Optimisation of the Rugby Wheelchair for Performance

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Abstract

Equipment design in para-sport has a substantial impact on athlete performance. Subsequently, wheelchair designs have progressed to reflect the requirements of their sports; for wheelchair rugby, this has resulted in features including reinforced frames to withstand the frequent high impacts and cambered wheels for improved agility and stability. Whilst these aspects of wheelchair design have advanced, there is currently no accepted method for optimising an individual's wheelchair configuration (e.g., setting of seat height/seat angle); instead, players rely on their previous experience and support staff in trial-and-error approaches to prescribing set-ups. This is likely due to a number of factors, including: the range of impairment types and severities in the sport, hence optimal set-ups differing across players; difficulty in assessing on-court performance and propulsion kinematics; limited knowledge of the effects of set-up parameters on key performance and propulsion factors; and the substantial time and cost associated with new chair prescriptions. To address this issue, this research aims to improve the knowledge regarding the effect configuration parameters have on performance and propulsion in wheelchair rugby.

To achieve this, an improved understanding of current player set-ups and their propulsion approaches is required. Large participant groups (n=16 and 25, for set-up and propulsion analysis respectively) allowed for statistical assessments based on classification groups (high-, mid-, and low-point groups). Significant differences were found in both set-up and propulsion approaches across classifications. The majority of these differences reflect the levels of the player's activity limitation (i.e., high-point players with greater trunk range of motion used flatter seat angles, and contacted the wheel closer to top dead centre than low-point players). Additionally, a potential trend

towards increasing release angles and greater peak accelerations was identified. More detailed individual assessments of propulsion were also performed that revealed variations in intra-stroke acceleration profiles of three players. This information can aid in wheelchair prescription by identifying regions of strength for an individual, with this then emphasised by the wheelchair set-up.

To assess the effect of set-up parameters on performance and propulsion measures, a robust design approach using an adjustable wheelchair was implemented with six elite players. This approach required reduced amounts of field testing whilst maintaining the ability to identify the effect of the specific settings of seat height, seat depth, seat angle, and tyre pressure. Half the players reported a blinded preference for a recommended set-up following this testing, while remaining players reported a preference based on 'comfort' despite similar results.

Finally, a linkage model and regression approach were developed that accounted for individual anthropometrics, propulsion approach, and wheelchair set-up and successfully predicted a performance measure for some players. Overall, this research has improved the knowledge surrounding the effect of wheelchair rugby set-up parameters on performance and propulsion at both group and individual levels. Optimisation of wheelchair set-up should occur at an individual level and consider functional abilities and on-court role; approaches such as the robust design and modelling methods presented in this thesis improve the ability to achieve this in practise.

Contents

Declaration

I certify that this work contains no material which has been accepted for the award of any other degree or diploma in my name, in any university or other tertiary institution and, to the best of my knowledge and belief, contains no material previously published or written by another person, except where due reference has been made in the text.

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I acknowledge the support I have received for my research through the provision of an Australian Government Research Training Program Scholarship.

Name: David Haydon Date: August 23rd, 2018

Signed:

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Glossary

Abbreviation	Term
AccFrame	Frame acceleration
AHRS	Attitude and Heading Reference System
ContAng	Contact angle
CycTime	Cycle time
EMG	Electromyography
GPS	Global positioning system
ICC	Intra-class correlations
IMU	Inertial Measurement Unit
ITS	Indoor tracking system
MDL	Miniaturised data logger
MEMS	Microelectromechanical systems
PCA	Principal component analysis
PLS	Partial least squares
RecTime	Recovery time
RelAng	Release angle
SA	Seat angle
SDep	Seat depth
SH	Seat height
StrokeAng	Stroke angle
StrokeTime	Stroke time
TDC	Top dead centre
TP	Tyre pressure
WCB	Wheelchair basketball
WCR	Wheelchair rugby
WMP	Wheelchair mobility performance

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Chapter 1: Introduction

This chapter provides a brief background of the current knowledge on the effect of wheelchair configuration on performance in para-sport, particularly in wheelchair rugby, and the issues and challenges in optimising wheelchair configurations at an individual level. Structure and details of the thesis are provided to conclude the section.

1.1 Overview

Para-sport has received increased levels of published research in recent years [1-3], as well as greater exposure at major events such as the Paralympic and Commonwealth Games. Due to the nature of athlete impairments in para-sport (e.g., impaired muscle power due to spinal cord injuries, limb deficiencies, hypertonia or athetosis due to cerebral palsy) and adapted rules, there is often a greater reliance on equipment than in able-bodied sport, with this adding another layer of complexity in achieving high levels of performance. Improvements in equipment design can have substantial effects on performance: for example, energy-storing sprinting prostheses can even provide a mechanical advantage over able-bodied athletes [4], while specific designs for seated throwing frames [5], and racing wheelchairs [6] have also increased performance standards.

Wheelchair court sports – wheelchair rugby (WCR), wheelchair basketball and wheelchair tennis – have been amongst the most investigated para-sports [1, 7-10]. Research in wheelchair sport has typically focused on three key areas: (i) the athlete; (ii) the wheelchair; and (iii) the athlete-wheelchair interaction. The athlete relates to the physical and psychological capabilities of the individual, such as the severity and type of impairment. The wheelchair relates to purely mechanical aspects of the wheelchair; these may include the overall design of the wheelchair (i.e., racing or rugby wheelchair) or factors such as mass or rolling resistance, which can be altered largely in isolation from the individual. The third aspect (athlete-wheelchair interaction) requires consideration of how changing the wheelchair will affect the individual and their on-court performance. For example, altering the angle of the seat will place the athlete in a different position and therefore likely alter their interaction and force applied to the handrim and/or wheel. This project will focus primarily on the athlete-wheelchair interaction in WCR; while there is a need to consider some general aspects of the wheelchair, consideration of the physiological adaptations is beyond the scope of the project.

Individual chair optimisation in WCR refers to the set-up of the chair to reflect the individual impairment and team role, both of which can differ substantially across athletes. Impairments often include impaired muscle power from spinal cord injuries (primarily at the cervical level) and limb deficiencies of various severities [11] among others, and subsequently affecting the on-court roles to which players are best suited. An individual's sport-specific activity limitation is accounted for using a classification system that allocates a player a point-score ranging from 0.5 to 3.5 points. In practice and research environments, players are often separated into low- (0.5-1.5 points), mid- (2.0-2.5 points), and high- (3.0-3.5 points) classification groups. This provides benefits in assessing trends in performance factors such as distance covered or number of passes for players with varying classifications.

Chair designs in WCR have primarily focussed on the functional aspects of the chair to reflect the requirements of the sport: characterised by intermittent power, with short sprints and frequent changes of direction, as well as aerobic demands [12-14]. The high-impact nature of WCR has resulted in reinforced frames and 'hooks' that promote trapping of opposition players. Sports wheelchairs in general are typically designed to allow for improved manoeuvrability in comparison with daily wheelchairs.

1.2 Challenges

In addition to the various impairments and on-court roles, optimisation of chair configuration has been limited by the ability to perform detailed analysis of propulsion under conditions representative of athletes' competition demands – due to instrumentation and testing approaches. Previous work has focused on the optimisation of wheelchair court sports chairs [3, 15-17]; however, these have often focused on single parameter studies across group situations often in highly controlled laboratory environments including on ergometers or treadmills [2, 18]. These are important and add to the experiential knowledge among coaches and athletes but are often difficult to apply in practical settings. Due to the range of performance requirements of the sport and the various trade-offs for wheelchair parameter settings (e.g., increasing seat height might improve ball handling but limit the players access to the pushrim/wheel [12]), optimisation of configuration should consider the effects of a range of parameters. Additionally, optimising individual parameters one at a time requires substantial time commitments. For elite players, it often takes multiple years and wheelchairs to find a near optimal set-up. Major adjustments to wheelchair set-up when ordering new wheelchairs are rare due to the risk of causing detrimental changes to performance (and associated substantial cost), while small changes (i.e., increasing or decreasing seat depth) are difficult (and in some cases impossible) due to the reinforcement of frames to withstand the large impacts. While testing multiple parameters at once can reduce the time requirements associated with testing, it can introduce difficulties in assessing the specific parameters that cause changes in performance. This is one of the strengths of the controlled laboratory methods, where clear cause and effects are visible. Effective assessments of various parameters in a time- and cost-efficient manner is therefore clearly a crucial practical component of determining optimal set-ups in WCR.

The ability to accurately measure on-court activity (including detailed tracking and propulsion) has remained a large obstacle in quantifying performance. Recent work in this area has improved these capabilities [19, 20], with the potential for detailed assessments of on-court movement now improved through technology such as inertial measurement units and video recording and processing advancements. These assessments have provided increased insight into requirements for physiological training, and allowed researchers to quantify differences in performance. For example, performance mobility measures identified for wheelchair basketball included average and best rotational and linear accelerations and peak speeds [3]. Due to similar on-court requirements to WCR [12], it is expected that the majority of these mobility measures could aid in the assessment of WCR performance. Improving knowledge of these areas allows for improved assessment of on-court performance, an important consideration when optimising wheelchair configurations.

Despite the improved ability to monitor mobility performance, there remains limited understanding of propulsion approaches that are used in WCR. Propulsion approaches refers to the temporal and angular parameters within a wheelchair stroke – i.e., how long they contact the wheel/pushrim and where on the wheel they contact and release. The impact of these parameters has received limited research in terms of their effect on overall and within stroke performance. Assessments of propulsion approaches across a large representative sample have not been reported for this sport, with this information an important first step to the understanding of propulsion variations across players. Understanding various individual propulsion approaches and athlete perceptions can potentially aid in selection of wheelchair configurations, with stronger regions of an individual's stroke – likely related to their trunk and arm activity limitation – maximised by adjusting parameters such as seat depth or angle. While assessments of performance changes for some configuration parameters have been performed [16, 21], these have not considered the resultant changes in propulsion technique.

To accurately assess changes in propulsion, testing approaches need to be representative of on-court activity. For WCR, there should be an emphasis on the first 3-4 strokes of a linear sprint as players rarely travel extended periods at maximal effort without a change in direction [13]. There is now increased awareness of the importance of representative test designs [7] and less reliance on testing conditions using ergometers or treadmills [21] which have been shown to alter propulsion technique and physiological responses of athletes [22]. These alternate testing philosophies may allow greater translation of testing results to on-court performance. Whilst testing protocols have been developed for wheelchair basketball, there is currently no accepted method for assessing a range of performance factors through testing protocols in WCR.

In summary, optimising WCR configurations currently has limited quantitative consideration of performance and propulsion approach for various set-up parameters and requires substantial time commitments. This primary aim of this work is therefore to:

> *Develop a greater understanding of the influence of wheelchair configuration parameters on WCR performance and propulsion at an individual level.*

To achieve this, the following objectives were formulated:

1. Improve understanding of propulsion approaches across a range of players and the impact various approaches have on performance.
2. Improve testing assessment methods for on-court performance in WCR.
3. Implement and assess methods that reduce the time associated with wheelchair prescription under practical conditions.

This objectives were achieved throughout numerous studies, including investigations into robust design approaches [18] and on-court, representative test designs utilising improved instrumentation approaches and analysis methods. This will improve the ability for coaches and support staff to provide detailed information to individual players on current or future set-ups with quantitative support. Achieving an optimal set-up through reduced time and testing can improve performance of elite players, but also allow new players to find an appropriate configuration earlier in their development.

1.3 Thesis Outline

To effectively optimise the wheelchair configuration for an individual, knowledge regarding their current wheelchair set-up, their propulsion technique, and how key parameters affect performance and propulsion is required. To address the objectives developed and achieve the overall aim of the thesis, six research studies were performed which are presented as the following chapters (Chapters 3 to 8). Brief outlines of these studies are provided below, detailing the process of how this information was obtained and reporting key findings at both group and individual levels.

Chapter 2: Literature Review

A detailed assessment of current knowledge for key areas of this research is performed. This includes on-court instrumentation approaches and testing, wheelchair propulsion, the effect of set-up parameters on performance, and current modelling approaches for wheelchair propulsion. Gaps in the literature are identified and used to inform the direction of this research.

Chapter 3: Elite Wheelchair Rugby: A Quantitative Assessment of Chair Configuration in Australia

Despite a focus on configurations in WCR in previous literature, there has been no assessment of wheelchair configurations across a large elite population. This chapter reports on the range of configurations across classification groups in WCR in an elite population, as well as player reported expectations how altering these parameters would affect performance. Documenting current player configurations and how they perceive effects of configuration changes will aid in future understanding of performance changes with configuration.

Chapter 4: Overground Propulsion Kinematics and Acceleration in Elite Wheelchair Rugby

Limited research exists for propulsion approaches that are used within wheelchair court sports. This work aims to develop a better understanding of WCR propulsion approaches across a range of impairment types during maximal acceleration from standstill. This information aids in assessing propulsion techniques used by individual athletes and how this is likely to influence their acceleration from standstill – a crucial aspect of WCR on-court performance.

Chapter 5: Intra-stroke Acceleration Profiling of Elite Wheelchair Rugby Players

While *Chapter 4* provides insights into propulsion approaches of classification groups, intra-stroke profiling improves the ability to assess various accelerations within a stroke at an individual level. Through the use of inertial measurement units, the wheelchair acceleration throughout the first three strokes of acceleration from standstill was investigated. In conjunction with video analysis, this allows for hand position at maximum and minimum acceleration to be identified. Three case studies of players from varying classification groups reveals substantial difference in intra-stroke acceleration profiles. Understanding how a propulsion stroke influences wheelchair acceleration is vital when optimising wheelchair configuration as regions of increased acceleration can be maximised for each individual.

Chapter 6: Test Design and Individual Analysis in Wheelchair Rugby

Due to the improvements in monitoring capabilities in wheelchair court sports (e.g., inertial measurement units, processing algorithms), greater understanding of on-court activity is possible. Testing protocols should therefore attempt to be representative of on-court activity, rather than utilising highly controlled testing protocols. A small change in test design to be more representative of on-court performance is compared with a standstill sprint. Comparisons of sprint time to five metres, peak accelerations, and propulsion angles reveals substantial differences between the two testing protocols. Testing protocols should, therefore, carefully consider how to enhance current test designs to ensure the translation of test findings to on-court performance.

Chapter 7: Wheelchair Rugby Chair Configurations: An Individual, Robust Design Approach

Using knowledge developed from previous chapters, this chapter focuses on the use of a robust design approach to optimising wheelchair set-ups. Using a custom designed adjustable wheelchair, testing was able to manipulate configuration parameters including seat height, seat depth, seat angle, and tyre pressure. To assess the effects of these parameters while reducing testing time requirements, an orthogonal design was implemented that reduced the number of tests required whilst maintaining the ability to identifying the impact of specific parameter settings for each individual (e.g., how increasing seat height affects performance). Propulsion changes are monitored, and performance is assessed by test times, acceleration changes in linear sprints, agility, and ball-handling testing. The overall method is presented for six players of varying physical impairment, with a detailed process presented for one player as a case-study. This method provides the potential to achieve improved wheelchair set-ups with reduced testing for both elite and developing players.

Chapter 8: Predicting Sprint Performance in Wheelchair Rugby using a Linkage Model

Despite the decreased testing requirements presented in *Chapter 7*, this method still requires 3-4 hours of detailed testing for each individual athlete.

Advancing previous propulsion linkage models used for submaximal propulsion to predict maximal effort propulsion parameters has the potential to further reduce testing requirements. A linkage model was advanced to account for trunk motion and was then implemented to predict individual propulsion kinematics and the associated wheelchair set-up. This allowed for the prediction of propulsion kinematics based on individual anthropometrics, wheelchair set-up, and the individual's propulsion approach. Principal component analysis and regression approaches were then used to assess relationships between chair set-up, propulsion kinematics, and performance for future predictions of linear sprint time. Comparisons between predicted and actual performance show potential benefits for using this approach. Future work should focus on increasing the quality and quantity of data to improve the reliability of prediction methods.

1.4 Thesis Details

This thesis is submitted in a *Thesis by Publication* format in accordance with requirements of the School of Mechanical Engineering at The University of Adelaide. The journal papers which follow satisfy the standard requirements, with additional journal articles and conference proceedings from this research also listed. All journal articles are included in the main body of the thesis (including the *Journal Articles* listed in *Additional Outcomes*).

1.4.1 Journal Articles

Satisfying University Requirements

1. **Haydon, D.S.**, Pinder, R.A., Grimshaw, P.N., Robertson, W.S.P., 2018. Overground Propulsion Kinematics and Acceleration in Elite Wheelchair Rugby. Int J Sports Physiol Perform 13, 156-162. DOI: doi.org/10.1123/ijspp.2016-0802. (Chapter 4)
2. **Haydon, D.S.**, Pinder, R.A., Grimshaw, P.N., Robertson, W.S.P., 2018. Test design and individual analysis in wheelchair rugby. J. Sci. Med.

Sport. In Press, published online 7 April, 2018. DOI: doi.org/10.1016/j.jsams.2018.04.001. (Chapter 6)

3. **Haydon, D.S.**, Pinder, R.A., Lewis, A.R., Grimshaw, P.N., Robertson, W.S.P., Under Review. Intra-Stroke Acceleration Profiling of Elite Wheelchair Rugby Players. Submitted to International Journal of Sports Medicine, August 2018. (Chapter 5)
4. **Haydon, D.S.**, Pinder, R.A., Grimshaw, P.N., Robertson, W.S.P. Predicting Sprint Performance in Wheelchair Rugby Using a Linkage Model. Submitted to Journal of Biomechanics, August 2018. (Chapter 8)

1.4.2 Additional Outcomes

Journal Articles

1. **Haydon, D.S.**, Pinder, R.A., Grimshaw, P.N., Robertson, W.S.P., 2016. Elite wheelchair rugby: a quantitative analysis of chair configuration in Australia. Sports Eng. 19, 177-184. DOI: doi.org/10.1007/s12283-016-0203-0. (Chapter 3)
2. **Haydon, D.S.**, Pinder, R.A., Grimshaw, P.N., Robertson, W.S.P., 2018. Rugby Wheelchair Chair Configurations: An Individualised, Robust Design Approach. Submitted to Sports Biomechanics, August 2018. (Chapter 7)

Conference Papers

3. **Haydon, D.S.**, Pinder, R.A., Grimshaw, P.N., Robertson, W.S.P., 2018. Using a Robust Design Approach to Optimize Chair Set-up in Wheelchair Sport. Proceedings 2, 482. Presented at International Sports Engineering Association Conference, 2018, Brisbane, Australia.

Conference Presentations

4. **Haydon, D.S.**, Pinder, R.A., Grimshaw, P.N., Robertson, W.S.P., 2016. Propulsion in Elite Low-point Classification Rugby Wheelchair Athletes. Poster at International Society of Biomechanics in Sport 2016, Tsukuba, Japan.
5. **Haydon, D.S.**, Lewis, A.R., Pinder, R.A., Grimshaw, P.N., Robertson, W.S.P. Alterations in pressure distribution during agility activities in

wheelchair rugby. Poster at International Society of Biomechanics 2017, Brisbane Australia.

6. **Haydon, D.S.**, Pinder, R.A., Grimshaw, P.N., Robertson, W.S.P. Intra-stroke acceleration profile of an elite mid-point wheelchair rugby player. Poster at International Society of Biomechanics 2017, Brisbane Australia.
7. **Haydon, D.S.**, Pinder, R.A., Grimshaw, P.N., Robertson, W.S.P. Representative experimental and practice task design in wheelchair rugby. Presented by Ross Pinder at VISTA 2017, Toronto, Canada.

1.5 Summary

Current methods of optimising individual wheelchair configurations rely on anecdotal coach and player experiences. This results in players often being involved in the sport of WCR for multiple years before finding an appropriate wheelchair set-up. Determining optimal wheelchair set-ups has previously been limited by instrumentation approaches for on-court testing, limited knowledge of propulsion approaches, and time requirements for testing purposes. This work aims to improve knowledge in this area by assessing wheelchair configurations and propulsion techniques across a range of elite players. Detailed assessment methods of intra-stroke profiles and representative test designs are presented, with results aiding individual analysis of wheelchair set-ups. To reduce the time requirements for testing a range of wheelchair parameters, a robust design approach is initially presented that allows for assessment of a range of performance factors. Further, a potential method that accounts for individual anthropometrics and propulsion approach and predicts linear sprint performance based on changing wheelchair set-up is established. The combined impact of these studies produces greater knowledge across a range of areas, including: current configurations and propulsion approaches in elite WCR; acceleration profiles of wheelchairs for various strokes; impact of test design on performance; and the effect of altering wheelchair configuration settings on performance and propulsion. These advances increase the ability of coaches and technical support staff to provide support to players who wish to optimise their WCR chair configuration, potentially resulting in substantial improvements in performance while reducing the associated time and cost.

Chapter 2: Literature Review

This chapter provides a detailed literature review into the current state of knowledge regarding WCR. This includes on-court athlete and match demands, wheelchair design and instrumentation, and the interaction between the athlete and wheelchair. Interaction includes propulsion (where the user applies forces to the wheels, causing the rotation that results in motion), configuration effects on performance and propulsion, with various testing approaches and potential methods for reducing testing requirements also discussed.

2.1 The Athlete

2.1.1 Impairment and Classification

WCR is a fast-paced court sport for both male and female players with a wide range of health conditions. The aim is to carry a ball over the opponent's goal line in order to score. To achieve this, players require frequent short sprints and rapid changes of direction, as well as involving large impacts between players due to blocking [1]. Blocking includes 'screening' of opposition players to aid teammates in gaining space when their team is in possession of the ball, as well as attempting to prevent movement of opposition players when defending. The impacts between wheelchairs plays a major role in on-court success by forcing opposition players into positions and actions which aid offensive and defensive game plans. Each team is allowed four players on-court but are limited to a total of eight classification points [2]. Classification points are assigned to each eligible individual to account for their sport-specific activity limitation. For an individual to be eligible for WCR, they require an impairment that affects at least three limbs [3]. A 'classification' process assigns point scores based on trunk, arm, and hand *function* (where *function* refers to strength, range of motion, and coordination), which results in a player's overall point score [4]. Classification scores range from 0.5- to 3.5-points, where a larger classification indicates the reduced impact a player's impairment is likely to have on their on-court performance [2]. Players can be assigned similar classification scores despite having varying impairment types such as impaired muscle power (due to spinal cord injuries (SCI)) or limb deficiencies (amputations etc.) due to the various effects an impairment type and severity can have on on-court activity limitation [5]. For example, a player may have limb deficiencies that result in low scores for arm and hand function, but higher scores in trunk function. Conversely, a player with a SCI may receive a low trunk function score, but higher scores in arm or hand function. The classification process is currently under-going a transition to a more evidence-based approach [3, 6, 7], particularly for trunk function, to better reflect the impairment types now involved in the sport [7, 8]. Due to the range of impairments involved and limited number of elite players for these impairment types, it remains unclear how specific impairments affect performance and propulsion, as well as the effects various wheelchair set-ups have on these parameters.

2.1.2 Match Demands

High-point players (3.0- and 3.5-point scores) are typically tasked with ball-handling responsibilities, whereas low-point players (0.5- to 1.5-point scores) more often block opposition players to create space for the high-point players [9, 10]. Mid-point players often perform a combination of these two roles [11]. Performance factors for the various point scores therefore differ slightly, although there is a large overlap in features that are seen as preferential for performance. Acceleration from standstill has been reported by players to be one of the most important factors, as it allows players to either escape or execute blocks depending on their role [12]. Findings from Sporner et al. [13] support this, where analysis suggested an average of 242 stop-starts per game for players across all classifications (n=18 players). Manoeuvrability or agility is also crucial for all players, as performing fast turns either on the spot or whilst in motion again allows players to escape or execute blocks. However, ball-handling capabilities (i.e., catching, passing, dribbling) are of greater importance to higher-point players as they are usually the primary ball-handlers [4]. Conversely, low-point players generally place a greater emphasis on stability – often due to their reduced trunk function [12]. Wheelchair tracking (i.e., the ability of the wheelchair to accurately continue in the desired direction) has also been reported as a key performance factor for both high- and low-point players [14], as good tracking allows players to perform their role with reduced effort. Top end speed is viewed to be of lesser importance in WCR as players are rarely required to perform a straight-line sprint long enough to reach and maintain top end speed in match situations [12]. Using data from Sporner et al. [13], where distances travelled averaged 2364m per game, players travel only 9.8m per stop-start under match conditions. This emphasises the requirements of WCR are focused around short, fast changes of direction and speed, allowing players to avoid or catch the opposition.

Published work quantifying detailed wheelchair mobility during representative match environments has increased in the last 5 years. For example, Rhodes et al. [15, 16] and van der Slikke et al. [17] have used recent improvements in technology to assess key mobility measures in WCR and wheelchair basketball, respectively. These studies have supported early work by Sarro et al. [18] that found WCR players travel distances of 3500-4600m during matches. The relative distances covered by WCR teams that received mid and high rankings (i.e., the top teams in terms of world rankings) does not vary significantly, although low ranked teams had significantly more substitutions [15], potentially due to a reduced physical capacity. High ranked teams achieved greater peak speeds in match-play than both low and mid ranked teams [15], with this potentially revealing success is influenced by the ability to consistently perform at high intensities (i.e., importance of ability to repeatedly accelerate maximally). In terms of individual player mobility, van der Slikke et al. [17] investigated 22 kinematic outcomes in relation to forward and rotational movement in match-play for wheelchair basketball players. Principal component analysis (PCA) was then used to reduce the number of variables required to describe a player's wheelchair mobility performance. This resulted in 6 parameters being selected: 1) average of the best five rotational speeds in a turn (where a turn was defined as between linear speeds of -1.5 to 1.5 m/s); 2) average rotational acceleration; 3) average forward acceleration in the first 2m from standstill; 4) average forward speed; 5) average rotational speed in a curve (>1.5m/s forward speed); and 6) average of five best forward speeds. As WCB has similar requirements to WCR (albeit with less contact between wheelchairs), it is expected the majority of these variables are transferrable to WCR due to the similar sporting requirements [12]. This is supported by national WCR teams using testing protocols that incorporate a number of these variables, including full-court sprints, up and backs, and slalom testing [19]. These quantitatively determined performance factors support those reported by athletes and coaches [12, 14], reinforcing their importance to on-court performance in WCR; research should also aim to confirm this. Further work is required to advance on the work by Molik et al. [4] regarding technical performance (i.e., balls passed/caught) in WCR; this could include, for example, investigating the ratio of long and short passes completed by the various classifications [20].

2.2 The Wheelchair

2.2.1 Overview and Design

The design of rugby wheelchairs has progressed substantially from the initial wheelchairs used for WCR, which was developed in the 1970s [21], with numerous advancements resulting in wheelchairs that are clearly designed for WCR. Wheelchairs for court sports utilise a greater camber – the angle of the wheels from the vertical [22] – on the main wheels compared with daily or racing wheelchairs [23]. This allows for both improved stability and manoeuvrability, as well as increasing a player's ability to block opposition and hand protection. Wheel sizes typically vary from 24-26 inches across players, with low-point players generally preferring smaller wheels due to a perceived improvement in initial acceleration [12]. Due to the frequent collisions, rugby wheels utilise guards that protect the spokes from damage. In addition, players may use minimal push- or handrims; partly due to the wheel guards, which allow for increased contact between the hand and a purely spoked wheel. For high-point players, this may result in push- or handrims that are barely prevalent, with a large majority of contact occurring between the hand and wheel guard/tyre. Low-point players, due to their typically limited hand function, are still likely to utilise push- or handrims.

An additional consequence of frequent – and often large – impacts, is the structure of the frame. While other sports wheelchairs are designed with reduced structure to minimise mass, rugby frames are reinforced with additional beams to provide the required strength and durability. This leads to increased mass of rugby wheelchairs (16-20kg) compared with racing (~5-6 kg) and basketball (~9-10 kg, [24]) wheelchairs as well as typical daily living wheelchairs. Material selection plays an important role in determining both the strength and mass of the wheelchair, with current chairs often made from aluminium (potential to be heat treated) or titanium. While reductions in mass are generally associated with improving performance, large masses in WCR have the potential to help players hold their position during impacts. Therefore, consultation with players and coaches around such factors are crucial in optimising on-court performance.

Rugby wheelchairs have also undergone design transformations to reflect the varying on-court roles, resulting in two primary chair types: offensive and defensive. Offensive chairs are typically used by higher point players and are designed to avoid opposition blocks. They are therefore shorter and use shrouds over the front beams of the frame to prevent hooking [25]. In contrast, defensive chairs, typically used by low-point players, are longer and utilise bumpers that improve their ability to capture or disrupt the path of offensive chairs as seen in Figure 2.1.

Figure 2.1: Defensive chairs (left) are considerably longer and utilise a pick bar, whereas offensive chairs (right) are shorter and attempt to deflect blocking attempts.

The set-up of wheelchair parameters also differs, due to player activity limitation, anthropometrics, and on-court role. Major design parameters are presented in Figure 2.2, with the two major components of seat position and wheel design. Seat position contains parameters such as seat height (distance from ground to bottom of back of seat), seat depth (distance behind the centre of the wheel axle), and seat angle (angle of the seat above the horizontal); wheel design contains wheel size, camber position, tyre type and pressure. Low-point players typically utilise a seat position that priorities stability, whereas high-point players will use a seat position that allows for improved ball-handling. Further details on how specific parameters effect performance are provided in Section 2.3.2.

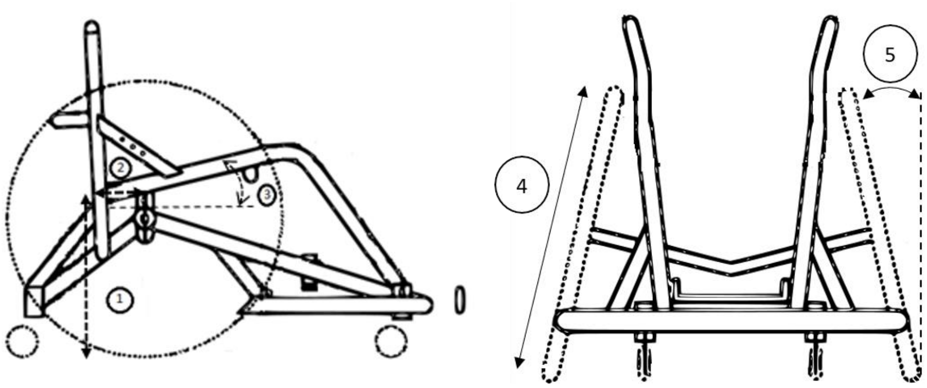

Figure 2.2: Key configuration parameters can be classified as influencing the seat position, such as (1) seat height, (2) seat depth, and (3) seat angle, or the wheel, including (4) wheel diameter and (5) camber angle.

While rugby wheelchairs have undergone a substantial transformation to relate specifically to on-court roles in WCR, further development of chair design is still required. This includes optimising the interaction between the athlete and wheelchair through altering parameters such as seat position and wheel design.

2.2.2 Instrumentation

Manual wheelchairs are often used by individuals who are suffering from a range of health conditions from spinal cord injuries to multiple sclerosis and cerebral palsy [26]. Due to the high usage level, a significant amount of work has been performed in an attempt to understand wheelchair propulsion tendencies (i.e., how forces are transferred to the wheels) with an aim to minimise the risk of injury by adjusting wheelchair configuration; however, a large majority of this research has focussed on assessing people in daily living, or in sports but with propulsion at a sub-maximal intensity [27-31]. As outlined above, WCR involves frequent maximal intensity efforts; hence it is important that research aims to address this obvious gap. Previous testing methods of wheelchair propulsion have also varied considerably, with on-court monitoring providing the most representative and relevant results [32].

However, monitoring propulsion kinetics and kinematics [33], as well as wheelchair motion [13], is difficult to achieve in representative match or on-court situations [32].

Instrumented wheel systems [34-36], and in particular the SMARTWheel [33, 37], provide the ability to monitor propulsion kinetics, which is important in identifying the risk of injury. Goosey-Tolfrey et al. [36] adapted their system for use with wheelchair sport, using a strain-gauge based measurement system developed for wheelchair racing. This allowed for measurement of forces (rotational and lateral) exerted on the handrim for varying speeds of propulsion on an ergometer. Increased stroke frequency was found at greater velocity, as well as slight increases in release and stroke angles [36]. While some requirements for instrumentation are consistent across wheelchair racing and court sports (e.g., small mass, wireless, etc.), such a system is not directly transferrable to WCR due to the use of a handrim (i.e. a protruding rim on the outside of the wheel used for hand propulsion of the chair). Wheelchairs in WCR often have a minimal handrim, with players often simply contacting the wheel for acceleration; hence, the introduction of a handrim for measurement of kinetic forces would alter the propulsion approach and therefore validity of the results. Additionally, this wheelchair racing system was not capable of measuring radial forces during a propulsion stroke. While the SMARTWheel is considered a gold-standard for clinical situations, it is not practical for use in WCR due to the similar use of a handrim, the additional mass, cost and transferability issues [38].

Due to the importance of monitoring on-court movement (i.e., for prescription of individual, specific training programs [16]), multiple tracking solutions have been attempted for wheelchair court sports. Global positioning systems (GPS) have become the most practical method for monitoring player movement in team sports [39], however are limited to outdoor use, while image-based processing techniques require time-consuming analysis post-event [40]. Alternative methods to these two popular systems have therefore been investigated. Early solutions used miniaturised data loggers (MDL) which were capable of monitoring variables such as distance travelled and the number of stop-starts [13]. However, the MDL used by Sporner et al. [13] utilised reed switch activation to monitor propulsion speed, with switches located at 60 degrees around the wheel – potentially resulting in low frequency measures. An alternative to this approach is the use of a radio-frequency based indoor tracking system (ITS) operating at 8-16Hz. This approach utilises ultra-wideband signals and requires stations to be positioned around the court that monitor the location and orientation of tags placed on players wheelchairs [16, 41]. Using such systems elicited mean horizontal positioning errors of 0.37m, distance travelled errors <0.5% [41], and peak speed errors <2.0% [16]. These favourable results support use in indoor sports in general, although the tracking frequencies available still do not allow for detailed analysis of speed and acceleration that have been identified as crucial for successful performance [15, 42].

Outdoor sports, such as wheelchair racing, can use GPS monitoring, but this remains limited by low frequency measurements. Higher frequency, in-field measurements have been possible through the use of a specially fitted telemetry-based velocometer. This device utilised an optical encoder to monitor wheel rotation [43] and was hence able to measure velocity throughout a linear sprint start as well as perform assessments on initial acceleration and braking [44]. As the unit used in that study was mounted on the frame using two aluminium alloy plates and secured with clamping blocks, it is difficult to transfer across wheelchairs. Additionally, the attachment adds substantial mass to the wheelchair system and lacks the ability to track multidirectional movements that are important in court sports.

Achieving higher frequency measurements is possible through the use of microelectromechanical systems (MEMS) technology [45]. Various approaches using associated sensors have been implemented, with single accelerometer methods used to identify different activities (no/low level activity, pushing, collisions) using fractal dimensions [46] and turning radii [47] in WCR. Bergamini et al. [48] also secured one inertial measurement unit (IMU) to a wheelchair, in addition to another on each of the player's hands. This approach allowed for a more detailed propulsion assessment of a 20m sprint, including factors such as bilateral symmetry and timing parameters during linear accelerations. In order to perform tracking assessments, IMUs have also been positioned on the wheels [42, 45, 49, 50]. Using accelerometer and gyroscope components of the IMU then allows monitoring of individual wheel rotation and frame orientation. Provided a correction is applied for the camber angle of the wheels [49], this information can then be processed in order to provide estimations of the frame motion. Further, van der Slikke et al. [50] reported that consideration of wheel skid is crucial to correct measurement of on-court motion – this is achieved through a correction when a threshold difference ($>2.5m/s^2$) was exceeded between the measured forward acceleration from the frame IMU and the derived forward acceleration measured from wheel IMUs. Using this approach for variables such as linear speed, rotation centres, and rotational speed, intra-class correlation coefficients (ICC) were >0.9 when compared with the gold standard of motion capture. However, when higher intensity exercise was performed including collisions and skidding, the system lost accuracy [42].

Shepherd et al. [45] utilised an Attitude and Heading Reference System (AHRS) algorithm with IMUs that provides a global orientation measure. An open-source implementation of such an approach by Madgwick et al. [51] was used to monitor both frame and wheel change in orientation, with these orientations then used to determine mobility variables. Good accuracy was again reported for distance, velocity, and trajectory estimations with mean errors of 1.62% across all trials [45]; however, these measurements were performed by researchers pushing an unmanned wheelchair at a relatively constant pace along a pre-defined path and did not consider the frequent changes in velocity and direction present in match-play.

In summary, recent work with IMU approaches has shown strong potential for tracking purposes in on-court situations – an improvement on previous instrumentation approaches. However, they are currently unable to provide important propulsion details such as contact and release position of the hand, and unable to provide detailed information on court location – important as movements may vary depending on location. The synchronisation of high-speed video with IMU measures has the potential to allow for detailed analysis of linear propulsion in representative on-court settings during maximal acceleration. This could provide greater insight into the propulsion methods used and how these effect performances.

2.3 Athlete-Wheelchair Interaction

2.3.1 Propulsion

Wheelchair propulsion considers the interaction of the hands and wheels that results in wheelchair motion. Wheelchair users exert forces onto the wheels that cause rotation: forward rotation of both wheels for forward movement; conversely backward rotation for backward movement; and for turning, greater rotation of one wheel which depends on the direction of the turn. Effectively converting the forces exerted onto the wheel into rotation (so users get maximum return on the physiological effort) is a key component of wheelchair propulsion. Despite wheelchair propulsion being a common method of mobility, propulsion approaches employed are generally inefficient with mechanical efficiencies of ~11.5% [52, 53], smaller than those for running (~50% [54]) and cycling (~20% [55]). This is likely due to the relatively low muscle mass of the upper limbs compared with the overall mass of the wheelchair/user [32]. Propulsion can be defined through a range of variables, particularly regarding the hand placement around the wheel. The hand position at contact can be defined as the contact angle and measured as the angle from top dead centre (TDC) of the wheel. The final position of the hand before it ceases contact with the wheel is also measured as the angle from TDC and defined as the release angle. These two measures can be combined and result in the stroke angle. Additionally, the time in contact with the wheel is the contact time and time between strokes is the recovery time; these times combined result in the cycle time. These kinematic measures help to define the propulsion approaches used by individuals and identify similarities or differences.

Assessment of propulsion approaches often includes investigating two components that are present within a single propulsion stroke: pull and push [56]. Pull refers to the region of a stroke in which the hand is in contact with the wheel and the elbow angle is decreasing – typically when the hand is moving up the wheel towards TDC (see Figure 2.3). Push is defined as the region following when the minimum elbow angle has been passed and the elbow angle is increasing. Proportions of these components provide further insight into propulsion approaches and are likely influenced by an individual's impairment. Whilst the push component occurs with triceps brachii extension, it can also be aided by gravitational forces to counterbalance the trunk-extension reaction forces generated by the hand pushing on the wheel [57]. However, using a greater proportion of push is likely to limit the stroke angle, hence the adoption of propulsion approaches towards being push oriented should be determined on an individual basis depending on the athlete's particular activity limitation.

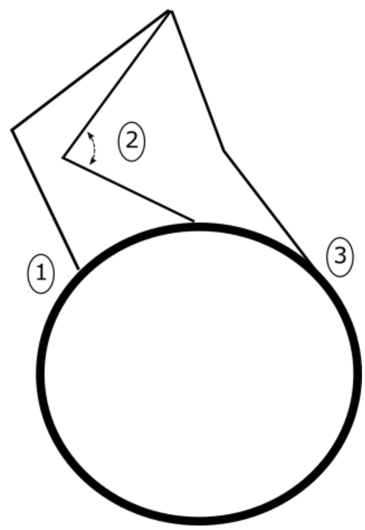

Figure 2.3: Regions throughout the stroke, where 1 is contact, 2 is the minimum elbow angle, and 3 is release. 'Pull' is from contact to minimum elbow angle, and 'push' is from minimum elbow angle to release.

Moss et al. [58] provides one of few examples to consider detailed within-stroke analysis for wheelchair sports. In this work, the first six strokes of a wheelchair racing start were monitored for a single athlete using a velocometer to track wheelchair velocity. For wheelchair racing sprint events, the start is crucial to performance, similar to the metric of repeated accelerations from standstill in WCR. Clear variations in propulsion kinematics such stroke time, recovery time, contact angle, and release angle were evident across the first three strokes, with signs of steady-state propulsion being reached for strokes five and six [58]. Despite the large variations in propulsion styles (racing athletes generally adopt a kneeling position, whereas WCR players adopt a more conventional seating position for ease of ball-handling), this analysis provides insights into the detailed assessments than can be performed on an individual's propulsion approach.

Differences in intra-stroke profiles in WCR would be expected to vary substantially due to the range in type and severities of impairment [2]. Knowledge of an individual's intra-stroke profile can potentially provide further information on how to adjust wheelchair set-up and seat position to emphasise the stronger regions of the stroke.

Despite the importance of the first three strokes to WCR performance, few studies have investigated kinematic variables under testing conditions representative of performance demands [6]. However, even studies that have investigated maximal effort propulsion from standstill have generally focused on either very specific features of the technique [59, 60] or changes in propulsion due to alterations to configuration [6]. Vanlandewijck et al. [6] investigated the relationship between seat angle, trunk motion, and acceleration in the first three seconds of acceleration from standstill for non-disabled, recreational wheelchair basketball and tennis players. Results showed that increasing seat angle reduced trunk motion and acceleration during this period. While this work provides insight into potential configuration effects, the study used able-bodied participants on a wheelchair ergometer. This test design was deemed appropriate for the assessment of trunk motion from an evidence-based classification perspective and was therefore not intended to provide detailed information on representative trunk kinematics in WCR. Recent work by Altmann et al. [61] in WCR does however support this work, with trunk impairment found to be most pronounced in the first metre of acceleration, with arm impairment becoming a larger influence than trunk impairment between 2 and 3 metres.

