laboratory conditions would also be recommended. It would also be desirable to control areas of wheelchair configuration, as factors such as seating position are likely to influence the point and amount of the wheel that is available to push on. This could allow the influence of factors such as glove type, hand-rim configuration and glue/handball wax to be studied more in isolation to improve the understanding about each factors contribution towards mobility performance. Although limited previous research has been conducted on the influence of different user to hand-rim interfaces, it was evident that the greater grip and friction that could be generated would predispose to greater force transmission (Gaines and La, 1986; Woude et al., 2003). Therefore, a force application investigation under dynamic pushing conditions specific to the movements of WCR should provide a useful insight into the impacts observed directly at the wheelchair-user interface and the effectiveness of force transferred to the wheels when using different types of gloves. An EMG analysis of the muscular activity observed around the wrist, elbow and shoulder joints would also be of value. The strain these muscles are under would offer information concerning the likelihood of injury prevalence and as a result should provide a more detailed evaluation of each glove’s protective properties. It could be hypothesised that a glove that can minimise the impact force and require less grip effort may improve the efficiency of propulsion for WCR players that could also translate into active daily living.

In order to assist manufacturers with the design and development of gloves specific to the impairment and propulsion technique of individual users, more needs to be known about the exact location of pressure distributed on the hands during dynamic wheelchair propulsion. Through the use of force sensing transducers located at various points on the hand, the specific areas of force and pressure distribution could be identified. This is a method which has been employed looking at the plantar pressure distribution witnessed during running manoeuvres to help assist the development of sports footwear (Wong et al., 2007). This should assist manufacturers with identifying which areas of a glove are in need of reinforcement. To support this, a better understanding of which materials generate the greatest amount of friction with the materials of the hand-rim is also desirable to inform manufacturers of the materials needed to reinforce these gloves with. This would ultimately serve to optimise the coupling between the hand and hand-rims, which should ultimately lead to improved mobility performance of WCR players.

4.6 CONCLUSIONS

Glove performance was significantly improved when wearing gloves that have been modified and adapted by the players for the specific demands of WCR. However, there are numerous possible explanations as to why this may be and subsequently further research would be advisable particularly to establish the role of the various hand-rim configurations and their interaction with different glove types. A glove designed specifically for the demands of WCR could still be advantageous, yet further research would again be necessary, as different gloves with different properties maybe needed to suit the noticeably different propulsion techniques between HP and LP. However, until this can be established, the results of the current study would encourage players to modify their ‘off the shelf’ gloves in a way in which they feel best suits their individual needs. Since coupling between the hand and hand-rim is also of relevance to daily life wheelchair propulsion, the findings of the current investigation may be of value to non-wheelchair athletes to help minimise the overuse injuries that these users often experience (Gellman et al., 1988) and to potentially improve the efficiency of wheelchair propulsion.

Chapter Five

The Effects of Camber on the Ergonomics of Propulsion in Wheelchair Athletes

Mason BS, Woude van der LHV, Groot de S & Goosey-Tolfrey VL (2010). The effects of camber on the ergonomics of propulsion in wheelchair athletes. *Medicine & Science in Sports & Exercise*, 43, 319-326.

5.1 ABSTRACT

Purpose: To examine the effects of rear-wheel camber on the physiological and biomechanical responses during hand-rim wheelchair propulsion in highly trained wheelchair athletes. ***Methods:*** Participants (N = 14) pushed on a motor driven treadmill (2.2 $m \cdot s^{-1}$, 0.7% gradient) in four standardised camber conditions (15°, 18°, 20°, 24°). Standardisation was achieved by controlling seat height, the distance between TDC of the main wheels and toe-in toe-out across all camber settings. External P_O and cardio-respiratory measures were collected for each camber setting. Three-dimensional upper body joint kinematics were analysed via two high speed video cameras (100 Hz). One-way ANOVA with repeated measures were applied to all data with statistical significance accepted at $P < 0.05$. ***Results:*** Significantly higher P_O was observed for 24° camber (24.3 ± 5.4 W) in relation to 15° (20.3 ± 4.0 W) and 18° (21.3 ± 4.4 W) and also for 20° (23.3 ± 5.3 W) in relation to 15°. This resulted in improved ME for both 24° (6.8 ± 1.4%) and 20° (6.7 ± 1.0%) compared to 15° (5.9 ± 0.7%). However, significantly higher $\dot{V}O_2$ (reduced economy) and HR responses were observed for 24° (1.04 ± 0.22 $L \cdot min^{-1}$; 105 ± 10 $beats \cdot min^{-1}$) compared to 15° (0.98 ± 0.18 $L \cdot min^{-1}$; 102 ± 9 $beats \cdot min^{-1}$) and 18° (0.97 ± 0.21 $L \cdot min^{-1}$; 102 ± 9 $beats \cdot min^{-1}$). Significantly greater RoM were observed for the degree of shoulder flexion and elbow extension during the push phase for 24°, which was likely to have contributed towards the reduced economy. ***Conclusions:*** Larger camber (20° and 24°) improved the ME of wheelchair propulsion in highly trained wheelchair athletes, yet increased external power requirements and reduced pushing economy.

5.2 INTRODUCTION

Rear-wheel camber is now a particularly common feature of wheelchairs used within the wheelchair court sports. Camber has been defined as the angle of the main wheels in relation to the vertical, whereby the distance between the top points of the main wheels is less than the distance between the bottom points (Higgs, 1983; Frank and Abel, 1993). The mechanical benefits derived from camber have been well documented, with the wider wheelbase thought to provide greater stability and improved hand protection (Trudel et al., 1995; Trudel et al., 1997; Cooper, 1998; Woude et al., 2001). However, the effects of camber on dynamic aspects of mobility performance are slightly more ambiguous, as documented in Chapter 3.

From an ergonomic perspective there has been a paucity of research that has focused on the adaptations that occur as a result of camber adjustments, with varying results reported. The biomechanical responses to camber adjustments have received a limited amount of research attention. Alterations in temporal parameters associated with wheelchair propulsion have been observed with increasing camber (Veeger et al., 1989b; Faupin et al., 2004), as has the degree of shoulder abduction exhibited (Veeger et al., 1989b). From a physiological perspective, $\dot{V}O_2$ and HR responses have been shown to increase with increasing camber during sub-maximal wheelchair propulsion (Buckley and Bhambhani, 1998), whilst no significant effect of camber on these physiological variables has been identified by others (Veeger et al., 1989b; Perdios et al., 2007).

The differences in the physiological responses that have previously been identified as a result of camber adjustments were likely to result from the different methodologies utilised, particularly surrounding the standardisation methods applied. Increasing camber can directly influence other areas of wheelchair configuration and in order to establish reliable cause and effect relationships between camber and mobility performance, these areas need to be strictly standardised. For instance, manipulating camber can alter the position of the shoulder in relation to TDC of the wheel, which means that sitting position may influence the results. However, only one previous study has accounted for this and controlled seat height between camber settings (Veeger et al., 1989b). The distance between TDC of the main wheels will also differ between camber settings if camber bars of fixed lengths have been utilised, yet this too has only been considered by one previous investigation (Faupin et al., 2004). Finally, the alignment of the main wheels in the transverse plane, often referred to as “toe-in toe-out” (Frank and Abel, 1993; Ball, 1994; Cooper, 1998) needs to be constant across camber

settings, since as little as 2° of toe-in toe-out has been shown to double the rolling resistance experienced during wheelchair propulsion (O'Reagan et al., 1981).

Disparity regarding the effects of camber on rolling resistance that have previously been reported would appear to be strongly related to the standardisation methods used, particularly relating to wheel alignment. Small, yet significant decreases in rolling resistance with increasing camber have been reported when toe-in toe-out was not controlled for (Veeger et al., 1989b). Alternatively, residual torque has been shown to increase significantly with camber, thus demanding a higher P_O from the user, when this variable has been controlled between camber settings (Faupin et al., 2004). The greater workload caused by a higher rolling resistance could explain the different physiological responses to camber, given the positive linear relationship that exists between P_O and both $\dot{V}O_2$ and HR (Woude et al., 1988a). Veeger at al. (1989b) previously kept P_O constant between camber settings and although differences in upper body joint kinematics were still observed, no differences in physiological demand were established. However, controlling P_O is not an advisable approach from a sporting perspective, given that P_O cannot be controlled by players on court, it subsequently fails to accurately simulate over-ground propulsion.

The demands of the wheelchair court sports are characterised by aerobic activity, interspersed with bouts of high intensity propulsion (Coutts, 1992; Bloxham et al., 2001; Vanlandewijck et al., 2001; Goosey-Tolfrey et al., 2006). It has recently been established that the mean velocity and distances covered by a group of elite WCR players decreases in the second half of competitive matches (Sarro et al., 2010). Therefore fatigue would appear to be a contributing factor towards performance. Subsequently, a wheelchair configuration that could improve the efficiency and/or economy of propulsion may be a valuable commodity to these athletes.

Sport-specificity has been a significant omission from previous research into the effects of camber on the ergonomics of wheelchair performance, as the majority of studies have incorporated inexperienced, AB participants (Veeger et al., 1989b; Buckley and Bhambhani, 1998; Perdios et al., 2007). In addition, the range of camber angles previously investigated has been limited between 0° and 15° (Veeger et al., 1989b; Buckley and Bhambhani, 1998; Faupin et al., 2004; Perdios et al., 2007). Whereas the camber angles selected by wheelchair athletes engaged in court sports currently range from 15° to as much as 24° (Chapter 3). Also, only Faupin et al. (2004) have incorporated the use of wheelchair athletes when evaluating the effects of camber. However, despite this study controlling the

distances between TDC and monitoring toe-in toe-out between camber settings, the incidental variations in sitting position were not accounted for. Moreover, this investigation was conducted on a roller WERG, which is not as representative of over-ground propulsion since the rolling resistances can be amplified (Faupin et al., 2008).

The aim of the present study was to examine the effects of sport specific and standardised rear-wheel camber settings on the physiological and biomechanical responses elicited during wheelchair propulsion in highly trained wheelchair athletes. Given the increases in P_O that have previously been observed with increases in camber (Faupin et al., 2004), P_O was expected to increase with camber in the present investigation. It was hypothesised that this would lead to an associated increase in physiological demand, particularly in the most extreme setting. It was anticipated that the biomechanical data could assist the interpretation of any physiological adaptations resulting from camber adjustments.

5.3 METHODS

5.3.1 Participants

An *a-priori* power analysis based on the data of Faupin et al. (2004) recommended a sample size of $N \geq 12$. Fourteen highly trained wheelchair athletes (male, $N = 11$; female, $N = 3$; age 23 ± 6 yr; mass 66.9 ± 14.3 kg; playing experience 8 ± 5 yr) volunteered to participate in the current study. Participants competed in either WCB ($N = 11$) or WCT ($N = 3$) with impairments ranging from lower limb amputees to an SCI no higher than T9. Participants self-selected camber settings ranged from 16° to 22°. Procedures for the current investigation were approved by the University's Ethical Committee and all participants provided their written informed consent prior to testing.

5.3.2 Experimental design

Participants were tested in an adjustable sports wheelchair (Top End Transformer, Invacare: mass 11.6 kg; wheel size 0.635 m; tyre pressure 120 psi) in four standardised camber settings (15°, 18°, 20° and 24°), which took place in a randomised, counter-balanced order. The degree of elbow extension each participant elicited when sitting upright in their own sports wheelchair with their hands at TDC was measured (Veeger et al., 1989b). Minor adjustments were made to the seat height to replicate this degree of elbow extension for each participant across camber settings in the adjustable wheelchair. Each camber bar differed in length, enabling the distance between TDC of both wheels to be kept constant (0.48 m) across

camber settings. Toe-in toe-out was also strictly monitored using a set of alignment gauges (RGK Wheelchairs, Cannock, UK) to ensure that wheel alignment was correct and consistent between each setting.

Testing was conducted on a motor driven treadmill (H/P/Cosmos Saturn, Nussdorf-Traunstein, Germany) and prior to each experimental trial, P_O was calculated for each camber setting using a separate drag test (Woude et al., 1986). The drag test set-up consisted of a strain gauge force transducer, attached at the front of the treadmill to the front of the wheelchair. Participants were instructed to remain stationary whilst the treadmill was raised over a series of gradients at a constant velocity (v). The drag test measures the drag force (*Fdrag*) exerted, which was the sum of the rolling resistance, internal friction and gravitational force experienced and enables P_O for a given speed and slope to be determined through the following equation:

$$P_O\,(W) = Fdrag \cdot v \qquad \text{(Woude et al., 2001)}$$

5.3.3 Physiological measures

The four wheelchair propulsion trials consisted of 4-minute bouts on the treadmill at a constant speed and gradient (2.2 $m \cdot s^{-1}$; 0.7%) in each camber setting. During the final minute expired air was collected using the Douglas bag technique (Cranlea, Birmingham, U.K.) and HR was recorded using radio telemetry (PE4000 Polar Sport Tester, Kempele, Finland). On completion of each trial, RPE was obtained for localised, centralised and overall fatigue using the Borg scale (Borg, 1970). Fifteen minutes of recovery was allowed between each trial.

Expired air was analysed for oxygen and carbon dioxide concentrations (Servomex 1400 Gas Analyser, Sussex, UK) and volume (Harvard Dry Gas Meter, Harvard Apparatus, Kent, UK). Oxygen uptake, expired minute ventilation (V_E) and respiratory exchange ratios (RER) were calculated for each camber setting. Heart rate was collected and averaged over 5-second intervals during the final minute. Gross mechanical efficiency was calculated as the ratio of the external work produced to the energy expended (Whipp and Wasserman, 1969). External work was derived from the drag test and energy expenditure was calculated using $\dot{V}O_2$ and the oxygen energetic equivalent of the RER values obtained through a standard conversion table (Peronnet and Massicotte, 1991).

5.3.4 Biomechanical measures

Two synchronized, high speed (100 Hz) gigabit ethernet video cameras (Basler piA640-210gc) were used to capture footage during each camber trial. Cameras were positioned to the left of the treadmill, 4 m back from the performance area to give an optical axis of approximately 60°. Both cameras were calibrated using a 3D 20-point calibration frame with known coordinates (1.0 x 1.4 x 1.4 m). Seven reflective joint markers (19 mm diameter) were placed on five anatomical landmarks (C7 – neck, acromion process – shoulder, lateral epicondyle – elbow, midpoint of radio-ulna styloid process – wrist, 3rd metacarpalphalangeal [MCP] joint – finger) on participant's left hand side by the same investigator, with two markers located on the wheelchair (seat base-backrest intersect and rear-wheel axle). Ten seconds of video footage was collected for each camber setting, mid-way through each 4-minute trial.

A 3D motion analysis system (SIMI Reality Motion Systems, Unterschleissheim, Germany) was used to digitise each of the joint markers from both cameras. A total of five complete push cycles were digitised by the same experienced investigator, with the middle three push cycles used for analysis to avoid inaccuracies with the extreme data points (Challis et al., 1997). Data was analysed using a low-pass 2nd order Butterworth filter with a 6 Hz cut-off frequency (Cooper et al., 2002). The direct linear transformation method was used to transform the 2D coordinates of each camera into 3D coordinates (Abdel-Aziz and Karara, 1971). The digitisation process was repeated for one propulsion cycle of a randomly selected participant to assess the mean experimental error. The mean error between the location of the x, y and z coordinates of the markers was 0.8 mm.

Key events during the propulsion cycle were identified from the biomechanical analysis including HCon and HRel. These events were established using visual identification of the 3rd MCP marker in relation to the hand-rim and enabled the following temporal parameters to be determined. Push time defined the time in which the hand was in contact with the hand-rim (HCon to HRel). Recovery time was characterised by the time in which the hand was not in contact with the hand-rim (HRel to HCon). Cycle time described one complete propulsion cycle (HCon to HCon) and was the sum of PT and RT. Push angle was also derived from the kinematic data using the position of the 3rd MCP at HCon and HRel in relation to the rear-wheel axle marker. Push angle was further divided into start angle (SA - angle created by the 3rd MCP at HCon and TDC in relation to the rear-wheel axle marker) and end angle (EA - angle created between 3rd MCP at TDC and HRel in relation to the rear-wheel axle marker).

The angular displacement of the upper body segments was also investigated in reference to the anatomical position, whereby the arms were positioned by the side of the body and the palms of the hands facing inwards. Shoulder flexion, abduction, elbow and trunk flexion and wrist extension were represented by positive displacement values. Trunk motion was determined as the angle created between the C7, seat base-backrest intersect marker and the vertical in the sagittal plane. The linear velocity of the wrist was also monitored throughout the cycle. For analysis purposes, the cycle was divided into two phases, the push phase and the recovery phase, to identify where any effects of camber were occurring. The angular displacements of all joints were investigated at HCon and HRel. The maximum and minimum displacements were also obtained during both phases and the RoM was calculated using these values.

5.3.5 Statistical analyses

Means and standard deviations (SD) were calculated for all variables. The Statistical Package for Social Sciences (SPSS; Chicago, IL, USA) was used for all statistical analyses. Shapiro Wilks' tests were conducted on all variables to check the distribution of the data. Due to the parametric nature of the data, a series of one-way ANOVA with repeated measures were conducted to examine any differences for all physiological and biomechanical variables between camber settings. Since 24° was hypothesised as being the least ergonomically favourable camber setting, planned simple contrasts were used to compare this setting to all others. Pairwise comparisons with a Sidak adjustment were utilised as further post-hoc tests to identify any significant differences between other camber settings. A follow up analysis was conducted to establish whether any significant differences were the direct result of camber changes or due to any changes in P_O associated with these changes. Subsequently, any variable that was shown to be significantly affected by camber was then normalised by P_O, with the statistical tests repeated. Statistical significance was accepted at $P < 0.05$. Effect sizes were calculated to determine the meaningfulness of any differences, whereby an r value > 0.5 reflected a large effect (Cohen, 1992).

5.4 RESULTS

As demonstrated in Figure 5.1, increases in camber led to a higher P_O, with 24° (24.3 ± 5.4 W) shown to elicit a significantly greater workload than 15° (20.3 ± 4.0 W; $P = 0.004$; r

= 0.69) and 18° (21.3 ± 4.4 W; P = 0.006; r = 0.67). Pairwise comparisons also revealed a significantly higher P_O for 20° (23.3 ± 5.3 W) than in 15° (P = 0.013).

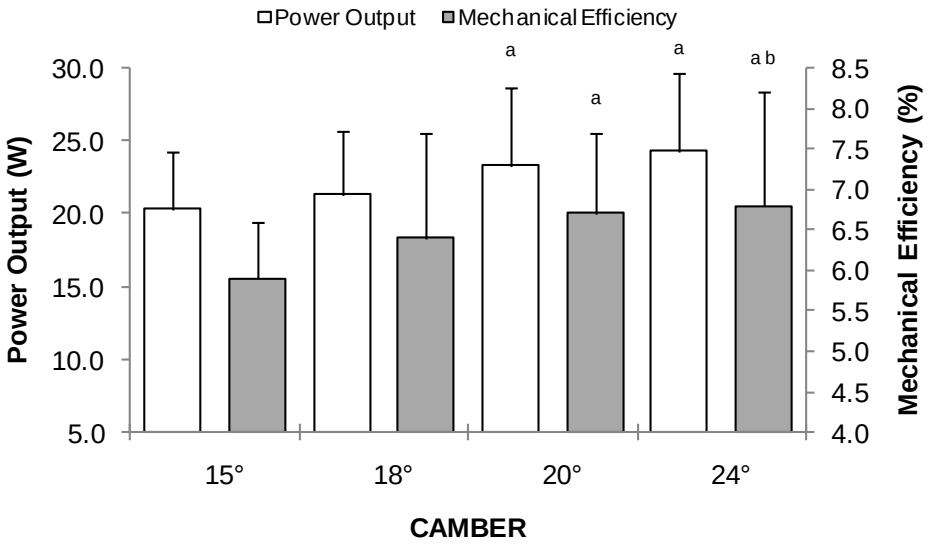

Figure 5.1 – Mean (± SD) power output and mechanical efficiency values across camber settings. [a] denotes a significant difference to 15°; [b] denotes a significant difference to 18°, *P* < 0.05.

5.4.1 Physiological measures

Increasing camber also resulted in ME improvements (Figure 5.1), with significantly lower values displayed for 15° (5.9 ± 0.7%) compared with 20° (6.7 ± 1.0%; P = 0.017) and 24° (6.8 ± 1.4%; P = 0.045; r = 0.53). Heart rate (P = 0.009) and $\dot{V}O_2$ (P = 0.037) were also both significantly increased with greater camber (Figure 5.2). Heart rate responses followed an identical trend to P_O with regards to camber adjustments, with 24° (105 ± 10 beats·min^{-1}) shown to elicit significantly higher values than 15° (102 ± 9 beats·min^{-1}; P = 0.032; r = 0.56) and 18° (102 ± 9 beats·min^{-1}; P = 0.030; r = 0.56) with 20° (106 ± 12 beats·min^{-1}) also significantly higher than 15° (P = 0.048). Oxygen uptake was significantly higher for 24° (1.04 ± 0.22 L·min^{-1}) than both 15° (0.98 ± 0.18 L·min^{-1}; P = 0.019; r = 0.59) and 18° (0.97 ± 0.21 L·min^{-1}; P = 0.043; r = 0.53). However, ME was shown to increase in the 20° and 24° settings in comparison to 15° due to a greater increase in P_O in these settings as opposed to $\dot{V}O_2$ (Table 5.1). Camber was not found to have any significant bearing on localised (P = 0.239), centralised (P = 0.422) or overall RPE (P = 0.264).

