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# Adapted sports: Wheeled-mobility, exercise and health


**Edited by**
Dirkjan Veeger, Riemer J. K. Vegter, Victoria Louise Goosey-Tolfrey and Christof A. Leicht



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# Adapted sports: Wheeled-mobility, exercise and health

**Topic editors**
Dirkjan Veeger — Delft University of Technology, Netherlands
Riemer J. K. Vegter — University Medical Center Groningen, Netherlands
Victoria Louise Goosey-Tolfrey — Loughborough University, United Kingdom
Christof A. Leicht — Loughborough University, United Kingdom

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EDITED BY
Reuben Escorpizo,
University of Vermont, United States

*CORRESPONDENCE
Riemer J.K. Vegter
r.j.k.vegter@umcg.nl

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# Editorial: Adapted sports: Wheeled-mobility, exercise and health

Riemer J.K. Vegter[1,2]*, DirkJan H.E.J. Veeger[3], Vicky L. Goosey-Tolfrey[2] and Christof A. Leicht[2]

[1]Center for Human Movement Sciences, University of Groningen, University Medical Center Groningen, Groningen, Netherlands, [2]Peter Harrison Centre for Disability Sport, School of Sport, Exercise & Health Sciences, Loughborough University, Loughborough, United Kingdom, [3]Department of Biomechanical Engineering, Delft University of Technology, Delft, Netherlands

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Editorial on the Research Topic
Adapted sports: wheeled-mobility, exercise and health

by Vegter RJK, Veeger DHEJ, Goosey-Tolfrey VL and Leicht CA. (2002) Front. Rehabilit. Sci. 3: 1015179. doi: 10.3389/fresc.2022.1015179

## Introduction

Persons that use a manual wheelchair depend on their upper body for daily mobility as well as for the sports they participate in. Numerous adapted sports exist. In some sports, modified wheelchairs are used for propulsion (e.g., wheelchair tennis, basketball, rugby and racing), others rely on other forms of cyclic upper-body exercise, like arm cranking movements (e.g., handcycling). Different adapted sports can also have an important impact on the upper body without a wheelchair involved, such as archery, paracanoe or swimming. What differentiates the abovementioned adapted sports from most able-bodied sports is the focus on the upper body for propulsion, which may result in different biomechanical and physiological responses when compared with the lower body.

The current Research Topic of Frontiers in Rehabilitation Sciences focuses on the performance and health aspects of participating in adapted sports and exercise for manual wheelchair users. 68 authors contributed to this special issue with 15 articles. They are spanning three broad topic areas:

(1) Shoulder-related responses and injuries resulting from upper-body exercise and wheelchair propulsion.
(2) Applied wheelchair sport research on wheelchair propulsion kinetics and kinematics.
(3) Elite sport and performance, considerations related to Paralympic sports specific classification.

## Shoulder-related responses and injuries resulting from upper-body exercise and wheelchair propulsion

The studies in this topic area put a spotlight on the shoulder as an anatomical structure that must cope with the very specific strains experienced during upper-body exercise, and more specifically, wheelchair propulsion. They provide new knowledge regarding activities that may be particularly stressful to the shoulder and surrounding structures and highlight factors that may help reduce strain, and as a result, minimise the risk for shoulder injury.

Bossuyt et al. investigated the acute shoulder tendon adaptations following maximal exercise in wheelchair rugby athletes. They provide evidence for exercise-mediated fluid inflow into the tendon, possibly because of the overload and acute inflammation. Arnet et al. demonstrated the benefits of fitness (assessed as anaerobic capacity) in the stabilisation of the shoulder during lifts, as it helped maintain the acromio-humeral distance following lifting. Aissaoui and Gagnon performed wheelchair propulsion training with haptic biofeedback with the aim to increase mechanical effectiveness and found the tangential push-rim force component to increase substantially, whilst also slightly increasing shoulder moments. Chénier et al. investigated sprinting with and without dribbling in wheelchair basketball, reporting higher speeds and shoulder loads when sprinting without dribbling. Finally, Mayrhuber et al. present a scoping review on shoulder injuries in wheelchair tennis. They identify possible risk factors as overhead movements, repetitive activation of the anterior muscle chain and internal rotators, as well as a higher spinal cord injury level.

## Applied wheelchair sports research on wheelchair propulsion kinetics and kinematics

Wheelchair sports and disability characteristics come in many shades, which results in a wide range of movement patterns. Kinetic and kinematic analysis allows to quantify impacts of equipment setup, practice and training interventions, with the aim to improve performance and avoid injury.

Three studies in this topic area investigate wheelchair-sport specific skills. Alberca et al. compared the impact of holding a badminton racket on wheelchair propulsion, reporting patterns associated with reduced propulsion effectiveness and higher injury risk. De Klerk et al. investigated wheelchair racing propulsion acquisition skills during three weeks of wheeling practice, reporting pronounced improvements in metabolic strain, push and cycle times. In a systematic review, Altman et al. identify tests for throwing maximal distance, throwing precision, and dribbling the ball to determine ball-handling proficiency in wheelchair sports.