Yang et al. [60] and Schantz et al. [59] focused on the electromyography (EMG) activity for the trunk and upper extremity muscles respectively – although Yang et al. [60] tested with unimpaired participants while Schantz et al. [59] performed testing with daily living wheelchairs with para- and tetra-plegia participants. EMG recordings monitored the activation levels of muscles, allowing for improved analysis of particular muscle contributions and timings throughout propulsion. Although Schantz et al. [59] predominantly concentrated on upper extremity muscle activation times, the range of motion of the trunk throughout propulsion was identified. It was found that, during the initial pushes in acceleration from standstill, the trunk range of motion for participants with paraplegia was 20-30°. This greatly exceeds the common trunk movements involved in maximal velocity propulsion [59, 60]. It was hypothesised that this trend occurs as the trunk motion is efficient at generating force at low-speed but comes at a high energy cost to the individual and is therefore used sparingly.

Using the EMG data and synchronised video recordings, Schantz et al. [59] also focused on propulsion parameters such as push times and push angles. It was identified that participants shortened their contact time with the hand-rim during maximal acceleration, as well as using a larger proportion of pushing rather than a pulling method. The trend towards a push method of propulsion has been identified previously with increasing speeds [53] however this was at steady state propulsion rather than acceleration. Whilst these trends have been noted in steady state propulsion, to the author’s knowledge, the effect of various pushing strategies for maximal acceleration has not been investigated in WCR. Despite the inefficient nature of wheelchair propulsion, there is limited knowledge surrounding propulsion approaches or the influence of wheelchair configuration on propulsion and performance wheelchair configurations that are able to improve this [62, 63].

Further work is required to investigate maximal effort acceleration from standstill – particularly for WCR. Limited quantitative knowledge exists for the various propulsion approaches used by players with varying classifications and impairments, with this including kinematics (contact/release angles), technique (proportions of pull/push), and intra-stroke profiles. Additionally, the effect of wheelchair configuration changes on these propulsion parameters remains under researched.

2.3.2 Configuration Effects on Performance and Propulsion

Numerous wheelchair configuration parameters have been shown to substantially influence mobility performance [11, 32, 64, 65]. The majority of investigations into the effects of specific configuration parameters (i.e., seat height, seat depth) have been related to sub-maximal propulsion for daily living and reducing the possibility of injury for the user [65]. The focus on reducing injury risk differs to applied research in wheelchair sport where optimal performance usually focusses on increases in speed and acceleration while managing injury risk. More recently, assessments of the influence of parameters on maximal effort propulsion, such as that required in wheelchair court sports including WCR, has increased [66-69].

The two main aspects of wheelchair configuration that have been investigated relate to seat position and the main wheels. Seat position parameters include height, depth, and angle, while the main wheels can be altered in terms of size and camber [12, 65]. While the rate of published research has increased for wheelchair sports in recent years, a large proportion of wheelchair configuration research has been completed on propulsion in daily living [29, 30, 70-72]; consideration of previous research, therefore, still has some reliance on daily propulsion studies, despite the potential lack of translation to wheelchair sport. Further, when assessing parameter effects at an individual level, there is often a trade-off between performance factors. For example, reducing the seat height can increase the amount of hand-rim available to the individual but reduce the ball handling ability of players [12]; assessments of optimal parameter settings should therefore consider the individual player as well as their on-court role.

Seat Position - Height

Seat height has been shown to have a significant influence on propulsion in a number of sub-maximal studies [66, 73-75]. These studies found that lower seat position allows the individual more access to the hand-rim, increasing the possible stroke time and angle, and thus decreasing the push frequency [66], with reductions in push frequency thought to reduce the likelihood of injury [76]. From these studies, it is thought that the optimal seat height for daily propulsion produces an elbow angle between 100° and 130° when the hand is at TDC of the wheel [75, 77] – although this does not necessarily translate to wheelchair sports. The selection of this range of angles was based on measures of physiological performance (namely VO_2 and mechanical efficiency).

For WCR players, an increased seat height allows improved ball handling capabilities and a better view of the court [12, 78], which are important factors in performing game skills. With an increase in seat height, there is an associated increase in the height of the centre of mass; this makes the chair more susceptible to tipping and subsequently increases the risk of injury of players. There is, however, limited quantitative research into the effect of seat height on maximal effort propulsion. Usma-Alvarez et al. [64] identified seat height as the most important factor for performance in short sprints using a rugby wheelchair. Whilst an increased seat height resulted in improved acceleration, testing was performed on an ergometer, potentially altering the propulsion approach, and there was minimal consideration of propulsion kinematics. Walsh et al. [79] investigated seat height (and fore-aft position/ seat depth) effects on maximal linear velocity in a racing wheelchair but found no significant effects. More recently, van der Slikke et al. [80] investigated seat height effects in 20 elite wheelchair basketball players using the field test and IMU sensor set-up discussed previously [42, 81]. Seat height was increased and decreased by 7.5% of the initial setting, measured from the top of the head.

Seat height was found to influence forward and rotational movement, with both lowered and increased seat heights showing slight decreases in testing times compared with the neutral setting [80]. Further, the average forward speed in the wheelchair mobility performance (WMP) test showed a small decrease for the elevated seat condition in comparison with the lowered seat condition – essentially, an elevated seat resulted in a reduction in linear speed. This result demonstrates that while wheelchair mobility is a key consideration in wheelchair set-ups, sporting requirements such as improved ball handling remain the primary considerations.

An aspect that needs to be considered when altering seat height is the level of trunk function of the individual. For increases in seat height, greater trunk ranges of motion are required to achieve similar levels of access to the push-rim [82]. Hence increasing seat height for players with reduced trunk function is likely to reduce their access to the push-rim. Greater trunk motion is likely to result in decreased propulsion mechanical efficiency due to the increased activation of trunk flexor (rectus abdominis) and extender (erector spinae) muscles [82]; however, players (and coaches) are likely to prefer acceleration and manoeuvrability over efficiency in wheelchair sport. Previous research suggests that an optimal seat height exists for wheelchair propulsion and performance; while work has been conducted in this area, optimising this condition at an individual level in WCR still requires further attention.

Seat Depth

The horizontal position of the seat relative to the wheel axle, perhaps more commonly known as the balance point or fore-aft position, again has been shown to influence wheelchair propulsion and performance. Increased stroke angles can be achieved through anterior seat positions [25, 74, 83] as well as posterior seat positions [75, 76] compared to seat positions directly above the wheel axle. While this may have benefits for reducing stroke frequency and injury risk [27], the effects on performance on factors such as acceleration from standstill is unknown. These conflicting results likely result from different methods: anterior seat positions allow the user to reach further past TDC on the wheel, increasing the release angle; more posterior positions allow the user to access more of the rim before TDC and likely increase the contact angle. These effects are highly dependent on the impairment of the individual, with factors such as limb length and trunk function also affecting access to the rim. Further, for those with limited or no triceps function, a more posterior position will promote the use of biceps and hence a greater proportion of pull throughout the propulsion stroke.

The seat depth will also influence the stability and manoeuvrability of the wheelchair, with an anterior seat position causing increased stability but decreased manoeuvrability and vice versa for a more posterior position [83]. Due to their reduced trunk function, low point athletes in WCR tend to have a more posterior seat position than high point players [12], which is likely a reflection of the increased access to the wheel it provides [83]. High point players desire a seat position that is closer to the centre of gravity of the wheelchair-athlete system, as this reduces the rolling resistance [12, 32] allowing for improved acceleration and velocity. In addition to these sub-maximal findings, Usma-Alvarez et al. [64] identified a position behind the wheel axle as beneficial to maximal effort performance (power output and acceleration), but with similar limitations of the test methodology discussed previously. Currently, evidence suggests that the seat depth is important and should be carefully considered [11], however there are no clear findings on optimal position for wheelchair sports or individual athletes [64]. Research should consider appropriate field tests that capture information on the range of performance factors for WCR, as well as monitoring propulsion kinematics in linear sprinting.

Seat Angle

The seat angle of the wheelchair is seen as important by players, coaches and manufacturers [12, 14], with the clearest effects related to stability. A greater seat angle places the player in a more reclined position, increasing their stability and ball handling [11, 25] but players have reported perceived effects of hindering their trunk mobility and hence contribution to propulsion [12]. A reduction in trunk contribution is supported by Vanlandewijck et al. [26], who found increasing seat angle results in decreased acceleration over the first three seconds of propulsion from a standstill position. However, as previously mentioned, their work focused on establishing evidence-based classification standards and able-bodied participants were used, with testing conducted on a wheelchair ergometer. In WCR, low-point players often rely on an increased seat angle compared to high point players [12] due to their increased reliance on trunk support for stability. This obviously depends on the type and severity of impairment, as players with limb deficiencies can receive relatively low classification scores despite good trunk function. Seat angle (and seat height) become particularly important in these cases, as players will rely substantially on trunk motion to achieve adequate access to the wheel/push-rim. On-court testing with wheelchair athletes could complement findings from Vanlandewijck et al. [26] regarding the effects of seat angle.

Wheels – Camber Angle

The camber angle is greatly increased in court sports compared to wheelchair racing and daily wheelchairs [23]. Camber angles in WCR can be as large as 24° [11, 84], creating a much wider base of support than in daily wheelchairs which often have no camber [85]. The wider support base has the primary role of increasing stability, but also increases the ability of low point players to block opposition players and prevent them from advancing down the court [22]. In addition to these benefits, an increased camber angle improves turning capabilities of the wheelchair [11, 25, 75, 83] and hand protection from other chairs [11, 32, 75].

Despite some clear benefits of increasing camber angle, there are limitations to the amount of camber that is beneficial. Increased camber results in an increased contact area between the tyre and the court, which increases the rolling resistance and causes a reduction in linear speed [22, 75]. Mason et al. [68] found that for the subjects tested, an 18° camber resulted in improved performance in linear testing compared to 24°, and improved manoeuvrability compared to 15°. However, it is stated that this is unlikely to result in improved performance for all individuals and should be considered a guide for new players only [86]. Research can continue to investigate camber angle and potential optimal settings for a range of players and impairments to provide more detail on performance effects.

Wheel Diameter

Players believe that using a smaller wheel size may lead to improved initial acceleration while a larger wheel size improves the maximum velocity achievable [12]. Players also report selecting larger wheels to achieve a higher seat height whilst still maintaining access to the wheels, while some low point players noted they find it more difficult to accelerate the chair using large wheels due to the increased force required [12]. However, studies vary in their reporting of the effect of wheel size on initial acceleration. Usma-Alvarez et al. [64] found that the wheel size had a moderate to large effect on acceleration, while Mason et al. [68] found no significant difference in the initial acceleration during sprinting between wheel sizes during acceleration from standstill. The Usma-Alvarez et al. [64] study has limitations in using an ergometer for testing purposes, while Mason et al. [68] were unable to keep all other configuration parameters, such as distance between the top of the wheels, consistent. Further, the construction of test designs differed. First, Mason et al. [68] investigated WCB players, who typically have greater function that WCR players. In conjunction with the small change in inertia between wheel sizes being minimal, there is the potential for small effects to be missed. Alternatively, Usma-Alvarez et al. [64] tested WCR players, with wheel size potentially a more important consideration for these athletes due to their comparatively reduced function. Due to the lack of evidence surrounding the effect of wheel size on initial acceleration, a key performance factor for WCR, further research has been recommended [86].

Backrest Height

As with seat angle, the backrest height is often dependent on the trunk muscular function of the individual, with high point players generally having lower backrest heights. The lower backrest height allows greater mobility of the trunk but provides less stability [12, 83]. To the author's knowledge, no detailed research has been conducted into the effect of backrest heights despite large variations across players.

Tyres

The tyre pressure used is often based on individual preferences. Research has previously used 120psi [86] however players often use pressures in excess of this [87]. Increased pressures result in less frictional resistance due to the decreased contact area between the tyre and court, whilst a lower pressure results in improved grip. Tyre pressure above the recommended level can result in reduced grip and increased wheel spin [12], reducing the effectiveness of propulsion. This can occur for players with good trunk function and who are therefore able to have greater adjustments to their mass distribution. These adjustments to mass distribution often occur during turns and aid the player's ability to control the wheelchair whilst in control of the ball. However, if the mass is distributed incorrectly, it can result in large instances of wheel spin when accelerating or attempting to turn from stationary positions.

Tyre type and orientation will also influence performance, with pneumatic tyres at a high pressure and aligned in the direction of camber reducing deformation, and thought to provide the least rolling resistance [87]. A large range of tyres are used across players, with personal preferences depending on edge prevalence (often contact occurs between tyre and wheel guard), wear, and performance. Determining optimal tyre pressure therefore relies on the tyre in use as well as the individual player. Statistical testing is therefore difficult; however, tyre pressure should remain a parameter of interest for individual player approaches.

Mass

In addition to these factors, the mass of the wheelchair is a crucial factor in performance. Due to performance relying on high levels of acceleration and manoeuvrability [12], a lighter wheelchair has the potential to improve performance factors. Currently, rugby wheelchairs are in the range of 16-20kg [82]. The large mass in comparison to basketball wheelchairs (9-10kg, [24]) is due to the modifications made to the frames to withstand the high impact forces [11]. There is limited evidence to suggest that the current mass of rugby wheelchairs has been considered in terms of on-court performance [78]. However, there is a trade-off between achieving high acceleration and agility and allowing players to take opposition impacts and maintain their position, which is improved in a heavier wheelchair.

van der Slikke et al. [80] investigated adding ~7.5% of total mass to WCB chairs for both centrally located mass and distributed mass (~45% located at front, 45% at rear, 10% as part of a custom-made clamp to secure mass). As was expected, forward average acceleration was reduced in nearly all scenarios. However, for the distributed mass condition, the WMP test also showed negative effect sizes for the rotational acceleration [80]. Negative effect sizes indicate more effort would be required for rotation; this is beneficial in linear sprints where any slight rotations detract from linear speed, but this rarely occurs for extended periods in WCR. Wheelchair mass remains a parameter that can have a large impact on performance factors such as acceleration and agility; however, optimising this based purely on non-contact testing is difficult due to on-court requirements. Methods for quantitatively assessing changes in acceleration or ability to hold position are required, as well as discussions with individual players and coaches to achieve an appropriate wheelchair mass.

Gloves

Glove types in WCR are often customised by the player to meet their individual needs. This may relate to improving the interaction with the wheel, or to aid ball-handling. As transferring force to the wheel is crucial to WCR performance [65], the effect of glove type has previously been investigated [44, 80, 89]. Testing of gloves has included comparisons with American football gloves, building gloves, multi-purpose gloves, and prototype gloves. Lutgendorf et al. [89] found that American football gloves were preferable to building, multi-purpose, or no gloves for able-bodied participants in sprint and agility drills. However, Mason et al. [44] reported improved performance using the players' own choice of glove for both sprint and agility tests against American football, building, and a hybrid glove. As would be expected for individually customised gloves, players also reported higher ratings of comfort. van der Slikke et al. [80] investigated rubberised gloves that were intended to provide additional grip. Quantitative results showed small effects of glove type on performance variables in WMP and player feedback differed substantially. This likely reflects the individual requirements for gloves, with some individuals able to achieve better customisation than others. Customisation is also likely to consider propulsion approaches; low-point players may transition to a back-hand propulsion approach [44] after the initial strokes, whereas this is uncommon among high-point players. Glove type remains an area for further research, with WCR specific gloves that can be customised for specific individuals a potential area for improvement.

Abdominal Binding

Abdominal binding is an approach that can potentially increase the trunk range of motion and have cardiorespiratory benefits in WCR for players with cervical spinal cord injuries [69]. Use of abdominal binding in on-court testing showed positive results across a range of variables. For an acceleration-deceleration test, a significant 1.7% reduction in time taken was identified – this was equivalent to ~0.36m increase in distance across the same time. This improvement may be associated with improvement in the ability to transition between forward and back propulsion, as no differences were found between 5m sprint times with binding. No differences in agility performance were identified with binding, but distance covered in a 4-minute maximal push test was significantly increased. The mean population increase was potentially as high as ~49m, with the population including classification players from 0.5-2.5 points. In addition, West et al. [69] considered temporal and angular parameters of the propulsion approach for abdominal binding and non-binding conditions. No significant findings were identified throughout the analysis for angular parameters such and contact and release angle, or for trunk motion throughout the stroke and recovery. Temporal parameters showed a slight, non-significant (p=0.055) reduction in recovery time for the bound condition during the first three pushes of an acceleration.

Summary

A wide range of variables influence performance in WCR, with seat position and wheel factors largely affecting both performance and propulsion [88]. Other factors such as backrest height, tyre pressure and type, wheelchair mass and glove type are also likely to have an influence on performance [12]. Recommendations on parameters often still rely on submaximal propulsion results or small sample sizes. Given the large variations in individual impairment types, severities, and muscular functions, optimisation of wheelchair and equipment settings is recommended to occur at an individual level. This remains a difficult problem to solve due to the large number of parameters that can be adjusted (e.g., seat height, depth, angle, camber, wheel size). Substantial time commitments from players are required for appropriate and relevant testing, with optimal positions difficult to identify because of the small adjustments (often ~0.05-0.1m [65, 90]) required amongst elite players. For parameters that are not actively being changed during testing, further effort should focus on maintaining their constant setting (e.g., same glove type, same binding approach). Detailed optimisation of wheelchair configuration can improve athlete on-court mobility and performance, increasing the standard of competition and potentially influencing a team's ability to win major tournaments.

2.4 Testing Approaches

With advancements in instrumentation approaches as discussed above, testing has begun to transition to the preferred setting of in-field measures. Prior to this, many studies incorporated laboratory testing protocols that allowed for increased monitoring. These additional measures include power output and allows for motion capture to assess kinematics in detail [22, 77, 91, 92]; however, it is now known that testing on ergometers and treadmills does not accurately replicate overground propulsion [38, 93]. Treadmill testing has been suggested to be superior to ergometer testing as: the wheelchair is not rigidly tethered to a treadmill as it is the belts for an ergometer; contributions of trunk oscillations are accounted for; and it permits small lateral steering movements [56, 94]. Laboratory testing has shown increased stroke angles and times compared with overground propulsion [38], as well as limiting testing to be for linear propulsion. Although Mason et al. [38] recommended using a treadmill at 0.7-1% gradient for laboratory testing, this considered steady-state propulsion rather than maximal acceleration from standstill. Further propulsion changes may be expected when testing acceleration from standstill using laboratory settings, as done by Vanlandewijck et al. [6]. Ideally, and provided adequate assessment of propulsion and performance can be achieved, testing should be performed in experimental settings that closely replicate on-court behaviour [32, 95].

Current testing protocols for wheelchair court sports are also transitioning towards more representative test designs, such as testing acceleration from standstill on-court rather than in laboratory context [44, 67-69, 80] under (or in addition to) steady-state propulsion. In addition, many of these testing protocols have also begun to consider agility and braking aspects – crucial for effective on-court performance. van der Slikke et al. [80] utilised a testing approach that consisted of 15 small tests that were developed as a method to assess the WMP of wheelchair basketball players [81]. These test items consist of linear sprints and stop-starts, pushing around a curve, turning on the spot and combined activities, some of which were performed with a ball. This combination of tests covers the key performance variables previously discussed including acceleration from standstill, agility, and ball-handling [12]. A validity and reliability assessment of this series of tests highlighted that overall performance time provides a reliable measure of mobility performance – however, turning on the spot test items showed low reliability [81]. Similar approaches have previously been used in testing protocols – acceleration, braking, and agility tests [44, 68] – although the exact execution of tests has differed. Many elite teams have their own testing protocols that they use over various time-scales to track player performance. This will often include a sprint and agility test – in WCR, these may be a full court sprint and a slalom-based agility test [19]. These tests are used with the aim of replicating key movement demands of WCR such as acceleration from standstill and agility under controlled conditions; this allows assessment of changes in performance of individuals and provide insights into key factors of acceleration and agility. Full court sprint times are recorded at 5m, 14m and 28m and allow initial acceleration and top speed to be investigated [19] and the agility test replicates the frequent changes of directions in WCR. There remains a need to focus on these on-court situations which will allow for improved translation of research findings to relevant information for coaches.

Manoeuvrability with a ball and passing has previously been considered as part of a WCR test battery [96]. This has been used to assess passing abilities of players across varying classifications for short and long passes [19, 96, 97]; however, passing whilst in motion has not been considered in previous WCR testing. While de Witte et al. [81] considers ball handling whilst in motion during some activities of the WMP test, this involves dribbling the ball rather than passing (and is for basketball, not WCR). As WCR differs in dribbling requirements, and in how passes are completed – anecdotally, WCR uses more short passes to players who then attempt to escape opposition – there remains a need to develop an appropriate ball handling and passing test for WCR.

These controlled testing protocols are important for reliability, and for the assessment of meaningful changes in physical capacity and performance [98].

However, over-constraining test designs can potentially result in kinematic and performance outcomes that are not reflective of performance contexts and thus provide reduced insight into the execution of on-court activities. Designing representative tests, however, is not a simple task [98, 99], with recent research in soccer questioning the validity of passing test designs in controlled environments [100]. Including more representative tasks in testing, along with the controlled testing, can allow researchers and practitioners to assess skill improvements as well as physical improvements of individual athletes. Although clearer definitions and examples of representative task designs are still required [98, 99, 101], simple additions to current tests (such as including a pass or catch of a ball) can create more representative tasks and aid in translating findings from testing to on-court performance.

2.5 Robust Design Approaches

One potential method to reduce the amount of time associated with wheelchair prescription is the use of robust design approaches. Robust design approaches provide an effective way to balance statistical testing and compressed time schedules [102], with these requirements often difficult to balance under practical conditions. Robust design or Taguchi optimisation has generally been used in upstream product engineering to focus on product quality, rather than a downstream problem-solving approach [103]. This method requires parameters to be adjusted independently and attempts to determine their associated effects on performance measures. Once a number of design parameters have been selected, and the number of different settings selected (referred to as testing 'levels'), an orthogonal array for testing can be developed. An orthogonal array investigates each parameter and level in isolation by testing this against all other parameters and levels. The effect of the individual parameter level can then be assessed by averaging the performance measure for all set-ups that involved that setting. For example,

Table 2-1 contains an L9 orthogonal array with four parameters, with each varied at three levels. As can be seen, Parameter A – Level 1 (A1) is in three configurations (Tests 1, 2, and 3). These three configurations also contain each level (1,2, and 3) for all other parameters (B, C, and D). As A1 is tested in configurations with all other parameter levels, the effect of A1 can be assessed in isolation. An orthogonal array implements this for all parameters and levels. This can substantially reduce the amount of testing required: for this example, the number of test trials can be reduced from 81 to 9.

Table 2-1: An example of an L9 orthogonal array, where three parameters at three levels are tested across 9 trials.

Parameter	**Test 1**	**Test 2**	**Test 3**	**Test 4**	**Test 5**	**Test 6**	**Test 7**	**Test 8**	**Test 9**
A	A1	A1	A1	A2	A2	A2	A3	A3	A3
B	B1	B2	B3	B1	B2	B3	B1	B2	B3
C	C1	C2	C3	C2	C3	C1	C3	C1	C2
D	D1	D2	D3	D3	D1	D2	D2	D3	D1

This approach has potential use in equipment design, where a number of independent configuration parameters can be altered with a focus of a limited number of performance outcomes. Usma-Alvarez et al. [64] implemented such an approach with WCR. This involved investigating seat height, seat depth, wheel diameter and camber angle. Each parameter was varied at three levels excluding wheel diameter, which was varied at two – therefore an L9 orthogonal array was used. Testing was performed with five players of varying point-scores, with individual case-studies showing potentially positive results. However, outcomes (in terms of recommended wheelchair set-ups) were not confirmed using follow-up testing, and translation to on-court performance was very limited: only linear sprinting was considered, with this performed on an ergometer with little consideration of propulsion kinematics. For such an approach to be practical, testing needs to consider the various on-court performance factors and any recommended set-ups tested before confirming a recommended set-up. This relies upon instrumentation approaches that allow for detailed on-court monitoring, and assessment methods that are able to reflect important changes in athlete propulsion and performance.

2.6 Modelling

A complimentary approach to reduce the reliance on substantial player testing while still providing detailed assessments of wheelchair configurations is through propulsion modelling. A range of models attempting to replicate wheelchair propulsion have been developed, with these again focused on sub-maximal propulsion. These models vary between four-bar linkage systems [104, 105] and musculoskeletal models [106, 107] and aim to predict the change in joint torques or muscular effort caused by changing seat positions.

The four-bar linkage model initially developed by Veeger et al. [108] was constructed to investigate the load on the upper extremity load by utilising inverse dynamics calculations. Richter [104] extended this to investigate the effect of seat position on sub-maximal propulsion. This system consisted of an upper arm, forearm-hand segment (based on a 50^{th} percentile male), handrim, and wheel, as shown in Figure 2.4, with mass considered negligible. The model was simplified to be quasi-static and two-dimensional such that it replicated propulsion on a dynamometer where the wheels rotate but the wheelchair does not move. The shoulder joint and wheel hub were fixed (Figure 2.4) and the anatomical measurements based on a 50^{th} percentile male [104].

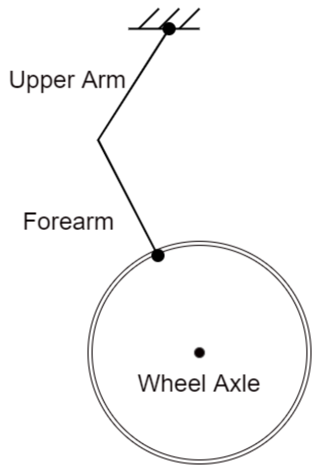

Figure 2.4: The four-bar linkage system developed by Richter [104] and adjusted by Leary et al. [105] consisted of an upper arm, forearm-hand segment, handrim and wheel.

Testing with a SMARTWheel system involved five participants to gain push force data, with an average handrim force profile across participants used as the input throughout the push simulation. From the model, it was possible to determine changes in propulsion kinematics such as contact angle and release angle after adjusting the relative positions of the shoulder joint and wheel axle [104]. The contact and release angle estimations are based on the angle between shoulder position and wheel hub (θ_{HS}), the distance between shoulder and wheel hub (L_{HS}), length of upper arm (L_{UA}) and forearm-hand (L_{FA}) segments, and the radius of the handrim/wheel (R_{HR}). Using the cosine rule, the hand contact (θ_C) and release (θ_R) can be calculated [104]:

$$\theta_C = \theta_{HS} - cos^{-1}\left[\frac{L_{HS}^2+(R_{HR}+L_{FA})^2-L_{UA}^2}{2R_{HR}L_{HS}}\right] \qquad \text{(Eq. 1)}$$

$$\theta_R = \theta_{HS} + cos^{-1}\left[\frac{R_{HR}^2+L_{HS}^2-(L_{UA}+L_{FA})^2}{2R_{HR}L_{HS}}\right] \qquad \text{(Eq. 2)}$$

The model by Richter [104] was analysed by Leary et al. [105] and found to produce incorrect measures of the shoulder torque. Leary et al. [105] then altered the model such that the shoulder torque sums to zero about the fixed shoulder joint, allowing more accurate assessment. These models can be used to investigate the effects of seat position on joint torques and identify optimal positions during sub-maximal propulsion in an attempt to avoid injuring particular muscle sets [105].

However, this model was intended for use with submaximal propulsion. As previously discussed, the propulsion methods between submaximal and wheelchair sport differ substantially. In this case, a major limitation of this model is the assumption of a fixed shoulder position. Multiple studies have shown there is substantial trunk activity and motion throughout the initial strokes when maximally accelerating [26, 59, 60]. For a propulsion model to accurately replicate maximal acceleration, trunk motion must be considered. Further, the current linkage models assume that each stroke is consistent. As Moss et al. [58] found for wheelchair racing, the initial strokes are likely to vary and a useful model should account for this as well as other factors such as impairment and individual anthropometrics.

Slowik et al. [107] also used equations based on Richter [104] to monitor contact and release angles in their musculoskeletal model, although these were multiplied by a factor of 0.9 because individuals were not expected to use their full range of motion during propulsion. From these values, further variables such as the push angle (θ_{PA}) and push frequency (f_P) were calculated in both models [104, 107]. Research by Richter [104] then focused on determining joint torques and forces at the shoulder and elbow, while Slowik et al. [107] focused on upper extremity energy demand.

Musculoskeletal models used by Rankin et al. [106] and Slowik et al. [107] have developed a more detailed simulation of the upper extremity to focus on energy demand. These models, developed using Software for Interactive Musculoskeletal Modelling (SIMM, Musculographics Inc., USA), used rigid segments to represent the trunk, upper arm, forearm and hand. Segment interactions were defined by six rotational degrees of freedom, which were trunk lean, shoulder plane of elevation, shoulder elevation angle, shoulder internal-external rotation, elbow flexion-extension and forearm pronation-supination. The model was driven by 26 hill-type musculotendon actuators that represented the major upper extremity muscles. The activation times of these actuators were determined through experimental EMG data [106]. The model's equations of motion were generated using SD/FAST (Parametric Technology Corp., Needham, MA, USA) [107], with the simulation model used in a variety of ways. Work by Rankin et al. [106] investigated the efficiency of propulsion by investigating a single participant’s common propulsion approach, as well as conditions thought to maximise sub-maximal propulsion efficiency such as altering cadence, peak forces at the push-rim and contact angles.

For musculoskeletal models investigating energy demand, prediction of muscle function is an important consideration. Muscle prediction generally occurs through two methods: static optimisation approaches which can be performed using MATLAB [109, 110]; or dynamic optimisation approaches performed in SIMM [106, 107]. The optimisation approaches are required to estimate indeterminate muscle forces, with static optimisation having a much lower computational cost but, unlike the dynamic optimisation, does not account for the time-dependant physiological nature of muscles [109]. Morrow et al. [109] investigated the effects of using a static optimisation approach compared with a dynamic approach with varying results. While the static optimisation was able to predict muscle forces that produced the appropriate motion at the shoulder, there were differences in the forearm motion and push-rim forces. The dynamic optimisation approaches require detailed musculoskeletal systems to be built in interfaces such as SIMM or OpenSim, adding substantial complexity and computational cost to the models. For optimisation across WCR squads, this becomes difficult as substantially different models are required – in terms of limb lengths, muscular function, and propulsion approach.

Sport specific models include work by Masson et al. [111], who investigated the effect of upper limb momentum to pushing power in wheelchair racing. While testing was performed at a relatively high velocity (10 m/s), it was still suggested that muscle contribution to propulsion was substantial. Muscle contribution is therefore expected to be the predominant propulsion mechanism at all times in WCR, where accelerations and changes of direction are frequent. This suggests that, despite the computational cost, inclusion of a player's upper limb and trunk muscular function may be important.

The propulsion models currently developed provide a base to allow future development of models. Leary et al. [105] and associated linkage models are able to investigate propulsion kinematics based on the shoulder position relative to the axle based on relatively simple calculation approaches. Musculoskeletal models [106] can be extended further and are able to account for muscle function to investigate energy demand but may not be suitable for individual optimisation approaches due to their added complexity. While current calculation-based models [104, 105, 110] are able to investigate two-dimensional motion of a single arm and trunk, they are unable to simulate the asymmetrical propulsion present in some WCR players or investigating turning capabilities [107]. These models have also only been applied to sub-maximal propulsion, with WCR incorporating maximal effort propulsion and frequent changes of directions. Improvement of propulsion models would allow reduced reliance on human testing to determine an optimal wheelchair set-up based on an individual's anthropometrics and level of muscular function. Musculoskeletal models provide an opportunity to account for this function, although at an additional computational and time cost compared with linkage models. Despite the limitations of a linkage model (namely not accounting for muscular function), it potentially provides an efficient manner in predicting an individual's propulsion approach. Achieving a subsequent performance measure – whether this is a time estimate or an energy cost prediction – would be critical in then utilising this method to assess various wheelchair set-ups.

2.7 Summary

Wheelchair court sports, along with wheelchair racing, have received substantial research interest for Paralympic sports. This research has included the attempted monitoring of on-court motion [13, 16, 20], physiological adaptations, equipment use [69, 89], and wheelchair configuration [64, 68, 80]. Only recently have instrumentation and processing methods [42, 45] allowed improved representative testing methods to be employed during testing protocols. This has aided in assessments of important performance factors; however, a number of gaps for further research remain. These include:

- No quantitative assessments of current wheelchair configurations or propulsion approaches across an entire squad
- Lack of knowledge surrounding detailed analysis of propulsion approaches, particularly for acceleration from standstill
- Assessments of how changing configuration parameters alters propulsion and performance in representative testing
- Current propulsion prediction models do not consider maximal effort propulsion, or account for trunk motion
- No prediction method for monitoring performance outcomes such as sprint time for varying configurations or propulsion approaches

This thesis aims to increase the knowledge of the effect configuration parameters have on performance and propulsion in WCR by addressing the above limitations. It is intended that this will result in an approved ability to optimise individual wheelchair configurations with reduced testing time and effort.

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Chapter 3: Elite wheelchair rugby: a quantitative analysis of chair configuration in Australia.

This chapter details the wheelchair configurations across an entire elite squad, analysed in terms of classification groups. No previous work has reported this data which provides greater insight into current configurations and views of elite players.

This chapter has previously been published (see below details) and has been reformatted for the purpose of this thesis. This publication does not satisfy The University of Adelaide requirements for inclusion in a Thesis by Publication.

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i. the candidate's stated contribution to the publication is accurate (as detailed above);

ii. permission is granted for the candidate in include the publication in the thesis; and

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3.2 Abstract

Limited recommendations of wheelchair configurations for court sports have been identified in published literature. To accommodate the wide range of impairments in wheelchair rugby, players are given a point score which reflects their impairment. Players have regularly been grouped as high-, mid- or low-point players in research, with high-point players having greater levels of muscle function compared with other classifications. This research documented the wheelchair configurations of elite Australian wheelchair rugby players across classification groups. Significant differences (p<0.05) were found for increased seat height and decreased seat depth for high-point players compared with low- and mid-point groups, respectively. Low-point players displayed reduced wheelchair mass compared with high- and mid-point players, as well as increased frame length. Camber angles showed no significant differences across the classification groups. The incorporation of anthropometric measures, such as the elbow angle at the top dead centre, were also investigated. While elbow angle showed no significant differences, seat height-to-total arm length ratio was higher for high-point players. Participants also completed surveys detailing their perception of the effect of altering wheelchair configurations. It is suggested that wheelchair configurations should consider an individual's anthropometrics, impairment, training history and court role to promote optimal performance, with predictive modelling having the potential to reduce the associated time and cost.

Keywords

Wheelchair rugby; Wheelchair sports; Configurations

3.3 Introduction

The configuration of different wheelchair parameters can have a significant effect on propulsion and sport performance [1, 2]. The effect of different wheelchair configurations on propulsion in activities of daily living have been previously investigated [3-8], with the aim of minimising the risk of injury during sub-maximal propulsion [9]. Conversely, wheelchair configurations used in sports aim to maximise 'performance' [1], which have been the focus of many recent publications [10-15]. A number of these studies have focused on wheelchair rugby (WCR), a fast-paced, high contact sport [16] for athletes with physical impairments of at least three limbs [17], such as spinal cord injury and multiple amputations [18]. Athletes are classified based on their trunk and arm function (indicative of their impairment), with scores ranging from 0.5 to 3.5 points. Each team is limited to a maximum of 8 points on court at any one time [19]. 'High-point' players (i.e., 3.0–3.5 points) are typically offensive players who receive the ball and attempt to score, due to their greater muscle and trunk function compared with other classifications. 'Low-point' players (i.e., 0.5–1.0 points) are defensive players, who try to 'hook', or block opposing team members. 'Mid-point' players (i.e., 1.5–2.5 points) usually perform aspects of both defensive and offensive roles [20].

A variety of wheelchair configurations for court sports have been proposed in the literature [2, 20, 21]. These recommendations have been typically developed through qualitative studies involving players and coaches [22, 23], and are limited to only a small number of wheelchair parameters (see Section 2) with small sample sizes [8, 11, 12, 15, 24, 25]. To the best of the authors' knowledge, a quantitative investigation into the preferred wheelchair configurations for an elite WCR squad has not been performed. This is seemingly an important consideration given the wide variance in on-court roles both within and between classifications. Therefore, the aim of this study is to perform a quantitative investigation of the various WCR wheelchair configurations of an entire elite squad.

3.4 Background

3.4.1 Wheelchair Parameters

The configurations of different wheelchair parameters have been suggested to affect performance in various wheelchair sports, including seat height, seat depth, seat angle, wheel diameter, camber angle, frame length and backrest height [14, 20, 21]. Wheelchair mass has received limited attention [18], and little is known regarding the relationship between wheelchair configurations and individual anthropometrics [1]. Previous studies have investigated key performance indicators (e.g., acceleration and agility) for each classification, as identified by players [22, 24]. It is currently unknown, however, how the athlete perceptions align with actual chair measurements.

Wheelchair measurements encompass measurements of the seat position, such as height, depth and angle, as well as measures such as camber angle and wheel diameter. Seat height has been one of the most investigated parameters in wheelchair sports. Usma-Alvarez et al. [14] suggested that seat height is the most influential parameter for short sprints in WCR. Reduced seat height can allow a given athlete to remain in contact with the pushrim/wheel for longer [26], thus increase the time for force application. The trade-off for wheelchair court sports is that a higher seat position allows for improved ball handling ability [18]. Ball handling constitutes passing, catching, and intercepting the ball [1]. High-point players reportedly prefer higher seat heights compared with low-point players due to their court role and improved trunk function [22-24], although measures of seat height across classifications have not previously been performed. Seat depth can affect the stability and manoeuvrability of the wheelchair [27]. Posterior seating positions, wherein the seat position is further behind the wheel axle, may improve stability at the expense of manoeuvrability. Low-point players, who have reduced trunk function, have been reported to prefer a more posterior seat position compared with mid- and high- point players [22], however this has not been measured across an entire squad. It is hypothesized that high-point players will have the smallest seat depth due to their decreased reliance on stability and attempts to maximise linear acceleration [22]. To further aid their stability, low-point players are expected to use larger seat angles [20, 22, 28] and backrests [22, 27] compared with high-point players. However, there are trade-offs, as increased seat angles are associated with decreased maximal acceleration during propulsion from standstill [15] and increased backrest heights associated with decreased trunk mobility [22, 27].

Camber angle is the relative angle between the wheels and the vertical axis [29]. While daily living wheelchairs often have little-to-no camber [30], rugby wheelchairs can reach values as large as 24° [11, 20, 25]. Increased camber angles provide a greater base of support and improve manoeuvrability [11, 20, 27-29, 31]. However, increased camber angle can result in increased contact area between the wheel and the surface, thereby increasing rolling resistance. Mason et al. [11] recommended a camber angle of 18° for new WCR players; however, there is little evidence as to how/if these recommendations are applicable across classifications. There is also little information regarding the effect of classification on wheelchair frame length. Low-point players usually have longer frame lengths due to the defensive chairs utilising a 'hook' to hold opposing players (Figure 3.1). Offensive players are expected to have a shorter wheelchair frame with a guard to prevent hooking [28], although the length of mid-point wheelchairs may vary. WCR players have suggested that using a smaller wheel diameter improves linear acceleration, while larger wheels increase maximum linear velocity [22]. Wheel diameters from 24-26 inches have been investigated; Usma-Alvarez et al. [14] reported wheel diameter has a moderate to large effect on linear acceleration and Mason et al. [25] suggested smaller diameter wheels increase the physiological demand. However, low-point players have reported a preference for smaller diameter wheels to aid in their acceleration from standstill [22], while high-point players may use larger wheels in conjunction with higher seat height, so as to maintain a similar ability to contact the pushrim/wheel.

Figure 3.1: The rugby wheelchairs used by low-point (left) and high-point (right) players.

The effect of varied wheelchair mass has received limited attention. WCR wheelchairs often have masses ranging between 16-20kg [24]. In comparison, wheelchair basketball chairs typically range between 9-10kg [32]. The added mass in WCR wheelchair's comes from the outer reinforced frame structure that protects against the high impact forces [20]. It is hypothesized that high-point players will have lighter wheelchairs compared with low-point players to allow for greater manoeuvrability and acceleration. The mass of the wheelchair is limited by the wheelchair manufacturer.

3.4.2 Anthropometrics and Configuration

Previous research has related wheelchair configurations to an individual's anthropometrics [1]. For example, elbow angle at 'top dead centre' (TDC) of the wheel has been suggested as a method for assessing seat height [8]. van der Woude et al. [8] showed that elbow angles between 100° and 130° at TDC were associated with increased cardio-respiratory and mechanical efficiencies during sub-maximal propulsion tests. While sub-maximal propulsion is frequent in WCR, high-intensity efforts are common and crucial to escaping or performing blocks [33] and therefore critical for enhanced on-court performance. Investigations of seat height using a standardisation method such as the elbow angle at the TDC in configurations used for maximal effort propulsion are limited [1]. It is hypothesized that high-point players will have a greater elbow angle at TDC due to the expected increased seat height. Determining seat height using the percentage of upper-and-lower arm segment lengths has also been proposed in the literature [5], although not as widely as elbow angle at the TDC. Further standardisation methods, such as considering trunk lengths and thigh lengths, will be considered in this paper.

3.5 Method

3.5.1 Participants

Sixteen athletes (age 30±6.3 years, international experience 5.5±5.1 years) provided informed consent before participating in the study. The sample included all the members of the Australian WCR squad, who are the current World and Paralympic champions. Athletes were grouped based on their point classification; a high-point group with athletes from 3.0 and 3.5 classes (n=6), a mid-point group of 1.5 to 2.5 (n=5), and a low-point group of 0.5 and 1.0 classes (n=5).