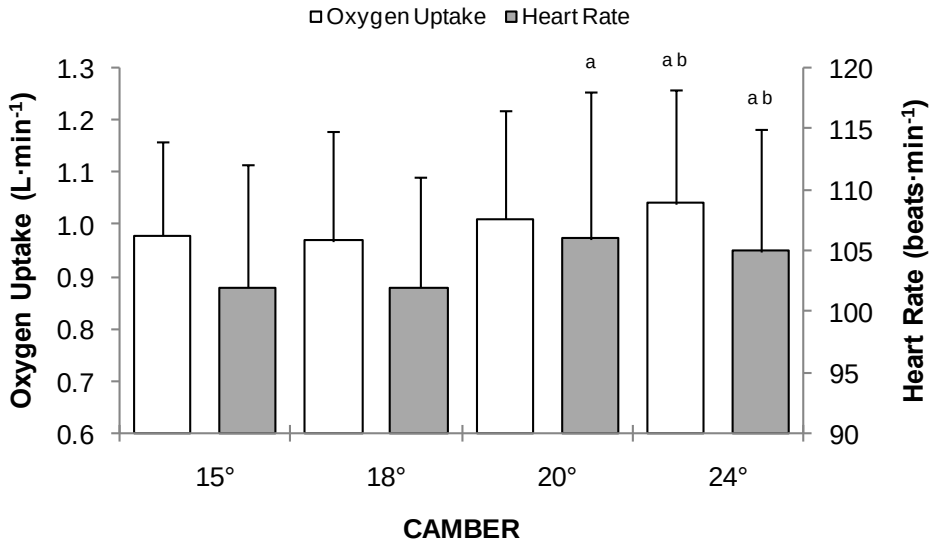

Figure 5.2 – Mean (± SD) oxygen uptake and heart rate responses to different camber settings. [a] significantly higher responses compared to 15°. [b] significantly higher responses to 18°, *P* < 0.05.

Table 5.1 – Relative increases (%) in mechanical efficiency, power output and oxygen uptake in relation to the 15° camber setting.

	15° to 20°	**15° to 24°**
ME (%)	11.4	10.4
P_O (W)	11.7	14.1
$\dot{V}O_2$ (L·min^{-1})	2.7	5.4

5.4.2 Biomechanical measures

The temporal parameters of wheelchair propulsion (PT, RT and CT) were not significantly affected by camber and neither were PA, SA, EA or the push frequencies (Table 5.2).

Table 5.2 – Effects of camber on temporal parameters. Values are means (± SD).

	***P* value**	**15°**	**18°**	**20°**	**24°**
Push times (s)	0.797	0.24 (0.04)	0.25 (0.04)	0.26 (0.03)	0.26 (0.03)
Recovery times (s)	0.709	0.78 (0.11)	0.80 (0.14)	0.77 (0.10)	0.79 (0.11)
Cycle times (s)	0.682	1.03 (0.14)	1.05 (0.16)	1.03 (0.12)	1.05 (0.13)
Start angles (°)	0.424	18.8 (8.3)	20.0 (6.8)	18.9 (8.2)	20.5 (8.6)
End angles (°)	0.246	62.8 (6.6)	62.7 (5.0)	65.3 (7.1)	64.3 (5.2)
Push angles (°)	0.121	81.6 (12.0)	82.7 (10.0)	84.2 (9.2)	84.8 (9.6)
Push frequency (strokes·min^{-1})	0.438	54.6 (11.1)	55.0 (9.9)	55.6 (9.8)	53.4 (9.2)

The biomechanical analysis also revealed that camber adjustments had no significant bearing on upper body joint kinematics at HCon or HRel (Table 5.3). However, the RoM of the shoulder and elbow in the sagittal plane differed significantly between camber settings during the push phase (Table 5.4). The degree of shoulder flexion was significantly greater for 24° in relation to 15° (P = 0.004, r = 0.76), 18° (P = 0.003, r = 0.77) and 20° (P = 0.006, r = 0.74). The difference in shoulder flexion RoM between 15° and 24° appeared to be due to significant differences (P = 0.048, r = 0.58) in the minimum angles displayed at this joint in these settings. The 24° setting induced a more extended shoulder position (-49.5 ± 8.4°) in relation to the 15° setting (-45.9 ± 5.7°). A larger RoM was also evident at the elbow for 24° in relation to 15° (P = 0.001, r = 0.82) and 18° (P = 0.023, r = 0.65) during the push phase. This difference between the 15° and 24° setting could be explained by the significant differences in maximum flexion angles that were revealed between these camber settings (P = 0.041, r = 0.59). Greater degrees of elbow flexion were found for 24° (89.9 ± 10.6°) in relation to 15° (86.5 ± 9.9°). No further differences in RoM, maximum or minimum angular displacements were evident for any other joints between camber settings during the push phase. Camber setting was also shown to have no significant bearing on any joint's RoM, maximum or minimum angular displacement values during the recovery phase of the propulsion cycle.

Table 5.3 – Upper body joint kinematics at hand contact and hand release across camber settings. Values are means (± SD).

	P value	15°	18°	20°	24°
HCon:					
Shoulder flexion (°)	0.188	-46.5 (5.7)	-47.2 (8.5)	-49.7 (7.6)	-48.9 (8.7)
Shoulder abduction (°)	0.197	25.6 (8.7)	25.2 (10.5)	23.8 (9.6)	23.1 (10.3)
Elbow extension (°)	0.467	108.6 (8.7)	108.6 (8.7)	108.4 (10.4)	107.4 (7.8)
Trunk (°)	0.368	0.5 (5.4)	-0.2 (7.5)	-0.5 (6.7)	1.0 (7.3)
Wrist (°)	0.917	40.2 (8.4)	41.1 (8.5)	41.1 (10.9)	40.7 (9.3)
HRel:					
Shoulder flexion (°)	0.198	3.8 (7.6)	5.7 (6.9)	3.7 (8.9)	6.7 (9.0)
Shoulder abduction (°)	0.268	20.8 (4.3)	21.8 (3.8)	21.7 (4.2)	21.9 (3.7)
Elbow extension (°)	0.171	143.8 (5.9)	145.1 (7.4)	145.4 (7.0)	146.8 (5.7)
Trunk (°)	0.489	-0.6 (7.4)	0.6 (8.7)	-0.6 (8.5)	1.4 (8.8)
Wrist (°)	0.095	28.2 (8.5)	30.5 (8.1)	29.9 (8.8)	30.5 (7.0)

Table 5.4 – Active range of motion of upper body joints across camber settings. Values are means (± SD).

	P value	15°	18°	20°	24°
Shoulder flexion (°)	0.009	48.1 (8.4)	50.6 (9.8)	50.3 (9.6)	53.7 (10.7)[a,b,c]
Shoulder abduction (°)	0.985	10.7 (5.8)	11.0 (4.0)	11.1 (3.5)	10.8 (3.6)
Elbow extension (°)	0.020	47.1 (11.7)	48.9 (10.1)	49.8 (9.6)	52.3 (11.9)[a,b]
Trunk (°)	0.715	3.0 (2.8)	2.8 (2.1)	2.5 (1.2)	2.1 (1.3)
Wrist (°)	0.799	18.3 (10.4)	17.2 (10.7)	18.6 (12.2)	17.1 (8.5)

Key: [a] significantly different to 15°
[b] significantly different to 18°
[c] significantly different to 20°

Normalising the variables that were originally shown to be significantly affected by camber by P_O revealed that $\dot{V}O_2$ (P = 0.061) and HR (P = 0.072) were no longer statistically significant. To an even greater extent, the RoM of the shoulder (P = 0.501) and elbow (P = 0.584) in the sagittal plane during the push phase were also no longer affected by camber when normalised by P_O.

5.5 DISCUSSION

This investigation revealed that the range of camber settings currently used in the wheelchair court sports can have a significant bearing on the ergonomics of wheelchair performance in a group of highly trained wheelchair athletes. The results supported the primary hypothesis that when other areas of wheelchair configuration were standardised, P_O still increased with camber, as demonstrated by the significantly higher values for 24° in comparison to 15° and 18° and also for 20° in relation to 15°. This relationship was in accordance with the findings of Faupin et al. (2004), as opposed to Veeger et al. (1989b) and subsequently highlighted the importance of controlling toe-in toe-out between camber settings. The observed increases in P_O are a direct product of an increased *Fdrag*. It has previously been stated that in un-standardised camber settings ranging from 0° to 10°, that camber increments had little effect on rolling resistance (O'Reagan et al., 1981). However, under more extreme and more standardised camber settings, the current study has opposed these findings. It was anticipated that the greater *Fdrag* experienced with increasing camber was due to two potential factors that have previously been proposed (Veeger et al., 1989b; Faupin et al., 2004). Firstly, an increase in internal friction is likely, due to the increased loading and stress on the bearings of the main wheels. In addition to this, the increased loading in greater camber is likely to cause an increased tyre deformation, culminating in a greater contact area between the tyre and the ground and creating a larger rolling friction and thus *Fdrag* (Veeger et al., 1989b; Faupin et al., 2004).

The main finding to emerge from the physiological investigation into rear-wheel camber was that ME improved with greater camber, yet this also resulted in increased $\dot{V}O_2$ (reduced economy) and HR responses. The greater ME revealed in both 20° and 24° camber in relation to 15° was due to the greater relative increases in P_O than $\dot{V}O_2$ with camber. Changes in physiological responses were largely influenced by the associated changes in P_O given that $\dot{V}O_2$ and HR were no longer significantly affected by camber setting when they were normalised by P_O. Yet the fact that both these variables still approached statistical significance (P = 0.061 and 0.072 respectively) demonstrated that other factors aside from increased P_O contributed to the greater physiological demand in the greater camber settings. Associated changes in the upper body joint kinematics with increased camber, as observed in the current investigation and previously by Veeger et al. (1989b) appeared to support this. The physiological findings from the current investigation differed substantially to previous investigations into rear-wheel camber (Veeger et al., 1989b; Buckley and Bhambhani, 1998;

Perdios et al., 2007). However, as previously alluded to, these studies differed in terms of methodology, through a combination of reduced camber angles, slower speeds, inexperienced AB participants and in one instance, fixed P_O (Veeger et al., 1989b). This was thought to be the primary reason for the differences observed in physiological responses to camber, as the current study was the first to test a sport-specific range of camber angles in an athletic, disabled population.

Caution must still be exercised when interpreting the results of the physiological data and assessing their implications for wheelchair athletes during game play. It was potentially valuable to recognise that 20° and 24° camber settings can improve ME during sub-maximal bouts of propulsion due to the higher P_O. Alternatively though, it is clear that pushing economy is reduced in these settings given that $\dot{V}O_2$ increased whilst the speeds remained constant. It has previously been stated in the running and cycling literature that successful performance in these endurance based sports are more dependent on economy, as opposed to the work done (Conley and Krahenbuhl, 1980; Cavanagh and Kram, 1985). This may also be true of the wheelchair court sports and therefore being more efficient in a greater camber configuration maybe of little use to athletes if their $\dot{V}O_2$ also increases, as oxygen resources would be depleted quicker and the onset of fatigue would potentially occur sooner.

Significant differences in upper body joint kinematics were revealed between the two extreme camber settings. The flexion/extension RoM at the shoulder and the elbow were increased during the push phase for the 24° setting in comparison to 15°. The greater active RoM for shoulder flexion and elbow extension in 24° camber may explain the reduced economy in this setting. It was likely that an increased activation of the muscles responsible for these movements was instigated over the larger RoM and that this contributed to the increased $\dot{V}O_2$ and HR responses. This was based on the fact that the temporal parameters of propulsion were unaffected by camber. Therefore, it would appear as though force was being applied over similar ranges and times across camber settings, yet due to the greater P_O and RoM in 24° it could be anticipated that the forces applied maybe greater in this setting.

The greater active RoM for shoulder flexion and elbow extension in the 24° camber setting appeared to be strongly associated to the increased P_O in this setting, as indicated by the insignificant results established when these variables were normalised by P_O. The increased active RoM at these joints may have also served as a compensatory motion due to the absence of any variation in shoulder abduction with increasing camber, as initially anticipated. Veeger et al. (1989b) revealed that camber influenced the maximum abduction

values produced. However, the use of inexperienced AB participants and the fact that the distance between TDC of the main wheels was not controlled in this study brings the validity of this finding into question, as this would strongly influence movements occurring in the frontal plane. Subsequently, when camber settings were strictly standardised, it would appear that settings between 15° and 24° do not affect the abduction of the upper arm in highly trained wheelchair athletes during sub-maximal propulsion.

The results of the biomechanical analysis also revealed that camber adjustments had little bearing on trunk or wrist motion. User safety/health is an important ergonomic consideration for adaptive sports equipment. The wrist in particular is an area that is especially prone to injury in wheelchair users, with greater peak extension values having been identified as serious risk factors for carpal tunnel syndrome (Veeger et al., 1998). However, the fact that maximum, minimum and RoM values for the wrist were unaffected by camber, suggested that at the speeds and range of cambers investigated, participants were not being placed at a higher risk of wrist injury. Alternatively, it is possible that the shoulder and elbow may be more susceptible to injury when increasing camber, especially in 24°, as it has been previously suggested that wheelchair users who demonstrate larger joint excursions during the push phase may predispose towards an increased risk of injury at the relevant joints (Finley et al., 2004). However, further investigations would be required to explore this assumption. A kinetic analysis in conjunction with the kinematic analysis could identify the force application patterns during these camber conditions. The use of inverse dynamics could then be applied to calculate the moments occurring at specific joints to inform athletes about their risk of injury when selecting camber settings. It would also be recommended for future investigations to examine these biomechanical responses to the current range of camber settings when performing maximal effort bouts of propulsion when force application will be amplified.

5.6 CONCLUSIONS

The current investigation has provided a detailed insight into some of the physiological and biomechanical adaptations that occur as a result of standardised camber adjustments. At the sub-maximal speeds and the subsequent workloads imposed, ME improved in the greater camber settings. However, due to an associated increase in $\dot{V}O_2$ (reduction in economy), 20° and 24° camber could not be recommended as ergonomically optimal settings. The absence of any curvilinear relationships prevented any optimal camber settings being recommended for the participants as a whole. This was likely to be due to the

vast differences in anthropometrics and impairment levels that existed amongst the current participants and the fact that the self-selected seat heights varied dramatically too. Therefore, it was incredibly unlikely that one camber setting would be optimal for all participants. Subsequently, future investigations would be recommended to group participants based on variables such as sitting position, classification or impairment level in an attempt to identify optimal camber settings and to ascertain how these differ dependent on certain grouping criteria.

The aim of the following study (Chapter 6) was to extend the results of the current investigation by exploring the effects of rear-wheel camber in a field environment under conditions of maximal effort, over-ground propulsion. Only then can a more holistic appraisal of the effects of camber be obtained to provide guidelines that inform athletes about optimising their wheelchair configuration selections. Providing an evidence base for athletes is vital given that the wheelchair selection process has been identified as an incredibly subjective process based on user perceptions (Chapter 3).

Chapter Six

The Effects of Rear-Wheel Camber on Optimal Mobility Performance in Wheelchair Athletes

Mason BS, Woude van der LHV, Tolfrey K & Goosey-Tolfrey VL. An investigation into the role of rear-wheel camber on optimal mobility performance in wheelchair athletes. *International Journal of Sports Medicine*, Under Review.

6.1 ABSTRACT

Purpose: To explore the effects of rear-wheel camber on sport-specific aspects of mobility performance. ***Methods:*** Highly trained wheelchair court sport athletes (N = 14) were assigned to two groups according to self-selected seat height (high and low). Participants performed a battery of field tests in four camber settings (15°, 18°, 20°, 24°) in an adjustable sports wheelchair. A series of performance measures were collected and analysed using a series of two-way ANOVA with repeated measures. Statistical significance was accepted at $P < 0.05$. ***Results:*** Twenty metre sprint times (P = 0.016) and linear mobility times (P = 0.004) were significantly lower in 18° camber (5.89 ± 0.47s; 16.06 ± 0.97s) compared with 24° (6.05 ± 0.45s; 16.62 ± 1.10s). Improved manoeuvrability performance was also evident for 18° compared to 15° (r = 0.72). Significant interactions between camber, seat height and decelerations in between pushes revealed that players with a high seat position experienced an increased rate of deceleration in 24° compared to 15°, whereas those with a low seat height experienced similar rates (P = 0.048; r = 0.62). ***Conclusions:*** It was concluded that one camber setting would not be optimal for all individuals, yet 18° may be a beneficial setting for young, inexperienced players given its superior performance for aspects of both linear and non-linear mobility performance.

6.2 INTRODUCTION

Considerable advances in the design of wheelchairs used for the wheelchair court sports have taken place over recent years, largely owing to improvements in technology, which has seen these chairs become substantially lighter (LaMere and Labanowich, 1984; Yilla, 2004; Ardigo et al., 2005; Burkett, 2010). Specific changes to areas of wheelchair configuration have also been noticeable. An increase in the degree of rear-wheel camber has proven to be one of the most prominent developments in configuration (Cooper, 1998; Ardigo et al., 2005). Despite these developments, very little is known about the influence of camber on aspects of mobility performance specific to the wheelchair court sports, as the focus of previous camber research has centred around daily life propulsive conditions (Veeger et al., 1989b; Buckley and Bhambhani, 1998; Perdios et al., 2007). The qualitative approach adopted in the first experimental study (Chapter 3) revealed that highly experienced wheelchair athletes unanimously reported that greater camber led to improved manoeuvrability performance, yet its impact upon straight line performance yielded mixed responses. Unfortunately, little empirical research exists to verify or refute these claims (Faupin et al., 2002; 2004).

Maximal sprinting performance has been shown to decrease linearly with increases in camber (9°, 12° and 15°), as demonstrated by lower mean velocities during an 8 second sprint (Faupin et al., 2004). The application of these findings to over-ground propulsion may be limited since these results were obtained from wheelchair ergometry. Alternatively, the earlier findings of Faupin and colleagues (2002) over the same range of camber settings, which revealed that 15° improved turning performance without influencing linear performance during over-ground propulsion, would seem more valid. It is important to note that although both these studies focused on the performance of wheelchair athletes, the camber settings investigated were not representative of the 15° to 24° range that is commonly used in the court sports today (Chapter 3). Recent work from our laboratory has revealed that 24° camber significantly increased the P_O requirements for athletes, as well as $\dot{V}O_2$ and HR responses, whilst affecting the active RoM at the shoulder and elbow during sub-maximal propulsion (Chapter 5). However, the effect that this range of camber settings can have on aspects of mobility performance specific to wheelchair sports in a field environment remains unknown.

Large inter-individual variability exists amongst wheelchair athletes in terms of their impairment, classification and anthropometrics (Goosey-Tolfrey and Price, 2010). Subsequently, athletes are frequently seen to adopt different sitting heights in order to

accommodate these differences in physical characteristics and to meet the requirements of their positional role on court (Yilla et al., 1998). However, previous studies investigating rear-wheel camber have often utilised fixed seat heights across camber settings for all participants (Veeger et al., 1989b; Faupin et al., 2004). Given the aforementioned variability that exists between individuals, it was unlikely that one camber setting would be optimal for all athletes. In order to provide guidelines for the optimisation of wheelchair configuration, researchers may need to group participants in order to make the findings relevant to individual's specific physical characteristics. Although at the same time it is vitally important to standardise other areas of wheelchair configuration when manipulating camber in order to identify any direct cause and effect relationships between camber and mobility performance (Chapter 5).

The principal aim of this study was to examine the effects of camber on sport-specific mobility performance in a group of highly trained wheelchair athletes. A secondary aim was to determine whether optimal camber settings could be established for athletes when (a) ungrouped or (b) categorised by self-selected seat height. It was hypothesised that increasing camber may lead to decreased linear performance, especially for 24°, given the greater resistance that has been associated with this setting (Chapter 5). Alternatively, an improvement in manoeuvrability performance was also hypothesised with increasing camber based on previous trends (Faupin et al., 2002).

6.3 METHODS

6.3.1 Participants

Fourteen highly trained wheelchair athletes (age 23 ± 6 yrs, mass 66.9 ± 14.3 kg) who played WCB (N = 11) and WCT (N = 3) volunteered to participate in the study. The range of participants' impairments spanned from lower limb amputees to an SCI no higher than T9. Participants were assigned to two 'seat height' groups based on the angle induced at the elbow when the hands were placed on TDC, when sat in their own sports wheelchair. Using the midpoint of the range of elbow angles elicited, a cut off setting of 105° elbow extension was established. Participants exhibiting elbow extension > 105° were defined as having a high seat height (N = 6), whereas those with ≤ 105° were deemed to have a low seat height (N = 8). Approval for the procedures involved, was obtained from Loughborough University's Ethical Committee and all participants provided their written informed consent prior to testing.

6.3.2 Equipment

Participants were tested in an adjustable sports wheelchair (Top End Transformer, Invacare: mass 11.6 kg, wheel size 0.635 m, tyre pressure 120 psi) in a total of four standardised camber settings (15°, 18°, 20°, 24°). The distance between TDC of both wheels was kept constant (0.48 m) between camber settings, as was tyre pressure. The degree of toe-in toe-out was strictly monitored through the use of alignment gauges (RGK Wheelchairs, Cannock, UK). Seat height was also standardised using the aforementioned degree of elbow extension that was elicited by each participant and this was rigorously replicated across camber settings.

A velocometer (Moss et al., 2003) sampling at 200 Hz was attached to the wheelchair throughout testing. This enabled the intra-push profiles during over-ground manual wheelchair propulsion to be collected for each camber setting. Wireless timing gates (Brower, Utah, USA) were used to record the times taken to perform each of the field tests described later.

6.3.3 Experimental design

Participants performed a battery of field tests that comprised the three drills (adapted from Chapter 4) incorporating movements specific to the wheelchair court sports in a sports hall with wooden sprung flooring in each camber setting. The following drills were completed in the same order yet the camber conditions were randomised and counter-balanced. Drills 1 and 2 were performed once for each camber setting, with Drill 3 repeated twice (once with a left turn at the top and once with a right turn) with the resultant times averaged. Participants were familiarised with all the field tests prior to data collection. Each trial commenced when participants indicated they were ready. Sufficient rest was permitted between trials to ensure that fatigue did not influence the results.

6.3.3.1 Twenty metre sprint

Participants performed a maximal effort linear sprint from a standstill. Performance was assessed by the times taken to perform the drill and the velocities reached. Acceleration performance was calculated over the first two and three pushes. Mean and peak velocity, the number of pushes performed and the mean deceleration in between pushes were also derived from the velocometer traces.