3.5.2 Measurements and Athlete Preferences

The wheelchair parameters measured included: seat height, seat depth, seat length, seat angle above the horizontal, wheel diameter, camber angle, frame length, backrest height, and wheelchair mass. Linear measurements were taken using a standard measuring tape (Festool Metric/Imperial Tape Measure 3m). Seat height was measured as the distance from the floor to the lowest point at the rear of the seat. Seat depth was measured from the wheel axle to the bottom rear of the seat. Frame length was measured from the most forward point of the frame to the centre of the wheels. Seat angle was determined through measurements of rear and front seat heights along with the seat length of the wheelchair. Seat angle was confirmed through the use of calipers to measure the seat angle relative to the horizontal axis. Camber angle was calculated by the top and bottom wheel separations, along with the wheel diameter. These measures were compared with video analysis using a rear view of the wheelchair. Measurements were performed in accordance with Melrose Kiwi Concept Chairs scripts [34].

Anthropometric measures included the lengths of the right upper arm (i.e., acromion to olecranon), lower-arm (i.e., olecranon to radial-ulnar processes), torso (i.e., anterior superior iliac spine (ASIS) to acromion), and thigh (i.e., ASIS to lateral epicondyle of the femur). Several wheelchair-to-anthropometric ratios were also calculated, including: backrest height-to-torso length, seat depth-to-thigh length, and seat height-to-total arm length. Elbow angle when the hand was positioned at the TDC of the wheel was calculated using the height of TDC on the wheel, upper arm length, lower arm length, shoulder height and shoulder depth. Shoulder depth was calculated using the measured seat depth, back angle, and torso length. From the shoulder position, the distance between this location and TDC could be determined, with the elbow angle calculated using trigonometry based on the upper and lower arm lengths. This approach enabled quick distance measurements to be used as opposed to a mechanical/electrical goniometer, which was deemed to be more time consuming. Where possible, these angles were confirmed using video analysis.

Participants also completed a survey detailing their perception of how specific wheelchair parameters affect sport performance metrics. Participants reported if they perceived sport performance would be improved or limited by changing their current configuration. The inclusion of both quantitative wheelchair measures and qualitative results from a survey has not previously been reported in one study.

3.5.3 Analysis

Statistical analysis was performed using IBM SPSS Statistics 21. An ANOVA investigation followed by Tukey post-hoc test was used to assess differences across high-, mid- and low-point classification groups. Significant differences ($p \leq 0.05$) across classifications were identified at a 95% confidence level. Wheel diameter was analysed based on the frequency of selected sizes within classifications, and hence investigated using Fisher’s test between the rank and wheel diameter.

3.6 Results

3.6.1 Wheelchair Configurations

Table 3-1 presents the means ± standard deviations of the measured wheelchair configurations for each classification group. Low-point players had a lower chair mass than both high- and mid-point groups. High-point players had significantly higher seat heights compared with low-point players, and reduced seat depth relative to mid-point. Seat height and depth for mid-point players showed no significant differences compared with low-point players. The coefficient of variance for seat height within classifications ranged from 7% for high-point players to 16% for low-point players. Seat depth had similar variances, from 5% for high-point players to 10% for low-point players. High-point players had smaller seat angles and backrest heights compared with mid- and low-point groups, with these parameters not significantly different between mid- and low-point players. Seat angles for high-point players showed a coefficient of variance of 22%, while low-point players had 6%. Backrest heights coefficient of variance was 10% for mid- and low-point players, but 16% for high-point players. The length of the wheelchair frames was longer for low-point players compared with both high- and mid-point groups. Coefficients of variance for frame length ranged from 3% for high- and mid-point players to 6% for low-point players. There was no significant difference found in camber angle across classifications. Wheel diameters selected across classifications varied, with high- and mid-point players often selecting 0.635m diameter wheels (n=5 for both high- and mid-point groups) while most low-point players (n=4) used 0.61m diameter wheels. One high-point player also used 0.66m diameter wheels and a single low-point player used 0.635m wheels.

Table 3-1: Mean (±SD) measurement values for wheelchair configurations of high-, mid- and low-point classification groups, with statistical significance amongst classification groups.

Group	Chair Mass (kg)	Seat Height (m)	Seat Depth (m)	Seat Angle (°)	Camber Angle (°)	Frame Length (m)	Backrest Height (m)
High-point	19.6 (0.7)	0.29 (0.02)	0.14 (0.01)	25.0 (5.6)	16.2 (1.2)	0.53 (0.02)	0.29 (0.05)
Mid-point	19.2 (0.5)	0.25 (0.02)	0.16 (0.02)	37.3 (3.8)	17.0 (1.2)	0.54 (0.02)	0.41 (0.04)
Low-point	17.4 (0.9)	0.24 (0.04)	0.15 (0.01)	31.9 (1.9)	16.8 (0.4)	0.72 (0.04)	0.42 (0.05)
			Tukey Post-Hoc Test Results (P-values)				
High-Mid	0.640	0.192	**0.046***	**0.001***	0.397	0.858	**0.002***
High-Low	**0.001***	**0.037***	0.138	**0.045***	0.577	**0.001***	**0.001***
Mid-Low	**0.003***	0.641	0.824	0.147	0.949	**0.001***	0.987

*Significant differences using p<0.05

3.6.2 Wheelchair-Anthropometric Ratios

The calculated wheelchair-to-anthropometric ratios are presented in Table 3-2. There were no significant differences between classification groups in regards to anthropometric measures (i.e., mass, shoulder height, total arm length), with coefficients of variance no larger than 13% for mass and 5% for shoulder height and total arm length. There were also no significant differences across classifications for elbow angle at the TDC and seat depth-to-thigh length. Coefficients of variance for elbow angle at the TDC varied from 4% for high-point players to 13% for mid-point players. Larger coefficients of variance were present for the seat depth-to-thigh length ratio, with a minimum of 10% for mid-point players and a maximum of 24% for high-point players. The ratio of the backrest height-to-torso length was significantly smaller for the high-point group compared with the mid- and low-point groups, as well as between mid- and low-point groups. Coefficients of variance ranged from 9% for mid- and low-point players to 15% for high-point players. The ratio of seat height-to-total arm length was significantly greater for the high-point players compared with low-point players, with coefficients of variance from 8% for high-point players to 13% for low-point players.

3.6.3 Survey Results

Survey responses varied between classifications. Mid-point players suggested increasing seat height would limit linear acceleration (80%) and 'agility' (80%), low-point players felt it would improve their acceleration (60%), and high-point players suggested it would improve agility (67%). 67% of high-point players also perceived that increasing seat height would improve their stability (i.e., the ability to maintain balanced), while mid- and low-point players perceived it would limit stability (100% and 80%, respectively).

Increasing seat depth also produced varying results across classifications, with high-point players perceiving reduced acceleration (67%) and agility (67%), compared with mid-point players perceiving improved acceleration (80%) and agility (80%). Low-point players suggested limited acceleration (60%) but improved agility (80%) for increased seat depth.

All classifications perceived improved stability for increasing seat angle, with high-point players also perceiving that it would limit acceleration (67%). All classifications also reported increased camber angles would improve agility and stability. However, low-point players perceived that it would limit acceleration (100%) and maximum linear velocity (80%). Increasing backrest height was perceived to improve stability across all classifications, though high- and mid-point players also suggested limited agility (83% and 60%, respectively). All classifications suggested that increasing wheel diameter would limit linear acceleration but improve the maximum linear velocity. High-point players also perceived a reduction in agility (67%).

3.7 Discussion

The aim of this research was to perform a quantitative investigation of the various WCR wheelchair configurations of an entire elite squad.

3.7.1 Wheelchair Measures

High-point players having increased seat heights compared with low-point players is supported by the measurements of the Australian WCR squad. Increased seat height for high-point players is thought to provide improved ball handling (e.g., throwing, catching, or dribbling) without large detriments to propulsion [18, 22]. For a given seat height, high-point players can remain in contact with the pushrim/wheels for a greater period of time relative to low-point players due to their increased trunk function. Despite the increased seat height, high-point players perceived the level of seat height as advantageous to improved stability (67%) and agility (67%). This differed from the mid-point players, who suggested that stability and agility would be negatively affected by higher seat heights.

Table 3-2: Mean (±SD) of anthropometrics and ratios of anthropometric and wheelchair parameters for classification groups, with statistical significance amongst classification groups.

Group	Individual Mass (kg)	Shoulder Height (m)	Total Arm Length (m)	Elbow Angle at TDC (°)	Seat Depth – Thigh Length	Backrest Height – Torso Length	Seat Height – Total Arm Length
High- point	71.7 (5.2)	0.87 (0.03)	0.54 (0.03)	70.3 (3.0)	0.37 (.09)	0.73 (.11)	0.53 (.04)
Mid- point	76.4 (10.0)	0.86 (0.02)	0.54 (0.02)	79.4 (10.3)	0.37 (.04)	0.88 (.08)	0.47 (.04)
Low- point	73.3 (2.5)	0.83 (0.03)	0.53 (0.02)	73.3 (9.1)	0.33 (.06)	1.04 (.09)	0.45 (.06)
	Tukey Post-Hoc Test Results (P-values)						
High-Mid	0.520	0.768	0.893	0.178	0.999	**0.038***	0.096
High-Low	0.918	0.098	0.811	0.814	0.742	**0.001***	**0.031***
Mid-Low	0.753	0.326	0.572	0.458	0.670	**0.039***	0.824

*Significant differences using $p<0.05$

It was hypothesized that high-point players would have seat depths closer to the wheel axle compared with other classifications, since this configuration is associated with decreased rolling resistance [1, 22, 27]. The results of this research confirmed this hypothesis with respect to mid-point players. However, there was no significant difference in seat depths between high-point players and low-point players, or between mid- and low-point players. While this may be a result of the limited sample size (n=16), it may also be an indication that mid-point players require a more posterior seating position for stability while maintaining access to the pushrim/wheel, which low-point players seemingly cannot. A posterior seat position also increases the 'push angle' (i.e., the angle the hand travels while in contact with the wheel [35]) and promotes a 'pull' propulsion technique [9, 22] (i.e., utilises a greater amount of biceps brachii function). The survey results showed that mid- and low-point players felt an increased seat depth may result in improved agility (80%). Compared with other classifications, the high-point players had significantly smaller seat angles and backrest heights, which concurs with previous literature [15, 22]. Lower backrest heights for high-point players allows for greater trunk mobility; higher backrests for mid- and low-point players provides greater postural stability. These results were supported by the survey responses, with 93% of players across all classifications suggesting that increasing backrest height would improve their stability, but limit their agility (56%). Interestingly, there was no significant difference between mid- and low-point players. It was expected that mid-point players would have lower backrest heights. There were also no significant differences in seat angles between mid- and low-point players. Mid-point players may conceivably achieve greater pushrim/wheel contacts with similar seat angles compared with low-point players due to their greater trunk function. The frame lengths for high- and mid-point players were similar (0.53±0.02m and 0.54±0.02m, respectively), but both were significantly smaller than those used by low-point players (0.72±0.04mm). This is likely due to differences in on-court roles, considering low-point players are primarily responsible for blocking opposing players [20]. The shorter frame lengths used by high- and mid-point players would allow them to manoeuvre effectively on offence while avoiding defence players [20].

While many of the aforementioned findings concur with previous literature [2, 15, 22], the wheelchair mass and camber angle results did not align with the original hypotheses. In an attempt to maximise linear acceleration, it was expected that high-point players would have smaller wheelchair masses than low-point players. Moreover, the longer frame length used by low-point players was expected to result in increased wheelchair mass. The higher wheelchair mass used by high-point players may be explained by a need for increased frame strength – high-point players can reach higher linear velocities compared with low-point players [19], thus the impact forces between opposing high-point players is expected to be higher, though this has never been formally documented. There was no significant difference in camber angle between the different classifications. High-point players were expected to have reduced camber to maximise linear velocities and low-point players to have increased camber angles for greater stability. 75% of the WCR players claimed that increased camber angles improved agility. However, 56% believed large camber angles would limit their linear acceleration. As expected, players from all classification groups suggested that increased camber angle would provide stability benefits (87.5%). The camber angles measured in this work were slightly smaller than those previously suggested for novice players [11]. Further research is needed to assess the 'optimal' camber angles for novice and elite WCR players of different classifications. This could potentially be achieved through the use of predictive modelling, where subject-specific optimal camber angles can be determined using forward dynamics.

Wheel diameter was positively associated with classification, with low-point players choosing the smallest diameters (i.e., 0.61m) and high-point players with the largest diameters (i.e., 0.635m or 0.66m). The larger wheels selected by high-point players likely allows the higher seat position whilst maintaining contact with the pushrim/wheel. Low-point players have previously reported difficulty in initiating motion using larger wheel diameters [22]. Due to the increased muscular function of high-point players, they are likely to be able to initiate movement with larger wheel diameters than low-point players.

3.7.2 Wheelchair-Anthropometric Measures

There was no significant difference in total body mass between the classifications. There was no consistent trend between total arm lengths and classification levels. There was a slight relationship between classification level and resting shoulder height, with high-point players having the highest shoulder height and low-point players the lowest. This finding is interesting considering there were significant differences in seat heights between classifications. There were no significant differences in the relative elbow angles at the TDC of the wheel between classifications. Although studies have investigated the elbow angle at the TDC of the wheel for wheelchair court sports [36], angles during activities of daily living are more commonly cited. The single study also used basketball wheelchairs with a handrim, thus to the authors knowledge, the elbow angle at TDC of the wheel has not been considered without handrims. van der Woude et al. [8] found that angles of 100–130° were associated with improved cardio-respiratory and 'mechanical efficiencies'. All relative elbow angles measured in this work were substantially lower than the aforementioned results, with group means ranging from 70–79°. While it was expected that high-point players would have greater elbow angles at the TDC due to higher seat heights compared with low-point players, the smaller seat depth likely reduced this effect. High-point players had a higher ratio for seat height-to-total arm length compared with low-point players, which reflects the higher seat heights for high-point players.

Using wheelchair-anthropometric ratios can provide a method of standardisation across individuals. There was no significant difference in seat depth-to-thigh length ratios between the different classifications. This was unexpected, since the seat depth was significantly smaller for high-point players compared with other classifications. These results may be influenced by the reduced number of measurements taken for high-point athletes, of which some measurements were not possible due to their impairments (i.e., amputation at thigh level). Compared with other classifications, high-point players had a significantly lower ratio for backrest height-to-torso length. Mid-point players had a lower backrest height-to-torso length ratio compared with low-point players. This provides greater detail on the backrest heights across classifications than considering only measurements of the backrest height.

Wheelchair features such as seat depth were based on identifying the centre of the wheel axle. Inaccuracies in the estimation of the centre of the axle have the ability to influence the measurements and subsequent angle calculations. The angles were calculated using distance measures – i.e., camber angle was calculated using top and bottom wheel separations, along with the wheel diameter. Inaccuracies in the linear measurements will affect the calculated angles such as the seat, camber, back and elbow angles. Due to time restrictions, it was desired to perform quick distance measurements as opposed to using a mechanical/electrical goniometer. Where possible, the calculated angles were compared with those derived via video analysis.

Identifying the hip joint centre was particularly difficult; the wheelchair frame and clothing surrounding the hip caused difficulties in accurately accessing the joint centre.

The current study used 16 elite players across three classifications, which reduced the likelihood of achieving statistically significant differences across classifications for the various parameters. Larger samples sizes have the potential to overlook useful individual characteristics due to the varying degrees of impairment within a specific classification, causing the results to be counterproductive to the optimisation of individual performance [37]. However, performing detailed investigations into individual wheelchair configurations is currently achieved through trial-and-error approaches [1] that are time consuming and expensive. Systematic testing methodologies [14] reduce these limitations while still providing detailed information regarding configuration effects. While this approach is an improvement on current methods, predictive models have potential for further developments. Predictive models have the potential to substantially reduce the amount of player testing involved in determining the 'optimal' wheelchair configuration for each individual. Whilst sub-maximal models have been developed [38, 39], to the authors knowledge, a predictive model for maximal effort propulsion has yet to be achieved.

3.8 Conclusion

Wheelchair configurations used in court sports have received limited attention. This research documented wheelchair configurations used by the Australian WCR, who are the reigning World and Paralympic Champions. Higher seat heights and seat depths were used by high-point players compared with low- and mid-point players, respectively. Seat angle and backrest height were significantly smaller for high-point players compared with both mid- and low-point players. Low-point players used wheelchairs with reduced mass and frame lengths compared with both high- and mid-point players. These findings are reflective of the various impairments across classifications, as well as on-court roles. Consideration of anthropometrics in wheelchair configurations is seemingly important, with ratios of seat height-to- total arm length and backrest height-to-torso length differing across classifications. Findings from this work provide quantitative data on the wheelchair configurations used by elite WCR players, and promote the consideration of an individual's anthropometrics, impairment, training history and court role, with sports engineering approaches such as predictive modelling providing potential benefits in reducing the time and cost associated with determining optimal wheelchair configuration.

Conflict of Interest

The authors declare no conflicts of interest.

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Chapter 4: Overground-Propulsion Kinematics and Acceleration in Elite Wheelchair Rugby

Detailed propulsion kinematics are limited for WCR, with this chapter addressing this gap by reporting kinematics across an elite squad during acceleration from standstill. Potential performance implications of propulsion approach are also presented.

This chapter has previously been published (see below details) and has been reformatted for the purpose of this thesis. This publication satisfies University of Adelaide requirements for inclusion in a Thesis by Publication.

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Name of Principal Author (Candidate)	David S. Haydon		
Contribution to the Paper	Led testing design, conducted testing, completed associated video analysis, interpretation of results, and the preparation of the manuscript.		
Overall percentage (%)	55		
Certification:	This paper reports on original research I conducted during the period of my Higher Degree by Research candidature and is not subject to any obligations or contractual agreements with a third party that would constrain its inclusion in this thesis. I am the primary author of this paper.		
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Co-Author Contributions

By signing the Statement of Authorship, each author certifies that:

i. the candidate's stated contribution to the publication is accurate (as detailed above);

ii. permission is granted for the candidate in include the publication in the thesis; and

iii. the sum of all co-author contributions is equal to 100% less the candidate's stated contribution.

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4.2 Abstract

Purpose: Maximal acceleration from standstill has been identified as a key performance indicator in wheelchair rugby; however, the impact of classification and kinematic variables on performance has received limited attention. This study aimed to investigate kinematic variables during maximal acceleration, with level of activity limitation accounted for using sport classification scores. **Methods:** 25 elite wheelchair rugby players were analysed in high-, mid-, and low-point groups based on their sporting classification score, which reflects the combined trunk, arm, and hand function, before completing five 5m sprints from a stationary position. Inertial measurement units and video analysis were used to monitor key kinematic variables. **Results:** Significant differences in kinematic variables were evident across the classification groups, particularly for the first stroke contact angle (one-way ANOVA, $F(2,122)=51.5$, $p<0.05$) and first stroke time ($F(2,124)=18.3$, $p<0.05$). High-point players used a first stroke contact angle that was closer to top dead centre of the wheel than both other groups, while also utilising a shorter overall stroke time than low-point players. A linear mixed effects model investigated how kinematic variables influenced performance, with results suggesting that increased release angles (i.e., further around the wheel) and decreased stroke angles resulted in larger peak accelerations. Further investigation revealed that these results are likely influenced by strong relationships for the high-point group, as there was often no clear trend evident for mid- and low-point groups. **Conclusion:** Findings show that various propulsion approaches exist across classification groups, with this information potentially informing individual wheelchair set-ups and training programs.

Keywords

Classification; Propulsion technique; Sprint; Paralympic sport

4.3 Introduction

Wheelchair propulsion kinematics have been investigated across a range of activities, including wheelchair basketball [1], wheelchair racing [2] and daily living [3]. Variables such as contact and release angles, as well as stroke and recovery times have been used to assess variations across athlete classifications [1] as well as performance outcomes [2]. Despite an increase in popularity and research in wheelchair rugby (WCR), there is currently a limited understanding of how the level of activity limitation affects key kinematic variables and their impact on chair acceleration and sprint performance, particularly when investigating on-court testing [4]. In WCR, players with specific impairment types (e.g., limb deficiencies, impaired muscle power) are eligible to be classified [5] with individual classes based on the sport specific activity limitation as a result of these impairments. Players are grouped using a point system from 0.5-3.5 where a higher point score indicates a greater degree of overall combined strength, range of motion, and co-ordination (referred to throughout this study as 'function') of the trunk, arm, and hand [6]. In the case of limb deficiencies, particularly multiple amputees, the limb length is also considered (for example, longer leg length corresponds to higher trunk score) [6]. While players within the same class may have different impairment types, they are deemed to have a similar level of activity limitation [7]. Each team is allowed 4 players on-court at a time, with their total point score limited to 8 points [8]. In research, players are often separated into broader groups than these classification scores; a high-point group of 3.0 and 3.5-point players; mid-point group of 2.0 and 2.5-point players; and low-point players of 0.5, 1.0 and 1.5-point scores [9, 10]. Propulsion techniques are expected to differ across classifications due to differences in trunk, arm, and hand function [11]. It is currently unclear whether propulsion kinematics differ substantially within classification groups and how this affects key performance variables such as acceleration and sprint performance.

Acceleration from standstill has previously been identified as a key performance factor in sports including wheelchair racing [2], basketball [12] and WCR [13, 14]. For WCR, this is due to the large number of high intensity efforts [15] and stop-starts involved [16]. As a result, studies have begun to focus on maximal acceleration from standstill [4, 17-19], rather than steady-state propulsion [3, 20, 21]. However, most of these studies that have monitored kinematic variables in acceleration from standstill have been performed on ergometers [4, 17]. Using ergometers and treadmills in laboratory testing conditions has been shown to alter kinematics compared with overground propulsion [22, 23], with stroke angles and times increased using laboratory testing methods [22]. Propulsion studies have also considered the effect of altering variables such as seat angle [17] or the inclusion of abdominal binding (the use of an elastic binder to apply an external abdominal compression) on performance [4] rather than variations in propulsion across different classifications. This has resulted in limited investigations using conditions representative of performance in WCR to assess kinematics and the effects of propulsion technique on sprint performance.

This study aims to investigate the effect of activity limitation on key kinematic variables during on-court maximal acceleration from standstill for highly experienced WCR players. The relationship between key kinematic variables for the first three strokes, identified in previous research as crucial to acceleration [4], and resulting acceleration (5m sprint time and peak stroke acceleration) was investigated. Trunk function has been shown to have the largest impact on the first metre when accelerating from standstill, before arm function became the predominant influence [24]. Additionally, Vanlandewijck et al. [17] found that, when considering seat angle variations, positions that allow for greater trunk motion promoted a contact position with increased trunk flexion; however, it should also be noted this study used able-bodied participants. This propulsion approach has the advantage of utilising gravitational forces to counterbalance the trunk extension reaction forces generated by the hand pushing on the wheel [25], but limits the stroke angle as contact occurs further around the wheel. It was therefore expected that high-point players (those with combined trunk, arm, and hand functions that have a reduced impact on sport performance) would use shorter stroke angles and stroke times compared with both mid-point and low-point players to maximise this benefit. This information would provide greater understanding of propulsion approaches to maximise wheelchair set ups and training designs for various sports classes based on trunk and arm function.

4.4 Method

4.4.1 Participants

25 WCR players (age 30.5±7 years) with a minimum of at least two years of national level experience (n=5, 3.6±1.9 years) participated in testing, with most having previous international experience (n=20, 7.7±6.5 years). Testing was performed after ethical approval from the required institutions and all players provided written informed consent (H-2015-127). All players had previously undergone national and/or international WCR classification processes, performed by a certified classifier. This involves assessment of arm function and an evidence-based approach assessing trunk function (muscle strength of the trunk and the legs, length of the legs, range of movement of the hips and the trunk, and coordination of trunk movements), as well as on-court activity observations [8]. Players were grouped by classification scores to align with previous research [9]: a high-point group (n=7) of 3.0 (n=3) and 3.5 point players (n=4); a mid-point group (n=9) 2.0 (n=7) and 2.5 (n=2) point players;

and a low-point group (n=9) of 0.5 (n=6), 1.0 (n=2) and 1.5 (n=1) point players. This grouping approach placed players with proximal arm weakness together (low-point group), and separated the recommended 'offensive' class (2.0-3.5-points) [6] based on activity limitation effects on sporting performance. All 3.5-point players in the current sample had limb deficiencies, while no other players had this type of impairment.

4.4.2 Experimental Design

Players performed five 5m sprints following established testing protocols [18, 19] and initiated trials in their own time. The testing was performed as the final stage of a warm-up before competition, with participants instructed to complete each trial as quickly as possible and given sufficient break between trials to ensure fatigue was not a contributing factor. For each sprint, video (100Hz, Sony HDR-PJ 430) was recorded from both side and rear views of the participant for kinematic analysis [4, 17]. Both cameras had a fixed view, with the side camera focused on capturing the first three strokes for each player. Two-dimensional analysis of hand contact points was deemed to be appropriate for linear wheelchair sprints, with analysis focussed on the motion of key variables primarily in a single plane (i.e. placement and release of hand on the wheel) [4, 17].

4.4.3 Performance variables

The 5m sprint time was recorded using a laser timing system (Kinematic Measurement System, Fitness Technology, Australia). Peak acceleration for each stroke was recorded using a tri-axial accelerometer (x8m-3mini, Gulf Data Concepts, USA) which was secured to the front and centre of the footplate on the wheelchair frame. Data was recorded at 100Hz and low-pass filtered (2nd order, Butterworth) at 10Hz [26], with each stroke identified and the associated peak acceleration determined.

4.4.4 Kinematic Analysis

Kinematic variables including hand position at the first instant of contact with the wheel (ContAng), release angle (RelAng), stroke angle (StrokeAng), stroke time (StrokeTime), recovery time (RecTime), cycle time (CycTime) and stroke distance (StrokeDis) were then calculated using Kinovea [27] (Version 0.8.15, kinovea.org) on each of the first three propulsion strokes of the sprint. The calibration of the video was performed using a distance measure positioned in the centre of the video image and in-line with the plane of motion. Parallax and perspective errors were minimised by using the 3-4-5 triangle approach to distance, and ensuring the camera was perpendicular to the player's plane of motion throughout the trial. Additionally, the camera was zoomed in to fill the field of view [28].

Timing of hand contact and release were identified using the first and final point of contact with the wheel, respectively [4]. Synchronised side and rear cameras were used to identify the first frame the hand was deemed to be in contact with the wheel. Reflective markers aided the estimation of joint positions. ContAng and RelAng were measured as the angles between the top dead centre (TDC) of the wheel (i.e., 0° or vertical) and the approximate location of the second metacarpophalangeal joint on the right hand [29], with StrokeAng being the difference between the two. Additionally, this allowed for the determination of StrokeTime (the time in which the hand is in contact with the wheel), RecTime (the time between the release of one stroke and contact of the next) and CycTime (stroke plus recovery time); variables previously identified by West et al. [4] as key outcome measures in the propulsion cycle. StrokeDis was the horizontal distance the axis of the wheel moved throughout the stroke. All timing variables were confirmed using a combination of side and rear view cameras. Intra- and inter-evaluator reliability of kinematic analysis was assessed using the technical error of measurement (TEM) [30]. A random selection of 20 trials was chosen and re-analysed by the lead researcher two-weeks after initial analysis, as well as by an additional experienced researcher. Reliability was deemed to be good to moderate (2.6-9.7% TEM) for all variables for both intra- and inter-evaluator reliability [31].

4.4.5 Statistical Analysis

One-way Analyses of Variance (ANOVA) used to assess the effect of classification grouping on both performance and kinematic variables. Bonferroni corrections were performed to control for Type I errors while Games-Howell testing was completed for instances of unequal homogeneity, with effect sizes calculated as Cohen's *d* [32].

A linear mixed effects model with random slope and intercept was used to investigate the influence of kinematic variables on performance using the peak acceleration for the corresponding stroke. Covariates initially included ContAng, RelAng, StrokeAng, StrokeTime, RecTime, CycTime, and StrokeDis, with significant covariates determined using backward elimination. Factors including the classification group and repeated trials were also accounted for. All statistical analysis was completed using IBM SPSS Statistics 21 (2012).

In addition to the model, scatter plots were produced to aid in the visualisation of the effects of specific kinematic variables. Kinematic variables were selected as the significant covariates from the mixed effects model. As a direct result of this visualisation, specific individual point scores corresponding to each extreme on the classification scale (0.5– and 3.5–points), as well as the median classification score (2.0–points) were investigated *a posteriori*. These sub-group sizes warranted additional analysis, and were deemed to be important given the study group and clear masking of potentially important findings based on typical larger research groupings previously used.

4.5 Results

4.5.1 Performance

A one-way ANOVA revealed significant differences for sprint time between classification groups ($F(2,124)=176.2$, $p<0.05$), with high-point players achieving faster sprint times (1.83±0.22s) than both the mid-point (2.06±0.13s, $p<0.05$, $d=-1.3$) and low-point (2.58±0.20s, $p<0.05$, $d=4.07$) groups. Mid-point players also achieved a faster sprint time than low-point players ($p<0.05$, $d=2.81$).

4.5.2 Kinematic Variables

ContAng and RelAng across the first three strokes were found to vary for classification groups (ContAng1: $F(2,122)=49.7$, $p<0.05$; ContAng2: $F(2,124)=27.7$, $p<0.05$; ContAng3: $F(2,124)=22.7$, $p<0.05$, RelAng1: $F(2,124)=32.1$, $p<0.05$; RelAng2: $F(2,124)=28.8$, $p<0.05$; RelAng3: $F(2,123)=24.6$, $p<0.05$) with magnitudes and significant differences between groups presented in Figure 4.1. StrokeAngs also differed, with high-point players using smaller StrokeAngs (Stroke 1: 92±19˚; Stroke 2: 96±15˚; Stroke 3: 102±18˚) than mid-point players (Stroke 1: 101±23˚; Stroke 2: 112±22˚; Stroke 3: 113±21˚) for strokes two and three ($p<0.05$, $d>0.62$) and low-point players (Stroke 1: 123±29˚; Stroke 2: 113±15˚; Stroke 3: 116±18˚) for all strokes ($p<0.05$, $d>0.78$). Mid-point players had a smaller StrokeAng than low-point players for stroke one ($p<0.05$, $d=0.93$). StrokeAngs are evident in Figure 4.1, however only ContAng and RelAng significant differences are shown.

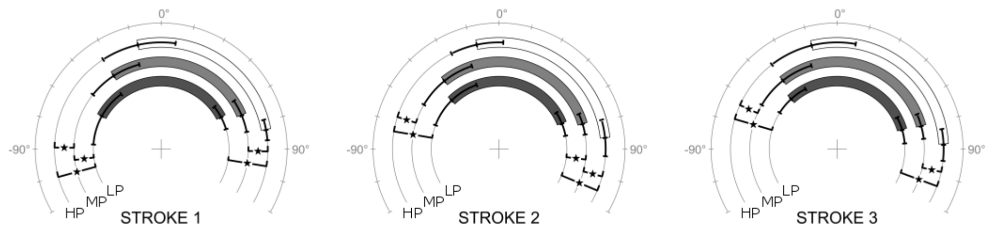

Figure 4.1: Mean (± SD) and significance (at 0.05 level, shown by starred identifiers) across classification groups for ContAngs and RelAngs for all strokes. The stroke direction is to the right, with values presented visually where each bar represents a classification group.

Table 4-1 presents the magnitudes and significant differences for the StrokeTimes, RecTimes, CycTimes, and StrokeDis for all strokes. Variations were identified across groups, with high-point players using the fastest stroke approach, and low-point players using the slowest. The differences in techniques result in varying proportions of 'push' and 'pull' throughout the strokes. The 'push' segment of the stroke refers to the region or time in which the players are pushing down on the wheel, while the 'pull' approach refers to when the player has contacted and is moving the hand up with the wheel, and is from initial hand contact until the minimum elbow angle is reached [11]. High-point players displayed smaller, and often positive, ContAngs (i.e., closer to TDC of the wheel) than both mid-point and low-point players, as well as larger RelAngs for all strokes. This resulted in the high-point group reaching a minimum elbow angle at hand locations ranging from 20-25° clockwise from TDC across the three strokes, corresponding to proportions of 'pull' of 33-36%. In contrast, mid-point players consistently showed proportions of 'pull' of 47-50% across all strokes, while low-point players showed greater variation with 59% 'pull' for the first stroke, but 46% for the third stroke.

Table 4-1: Mean (±SD) for each of the classification groups for StrokeTime, RecTime, CycTime and StrokeDis for the first three strokes. Differences between groups following post-hoc testing are also presented.

	High-point	Mid-point	Low-point	Significant Differences (Post-hoc)
StrokeTime1 (sec)	0.59 (.12)	0.64 (0.12)	0.77 (0.10)	**,***
StrokeTime2 (sec)	0.27 (0.06)	0.35 (0.06)	0.41 (0.05)	*,**,***
StrokeTime3 (sec)	0.22 (0.05)	0.28 (0.05)	0.36 (0.05)	*,**,***
RecTime1 (sec)	0.22 (0.04)	0.23 (0.03)	0.30 (0.04)	**,***
RecTime2 (sec)	0.22 (0.03)	0.23 (0.02)	0.30 (0.05)	**,***
RecTime3 (sec)	0.22 (0.03)	0.24 (0.03)	0.29 (0.05)	**,***
CycTime1 (sec)	0.81 (0.15)	0.88 (0.14)	1.05 (0.15)	**,***
CycTime2 (sec)	0.48 (0.08)	0.58 (0.08)	0.71 (0.08)	*,**,***
CycTime3 (sec)	0.44 (0.07)	0.51 (0.07)	0.64 (0.08)	*,**,***
StrokeDis1 (m)	0.53 (0.08)	0.57 (0.11)	0.64 (0.10)	**,***
StrokeDis2 (m)	0.55 (0.07)	0.67 (0.14)	0.65 (0.08)	*,**
StrokeDis3 (m)	0.58 (0.07)	0.68 (0.15)	0.68 (0.09)	*,**

* signifies difference between high- and mid-point groups; ** signifies difference between high- and low-point groups; and *** signifies difference between mid- and low-point groups.

4.5.3 Mixed Effects Statistical Model

Results for the mixed effects model produced significant covariates of the RelAng and StrokeAng. For the third stroke, an increase in RelAng was associated with an increase in peak acceleration (0.048±0.018m/s^2, $p<0.01$). This result indicates there is an expected increase in peak acceleration of 0.049m/s^2 for each degree the RelAng is increased by for the third stroke. Conversely, an increase in the third StrokeAng was associated with a decrease in peak acceleration ($-0.037\pm0.013 m/s^2$, $p<0.01$). The first and second strokes produced no significant relationships with the resultant peak acceleration. Peak acceleration was also influenced by classification group, with significant differences between groups for all strokes. High-point participants produced greater peak acceleration than the low-point group for all strokes ($p<0.001$), and the mid-point group for strokes one and two ($p<0.013$) as well as a trend for stroke three ($p<0.054$). There were no significant differences between the mid- and low-point peak accelerations for any investigated strokes.

The results from this statistical model are supported by Figure 4.2, which displays scatter plots for all participants and trials for the comparison of high, mid-, and low-point groups. RelAngs are presented against the corresponding peak acceleration for the third stroke, which was identified as a significant covariate. Although the lines of best fit for the mid- and low-point groups show no obvious trend with increasing RelAng, there is a positive trend for increasing RelAng and the peak acceleration for high-point players. Following the secondary analysis for specific point scores, this was found to be related to the varying propulsion approach of the 3.5-point players.

4.6 Discussion

This study investigated the impact of activity limitation (reflected in the classification scores) on kinematic variables and resultant wheelchair acceleration during on-court propulsion in WCR. This is the first study to focus on the differences in kinematic variables in maximal effort sprinting from standstill between classification groups, as well as utilising a large, experienced participant sample [4, 19].

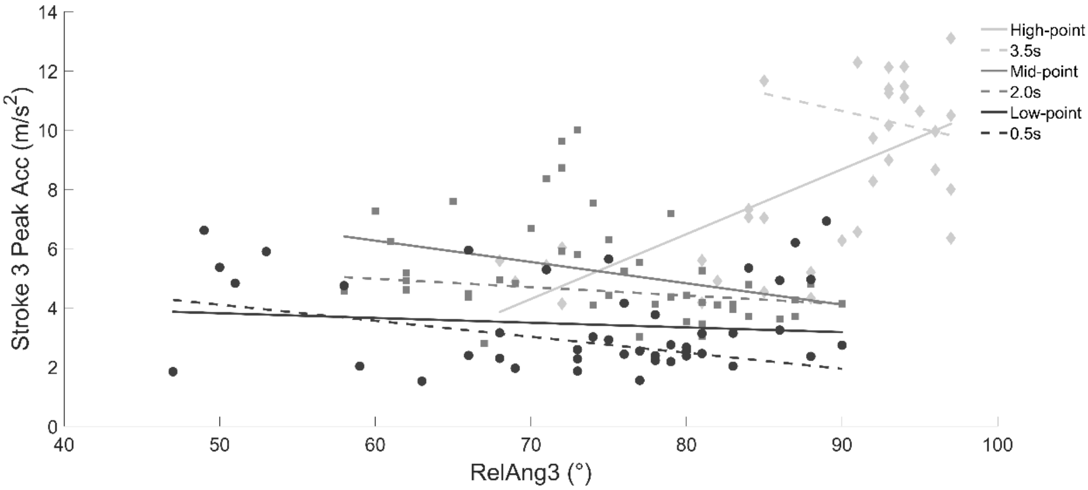

Figure 4.2: Scatter plot for the RelAng against the peak acceleration for the first three strokes of the 5m sprint. Lines of best fit (least squares approach) are plotted for the high-, mid-, and low-point groups, as well as the specific point scores of 3.5, 2.0, and 0.5 players. Data points for high-point players are represented by diamonds, mid-point players by squares, and low-point players by circles.

Results suggest there is significant differences in key kinematic variables between research groups, highlighting two key components; 'push' and 'pull'. High-point players used a greater proportion of 'push' throughout their strokes which is likely due to greater trunk function and impairment type, compared with mid-point and low-point groups. The high-point group in this study included players with limb deficiencies (all 3.5 players in the sample) and incomplete spinal cord lesions, therefore displaying greater trunk function. This allows the player to lean forward and reach further around the wheel, increasing the RelAng. The increased function also allows for greater variation in wheelchair set-up with set-up known to influence propulsion performance and kinematics [4, 17]. For high-point players, this includes parameters such as the seat height, where improved trunk strength and range of motion allows for greater regions of contact with the wheel compared with low-point players when seat height is increased [33]. Utilising this promotes the 'push' technique and allows high-point players to apply increased force to the wheel due to the contribution of both trunk and arm motion. As the trunk motion benefits propulsion for a limited range around the wheel, it is expected the smaller StrokeAng and StrokeTime used by high-point players attempts to maximise this zone. A shorter StrokeTime reflects both the reduced StrokeAng and increased muscular function of high-point players. Increased muscular function reduces StrokeTime as greater forces can be applied to the wheels, increasing peak acceleration and hence faster movement of the hand around the wheel.

Mid- and low-point groups used a greater proportion of 'pull' than high-point players. This enables players to maximise their bicep function, promoting a 'pull' technique, which is maximised by using a ContAng further anti-clockwise from TDC. In addition, a larger RelAng was evident for the mid-point group compared with the low-point group in stroke two, due to greater triceps contribution in the mid-point group.

StrokeTimes for mid-point players were shorter for all strokes compared with low-point players. Increased muscular function of mid-point players compared with low-point players allows greater force application and faster wheel rotation, as measured through decreased contact times, increased peak acceleration and reduced sprint times. RecTime also differed, with the low-point group having longer RecTimes than high- and mid-point groups; with low-point players reduced muscular function, and increased StrokeAngs (i.e., distance for hand to travel) compared with high-point players, inhibiting their ability to match the short recovery times of other groups. Minimising the RecTime has performance benefits, as it reduces the time in which the frictional resistance between wheels and court is the main force acting on the wheelchair. CycTimes reduced with each classification group (i.e., increasing level of activity limitation) with relatively stable trunk positions following the first stroke, as have previously been reported [17]. This likely reduces the

RecTime and CycTime as repositioning of the trunk in preparation for the stroke phase is not required.

The linear mixed effects model findings demonstrate that increases in RelAng and a decrease in total StrokeAng significantly influence peak acceleration for the third stroke. As the third stroke is likely to be similar to the following strokes throughout a sprint, increasing the RelAng but shortening the StrokeAng will benefit acceleration and performance. Although not significant, similar trends were present for strokes one and two, where increased RelAng, and decreased StrokeAng are associated with greater acceleration for the high-point group. To achieve this, a ContAng closer to or clockwise of TDC is required, maximising the proportion of 'push'. This promotes a short, fast stroke approach to initiate motion in a linear sprint. Results here are supported by acceleration data; compared with the mid-point group, the high-point group achieved 1.96 m/s^2 more acceleration for each stroke, while this was increased to 3.14 m/s^2 compared with the low-point group.

Figure 4.2 presents the peak acceleration against the RelAng for the third stroke. The lines of best fit for the mid- and low-point groups are relatively horizontal, demonstrating no clear relationship. However, the high-point group show strong positives relationship for the RelAng; therefore, the mixed effects model results appear strongly influenced by the high-point players trends. By considering the specific point-scores presented in Figure 4.2, it is evident that differences between 3.0 and 3.5-point players exist, and it is therefore important to ensure that measurements accurately reflect individual characteristics rather than simply assessing group tendencies [34]. While previous research has considered the broader high-point, mid-point and low-point groups, a point specific analysis allows for analysis of similar levels of activity limitation and the impact on performance. The range in magnitude for

3.5-point players is reduced (shown by the horizontal length of the line) compared with the overall high-point group data showing that 3.5-point players investigated use a similar propulsion approach to each other. The slope of the line also varies for the 3.5-point group, showing a weaker relationship than the high-point group. Variance in kinematic variables between groups may be a reflection of the range of activity limitation within each group. Players can be given similar classification scores despite large differences in physical impairments. All 3.5-point players in this study group had limb deficiencies, whereas 3.0-point players all had impaired muscle power due to incomplete spinal cord injuries. This results in varying levels of trunk function between the two point-scores, yet they have previously been considered as part of the same group [9, 35]. Although this distribution of impairments does not represent the international populations for the 3.5-point group (as the higher classification scores can be reached through various combinations of trunk and arm scores), six of eight 3.5-point players at the 2016 Paralympic Games had limb deficiencies, representing a large proportion of this classification. For 3.5-point players with limited trunk function, but good arm and hand function, the trends discussed may not be evident. Conversely, low-point players have a similar impairment (impaired muscle power), and may account for findings presented here. For example, in Figure 4.2 when comparing the low-point and 0.5-point groups, both the range of the variables and slope of the line appear similar; supporting the similarity in techniques across the low-point classification group. Group designs and mixed effects models may be masking important differences between classification groups in similar research, and statistical models for each group are likely to provide greater insight into the effects of propulsion technique. However, the relatively small population size of elite and experienced WCR players limits studies in acquiring adequate statistical power; alternative analyses such as small group designs and individual case studies in elite populations would allow for a greater understanding of how to optimise chair propulsion for athletes with varying activity limitation.