6.3.3.2 Linear mobility

Repeated bouts of acceleration, braking and backwards pulling manoeuvres were performed during this drill. A total of three incremental distance sprints were performed during forwards propulsion (5 m, 7.5 m and 12.5 m). Participants were then required to stop sharply at the end of the first two sprints, prior to backwards pulling over a constant distance of 2.5 m at two points (Figure 6.1). Performance was assessed by the overall times taken to complete the drill. Braking performance was assessed after 5 m and 7.5 m of using the deceleration values calculated from the time at which deceleration commenced until the wheelchair had reached a standstill at point ‘a’ and ‘b’ in Figure 6.1. The mean and peak velocity exhibited during each phase of the forward and backward propulsion sections were also analysed from the velocometer traces.

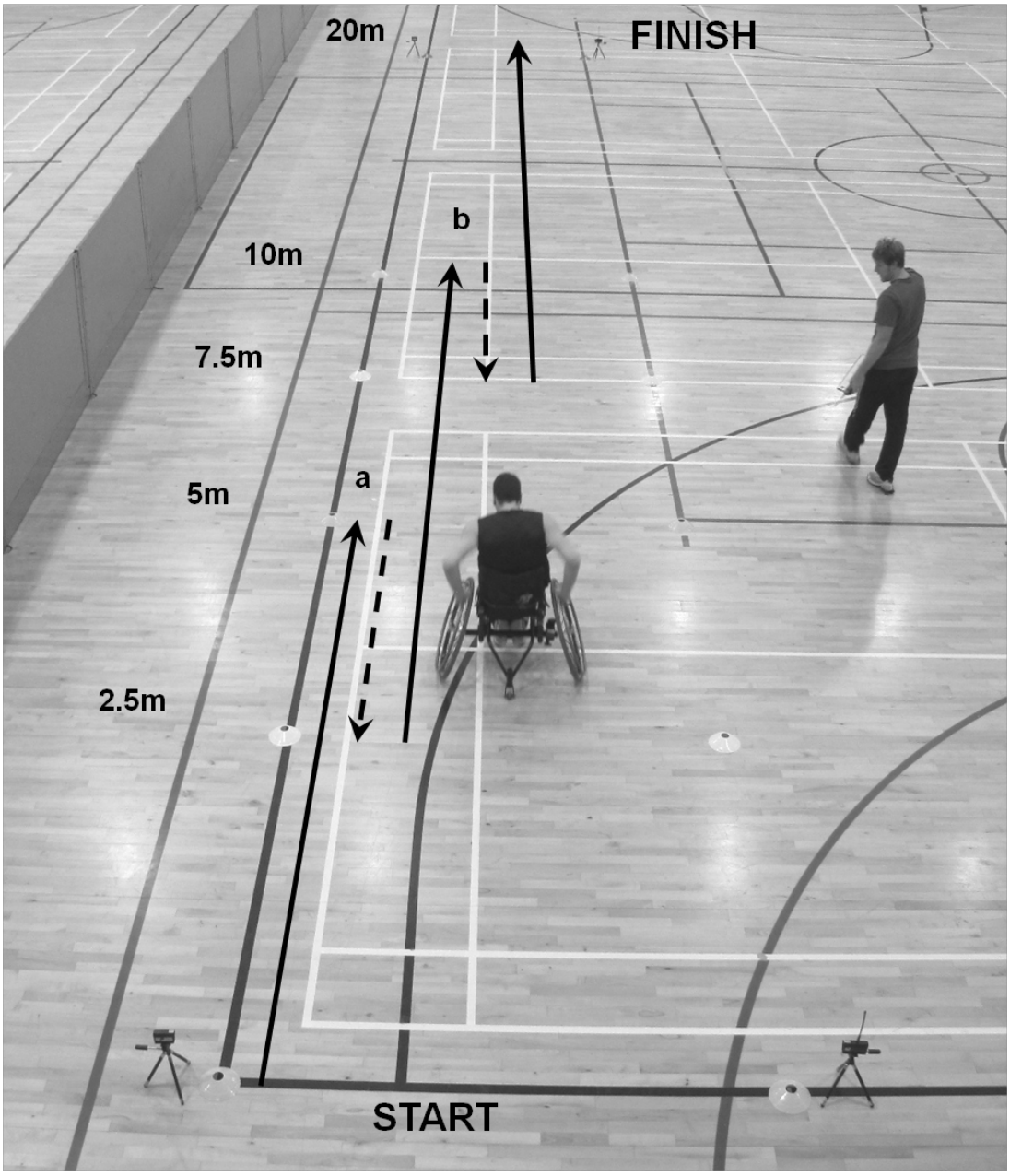

Figure 6.1 – The linear mobility drill (solid arrows represent forward propulsion and broken arrows represent backward pulling).

6.3.3.3 Manoeuvrability

This drill was adapted from Chapter 4 and was divided into two sections to assess both sprinting and manoeuvrability performance, as determined by the times taken. The first section allowed linear times to be established for a 9 m sprint after a 5 m rolling start. The second section commenced on completion of this and after performing a sharp turn (approximately 230°), manoeuvrability performance was measured by the time taken to perform a series of ‘slalom’ movements (Figure 6.2).

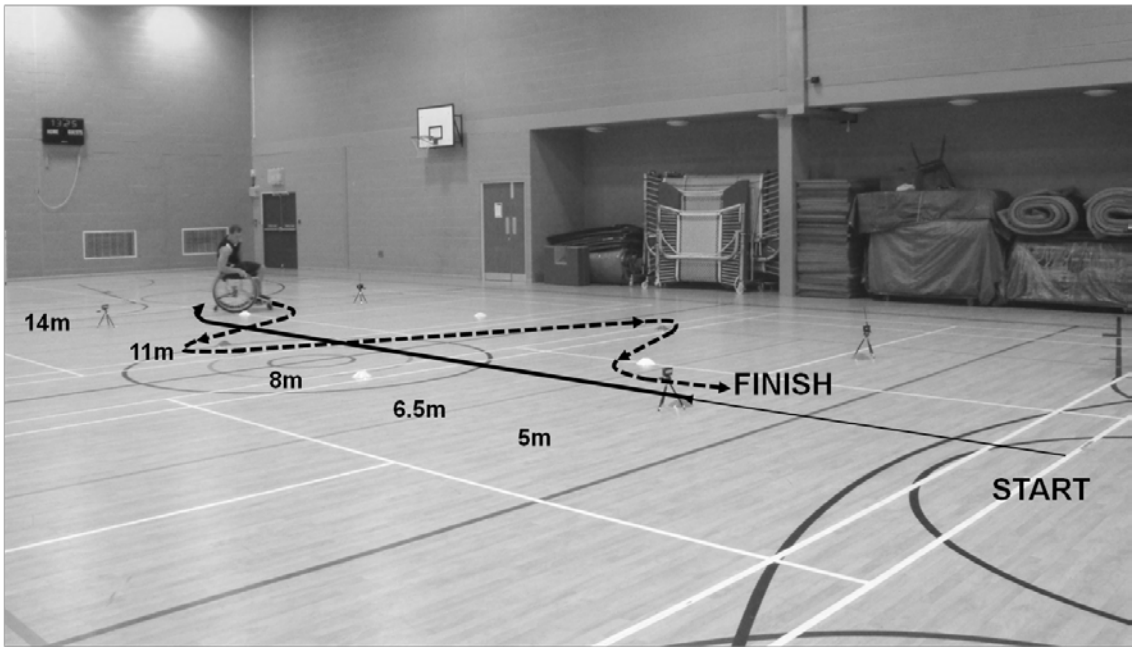

Figure 6.2 – The manoeuvrability drill (solid arrow represents 'linear section' and broken arrow represents 'agility section' of the drill).

6.3.4 Statistical analyses

Means and standard deviations (SD) were computed for all variables. The Statistical Package for Social Sciences (SPPS; Chicago, IL, USA) was used for all the statistical analyses. A two-way ANOVA with repeated measures was conducted on all performance parameters to examine any differences in performance resulting from camber adjustments. Differences between the two seat height groups and any interactions between camber settings, sitting height and performance were also investigated by the ANOVA. Since 24° was hypothesised to perform differently to other camber settings, simple planned contrasts were applied to explore this hypothesis. Effect sizes were calculated to determine the meaningfulness of any differences, whereby $r > 0.5$ reflected a large effect (Cohen, 1992). Pairwise comparisons with a Sidak adjustment were also used to identify any differences between other camber settings. All data were accepted as statistically significant whereby $P < 0.05$.

6.4 RESULTS

The results revealed that camber had a significant effect on certain aspects of mobility performance.

6.4.1 Twenty metre sprint

The time taken to complete the 20 m sprint from a standstill was significantly influenced by camber setting (Figure 6.3). Sprint times were quicker for 18° (5.89 ± 0.47 s; P = 0.016, r = 0.63) and 20° (5.93 ± 0.47 s; P = 0.041, r = 0.55) in comparison to 24° (6.05 ± 0.45 s). Table 6.1 displays the performance parameters derived from the velocometer during the 20 m sprint. It was revealed that significantly higher mean velocities were maintained in the 18° condition compared to the 24° condition (P = 0.006, r = 0.80). The results of the mean decelerations showed that 24° camber decelerated the wheelchair at a greater rate than both the 18° (P = 0.009, r = 0.77) and 20° conditions (P = 0.001, r = 0.89). No effect of camber was observed for the peak velocity reached, the number of pushes performed or initial acceleration over the first two and three pushes (Table 6.1).

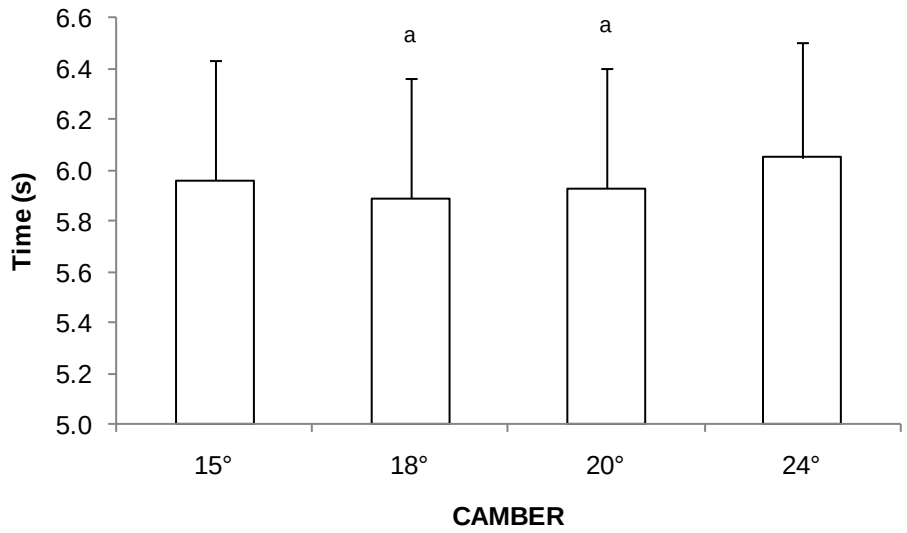

Figure 6.3 – Linear 20 m sprint times in different camber settings. [a] represents a significant difference with the 24° setting.

Table 6.1 – Performance parameters assessed during the linear 20 m sprint. Values are means (± SD).

	***P* value**	**15°**	**18°**	**20°**	**24°**
Mean velocity ($m \cdot s^{-1}$)	0.021	2.92 (0.27)	3.00 (0.26)[a]	2.92 (0.18)	2.85 (0.25)
Peak velocity ($m \cdot s^{-1}$)	0.715	4.45 (0.47)	4.53 (0.51)	4.46 (0.47)	4.48 (0.53)
Mean decelerations ($m \cdot s^{2}$)	0.025	4.52 (1.51)	4.56 (1.95)[a]	4.37 (1.96)[a]	4.95 (1.94)
Number of pushes	0.104	12.4 (1.7)	12.0 (1.8)	12.3 (1.5)	12.5 (1.6)
Acceleration: two pushes ($m \cdot s^{2}$)	0.351	2.09 (0.43)	2.19 (0.52)	2.12 (0.28)	2.07 (0.52)
Acceleration: three pushes ($m \cdot s^{2}$)	0.316	1.80 (0.37)	1.84 (0.37)	1.81 (0.24)	1.75 (0.36)

Key: [a] significantly different to 24°

6.4.2 Linear mobility

The time taken to complete the linear mobility drill was also significantly affected by camber (Figure 6.4). Quicker times were displayed in both 15° (16.09 ± 0.84 s, $P = 0.014$, $r = 0.63$) and 18° (16.06 ± 0.97 s, $P = 0.004$, $r = 0.71$) compared to 24° (16.62 ± 1.10 s). Table 6.2 documents the sub-parameters of performance, as derived from the velocometer, to account for the differences in times taken between conditions. It was revealed that braking performance was not influenced by camber setting and neither was the mean or peak velocity exhibited during the backward propulsion phases. The mean velocity achieved during the phases of forward propulsion was influenced by camber, however, the peak velocity reached in each of these phases was unaffected by camber. The mean velocity during the 1st sprint phase (5 m) was significantly lower for the 24° setting in comparison to 18° ($P = 0.034$, $r = 0.57$). Although not statistically significant, large effect sizes were revealed for the 24° to also elicit lower mean velocities during the 2nd (7.5m) sprint phase compared to 15° ($r = 0.51$) and 18° ($r = 0.53$). A similar pattern was also observed during the 3rd (12.5 m) sprint phase between the 24° condition and both 15° ($r = 0.63$) and 18° camber ($r = 0.53$).

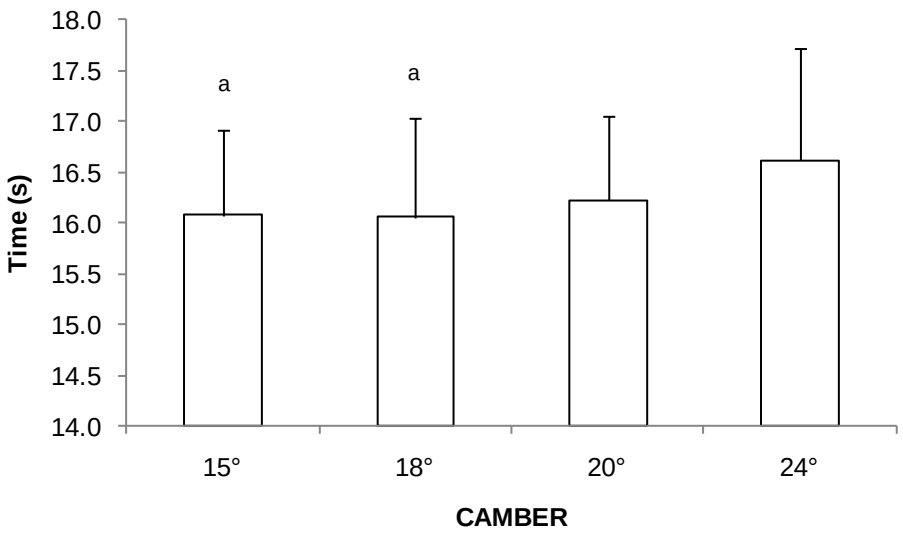

Figure 6.4 – Times taken to complete the linear mobility drill in different camber settings. [a] represents a significant difference with the 24° setting.

Table 6.2 – Performance parameters assessed during the linear mobility drill. All values are means (± SD).

	***P* value**	**15°**	**18°**	**20°**	**24°**
Braking after 5 m ($m \cdot s^{2}$)	0.332	3.45 (0.61)	3.56 (0.86)	3.74 (0.80)	3.52 (0.90)
Braking after 7.5 m ($m \cdot s^{2}$)	0.678	3.26 (0.68)	3.37 (0.62)	3.42 (0.75)	3.29 (1.01)
Mean velocity during 1st sprint ($m \cdot s^{-1}$)	0.006	1.69 (0.15)	1.72 (0.23)[a]	1.55 (0.10)	1.60 (0.18)
Mean velocity during 2nd sprint ($m \cdot s^{-1}$)	0.189	2.07 (0.16)	2.09 (0.15)	2.03 (0.12)	1.98 (0.12)
Mean velocity during 3rd sprint ($m \cdot s^{-1}$)	0.069	2.60 (0.21)	2.56 (0.19)	2.56 (0.13)	2.49 (0.20)
Peak velocity during 1st sprint ($m \cdot s^{-1}$)	0.654	3.14 (0.33)	3.09 (0.34)	3.08 (0.26)	3.13 (0.33)
Peak velocity during 2nd sprint ($m \cdot s^{-1}$)	0.275	3.52 (0.37)	3.50 (0.37)	3.42 (0.27)	3.43 (0.28)
Peak velocity during 3rd sprint ($m \cdot s^{-1}$)	0.568	4.06 (0.47)	3.98 (0.43)	3.95 (0.35)	3.95 (0.41)
Mean velocity during backwards propulsion ($m \cdot s^{-1}$)	0.230	1.27 (0.14)	1.26 (0.16)	1.25 (0.19)	1.22 (0.14)
Peak velocity during backwards propulsion ($m \cdot s^{-1}$)	0.218	2.19 (0.35)	2.12 (0.29)	2.19 (0.35)	2.11 (0.33)

Key: [a] significantly different to 24°

6.4.3 Manoeuvrability

No statistically significant differences in manoeuvrability performance were revealed between camber settings in the manoeuvrability drill, although some strong trends were again evident (Table 6.3). The 18° setting led to slightly quicker times during the linear section of the drill in comparison to 24° and the large effect size suggests this could be meaningful (r = 0.65). This would reiterate the superior linear performance demonstrated for 18° compared to 24° as previously seen during the 20 m sprint and linear mobility drills. The time taken to complete the agility section (r = 0.72) and the overall drill (r = 0.68) were also slightly improved in the 18° setting, this time in relation to the 15° setting.

Table 6.3 – Times taken to complete the manoeuvrability drill across camber settings. Values are means (± SD).

	***P* value**	**15°**	**18°**	**20°**	**24°**
Linear section (s)	0.139	2.59 (0.22)	2.57 (0.22)	2.59 (0.21)	2.62 (0.21)
Agility section (s)	0.372	9.31 (0.52)	9.18 (0.56)	9.11 (0.58)	9.25 (0.68)
Overall (s)	0.369	11.90 (0.67)	11.75 (0.71)	11.69 (0.71)	11.86 (0.82)

Although camber was shown to influence aspects of mobility performance, the between-subject effects revealed that seat height did not have any significant bearing upon any of the performance variables measured. Despite this, a significant interaction (P = 0.032) was identified for the mean decelerations between pushes during the 20 m sprint in the 15° and 24° settings (P = 0.048, r = 0.62). Participants with a low seat height experienced a similar rate of deceleration in between pushes in both these settings, whereas those with a high seat height experienced a greater rate of deceleration in the 24° setting (Figure 6.5). Braking performance after 7.5 m of propulsion in the linear mobility drill (P = 0.090) and the time taken to complete the linear section of the manoeuvrability drill (P = 0.084) also approached statistical significance for an interaction between camber and seat height, again involving the two extreme settings (Table 6.4). Braking performance appeared slightly improved for those with a high seat height in the 24° setting in comparison to 15°, whereby participants with a low seat height demonstrated an opposite trend (r = 0.53). In addition, participants with a high seat position appeared to perform the linear section of the manoeuvrability drill quicker in the 24° setting in comparison to 15°, whereas those with a low seat height again directly opposed this trend (r = 0.55).

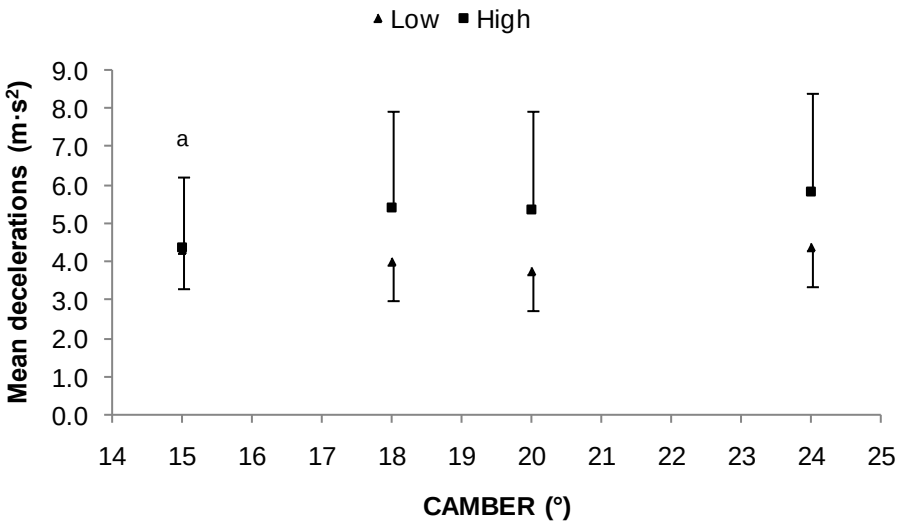

Figure 6.5 – Interactions between camber, seat height and the mean decelerations experienced between pushes during the 20 m sprint. [a] represents a significant interaction between group and the 24° camber setting.

Table 6.4 – Areas of mobility performance demonstrating interactions with camber and seat height. All values presented are means (± SD).

	15°		18°		20°		24°	
	Low	**High**	**Low**	**High**	**Low**	**High**	**Low**	**High**
Braking after 7.5 m ($m \cdot s^2$)	3.30 (0.74)	3.21 (0.67)	3.28 (0.67)	3.48 (0.60)	3.14 (0.52)	3.78 (0.89)	2.91 (0.45)	3.79 (0.36)
Linear section times (s)	2.52 (0.17)	2.70 (0.25)	2.51 (0.18)	2.65 (0.25)	2.54 (0.18)	2.66 (0.24)	2.58 (0.20)	2.67 (0.24)

6.5 DISCUSSION

The results of this investigation somewhat supported the hypothesis, that increasing camber would negatively affect aspects of linear performance. It was evident that 24° was commonly the least effective camber setting for aspects of linear performance. The current study revealed that both 18° and 20° significantly reduced the times taken to perform the 20 m sprint and 18° and 15° reduced the times taken to perform a linear mobility drill in comparison to 24° camber. Neither maximal sprinting nor initial acceleration performance appeared to be responsible for these improved times, since peak velocity reached and acceleration over the first two and three pushes was unaffected by camber during the 20 m sprint. Therefore, the greater *Fdrag* previously experienced in the 24° camber setting (Chapter

5) did not impact on the ability to accelerate or reach high top end speeds. This may have been due to the highly trained status of the participants investigated. Their ability to generate sufficient force to overcome any increases in resistance may have been more substantial than the actual differences in *Fdrag* between camber settings.

Alternatively, it seemed as though decreases in the mean velocity produced during both linear drills were the main cause for the increased times taken in the 24° camber setting. Therefore, the increased *Fdrag* previously identified in this camber setting, may have had more of a bearing during the recovery phase of propulsion given that the wheelchair was shown to decelerate at a greater rate in between pushes in the 24° setting. Based on the consistent number of pushes performed across camber settings, it seemed likely that the greater rate of deceleration in between pushes contributed to the reduced mean velocities and subsequently the increased times taken in the 24° setting.

The lower mean velocity and greater rate of deceleration in between pushes that were associated with 24° camber may support the earlier findings in Chapter 5 relating to the economy of propulsion. It was revealed that pushing economy was reduced in the 24° setting, when tested at a constant, sub-maximal speed on a motor driven treadmill. The fact that the wheelchair decelerates quicker when configured with 24° camber implies that the athlete will have to work harder with each push in order to accelerate the wheelchair. Over the course of a WCB or WCT match, this may trigger an earlier onset of fatigue. It has already been shown that the mean velocity and distance covered by highly trained WCR players significantly decreases during the second half of competitive game play (Sarro et al., 2010). Therefore, if a wheelchair configured with 24° camber could potentially accelerate the onset of fatigue and hinder performance later on in matches, this could be a valuable piece of information to athletes and coaches alike.

Interestingly, no statistically significant differences in agility section times or overall times were revealed between camber settings for the manoeuvrability drill. However, despite the absence of any statistical significance, there were very strong trends, supported by large effect sizes, for 18° camber to allow for quicker agility and overall times than the 15° condition. Speed and Andersen (2000) have stated that when researching elite athletes, small improvements in times may not be statistically significant, yet practically they may be very significant and therefore the use of effect sizes is likely to be of more value. This may be extremely applicable to the elite athletes sampled in the current investigation and subsequently on a practical level manoeuvrability performance may become meaningfully restricted in 15° camber.

During game play these small effects accumulate and may lead to increased deterioration of performance over the course of a competitive match, as previously mentioned. It was also interesting to see that no further improvements in manoeuvrability performance were achieved from selecting camber settings greater than 18°. This differed to the linear relationship revealed by Faupin et al. (2002), whereby increasing camber was associated with improved turning performance. However, differences in the approaches used for determining manoeuvrability performance may also provide an explanation for the different trends identified between studies. The current study measured manoeuvrability performance by the time taken to perform a series of slalom movements to simulate advancing up court and avoiding other opponents. Alternatively, Faupin et al. (2002) quantified manoeuvrability performance by the time taken to pivot around one cone. Both these movements would appear to be extremely relevant to WCB performance in particular (Brasile and Hedrick, 1996). Therefore, it may be that the ability to pivot on the spot does increase linearly with camber and may have done so up to the 24° setting if this had been assessed by the current study. Yet it would appear that more dynamic manoeuvrability performance, as assessed by the current study, does not improve in settings over 18°. Further investigations would be warranted to investigate this assumption and the use of a more comprehensive battery of field tests could be employed in the future. If this proposal was proven to be true then it would appear that players would need to consider which aspect of manoeuvrability performance is most important to them based on their role on court, as to which camber setting may be the most suitable.

Linear relationships have previously been identified between camber and both manoeuvrability (Faupin et al., 2002) and linear performance (Faupin et al., 2004). However, these relationships were established between 9° and 15° camber settings, which are not representative of the ranges that are commonly selected by athletes from the wheelchair court sports or the range investigated by the current investigation. The fact that curvilinear relationships between camber and both linear and non-linear measures of performance were identified, implied that optimal camber settings may exist within the current settings examined. There was a recurring theme for 18° to allow for the greatest linear and non-linear performance during the current investigation. For all significant and meaningful differences established, 18° was always shown to perform greater than 24° for aspects of linear performance and greater than 15° for manoeuvrability performance. This was not to suggest that 18° would be the optimal camber setting for all individuals. The fact that 15° allowed for significantly quicker linear mobility times and 20° allowed for quicker 20 m sprint times, both

in comparison to 24°, emphasises this. However, due to the frequency with which superior performance was associated with this setting could demonstrate that 18° camber may be the best setting to recommend to young or inexperienced athletes. The process of configuring a new sports wheelchair is currently extremely subjective and selections are often based on trial and error (Chapter 3). Therefore, young and inexperienced athletes, who have no previous experience to base their selections on may benefit from this information.

Since it was anticipated that one camber setting was unlikely to be optimal for all individuals, the current investigation grouped participants based on their self-selected seat heights. To the authors' knowledge, other than Chapter 4, no previous investigations have attempted to group athletes when investigating the ergonomics of different wheelchair user-combinations on mobility performance. The current investigation revealed that a significant interaction existed between camber and seat height for the mean decelerations experienced in between pushes during the 20 m sprint. It has already been stated that the rate of decelerations during the recovery phase increased in the 24° camber setting. However, it was also clear that this rate of deceleration was exacerbated for individuals with a high seat height compared to those with a low seat height, who experienced similar decelerations across camber settings. It was proposed that as these individuals often possess greater trunk stability it was possible that a greater degree of trunk extension during the recovery phase was performed. This could increase the rate at which the wheelchair decelerates in between pushes as this movement is occurring in a direction that opposes the direction of propulsion. Therefore, less camber may be advisable for individuals who adopt a high seating position, whereas for those who sit low, it appeared less critical during maximal effort linear sprinting.

Alternatively, interactions approaching statistical significance were also revealed for braking performance after 7.5 m of propulsion (linear mobility drill) and the linear section times of the manoeuvrability drill between the two extreme camber settings. In both instances it was shown that participants with a high seat height were more suited to greater camber than those with a low seat height, who were more effective with less camber. It could be proposed that athletes who adopted a high seat height performed favourably in the larger camber setting due to a greater feeling of stability created by the larger wheelbase (Trudel et al., 1997; Cooper, 1998). The two measures where interactions were observed required braking after maximal effort propulsion (linear mobility drill) and the linear section of the manoeuvrability drill, which was immediately followed by a sharp turn. Therefore, it was possible that confidence concerns about performing these manoeuvres with less camber could have impaired the performance of these measures for those with a high seat due to a combination of

a higher centre of mass and a restricted wheelbase. Future investigations would be recommended to continue to group participants and to establish the most effective means for doing so. This would enhance the specificity of feedback when informing athletes about optimal wheelchair configuration settings. It is possible that severity of impairment may have been a confounding factor towards seat height group in this particular study, as the most severely impaired individuals were also shown to adopt a low seat position. However, given that seat height is strongly governed by impairment level in the wheelchair court sports, the results relating to camber would still be valid (Chapter 3).

6.6 CONCLUSIONS

Currently, limited evidence based information is available to assist athletes with their selections when configuring a sports wheelchair. In association with the results of Chapter 5, where it was established that 24° was not an energetically favourable camber setting during sub-maximal wheelchair propulsion, the current study also identified that this setting was not favourable for performing maximal effort bouts of sport-specific over-ground propulsion. The results of the current study may also provide athletes with some guidelines to facilitate their camber selections when configuring a new sports wheelchair. Inexperienced athletes who are uncertain of their optimal setting could be advised to select 18° camber, due to the improved main effects for both linear and non-linear aspects of performance. Additionally, there was evidence to suggest that athletes who adopt a high seating position may benefit from greater camber in certain situations due to the improved stability these settings may offer, yet at the same time experience a greater resistance in between pushes as a trade off. Although the effects of camber on other aspects of performance such as ball handling and stroke production may need to be considered, the current investigation has provided a detailed insight into how camber may optimise mobility performance specific to the wheelchair court sports.

The previous and current chapter (Chapters 5 & 6) have provided a thorough examination of the effects of camber on the ergonomics of sports wheelchair performance. The following two studies adopt a similar multidisciplinary approach to investigate the effects of different wheel sizes with fixed gear ratios on aspects of sub-maximal (Chapter 7) and maximal effort (Chapter 8) mobility performance.

Chapter Seven

Effects of Wheel and Hand-rim Diameter on Sub-maximal Wheeling Performance in Elite Wheelchair Athletes

Mason BS, Woude van der LHV, Lenton JP & Goosey-Tolfrey VL. Effects of wheel and hand-rim size on sub-maximal propulsion in wheelchair athletes. *Medicine & Science in Sports & Exercise*, Under Review.

7.1 ABSTRACT

Purpose: To investigate the effects of fixed gear ratio wheel sizes on the physiological and biomechanical responses to sub-maximal wheelchair propulsion. ***Methods:*** Highly trained WCB players (N = 13), grouped by classification, propelled an adjustable sports wheelchair on a motor driven treadmill in three different wheel sizes (24”, 25”, 26”). Each wheel was equipped with force sensing hand-rims (SMART^{Wheel}) which collected kinetic and temporal data. Oxygen uptake, HR and RPE responses were monitored, with high speed video footage collected to determine 3D upper body joint kinematics. A two-way ANOVA with repeated measures was conducted on all variables. ***Results:*** Mean P_O and work per cycle increased between 24” (25.3 ± 6.6 W, $P < 0.0005$; 35.1 ± 29.6 J, $P = 0.015$) and 25” (22.0 ± 6.5 W; 29.6 ± 24.6 J) and 25” and 26” wheel sizes (19.1± 5.9 W, $P < 0.0005$; 26.0 ± 23.9 J, $P = 0.002$) respectively. Oxygen uptake (0.98 ± 0.20 L·min^{-1}, $P = 0.035$) and HR responses (105 ± 9 beats·min^{-1}, $P = 0.010$) were elevated in 24” compared to 26” wheels (0.87 ± 0.16 L·min^{-1}; 99 ± 8 beats·min^{-1}). Mean *Fres* ($P = 0.006$) and *Ftan* ($P = 0.007$) increased in 24” (46.8 ± 19.5 N; 28.4 ± 16.6 N) compared to 26” wheels (39.7 ± 20.3 N; 24.5 ± 16.7 N). No changes in temporal or upper body joint kinematics were observed between wheel sizes and no interactions existed between wheel size, classification and performance. ***Conclusions:*** A greater power requirement owing to a greater rolling resistance in 24” wheels increased the physiological demand and magnitude of force application during sub-maximal wheelchair propulsion.

7.2 INTRODUCTION

It has long been acknowledged that an optimisation of wheelchair geometry to complement the user's physical characteristics is a necessity for improving the ergonomics of manual wheelchair propulsion and sport performance (Higgs, 1983; Woude et al., 1989a). It was identified by a group of elite wheelchair sportsmen in the first experimental study (Chapter 3) that the size of the main wheels can have a strong bearing on mobility performance during the wheelchair court sports. Larger wheels were described as being more economical due to the reduced stroke frequency they required. However, evidence based research into the ergonomics of wheel size manipulations is lacking.

From a theoretical perspective, the mechanical effects of adjusting wheel size are clear, with smaller wheels known to increase rolling resistance (Kauzlarich and Thacker, 1985). When other areas of the wheelchair-user combination and propulsion velocity are controlled, increased rolling resistance increases the external P_O for the user (Woude et al., 1988a). Investigations into the effects of different hand-rim diameters have been conducted in relation to fixed wheel sizes (Woude et al., 1988b; Gayle et al., 1990; Guo et al., 2006; Costa et al., 2009) whereby rolling resistance and subsequently P_O are likely to be unaffected. Despite this, increases in physiological demand were still established in larger hand-rim diameters (Woude et al., 1988b; Gayle et al., 1990). This was thought to be due to the biomechanical adaptations evoked by the different gear ratios that are introduced when hand-rim diameter is manipulated in relation to fixed wheel sizes (Woude et al., 1988b; Veeger et al., 1992b, Guo et al., 2006). However, in the wheelchair court sports, increases in wheel size are commonly accompanied by proportionate increases in hand-rim diameter and subsequently changes in gear ratio are unlikely (Chapter 3).

Although an increase in physiological demand is anticipated with smaller wheels, due to the relationship previously alluded to between P_O and physiological demand (Woude et al., 1988a), the effects of adjusting wheel size with a fixed gear ratio on biomechanical parameters remains unknown. It is unclear whether the mechanical effects of wheel size modifications account for the changes in physiological demand, or whether there are associated adaptations in propulsion biomechanics that could further explain any changes. A biomechanical investigation also enables an assessment of injury risk to be established in different wheel size conditions. Kinetic investigations in particular can supplement the assessment of user safety/health, given that variables such as the magnitude and rate of force development can both affect the risk of upper extremity injury in wheelchair users (Boninger et al., 1999; 2002). Yet, to the author's knowledge, investigations into force application have

only been utilised when manipulating wheelchair configurations specific to daily life wheelchair propulsion (Linden et al., 1995; Boninger et al., 2000; Kotajarvi et al., 2004; Mulroy et al., 2005; Cowan et al., 2009).

The principal aim of the current investigation was to determine the effects of different wheel sizes, with fixed gear ratios, on the physiological and biomechanical responses to sub-maximal wheelchair propulsion in a group of highly trained athletes. Based on the limited quantitative literature it was hypothesised that smaller wheels would increase the P_O requirements and the physiological demand from the user. Increases in stroke frequency, upper body joint excursions and hand-rim kinetics were also expected to account for any changes in physiological demand resulting from wheel size manipulations. These anticipated changes in propulsion biomechanics were also hypothesised to potentially exacerbate the risk of injury in smaller wheel sizes. A secondary aim was to investigate whether any ergonomically optimal wheel sizes could be established and whether any relationships existed when athletes were grouped according to their IWBF classification level.

7.3 METHODS

7.3.1 Participants

Thirteen highly trained WCB players (24 ± 7 years; 66.6 ± 15.6 kg) volunteered to participate in the current study. Given that large inter-individual differences exist between wheelchair athletes, one configuration is unlikely to be optimal for all (Chapter 6). Player's role on court and level of impairment in particular can influence selections relating to wheelchair configuration (Yilla et al., 1998). This also applies to wheel size as the more physically able, "high point players" are often seen to select a larger wheel size compared to more impaired, "low point players" (Hutzler et al., 1995). Subsequently, participants were grouped according to impairment using the IWBF classification system (IWBF, 2009). Participants with a classification of ≤ 2.5 were termed "low point players" (LP, N = 7), whereas those with a classification > 2.5 were termed "high point players" (HP, N = 6). Details of participants' physical characteristics are documented in Table 7.1. Procedures for the current investigation were approved by Loughborough University's Ethical Committee and all participants provided their written informed consent prior to testing.

Table 7.1 – Details of participants' physical characteristics, their sporting involvement and self-selected wheelchair configurations.

Sex (M/F)	Age (yrs)	Mass (kg)	Playing Experience (yrs)	Impairment	Classification	Group	Wheel Size (")	Elbow Angle (°)
M	35	65.1	18	T9/10 inc. SCI	1.5	LOW	24	92
M	27	59.9	16	T12 com. SCI	2.0	LOW	25	92
M	26	57.8	13	T12 com. SCI	2.0	LOW	24	85
M	24	74.2	11	T12 com. SCI	2.0	LOW	24	102
M	17	46.1	9	L1-4 inc. SCI	2.5	LOW	25	75
F	16	52.1	7	T12 com. SCI	2.5	LOW	24	95
F	18	52.8	5	Motorneuropathy	2.5	LOW	24	105
F	27	56.4	3	L1 inc. SCI	3.0	HIGH	25	106
M	28	74.1	15	Spina Bifida	3.0	HIGH	26	109
M	21	72.3	14	Brittle Bones	4.0	HIGH	26	109
M	37	94.1	2	Amp (LAK)	4.0	HIGH	27	121
F	20	62.8	4	Myalgic encephalomyelitis	4.0	HIGH	25	139
M	20	97.6	5	Club Foot	4.5	HIGH	26	150

Key: Sex – (M) Male, (F) Female; **Impairment** – (SCI) – Spinal cord injury [com. – complete; inc. – incomplete], (Amp) Amputee [AK – above knee; BK – below knee]; **Classification** - Based on IWBF classification system.

7.3.3 Experimental design

All participants were tested in an adjustable sports wheelchair (Top End Transformer, Invacare) for three different wheel sizes commonly used in the wheelchair court sports (24” – 0.592 m; 25” – 0.614 m; 26” – 0.646 m). The hand-rim diameters of these wheel sizes were 24” – 0.533 m, 25” – 0.552 m and 26” – 0.585 m. Veeger et al. (1992b) calculated the gear ratio used during propulsion by dividing the hand-rim radius by the wheel radius. Applying this principle to the wheel sizes investigated in the current study confirmed that a fixed gear ratio of 0.9 was present for each wheel size condition.

Each wheel size was configured with a separate force sensing hand-rim (SMART^{Wheel}, Three Rivers Holdings, Mesa, AZ) to determine the 3D forces and moments applied during propulsion. Each SMART^{Wheel} weighed approximately 4.7 kg. In order to counterbalance the greater weight and moment of inertia of the SMART^{Wheel} in relation to a conventional wheel, additional weight was added around the hub of the opposing wheel, resulting in an overall wheelchair mass of 18 kg. Each wheel size condition was investigated during a 4-minute bout of propulsion at a fixed velocity and gradient (2.2 $m \cdot s^{-1}$; 0.7%) on a motor driven treadmill (H/P/Cosmos Saturn, Nussdorf-Traunstein, Germany). Seat height was determined for each participant in their own sports wheelchair, using the degree of elbow extension induced when the hands were positioned at TDC of the hand-rims. These elbow extension values were then replicated in the adjustable wheelchair and controlled across wheel size conditions by making minor adjustments to the seat height. The only area of configuration that could not be standardised between conditions was the distance between TDC of both wheels due to the use of a fixed length camber bar, which subsequently led to minor differences (24” – 0.496 m; 25” – 0.477 m; 26” – 0.468 m). Rear-wheel camber was kept constant at 18° due to the favourable performance of WCB players in this setting (Chapters 5 & 6), with tyre pressure controlled at 120 psi and toe-in toe-out strictly monitored. The order for testing was conducted in a randomised order.

7.3.4 Cardio-respiratory measures

During the final minute, expired air was collected using the Douglas bag technique (Cranlea, Birmingham, U.K.) and HR was recorded at 5-second intervals using radio telemetry (PE4000 Polar Sport Tester, Kempele, Finland). On completion of each condition, localised, centralised and overall RPE was assessed using the Borg scale (Borg, 1970). Expired air was analysed for oxygen and carbon dioxide concentrations (Servomex 1400 Gas Analyser, Sussex, UK) and volume (Harvard Dry Gas Meter, Harvard Apparatus, Kent, UK).

Subsequently $\dot{V}O_2$ and RER were calculated for each wheel size condition. Gross mechanical efficiency was also calculated using the mean P_O values derived from the SMART^{Wheel} and energy expenditure from the $\dot{V}O_2$ and the oxygen energetic equivalent of the RER values using a standard conversion table (Whipp and Wasserman, 1969; Peronnet and Massicotte, 1991).

7.3.5 Kinetic measures

Each SMART^{Wheel} was positioned on the left hand side of the wheelchair and had been individually calibrated with known weights suspended from the hand-rims when the wheels were positioned vertically. A calibration constant for each wheel could then be calculated to determine the raw forces and moments (Asato et al., 1993). Thirty seconds of data were collected via an infrared wireless transmitter sampling at 240 Hz, 2-minutes 30-seconds into each 4-minute bout, with participants instructed to push by the hand-rim alone. Kinetic data were filtered using a 32-tap finite impulse response (FIR) low-pass digital filter with a 20 Hz cut-off frequency, which enabled the filtered forces and moments to be determined during each wheel size condition. All forces (*F*) and moments (*M*) derived from the SMART^{Wheel} were defined as follows: *Fx* – horizontally forward; *Fy* – vertically downward; *Fz* – horizontally inwards; and *Mz* - referred to the moment produced around the hub in the plane of the wheel (Asato et al., 1993). Mean P_O was calculated from the mean *Mz* and the mean angular velocity (ω) of the wheel:

$$P_O\ (\mathrm{W}) = Mz\ (\mathrm{N{\cdot}m}) \cdot \omega\ (^\circ{\cdot}\mathrm{s}^{-1}) \qquad \text{(Niesing et al., 1990)}$$

The mean work per cycle was then derived from the mean P_O and stroke frequency (f) values:

$$\mathrm{Work}\ (\mathrm{J}) = P_O / f\ (\mathrm{strokes{\cdot}s}^{-1}) \qquad \text{(Woude et al., 1986)}$$

The vector sum of the SMART^{Wheel} force components (*Fx, Fy* and *Fz*) enabled the resultant forces (*Fres*) applied to the hand-rims to be determined:

$$Fres\ (\mathrm{N}) = \sqrt{\left(Fx^2 + Fy^2 + Fz^2\right)} \qquad \text{(Cooper et al., 1997)}$$

The force that directly contributed to the rotation of the wheels, referred to as the tangential force (*Ftan*) was also calculated whereby Rr^{-1} refers to the hand-rim radius:

$$Ftan\ (\mathrm{N}) = Mz / Rr^{-1}(\mathrm{m}) \qquad \text{(Robertson et al., 1996)}$$

Using the previous two equations, the fraction of effective force (*FEF*), which describes the ratio of force that contributes towards forward motion (*Ftan*) in relation to the resultant force (*Fres*) was established:

$$FEF\ (\%) = (Ftan / Fres) \cdot 100 \qquad \text{(Cooper et al., 1997)}$$

To obtain an indicator of injury risk, the rate of force development was also calculated as the ratio between the change in *Fres* from initial HCon to the peak *Fres* and the change in time between these two events (Boninger et al., 1999).