Future research should continue to look at smaller sub-group and individualised analyses where possible to provide insights into the impact of activity limitation on key performance measures. Here, we advocate for the careful consideration of classification groupings in research, and suggest a greater emphasis on specific physical impairments, particularly in high-point groups. In addition, it should be noted that linear sprints represent only one of many activities in WCR, with agility and ball handling also crucial to performance.

4.7 Practical Applications

Greater understanding of propulsion techniques across classification groups in elite populations, as well as factors that influence performance, will aid athletes and coaches in improving individual propulsion approaches. The identification of strong and weak regions of a stroke can also influence training interventions to promote the desired technique. Specific propulsion strategies should also be considered when altering wheelchair set-ups, with wheelchair configuration repeatedly shown to affect propulsion techniques. For high-point players, this may involve reducing the fore-aft position to promote a reduced StrokeAng [36]. Performance factors, such as ball handling and stability, are also crucial to on-court success and should be considered during wheelchair set-up and testing.

4.8 Conclusion

Data from the current study suggests that overall level of activity limitation in WCR leads to notable differences in propulsion techniques (kinematic variables) during maximal effort sprints from standstill. High-point players use a greater proportion of 'push' throughout their stroke, while low-point players utilised an increased amount of 'pull' in their technique. These differences are related to the activity limitations of individuals, with high-point players in the current sample having higher levels of trunk function allowing increased RelAngs. Linear mixed effects model results showed that increased RelAng and decreased StrokeAng resulted in improved peak accelerations for the third stroke. This could be achieved through a ContAng closer to or clockwise of TDC on the wheel. Further investigation into these results revealed that these results are likely influenced by strong relationships for the high-point group due to differences in propulsion technique of 3.0 and 3.5 point players, as there was often no clear trend evident for mid- and low-point groups. Findings have implications for the design of training interventions and wheelchair design to maximise propulsion and sprint performance, as well as the adoption of an individualised analysis approach when considering wheelchair set-up and propulsion. Future studies are required to assess kinematic responses both within and between classification groups or for specific physical impairments, particularly for high-point classifications. To achieve this, testing of players across multiple elite pathways and collaborations between countries is likely required.

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Chapter 5: Intra-Stroke Acceleration Profiling of Elite Wheelchair Rugby Players

Intra-stroke profiling, in addition to monitoring propulsion kinematics as reported in Chapter 4, provides a method for more detailed assessment of propulsion approaches on performance. This chapter details the differences seen across three players of varying classifications and key information that this can provide.

This chapter has been submitted (see below details) and has been reformatted for the purpose of this thesis. This submission satisfies University of Adelaide requirements for inclusion in a Thesis by Publication.

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Overall percentage (%)	50%		
Certification:	This paper reports on original research I conducted during the period of my Higher Degree by Research candidature and is not subject to any obligations or contractual agreements with a third party that would constrain its inclusion in this thesis. I am the primary author of this paper.		
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i. the candidate's stated contribution to the publication is accurate (as detailed above);
ii. permission is granted for the candidate in include the publication in the thesis; and
iii. the sum of all co-author contributions is equal to 100% less the candidate's stated contribution.

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5.2 Abstract

Intra-stroke acceleration profiling could provide detailed information for enhancing performance in wheelchair sports. To date, no research has investigated this, likely due to the difficulty in monitoring acceleration in on-court performance, and the individual nature of propulsion kinematics of players in sports such as wheelchair rugby. This study investigated the potential of using inertial measurement units for intra-stroke acceleration profiling. Three elite wheelchair rugby players (international experience: 5 ± 3.5 years) each with a different level of physical impairment, completed six five-metre linear sprints from a stationary position in their own wheelchair. High speed video was recorded from both side and rear views, with inertial measurement units secured to the frame and each wheel, to monitor overall acceleration and specific left-hand and right-hand contacts. Individual intra-stroke acceleration profiling was deemed successful with the current method and use of IMUs. Differences were demonstrated for timing of peak accelerations, and hand position around the rim aligned with the particular peaks. Peak accelerations are able to identify regions of the propulsion stroke that are indicative of individual skill execution. Intra-stroke acceleration profiling can provide coaches with detailed feedback on the significance of changes to wheelchair set-ups or technical changes at an individual level.

Keywords

Propulsion technique; Wheelchair sprinting; Paralympic sport; inertial measurement units; propulsion kinematics

5.3 Introduction

In recent years, propulsion in wheelchair court sports has received increased levels of published research [1, 2]; however, detailed analyses into the propulsion stroke have been limited. In wheelchair rugby (WCR), propulsion has been shown to differ significantly across players with varying classification scores [3]. Players are eligible for WCR if they have an impairment that affects at least three limbs and the trunk – such as impaired muscle power (i.e., from spinal cord injuries (SCI)) or limb deficiencies [4]. Players are then assigned a classification score which represents the level of sport-specific activity limitation. Scores range from 0.5-3.5 points depending on the player's trunk, arm, and hand function (where 'function' considers strength, range of motion, and coordination), with higher scores indicating reduced limitations to sport performance [5]. Each team is allowed four players on-court at any one time, with a maximum combined score of eight points.

Due to the broad differences in activity limitation across classifications, players adapt propulsion techniques to maximise their individual muscular function. Previous work has shown variations in technique across low- (0.5–1.5 points, i.e., impairments that have a greater impact on performance), mid- (2.0–2.5 points) and high-point players (3.0–3.5 points, i.e., impairments that have a reduced impact on performance). These differences include variations in contact, release, and stroke angles, as well as stroke and recovery times [6]. High-point players with greater trunk function have also been shown to use a greater proportion of 'push' throughout their propulsion stroke [3]. 'Push' refers to the region of the stroke following the minimum elbow angle (often occurring near top dead centre (TDC) of the wheel) and the elbow is extending [7]. The region of the stroke before minimum elbow angle is reached (i.e., while the elbow joint is flexing) is referred to as 'pull', and this approach is more prevalent amongst low-point players [3].

Intra-stroke profiles have received limited attention in wheelchair sports (see Moss et al. [8] for an exception in wheelchair racing), particularly at an individual level of analysis. This may be due to previous instrumentation approaches, such as the SMARTWheel [9], being inappropriate for on-court monitoring. Recent advancements in sensor technology and algorithms [10-12] can allow for analyses in on-court situations with minimal and no impact on athlete performance. Detailed intra-stroke acceleration profiling has the potential to provide highly detailed information on an individual player's stroke (technique), including highlighting regions in which acceleration events occur, such as peaks and troughs – where a trough is a temporary decrease in acceleration occurring during the transition from biceps brachii to triceps brachii contributions [13]. These variables have previously been shown to differ across the first three strokes of a wheelchair racing start, before showing greater consistency from strokes four to six [8]. This is expected to be highly reliant on hand position on the wheel, reflecting regions in which the individual's level of physical function is maximised; in WCR, this is based on the contribution of the trunk, arm, and hand.

Asymmetries in propulsion approaches have also been investigated recently [14], with assessment of these on-court possible with inertial measurement units (IMUs). Asymmetries during linear sprinting can cause difficulties in maintaining the intended direction, which has obvious performance implications. Understanding asymmetries – whether as an inherent part of the player's impairment, or an area which can be improved through awareness and training – has the potential to allow for improved performance in linear sprinting.

By identifying regions of propulsion strokes which maximise an individual player's muscular strengths, performance staff (e.g., coaches, biomechanists) can potentially inform technique changes or modifications to wheelchair set-ups to further enhance acceleration and performance. Therefore, the initial aim of this study is to investigate the intra-stroke acceleration profiles and identify differences between players of differing classification scores in WCR. It was hypothesised that combining high speed video analysis and IMU data would identify variations in individual profiles, in particular the timing and magnitude of peak acceleration for the first three strokes. Furthermore, the presence of propulsion asymmetries (i.e., varying hand contact and release times, as well as positions) was investigated due to the lack of associated research and the potential performance implications due to factors such as steering compensation. Greater asymmetries were expected for low-point players due to the increased levels of activity limitation and subsequent potential discrepancies in symmetric muscle function.

5.4 Methods

5.4.1 Participants

Ethical approval was provided by the required institutes and written, informed consent obtained from all players prior to commencing testing. Three elite WCR players (age: 26±3 years; all male) from the Australian WCR team were recruited and completed the testing protocol. Player information including classification, impairment, international experience, and wheelchair configuration is presented in Table 5-1.

5.4.2 Testing Protocol

Players completed six, five-metre linear sprints on-court from a stationary position in their own rugby wheelchair. Each trial was initiated by the player in their own time. Players were given sufficient break between trials to ensure that fatigue had no effect. Three IMUs (500Hz, acceleration range of ±16g, mass of 12 grams, IMeasureU, NZ) were secured to the wheelchair: one to the centre-front of the frame, and one on each wheel to monitor any asymmetries in timings. The position of the IMU on the wheel was selected in consultation with the participant to ensure no interference with the stroke would occur. Synchronised video (120Hz, Go Pro Hero 3+, California, U.S.) was recorded from left and right sides and the rear view, with side cameras positioned to capture the first three strokes of the sprint; this was deemed to be important, as repeated acceleration from standstill is a key component of WCR performance [11, 15]. Cameras were positioned perpendicular to the player's plane of motion throughout the linear sprint, reducing the effect of perspective errors.

Prior to initiating the sprint, a sharp strike (a physical impact from a researcher) was exerted on the wheelchair frame causing a clearly identifiable peak in the acceleration trace of all IMUs (synchronisation event). The player remained stationary before and after the strike to ensure the strike was evident in both the IMU trace (Figure 5.1) and video. The synchronisation of these monitoring systems allows for the identification of the hand position during specific regions of acceleration throughout the stroke. Sprint times were also recorded using laser timing gates (SpeedLight, Swift Performance) as an overall measure of performance and to ensure players continued at maximal effort throughout the testing session.

5.4.3 Analysis

A custom MATLAB (R2016a) script was written to analyse the data with user prompts for each major step. The peak of the synchronisation event was used to synchronise all IMU data which was then down-sampled to match the video frequency, with video synchronisation using the frame where impact was first evident. Following synchronisation, the frame IMU data was low-pass filtered at 20Hz (Butterworth filter, order 5, bidirectional, -6dB cutoff frequency), well above the recommended cut-off for daily activities [16] and similar to previous wheelchair sport studies [17]. As the accelerations measured from the wheel IMUs were critical in identifying the high frequency behaviour due to contact and release points, as well as the start of the sprint, this data was not low-pass filtered. The contact and release events were evident in the raw wheel acceleration data as momentary alterations to the cyclic nature of the acceleration profile of the wheel (Figure 5.1). Manual selection of expected contact and release points prompted the corresponding video frame for the side and back view to be displayed on-screen, with a researcher experienced with wheelchair propulsion then confirming if this was the point of contact (first frame hand contacts the wheel) or release (last frame of contact between hand and wheel before releasing) [3]. The hand position in the specific frame of interest was then analysed using the custom MATLAB code. These angles, positive in the direction of wheel rotation, were measured from TDC of the wheel (defined as 0°) to approximately the second metacarpophalangeal joint, individualised for each player depending on the use and type of gloves. Key points of the filtered frame acceleration trace (AccFrame), such as peaks and troughs, were then investigated using a similar approach. Hand positions (for both right and left hands) at the corresponding video frames for the peak and trough events were then calculated using wheel and hand positions.

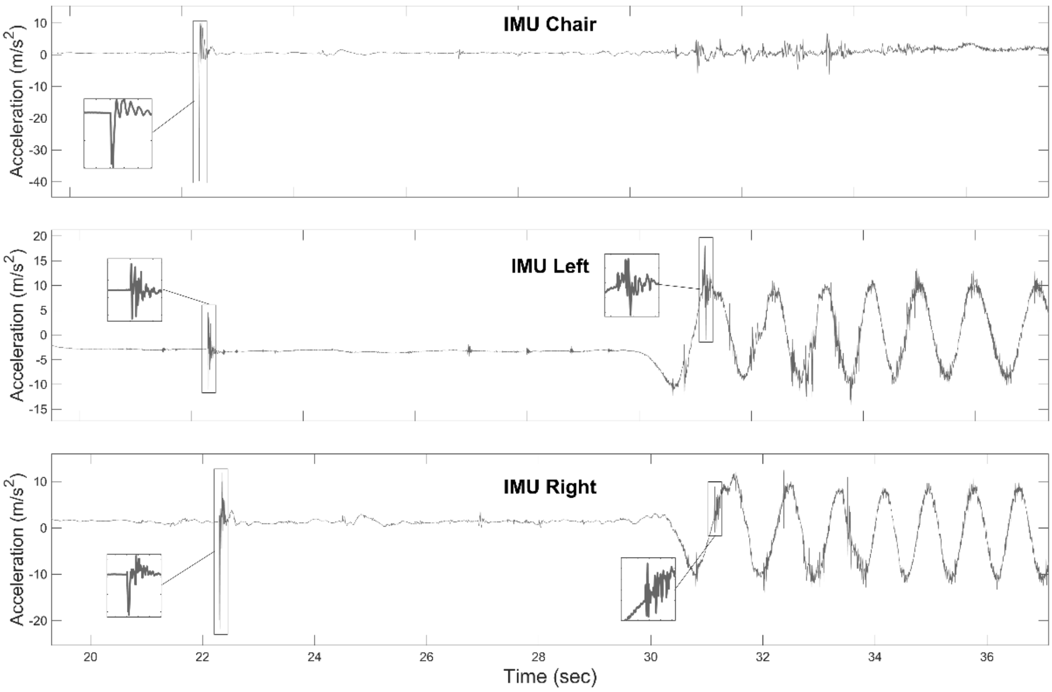

Figure 5.1: The IMUs were synchronised using a strike that caused a peak acceleration that was preceded and followed by stationary periods. In addition, contact and release points were identified with the aid of wheel IMUs, where alterations to the cyclical acceleration trace represented the left and right hands separately.

For each sprint (N=6), AccFrame in the direction of travel was expressed in normalised time from sprint start to fourth contact point, capturing the first three stroke cycles of each player. The normalised time traces were re-sampled onto a regular grid (1000 data points in total, where possible), with these used to calculate mean and standard deviation acceleration profiles for each player respectively. The normalised time steps corresponded to increments of 0.0029, 0.0022, and 0.002 seconds between data points, according to the times taken for the first three stroke cycles of each sprint (2.86±0.19, 2.20±0.14 and 1.78±0.05s, for the 0.5-, 2.0-, and 3.0-point player respectively).

A random selection of six trials, including some from each player, were re-analysed by both the lead and an additional researcher to investigate the reliability of the approach. Bland-Altmann analysis showed no differences in the identification of contact and release frames (no systematic bias following the completion of a one-sample t-test showed difference (p=0.93), and identification of 95% confidence limit of ±4 frames – equating to 0.03 seconds) between the custom MATLAB code approach and a previously used video analysis method with Kinovea (Version 0.8.15, kinovea.org) [3]. The technical error of measurement (TEM) was also investigated and displayed good to moderate reliability [18] for the MATLAB approach (absolute TEM: 1.9% intra-rater, 5.9% inter-rater) for frame identification, and the calculation of angles (absolute TEM: 4.6% intra-rater, 4.5% inter-rater).

5.5 Results

Contact and release angles for each player and the first three strokes are presented in Table 5-2, with the variations supporting the expectation of varying propulsion approaches amongst these players. Asymmetries are evident, particularly for Player 2, where contact occurs further from TDC for the left hand, but the right-hand releases further around the wheel. Further asymmetries may be evident for the third stroke contact for Player 1, although there were no clear trends for Player 3.

Table 5-1: Player classification, impairment, international experience, and key wheelchair configuration parameters.

Player	Classification	Impairment	International Experience (years)	Seat Height (mm)	Seat Depth (mm)	Seat Angle (°)	Wheel Diameter (Inches)
1	0.5	Impaired muscle power (SCI)	4	260	160	34	24
2	2.0	Impaired muscle power (SCI)	10	280	155	28	25
3	3.0	Impaired muscle power (SCI)	4	310	140	24	26

Table 5-2: Contact and release angles for all players and the left and right hand. Negative angles denote the hand position is before TDC of the wheel, while positive is after TDC of the wheel.

	ContAng1 (°)		ContAng2 (°)		ContAng3 (°)		RelAng1 (°)		RelAng2 (°)		RelAng3 (°)	
Player	**Left**	**Right**	**Left**	**Right**	**Left**	**Right**	**Left**	**Right**	**Left**	**Right**	**Left**	**Right**
1	-69.4	-64.4	-69.5	-68.5	-63.6	-47.7	58.7	49.7	67.3	63.1	71.5	74.6
	±21.3	±12.1	±2.6	±7.9	±6.2	±8.6	±5.0	±4.9	±3.7	±7.5	±2.8	±7.3
2	-45.9	-48.0	-73.3	-59.5	-72.2	-48.9	54.1	57.4	69.5	75.9	68.8	81.6
	±16.1	±4.4	±2.8	±6.2	±5.3	±4.6	±6.0	±3.6	±6.4	±10.3	±6.8	±8.4
3	-11.8	-10.8	-11.2	-10	-8.5	-7.3	74.3	70.7	81.2	84	83.5	89.7
	±2.6	±3.1	±1.5	±3.2	±1.6	±3.8	±5.8	±2.9	±4.7	±2.8	±2.1	±1.5

Intra-stroke acceleration profiles for the three players are presented in Figure 5.2. As expected, for players of different point classification clear differences are evident in the timing of peak accelerations during each stroke, as well as the magnitude of these peaks and the general shape of the acceleration traces. For all traces, the magnitudes of acceleration for the first stroke are reduced compared to strokes two and three, which show a trough before a large acceleration peak, then slight decreases in acceleration prior to release.

The positions of the left and right hands at identified peaks and troughs for all players are presented in Figure 5.3. Player 3 displayed the smallest asymmetries (all mean differences less than 7.5°), while Player 1 displayed the greatest asymmetries (smallest mean difference of 11.2°). Despite showing greater asymmetries at contact and release, Player 2 displayed relatively symmetrical hand locations at peaks (differences of 10.3°, 9.3°, and 2.2°) and troughs (differences of 8.2°, 5.5°, and 6.0°). Due to the increased trunk, arm, and hand function compared with other players, the high-point player was able to lean further forward, hence the greater angle at the trough and for peak acceleration.

Each specific stroke was also normalised to 0–100% from contact to release to allow for timing comparisons between strokes (Table 5-3). The high-point players peak accelerations occurred later in each stroke compared with the two lower-point players (relative to the normalised stroke cycle time). However, the trough occurred before the peak for the 2.0- and 3.0-point players in strokes two and three, whereas the 0.5- point player displayed the trough following the peak.

Table 5-3: The timing (mean±SD) of peaks and troughs for each of the first three strokes, as a percentage of the specific stroke time (e.g., the 0.5-point player's peak acceleration for stroke one occurred at 82±6% of the first stroke length).

	Stroke 1 (%)		Stroke 2 (%)		Stroke 3 (%)	
	Peak	Trough	Peak	Trough	Peak	Trough
1	82 ±6	62 ±28	23 ±6	41 ±10	31 ±5	48 ±7
2	69 ±17	73 ±25	54 ±5	37 ±8	67 ±6	43 ±8
3	42 ±15	59 ±6	68 ±3	31 ±2	86 ±6	32 ±11

5.6 Discussion

The intra-stroke acceleration profiles in wheelchair court sports have not been previously investigated, despite the potential performance improvements through wheelchair design or technique analysis. As previously discussed, large variations in player impairments (and therefore level of physical activity limitation) in WCR results in significant differences in propulsion kinematics [3], and has likely been a barrier in exploring individual analyses in research. This study assessed the potential for combining high speed video analysis and IMUs to assess and interpret intra-stroke acceleration profiles for three elite WCR players. The current study successfully demonstrated that this method can be used to identify key events within specific strokes. Due to the small mass of the IMUs (12 grams), the ease of securing to- and transferring between wheelchairs, and ability to reduce the processing time compared with previous video analysis (able to quickly identify contact and release timings), they can increase the amount of detailed information provided to coaches [11, 12].

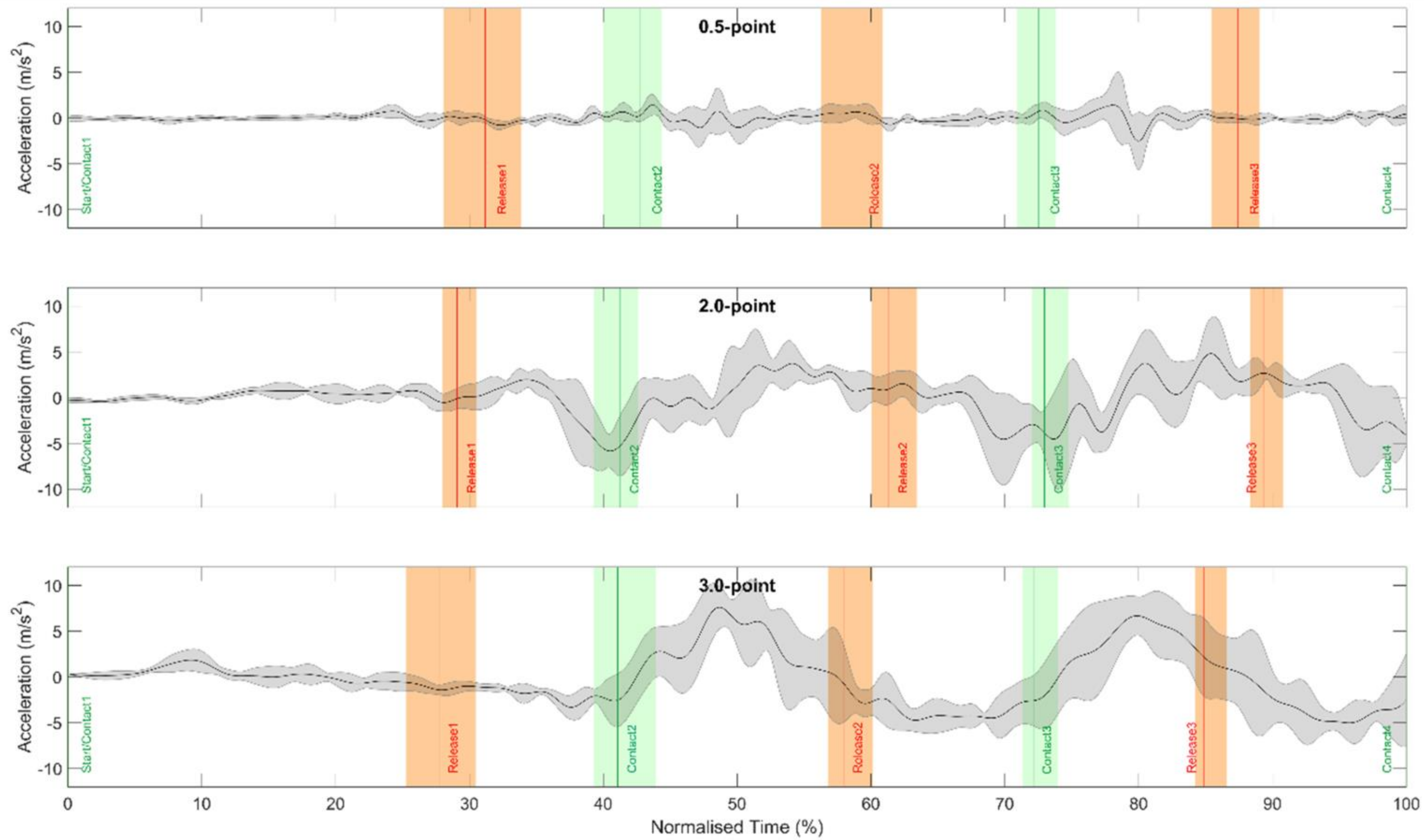

Figure 5.2: Mean intra-stroke acceleration profiles against normalised time for each participant, where the black line represents the average AccFrame in the direction of propulsion and shading represents ± one standard deviation. Contact and release points are shown by the green and red zones, with the shaded region representing the range across the six trials. The regions between a contact and release is the stroke time, whilst the region between release

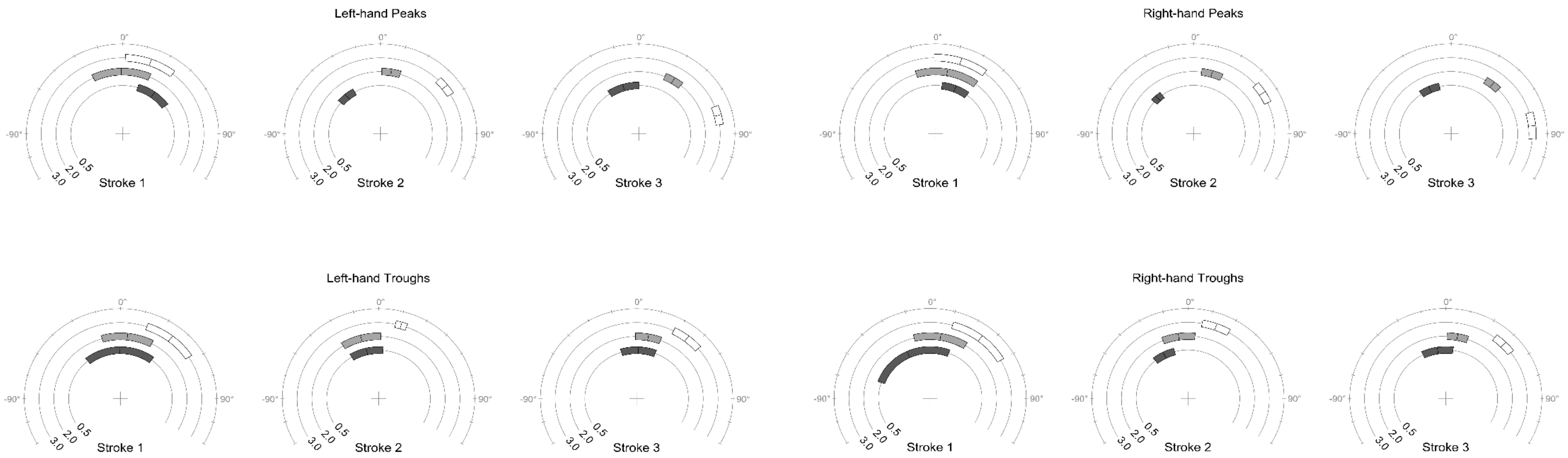

Figure 5.3: The left and right-hand locations on the wheel (mean±SD, indicated by the black line and the surrounding box respectively) at the peak (top) and trough (bottom) timings for each of the three players investigated. Left-hand positions are on the left, and propulsion direction is to the

As demonstrated by Figure 5.2, distinct differences can be identified between the averaged acceleration profiles of the first three stroke cycles for each player, as hypothesised based on the varying propulsion approaches (Table 5-2). This supports the use of IMUs to identify differences across propulsion approaches in WCR. As classification scores increase, there are clear increases in the magnitudes of acceleration, as expected. The greater magnitudes of acceleration result in players reaching higher velocities earlier, and therefore improves their ability to escape or execute blocks during gameplay. However, the general shape of the acceleration profile also differs for the average of the first three strokes, supporting the findings by Moss et al. [8] for wheelchair racing starts. For the 0.5- and 2.0-point players, there are clear troughs or dips in acceleration during the stroke phase; conversely, the 3.0-point player primarily displays a single peak for the third stroke, and a slight trough during the second stroke (see Figure 5.2). This further separates the propulsion techniques across these players, with the peak and trough locations and times for the low-point player suggesting a preference for greater proportions of pull than the other players, as previously reported [3].

The difference in propulsion approaches likely reflects the impairments of the individuals. The 0.5-point player has limited to no triceps brachii or trunk function, and therefore aims to maximise the use of their biceps brachii function during the pull motion [15]. The trough then occurs closer to TDC of the wheel, where the player is closer to reaching their minimum elbow angle and transitioning into the push phase (i.e., from biceps brachii to triceps brachii contributions [13]). This propulsion approach likely benefits from a wheelchair set-up with greater seat depth, with low-point players reporting it allows them to position their hands further back on the wheel to maximise the pull phase [15]. While troughs are always likely to be present, reducing their effect (as well as negative accelerations during the recovery phase) is important [8]. Identifying the timing and location of troughs, through the implementation of intra-stroke profiling, provides opportunities to supplement strength sessions with movements that promote more efficient transitions between pull and push, and therefore a reduction in the amount of deceleration during the stroke. This information would also provide coaches with guidelines for maximising propulsion skills; for example, moving the contact angle closer to or forward of TDC if the peak acceleration only occurs after the trough, such as for Player 3.

Left and right-hand variations were evident for the contact and release angles, particularly for the 2.0-point player (Table 5-2). Asymmetries can potentially be detrimental to performance as any discrepancies (potentially due to physical activity limitations cause by impairments) between wheel rotations require a greater focus on steering [19]. In this case, larger contact angles occurred with the left hand, but larger release angles for the right hand. It is unclear whether this is a result of steering considerations within the stroke or is representative of the players usual propulsion approach. Hand locations at peak and trough locations were also monitored, with the greatest differences occurring for Player 1 (Figure 5.3). This result supported the hypothesis, where greater asymmetry was expected due to the greater activity limitation, and hence greater likelihood of asymmetrical arm and hand function of the athlete. Despite the contact and release variations for Player 2, there was a reduction in asymmetry at the peak and trough locations. These results contradict findings from Goosey-Tolfrey et al. [14], where high-point WCR players showed greater asymmetries in distance, speed, and power during sprint. This may occur due to different classification groupings or types of impairment (high-point group was ≥2.0-point players) or testing protocols as testing was completed on an ergometer over 28 metres. Tracking of each hand’s location throughout the entire stroke, as well as three-dimensional analysis of factors such as elbow and shoulder locations and angles, is required in future research to more accurately assess any presence or implications of asymmetry. This includes monitoring of wheelchair motion and any slightly changes in direction potentially related to asymmetric propulsion.

By considering individual propulsion acceleration profiles, coaches and analysts can assess the stronger and weaker regions within a player's stroke. Individual profiles, therefore, have the potential to guide physical and technical changes to propulsion strokes for wheelchair athletes. Conversely, it can also be beneficial when considering the impact of wheelchair set-ups, where the pull and push phases are influenced by design parameters such as the seat depth [15]. When assessing or trialling wheelchair set-ups, the effect parameter changes have on the intra-stroke acceleration and factors such as timing and magnitude of peak acceleration can be monitored for a greater understanding of the influence on performance. These considerations need the input of experienced coaches and inter-disciplinary teams, as an individual's capabilities are often restricted by their impairment and subsequent activity limitation [4], limiting the range of potential wheelchair set-ups.

While this study provides the first insights into intra-stroke acceleration profiles in WCR, the case-study approach of three players does not represent the wider variety of propulsion approaches used by players in WCR. However, the primary aim to assess the use of IMUs in performing intra-stroke acceleration profiling has been achieved, using players from across the range of classification scores (excluding 3.5-point players). Therefore, it is posited that the current method provides the ability to develop individual profiles for the majority of WCR players. Due to the range of impairments in WCR [4], it is recommended that detailed studies into propulsion focus on case-study approaches as introduced here. While this has limitations in terms of statistical assessment, grouping of players based on classification scores can potentially hide crucial information about subsets or individual players [3]. Players initiated the sprint from a stationary position in their own time. Whilst this is part of their regular training and sprinting from a stationary position is common in WCR [11], future work should also consider kinematics under more reactive situations representative of match play. As testing was completed on-court and in the player's own wheelchair, kinetic data was not recorded. Although this information would be beneficial in the analysis of intra-stroke profiles, current methods of obtaining this information would likely alter the propulsion approach and hence reduce the validity of the results. The current trade-off between highly detailed analysis and representative test designs with practical applications remains an area for development in wheelchair sport. The method implemented allows for simple instrumentation across wheelchairs that allows increased analysis whilst maintaining a relatively representative on-court, as well as improving current processing times.

Individual intra-stroke accelerations provide the potential for detailed assessments of propulsion technique. This information can be used by coaches or analysts to assess regions of strength and weakness within a stroke, as well as implications of wheelchair set-up. The use of IMUs increases the ability to monitor these performance measures with minimal interference and therefore promote on-court testing, resulting in increased validity of outcomes. When IMUs are used in conjunction with high speed video analysis, they able to identify clear variations in intra-stroke accelerations, such as timings and hand location on the wheel for key events in the trace. Intra-stroke profiling for individual WCR players has the potential to provide increased information to coaches about an individual's stroke, as well as monitor changes due to skill development or wheelchair set-up alterations.

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Chapter 6: Test Design and Individual Analysis in Wheelchair Rugby

This chapter investigates the importance and consideration of test design for simple tasks, with implications presented for a 5m sprint in WCR. This includes a focus on sprint time, propulsion kinematics, and peak accelerations.

This chapter has previously been published (see below details) and has been reformatted for the purpose of this thesis. This publication satisfies University of Adelaide requirements for inclusion in a Thesis by Publication.

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Contribution to the Paper	Involved in test design, then led testing procedure, video and accelerometer analysis, interpretation of results, and preparation of manuscript.		
Overall percentage (%)	50		
Certification:	This paper reports on original research I conducted during the period of my Higher Degree by Research candidature and is not subject to any obligations or contractual agreements with a third party that would constrain its inclusion in this thesis. I am the primary author of this paper.		
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By signing the Statement of Authorship, each author certifies that:

i. the candidate's stated contribution to the publication is accurate (as detailed above);
ii. permission is granted for the candidate in include the publication in the thesis; and
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6.2 Abstract

Objectives: Use a task vehicle of sprint testing in wheelchair rugby (WCR) to explore the impact of small changes to test design using both group and individual analysis. **Design:** Exploratory, repeated measures, on-court study. **Method:** 25 national or international level wheelchair rugby players completed 5× 5m sprints under two conditions: (i) an acceleration from standstill in their own time, and (ii) an 'active' start, simulating a key aspect of performance. Video analysis and accelerometer data were used to measure key kinematic and performance variables with a focus on the first three strokes. Each player was grouped into a high-, mid-, or low-point group based on their sport-specific classification score. Group (paired sample t-tests) and individual (meaningful differences, performance coefficients, and Cohen's *d* effect sizes) analysis assessed differences between the two conditions. **Results:** The low-point classification group performed significantly slower in the active start ($p<0.05$). There were no differences in sprint time for the high- and mid-point groups. Mid-point players achieved greater peak accelerations for strokes two and three in the active start ($p<0.05$). Individual sprint performances varied substantially, ranging from 8% decrease to 14% increase in sprint time for the active start. Meaningful differences in peak accelerations were demonstrated for 23 out of the 25 players. **Conclusions:** Small amendments to test design can lead to significant differences in individual athlete performance. Traditional group analyses masked important individual responses to testing conditions. There is need to further consider representative test design, and individual analysis for monitoring physical and skill performance.

Key words:

Representative design; Paralympic sport; wheelchair propulsion; impairment; skill

6.3 Introduction

The use of skill testing in sports has been a topic of recent interest, particularly in talent identification and development [1-5]. Much of the literature, however, has focussed on isolated physical performance aspects, often neglecting important components critical to skill performance in competition contexts. Research is required to explore and demonstrate the impact of small changes in test design to provide options to complement existing physical performance tests. Para-sport contexts provide unique opportunities to demonstrate the potential impact of test design on individual performance responses, with the range of physical activity limitation caused by athlete impairments akin to wide range of skill levels across sport development pathways.

Wheelchair court sports, including wheelchair rugby (WCR), basketball and tennis, are all characterised by repeated intermittent, high intensity activity [6-8]. Critical for successful performance in such sports is the ability to apply force to the wheel (through a combination of push-rim and wheel contact with the hand, where push-rims are more prominent for more impaired athletes), maximally accelerate the chair [9] and change direction quickly [8]; indeed surveys of WCR players and coaches have rated acceleration from a standstill and fast turning on the spot as the most crucial performance aspects in the sport [7]. To be eligible for WCR, players are required to have a physical impairment affecting at least three limbs (e.g. impaired range of motion, limb deficiencies) [10, 11], hence the level of sport specific physical activity limitation as a result of the impairment differs significantly between players. Players are 'classified' under a point system of 0.5-3.5 (with teams made up of 4 players of no more than 8 points on-court), and where a higher point score is indicative of a greater amount of strength, coordination and range of motion of the trunk, arm and hand [11]. Since the sports appearance at the Paralympic

Games in 2000, there has been a marked increase in research investigating the effect of impairment on physical performance variables. In both research and practice, players are often assessed in three broader groups, reflective of similar levels of physical activity limitation and on-court roles (e.g., low-point: 0.5-1.5; mid-point: 2.0-2.5; high-point: 3.0-3.5) [12]. Recent research in WCR has investigated the effect of abdominal binding [6] and changes to key wheelchair parameters such as seat angle on performance [9]. Additionally, research across multiple wheelchair sports has begun to consider and reflect the characteristics of on-court performance [13], moving away from steady state assessments to focus on maximal efforts and repeated accelerations from standstill [14, 15]. For example, it has been demonstrated that performance, propulsion strategies and upper limb kinematics of sprinting from a standstill in elite WCR players differ significantly across classification groupings [16]. There is, however, a reliance on highly controlled tests (e.g., linear 'self-paced' sprints, or pre-planned agility) [14-16], and experimental designs which may not capture performance that is representative of competition (e.g. use of ergometers [17]). Isolated performance tests are important, as they allow for a high degree of control and the reliable measurement of physical changes [1]. However, current test designs can often (overly) constrain athlete behaviour, resulting in kinematic and performance outcomes that are not reflective of competition contexts and thus provide little insight into execution of critical on-court skills (e.g., ability to perform a pass or turn followed by a sprint, representative of match play in WCR). The design of representative and efficient testing is not an easy or a trivial issue to solve [1, 5]. For example, recent research in soccer has questioned the validity of passing test designs in controlled environments [18]. Furthermore, traditional group analyses have been suggested to be masking important and meaningful individual differences in biomechanics and motor learning assessments [19]. These differences are likely to be magnified when assessing elite athletes with disabilities due to significant differences in the level of sport specific activity limitation caused by the athletes' impairments [16, 20]. Due to the restrictions in the peer-review process, where statistical power and experimental control often outweighs the value of individual assessments and representative testing, research in disability sport can have minimal practical application (cf. Churton et al. [21]; Paulson et al. [13]). There is a clear need to consider the impact of test design at an individual scale of analysis, and consider the feasibility for supporting the design of more representative tasks [1, 22]. The aim of this study, therefore, was to use a task vehicle of sprint testing in WCR to explore the impact of small changes to test design using both group and individual analysis approaches. Performances of elite WCR athletes were assessed through propulsion kinematics, acceleration and sprint time. It was expected that a small amendment to an existing test design (aimed at making the test more representative of on-court activity) would result in meaningful differences in key performance variables for athletes, highlighted through individual descriptions and supplementary statistics [23]. Furthermore, it was predicted that these changes would be largely masked in traditional group statistical analyses.

6.4 Method

Elite wheelchair rugby players (n=25; age: 30.5±7.0 years) were recruited at a national event, provided informed consent and participated in the study. The majority of players had substantial international experience (n=20, 7.7±6.5 years), with the remaining players having a minimum of two years of experience competing at a national level in Australia (n=5, 3.6±1.9 years). Accordingly, all had a national and/or international WCR classification as confirmed by certified classification panels [24]. Participant information is provided in Table 6-1. Low-point (those with impairments that cause a greater sport specific activity limitation) and mid-point players all had spinal cord injuries, while high-point players had spinal cord injuries (all 3.0–point players) and limb deficiencies (all 3.5–point players). Ethics committee approval was obtained for human investigation through the required institution (H-2015-127), and approved testing protocols and guidelines were followed by the investigators.

Players completed five 5m sprints in their own wheelchair rugby chair, in each of two conditions: (i) an acceleration from standstill in their own time (a typical sprint testing protocol [14, 15]), and (ii) an 'active' acceleration, to simulate a key aspect of competitive performance. The active start was designed to be a feasible amendment to an existing test, based on analysis of elite WCR competition. The test aimed at simulating a common skill that incorporated both a turn and sprint as identified as critical for performance [7, 8]. For this, players faced away from the intended sprint direction, received an 'inbound' (a pass to restart the game following a goal from the defending team's own goal line) and were required to perform a 'give-and-go' requiring a quick two handed chest pass back to a teammate, before turning 180 degrees on the spot and completing a sprint. In performance contexts, this would be to create a block or screen (often low-point players) for the ball-carrier, or to find space as a passing option (more often high-point players). For the active start, the 'give-and-go' pass was completed by the same person each time, passed from a short distance to ensure a consistent receiving position for the participant, and had to be a successful return pass to count as a valid trial. Players were told that the total time from receiving the pass until completion of the sprint was also being recorded and were instructed to complete all tests as quickly as possible. No additional instructions were provided (i.e. turning technique or direction was not prescribed). Players were given two familiarisation trials for the active start and were all experienced with the standstill start as part of their regular performance testing. Players completed the tests in a counterbalanced design following a standardized warm-up prior to competition and given sufficient break between trials to ensure fatigue was not a contributing factor.

Table 6-1: Participant (player) information and sprint performance (averaged across five trials ±SD). I = international, N=national. P-C is performance coefficient, P=D is performance difference, and M-D is meaningful difference. Effect Size presented is Cohen's d.