All forces and moments were expressed as mean values per stroke, which were then averaged over the total number of strokes produced during the 30-second data collection period. The only exception was in the calculation of mean P_O, whereby the recovery phase was accounted for with *Mz* and the angular velocity of the wheel averaged from the onset of the first push to the completion of the final push.

The temporal parameters associated with propulsion were also computed from the kinetic data. Stroke frequency (strokes·s^{-1}) was calculated by dividing the number of strokes during the 30-seconds by the change in time from the beginning of the first push to the end of the last push. Push times were defined as the period of time that the hand exerted a positive *Mz* around the hub of the wheel. Push angles were also derived from the SMART^{Wheel} and described the relative angle over which the push occurred.

7.3.6 Kinematic measures

Full details for the kinematic procedures are documented in Chapter 5 (section 5.3.4). All calibration, data collection and analysis procedures were replicated. The only omission was that temporal parameters associated with propulsion were no longer assessed from the high speed video analysis, as previously mentioned these were now obtained directly from the SMART^{Wheel}.

7.3.7 Statistical analyses

Means and standard deviations (SD) were calculated for all variables. The Statistical Package for Social Sciences (SPSS Version 16.0; Chicago, IL) was used for all statistical analyses. A two-way ANOVA with repeated measures examined any differences for each physiological, kinetic and kinematic variable across wheel size conditions and was used to identify any interactions with classification level. Pairwise comparisons with a Sidak adjustment were conducted to establish where any significant differences between wheel sizes existed. Statistical significance was accepted at $P < 0.05$. Effect sizes were calculated to identify the meaningfulness of any differences, whereby $r > 0.5$ reflected a large effect size (Cohen, 1992).

7.4 RESULTS

7.4.1 Wheel size

As hypothesised, an inverse relationship existed between wheel size and P_O requirements (Figure 7.1). A significantly higher P_O was required during propulsion in 24" wheels (25.3 ± 6.6 W) compared to 25" wheels (22.0 ± 6.5 W; r = 0.93) and 26" wheels (19.1 ± 5.9 W; r = 0.95). External P_O was also significantly greater for 25" wheels compared to 26" wheels (r = 0.93). These changes resulted from identical patterns in *Mz* and angular velocity data, which were both significantly greater (P < 0.0005) in smaller wheel sizes.

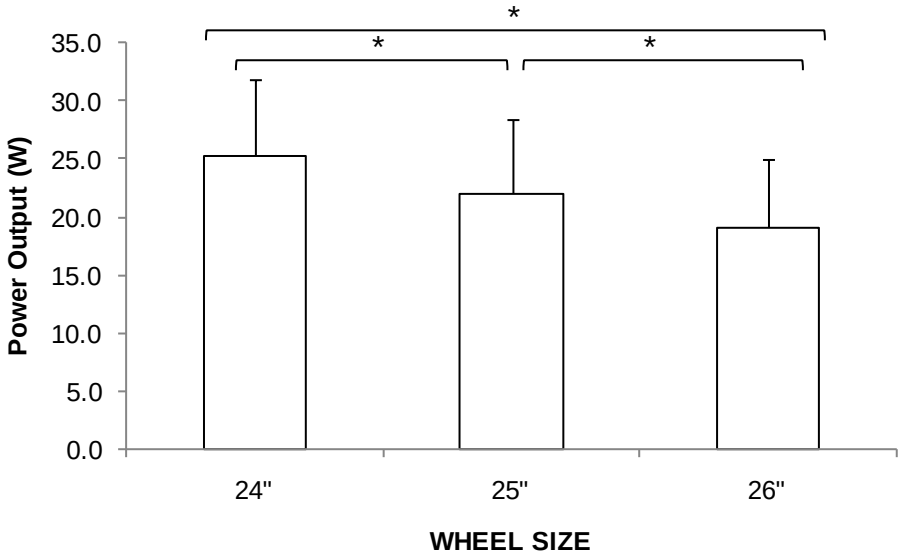

Figure 7.1 – Power output requirements between wheel size conditions. * represents a significant difference between wheel sizes, *P* < 0.0005.

The physiological demand of wheelchair propulsion was also reduced in larger wheel sizes (Table 7.2). An increased $\dot{V}O_2$ (reduced economy) was revealed for the 24" condition in relation to both the 25" (P = 0.014, r = 0.76) and 26" condition (P = 0.035, r = 0.75). Participants also displayed significantly reduced HR responses in 26" wheels compared to 24" (P = 0.010, r = 0.81) and 25" wheels (P = 0.040, r = 0.74). Alternatively, ME improved with decreasing wheel size, with a significantly higher ME demonstrated for 24" wheels compared to 26" wheels (P = 0.005, r = 0.82). No effects of wheel size on the localised, centralised or overall RPE were noted (Table 7.2).

Table 7.2 – Physiological responses to different wheel sizes. All values presented are means (± SD).

	P values	24”	25”	26”
$\dot{V}O_2$ (L·min^{-1})	0.005	0.98 (0.20)	0.90 (0.20)[a]	0.87 (0.16)[a]
HR (beats·min^{-1})	0.001	105 (9)	103 (8)	99 (6)[a b]
ME (%)	0.001	7.4 (1.8)	7.1 (2.0)	6.3 (1.7)[a]
Local RPE	0.616	9 (3)	8 (2)	8 (3)
Central RPE	0.548	8 (2)	8 (2)	8 (2)
Overall RPE	0.993	8 (2)	8 (2)	8 (2)

Key: [a] represents a significant difference to 24” setting.
[b] represents a significant difference to 25” setting.

Work per cycle was also shown to be significantly affected by wheel size manipulations (P = 0.001). As demonstrated in Figure 7.2, a greater amount of work per cycle was performed in 24” wheels (35.1 ± 29.6 J) compared to both 25” (29.6 ± 24.6 J; P = 0.015, r = 0.70) and 26” wheels (26.0 ± 23.9 J; P = 0.003, r = 0.79). A greater amount of work per cycle was also performed during the 25” condition compared to the 26” condition (P = 0.002, r = 0.84). The temporal parameters, stroke frequency and PT, were both unaffected by wheel size (Table 7.3). However, PA were shown to reduce with increasing wheel size, with greater displacements observed in 24” versus 26” wheels (P = 0.035, r = 0.74). Wheel size had no statistically significant bearing on upper body joint kinematics during the push or recovery phases for the group as a whole.

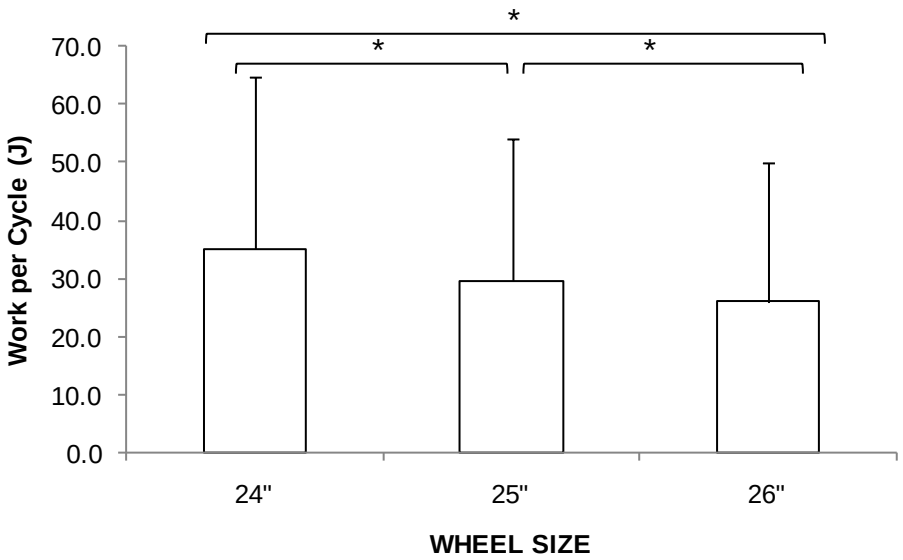

Figure 7.2 – Changes in work per cycle across wheel size conditions. * represents a significant difference between wheel sizes, *P* < 0.0005.

Table 7.3 – Temporal and displacement hand-rim parameters during different wheel sizes. Values are means (± SD).

	***P* values**	**24”**	**25”**	**26”**
Stroke frequency (strokes·s^{-1})	0.824	0.92 (0.31)	0.91 (0.26)	0.92 (0.26)
PT (s)	0.835	0.26 (0.03)	0.26 (0.03)	0.26 (0.03)
PA (°)	0.015	106.3 (11.0)	104.3 (11.2)	99.8 (12.9)[a]

Key: [a] represents a significant difference to 24” setting.

Hand-rim kinetics during the push phase also varied dependent on wheel size (Figure 7.3). Mean *Fres* was significantly influenced by wheel size ($P < 0.0005$) and was shown to increase in smaller wheels (Figure 7.3a). The 24” condition evoked a significantly greater magnitude (46.8 ± 19.5 N) than both the 25” (41.9 ± 20.1 N; $P = 0.009$, $r = 0.80$) and 26” conditions (39.7 ± 20.3 N; $P = 0.006$, $r = 0.81$). Although not statistically significant ($P = 0.090$) there were trends for the 25” condition to also demand a greater magnitude of *Fres* compared to the 26” condition, supported by a large effect size ($r = 0.63$). Figure 7.3b revealed that mean *Ftan* was also elevated in smaller wheels ($P = 0.002$). The 24” condition revealed a significantly greater mean *Ftan* (28.4 ± 16.6 N) than displayed in the 26” condition (24.5 ± 16.7 N; $P = 0.007$, $r = 0.76$) and although statistically insignificant ($P = 0.113$) a similar relationship was observed in comparison to the 25” condition (25.4 ± 17.1 N; $r = 0.63$). No significant effect of wheel size was observed for the mean *FEF* ($P = 0.924$), given the similar values reported in Figure 7.3c. No statistically significant differences existed between wheel size and the rate of force development ($P = 0.108$). However, trends existed for a reduced rate of force development to coincide with increasing wheel sizes (Figure 7.3d). This was exemplified by the slightly more rapid rate of force development evoked in the 24” condition (399.2 ± 175 N·s^{-1}) compared to the 26” condition (323.2 ± 125.2 N·s^{-1}; $r = 0.55$).

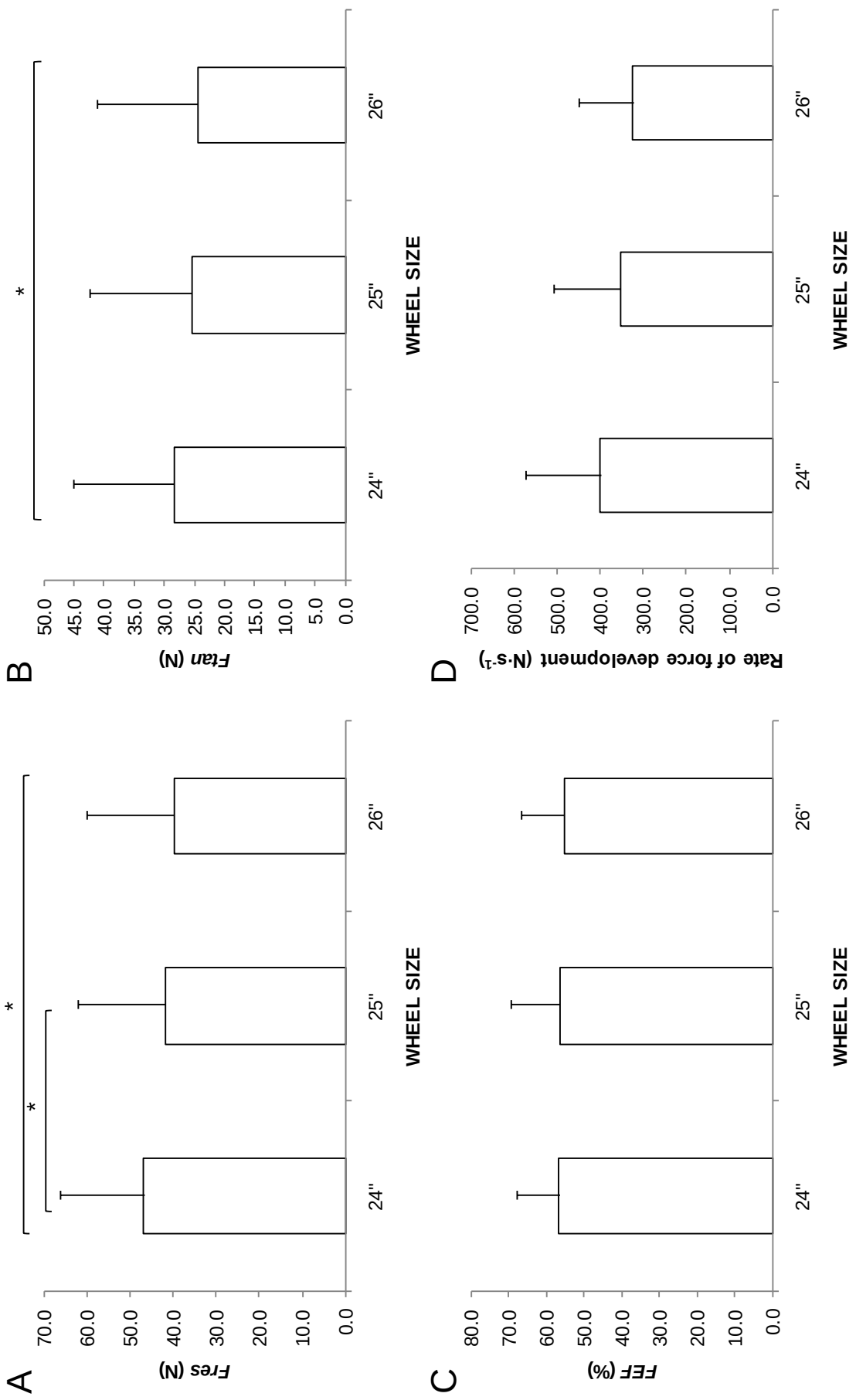

Figure 7.3 – Adaptations in hand-rim kinetics during different wheel size conditions: (a) mean *Fres*, (b) mean *Ftan*, (c) mean *FEF* and (d) rate of force development. * represents a significant difference between wheel size conditions, $P < 0.05$.

7.4.2 Wheel size and classification

No interactions between wheel size and classification were evident for any of the physiological or biomechanical parameters investigated. However, a number of significant between subject effects were evident between classification level and certain kinematic and kinetic parameters (Table 7.4). A greater amount of work per cycle, mean forces (*Fres* and *Ftan*) and a reduced stroke frequency was demonstrated by HP compared to LP. During the push phase HP also exhibited a larger active RoM at the trunk and the wrist. It was evident that LP actually displayed trunk extension during the push phase, whereas HP flexed their trunk. Finally, the maximum and minimum shoulder abduction values were significantly greater for LP when compared to HP during the push phase.

Table 7.4 – Mean (± SD) values for the kinetic and kinematic variables exhibiting (or strongly approaching) statistically significant differences between classification groups.

	***P* value**	**LP**	**HP**
Work per cycle (J)	0.093	20.3 (6.8)	47.6 (37.6)
Fres (N)	0.036	33.7 (8.9)	58.7 (23.2)
Ftan (N)	0.033	18.4 (4.2)	39.7 (20.8)
Stroke frequency (strokes·s^{-1})	0.023	1.05 (0.17)	0.69 (0.26)
Active trunk RoM (°)	0.023	-4.1 (2.0)	3.9 (2.1)
Active wrist RoM (°)	0.032	18.9 (2.2)	28.0 (2.9)
Maximum shoulder abduction (°)	0.009	38.0 (1.9)	27.3 (2.6)
Minimum shoulder abduction (°)	0.008	27.9 (1.8)	17.6 (2.4)

7.5 DISCUSSION

7.5.1 Wheel size

The results of the current investigation revealed that different wheel sizes with fixed gear ratios, significantly affected the physiological demand and propulsion kinetics in a standardised sports wheelchair configuration. These adaptations occurred predominantly through the dimensional changes evoked from wheel size adjustments and the consequential effects observed regarding P_O requirements. As expected an inverse relationship was identified between wheel size and P_O, as a consequence of a higher rolling resistance with

decreasing wheel size (Kauzlarich and Thacker, 1985), the P_O requirements of the user increased. Based on this and in association with the relationship between the energetic cost of wheelchair propulsion at constant velocities and increased P_O, physiological demand was elevated in smaller wheel sizes (Woude et al., 1988a). The increased P_O resulted from a higher angular velocity and *Mz* in smaller wheels, as, in order to maintain the constant velocity of the treadmill, these wheels needed to rotate at a greater rate and required a larger torque to achieve this. As a result of the increased *Mz* and reduced hand-rim radius of smaller wheels, a larger *Ftan* was also demanded in smaller wheel sizes.

As previously mentioned, an elevation in physiological demand was observed in smaller wheel sizes. This was demonstrated by the elevated $\dot{V}O_2$ (reduced economy) and HR responses in the 24" condition compared to the 26" condition. When utilising fixed wheel sizes, smaller diameter hand-rims have been shown to reduce these responses (Woude et al., 1988b; Gayle et al., 1990). However, the variations in gear ratio imposed by these investigations were likely to account for the increased physiological demand (Veeger et al., 1992b). Veeger et al. (1992b) revealed increases in $\dot{V}O_2$ and HR when P_O was increased, yet gear ratios remained constant. This appears to support the results of the current study whereby fixed gear ratios were maintained with decreases in wheel size, yet the P_O increases observed in these settings were likely to have evoked the increased physiological demand. Despite the increased $\dot{V}O_2$ and HR responses demonstrated in 24" wheels, ME was also shown to improve in this setting due to its dependence on P_O (Woude et al., 1988a). These findings are not unexpected, given the curvilinear relationship that has been demonstrated between P_O and ME (Woude et al., 1988a). Also, at the fixed gear ratios imposed by Veeger et al. (1992b), improvements in ME were observed when P_O was increased. Therefore, in the context of the present investigation it can be deduced that smaller wheel sizes increased the P_O requirements from the user more substantially than the energetic cost of propulsion, given that ME was still seen to improve. However, given the emphasis placed on aerobic fitness and endurance during wheelchair court sport competition (Vanlandewijck et al., 1999a; Goosey-Tolfrey, 2005; Roy et al., 2006; Bernardi et al., 2010), ME is not likely to be the best indicator for assessing the efficiency of certain wheel size conditions, as also seen with camber (Chapter 5). During constant speed propulsion, increases in P_O are highly unfavourable and subsequently reductions in P_O, oxygen cost and thus an improvement in economy may be of greater relevance to wheelchair court sport athletes, as seen in larger wheel sizes.

Limited kinematic adaptations to wheelchair propulsion in different wheel sizes existed with fixed gear ratios. In fact no significant differences in temporal parameters or upper body joint kinematics were observed between wheel size conditions. Only PA was shown to be affected, whereby decreases in wheel size were accompanied by marginal, yet significant increases in PA. Given that no differences were observed between wheel sizes for PT or the displacement of the hand during the push phase, it was likely that participants pushed over a similar linear trajectory across wheel size conditions. Subsequently, the changes in physiological demand with regards to fixed gear ratio wheel size adjustments were not the result of any concomitant alterations to upper body joint excursions or the timing of propulsion.

Investigations into varying hand-rim diameters and gear ratios have attributed increases in cardio-respiratory stress with larger diameters to kinematic adaptations such as increased RoM (Woude et al., 1988b) and linear velocities (Guo et al., 2006) of upper body segments. In addition to the fixed wheel sizes and hence different gear ratios, these studies have also utilised fixed seat positions, which alters the position of the user in relation to TDC of the hand-rim in each condition. Therefore kinematic adaptations would be expected, yet the influence of such on physiological demand cannot be directly attributed to hand-rim modifications, since variations in seat position are also introduced. The more stringent standardisation methods imposed during the present investigation, particularly relating to seat height, could explain the absence of any kinematic adaptations to wheel size conditions. Furthermore, it has also been established that experienced, elite WCB players possess a highly reproducible propulsion technique (Vanlandewijck et al., 1994b). Therefore, it was proposed that at the given sub-maximal velocities and well controlled testing conditions, slight changes in wheel size were not sufficient to evoke a change in propulsion kinematics in this participant group.

One kinematic parameter that was originally expected to be influenced by wheel size adjustments was stroke frequency. It was anticipated that a higher stroke frequency would be required in smaller wheels in order to overcome the increased rolling resistance associated with these wheels and to maintain the constant velocity of the treadmill. Support was added to this hypothesis, since stroke frequency has been shown to display a curvilinear relationship with P_O (Woude et al., 1988a) and given that smaller wheels increase P_O, a higher stroke frequency was anticipated. Veeger et al. (1992b) also established that during propulsion under fixed gear ratios, increases in P_O increased stroke frequency. These results may again arise from the highly standardised protocols implemented and the experienced nature of the

participants sampled in the current investigations. The consistent stroke frequencies observed across wheel size conditions necessitated that a greater amount of work per cycle was performed in order to meet the P_O requirements of the smaller wheel sizes. This is in accordance with previous investigations, whereby work per cycle has been shown to elicit a curvilinear relationship with P_O (Woude et al., 1988a). Also under constant P_O conditions, Woude et al. (1988b) revealed that manipulating gear ratios through adjustments in hand-rim diameter did not affect the amount of work per cycle.

It appeared as though athletes employed a number of adaptations in hand-rim kinetics, which may have served as compensatory factors to meet the P_O requirements of smaller wheels in the absence of any increases in stroke frequency. It was clear that by maintaining a relatively constant stroke frequency, athletes alternatively applied a greater torque around the hub (*Mz*) of the wheel. Based on its relationship with hand-rim diameter, a greater magnitude of mean *Ftan* was subsequently demanded in smaller wheel sizes in order to generate a greater *Mz*. In addition to this an increased mean *Fres* was also observed in smaller wheel sizes. Veeger et al. (1992b) demonstrated that gear ratio had no effect on any of the subcomponents of *Fres* (*Fx*, *Fy* & *Fz*), yet increases in P_O at fixed gear ratios significantly increased each of these subcomponents. This again suggested that the increased P_O associated with smaller wheels may have been predominantly responsible for the greater magnitude of *Fres* in this wheel size. These increased force magnitudes in smaller wheel sizes may also explain the greater physiological demand. It was anticipated that given the absence of any adaptations in the joint excursions of upper body segments resulting from wheel size adjustments, the increased force application and subsequently physiological demand may be the result of an increased muscular activity. Although an EMG analysis would be warranted to determine this, it appeared likely that the forces at which the muscles are contracting maybe greater in smaller wheels, even though they would appear to be doing so at similar lengths and velocities.

No changes in the effectiveness of force application were revealed between wheel size conditions, as even though *Fres* increased with decreasing wheel sizes, it was accompanied by seemingly proportionate increases in *Ftan*, resulting in no effects on mean *FEF*. This was reinforced by the findings of Veeger et al. (1992b) who established that when gear ratio was constant, yet P_O increased, no changes in mean *FEF* were observed. Alternatively, under constant P_O conditions a reduction in gear ratio was shown to significantly improve the effectiveness of force application (Veeger et al., 1992b).

The hand-rim kinetic data also offered a meaningful insight into injury risk. It has previously been stated that increases in cadence, force magnitude and rate of force development are all associated with an increased risk of injury in wheelchair users (Boninger et al., 1999; 2002). The fact that no changes in stroke frequency were observed suggested that wheel size had no influence on overuse injuries during sub-maximal wheelchair propulsion. However, the greater magnitudes of force application observed in 24" wheels implied that smaller wheels may increase injury risk as a result of the increased forces imparted on the skeletal system. In support of this, trends existed for smaller wheels to also increase the rate of force development. In order to develop a more comprehensive understanding of the role of wheel size on injury risk in wheelchair athletes, acceptable limits of force magnitude are desirable. Also, investigations at higher velocities that are more reflective of wheelchair sports propulsion will be warranted when force application will be augmented (Veeger et al., 1992b; Boninger et al., 1999).

7.5.2 Wheel size and classification

A secondary aim of the current investigation was to establish whether any physiological and biomechanical adaptations to wheel size adjustments were dependent on classification level. No interactions displaying or approaching statistical significance existed. This may have been the result of an insufficient sample size from which to divide participants into two groups. The absence of any interactions may also be due to participants already having a self-selected wheel size, which varied within classification groups, although there are strong trends for HP to opt for larger wheels than LP (Table 7.1). It is possible that a familiarisation period with wheel sizes different to the athletes' current choice is required. This may be valid during on court testing whereby fewer restrictions are enforced, yet to a lesser extent during propulsion at a constant velocity on a treadmill whereby a reliance on wheelchair handling skills are limited. The fact that no interactions existed between classification, wheel size and any of the physiological or biomechanical responses to wheelchair propulsion emphasises that wheelchair configuration is highly specific to each individual and is dependent on other factors in addition to classification. It was clear from the between-subject effects that classification level affected various kinetic and kinematic parameters during wheelchair propulsion. Differences between the two classification groups were unsurprising given the higher seat positions adopted by HP and subsequently the greater distance between the shoulder and TDC of the hand-rims. As a result, this group of participants demonstrated a greater degree of trunk and wrist motion during the push phase

and maintained a less abducted shoulder position throughout, whilst performing more work per cycle at a reduced stroke frequency and exerting larger forces on the hand-rim. The fact that classification level directly corresponded with seat height in the current investigation, supported what has previously been suggested in the literature (Hutzler et al., 1995). Although, investigating the role of classification level or seat height was not the principal aim of the current investigation it has further highlighted the importance of grouping participants in future studies investigating WCB players to make results more specific to the individuals.

7.6 CONCLUSIONS

The present investigation has demonstrated that decreasing wheel size, with a fixed gear ratio increases the physiological demand and magnitude of force application during sub-maximal wheelchair propulsion under highly standardised conditions. The changes in physiological and biomechanical responses were largely the result of the increased P_O associated with smaller wheels owing to an increased rolling resistance. No ergonomically optimal settings could be identified based on the findings of the current study due to the absence of any meaningful interactions between wheel size and classification level. However, for highly trained wheelchair athletes, the 26” condition evoked favourable responses to sub-maximal wheelchair propulsion, as demonstrated by the reduced P_O, physiological demand and magnitude of force application for the group as a whole. Subsequently, given this seemingly linear relationship between wheel size and the ergonomics of sub-maximal wheelchair performance, further research would be advised to extend this investigation to 27” wheels, since this wheel size is emerging on the wheelchair court sports scene. Future investigations may also benefit from exploring the effects of different wheel sizes with different gear ratios. Although these are currently fixed within the wheelchair court sports, further benefits in physiological demand (Woude et al., 1988b; Gayle et al., 1990; Veeger et al., 1992b) and effectiveness of force application (Veeger et al., 1992b) have been observed through reductions in gear ratio.

Finally, a field based investigation would further assist with the identification of optimal wheel sizes. Although unfavourable physiological and biomechanical responses were revealed, benefits in initial acceleration and manoeuvrability have been alleged with smaller wheels (Coutts, 1990 & Chapter 3). Therefore, the aim of Chapter 8 was to investigate the effects of wheel size on the performance of maximal effort, over-ground wheelchair propulsion.

Chapter Eight

The Effect of Wheel Size on Mobility Performance in Wheelchair Athletes

8.1 ABSTRACT

Purpose: To examine the effects of different wheel sizes with fixed gear ratios on maximal effort mobility performance during hand-rim wheelchair propulsion. ***Methods:*** Thirteen highly trained WCB players, grouped by classification level, performed a 20 m linear sprint, a linear mobility and an agility drill in an adjustable wheelchair with three different wheel sizes (24", "25" and 26"). Performance was assessed by the time taken to perform the drills, accelerations, decelerations and the mean and peak velocity displayed using a velocometer. ***Results:*** Twenty metre sprint times were improved in the 26" condition (5.58 ± 0.43 s, P = 0.029) compared with 24" (5.72 ± 0.40 s). The mean (P = 0.003) and peak (P = 0.078) velocity was also greater in the 26" condition (mean = 3.11 ± 0.19 $m \cdot s^{-1}$; peak = 4.77 ± 0.46 $m \cdot s^{-1}$) compared to 24" (mean = 2.99 ± 0.20 $m \cdot s^{-1}$; peak = 4.61 ± 0.40 $m \cdot s^{-1}$). The number of pushes required to complete the 20 m sprint was also lower in 26" wheels (11.7 ± 1.3, P = 0.033) compared to 24" wheels (12.6 ± 1.6). Acceleration performance over the first two (P = 0.299) and three (P = 0.145) pushes was unaffected by wheel size, as were the times taken to complete the linear mobility (P = 0.630) and the agility drill (P = 0.505). No significant interactions existed between wheel size, classification and any performance measure. ***Conclusions:*** Maximal sprinting performance of highly trained WCB players improved in 26" wheels. This was likely to be attributed to the reduced rolling resistances experienced and the lower angular hand-rim velocities required when developing high wheelchair velocities.

8.2 INTRODUCTION

The size of the main wheels is an area of wheelchair configuration that varies between wheelchair court sport athletes (Chapter 3 & 7). In WCB, it has been suggested that wheel size selection is influenced by athletes' IWBF classification (Hutzler et al., 1995). Chapter 7 established that the self-selected wheel sizes of elite WCB players ranges from 24" to 27", with HP often shown to select larger wheels. Wheel size, by mechanical definition, affects the rolling resistance of the wheelchair-user combination, with greater resistances experienced in smaller wheels (Kauzlarich and Thacker, 1985). It was shown in Chapter 7, during conditions of constant speed, sub-maximal wheelchair propulsion that reductions in wheel size increased the external P_O. Furthermore, these reductions in wheel size from 26" to 24" resulted in an increased physiological demand, work per cycle and magnitude of force application. It was also established that a greater mean angular velocity and torque around the hub was necessitated in smaller wheels in order to maintain the given speeds of the treadmill (Chapter 7).

To provide a valid assessment of the effects of certain wheelchair configurations such as wheel size, investigations must also be conducted in ecologically valid conditions whenever possible. For athletes participating in the wheelchair court sports, this requires the performance of maximal effort movements specific to the sports to be performed during over-ground propulsion (Vanlandewijck et al., 2001). These movements include rapid initial acceleration, manoeuvrability, braking and maximal sprinting (Vanlandewijck et al., 2001), which have been examined during Chapters 4 and 6. However, such investigations have never been extended to wheel size, even though this area of configuration was perceived to influence each of the aforementioned performance indicators (Chapter 3). During 10-second maximal effort sprint tests on a roller WERG, a group of WCB players were able to accelerate their wheelchairs quicker over the first push compared to wheelchair racers, who could alternatively reach higher peak velocities (Coutts, 1990). Coutts (1990) proposed that a combination of the smaller wheel and larger hand-rim diameters associated with the WCB wheelchairs largely accounted for these results. Obviously other areas of wheelchair configuration and design differ between court sport and racing wheelchairs. One area in particular that may have influenced these results was the different gear ratios that existed between the two types of wheelchair. Lower gear ratios (smaller rim to wheel ratio) are observed within wheelchair racing configurations (Woude et al., 1988b; Gayle et al., 1990; Guo et al., 2006; Costa et al., 2009), which can alter the linear velocity of the hand-rims and

introduce coordination and coupling issues at higher wheelchair velocities (Woude et al., 1988b; Veeger et al., 1992b). However, during wheel size manipulations for the wheelchair court sports a higher, yet fixed gear ratio remains (Chapter 7).

The principal aim of the current investigation was to determine the effects of different wheel sizes with fixed gear ratios on the mobility performance of WCB players during maximal effort, over-ground propulsion. It was hypothesised that larger wheels would improve maximal sprinting performance, yet potentially at the expense of impaired initial acceleration performance. Finally, to assist athletes with wheel size selection the current investigation also examined whether different wheel sizes were optimal dependent on classification, as suggested by Hutzler et al. (1995).

8.3 METHODS

8.3.1 Participants

Thirteen highly trained WCB players (24 ± 7 years; 66.6 ± 15.6 kg) participated in the current study. Participants were divided into two classification groups, as determined by the IWBF classification system. Participants with a classification of ≤ 2.5 were termed ‘low point players’ (LP, N = 7) and those with a classification > 2.5 were termed ‘high point players’ (HP, N = 6), as employed in Chapter 7. This information and other physical characteristics of the participants are listed in Table 8.1, alongside the breakdown of athletes’ self-selected wheel size configurations (Figure 8.1). Loughborough University’s Ethical Committee approved all the procedures involved in the current investigation and all participants provided their written informed consent prior to testing.

Table 8.1 – Physical characteristics of the participants and details of their sporting involvement and current wheelchair configuration settings.

Sex (M/F)	Age (yrs)	Mass (kg)	Playing Experience (yrs)	Impairment	Classification	Group	Wheel Size (“)	Elbow Angle (°)
M	35	65.1	18	T9/10 inc. SCI	1.5	LOW	24	92
M	27	59.9	16	T12 com. SCI	2.0	LOW	25	92
M	26	57.8	13	T12 com. SCI	2.0	LOW	24	85
M	24	74.2	11	T12 com. SCI	2.0	LOW	24	102
M	17	46.1	9	L1-4 inc. SCI	2.5	LOW	25	75
F	16	52.1	7	T12 com. SCI	2.5	LOW	24	95
F	18	52.8	5	Motorneuropathy	2.5	LOW	24	105
F	27	56.4	3	L1 inc. SCI	3.0	HIGH	25	106
M	28	74.1	15	Spina Bifida	3.0	HIGH	26	109
M	21	72.3	14	Brittle Bones	4.0	HIGH	26	109
M	37	94.1	2	Amp (LAK)	4.0	HIGH	27	121
F	20	62.8	4	Myalgic encephalomyelitis	4.0	HIGH	25	139
M	20	97.6	5	Club Foot	4.5	HIGH	26	150

Key: Sex – (M) Male, (F) Female; **Impairment** – (SCI) – Spinal cord injury [com. – complete; inc. – incomplete], (Amp) Amputee [AK – above knee; BK – below knee]; **Classification** - Based on IWBF classification system.

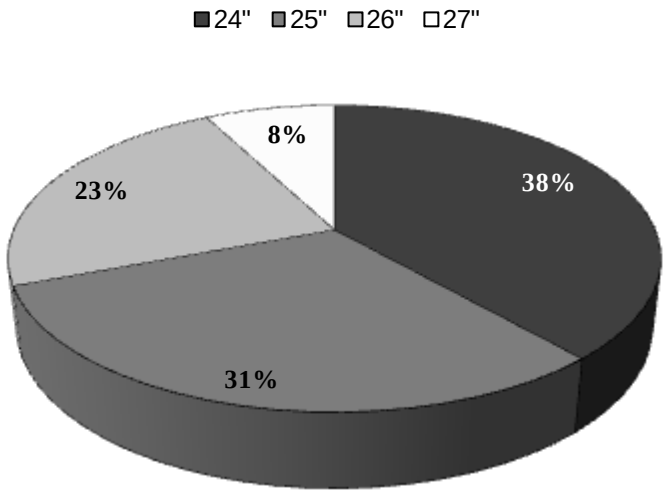

Figure 8.1 – Frequency distribution of participants' self-selected wheel sizes.

8.3.2 Equipment

All participants were tested in an adjustable sports wheelchair (Top End Transformer, Invacare: mass 11.6 kg) in three different wheel sizes (24" – 0.592 m; 25" – 0.614 m; 26" – 0.646 m). The hand-rim diameters of each of these respective wheel sizes were 24" – 0.533 m; 25" – 0.552 m; 26" – 0.585 m, giving a fixed gear ratio of 0.9 (Veeger et al., 1992b). All wheels were lightweight Spinergy Spox wheels complete with Primo V-Trak tyres, inflated to 120 psi. Seat height was standardised by measuring the degree of elbow extension elicited in participants' own sports wheelchair, as previously conducted in Chapters 5, 6 and 7. This degree of elbow extension was then replicated in the adjustable sports wheelchair and was maintained across wheel size conditions by making minor adjustments to the seat height (Table 8.1). Rear-wheel camber was also kept constant at 18° due to the favourable performance of wheelchair athletes in this configuration (Chapter 5 & 6). However, due to the use of a fixed length camber bar, the distance between TDC of both main wheels differed slightly between conditions (24" – 0.496 m; 25" – 0.477 m; 26" – 0.468 m). A velocometer (Moss et al., 2003) which sampled at 200 Hz was attached to the wheelchair throughout testing and was recalibrated for each wheel size condition. This enabled the velocities, accelerations and decelerations to be calculated. Wireless timing gates (Brower, Utah, USA) were used to record the times taken to perform each of the field tests.

8.3.3 Experimental design

Testing took place in a sports hall with wooden sprung flooring to replicate the playing surface used during WCB competition. To examine the influence of different wheel sizes on sports specific mobility performance, participants performed three field tests, as adapted from Chapter 6. Participants were allowed sufficient time to familiarise themselves with the drills and the wheelchair prior to data collection, which commenced when the participants indicated a state of readiness. The 20 m sprint and linear mobility drill were performed once in each wheel size condition, whereas the agility drill was performed twice with the resultant times averaged. The order of the field tests were kept constant, however the order for each wheel size condition was randomised amongst subjects. Sufficient rest in between trials was ensured to prevent the effect of fatigue, with participants responsible for indicating their physical state of readiness.

8.3.4 Statistical analyses

Means and standard deviations (SD) were computed for all variables using the Statistical Package for Social Sciences (SPSS Version 16.0; Chicago, IL). A two-way ANOVA with repeated measures was performed on each performance variable to examine any differences between wheel size conditions and to establish whether any interactions existed between wheel size and classification level. Pairwise comparisons with a Sidak adjustment to the alpha level were conducted to identify where any significant differences occurred. All data were accepted to be statistically significant at $P < 0.05$. Effect sizes were calculated to establish the meaningfulness of any differences, whereby $r > 0.5$ reflected a large effect size (Cohen, 1992).

8.4 RESULTS

Figure 8.2 revealed that wheel size had a significant effect on the time taken to perform the 20 m sprint ($P = 0.017$). Quicker times were observed for the 26” condition (5.58 ± 0.43 s; $P = 0.029$, $r = 0.76$) compared to the 24” condition (5.72 ± 0.40 s).

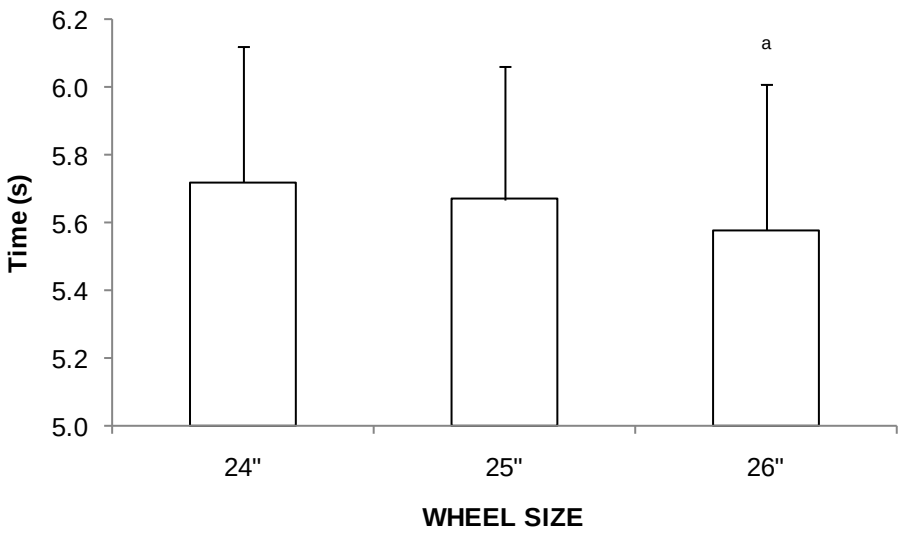

Figure 8.2 – The effect of wheel size on 20 m sprint times. [a] represents a significant difference to the 24" condition.

The performance parameters that were derived from the velocometer during the 20 m sprint were documented in Table 8.2. The mean and peak velocities reached were both affected by wheel size. Mean velocity was higher in 26" wheels compared to 24" (P = 0.003, r = 0.82) and 25" (P = 0.064, r = 0.65) wheels. Although not statistically significant, peak velocities were also greater in both 26" (P = 0.078, r = 0.63) and 25" wheels (P = 0.080, r = 0.60) compared to 24" wheels. A significantly reduced number of pushes were employed in the 26" condition compared to the 24" condition (P = 0.033, r = 0.67), with a meaningful effect also observed for 26" wheels to require a reduced number of pushes than 25" wheels (P = 0.054, r = 0.64). No differences were revealed for stroke frequency, as calculated from the number of pushes performed in relation to the 20 m sprint times in each wheel size condition. Initial acceleration performance was also unaffected by wheel size over the first two or three pushes from a standstill.

Table 8.2 – Mean (± SD) values for aspects of mobility performance measured during the 20 m sprint in different wheel sizes.

	***P* values**	**24”**	**25”**	**26”**
Mean velocity ($m \cdot s^{-1}$)	0.001	2.99 (0.20)	3.07 (0.25)	3.11 (0.19)[a]
Peak velocity ($m \cdot s^{-1}$)	0.022	4.61 (0.40)	4.63 (0.46)	4.77 (0.46)
Number of pushes	0.004	12.6 (1.6)	12.4 (1.3)	11.7 (1.3)[a]
Stroke frequency ($strokes \cdot s^{-1}$)	0.133	2.20 (0.34)	2.19 (0.28)	2.13 (0.26)
Acceleration over 2 pushes ($m \cdot s^{2}$)	0.299	2.10 (0.49)	2.25 (0.47)	2.21 (0.35)
Acceleration over 3 pushes ($m \cdot s^{2}$)	0.145	1.76 (0.34)	1.87 (0.35)	1.83 (0.26)

Key: [a] significant difference to the 24” condition

The push where peak velocity was achieved and the two preceding pushes were also analysed to determine how accelerations and absolute changes in velocity were affected by wheel size increments, during the push and recovery phases (Table 8.3). It was revealed that the mean rate of acceleration during the push phase and the decelerations during the recovery phase were unaffected by wheel size during maximal effort sprinting. However, greater absolute changes in velocity were evident during the push phase in 26” wheels compared to 24” wheels (P = 0.011, r = 0.67). A greater decline in velocity also existed during the recovery phase in 26” (P = 0.004, r = 0.82) and 25” (P = 0.018, r = 0.73) wheels compared to 24”.

Table 8.3 – Influence of wheel size on changes in wheelchair velocity during peak velocity propulsion. All values are means (± SD).

	***P* values**	**24”**	**25”**	**26”**
Mean accelerations ($m \cdot s^{2}$)	0.539	5.25 (1.97)	5.40 (2.57)	5.43 (2.86)
Mean decelerations ($m \cdot s^{2}$)	0.136	5.18 (1.90)	5.15 (1.06)	4.71 (1.39)
Push phase change in velocity ($m \cdot s^{-1}$)	0.002	0.93 (0.26)	0.98 (0.32)	1.12 (0.38)[a]
Recovery phase change in velocity ($m \cdot s^{-1}$)	0.001	-0.82 (0.27)	-0.91 (0.33)[a]	-1.04 (0.38)[a]

Key: [a] significant difference to the 24” condition

Figure 8.3 displays the intra-push profiles taken from one participant during the 20 m sprints, which demonstrates the differences in the velocity profiles between the 24” and 26” wheel sizes. Noticeable differences included the reduced overall times taken, number of pushes performed and higher peak velocities in 26” wheels. The greater absolute change in velocity during the push and decline during the recovery phases can also be observed in 26” wheels.