Player	Point Score	Experienc e (years)	Standstill 5m sprint (s)	Active 5m sprint (s)	P-C	P-D (s)	M-D (s)	Effect Size
1	3.5	13 (I)	1.79 (0.07)	1.75 (0.09)	1.02	-0.04	0.12	0.51
2	3.5	5 (I)	1.53 (0.03)	1.58 (0.04)	0.97	+0.05	0.05	1.37
3	3.5	3 (I)	1.91 (0.05)	1.78 (0.08)	1.08	-0.13	0.09*	2.04
4	3.5	1 (I)	1.60 (0.03)	1.62 (0.07)	0.98	+0.02	0.05	0.48
5	3.0	3 (I)	1.78 (0.02)	1.87 (0.06)	0.95	+0.09	0.04*	1.96
6	3.0	7 (I)	2.04 (0.02)	2.02 (0.06)	1.01	-0.02	0.04	0.47
7	3.0	10 (I)	2.17 (0.07)	2.35 (0.07)	0.92	+0.18	0.12*	2.60
8	2.5	19 (I)	2.06 (0.02)	2.15 (0.08)	0.96	+0.09	0.04*	1.54
9	2.5	10 (I)	1.96 (0.01)	1.96 (0.08)	1.00	0.00	0.02	0.08
10	2.0	11 (I)	1.97 (0.03)	2.09 (0.07)	0.94	+0.12	0.05*	2.34
11	2.0	3 (I)	2.08 (0.10)	2.14 (0.03)	0.97	+0.06	0.18	0.80
12	2.0	10 (I)	2.05 (0.01)	2.38 (0.15)	0.86	+0.33	0.02*	3.20
13	2.0	12 (I)	2.09 (0.04)	1.95 (0.03)	1.07	-0.14	0.07*	4.07
14	2.0	20 (I)	2.38 (0.08)	2.43 (0.06)	0.97	+0.05	0.14	1.09
15	2.0	7 (N)	2.05 (0.03)	2.26 (0.05)	0.91	+0.21	0.05*	4.94
16	2.0	20 (I)	1.95 (0.02)	1.95 (0.04)	1.00	0.00	0.04	0.22
17	1.5	3 (N)	2.32 (0.02)	2.33 (0.03)	1.00	+0.01	0.04	0.26
18	1.0	12 (I)	2.41 (0.03)	2.53 (0.08)	0.96	+0.12	0.05*	1.97
19	1.0	2 (N)	2.32 (0.12)	2.38 (0.05)	0.97	+0.06	0.21	0.65
20	0.5	5 (I)	2.74 (0.05)	2.91 (0.17)	0.94	+0.17	0.09*	1.42
21	0.5	16 (I)	2.84 (0.03)	2.95 (0.02)	0.96	+0.11	0.05*	3.88
22	0.5	5 (I)	2.86 (0.08)	3.00 (0.08)	0.96	+0.14	0.14	1.62
23	0.5	23 (I)	2.54 (0.06)	2.68 (0.08)	0.95	+0.14	0.11*	1.96
24	0.5	3 (N)	2.66 (0.01)	2.87 (0.08)	0.93	+0.21	0.02*	3.80
25	0.5	3 (N)	2.55 (0.04)	2.76 (0.04)	0.92	+0.21	0.07*	5.04

*Indicates meaningful difference between test design sprint times for the individual.

Sprint performance (time to 5m) was assessed using a laser timing system (Kinematic Measurement System, Fitness Technology, Australia). Overall acceleration of the wheelchair was monitored using a tri-axial accelerometer secured to the frame near the footplate (x8m-3mini, Gulf Data Concepts, USA, 100Hz). Peak accelerations in the sprint direction were determined for each of the first three strokes following application of a low-pass filter (20Hz, 2nd order, Butterworth, bidirectional) [25]. Fixed cameras (Sony HDR-PJ 430, 100Hz) captured the start of each sprint from both rear and side views, in line and perpendicular to the player's plane of motion, respectively. Kinematic analysis was restricted to the first three strokes and hand contact points and timings appropriate for linear sprint testing (i.e. post-turn for the active start) [16]. Kinematic variables were calculated using Kinovea (Version 0.8.15). Side and rear cameras were event synchronised to support identification of hand contact with the wheel, identified using first and last contact points, respectively [6]. Reflective markers were positioned on the players gloves approximately at the second metacarpophalangeal joint [26] to allow for the consistent estimation of hand contact and release. Hand contact (ContAng) and release angles (RelAng) were then assigned relative to the top dead centre (TDC - 0º) of the wheel. Stroke angle (StrokeAng) was calculated as the angle between ContAng and RelAng. Intra- and inter-rater reliability was completed on a random selection of trials for both active and standstill starts. Technical error of measurement (TEM) was calculated and deemed to be acceptable (5.6% and 5.3% for intra- and inter-rater, respectively).

For group analysis, players were grouped by classification score to align with previous research [12, 16]. The average performance (sprint) time and the average peak acceleration for each of the first three strokes were calculated for the standstill and active starts. Starting condition (standstill versus active start) was then compared using paired-sample t-tests for each of the classification groups. Propulsion kinematics (ContAng, RelAng, StrokeAng) were also compared between testing protocols using paired-sample t-tests, however this was only performed for the second and third strokes. The first stroke was not analysed as the use of 2D video analysis could potentially result in perspective issues due to the initial out-of-plane motion in the active start. As we have demonstrated in previous work the significant differences in kinematic variables for low, mid and high point players [16], no between group analyses were completed. Within subject Cohen's d effect sizes were calculated for each of these group comparisons, as well as each individual for sprint time. Performance coefficients, standstill start time divided by active start time, were also calculated for each player's sprint performance to allow for a comparison of test designs (a score above 1.0 indicates improved performance). All statistical analyses were completed using SPSS Statistics (v.21, IBM), with alpha level set at 0.05 with a Bonferroni correction. Plots allowing for individual comparisons of (i) peak acceleration, and (ii) StrokeAng for each of the first three strokes were produced using MATLAB (R2016a, Mathworks). Individual meaningful differences [27] were calculated for sprint times, peak accelerations (both 2× typical error as recommended) and propulsion variables (1.5× typical error due to influence of human error during measurements) to identify differences between standstill and active starting protocols. The presentation of individual results with supplementary statistics supports the recommendation of Dankel et al. [23] for exercise science research. This approach allows for greater reader interpretation for factors such as effect sizes, which are highly influenced by standard deviations, as well as simpler interpretation of magnitude changes between testing protocols.

6.5 Results

Sprint performances for the low-point group were significantly slower in the active compared with the standstill start (difference of –0.13 seconds, $p<0.001$, $d=2.64$). No significant differences in sprint performance between tests were noted for high-point groups (difference of –0.02 seconds, $p>0.05$, $d=0.53$) and mid-point groups (difference of -0.08 seconds, $p>0.05$, $d=2.18$). Individual sprint performances are provided in Table 6-1, along with performance coefficients and individual meaningful differences. Individual performance coefficients showed improvements of up to 8% during the active start (e.g., sprint performance improved in the active start for Player 3), while also detrimental effects of up to 14% (e.g., sprint performance worsened in active start for Player 12). Two players displayed a meaningful improvement to their sprint time during the active protocol, while twelve had a slower sprint time.

Peak accelerations in the active start were significantly higher than in the standstill start for the mid-point players' second stroke ($5.56 m/s^2$ compared with $4.81 m/s^2$, $p<0.001$, $d=0.50$) and third stroke ($5.82 m/s^2$ compared with $5.23 m/s^2$, $p=0.029$, $d=0.35$). No differences were evident for the low- and high-point groups for any of the first three strokes. Individual differences in peak acceleration across the three strokes are provided in Figure 6.1. Individual meaningful differences were identified for 23 of 25 players for at least one stroke, although only one player displayed a consistent trend across all strokes (Player 23 – decreased peak acceleration across all strokes for active start). Across all players and strokes, 24 meaningful increases to peak acceleration were identified, as well as 10 meaningful decreases.

Kinematic variations were also investigated for the second and third strokes. Significant differences were evident for low– and mid–point groups. The mid-point group had a ContAng further from TDC for the second stroke of the standstill start (-4.15±7.73°, $p=0.007$, $d=0.14$), while the low-point group displayed a larger StrokeAng for the third stroke (-7.11±19.1°, $p=0.025$, $d=0.36$). Individual exemplar plots for a high, mid-, and low-point player are provided in Figure 6.2. ContAng, RelAng, and StrokeAng show small, consistent changes to the propulsion approaches between test designs (small standard deviations – all less than 6° for all strokes and angles for these individuals).

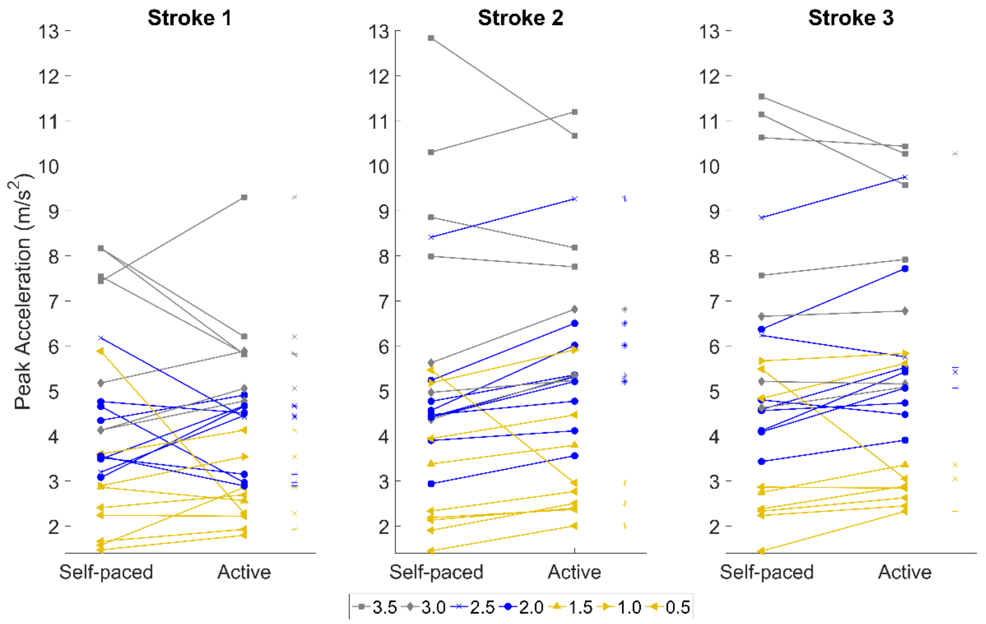

Figure 6.1: Peak acceleration (m/s²) during the first three strokes of standstill and active performance tests, where 0.5–1.5 points is considered the low-point group, 2.0–2.5 the mid-point, and 3.0–3.5 the high-point group.

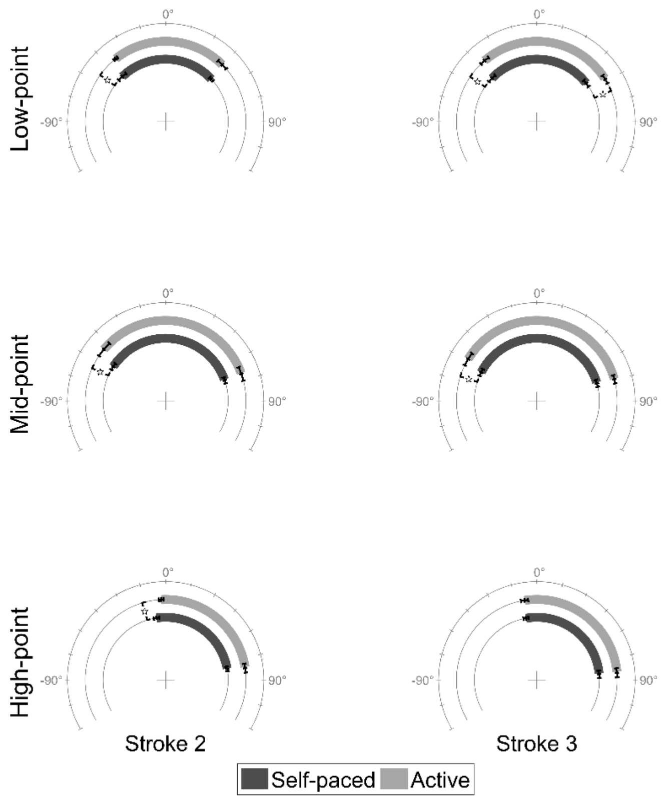

Figure 6.2: Exemplar kinematic data for three players. Average ContAng, RelAng, and StrokeAng for active and standstill task designs are shown for a low-, mid-, and high-point player across strokes two and three of the linear sprint. 'Bars' represent standard deviation (SD) of five repeated trials in each condition, and 'stars' represent the presence of individual meaningful differences.

6.6 Discussion

The aim of the current study was to demonstrate the impact of test design and analysis methods, using a task vehicle of sprint performance in WCR. Results demonstrate that small amendments to current test designs, in this case the inclusion of a pass and turn on the spot prior to a typical sprint test, can result in significant changes in kinematic and performance variables. For example, overall sprint performance for the low-point group reduced significantly when they were required to perform an active start simulating a common movement from a performance context (a finding that was not replicated for high- and mid-point groups). These results suggest that test design may have a clearer and more consistent impact on performance of athletes with greater levels of physical activity limitation, and further considerations for modifying tests may be required across heterogeneous population. While group analysis revealed no differences in sprint performance for mid-point athletes, peak accelerations in the active start were significantly higher than in the standstill start for the second and third strokes (see Figure 6.1). This difference (+0.45m/s^2 averaged across the first three strokes for all mid-point players) corresponds to traveling an extra 0.49m in the first moments following a turn, a change in performance that could be the difference in making or escaping a block. Similarly, individual players demonstrated various responses to test designs (see Table 6-1), with mid-point athletes ranging from a performance difference of +7% to –14%. Differences in peak acceleration suggest that while overall performance times were relatively stable in group analyses, how individual players achieved and contributed to those outcomes was highly variable between tests (14 players displayed a meaningful change in sprint time across protocols, and 23 of 25 showed a meaningful change in peak acceleration for at least one of the first three strokes). For example, across the high-point group for one stroke, acceleration changes in the active start varied from –5% to +41%. Given the role of these players as offensive ball carriers, this ability to maximally accelerate the chair and avoid defenders is of further interest for individual analysis and future research, and test designs focussed more specifically on chair skills would be of value. Equally, an experienced low-point player (#23) displayed large increases (80-158%) in peak acceleration across all strokes during the standstill start. This finding is suggested to be as a result of the development of a propulsion approach that maximises their performance when accelerating from standstill in testing situations, with this athlete's results in the active start comparable with other low point players. Both testing design and analysis methods clearly shape and constrain performance results in this elite sample of WCR players.

Controlling test design is important to assess changes in development and physical preparation of athletes. However, there is the potential of missing valuable information pertaining to execution of skills through the use of more representative tasks (see Robertson et al. [1] for an extended review and discussion). Both field and laboratory-based testing approaches should strive to improve the representativeness of tests to (i) increase translation of findings to on-court performance; (ii) decrease total testing required; and (iii) aid in monitoring the development of technical skills. While specific research questions may require more robust methods (i.e., laboratory testing), further work is needed to consider the concept of representative design in these conditions. For example, continual refinement of treadmill and roller design and protocols to better represent overground propulsion [28]. It is important to note that the current 'active' start is not necessarily promoted as a fully 'representative' task, but rather an insight into how small (and often overlooked) aspects of performance can lead to new insights, and may provide complimentary tests or ideas for solving current quandaries across a wider range of sports [1]. Indeed, it has been noted that a clear definition and examples of enhancing representative task designs in this area is needed [1] (also see Pinder et al. [5], Pinder et al. [20]), and researchers and practitioners should work together to develop skill tests which maintain key measurement properties. Some of these amendments may be simple and feasible additions to existing tests. For example, it is common for physical tests in wheelchair tennis and wheelchair basketball to be completed without a racket and ball, respectively. Inclusion of these performance objects, or comparison of tests with and without them could allow practitioners to support coaches' knowledge on individual areas for development (i.e., is there a need for physical and/ or technical development for this athlete?). However, until we can reach suitable control in more representative tasks, we should continue to include controlled tests as a measure of athlete physical development. Alternatively, the role of performance [29] or 'discrepancy' [30] profiling may help to provide further insights into gaps between coaches' current ratings of skilled performances and relative importance of those factors across varied performance contexts (e.g., different impairment groups, playing positions, development levels). Researchers and practitioners should then work closely with coaches to continue to consider ways to enhance test designs to promote tests that are highly specific to the goal and focussed on skill outcomes, including the consideration of specific impairment types and their effect on performance.

6.7 Conclusion

Small amendments to physical performance tests can lead to significant differences in individual athlete performances, demonstrated through this example in WCR. Future research should focus on enhancing the representative design of performance tests, or exploring feasible methods to complement current tests to ensure practitioners can continue to maintain suitable measurement properties.

6.8 Practical implications

- Findings demonstrate the importance of careful test design for capturing representative performance data for research, classification, and performance enhancement in wheelchair sports.
- Completion of both group and individual analysis is recommended, particularly for assessment of athletes with disabilities due to differences in sport specific activity limitation caused by impairments.
- Physical performance tests could be further complemented with amendments to test designs to allow for additional insights into skill performance and simulation of performance contexts.

Acknowledgements

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Chapter 7: Wheelchair Rugby Chair Configurations: An individual, Robust Design Approach

Bringing together knowledge developed through Chapters 3-6, this chapter reports on a method to adjust WCR chair configuration parameters in a timely manner whilst maintaining the ability to identify specific parameter effects. Effects are assessed on sprint, agility, and skill performance with a single case-study presented in detail. Results for remaining athletes are presented, although discussion and implications are not reported in large detail.

This chapter has been submitted as a journal article (see below details) and has been reformatted for the purpose of this thesis. This submission *does not* satisfy University of Adelaide requirements for inclusion in a Thesis by Publication.

> Haydon, D.S., Pinder, R.A., Grimshaw, P.N., Robertson, W.S.P., 2018. Rugby Wheelchair Chair Configurations: An Individualised, Robust Design Approach. Submitted to Sports Biomechanics, August 2018.

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Overall percentage (%)	55		
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i. the candidate's stated contribution to the publication is accurate (as detailed above);

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7.2 Abstract

Prescription of wheelchair rugby chairs is difficult due to the range of impairment types and severities in the sport, difficulty in adjusting wheelchair settings, and the assessment of on-court performance measures. Currently, players rely on their personal experiences and those of surrounding coaches to select an appropriate set-up. Technological advancements, such as with inertial measurement units and processing algorithms, as well as more representative, testing approaches, has improved the potential for achieving near optimal set-ups at an individual level. An orthogonal design approach was implemented using an adjustable wheelchair to investigate the effect of seat height, seat depth, seat angle, and tyre pressure on performance, mobility, and propulsion kinematics. Six elite wheelchair rugby players completed testing in nine individually tailored wheelchair set-ups while monitoring both quantitative and qualitative measures of performance. From this testing, a theoretical optimal set-up was compared with the current set-up for each individual. Three of six players reported a blind preference for the theoretical set-up, whilst others displayed similar performance. A single case-study approach shows how the assessment method can identify parameter settings that can potentially improve performance. This approach has the ability to improve upon the current prescription process for rugby wheelchairs.

Keywords

Paralympic sport; orthogonal design; wheelchair mobility; propulsion kinematics; wheelchair configuration

7.3 Introduction

Prescription of wheelchairs for court sports, such as wheelchair rugby (WCR) and wheelchair basketball (WCB), is a difficult process that predominantly relies on the expertise of coaches and players who have substantial experience in the sport [1]. Due to the variability in impairment types and severity in WCR, together with individualised optimal set-ups related to the on-court role [1], players can be involved in the sport for a decade before achieving a near optimal set-up. Substantial anecdotal knowledge regarding performance and configurations exists among coaches [2] and it is crucial to incorporate this into the process of wheelchair prescription; however further work is required to supplement this with a method that allows for quantitative assessments.

In WCR, impairment types include limb deficiencies and impaired muscle power, with each individual assigned a classification score based on the sport specific activity limitation caused by their impairment [3]. Each team is then restricted to 8 points on-court at a time. Individual classification scores range from 0.5– to 3.5–points, where a lower score indicates greater activity limitations. These are based on a range of assessments for strength, range of motion, and co-ordination of the trunk, arm, and hand [4]. Therefore, a specific classification score can be assigned to individuals with substantially different impairments.

Previous work has identified that various chair configuration parameters affect performance in WCR and WCB, with these including seat height, seat depth, and seat angle [5], wheel camber angle [6] and wheel diameter [7]. The selection of each parameter setting requires consideration of the trade-off effects; for example, lowering seat height can provide greater access to the pushrim/wheel, but potentially restricts ball handling skills which is a major component of successful performance, particularly for high-point players [8].

Alternative trade-offs may include: reduced seat depth (i.e., more anterior seat position) improving manoeuvrability at the cost of decreasing stability [9]; increased seat angle providing greater stability by limiting trunk motion and hence chair acceleration [5]; increased camber angles providing greater stability and manoeuvrability, while limiting linear speed [6]; and increased wheel diameters allowing for greater top end speeds [7], but potentially limiting acceleration [8]. Additionally, parameters such as tyre pressure are expected to influence performance despite a lack of associated research [10]. Tyre pressure influences the contact area between court and tyre, and hence both the amount of grip provided and rolling resistance encountered [10]. Players typically select their pressure based on 'feel' and personal preference, hence selection of optimal tyre pressure for individuals requires further research [10].

The assessment of these parameter effects has been investigated using a range of testing approaches. To maximise relevance and translation of research findings to practical outcomes, testing conditions should replicate players' on-court demands. In WCR, the ability to accelerate from a stationary position, execute quick and effective turns, and handle the ball are crucial to success [8, 11]. Recent testing approaches have focused on accelerations from standstill [10, 12, 13] and 'slalom' movements [6, 14], as well as the investigation of a larger testing regime consisting of various sprints and rotations to reflect on-court mobility in WCB [15]. These methods have improved upon previous testing that has been conducted on ergometers [5, 16], where propulsion approaches have been shown to be altered when compared with overground propulsion [17, 18].

Advancements in technology and instrumentation have also allowed for testing that is more representative of competition or on-court performance. Multiple approaches to improving tracking capabilities have been developed, namely a radio frequency-based indoor tracking system (ITS) [19], and the use of algorithms for inertial measurement units (IMUs) [20-22]. These algorithms utilise the acceleration and gyroscope components of IMUs secured to the wheels and frame of the wheelchair allowing estimates of the orientation and distance travelled. Higher measurement frequencies of IMUs (>100 Hz) compared with the ITS (up to 16 Hz) allows for calculation of key mobility measures. For WCB, a reduced number of mobility variables were found to accurately identify differences between classification groups [23]. These descriptors (average speed; maximum speed; 2m acceleration; average rotational speed in a curve; maximum rotational speed in a curve; rotational acceleration) align well with the reported performance factors in WCR [8, 24]. Furthermore, the use of IMUs also allows for the identification of hand contact and release timings on individual wheels and the assessment of individual intra-stroke acceleration profiles [12]. This allows for greater understanding of an individual athlete's propulsion approach, and the regions on the wheel where contact results in the greatest acceleration and performance. Due to these advancements in assessing on-court performance, there is the potential for improved monitoring of the effects of altering specific wheelchair parameters.

Whilst improved testing approaches provide a means to assess effects, due to the large number of parameters and the range of settings at which they can be adjusted, it has remained difficult to investigate a range of parameters in an efficient manner even for individual athletes. A user-centred, orthogonal design has been attempted previously [16], although testing involved only straight-line propulsion on an ergometer, no consideration of changes to propulsion kinematics, and no on-court translation. Orthogonal design allows for a range of specific parameters (i.e., seat height, seat depth, etc.) to be investigated at varying levels (i.e., increased, decreased, etc.) in a reduced number of trials using an array. In the array, each pair of columns is orthogonal, and each row represents the levels of the parameters for a specific set-up, the effects of each parameter level can be determined [25]. A detailed explanation and example of this approach is provided in Supplementary Material. In this study, this approach allows for the effects of specific wheelchair adjustments to be determined with a reduction on the amount of testing required for each athlete.

This study aims to assess a method for improving wheelchair prescription at an individual level in WCR. It is hypothesised that using representative on-court testing and IMUs in conjunction with an orthogonal design approach and feedback from athletes and coaches, optimal configurations can be developed at an individual level.

7.4 Materials and Methods

7.4.1 Participants

Six elite WCR players (all male) were recruited from the Australian Wheelchair Rugby team for testing based on their current wheel diameter and testing availability. Each player provided written, informed consent before completing testing. Individual participant information is provided in Table 7-1 (shown in the results).

7.4.2 Procedure

An adjustable rugby wheelchair was designed and manufactured to investigate the effect of a range of configuration factors on individual performance. The wheelchair was able to adjust a range of set-up parameters for the seat (height, depth, angle, dump – angle between base of seat and backrest) and footplate (vertical and horizontal position). The wheelchair was designed to suit 25-inch diameter wheels, with camber angle fixed at 16 degrees. As the wheelchair was not designed to withstand large impacts, the mass of the wheelchair was 14kg – a slight reduction in comparison to the participants typical sport wheelchairs [24]. Due to the individual customisation of wheelchairs, features such as ball carriers were not incorporated into the wheelchair design.

A robust design approach using an orthogonal array was implemented to substantially reduce the amount of individual testing required. Key wheelchair parameters of seat height (SH), seat depth (SDep), seat angle (SA), and tyre pressure (TP) were varied at three settings (see Figure 7.1): (i) the individual's current setting; (ii) a decrease and (iii) an increase to the parameter. Increments of ±15mm were used for SH and SDep, with SA and TP varied by ±5° and ±15psi, respectively – based on pilot testing and discussions with players and coaches. To accommodate the four parameters at three levels, an L9 orthogonal array (see Supplementary Material for more details) was selected. Chair set-ups were completed in a randomised order, and an overall set-up similar to the player's current set-up was incorporated as one of the nine tests without the player's knowledge. Players were also instructed to undergo a familiarisation period similar to their usual warm-up with each set-up before beginning testing. Throughout testing, all other configuration parameters were kept constant, with participants using their own wheels and strapping ensured to be consistent across all set-ups [13].

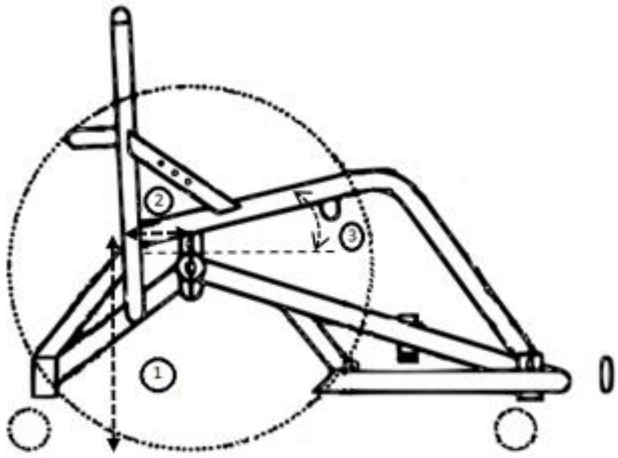


Figure 7.1: The parameters changed on the adjustable wheelchair included SH (1), SDep (2), and SA (3).

For each set-up, the player completed two 5m sprints from a stationary position, two Illinios agility tests, and a 'skill' test developed in conjunction with an experienced coach. The test was designed specifically to highlight how manipulations to chair parameters affected athletes' control of the chair while passing (or 'offloading') and receiving. While representative test designs have been shown to alter performance and propulsion [26], the combination of these test designs was preferred to their focus on specific components of performance. To allow for a suitable degree of control [26], no external players or coaches were involved in the test to remove this as a variable. For more information, see Supplementary Material. This combination of tests allowed for crucial performance factors such as acceleration from standstill, agility, and ball handling to be assessed [8]. Players were experienced with the 5m sprint and Illinois tests as part of their regular testing protocol, while they were instructed on and attempted multiple trials of the skill test in their own wheelchair prior to beginning testing. For each test, laser timing gates (SpeedLight, Swift Performance) monitored the performance time. Tests were short and distributed so as to ensure that fatigue was not a contributing factor. Following the completion of all tests for a single set-up, the player (and an experienced coach) provided feedback on their perceptions of the configuration to a researcher whilst the next set-up was prepared. This included self-reported ratings of key performance factors (without knowledge of performance times) such as acceleration, manoeuvrability, ball handling, top end speed and stability, as well as any specific comments they felt were of value (e.g., thoughts on how the set-up would translate to match performance). During this process, the player was unaware of the specific setting for each parameter.

7.4.3 Analysis

Three IMUs (500Hz, IMeasureU, NZ) were secured to the wheelchair throughout testing: one at the centre-front of the footplate, and one on each wheel near the axle to avoid interfering with the hand during the stroke phase. Data from the IMU located on the frame was low-pass filtered at 20Hz (Butterworth filter, order 5, bidirectional, -6dB cutoff frequency) and provided an overall assessment of the wheelchair motion. This focus included intra-stroke linear accelerations during the sprint [12], whilst also monitoring changes in orientation throughout agility and skill tests to allow for tracking assessments. During the 5m sprints, video (120Hz, GoPro Hero3+, GoPro, California, U.S.) from both side and rear views was recorded and synchronised with IMU data. The synchronisation was performed using a sharp strike to the front of the wheelchair frame, with this event clearly evident in both the video and acceleration trace of the IMU. The acceleration data was then used to select the region in which hand contact or release would have occurred for each of the first three strokes [12]. Using a custom MATLAB script (version R2016a), the selection of a point on the acceleration trace prompted the viewing of the corresponding side and rear video frames, as well as two frames before and after. A researcher experienced with analysis of wheelchair propulsion could then determine if any of these frames represented the moment of contact or release or re-select a point on the acceleration trace. If the desired frame was evident, contact (ContAng) and release angles (RelAng) were measured by selecting the centre of the wheel, top dead centre (TDC), and the hand location on the wheel [12, 27].

For the agility and skill tests, further mobility measures were selected based on the work of van der Slikke et al. [23]. This included the monitoring of average speed, average rotational velocity, peak rotational velocity, and average rotational acceleration. To ensure the accuracy of these measures, IMU tracking methods that incorporated previous work by van der Slikke et al. [20] and Shepherd et al. [21] were applied. This confirmed the accuracy of the IMU mobility measures, as well as allowing the separation of tests into specific sections (e.g., the 'weave' and sprint sections of the Illinois agility test – Figure 7.2).

Statistical assessment of changes was performed using an ANOVA with Bonferroni post-hoc testing. Significance was set at $p<.05$, with potential trends defined as $p<.10$. This allowed for the comparison of the three settings (decreased, current, and increased) for each parameter (SH, SDep, etc.). For each set-up, four types of variable groups were monitored: (i) performance times – sprint time, agility time, and skill time; (ii) mobility measures – peak magnitudes for linear sprint; and average speed, average rotational velocity, peak rotational velocity, and average rotational acceleration for agility and skill tests; (iii) contact and (iv) release angles for the first three strokes of the sprint. To ensure the even contribution of each of variable in each group, all variables were normalised from 0 (minimum value of variable across all set-ups) to 100% (maximum value of variable across all set-ups) in each set-up. Group variables were then combined for each set-up to provide a single value to represent each of the four groups (performance times, mobility measures, contact, and release angles). For example, set-up one sprint, agility, and skill times were all normalised from 0-100% and the average of these three measures was then used to summarise performance times for each set-up. Improved performance for performance times is indicated by lower values, while improved performance for mobility measures is indicated by higher values. Higher values of propulsion angles represent angles that are closer to or further past TDC of the wheel.

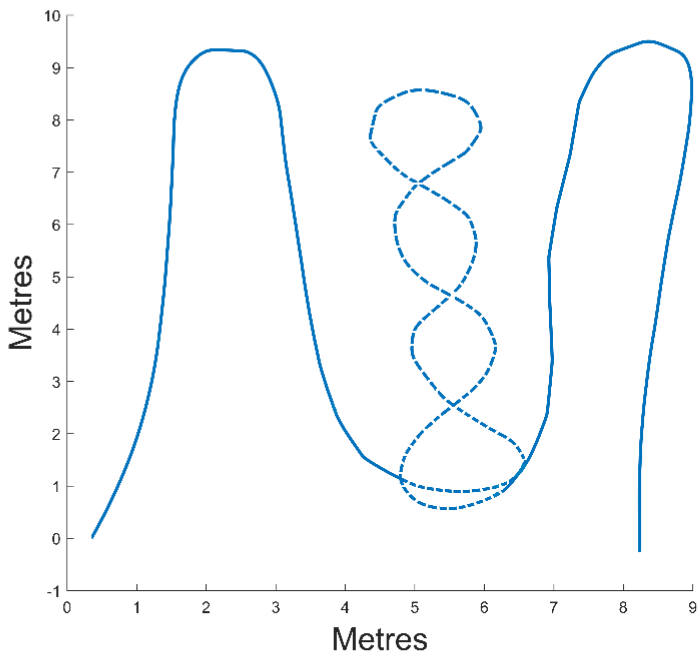


Figure 7.2: Using IMU tracking approaches (Shepherd, Wada, Rowlands, & James, 2016b; van der Slikke, et al., 2015), the path throughout the agility tests could be viewed, and key features such as the weave section (dashed) investigated in further detail. Data shown of a representative sample from the current study.

ANOVA testing involved comparisons of the four group values of all set-ups for each parameter (SH, SDep, SA, TP) across the three levels (increased, current, and decreased) for each player. Individual assessments were required due to the various impairment types and severities of participants, and parameter settings being based around the individual's current setting rather than a standardised measure [28]. Additionally, group analysis can mask important individual performance features – particularly in the initial strokes in acceleration from standstill [26] – further promoting the desire for individual assessments. However, attaining statistical significance for individual assessments is difficult due to the small sample sizes (in this case, three set-ups that contained each parameter level) and potentially small variations in performance.

Therefore, to aid in assessments of impacts on overall performance changes, radar plots were developed for each parameter to visualise the effects of the three settings on performance times and mobility measures. Although not providing a statistical assessment, visual interpretation across a wider range of variables can potentially provide similar influence. Improved performance is indicated by an increased distance from the centre of the plot. This aided selection of the preferred setting for each parameter. This was then compared with comments and feedback from the athlete and coach for the three of nine test set-ups that included the setting. A final decision on the parameter was then selected.

Through the selection of an optimal setting for each parameter, a theoretical best set-up that optimises the player performance was chosen for their specific on-court role. Each player then completed further testing – consisting of the same protocol and analysis in the adjustable wheelchair – in (i) their recommended and (ii) their current set-up. Player and coach then provided feedback on their preference between set-ups, only after which they were informed of the changes in the recommended set-up.

7.5 Results

The performance times for each player's follow-up testing in their current and recommended set-ups, as well as their blinded preference, is presented in Table 7-1. Three of six players preferred the recommended setting, while for others performance was similar despite preferring the current set-up (Players 4 and 6).

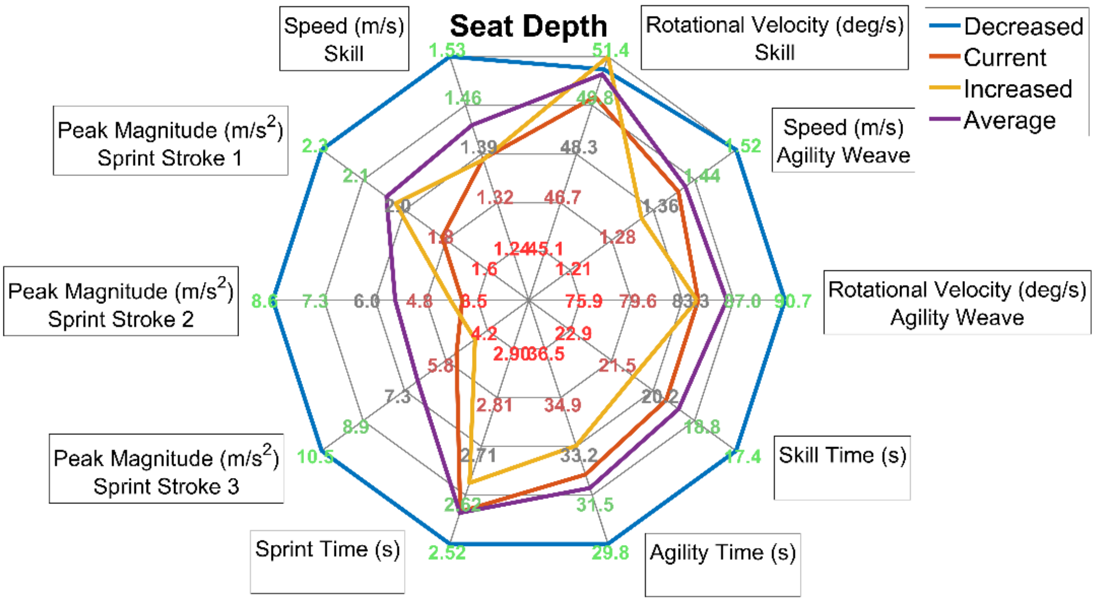
Due to the substantial amount of data analysed for each individual, a single case study will be presented and discussed in detail (Player 1). Analysis for the remaining players is available in supplementary material. Post-hoc testing revealed significant improvements for performance times for reducing SDep compared with increasing SDep (p=.05, difference of 55%, Cohen's d=2.87) – and a trend for mobility measures – for reduced SDep compared with both current (p=.09, difference of 44%, Cohen's d=2.34) and increased SDep (P=.10, difference of 44%, Cohen's d=2.34). There was also a trend towards contact angles closer to TDC of the wheel for the reduced SA compared with current (p=.07, difference of 40°, Cohen's d=3.03) and increased SAs (p=.09, difference of 37°, Cohen's d=2.48). Figure 7.3 displays the mean responses for specific variables for each of the wheelchair parameters (SH, SDep, SA, and TP) at each level (current, increased, and decreased) from the original robust design testing approach. For the provided case study, it is evident that a decrease in SDep and SA resulted in improved performance across a large majority of variables (Figures 7.3a and 7.3b, respectively) – supporting the statistical assessment. While increased SH appears to have benefits for peak acceleration magnitudes during the sprint, there is a reduction in the average rotational velocity during the skill test – likely influencing the slower time compared with the current SH (Figure 7.3c). A reduction in TP may also have benefits, with this setting showing improved performance across the majority of factors – particularly in comparison to an increase in TP. Propulsion kinematics for each parameter and setting are shown in Figure 7.4.

7.6 Discussion

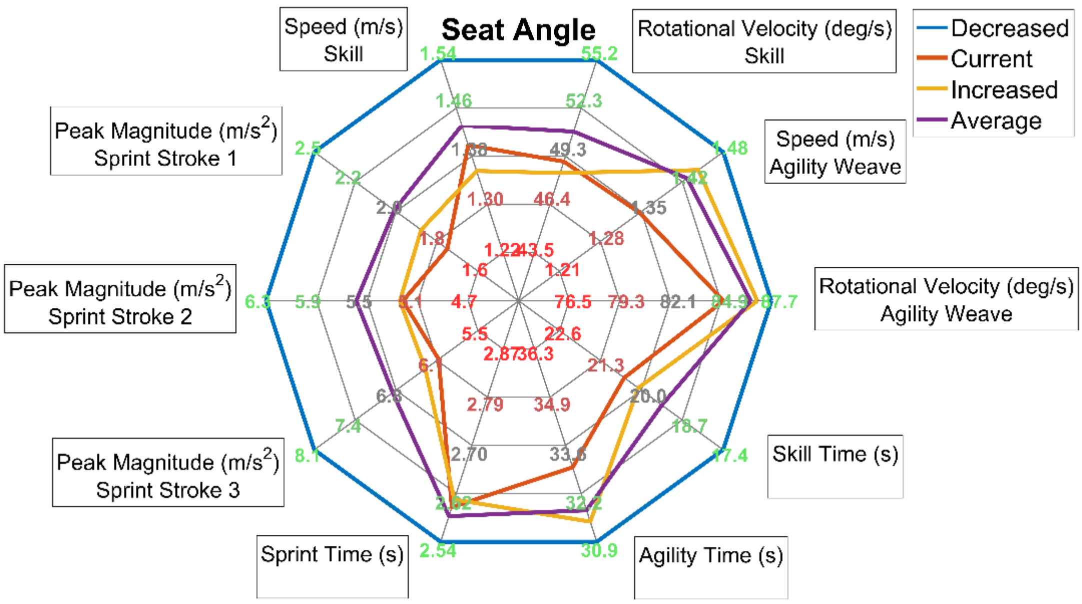
This study investigated a method for prescribing an optimal WCR set-up with specific parameter settings on performance measures, propulsion kinematics, and mobility measures. This was achieved by implementing an orthogonal test design with the use of representative on-court testing and monitoring of performance times, mobility measures with the use of IMUs (including average speed, rotational velocities), and any kinematic changes. Based on these results, a modifiable chair was used to test a recommended set-up and the athlete's current set-up. Of the six athletes who completed testing, three preferred the recommended configuration. Further, of the three who preferred their current set-up, one had similar performance through all tests for both set-ups (Player 6), while another had improved performance in the sprint and agility tests (Player 4). This method was attempted with elite players who had point classifications ranging from 0.5– to 3.5–points. Table 7-1 demonstrates players with varying point classifications, impairment types, and experience levels reported a blinded preference for the recommended set-up. This approach therefore has beneficial impacts across this elite population. It is also hypothesised that this approach could be implemented for developing athletes new to the sport, providing an opportunity to determine an optimal configuration earlier in their career.

Table 7-1: Individual player details and performance results for testing in the current (C) and recommended (R) settings. The faster of each timed measure are indicated with shading.