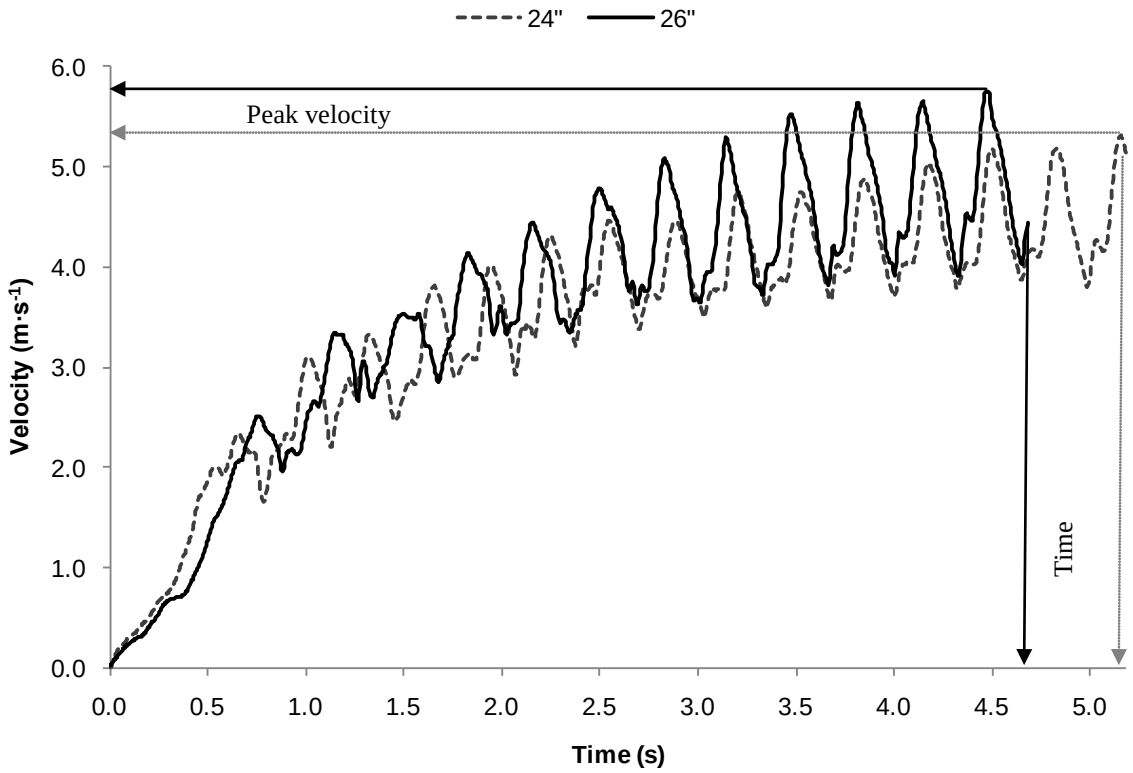

Figure 8.3 – A typical velocometer trace illustrating the intra-push profiles of one participant's 20 m sprint in 24" and 26" wheels.

Wheel size was not shown to have any significant bearing on the time taken to perform the linear mobility drill (P = 0.630) or the agility drill (P = 0.505). Time taken to perform the linear mobility drill (24" = 15.46 ± 1.03 s; 25" = 15.36 ± 1.15 s; 26" = 15.50 ± 1.25 s) and the agility drill (24" = 11.05 ± 0.58 s; 25" = 11.10 ± 0.67 s; 26" = 11.18 ± 0.87 s) were subsequently similar across wheel size conditions. No significant interactions were revealed between wheel size, classification level and any aspect of mobility performance. In addition to this, no between-subject effects existed across classification levels for any of the performance measures.

8.5 DISCUSSION

This study revealed that manipulating wheel size, whilst maintaining fixed gear ratios, affected the performance of a 20 m linear sprint, yet had no bearing on the performance of a linear mobility or agility drill in a group of highly trained WCB players. The times taken to

complete the 20 m sprint were significantly reduced in 26” wheels compared to 24” wheels. Factors contributing to the improved performance in 26” wheels appeared to stem from the higher mean and peak velocities attainable in this condition. This confirmed the original hypothesis that larger wheel sizes would improve maximal effort sprinting performance. However, the hypothesis that an improvement in sprinting performance with larger wheels would occur at the expense of impaired initial acceleration performance was rejected, since no significant differences were observed between wheel size conditions over the first two or three pushes.

Differences in vehicle mechanics between wheel sizes may have contributed towards the observed performance differences. The increased rolling resistance that is experienced in smaller wheels (Kauzlarich and Thacker, 1985) was shown to increase the P_O requirement from the user during sub-maximal wheelchair propulsion at constant velocities (Chapter 7). Increased resistances have been shown to have a negative impact on the mean velocity achieved during wheelchair sprinting (Veeger et al., 1991; Hintzy-Cloutier et al., 2002; Faupin et al., 2004). Therefore the impaired mean velocities in the 24” wheel size seemed likely to be associated to the greater resistance experienced in this condition. The increased P_O associated with 24” wheels during the sub-maximal investigation (Chapter 7) was unfavourable due to the increased physiological demand it induced. Alternatively, during maximal effort performance a higher peak P_O is often desirable due to the positive relationship it shares with sprinting performance (Coutts and Stogryn, 1987; Coutts, 1994; Dallmeijer et al., 1994; Woude et al., 1998). Although the peak P_O could not be quantified during the current investigation, the reduction in sprinting performance suggested that the increased rolling resistance and additional factors limited wheelchair velocity in the 24” condition.

A potential limiting factor during maximal effort sprinting in different wheel sizes may have resulted from coupling difficulties experienced between the user and the hand-rims. Coupling between the user and the hand-rims is a complex task, which is exacerbated at increasing linear velocities, as a shorter time span exists during which force can be transferred to a rapidly rotating hand-rim (Woude et al., 1998). The results of the current investigation suggested that coupling may be improved during maximal effort sprinting in larger wheels as a greater absolute increase in push phase velocity was observed in the 26” condition. During maximal effort wheelchair propulsion, an extremely limited variation in technique has been identified amongst wheelchair athletes (Roeleveld et al., 1994; Vanlandewijck et al., 1999b).

In addition to the highly standardised testing conditions imposed, it was anticipated that the elite athletes sampled may also have experienced minimal adjustments in technique between wheel size conditions. This was supported to some extent by the fact that no changes were observed in stroke frequency between conditions. It was also proposed that a 'ceiling effect' was likely to exist with regards to the maximal joint accelerations and velocities during maximal effort sprinting. This restriction on kinematic parameters, particularly linear hand velocity could be an important limiting factor due to its role in generating high tangential hand-rim velocities (Woude et al., 1988b). If the velocity of the hand is inferior to the angular velocity of the hand-rim, a braking force is applied, which causes a negative dip in the torque and power produced (Sanderson and Sommer, 1985; Veeger et al., 1991; Woude et al., 1998). A qualitative examination of the velocity trace presented (Figure 8.3) reveals a more abrupt impact peak during maximal sprinting pushes in 24" wheels compared to 26" wheels. The dip in velocity at the beginning of the push phase may be the result of a greater braking force that is applied in smaller wheels due to difficulties with coupling. A higher angular hand-rim velocity is of further importance in smaller, fixed gear ratio wheels in order to develop higher angular velocities at the wheel and to attain a similar velocity to larger wheels (Chapter 7). Based on the fact that 24" wheels demonstrated lower mean and peak velocities and lower changes in velocity during the push phase, it was proposed that sufficiently high angular hand-rim velocities are unachievable. This was thought to be strongly related to a minimising effect that occurs to joint accelerations and velocities during maximal effort sprinting.

A potentially similar propulsion technique may also result in an upper limit of force that can be applied during maximal effort sprinting. Torque around the hub is a very important indicator of sprinting performance (Roeleveld et al., 1994). Yet during constant velocity propulsion a greater magnitude of force and subsequently torque was required in smaller wheels to maintain the velocity in comparison to larger wheels (Chapter 7). If a similar *Fres* was applied across wheel size conditions due to the existence of an upper limit, then a greater torque would be anticipated in larger wheels, given its greater radius (Coutts, 1990). This could explain the improved sprinting performance resulting from the higher mean and peak velocities in the 26" wheel size. However, further kinetic data, which was not attainable within the scope of the current investigation, would be required to confirm or reject this hypothesis.

The current study established that wheel size had no significant effect on initial acceleration, as had originally been hypothesised. For a given force applied, a smaller wheel

would accelerate quicker due to the lower rotational inertia that would be experienced (Coutts, 1990). This was not observed within the current investigation which may be down to the highly trained nature of the participants sampled. It was possible that these athletes possessed sufficient strength to overcome the potentially small increments in rotational inertia in larger wheels. Again, the availability of hand-rim kinetic data would be able to verify this hypothesis and would be recommended for future investigations.

As previously mentioned, wheel size had no influence on stroke frequency during the 20 m linear sprint, which was in accordance with what was revealed during constant velocity, sub-maximal propulsion (Chapter 7). The consistent stroke frequency resulted from seemingly proportionate increases in both the number of pushes required and the times taken to perform the 20 m sprint in smaller wheels. The fact that a greater number of pushes were required to cover a set distance in smaller wheels may have permeations on injury risk. Using the data of Bloxham et al. (2001) and Sarro et al. (2010), it may be estimated that wheelchair court sport athletes perform maximal effort sprinting over approximately 400 m during competition. Extrapolating the results of the current study to such distances revealed that a 7.5% increase in the number of pushes would be required in 24” wheels compared to 26” wheels during match play. Therefore the potential for overuse injuries may be increased in smaller 24” wheels.

No interactions were observed between wheel size, classification level and performance during any of the field tests. This implied that optimal wheel size selection may not be dependent on classification level in WCB players, as had previously been proposed (Hutzler et al., 1995). Independent of wheel size, classification level was also shown to have no influence on any aspect of mobility performance, suggesting that factors other than classification affect individual’s optimal wheel size. This reiterated what was observed between classification groups during sub-maximal wheelchair propulsion in different wheel sizes (Chapter 7). Physical characteristics such as strength may be of benefit for future studies to examine in relation to wheel size and performance in the identification of optimal settings.

A limitation with the current investigation may be linked to the number of assumptions that were made when interpreting the results. Numerous contributing factors have been proposed as a rationale for the observed performance differences between wheel sizes, however these underlying mechanisms could not be directly quantified. This was largely due to the difficulties associated with the collection of biomechanical data during

maximal effort, over-ground propulsion. Also, during maximal effort propulsion responses are harder to predict and interpret, since variables such as velocity and/or workload are not controlled outside of standardised laboratory conditions. It has been reiterated throughout the discussion that a hand-rim kinetic investigation would have facilitated the interpretation of the results. Instrumented hand-rims such as the SMART^{Wheel} (Chapter 7) would have enabled the calculation of force magnitude, torque and angular velocity, which would not only have added a greater underpinning to the performance results but would also have offered a more detailed assessment of injury risk between wheel sizes. Unfortunately, the mass of a SMART^{Wheel} is currently far greater (approximately three fold) to that of a conventional sports wheel. This could have severely compromised the validity of testing when attempting to investigate the effects of wheel size in a sports specific environment and subsequently, further developments to such devices would be warranted. Nevertheless, the current investigation has provided extremely valuable information about the performance effects of wheel size adjustments on aspects of mobility performance, which has never previously been investigated. This information provides athletes, coaches and manufacturers alike with some evidence based data to inform their wheel size selections, which had previously been based on trial and error (Chapter 3).

Future investigations extending the current research would also be recommended to investigate the role of manipulating gear ratios in different wheel sizes. Decreasing the diameter of the hand-rim in relation to a given wheel size reduces the gear ratio (Veeger et al., 1992b). Veeger et al. (1991) suggested that a reduction in gear ratio may improve the transfer of joint angular velocities, thought to reach a maximum level during the current investigation, into wheelchair velocity. A reduction in gear ratio reduces the tangential velocity of the hand-rims, which can limit coupling issues between the user and the hand-rim (Veeger et al., 1992b). Woude et al. (1988b) and Costa et al. (2009) revealed that participants struggled to maintain the highest test velocity in higher gear ratios, potentially as a result of coordination problems with the higher tangential hand-rim velocities. Therefore, if participants could be instructed to accelerate from a standstill by pushing on the wheel, the larger diameter here would enable greater torque production for a given force. Once the initial moment of inertia has been overcome, switching to hand-rim propulsion on a reduced diameter hand-rim may further improve maximal sprinting performance.

8.6 CONCLUSION

The current investigation revealed that for the group of highly trained WCB players investigated, 26” wheels appeared to be the optimal setting based on the improved sprinting performance demonstrated in combination with no negative effects on acceleration or manoeuvrability. The improved performance in this wheel size was thought to be due to a combination of the lower rolling resistances experienced and improvements in coupling resulting from the lower angular velocities of the hand-rims that were expected at high propulsion velocities. The current results complemented the findings of Chapter 7 whereby it was identified that in addition to being less economical, 24” wheels also performed unfavourably for aspects of maximal effort mobility performance. Given that 69% of the participants self-selected either 24” or 25” wheels (Figure 8.1) the practical value of the current findings to athletes and coaches is demonstrated. Given the linear relationship between maximal effort mobility performance and the range of wheel sizes currently investigated, future investigations would also be advised to include 27” wheels, to determine whether further benefits result from increased wheel size.

Chapter Nine

General Discussion

The aim of the current thesis was to investigate how specific areas of wheelchair configuration can optimise the ergonomics of sports wheelchair performance in elite court sport athletes. As outlined in section 1.2, three main objectives were developed:

i) To investigate the self-selected configurations of elite wheelchair athletes to subjectively evaluate how modifications to configuration are perceived to affect mobility performance and to establish key areas of wheelchair configuration worthy of empirical research.
ii) To analyse the effects of manipulating several key areas of sports wheelchair configuration on the physiological and biomechanical responses of elite wheelchair athletes during sub-maximal wheelchair propulsion.
iii) To analyse the effects of manipulating these key areas of wheelchair configuration on the performance of maximal effort, sports specific manoeuvres.

The aim of the current chapter was to discuss the outcomes of the experimental chapters, addressing each of the aforementioned objectives (section 9.1, 9.2 & 9.3) and to integrate the findings in the context of the overall aim of the thesis. Optimising the ergonomics of sports wheelchair performance through wheelchair configuration is dependent on the physical characteristics of the wheelchair user and the subsequent interaction between the wheelchair-user combination. An integrated, multidisciplinary approach is necessitated to reliably investigate the efficiency, safety/health, comfort and sports performance of different wheelchair-user combinations (Murrel, 1965; Woude et al., 1989a). Figure 9.1 was adapted from the conceptual model originally developed in Figure 1.1 to illustrate the factors that influence the ergonomics of sports wheelchair performance and to highlight those that have been considered throughout the experimental chapters within the current thesis.

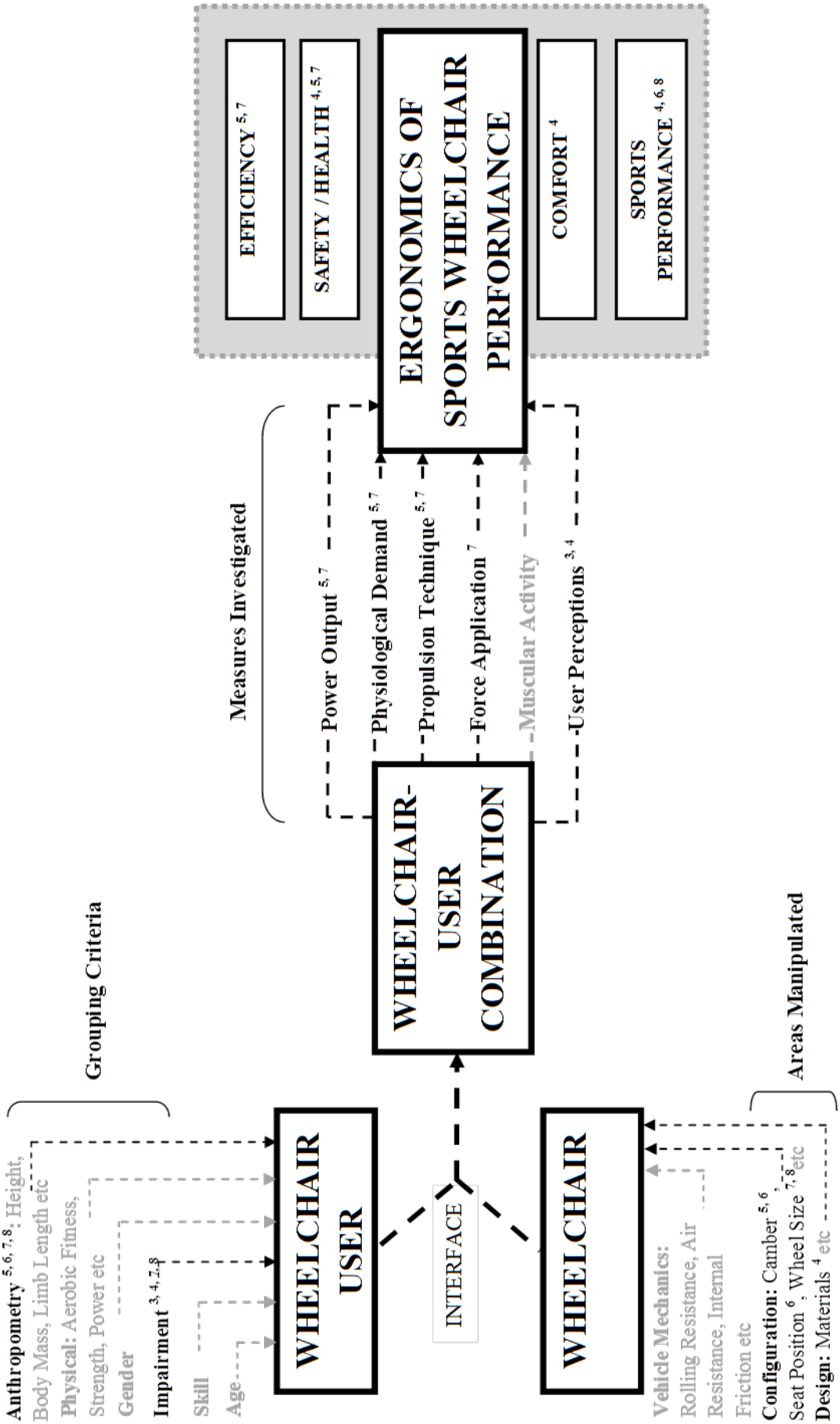

Figure 9.1 – Conceptual model adapted from Figure 1.1 illustrating the areas of the wheelchair-user combination investigated throughout the current thesis and the measures and approaches used to assess the ergonomics of sports wheelchair performance. Superscript numbers represent the relevant experimental chapter during which each component was examined.

9.1 Understanding the self-selected configurations of elite wheelchair athletes

Due to the numerous components to a sports wheelchair it was first necessary to develop an awareness of the key areas of wheelchair configuration. Given the limited evidence based data, this was addressed using a qualitative approach into highly trained athletes' perceptions on wheelchair configuration and mobility performance (Chapter 3). It was revealed that athletes strongly considered how wheelchair configuration affected elements of their ball handling (WCB and WCR) or stroke production (WCT). Athletes' selections were also largely influenced by factors including classification, the demands of the sport and their specific role during competition, as had previously been suggested (Yilla et al., 1998). Although these were all important considerations, the overall focus of the current thesis was to establish the effects of wheelchair configuration on mobility performance. Remarkable consistency was demonstrated amongst athletes' perceptions relating to the effects of adjusting the majority of areas of wheelchair configuration on aspects of mobility. Despite this unity, athletes' perceptions were rather ambiguous with a general understanding of how manipulating configuration affected performance. Unfortunately, when attempting to ascertain optimal configurations within an elite group of athletes, more detailed information is required and hence the rationale for the research conducted throughout the current thesis.

The primary aim of Chapter 3 was to establish where any discrepancies or uncertainties existed between athletes' perceptions. This identified the areas of wheelchair configuration in greatest need of empirical research and was extremely valuable in the context of the current thesis, as it served as a foundation for the ensuing experimental studies. Rear-wheel camber generated the most conflict amongst participants, particularly relating to its influence on linear mobility performance. This replicated the diversity in findings that have been established from the limited evidence based research into this area of configuration (Veeger et al., 1989b; Buckley and Bhambhani, 1998; Faupin et al., 2002; 2004; Perdios et al., 2007). Improving the coupling between the user and the hand-rims, along with the effects of wheel size and gear ratios were identified as important by athletes, but not consistently understood. In addition to this, the vertical and horizontal positioning of the seat were also identified as vital areas of sports wheelchair configuration, yet determining the optimal positioning again appeared an imprecise process.

9.2 The effects of manipulating key areas of wheelchair configuration on the physiological and biomechanical responses during sub-maximal propulsion

It was evident from Chapter 3 that elite wheelchair athletes predominantly focused on aspects of maximal effort performance when considering the effects of wheelchair configuration on mobility. The influence of wheelchair configuration on sub-maximal or endurance based performance was often overlooked, with the economy (or efficiency) of wheelchair propulsion only considered on one occasion, pertaining to the effects of wheel size. It was highlighted in Chapter 1 that manufacturers of sports equipment often neglect other important ergonomic considerations, such as efficiency and safety/health at the expense of performance optimisation (Frederick, 1984; Reilly and Lees, 1984; Reilly and Ussher, 1988). It seemed clear that elite athletes adopted a similar approach with regards to wheelchair configuration. However, aerobic conditioning is of particular importance to athletes competing in the wheelchair court sports (Goosey-Tolfrey, 2005; Roy et al., 2006; Bernardi et al., 2010), since high intensity propulsive actions only account for 10-20% of the movements performed during competitive game time (Coutts, 1992; Bloxham et al., 2001). The fact that the distances covered and the mean velocities maintained by elite WCR players decreases significantly during the second half of match play, reiterated the merit in maximising aerobic capacity through wheelchair configuration (Sarro et al., 2010).

The laboratory based investigations (Chapter 5 & 7) allowed such information to be ascertained in a controlled environment. During sub-maximal wheelchair propulsion it was revealed that larger camber settings (20° and 24°) and smaller wheel sizes (24”) increased the P_O requirements from the user. During sub-maximal propulsion, P_O increments are unfavourable due to the associated increases in physiological demand (Woude et al., 1988a). These unfavourable effects on physiological demand were also observed with the larger camber settings and smaller wheel sizes, with increases in both $\dot{V}O_2$ and HR evident. It was also apparent that the relative increases in P_O were greater than the relative increases in $\dot{V}O_2$ in these settings since ME was shown to improve in both larger camber and smaller wheels. Subsequently, ME was proposed as an inappropriate measure of efficiency when comparing different wheelchair configurations amongst wheelchair athletes, due to its dependence on P_O. The use of ME is subsequently a better indicator of efficient performance between conditions with constant P_O, as has previously been utilised during manipulation studies which have not affected P_O, such as seat height (Woude et al., 1990; Samuelsson et al., 2004; Woude et al.,

2009) or have controlled P_O (Veeger et al., 1989b). However, in order to assess the effects of manipulating wheelchair configurations in the most ecologically valid environment, controlling P_O is unfavourable, since it was clear that the manipulations studied (camber and wheel size) both influenced P_O, yet during competitive, over-ground propulsion this cannot be controlled. Alternatively, during constant speed protocols (yet varying P_O), as employed in Chapters 5 and 7, pushing economy is a more valid indicator of efficient performance. Pushing economy would subsequently be recommended in future wheelchair configuration investigations that evoke different workloads.

The kinematic analyses in Chapters 5 and 7 revealed that camber and wheel size adjustments had extremely limited effects on both the temporal parameters and the segmental excursions of the upper body during wheelchair propulsion. With regards to the temporal parameters, only PA was shown to be affected by increments in wheel size, whereby larger wheels reduced the PA. However, due to the greater circumference of these wheels and the consistency between other temporal parameters such as PT and stroke frequency, it was purported that wheel size manipulations had no effect on the linear trajectories of the push phase. In addition to the temporal parameters, no adaptations in the upper body joint excursions were observed between different wheel size conditions. Only the flexion/extension RoM of the shoulder and elbow were shown to be affected by camber increments, both of which increased during the 24° condition. The absence of any dramatic changes in the kinematics of wheelchair propulsion during Chapters 5 and 7 were the likely results of both the strict standardisation methods of the wheelchair-user combination characteristics (other than the manipulated experimental factor) and the highly trained status of the participants. Highly trained wheelchair athletes, such as those investigated throughout the current thesis, possess a highly reproducible propulsion technique (Dallmeijer et al., 1994; Roeleveld et al., 1994; Vanlandewijck et al., 1994b; 1999b). It was proposed that under the highly standardised and sub-maximal protocols employed, adjustments in both rear-wheel camber and wheel size were not substantial enough to significantly affect elite athletes' propulsion technique.

The use of an instrumented hand-rim ($SMART^{Wheel}$) in Chapter 7 was a novel and innovative approach into the ergonomics of wheelchair configuration, which has never been adopted before in sports wheelchairs with the relevant sports specific range of configurations. It enabled a more detailed investigation into the ergonomics of sports wheelchair performance evoked by changes in wheel size. A greater *Mz* was required in smaller wheels to overcome the greater P_O requirements, which culminated in a larger mean *Fres*, *Ftan* and amount of

work per cycle being performed in 24” wheels. The availability of such data facilitated the interpretation of the physiological results. The increased physiological demand experienced in 24” wheels was likely to result from a greater magnitude of force being required in these wheels in order to compensate for the greater rolling resistances and P_O requirements. Hand-rim kinetic data also offered a valuable insight into the safety/health of propulsion in different wheel sizes. In addition to being an inefficient wheel size, the 24” condition was also shown to place users at a slightly greater risk of injury. This was due to the larger magnitude of forces generated during standardised, sub-maximal wheelchair propulsion and the tendencies for these forces to develop at greater rates. Obtaining information relating to the magnitude of force application and the rate of force development is particularly valuable given the relationship of both to injury risk (Boninger et al., 1999; 2002) and the importance of user health/safety in ergonomics (Murrell, 1965; Reilly and Lees, 1984).

In summary, the laboratory based investigations established differences in the physiological and biomechanical responses to sports wheelchair configurations, which provided valuable practical information to athletes, coaches and manufacturers alike. Given the increased P_O and physiological demand associated with larger camber settings (20° and 24°) and smaller wheels (24”), it may be suggested that the selection of both these configurations should be avoided in the same wheelchair to prevent these elements becoming exacerbated. Also, the limited adaptations observed in propulsion kinematics through highly standardised adjustments to wheelchair configuration implied that it may be advisable to minimise the number of areas of configuration that are manipulated in comparison to the previous wheelchair. This was likely to diminish the likelihood of altering technique and unnecessarily increasing the risk of injury.

9.3 The effects of manipulating key areas of wheelchair configuration on the performance of maximal effort, sports specific manoeuvres

Investigating the effects of different wheelchair configurations under sub-maximal laboratory conditions facilitates the interpretation of the results due to the controlled environment it creates (Woude et al., 1989a; Vanlandewijck et al., 2001). However, the wheelchair court sports are categorised as intermittent activities requiring high levels of anaerobic capacity (Coutts, 1992, Bloxham et al., 2001; Sporner et al., 2009). Therefore, taking an ecologically valid approach to examine the effects of wheelchair configuration

necessitated that the performance of maximal effort, sports specific movements were assessed during over-ground propulsion. Movements such as initial acceleration, sprinting, manoeuvrability and braking are all key indicators of mobility performance specific to the wheelchair court sports (Vanlandewijck et al., 2001). The inclusion of the velocometer during the field based investigations (Chapters 4, 6 & 8) allowed these specific elements to be objectively analysed in greater detail.

9.3.1 Initial acceleration

Modifications to areas of the wheelchair-user combination investigated had no significant affect on initial acceleration performance over the first two and/or three pushes from a standstill (Chapters 4, 6 & 8). This was somewhat unexpected based on the perceptions of elite athletes (Chapter 3) regarding each of these areas of the wheelchair-user combination. When the wheels are stationary, it appeared as though the strength possessed by the highly trained athletes investigated was sufficient to overcome any changes in resistive force imposed by modifications to the wheelchair-user combination. Although other factors were likely to have an impact, it was purported that initial acceleration may be more significantly affected by alterations in the overall mass of the wheelchair-user combination, which can have more of a substantial effect on the resistive forces experienced (Fuss, 2009). However, given the improvements in technology over recent years, a substantial reduction in the mass of court sport wheelchairs have been observed, implying that further reductions in mass may be challenging (Ardigo et al., 2005; Burkett, 2010). It must be acknowledged that initial acceleration was assessed during one-off bouts in each field based investigation. This may be regarded as a potential limitation, since repeated bouts of acceleration are performed during wheelchair court sports competition (Coutts, 1992; Bloxham et al., 2001; Sporner et al., 2009). The effect of the increased P_O demands associated with larger camber and smaller wheels may have more of an impact on repeated acceleration performance over time.

9.3.2 Linear sprinting

In contrast to initial acceleration, a greater number of differences were observed in maximal linear sprinting performance as a result of the modifications to the wheelchair-user combination. Likely mechanisms for these differences existed during both the push and

recovery phase of the propulsion cycle. During maximal effort sprinting, the wheels and hand-rims rotate at very high rates, which introduce complexities in coordination and coupling between the user and the wheelchair. If sufficient friction cannot be developed at the wheelchair-user interface or the linear hand velocity is lower than the angular velocity of the hand-rim, an unfavourable braking force can be imparted (Sanderson and Sommer, 1985; Veeger et al., 1991; Woude et al., 1998). Chapter 4 identified improved sprint times in gloves that had been modified by each participant (CON) compared to a generic glove developed specifically for WCR (HYB). Not only did this indicate the importance of customising equipment to the individual needs of each athlete, but it also suggested that CON improved the coupling during the push phase at high tangential hand-rim velocities. Fixed gear ratio wheel size was also thought to have a strong bearing on coupling at maximal wheelchair velocities. Greater negative dips in wheelchair velocity were experienced at the onset of the push phase in smaller wheels, with lower absolute increases in velocity also observed. This suggested that difficulties in coupling during maximal effort propulsion were exacerbated in smaller wheels due to the need to develop higher angular hand-rim velocities in order to reach the same velocity of a larger wheel.

Alternatively, modifications to rear-wheel camber suggested that the observed changes in maximal sprinting performance between conditions may have occurred from resultant changes during the recovery phase (Chapter 6). It was previously mentioned that highly trained wheelchair athletes may possess sufficient strength to overcome the increased P_O requirement evoked by different wheelchair configuration settings, albeit at a physiological cost. The increased P_O requirements were the likely result of an increased rolling resistance, which cannot be influenced by the user during the recovery phase. Therefore the increased rolling resistance in the 24° camber setting may have been responsible for the increased the rate of deceleration in between pushes during the recovery phase, which ultimately may have limited the maximal linear sprinting performance.

9.3.3 Manoeuvrability

Numerous aspects of wheelchair configuration were perceived to have an influence on manoeuvrability performance during the qualitative investigation, including rear-wheel camber and wheel size (Chapter 3). However, it was not anticipated that glove type would have such a bearing on this area of mobility performance. Yet as demonstrated in Chapter 4,

CON were shown to improve manoeuvrability times in comparison to all other non-modified glove types. Given that the agility drill used to assess manoeuvrability performance included a rolling start, it could be suggested that sprinting performance was assessed more than acceleration during this drill. Therefore, it may have been that manoeuvrability was also improved through an enhanced coupling in CON when high tangential velocities would have been experienced at the hand-rims, as had previously been proposed (section 9.3.2). Rear-wheel camber was the main area of configuration thought to affect manoeuvrability performance (Chapter 3), with greater degrees associated with improved manoeuvrability. This was also supported by the limited empirical research, whereby increments from 9° to 15° camber increased turn velocity (Faupin et al., 2002). However, it was revealed in Chapter 6 that increments in excess of 18° camber yielded no further improvements in manoeuvrability performance.

9.3.4 Braking

Braking performance, defined as the ability to decelerate the wheelchair, was another key indicator of mobility performance that has never been accounted for during maximal effort wheelchair propulsion (Vanlandewijck et al., 2001). Although not statistically significant, Chapter 4 revealed that glove type displayed strong trends for braking performance to be improved in CON, particularly when compared to HYB. Alternatively no affect of camber was revealed for this facet of performance. Subsequently alterations directly to the wheelchair-user interface, such as glove type and/or hand-rim configuration may have more of an effect on braking performance than manipulating areas of wheelchair configuration.

9.4 Summary

The evidence based approach of the current thesis was a necessary and valuable step towards improving the ergonomic understanding of sports wheelchair performance through modifications in wheelchair configuration. In order to establish optimal wheelchair configurations from an ergonomics perspective all aspects of sub-maximal and maximal effort performance must be considered. Therefore the current experimental chapters have provided a valid environment to answer the research question of the thesis. The inclusion of equipment

such as the SMART^{Wheel} and the velocometer in both environments has enabled innovative and novel data to be collected to supplement the validity of the research. Table 9.1 summarises the main effects from each of the manipulation studies and compares the results to those that were anticipated by highly trained wheelchair athletes (Chapter 3). It was clear that a large percentage of athletes' perceived responses to such adjustments differed to those established from the objective data. Therefore, athletes' perceptions and selections pertaining to their current wheelchair configurations maybe inhibiting their efficiency and/or performance in addition to placing themselves at an increased risk of injury.

To further facilitate the identification of optimal wheelchair configurations throughout the current thesis participants were grouped by classification (Chapters 4, 7 & 8) and self-selected seat height (Chapter 6). The influence of glove type was investigated from a purely sports performance perspective, although a subjective appraisal of each gloves safety and comfort was obtained. It could be interpreted that players own specifically modified gloves (CON) were optimal for the demands of WCR, given that aspects of maximal effort mobility performance were maximised (sprinting, manoeuvrability, braking and subjective comfort) without any negative effects on other areas (initial acceleration and subjective safety).

Rear-wheel camber was investigated from both a sub-maximal and maximal effort mobility perspective, which although improved the validity of any optimal configurations identified, also increased the complexity in doing so. During sub-maximal wheelchair propulsion, 15° and 18° camber appeared favourable due to the lower P_O and physiological demand in relation to 24°. The 20° setting also appeared unfavourable from a sub-maximal perspective given the increased P_O and elevated HR responses compared to the 15° camber setting. However, the 20° camber setting appeared beneficial during aspects of maximal effort mobility performance, as demonstrated by the superior 20 m sprint times compared to 24°. Alternatively, the 15° camber setting which appeared favourable during sub-maximal propulsion displayed deficiencies in maximal effort manoeuvrability performance compared to the 18° setting. The results of Chapters 5 and 6 suggested that 18° camber seemed like the most optimal setting from an ergonomics perspective of sports wheelchair performance, based on its superior performance in both aspects of sub-maximal and maximal effort mobility. It also suggested that 24° camber seemed the least favourable configuration for the opposite reasons. However, interactions also existed between camber and seat height to suggest that athletes with a high seating position may be suited to slightly greater camber due to improvements in linear mobility and braking performance. This was likely to be due to the

greater degree of stability increased camber afforded these players. This highlighted the difficulties in terms of optimising areas of wheelchair configuration, yet identified important considerations for wheelchair athletes, coaches and manufacturers alike.

In accordance with camber, wheel size was also investigated during controlled sub-maximal propulsion and maximal effort, over-ground propulsion. The sub-maximal examination revealed that both 25” and 26” wheels were favourable from an efficiency perspective due to the reduced P_O, physiological demand and work per cycle compared to 24” wheels. From a safety/health perspective 24” wheels also seemed unadvisable given the larger magnitudes of *Fres* and the strong trends for these forces to develop at a greater rate. The field based investigation supported the notion that 24” wheels was an unfavourable setting due to the inferior performance they displayed for aspects of linear sprinting compared to 26” wheels. The 25” wheel size also appeared less favourable during maximal effort mobility performance than it did during sub-maximal performance, with reduced mean velocities displayed versus the 26” condition. Therefore, 26” wheels appeared like an optimal configuration for a group of highly trained WCB players based on the favourable efficiency and safety/health responses exhibited and for the superior maximal effort linear sprinting performance. This setting was also established as optimal for this group of athletes due to the absence of any negative effects on initial acceleration and manoeuvrability performance as had initially been hypothesised.

Grouping criteria used during Chapters 4, 7 and 8 also established that classification level had an impact on the ergonomics of sports wheelchair performance, independent of wheelchair configuration. During Chapter 7, significant differences in force application, joint excursions and temporal kinematics were revealed between HP and LP WCB players, although this did not affect maximal effort mobility performance (Chapter 8). Alternatively, during Chapter 3, HP WCR players were shown to perform superiorly for all aspects of maximal effort mobility performance compared to LP. Therefore, it appeared as though classification level had more of a bearing on maximal effort mobility performance in WCR players than WCB players. This may have been due to the increased severity of impairment in these individuals. Subsequently, it may be more beneficial for WCB players to be grouped by other criteria in Figure 9.1, such as strength in order to facilitate the identification of optimal configurations in this population group.

Table 9.1 – A summary of the perceived responses of elite athletes to glove, camber and wheel size adjustments compared to the sub-maximal and maximal evidence based data generated throughout the relevant experimental chapters.

		Perceived responses to self-selected gloves	Self-selected gloves	Perceived responses to increased camber	Increased camber (15° - 24°)	Perceived responses to increased wheel size	Increased wheel size (24" – 26")
		Chapter 3	Chapter 4	Chapter 3	Chapters 5/6	Chapter 3	Chapters 7/8
Sub-maximal - Laboratory	Power output				↑		↓
Physiology	Mechanical efficiency				↑		↓
	Oxygen uptake				↑	↓	↓
	Pushing economy				↓	↑	↑
	Heart rate				↑		↓
	Rating of perceived exertion				=		=
Kinematics	Push times				=		=
	Stroke frequencies				=	↓	=
	Push angles				=	↑	↓
	Active RoM:						
	Trunk – sagittal				=		=
	Shoulder – sagittal				↑		=
	Shoulder - frontal				=		=
	Elbow - sagittal				↑		=
	Wrist – sagittal				=		=
Kinetics	Resultant force				?	↑	↓
	Tangential force				?		↓
	FEF				?		=
	Rate of force development				?		= (↓ trends)
	Torque				?		↓
	Work per cycle				?		↓
Maximal - Field Performance	Sprinting - linear times	↓	↓	↑ / =	↑	↓	↓
	Initial acceleration	↑	=	↓ / =	=	↓	=
	Mean velocity		?		↓		↑
	Peak velocity	↑	=		=	↑	↑
	Decelerations between pushes		?		↑		=
	Braking	↑	= (↑ trends)		=		?
	Manoeuvrability		↑	↑	=	↓	=

Key: ↑ increase; ↓ decrease; = no change; ? not measured

9.5 Future directions

Given the novelty of the research conducted throughout the current thesis, there are obviously several key areas of study that future investigations could build on to further knowledge on the topic of sports wheelchair configurations. Table 9.1 highlighted the fact that there are still some unknowns even after the evidence based research presented within the current thesis. The availability of a force sensing hand-rim during Chapter 5 would have proved extremely useful in order to establish the force application patterns in different camber settings. This would not only have facilitated the interpretation of the physiological data, but it would also have provided a more detailed insight into the injury risk associated with each camber condition. Investigating the upper body joint kinematics provided an exploratory investigation into injury risk. However, by examining the magnitudes and rates of force development in conjunction with this, would allow for a more accurate assessment of safety/health to be obtained. Subsequently, future investigations into any area of wheelchair configuration would benefit from hand-rim kinetic data whenever possible.

The collection of hand-rim kinetic data during over-ground, maximal effort field testing would also be extremely advantageous. The velocometer used throughout the present field based investigations was invaluable for providing objective performance data about each area of wheelchair configuration in a sports specific environment. However, a recurring limitation throughout the current thesis was that the underlying mechanisms behind any of the changes in performance observed could not always be determined. The availability of kinetic data would again assist with interpreting the performance results and understanding the mechanisms behind why a certain configuration may affect performance differently to others. It would also enable an assessment of user safety/health to be ascertained during each configuration under the most realistic conditions and subsequently would have been a valuable addition to Chapters 4, 6 and 8. Unfortunately, this was not possible as it currently stands largely owing to the weight of current instrumented wheels such as the SMART^{Wheel}, which would affect the validity of the kinetic data collected. However, with developments in technology for Paralympic sports always improving (Burkett, 2010), lightweight devices suitable for use in the field may be imminent.

It was also clear from Figure 9.1 that EMG analyses can prove to be a valuable measure when considering the ergonomics of sports wheelchair performance. However, it was not possible to include such an analysis during any of the current experimental chapters.

Investigating the muscular activity of different wheelchair user combinations may provide a more detailed assessment of the efficiency and safety/health of certain configurations and its inclusion would subsequently be recommended by future investigations whenever possible.

One area that future investigations could build on from the sub-maximal and maximal effort responses to different wheel sizes (Chapters 7 & 8) would be to investigate the effects of different wheel sizes with different gear ratios. Chapters 7 and 8 revealed that larger wheel sizes with relatively high, fixed gear ratios (0.9) reduced the P_O, physiological demand and magnitude of force application and improved the maximal linear sprinting performance. The use of different diameter hand-rims is something that has been investigated in the wheelchair racing literature (Woude et al., 1988b; Gayle et al., 1990; Guo et al., 2006; Costa et al., 2009). Woude et al. (1988b) revealed that during fixed wheel sizes, reducing the hand-rim diameter and consequently the gear ratios from 0.7 and 0.83 to as little as 0.44, 0.52 and 0.56 significantly reduced the physiological demand and facilitated coupling at higher velocities, due to the lower tangential velocity of the smaller hand-rims. Therefore, potential benefits could be derived from both a sub-maximal and maximal effort perspective by manipulating wheel size and gear ratio in unison.

The current thesis examined some of the most prominent areas of wheelchair configuration that were in the greatest need of evidence based research from a sporting perspective. Yet, there still remain a number of areas of configuration that would benefit from further investigation. The vertical and horizontal positioning of the seat was identified as another important area of consideration when athletes configure a new sports wheelchair (Chapter 3). Chapter 7 revealed that differences existed between classification groups for the biomechanical responses to sub-maximal wheelchair propulsion. Strong correlations were also made between classification and self-selected seat height in this study, suggesting that the latter could also strongly impact on the ergonomics of sports wheelchair performance. Seat position has received a fair amount of previous research as documented in Table 2.1, although numerous deficiencies existed predominantly due to the daily life configurations, AB participants and/or lack of standardisation methods imposed. Future investigations would be encouraged to implement the standardisation methods previously employed for seat height (Woude et al., 1989b; 1990; 2009) and fore-aft (Hughes et al., 1992; Wei et al., 2003) adjustments to the anthropometrics of the user and to also consider their effects on maximal effort, over-ground propulsion.

The experimental chapters conducted throughout the current thesis have provided researchers, athletes, coaches and wheelchair manufacturers alike with some useful information regarding the consequences of adjusting certain areas of wheelchair configuration on mobility performance. Generic findings and recommendations have been derived to assist athletes with how certain manipulations can affect their performance. Grouping methods were imposed to add further specificity to these findings relating to the classification and sitting height of individual athletes. However, it was apparent that limited interactions between group, wheelchair configuration adjustments and performance were observed in the experimental chapters. Therefore future investigations may be advised to establish whether more suitable ways for grouping wheelchair athletes can be identified in order to further enhance the specificity of the findings. Alternatively, this may prove too complex an issue given the huge inter-individual differences that exist between wheelchair users (Goosey-Tolfrey and Price, 2010). Subsequently, in order to optimise the ergonomics of sports wheelchair propulsion, an individualised case study approach may also be appropriate when exploring elite wheelchair sports performance.

Until further research has been conducted, the evidence based findings from the current thesis would advise athletes, particularly those who are new to wheelchair sport or inexperienced, to configure their sports wheelchairs with 18° camber and 26" wheels. Also, for WCR players, it is advisable to modify a self-selected glove type to best suit the individual's specific requirements to optimise wheelchair handling skills.

Chapter Ten

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