Player	Point Score	Impairment	International Experience (Years)	Setup (C/ R)	Sprint Time (s)	Agility Time (s)	Skill Time (s)	Blinded Preference
1	1.0	Impaired muscle power	8	C	2.53	29.00	15.72	✓
				R	2.55	30.46	15.79	
2	2.0	Impaired muscle power	3	C	1.99	24.50	10.79	
				R	2.00	24.65	10.88	✓
3	2.0	Impaired muscle power	10	C	2.34	28.75	13.78	
				R	2.24	27.42	14.42	✓
4	2.0	Limb deficiencies	1	C	2.38	26.41	10.99	✓
				R	2.28	26.34	11.71	
5	3.5	Limb deficiencies	3	C	1.90	20.53	12.51	
				R	1.73	21.01	12.00	✓
6	3.5	Limb deficiencies	6	C	1.84	22.74	10.93	✓
				R	1.89	22.65	10.92	

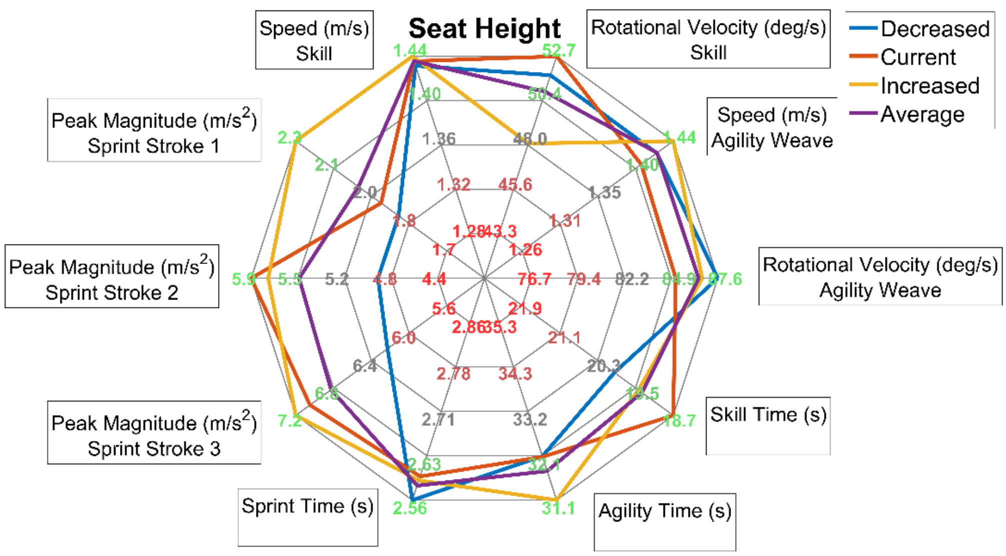


(a) Player 1 radar plot displaying the effects of the various SDep settings.

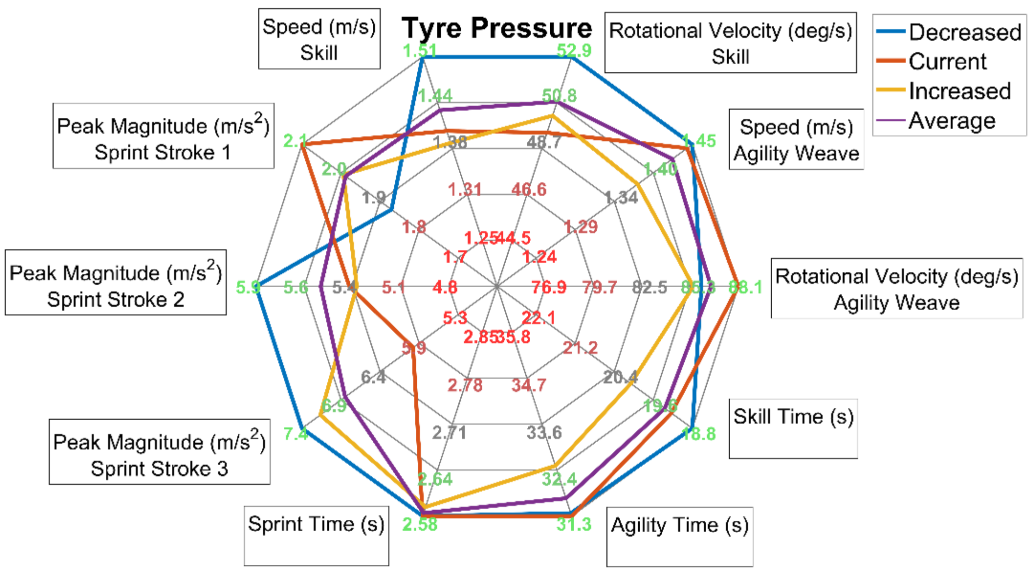


(b) Player 1 radar plot displaying the effects of the various SA settings.

Figure 7.3: Radar plots display the effects of the various levels of (a) SDep, (b) SA, (c) SH, and (d) TP on performance times, and mobility measures for sprint, agility, and skill tests. Axes are orientated with improved metrics at increased distances from the origin.



(c) Player 1 radar plot displaying the effects of the various SH settings.



(d) Player 1 radar plot displaying the effects of the various TP settings.

Figure 7.3: Radar plots display the effects of the various levels of (a) SDep, (b) SA, (c) SH, and (d) TP on performance times, and mobility measures for sprint, agility, and skill tests. Axes are orientated with improved metrics at increased distances from the origin.

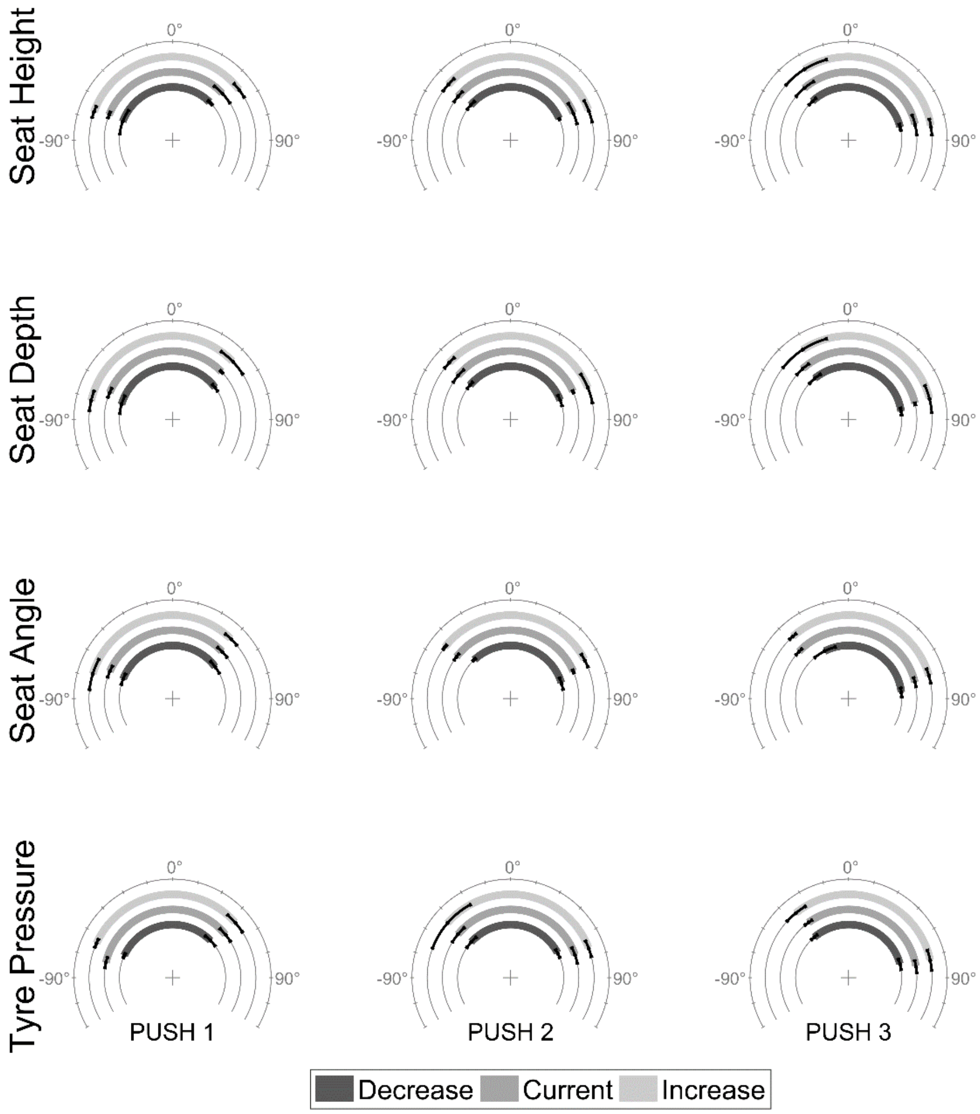


Figure 7.4: ContAng and RelAng positions for each factor and level across the first three strokes, with stroke direction to the right.

Detailed analysis for each individual includes assessment of performance times, propulsion kinematics, and mobility measures such as average speeds, rotational velocities and accelerations. For the case-study presented (Player 1), this identifies improved performance for reduced SDep (significantly improved performance times – $p<.05$ – and trend towards improved mobility measures – $p<.10$) and changes in propulsion technique with changing SA (trend of contact angles closer to TDC, $p<.10$). Although not a statistically supported approach, Figure 7.3b shows improved performance across all measures for the SA, and Figure 7.3a (SDep) shows improved performance for all measures excluding the rotational velocity during the skill test. These findings were expected, as reduced SA allows for greater peak accelerations [5] and reduced SDep results in improved manoeuvrability [8]. SH and TP did not have clear results, however there was potentially improved performance for increased SH and decreased TP. Due to the non-homogeneity of the WCR population and the need for individualised approaches, statistical comparisons across players and groups would not provide clear insights into wheelchair configuration effects.

Variations in propulsion kinematics were also evident across SA levels, with a trend of ContAngs closer to TDC of the wheel, and release angles further around the wheel for reduced SAs compared with the current and increased settings. As reducing the SA flattens the legs and promotes greater trunk motion [5], release angles would be expected to increase. Seat dump (the angle between the seat and backrest) was kept constant throughout testing, therefore a flatter SA produced a forward shift to the shoulder position, hence the ContAngs closer to TDC. Increased SH also resulted in ContAngs closer to TDC for the third stroke, likely due to the reduced access to the pushrim/wheel [8].

This analysis of performance times, mobility measures and propulsion kinematics leads to the selection of the setting for each parameter in the recommended set-up for Player 1. Despite the benefits evident in orthogonal design testing, the recommended set-up decreased performance in comparison to the current set-up in follow-up testing (see Table 7-1). This may be as a result of all changes to seating position (decreased SA and SDep, increased SH) reducing stability and increasing manoeuvrability [1]. The selection of these parameter levels therefore likely led to a seating position that was difficult to control, resulting in reduced force application by the player due to a lack of confidence in the set-up. There is the possibility that selection of reduced SA or SDep in isolation – the parameters that had greatest impact on performance – would improve performance, however this was not investigated. Further experience with this approach may aid the selection of recommended set-ups, where the interaction of parameters receives greater consideration. The inability to predict coupled effects from altering more than one variable at a time is a necessary limitation of the orthogonal design approach.

Throughout testing, feedback from players and coaches aided in the assessment of the method. The use of an adjustable wheelchair allowed for a range of players and configurations to be tested, however there are restrictions in how accurately an adjustable wheelchair is able to replicate finer characteristics of each individual's current wheelchair. This includes mass distribution, inclusion/exclusion of ball carriers, and inclusion of an individual's full strapping approach. Due to the variations across high- and low-point wheelchairs [24], a single adjustable wheelchair that accounts for all design possibilities is unrealistic. Instead, the ability to add or remove mass to specific areas of the adjustable wheelchair to replicate the player's current wheelchair mass distribution would be beneficial, although this was not performed in this study. Further, whilst the inclusion of various strapping approaches would potentially allow players to feel more secure in the wheelchair, a reduction in strapping (provided it is consistent throughout testing) may provide a clearer indication of the effect of the set-up on performance.

7.7 Conclusion

The implementation of an orthogonal design approach with representative on-court testing, monitoring of performance, mobility, and kinematic measures, has been demonstrated to identify the effects of various parameter settings. This information can then be used to select a near optimal set-up that results in improved performance for elite, experienced WCR players. Achieving improvements with the current participant group suggests that this approach could benefit new or developing WCR players in finding an optimal wheelchair set-up earlier in their sporting career. Future work should continue to advance the relevance of testing, adjustable wheelchair features, and ability to distinguish the crucial findings and apply these to a recommended wheelchair set-up at an individual level.

Acknowledgements

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7.10 Supplementary Material

7.10.1 Orthogonal Design

Orthogonal design is a robust design approach that aims to 'improve product quality and reduce cost efficiently in real-life industry applications' (Mori & Tsai, 2011). The approach focuses on improving product quality and reducing cost by improving the optimisation approach. Through the use of orthogonal arrays, an improved optimisation approach is achieved. The orthogonal arrays are used to assess an optimal level for each experimental factor through a reduced amount of testing, by considering the level average of each factor. Level averages can then be used against the grand average (average output value for all trials) to assess the effects of an experimental factor. Experimental factors can vary in the number of variations investigated (investigating seat height at 2 or 3 levels), but each variation is tested the same number of times throughout the orthogonal array. An L_9 (3^4) has 4 factors that are investigated, with each of these having 3 variations or testing conditions. This then requires 9 trials to complete the testing, with an example orthogonal array presented below.

An output measure for each trial is then monitored. In this case, this was the performance times, mobility measures, and propulsion kinematics. For each of these, the level average is calculated as an average of the output values for the specific level of an experimental factor (i.e. the level average of Seat Height A will be the average of all output values where Seat Height A is tested). The effect of Seat Height A can then be assessed against the grand average for Seat Height to see if a positive or negative influence on performance is present. The calculation of meaningful differences can then be applied to assess which parameters and settings have an influence. A theoretical optimal set-up can then be determined for each individual based on the optimal level for each position, with this position trialled using the adjustable wheelchair. For further information regarding this approach, see Mori and Tsai (2011).

Table 7-2: Example L9 orthogonal array, with four parameters varied at three levels throughout nine set-ups.

Set-up	Seat Height	Seat Depth	Seat Angle	Tyre Pressure
1	A	A	A	A
2	A	B	B	B
3	A	C	C	C
4	B	A	B	C
5	B	B	C	A
6	B	C	A	B
7	C	A	C	B
8	C	B	A	C
9	C	C	B	A

Mori, T., & Tsai, S.-C. (2011). *Taguchi Methods: Benefits, Impacts, Mathematics, Statistics, and Applications* New York: ASME.

7.10.2 Skill Test Design

The skill test developed focuses on an individual and their ability to control the wheelchair during turns, passing and receiving. This includes bounce passes against (around the cones marked 'X' on the figure below), and chest passes (around the 'O' cone) against a wall. The time taken to complete the test was recorded. Two skill tests were performed for each wheelchair set-up, with one test in the direction shown below, and the other in the reverse direction (i.e., around left 'X' cone first). If a pass or receive was not successful, the trial was not counted. This test also provided crucial feedback to the player and coach of the control for the particular set-up, which greatly influenced the feedback provided.

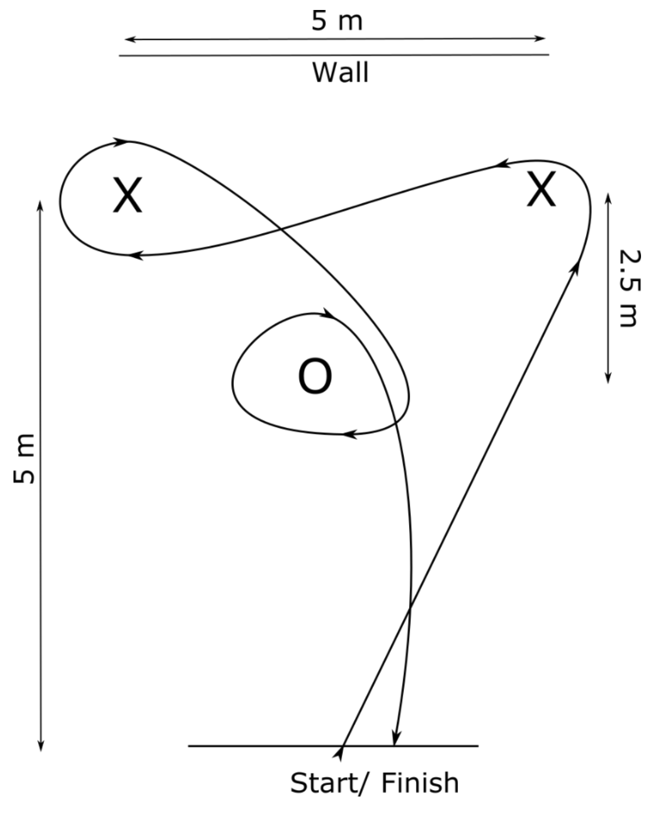


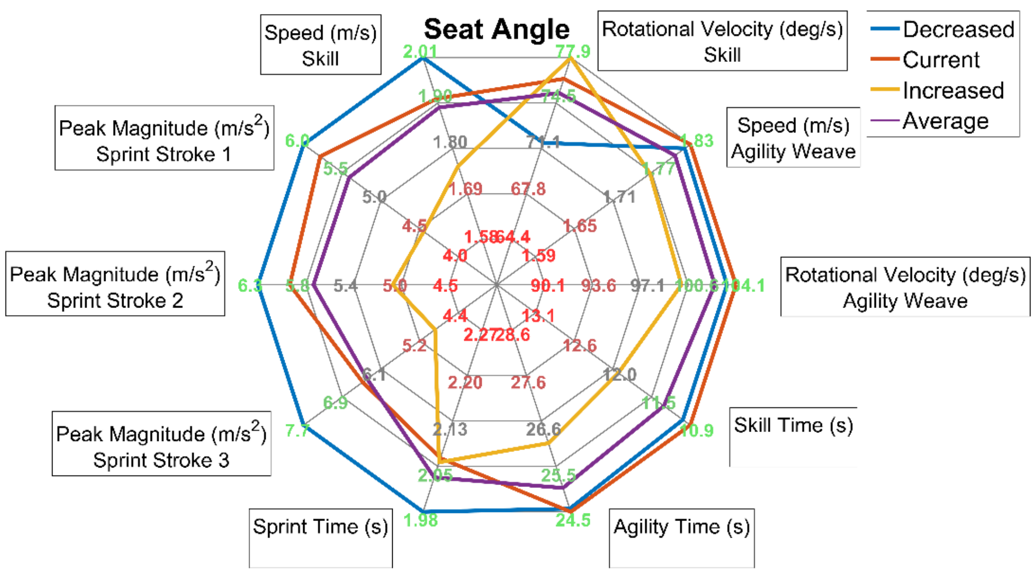
Figure 7.5: The diagram above details the path taken during the skill test. The 'X' markers represent cones at which the player had to execute and receive a bounce pass against the wall, while the 'O' marker represents a cone where the player performed a chest pass against the wall.

7.10.3 Additional Results

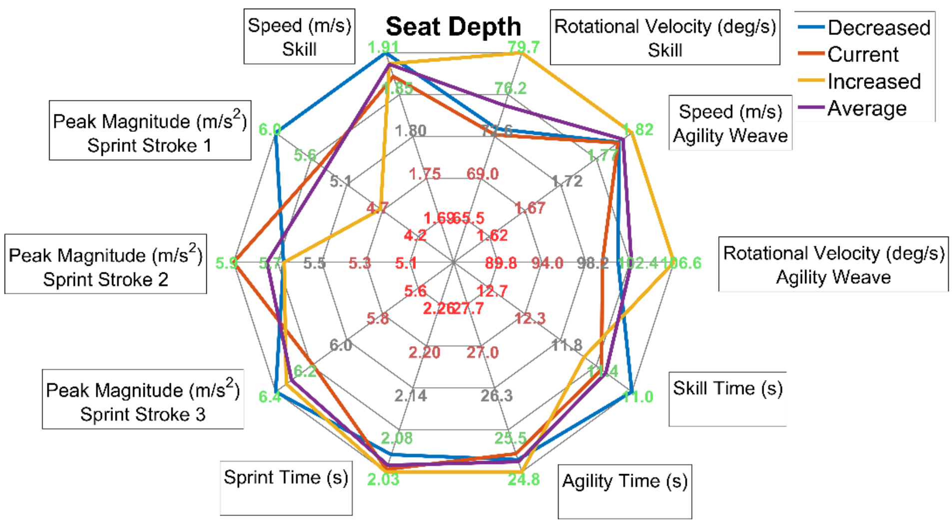
The following figures present the radar plots for Players 2-6 for each of the configuration parameters (seat height, seat depth, seat angle, and tyre pressure). In addition, any kinematic changes across the first three propulsion strokes for each individual are presented.

Player 2

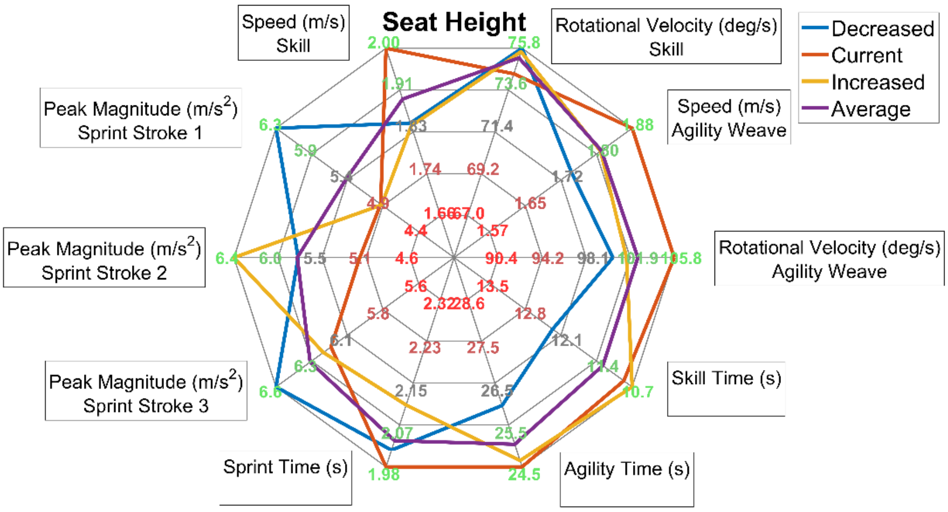
Decreased seat angle showed trends for improved mobility measures ($p=0.066$, difference is 35%) as suggested on radar plots below, particularly for peak magnitudes during sprinting. Trends for smaller release angles with higher seat angles were seen compared with low ($p=0.08$, difference is 47°) and mid ($p=0.056$, difference is 53°) angles. The improvements in peak magnitudes in sprinting likely aided the improved sprint time for decreased seat angle. Increased seat depth had no significant findings but radar plots show a potential trend for improved mobility measures from the skill and agility tests. This is reflected in the agility time, but not the skill time – potentially related to effects on ball-handling. Increased tyre pressure showed best performance for all variable, although these differences were often small. No clear, consistent patterns were seen for seat height.



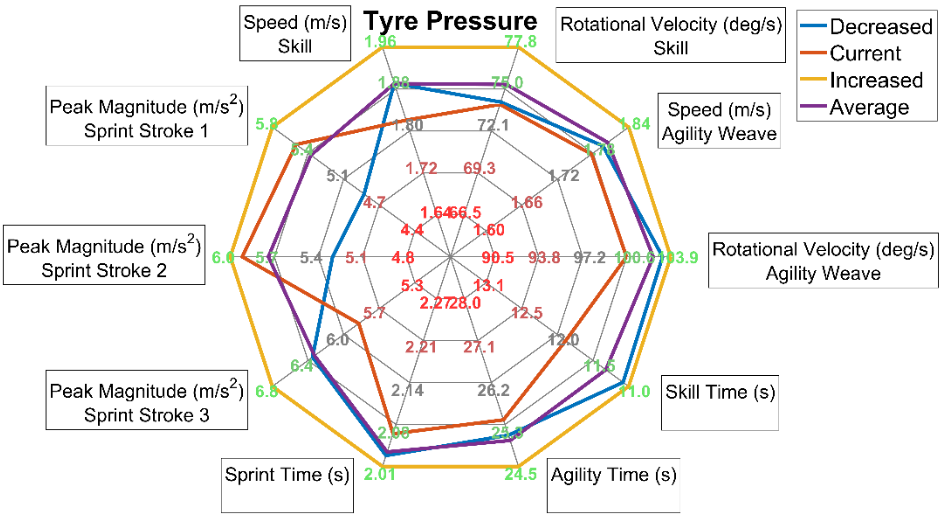
(a) Player 2 radar plot displaying the effects of the various SA settings.



(b) Player 2 radar plot displaying the effects of the various SDep settings.



(c) Player 2 radar plot displaying the effects of the various SH settings.



(d) Player 2 radar plot displaying the effects of the various TP settings.

Figure 7.6: Radar plots display the effects of the various levels of (a) SA, (b) SDep, (c) SH, and (d) TP on performance times, and mobility measures for sprint, agility, and skill tests. Axes are orientated with improved metrics at increased distances from the origin.

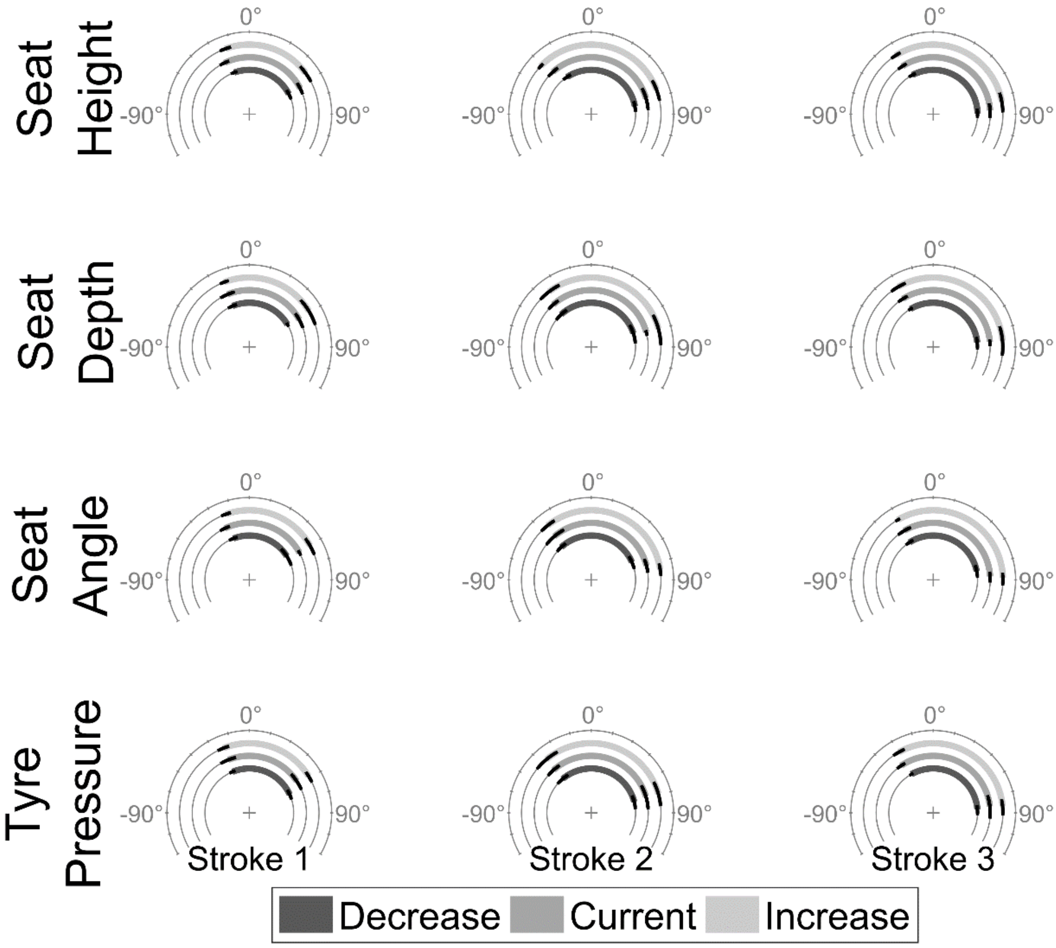
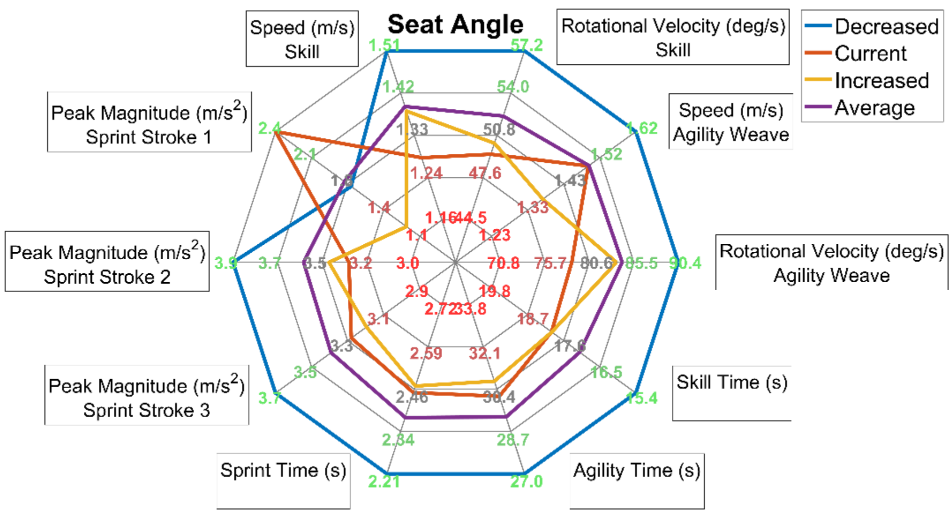


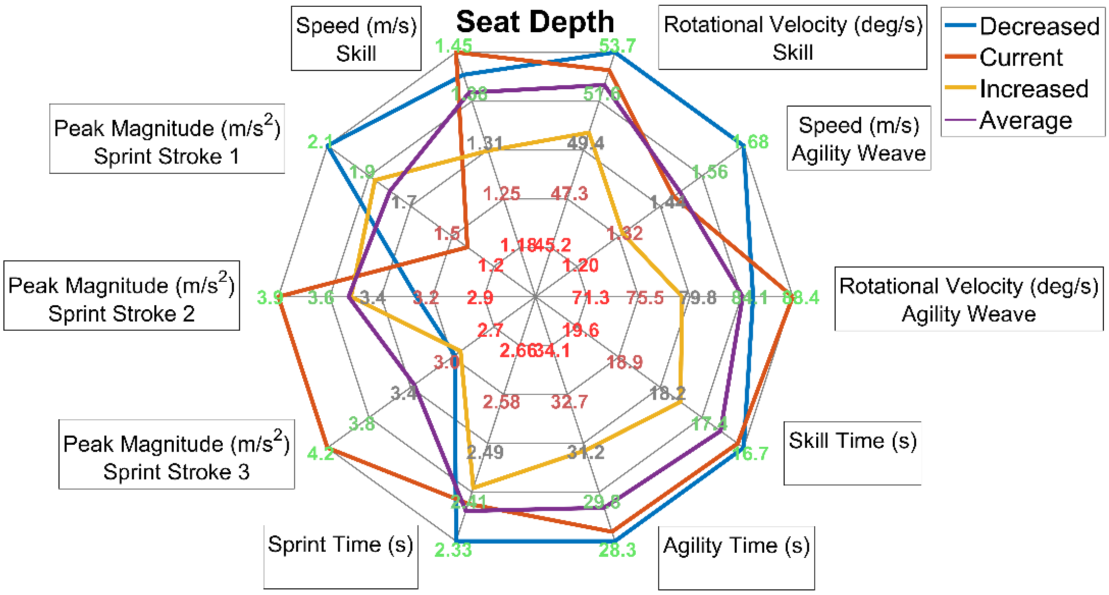
Figure 7.7: Player 2 ContAng and RelAng positions for each factor and level across the first three strokes, with stroke direction to the right.

Player 3:

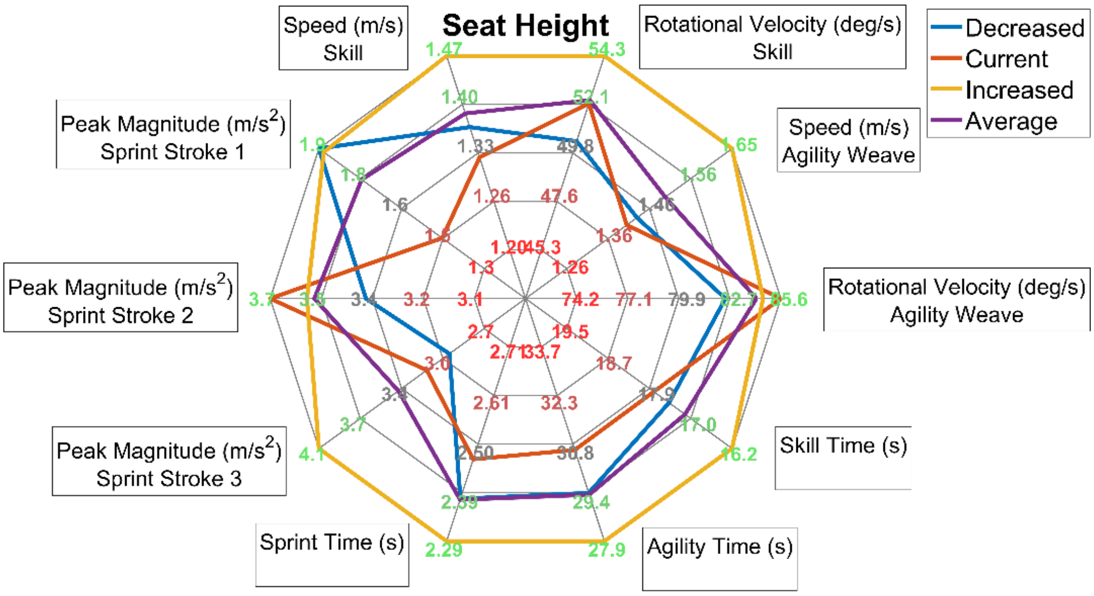
Despite clear patterns evident in radar plots for seat angle, seat height and tyre pressure, no significant differences were achieved. This is likely due to the variations in performance between settings, with this player appearing to be substantially affected by set-up and hence large standard deviation of variables achieved. However, a decreased seat angle improved performance for all variables bar peak magnitude for the first stroke. There also appears to be a reasonable gap for most variables compared with current and increased settings, hence a reduction on current seat angle was recommended. Increased seat height also showed improved performance for a large majority of variables, particularly the test times. Somewhat surprisingly, current tyre pressure achieved similar improved performance in the majority of variables – this may reflect greater attention to tyre pressure by this athlete. Based on these findings, a recommended set-up would consist of decreased seat angle, increased seat height and the current tyre pressure.



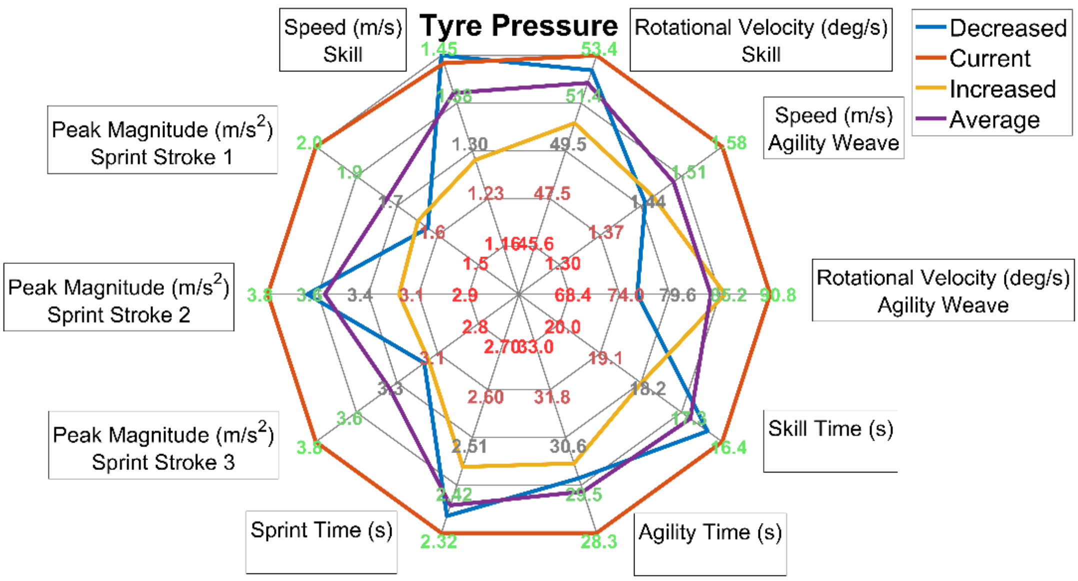
(a) Player 3 radar plot displaying the effects of the various SA settings.



(b) Player 3 radar plot displaying the effects of the various SDep settings.



(c) Player 3 radar plot displaying the effects of the various SH settings.



(d) Player 3 radar plot displaying the effects of the various TP settings.

Figure 7.8: Radar plots display the effects of the various levels of (a) SA, (b) SDep, (c) SH, and (d) TP on performance times, and mobility measures for sprint, agility, and skill tests. Axes are orientated with improved metrics at increased distances from the origin.

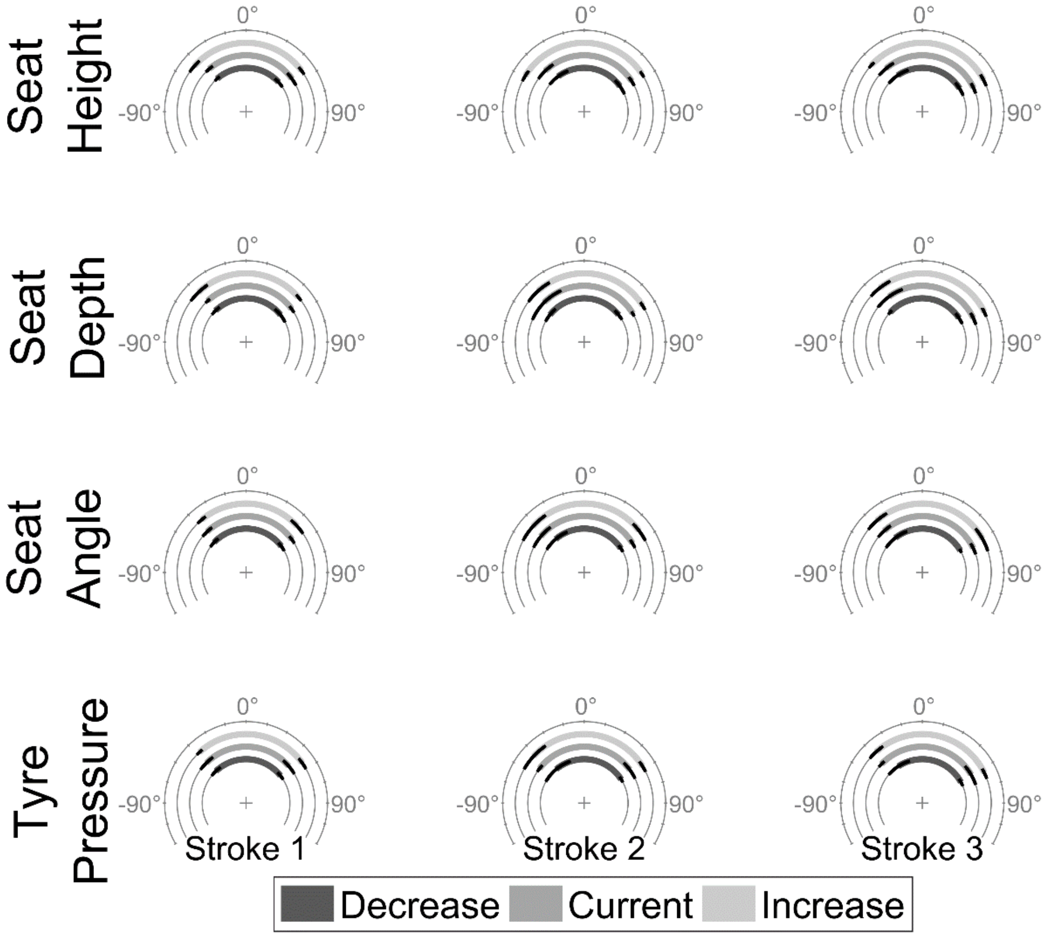
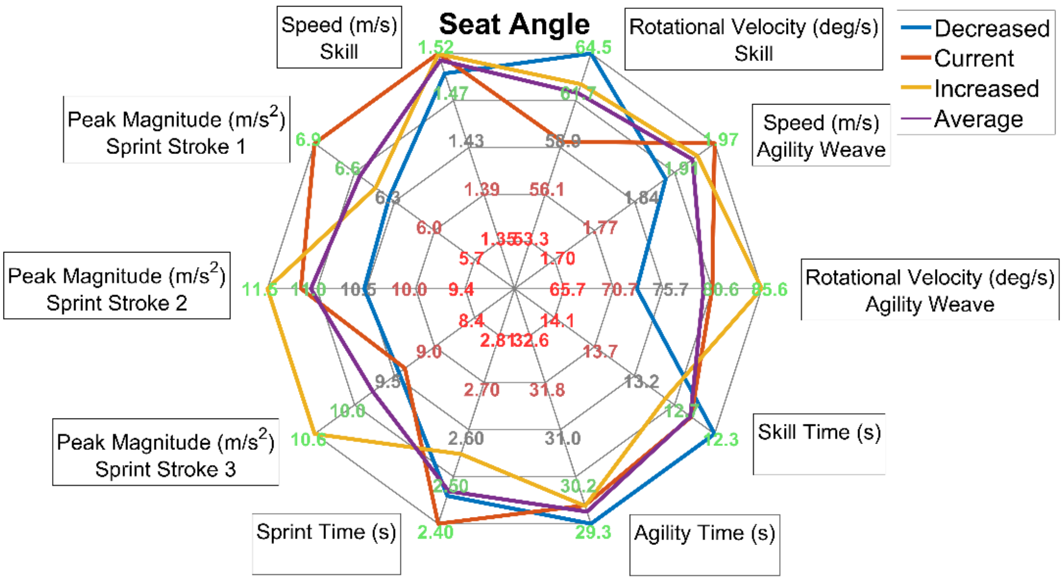


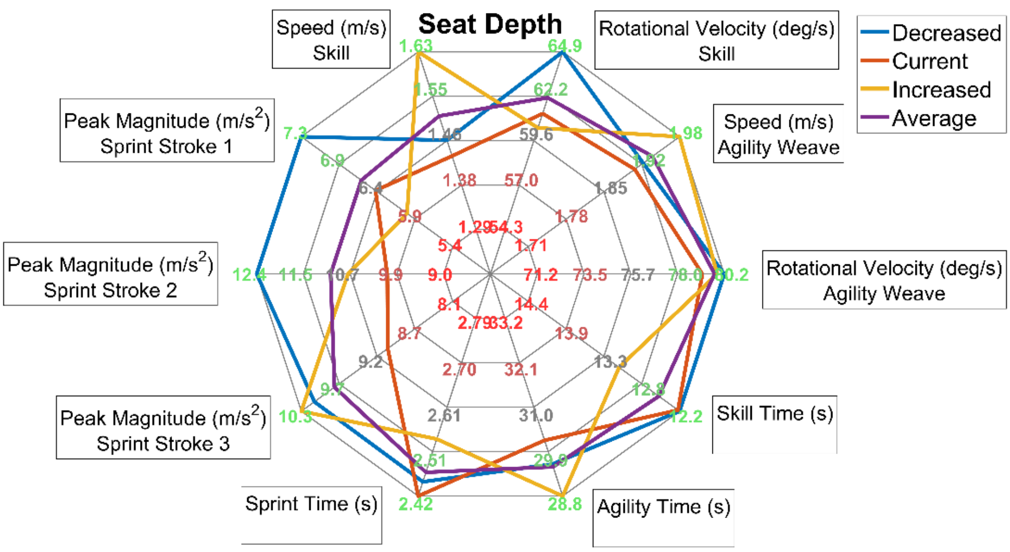
Figure 7.9: Player 3 ContAng and RelAng positions for each factor and level across the first three strokes, with stroke direction to the right.

Player 4:

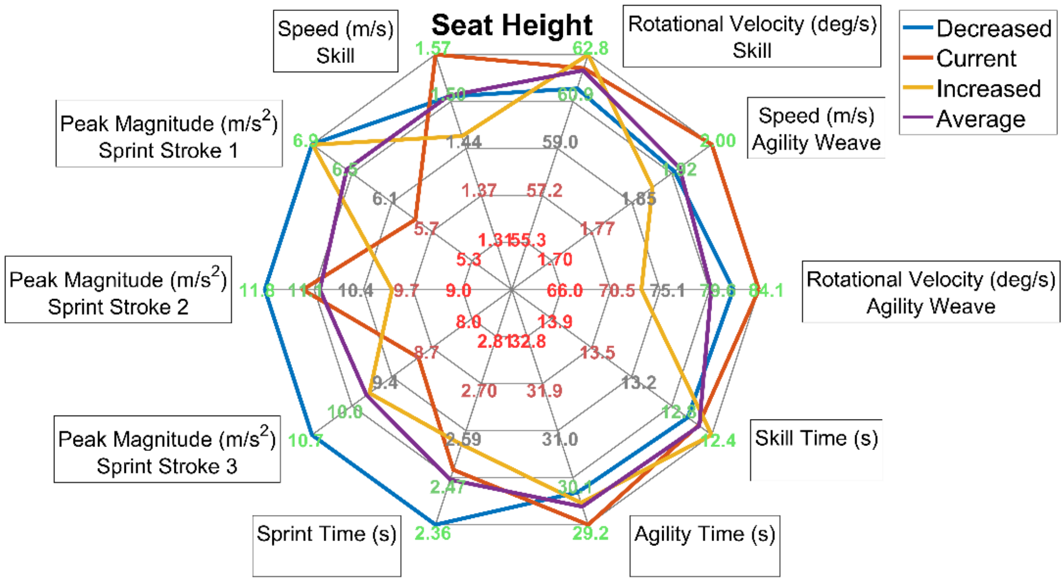
No significant findings were evident for any parameters or variables. Radar plots show few consistent trends, although both seat height and seat depth show larger peak magnitudes for the decreased setting. Increased tyre pressure achieved best times for all tests, but these differences were relatively small.



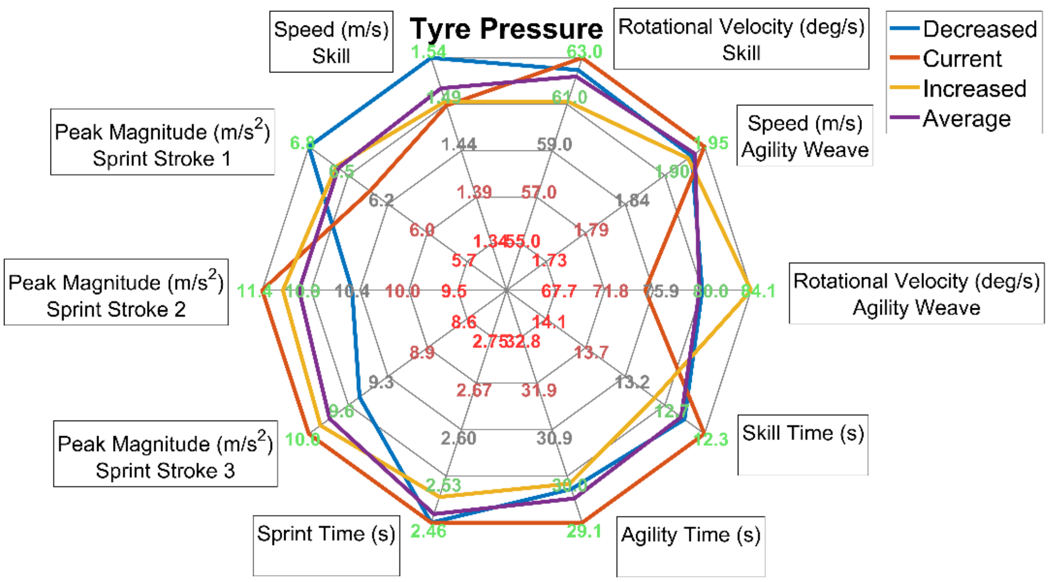
(a) Player 4 radar plot displaying the effects of the various SA settings.



(b) Player 4 radar plot displaying the effects of the various SDep settings.



(d) Player 4 radar plot displaying the effects of the various SH settings.



(d) Player 4 radar plot displaying the effects of the various TP settings.

Figure 7.10: Radar plots display the effects of the various levels of (a) SA, (b) SDep, (c) SH, and (d) TP on performance times, and mobility measures for sprint, agility, and skill tests. Axes are orientated with improved metrics at increased distances from the origin.

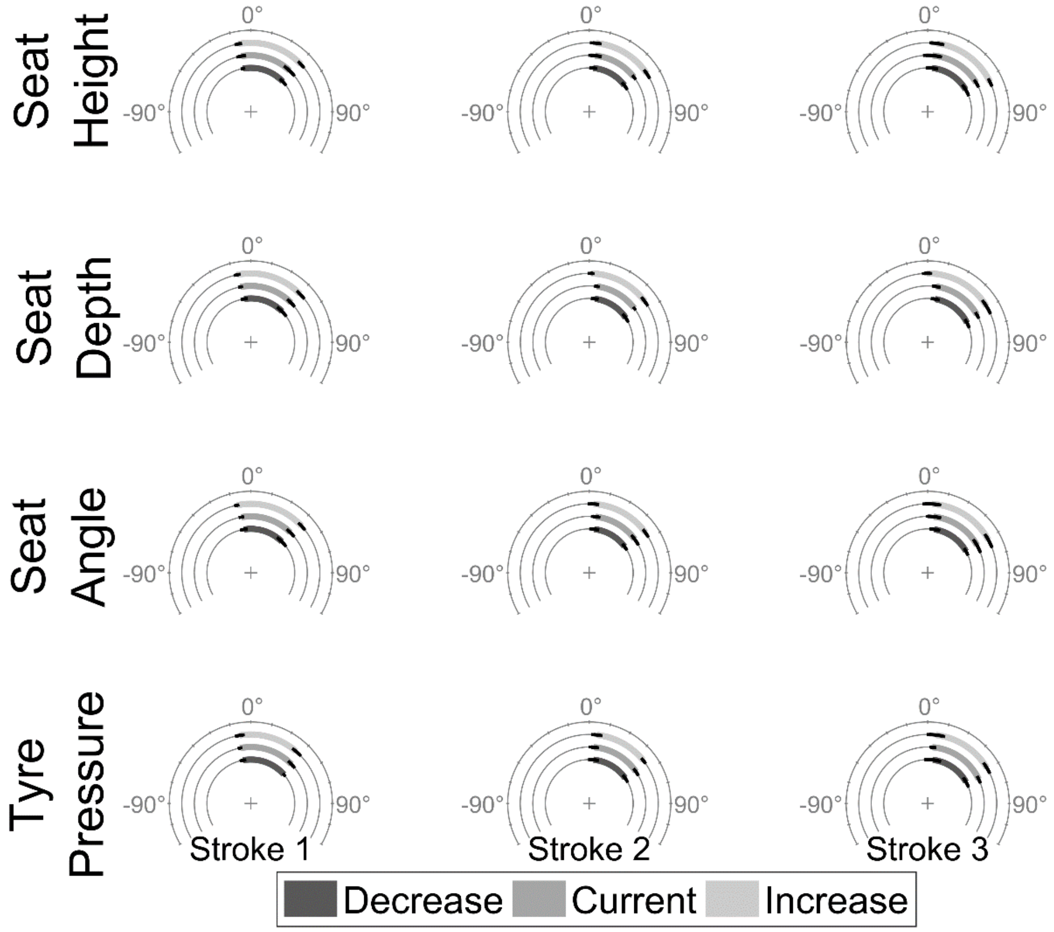
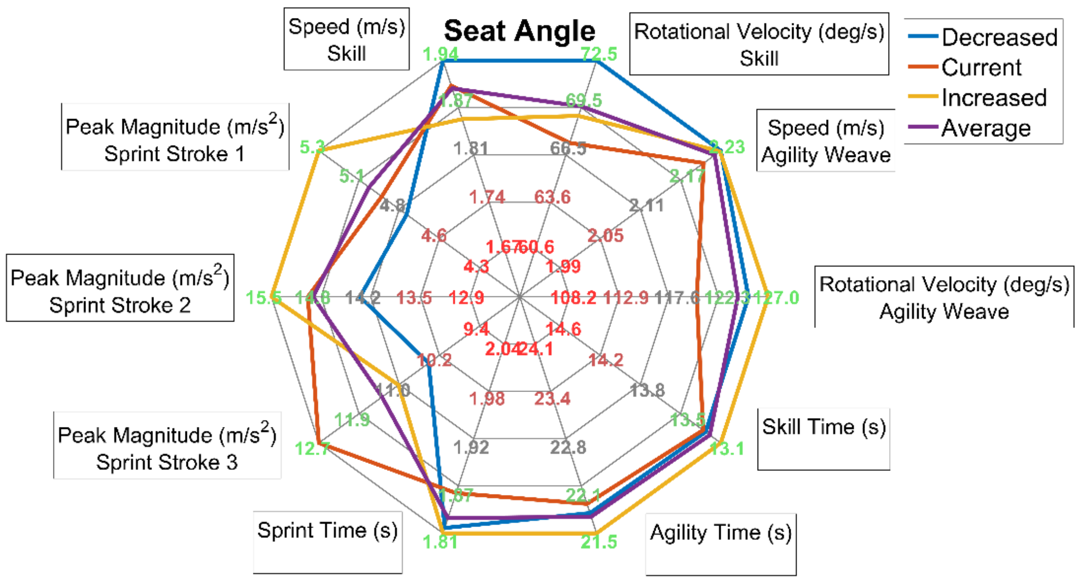


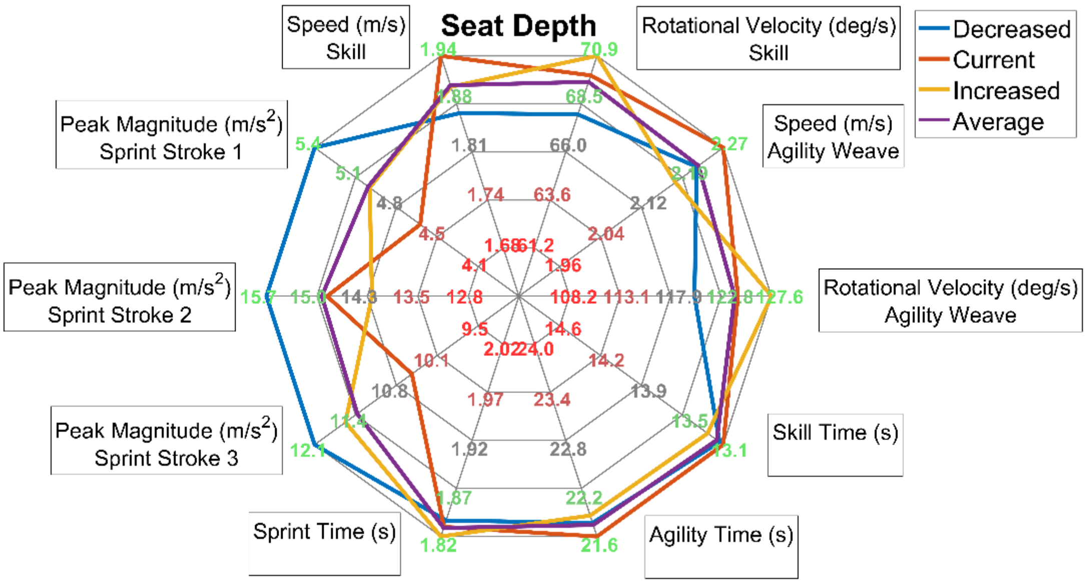
Figure 7.11: Player 4 ContAng and RelAng positions for each factor and level across the first three strokes, with stroke direction to the right.

Player 5:

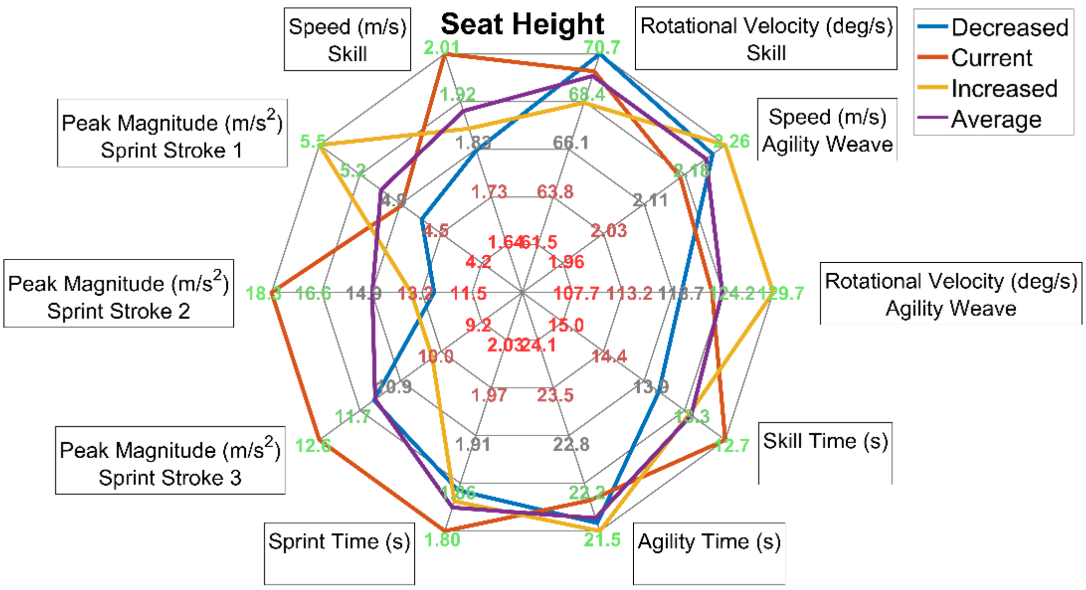
Seat height produced the only significant differences across settings. The current seat height result in improved mobility measures compared with the low (p=0.01, difference of 20.8%) and high (p=0.023, difference of 10.7%) settings. This is likely dependent on the large increase in peak magnitudes for strokes two and three of the linear sprint. There was also improvement for the high setting compared with low (p=0.029, difference of 10.1%). Decreased seat depth displayed a similar trend in peak magnitudes in the radar plot, while decreasing seat angle showed a contrasting pattern – improved speed and rotational velocity in the agility and skill tests but reduced peak magnitudes for linear sprinting. There was little difference in times for the various tests, and no clear trends for tyre pressure.



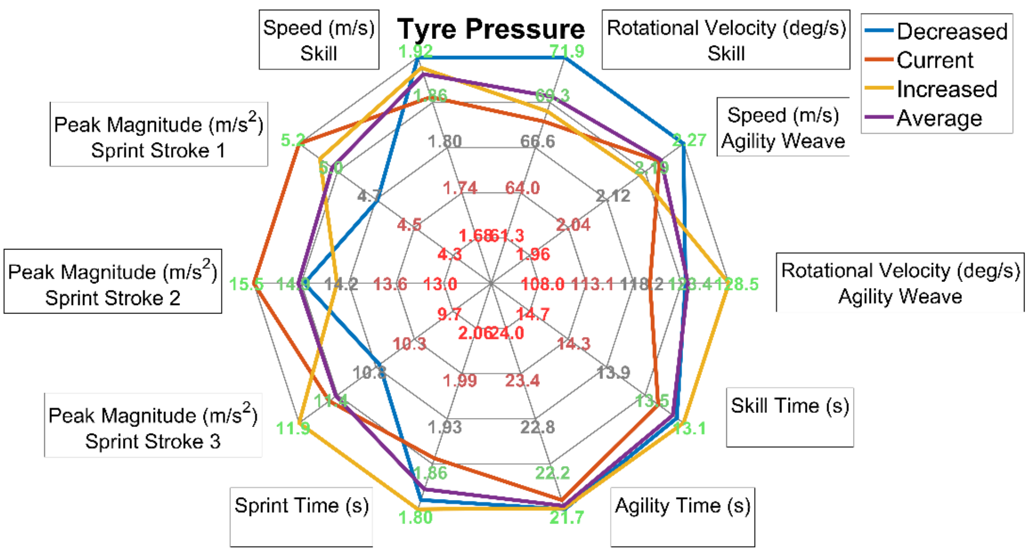
(a) Player 1 radar plot displaying the effects of the various SA settings.



(b) Player 1 radar plot displaying the effects of the various SDep settings.



(c) Player 1 radar plot displaying the effects of the various SH settings.



(d) Player 1 radar plot displaying the effects of the various TP settings.

Figure 7.12: Radar plots display the effects of the various levels of (a) SA, (b) SDep, (c) SH, and (d) TP on performance times, and mobility measures for sprint, agility, and skill tests. Axes are orientated with improved metrics at increased distances from the origin.

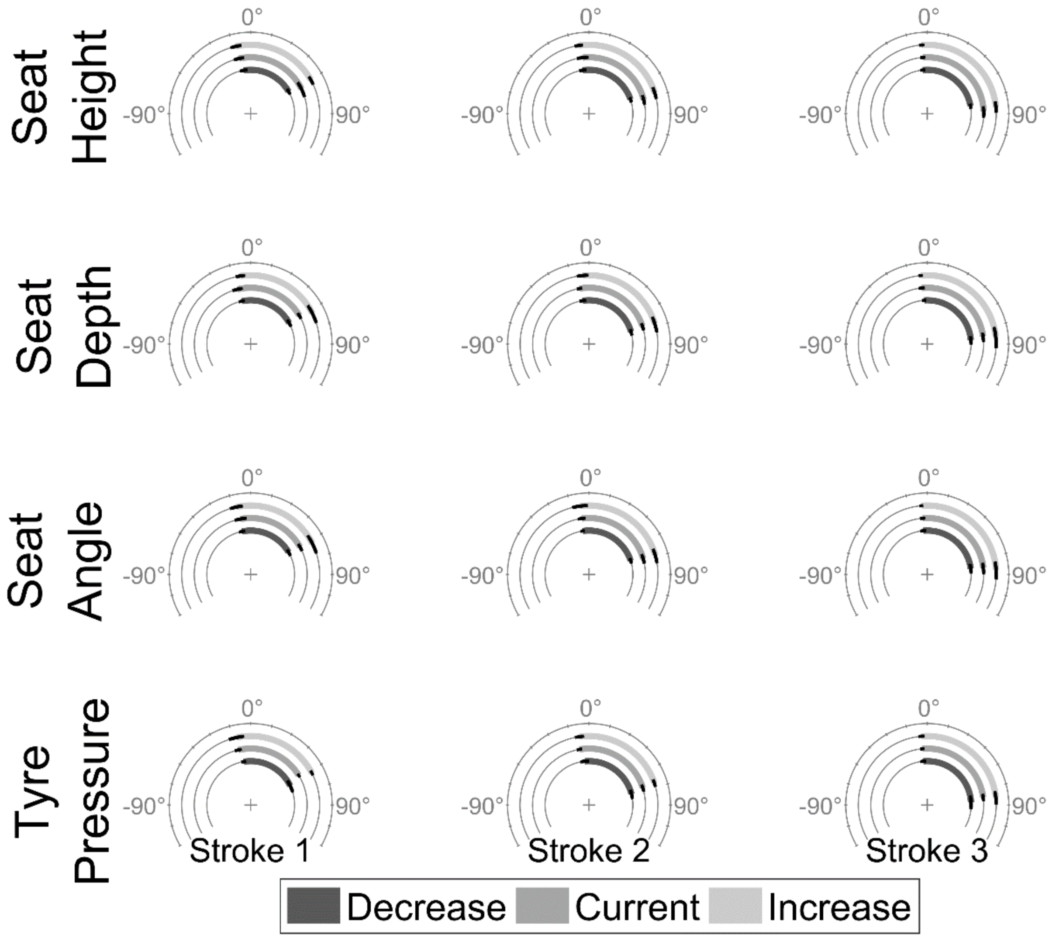
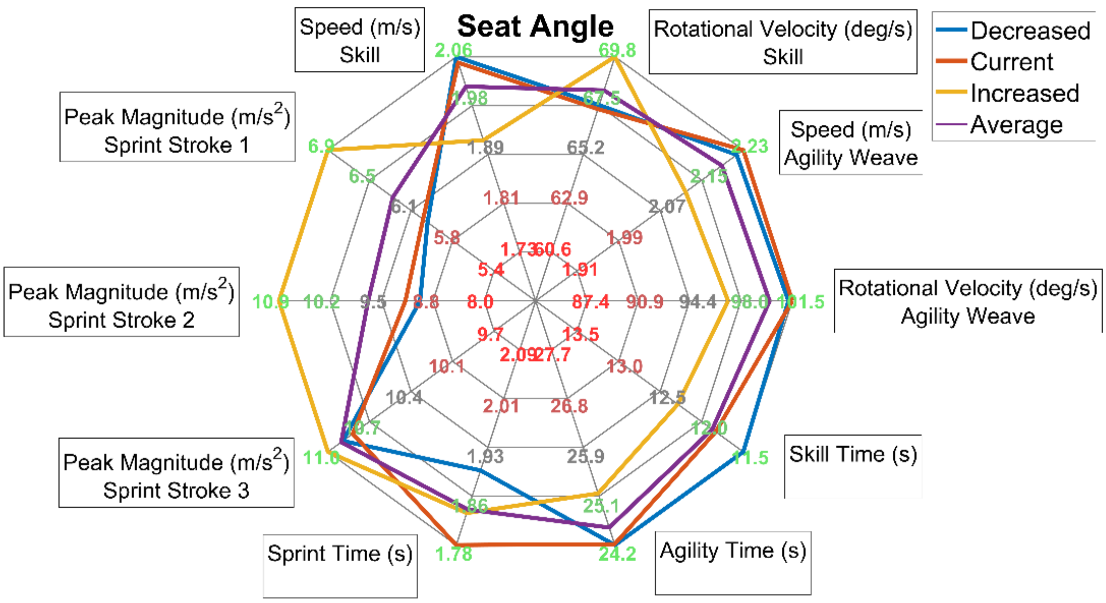


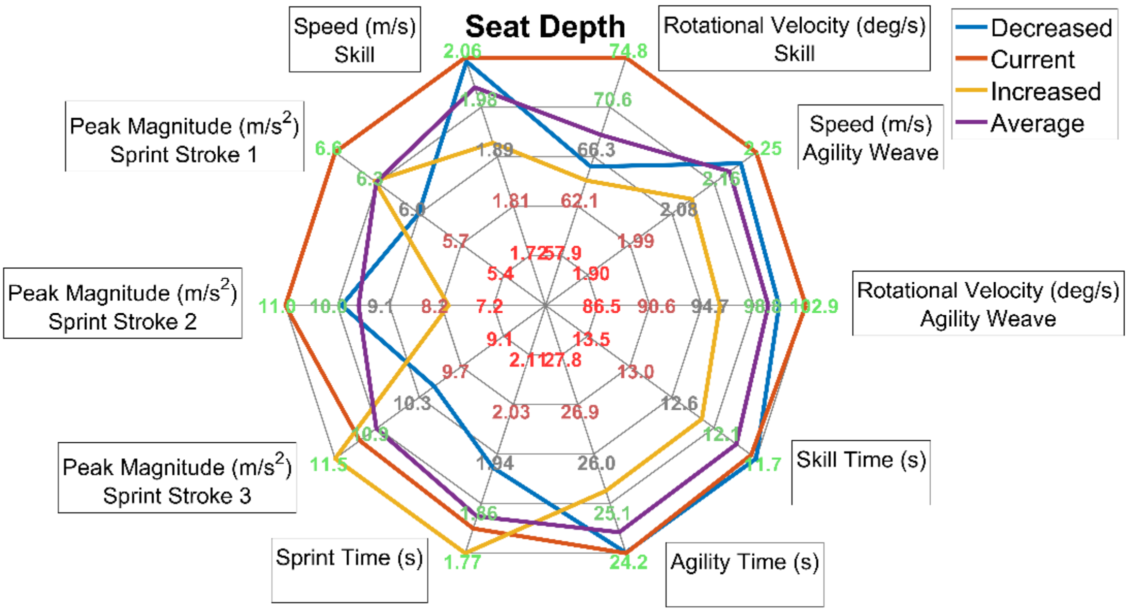
Figure 7.13: Player 5 ContAng and RelAng positions for each factor and level across the first three strokes, with stroke direction to the right.

Player 6:

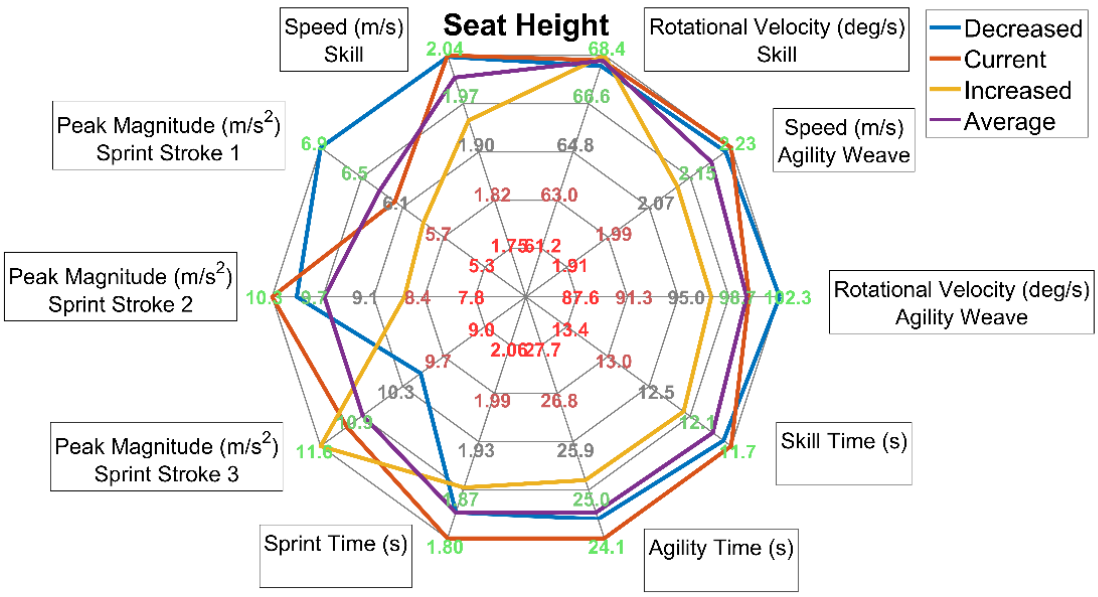
Significant improvements in mobility measures was evident for the current seat depth compared with the increased setting (p=0.034, difference of 35.1%). The seat depth radar plot shows improved performance for the majority of variables, although this did not translate to best times in any of the tests. No significant findings were found for any other parameter settings. Inspection of radar plot identifies increased seat angle surprisingly has potential benefits in peak magnitudes during linear sprinting, but limits performance during rotational activities as indicated by slow times in agility and skill tests. Decreased seat height potentially had contrasting effects – improving rotational performance but limiting peak magnitudes during sprinting. Current seat height was the selected setting due it's consistently strong performance across all variables. Lower tyre pressure potentially had limited performance benefits compared with both the current and increased setting.



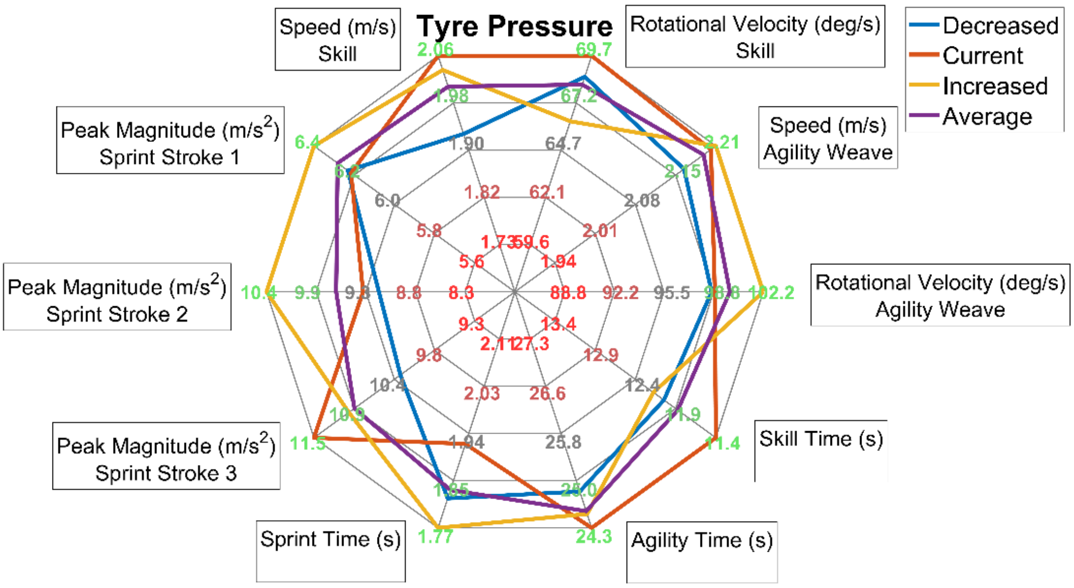
(a) Player 1 radar plot displaying the effects of the various SA settings.



(b) Player 1 radar plot displaying the effects of the various SA settings.



(c) Player 1 radar plot displaying the effects of the various SH settings.



(d) Player 1 radar plot displaying the effects of the various TP settings.

Figure 7.14: Radar plots display the effects of the various levels of (a) SA, (b) SDep, (c) SH, and (d) TP on performance times, and mobility measures for sprint, agility, and skill tests. Axes are orientated with improved metrics at increased distances from the origin.

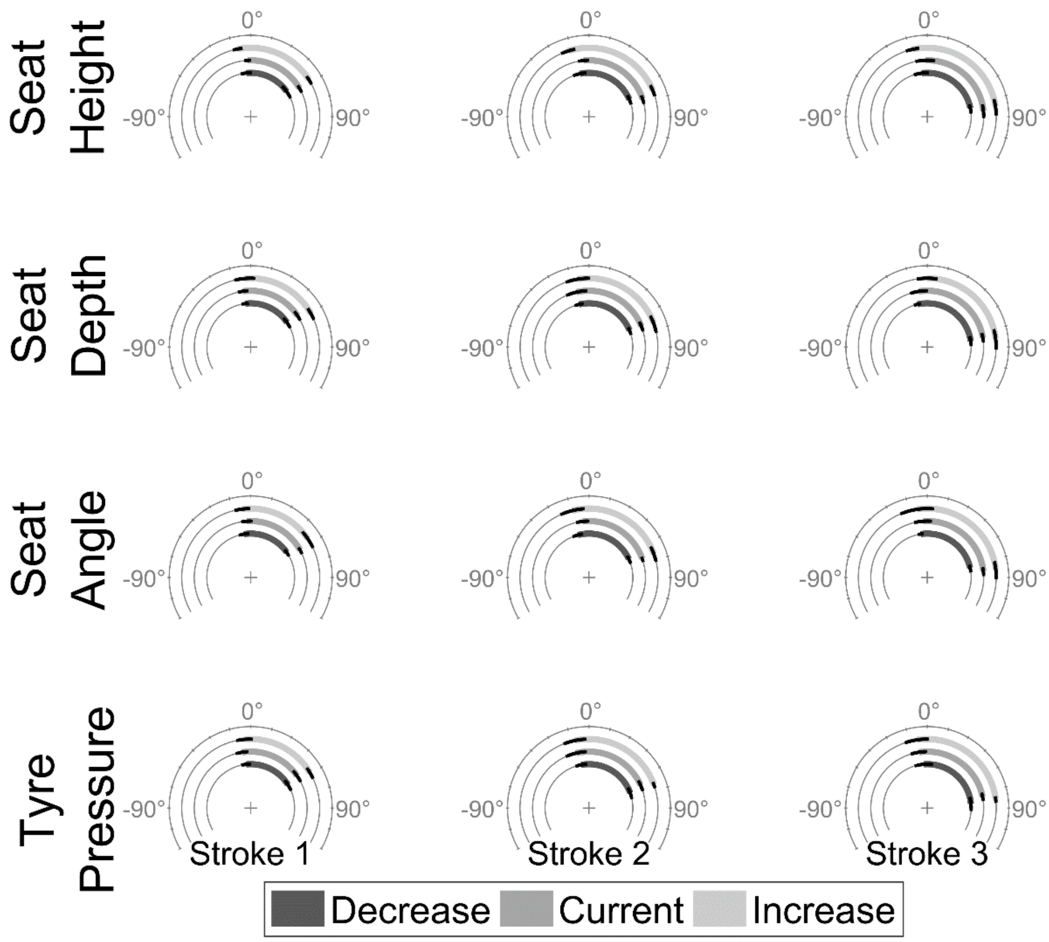


Figure 7.15: Player 6 ContAng and RelAng positions for each factor and level across the first three strokes, with stroke direction to the right.

Chapter 8: Predicting Sprint Performance in Wheelchair Rugby Using a Linkage Model

Despite improvements in testing time achieved in Chapter 7, further improvements are possible through the use of prediction modelling. This chapter details the use of a linkage model to predict propulsion kinematics for various wheelchair configurations based on individual propulsion techniques, and then uses these details to predict performance outcomes in a linear sprint.

This chapter has been submitted as a journal article (see below details) and has been reformatted for the purpose of this thesis. This submission satisfies University of Adelaide requirements for inclusion in a Thesis by Publication.

> Haydon, D.S., Pinder, R.A., Grimshaw, P.N., Robertson, W.S.P. Predicting Sprint Performance in Wheelchair Rugby Using a Linkage Model. Submitted to Journal of Biomechanics, August 2018.

8.1 Statement of Authorship

Title of Paper	Predicting Sprint Performance in Wheelchair Rugby Using a Linkage Model
Publication Status	☐ Published ☐ Accepted for Publication ☑ Submitted for Publication ☐ Unpublished and Unsubmitted work written in manuscript style
Publication Details	Haydon, D.S., Pinder, R.A., Grimshaw, P.N., Robertson, W.S.P. Predicting Sprint Performance in Wheelchair Rugby Using a Linkage Model. **Submitted** to Journal of Biomechanics, August 2018.

Principal Author

Name of Principal Author (Candidate)	David S Haydon		
Contribution to the Paper	Analysis of data input for model, execution of the modelling approach, analysis of results, and manuscript preparation.		
Overall percentage (%)	60		
Certification:	This paper reports on original research I conducted during the period of my Higher Degree by Research candidature and is not subject to any obligations or contractual agreements with a third party that would constrain its inclusion in this thesis. I am the primary author of this paper.		
Signature		Date	18/08/2018

Co-Author Contributions

By signing the Statement of Authorship, each author certifies that:

iv. the candidate's stated contribution to the publication is accurate (as detailed above);

v. permission is granted for the candidate in include the publication in the thesis; and

vi. the sum of all co-author contributions is equal to 100% less the candidate's stated contribution.

Name of Co-Author	Dr Ross A Pinder		
Contribution to the Paper	Supervision, analysis of results, and manuscript preparation.		
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Contribution to the Paper	Supervision, analysis of results, and manuscript preparation.		
Signature		Date	18/08/2018

Name of Co-Author	Dr William SP Robertson		
Contribution to the Paper	Supervision, analysis of results, and manuscript preparation.		
Signature		Date	18/08/2018

8.2 Abstract

Prediction of propulsion and performance in wheelchair sport has the potential to improve capabilities of individual wheelchair prescription while reducing testing requirements. Despite possible benefits, prediction methods have not been applied to maximal effort propulsion in wheelchair sports. A two-step approach to predicting the changing set-up effects for wheelchair rugby was developed, consisting of (i) predicting the participant's propulsion kinematics during a maximal effort 5m sprint using a linkage model; and (ii) development of principal component and partial least-squares regression relationships between wheelchair set-up, propulsion kinematics, and performance. Eight elite wheelchair rugby players completed testing in nine wheelchair set-ups, with seat height, seat depth, seat angle and tyre pressure altered and propulsion kinematics (contact and release angles) measured during the sprint. Accuracy was assessed through comparison of predicted and experimental propulsion kinematics (degree differences) and performance times (seconds differences). Results show good accuracy for kinematic measures, particularly for contact angles, with mean prediction errors less than 5° for 43 of 48 predictions. Performance predictions matched on-court results well for some participants, while others showed weaker prediction accuracy. Further work is required to account for individual impairments and propulsion approaches and develop strong predictors of sprint performance with limited player testing.

Keywords

Paralympic sport; wheelchair propulsion; regression; wheelchair configuration; propulsion kinematics

8.3 Introduction

Current procedures for prescribing wheelchair set-up parameters such as seat height and seat angle are limited in wheelchair sport, relying on previous coach and player experience [1], optimising parameters in isolation [2-4], or requiring substantial amounts of testing [5, 6]. These issues stem from difficulties in: monitoring on-court performance, where inertial measurement units (IMUs) only recently providing a reliable solution [7-10]; the substantial cost associated with wheelchair purchase (often $5,000-$10,000USD); adjusting wheelchair set-ups on current wheelchairs; and optimisation varying for individual players, where a greater focus on individual impairments can potentially improve the ability to achieve near optimal set-ups quickly.

In wheelchair rugby (WCR), players are assigned point classification scores ranging from 0.5-3.5 points depending on their sport specific activity limitation where a lower score indicates greater limitation. The classification process considers trunk, arm, and hand function (where 'function' includes strength, range of motion and co-ordination [11]) and hence players with varying impairment types (i.e., impaired muscle power – potentially due to spinal cord injuries (SCI) – or limb deficiencies) can be assigned equal classification scores. Optimising wheelchair set-up based on classifications is therefore not viable, as players will vary substantially even within a single point score [6, 12]. Hence methods that are able to provide detailed quantitative insight into the effects of specific set-up parameters on performance factors while reducing the amount of time and effort of on-court player testing are desired.

Ideally, on-court testing should be used for optimising wheelchair configurations, where athletes can be tested under conditions representative of competition demands [13]. This testing can then reveal significant differences in performance for set-up parameters such as wheel camber angle [3], seat angle and depth [6], and even glove type [14]. Slight improvements in performance can have large impacts on on-court results, with the difference between executing or missing blocks on opposition dependent on a number of centimetres [11]. To assess these impacts, improved sensor design and processing methods has enabled IMUs to provide detailed information such as position tracking [7-9], peak accelerations [15] and rotational velocities assessments [6, 9], and (when combined with high speed video) faster assessment of propulsion kinematics [11]. However, even with these recent developments, on-court assessments require set-up parameters to be investigated in isolation or with substantial amounts of testing. When assessing the impact of changing wheelchair parameters such as seat height, or seat angle, it is crucial to consider the affect each parameter has on critical on-court performance measures. This includes the player's ability to perform repeated maximal effort sprints from stationary positions, quickly and effectively change directions, ball-handling skills and chair stability [9, 16]. Optimising parameters in isolation requires detailed understanding of the interaction of different parameters, and large time commitments from players and support staff. Robust test design approaches have been investigated to reduce time requirements whilst maintaining the ability to assess effects of individual parameter settings [5, 6]. While using a robust design approach substantially decreases time requirements, it still requires a number of hours to test four parameters (seat height, seat depth, seat angle, tyre pressure) at three levels – with additional time to complete analysis and follow-up testing. Further developments are therefore desired in maximising efficiency in optimising wheelchair set-ups at an individual level: propulsion modelling provides a potential method to achieve this.

The majority of existing wheelchair propulsion modelling has focussed on musculoskeletal models attempting to quantify shoulder loads in daily propulsion to assess or reduce the likelihood of shoulder injuries [17-21]. This is clearly a crucial area for improving the well-being of wheelchair users, but it is unable to address performance aspects such as sprint or agility times. Due to the complexity of musculoskeletal models, individual representation of anthropometrics and muscular function is also a time-consuming and difficult process. To address this, a linkage model has previously been developed that is able to predict changes in propulsion kinematics (contact and release positions) for changing seat height and seat depth during daily propulsion [22, 23]. This method – despite not accounting for individual muscular function – appears to be a more realistic solution for optimising wheelchair set-up for performance due to the reduced time requirements and ease of adjusting for individual players. These considerations are important given a WCR squad regularly has more than ten players, with vastly different anthropometrics, levels of muscular function, and propulsion approaches [11]. However, assessing the relationship between propulsion kinematics and on-court performance measure is difficult, particularly when this varies across players.

The development of regression approaches, such as partial least squares (PLS) and principal component analysis (PCA), provide a potential method for quantifying the relationship between wheelchair set-up, propulsion kinematics, and performance. These regression approaches consider a number of predictor variables (such as wheelchair set-up, or propulsion kinematics) and construct new predictor variables or components. These predictor components can then be used to estimate performance factors such as sprint time. Whilst these two regression approaches operate in a similar way, the construction of the predictor components differ. For PCA, components are created to explain the observed variability without considering the response variable, whereas PLS accounts for the response variable during this construction [24]. These approaches have been used across a range of areas, including the form of running shoes and emotional reaction of consumers [24], pelvic shape prediction [25], determination of sport rock climbing performance [26], and technique analysis in sports [27, 28].

The aim of the current study, therefore, was to investigate PCA and PLS methods in predicting sprint performance based on individual wheelchair set-up and predicted propulsion approaches. To achieve this, an additional aim was to assess the prediction accuracies of propulsion kinematics of a linkage model in comparison against known propulsion kinematics. This method will implement a linkage model to predict alterations in propulsion kinematics with changing wheelchair set-up for elite WCR players, and then utilise PLS and PCA approaches to predict the effect this has on sprint performance. It was expected that the prediction of propulsion kinematics would be more reliable for players with lower movement variability throughout their propulsion approach, and this will aid the regression approaches in prediction of performance.

8.4 Method

8.4.1 Participants

Eight elite WCR players were recruited and provided informed, written consent before completing testing. Individual player details are summarised in Table 8-1.

8.4.2 Participant Testing

Testing consisted of an orthogonal design approach using an adjustable wheelchair. The adjustable wheelchair was designed as an offensive wheelchair for wheels of 25-inch diameter and camber angle of 16 degrees. It was capable of adjusting parameters associated with seat and footplate position, with a mass of 14kg. It allowed for individual strapping approaches, and for players to use their own wheels throughout testing. This allowed for the variation of four set-up parameters (seat height, seat depth, seat angle, and tyre pressure) at three levels (player's current level, an increase, and a decrease) using an L9 orthogonal array. Players then completed sprint, agility, and ball-handling tests in each set-up while monitoring performance measures, propulsion kinematics and mobility measures. For more details on testing implementation and analysis, see Haydon et al. [6]. For this work, the propulsion kinematics (contact and release angles) and performance time for the 5m sprints, along with the set-up information, was utilised. Additionally, as seat angle has previously been linked to trunk motion [4], trunk angles at contact and release for each of the first three strokes were investigated for the various seat angle levels. Angles were calculated from video data (120Hz, Go Pro Hero 3+, California, U.S.) that was analysed as part of a custom MATLAB (R2016a) script by selecting an approximate hip position, a point superior to this in the video frame, and the acromion. A flexed trunk position was defined as a positive trunk angle. These results were then used for each individual's trunk angle input in the linkage model, depending on the seat angle level.

Table 8-1: Player information, including impairment, classification, and experience information.

Player	Impairment	Classification Score	International Experience (years)	Contact Prediction Method
1	Limb Deficiency	3.5	14	Altered
2	Limb Deficiency	3.5	6	Original
3	Limb Deficiency	3.5	3	Altered
4	Impaired muscle power	2.0	3	Original
5	Limb Deficiency	2.0	1	Altered
6	Impaired muscle power	2.0	10	Original
7	Impaired muscle power	2.0	12	Original
8	Impaired muscle power	1.0	8	Original

8.4.3 Modelling

Performance predictions for various wheelchair set-ups from on-court testing results occurs in two main steps: (i) predicting propulsion changes when altering wheelchair set-up, and (ii) predicting performance for inputs of wheelchair set-up and propulsion kinematics. Step (ii) relies on propulsion predictions inputs from step (i) and regression equations developed from on-court testing to predict the performance measure of sprint time. The outline of this procedure is displayed in Figure 8.1 and is detailed in the following sections.

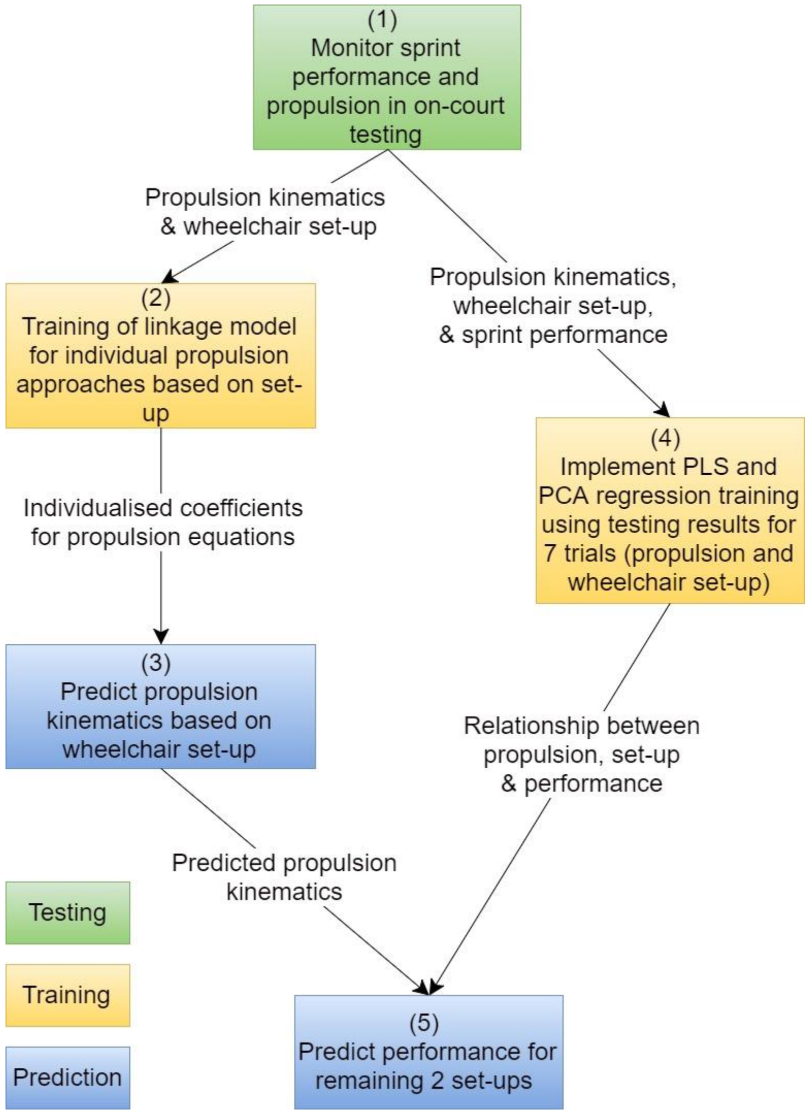

Figure 8.1: Outline of the procedure from on-court testing to performance prediction.

8.4.4 Propulsion Predictions

A sub-maximal linkage model [22, 23] was implemented that calculated contact and release angles based on individual anthropometrics and chair set-up. In advancing previous models to accurately predict maximal effort propulsion [4], the model included an additional trunk segment with motion fixed about the hip position. The equations for contact and release were derived to use shoulder position based on the trunk angle at contact and release. Trunk angular velocity (i.e., progression from contact to release angle) was assumed to be constant throughout the stroke phase. The assumption of contact occurring when the hand is perpendicular to the tangent of the wheel [22] was not valid for some players due to various propulsion techniques as seen in Figure 2. Players with greater trunk range of motion (i.e. in this participant group, players with limb impairments) generally utilised an approach with a greater proportion of 'push' (see Haydon et al. [11]). This approach requires the trunk to be in a flexed position at contact, and the forearm segment approximately parallel to the wheel tangent. For these players (detailed in Table 8-1 as *Altered*), a 90-degree addition was included for the prediction of the contact angle (Equation 8- 1).

Equation 8-1: $$\theta_c = \beta\left(\tan^{-1}\left(\frac{X_{hs}-L_{ua}\sin\theta_{TI}+L_{fa}\sin(90°-\theta_{TI})}{Y_{hs}-L_{ua}\cos\theta_{TI}+L_{fa}\cos(90°-\theta_{TI})}\right)\right)$$

Where β is a contact coefficient varied from 0.5 to 1.5 (see below for more details); θ_c is the contact angle; X_{hs} is the horizontal position of the shoulder relative to the wheel axle; Y_{hs} is the vertical position of the shoulder relative to the wheel axle; L_{ua} and L_{fa} are the length of the upper arm and forearm respectively; and θ_{TI} is the initial trunk angle. This enabled the prediction of contact and release angles based on an individual's anthropometrics and chair parameters (seat height, seat depth, and seat angle). As mentioned above, the seat angle setting influenced the trunk position at contact and release for each of the first three strokes.

The contact coefficient accounts for individual propulsion approaches by adjusting the ratio of angle prediction around the assumptions of perpendicular or parallel forearm segments at contact. During analysis of the nine set-ups tested by an individual, a contact coefficient was determined (to two decimal places) for each of the first three strokes that minimised the error between measured and predicted angles from the above equations. A contact coefficient for each of the first three strokes was then set for future predictions by averaging across the nine set-ups. A similar process to determine release angle coefficient for each of the three strokes using the prediction equation from previous work [22]. This approach not only accounts for differences across individuals, but also across the first three strokes within a sprint which have been shown to differ in accelerations from standstill [11]. Despite the potential asymmetry present in WCR propulsion [29], this process combined left and right propulsion kinematics to reduce the impact of any outliers in coefficient calculations.

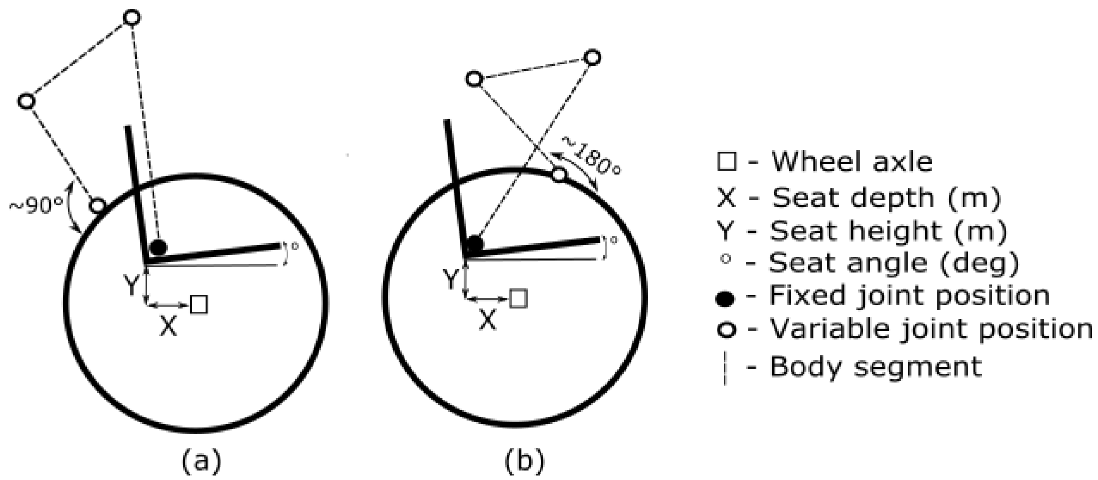

Figure 8.2: The propulsion model consisted of a trunk, upper arm, and forearm segments with a fixed hip position and variable seat height, seat depth, and seat angle. Contact angle estimation varied between the previous assumption of the forearm being perpendicular to the wheel tangent at contact (a), and a propulsion method where the forearm is close to parallel with the wheel tangent (b) at contact.

8.4.5 Performance Prediction

Partial least squares (PLS) and principal component analysis (PCA) training was implemented on the experimental data. These included thirteen input variables: seat height, seat depth, seat angle, tyre pressure, contact angles for the first three strokes, releases angle for the first three strokes, and the push angles for the first three strokes. These regression approaches were trained independently in Matlab (Mathworks, 2017b), with the first seven of the nine set-ups from experimental testing used to train the prediction methods (similar to typical training-test ratios of 70%-30% and 80%-20%). The performance of the prediction method was then assessed using the final two set-ups from experimental testing for each athlete. While the set-up parameter values (i.e., seat height, seat depth) were matched with those from experimental testing, the prediction approach was implemented using the predicted propulsion kinematics.

8.5 Results

Differences between measured and predicted contact and release angles are provided in Figure 3. Mean values show contact angles could be accurately predicted, with differences less than 0.5° for 18 of 24 (75%) contacts. However, while mean values provide the appearance of accuracy, the maximum differences between a measured and predicted contact angle varied by greater than 10° for 9 of 24 (37.5%) of contacts. Further, mean release angle prediction differences increase during later strokes (pushes) after sprint start. Maximum differences were also greater for the release angles compared with contact angles for the majority of players.

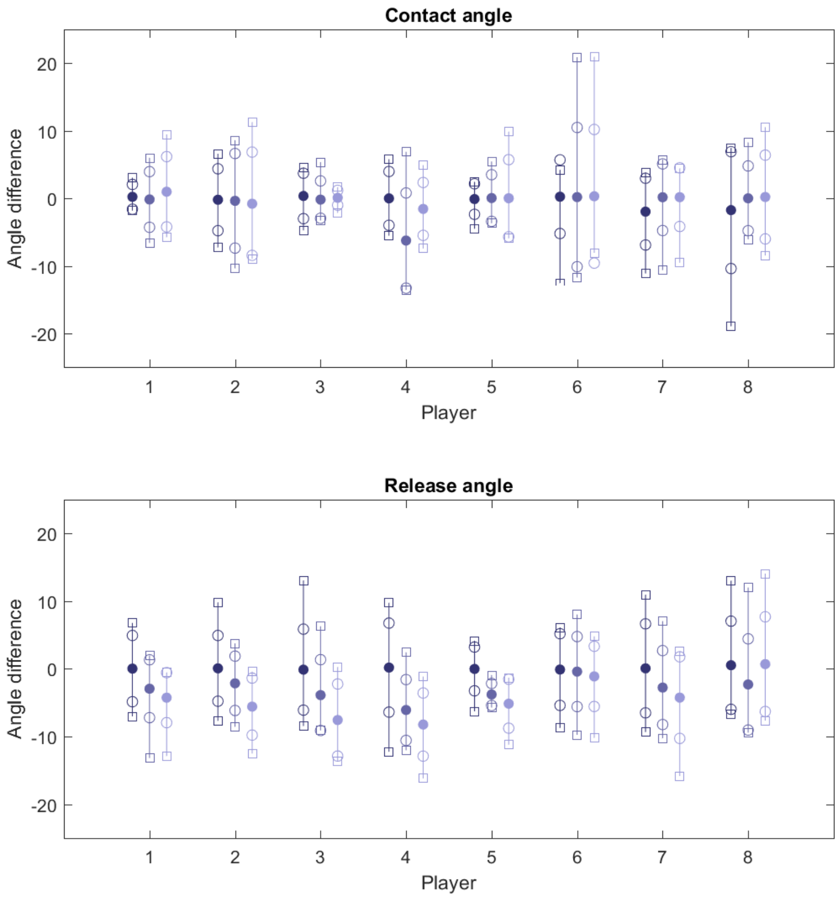

Figure 8.3: Contact and release angle prediction differences from testing results. The first three strokes for each player is presented on individual bars, with each bar containing the mean difference (filled circle), the standard deviation (open circle), and minimum and maximum differences from testing results (open squares).

Sprint time predictions were calculated for chair set-up parameters and predicted propulsion angles using both PLS and PCA regression approaches. Comparisons with actual (recorded) sprint time for the two set-ups that were not included in training the regression approaches are presented in Figure 4. The accuracy of the regression models varies between players and approaches: Player 4 results were predicted within 0.01 seconds for both set-ups, while Player 7 had variations of 0.21 and 0.59 seconds for the PLS and PCA methods, respectively.

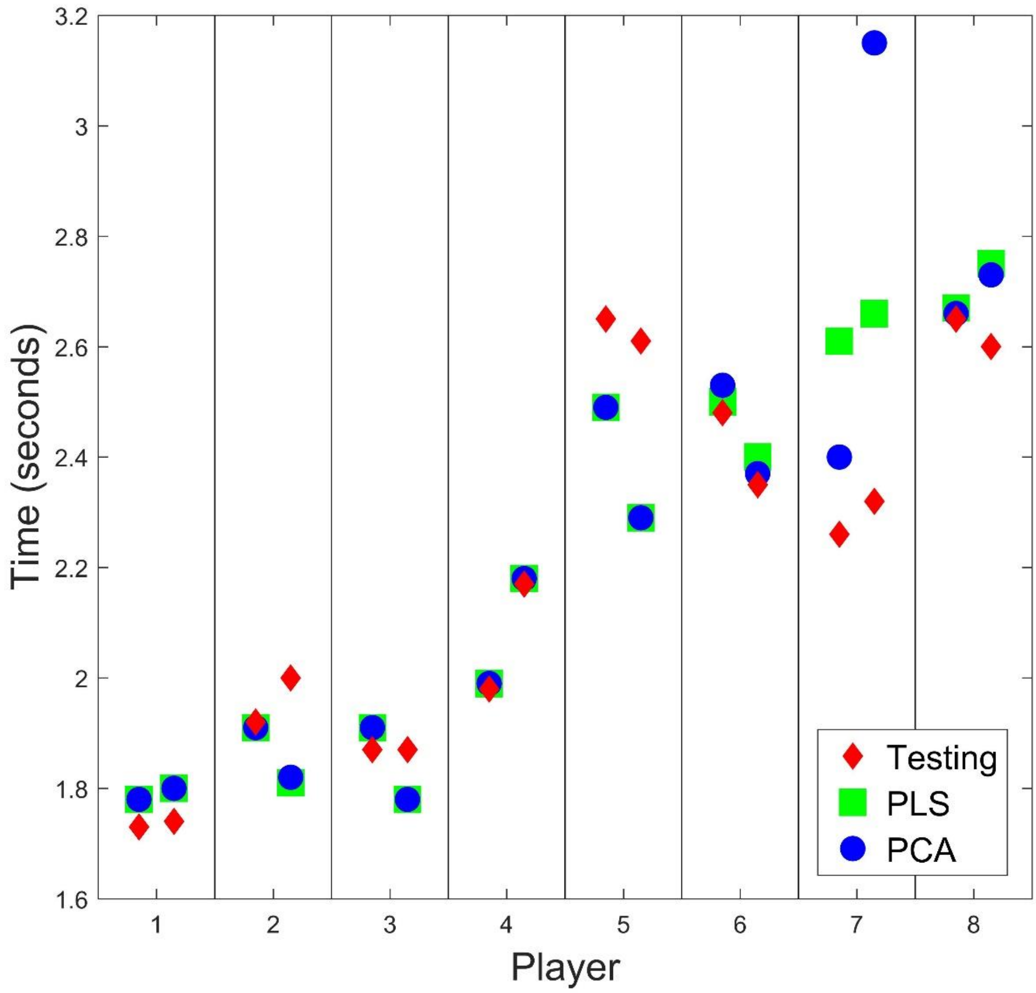

Figure 8.4: Comparison of sprint times from testing and the two regression approaches for all players.

8.6 Discussion

Modelling of wheelchair propulsion has the potential to decrease the amount of testing required whilst maintaining the ability to detect changes in propulsion and performance. This study investigated the ability of a linkage model to predict propulsion kinematics for a range of WCR players and use these results to predict performance using PLS and PCA regression analysis. On-court testing allowed propulsion kinematics and performance to be assessed across nine set-ups when using an adjustable wheelchair. All nine set-ups were used to develop predictions for contact and release angles for the first three strokes based on wheelchair set-up parameters, and seven set-ups used to train the regression approaches. Propulsion predictions and regression approaches were then used to predict performance in the final two set-ups.

Mean values for contact angle predictions show good agreement with on-court testing results; however, maximum differences for each player can vary substantially. These large differences likely occur due to the assumption that players attempt to employ the same propulsion technique regardless of wheelchair set-up – evident by using an average coefficient from all nine set-ups. If players alter propulsion approaches with altering set-ups, it is not accounted for using an average coefficient and may therefore result in large differences. Release angle mean prediction error appears to increase with stroke number following a sprint start for most of the players. For stroke one, mean prediction error is less than 0.51° for all players and less than 0.16° for 7 of 8. However, for the third stroke, only 2 players have an absolute mean prediction error less than 4.27° with a maximum of 8.25°. This likely occurs as the magnitudes or release angles are typically larger than those of the contact angles (i.e., contact angles can vary from -45° to +15°, compared with release angles which often vary from +70° to +105°) [11]. Using an average coefficient in the calculation is therefore troublesome as slight changes to propulsion technique result in larger differences in the predicted release angle. This hypothesis is supported by Player 8 having the smallest error for release angle estimation for the third stroke, as this player displayed the smallest release angles and therefore variations in coefficient value had less effect on the magnitude of the error. Additionally, inclusion of strength or impairment testing would provide greater detail on player capabilities, improving in particular the trunk motion predictions and likely performance estimates.

Regression results varied between players – good agreement was seen with testing results for some players (1, 4, 6) and inconsistent results for others (Players 2, 3, 7). Player 4 results display the most potential for continued use of this approach. Despite large differences in experimental performance time in set-ups eight and nine on-court, these changes in performance are predicted within 0.01 seconds by both regression models. This is likely influenced by a consistent relationship between wheelchair set-up, propulsion kinematics, and performance. These relationships refer to the influence changing parameters has on sprint time: in a consistent relationship, increasing contact angle is likely to have the same effect on sprint time in all set-ups. The development of this relationship occurs in the regression training (on the first seven set-ups), with the impact of wheelchair set-up and propulsion likely consistent in the tested (final two) set-ups. Although performance times for Players 1 and 6 don't match as accurately, the trend is of comparable magnitude and direction. As this approach is proposed as a method to assess the effect of various wheelchair set-ups, the ability to detect changes in performance is critical. Players 2 and 3 show occasions where both regression models were poor in predicting changes in performance. Both PLS and PCA regression approaches predicted improved performance for Player 2's set-up nine, but decreased performance was evident in on-court testing. Similarly, Player 3 had similar performance in set-ups eight and nine, but regression predictions expected performance to vary by 0.13 seconds. These prediction variations likely relate to regression training approaches not aligning with the relationships for tested set-ups. Greater variation in these relationships (i.e., increasing contact angle does not consistently improve/decrease sprint performance) makes performance predictions difficult; this training phase can be improved by including greater amounts of relevant data, however this is often difficult to achieve in practice.

Player 7 modelling displayed lower accuracy for performance prediction– whilst PLS regression displayed changes in performance of similar magnitude and direction (albeit 0.30 seconds difference from on-court results), PCA regression over-estimated sprint time by 0.80 seconds for set-up nine. This discrepancy in PLS and PCA predictions signifies the difference in regression prediction methods – PLS considering the response variable in the construction of predictor components [24]. Due to the relatively limited number of variables used in this approach, consideration of the response variable is likely required for accurate prediction.

This wheelchair prescription method relies on two distinct sections of prediction for changing wheelchair set-ups: (i) propulsion kinematics and (ii) sprint time performance. Propulsion kinematics were predicted based on a linkage model, with fixation about the hip an extension on previous models [22, 23]. Assessment of maximal effort propulsion from standstill in WCR requires consideration of trunk motion – due to trunk motion accompanying force generation [4] – and player specific approaches due to the substantial variations across classifications [11]. The PLS and PCA regression approaches can then be trained using on-court testing to produce a prediction method based on inputs of wheelchair configuration and propulsion kinematics – allowing a greater number of potential set-ups to be investigated for players with reduced amounts of on-court testing. After completing on-court testing, this modelling approach can be implemented by team support staff or biomechanists to identify set-ups of interest. These set-ups could be replicated on-court to confirm findings, giving the player more detailed information on the effect of altering their wheelchair set-up. This improves upon current implemented approaches, where small adjustments to wheelchair parameters are often made over long periods of time, which can result in players only achieving set-ups they are comfortable with (and are nearer to optimal for performance) after many years in the sport.

The linkage model used in this study appears the best approach to predicting propulsion measures due to the reduction in time for processing and relative ease of individualising compared with musculoskeletal models. Whilst musculoskeletal models can potentially account for specific muscle functions of an individual and perform more detailed propulsion assessment through incorporation of three-dimensional motion throughout multiple strokes [19], this is likely impractical for the range of players across a squad in WCR (typically 10-12 players). Individual customisation of the musculoskeletal models would require further processing time and more detailed on-court testing assessment including motion capture and electromyography. Additionally, the linkage model has previously estimated torque and power at the shoulder joint throughout push motion [22] – however more research is required to validate these estimations in a practical setting.

Currently, this approach requires two- to three-hours of on-court testing with various set-ups for each individual. With further progression of this method, there is the possibility to substantially reduce the amount of on-court testing required. This progression relies on increasing the number of players and therefore data on how particular classifications and impairments respond to changes in wheelchair set-up. For players of similar impairments and anthropometry there is a greater likelihood their response to changing set-ups will be similar. As regression approaches require increases in data to build their relationships improve reliability, international collaborations are recommended to increase the pool of elite wheelchair sport athletes.

8.7 Conclusion

The process of wheelchair prescription is currently a time-consuming process that relies on player and coach experience. This study presents a method to predict propulsion kinematics based on changing wheelchair set-ups for maximal effort sprinting. Regression approaches (PLS and PCA) can be trained using on-court testing results, and then applied with the propulsion predictions to estimate sprinting performance for WCR. Results found that the assumption of a consistent propulsion approach may not be appropriate, particularly for release angles. Regression approaches were inconsistent in their ability to accurately predict performance changes. Player 4 performance was predicted almost exactly despite the large variations present. However, other results were unable to achieve the same accuracy, likely due to errors in the propulsion predictions. Additionally, PLS appears to be better suited for this type of analysis as it considers the response variable when constructing components. This method shows potential to improve the process of wheelchair optimisation, although the accuracy of this method would be improved with increased data from players with similar impairments and activity limitation.

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Chapter 9: Summary and Future Work

This section summarises the findings from this research, the implications and contribution, and future work.

This research aimed to improve knowledge of the impact wheelchair configurations have on propulsion and performance using an individualised approach in WCR. Achievement of these aims can contribute to an individual optimising their wheelchair configuration earlier in their sporting career, and modelling to potentially reduce the testing time required. This section reports on the process taken to achieve the aim of the research, the contributions to literature in this area and future directions of research.

9.1 Discussion

To improve upon on the current process of wheelchair configuration in WCR – which currently relies upon trial-and-error approaches and experiential knowledge [17] – an understanding of current wheelchair configurations and views amongst elite players was needed. The current reliance on subjective approaches for individual players was expected to be an inefficient method to achieving optimal wheelchair configurations. However, little research had previously reported on current configurations across elite players or the propulsion approaches used in representative testing protocols. This lack of research may be attributed to lack of access to elite populations, or chair configuration and propulsion approaches being highly individualised based on player activity limitation (and for configurations, the player's on-court role); however, knowledge surrounding trends in these areas among elite players has the potential to identify consistent approaches across similar impairments and severities, or alternatively promote further research into potential reasoning behind similarities or differences. The collection and analysis of this data also acted as the base for this research, with more detailed assessments into propulsion approaches (such as intra-stroke profiling) and the effect of configuration on performance measures used for individual optimisations.

To add to the previous qualitative assessments of configuration approaches and effects [12, 23], this work consisted of a quantitative assessment across an entire elite squad. Expectations from players on the effect of configuration changes on performance factors were also reported, providing a greater representation of views within an elite squad. While some of this knowledge is likely available within the WCR community of coaches, experienced players, and support staff, reporting these details allows for greater clarity about potential trends based on previous player experiences. As reported in Mason et al. [17], players' configurations are often based around subjective trial-and-error approaches. The initial attempted configurations used are influenced by the knowledge or experiences of the player or practitioner involved, rather than a systematic or quantitative optimisation process. This then requires time and often multiple wheelchairs for the player's configuration to be refined to a near-optimal set-up. Information gained through this work supplemented experiential coach knowledge, as has previously been recommended [8], and provided quantitative insight into configurations for specific classification groups and impairment levels.

A detailed assessment was also performed for propulsion approaches across a large population (n=25), many of which were elite players. This addressed a gap in the literature detailing propulsion kinematics and the lack of propulsion kinematics reported in WCR, particularly for maximal acceleration from standstill [24], and a statistical assessment of how propulsion can potentially be altered to improve peak accelerations. This identified that high-point players contacted the wheel closer to TDC of the wheel and released further around the wheel than other classification groups. Even when accounting for the difference in propulsion styles across classification groups, a linear mixed-effects model found a significant increase in peak acceleration was expected for larger release angles and smaller stroke angles in the third stroke. In general, these findings suggest players should contact closer to TDC of the wheel to maximise the proportion of the stroke spent in the 'push' phase. However, intra-stroke acceleration profiling revealed this is likely not the case for all players. Whilst the mid- and high-point player in the intra-stroke analysis showed peak accelerations occurring past TDC (particularly for strokes two and three), the low-point player did not, with peak accelerations occurring during the 'pull' phase of strokes two and three. While it was shown in Chapter 7 and 8 that propulsion strokes are affected by configuration, the change in propulsion kinematics is expected to be relatively small. It is therefore expected that the trends seen in intra-stroke profiling, such as the elbow angle when peak accelerations occur, will remain relatively consistent for an individual. For the low-point player presented, major changes to their propulsion stroke would not be recommended; rather, changes to chair parameters that can maximise the stronger regions of the stroke should be investigated. In this case, slight increases to seat depth or seat angle may result in more effective propulsion. A limitation of the statistical model implemented across the elite participant group was the classification groupings. As presented in Chapter 5, the relationship between increased release angles and increased peak accelerations was likely influenced by the high-point group. The propulsion approaches investigated differed substantially between 3.0- and 3.5-point players, potentially due to the varying types of impairment in the participant group (impaired muscle power – typically from SCI – and limb deficiencies, respectively). As 3.5-point players achieved higher peak accelerations in sprints while using larger release angles than 3.0-point players, this influenced the statistical model results. This supports the recommendation of statistical group assessments to either be developed for specific impairment types and severities, or to account for these factors during the modelling approach.

In addition to the improved propulsion assessments possible when using more detailed propulsion assessments, this approach also has substantial benefits when considering the impact of parameter settings. Assessments can provide additional insight into how the parameter influences linear propulsion, and hence allows for greater understanding of how findings will translate to on-court activity. In this work, the detailed assessments of wheelchair propulsion from Chapter 5 were continued through the configuration testing using a robust design approach (assessment of propulsion kinematics and peak accelerations). Used in conjunction with assessments of performance factors such as sprinting, mobility, and ball handling, this can substantially improve the configuration assessments possible. Utilising this knowledge, as well as robust design approaches with an adjustable wheelchair, can improve the process of accessing a player's initial wheelchair set-up and potentially prescribing a near-optimal position.

In practice, players and practitioners can utilise this approach with either elite or development players. The process has been shown to be successful for elite players, with 50% of those who completed testing preferring the recommended set-up over their current set-up in the adjustable WCR chair. Of the other 50% players often displayed similar or improved performance measures between chair set-ups, and 'preference' comments often related to perceived levels of comfort – potentially related to their experience and familiarity in the current set-up. The implemented process requires detailed assessments of propulsion and mobility affects, with coach involvement to ensure emphasis is placed on the desired variables for the player's on-court role. For development players, implementing a robust design approach would allow them to achieve a near-optimal configuration much earlier in their development than previously. Throughout their career, an individual is also likely to undergo physical development and this approach can provide a more time efficient approach in finding near optimal configurations – particularly in the lead-up to key phases in the sporting cycle. This has benefits in terms of their individual performance, but also has the ability to improve team performance and increase quality and depth of players involved in the sport.

9.2 Limitations and Future Work

While the work presented in this research provided new insights into a range of areas (e.g., quantitative assessments across elite players, intra-stroke profiling, configuration testing and analysis, and prediction modelling), there remains a number of areas that require further work.

One of the major difficulties within para-sport research is the large variations in levels of physical activity limitation, particularly when considering team sports such as WCR. Throughout this work, the importance of individual assessments has been emphasised. However, published research typically remains focused on the ability to achieve statistical significance which often relies on group analysis. As evidenced by Chapter 6, group analysis can mask important individual findings. In research settings, there is therefore the difficulty of attempting to achieve statistical significance whilst maintaining clear individual outcomes. Future work can improve these possibilities through increasing the sample size. Although difficult due to the limited population size of WCR players (especially at an elite level), larger participant groups would allow for assessment of a greater number of players with similar impairment types and severities. This would increase the likelihood of (although not guarantee) consistent player trends or adaptations to particular configuration changes. The increase in size of participant groups could be achieved through international collaborations, or inclusion of national level players. Alternatively, case-study approaches or alternate analysis methods (e.g., radar plots of different statistical methods) can provide detailed insight at individual or small group levels.

Increases in sample size (or available data) also has the potential to improve modelling capabilities. More data for players of similar impairments and severities would improve the training capabilities of PLS and PCA regression approaches, resulting in more reliable prediction methods. Developing the model such that it has a number of levels of impairment and severity options can allow for initial assessments of an individual's propulsion approach to be followed by an appropriate prediction approach for various set-ups.

Using such an approach would also rely upon the standardisation of wheelchair configuration parameters to an individual. Standardisation approaches have previously been recommended [17]; however, they were not reported in this work outside of the quantitative assessments of configurations. Standardisation approaches include measures such as elbow angle when the hands are placed at TDC of the wheel and seat depth to thigh length ratio. The decision to not report standardisation measures was made as testing was performed around the player's current parameter settings. As elite players were involved in testing, it was assumed optimal configurations would be close to the player's current setting (as reported in Chapter 7, changes were typically ±10mm or ±5°). These small changes were expected to have little effect on the magnitude of standardised values. However, implementing standardised measures into a modelling approach would allow comparison of similar impairment severities for players of various anthropometrics. Due to the testing of elite players, additional considerations such as minimising the testing time and load were important, as well as ensuring minimal impact on preparation for competitions.

Other modelling improvements include the assessment of independent left and right contact and release positions. This was not included in the current analysis due to the limited data available and the need for as much data as possible to develop the regression equations. Inclusion of potential asymmetries in modelling predictions allows for a more thorough representation of propulsion, as well as improving regression predictions by accounting for asymmetries.

Assessments of propulsion kinematics throughout this work were completed by synchronising video recordings from multiple views. This was deemed appropriate as they were monitored during linear sprinting where cameras were perpendicular to the plane of motion of the player. The focus of hand position on the wheel at contact and release also allowed assessments using video to be used. For more detailed assessments, three-dimensional analysis options (such as motion capture) should be investigated. This would allow features such as shoulder angles and elbow angles to be monitored in more detail. Not only would this provide information on angles throughout the StrokeTime, it could potentially allow for assessment of hand recovery patterns – similar to the work of Boninger et al. [25].

In addition, use of motion capture options would allow for monitoring of propulsion in various conditions such as the turn and sprint implemented in Chapter 6. This would ensure that propulsion kinematics can be accurately monitored during the initial strokes of such testing protocols. Additionally, various manoeuvring approaches could be assessed for both turning from a standstill and weave patterns typical of match play. Motion capture investigations require greater time commitments during set-up, including calibration which needs to be maintained throughout testing. This was deemed difficult to employ under testing constraints outlined here, where time with the participant group was limited and ability to ensure the monitoring equipment (cameras, etc.) remained undisturbed throughout testing limited. Aligning with increased monitoring of propulsion, measurement of contact forces would be a beneficial addition. Acceleration profiles presented here provide a good indication of the outcome measure (wheelchair motion), whereas contact forces with the wheel would provide insight into the energy exerted onto the wheels. Measurements such as the fraction of effective force [26] could then be investigated to provide a measure of propulsion efficiency. This can help in assessments of configuration settings, with more effective positions expected to result in more efficient propulsion. Increased efficiency would also provide benefits in terms of the physiological demands (e.g., repeated sprints) throughout games [1], with less effort required for similar motion in efficient configurations.

It also recommended to continue to investigate representative test designs that are able to replicate common on-court activities. The example in this work of a catch, pass, turn, and sprint is performed regularly under match conditions. This could involve tests involving blocking and contact which are important factors in WCR performance [27]. This improves translation of findings from testing to on-court performance. In future, an assessment protocol similar to the WMP developed by de Witte et al. [7] for WCB could be developed for WCR to account for the slight variations between the two sports (e.g., executing/escaping blocks and impacts, ball carrying variations). This would standardise testing approaches to make findings more transferrable and replicable.

Finally, this work only investigated a small number of possible wheelchair configuration parameters. The parameters selected for testing (seat height, seat depth, seat angle, and tyre pressure) predominantly surrounded the seat position (excluding tyre pressure) in relation to wheel axle. These selections were made as these parameters were expected to have the large influences on both performance and propulsion. Findings supported this expectation, with seat depth and seat angle in particular showing large impacts for multiple players throughout configuration testing. Tyre pressure was selected as it had received limited attention in previous research despite potential impacts on factors including frictional resistance. The design of an adjustable chair specifically for WCR facilitated this testing, with the expectation that such chairs are likely to increase in their availability. The inclusion of such testing options with chair manufacturers in the prescription process would substantially improve the ability of players to assess various wheelchair set-ups before ordering. These assessments can also involve parameters such as wheel diameter, camber angle, and backrest height, which are all expected to affect performance measures. Future work would continue to add to this work and that of van der Slikke et al. [3] and Mason et al. [15] among others to assess the impact of various wheelchair configuration parameters on performance in representative test designs. Additionally, longitudinal assessments of player adaptations to new wheelchairs would be greatly beneficial – in terms of method of adjustments, and associated length of time before peak performance is achieved. This information would allow support staff to schedule new wheelchair prescriptions to optimise performance at major competitions.

9.3 Contribution and Conclusion

The work throughout this thesis has addressed a number of gaps in published literature. These included: greater assessments of configuration and propulsion approaches across an entire squad; more detailed assessments of propulsion approaches, including intra-stroke acceleration profiling and the impact of test design; effect of configuration parameters on propulsion and performance, as well as using prediction approaches to reduce the amount of player testing required.

Configuration and propulsion assessment across an entire elite squad provide greater detail on current practises within WCR. The assessment of configuration differences across classifications groups provides detail on how players' configurations currently vary based on point score. This does not account for the various impairment types or severities within classification groups but does provide the first quantitative assessment across an entire squad. Similarly, propulsion approaches for maximal acceleration from standstill had not previously been reported. Comparisons across classification groups aided in assessing the various propulsion approaches, while linear mixed-effects modelling suggests that increasing release angles and reducing stroke angles can increase peak accelerations, particularly for the third stroke. This information not only allows for improved knowledge of how propulsion approaches vary across players, but also potential methods on how players can improve their propulsion stroke in terms of hand location at or timing of peak accelerations.

Assessments of propulsion in WCR were further extended to consider the intra-stroke acceleration profiles for three case-studies. Intra-stroke profiles had received attention in wheelchair racing [28], however no investigations had focussed on this in WCR or wheelchair court sports. This work provided a method to consider the intra-stroke profiles, as well as providing detail on key features that likely affect sprinting performance. Clear variations in profiles and peak magnitudes were evident across players, with these then related to hand position on the wheel and components of push and pull within each stroke. For the athletes presented, this reiterated that propulsion magnitudes increased with larger release angles. The method presented also allowed for consideration of asymmetries for each individual, with this being a recent focus of other studies into sprint performance in WCR [2]. This approach can also be implemented when investigating configuration changes, where intra-stroke profiles and peak magnitudes will vary depending on the player's strength and muscular function and the relative position the wheelchair configuration places them in.

The impact of test design details the impact small changes to testing protocols can have on the results achieved. This supports careful consideration of test designs to ensure that findings translate to on-court performance. The ability to monitor and identify variables that are important to on-court performance has recently improved [14]; hence researchers should focus on test designs that better replicate these variables.

Following the work on propulsion assessment, greater understanding improved the ability to determine configuration parameter effects on performance. This involved implementing a method that allowed for quantitative assessment of a range of performance factors such as acceleration from standstill, agility, and ball handling [12] as well as input from players and coaches on the perception of each set-up on performance. This involved a robust design approach to assess performance factors including sprinting, agility movements, and ball handling, as well as propulsion changes in acceleration from standstill. Although individual case studies are required in this method, three of the six players who completed the protocol preferred the recommended setting over their current setting. Other players who completed testing showed similar or improved test performance, but preferred their previous set-up due to perceived levels of comfort. This is a positive result for this method; as all players were elite, it was expected that current configurations could already be relatively close to optimal. Therefore, implementing this approach with players new to the sport would likely result in finding a configuration closer to their optimal setting earlier in their development or at key milestones/phases during their career. This could also be important following strength and conditioning programs, where the physical strength of players may improve and alter the optimal configuration.

The final aspect of this work focussed on a prediction modelling approach to further reduce the required on-court testing. Encouraging results were evident for some players with changes in performance predicted accurately; however, changes in performance for other players were not well predicted. There remains potential for this method to be successful but requires improved learning mechanisms to be a reliable measure of performance.

With the range of propulsion assessments implemented and presented, clear contributions to the understanding of current propulsion approaches in elite WCR are evident. Importantly, the methods presented promote individual considerations of propulsion kinematics and intra-stroke profiles due to the wide range of activity limitations in WCR. These assessments were then included in the analysis of configuration parameters in the robust design approach. As this approach showed some elite players immediately preferred the recommended set-ups over their current set-up in an adjustable chair, and showed improved testing results, it appears successful. Therefore, the aims of this research to improve knowledge surrounding the effect changing configuration parameters on performance and propulsion have been achieved.

9.4 References

References used throughout Chapter 1 and Chapter 9. All other references are provided within the specific chapter.

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