A systematic review by Fritsch et al. outlines the methodologies used to study the impact of manual wheelchair configuration on biomechanical outcome measures. An applied example of a study using such methods is presented by Bakatschina et al., comparing kinematic variables between offensive and defensive wheelchair rugby wheelchairs in able-bodied participants. Perhaps surprisingly, they found that higher sprint velocities were achieved in defensive wheelchairs, indicating that that the higher performance observed in offensive vs. defensive wheelchair rugby players is a result of differences in disability, not wheelchair type. Staying within the sport of wheelchair rugby, Haydon et al. attempted to develop an algorithm to predict the impact of changing wheelchair setup on performance outcomes. Their on-court performance prediction was accurate for some, but less so for others, leading them to provide suggestions to improve accuracy further (e.g., inclusion of athlete activity limitations).

## Elite sports and performance, considerations related to Paralympic sports specific classification

Sound sports specific classification procedures are the basis for fair competition in Paralympic sports. Classification is an evolving field (as is the whole field of Paralympic sports), therefore adaptations to, or at times, completely new classification tests are required. Altman et al. investigated a test to determine arm coordination impairment (the spiral test). They found it useful and reliable to differentiate arm coordination impairment in people without impairment, making it a promising option for Paralympic classification.

The three other studies in the topic area of elite sport assess performance, and more specifically, how performance may be impacted by disability type and sport. Gee et al. provide an overview of the altered physiological response to exercise in disability and offer physiological considerations to benefit Paralympic performance, whilst highlighting research gaps. Gavel et al. address one of those gaps, namely the thermoregulatory response in National team wheelchair rugby players during international competition, relating thermal strain to movement time. Quittman et al. round this topic area off with a case report of a paratriathlete undergoing chronic myeloid leukaemia treatment, which dramatically reduced markers of physical capacity.

## Future perspectives

The articles in this special issue cover a range of approaches including experimental studies, systematic reviews, and a case study. Whilst broad in the topic areas covered, many provide evidence helping to maximise performance and/or minimise injury risk in sports suitable for manual wheelchair users. On the other hand, almost all presented articles show the continued need for research and are often only a starting point for gaining new knowledge about the multi-disciplinary impact of wheelchair sports on wheelchair users. Both the shown difficulty of measuring larger groups of participants, because of the relatively small and heterogeneous population, and the lack of strong longitudinal research designs hamper the level of evidence. One solution might be increased international collaboration between researchers, health and sports professionals, applying open science principles. Of course, at times this can be at odds with the competitive nature of top-level athletes. However, hesitancy to participate in such research may be overcome if overarching research questions are formulated with a goal to further professionalize adapted wheelchair sports as a whole. Buying into the research process by athletes, coaches and health professionals may further be facilitated by feeding back findings to the base, highlighting the relevance and applicability to the various stakeholders. In this context, the presentation of these 15 articles should be followed up with activities to engage lay audiences. Findings may be made palatable by presentations, summary videos or visual overviews aimed at specific target groups. Whatever the format, what unites the findings is their root in scientific principles. We are therefore grateful for this showcasing opportunity of the already high-quality research performed in this relative niche research area of adapted wheelchair sports—it certainly holds scope for further study.

## Author contributions

RV and CL drafted the editorial. All authors approved the submitted version.

## Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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REVIEW
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# Physiological Considerations to Support Podium Performance in Para-Athletes

*Cameron M. Gee[1], Melissa A. Lacroix[2], Trent Stellingwerff[1,3], Erica H. Gavel[2,4], Heather M. Logan-Sprenger[2,4] and Christopher R. West[3,5,6,7]**

*[1] Athletics Canada, Ottawa, ON, Canada, [2] Canadian Sport Institute-Ontario, Toronto, ON, Canada, [3] Canadian Sport Institute-Pacific, Victoria, BC, Canada, [4] Faculty of Health Science, Ontario Tech University, Oshawa, ON, Canada, [5] Faculty of Medicine, International Collaboration on Repair Discoveries, Vancouver, BC, Canada, [6] Department of Cellular and Physiological Sciences, University of British Columbia, Kelowna, BC, Canada, [7] Centre for Chronic Disease Prevention and Management, University of British Columbia, Kelowna, BC, Canada*

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**Edited by:**
*Christof A. Leicht, Loughborough University, United Kingdom*

**Reviewed by:**
*Kristin L. Jonvik, Norwegian School of Sport Sciences, Norway*
*Elizabeth Broad, Independent Researcher, Huskisson, Australia*

***Correspondence:**
*Christopher R. West*
*westchri@mail.ubc.ca*

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The twenty-first century has seen an increase in para-sport participation and the number of research publications on para-sport and the para-athlete. Unfortunately, the majority of publications are case reports/case series or study single impairment types in isolation. Indeed, an overview of how each International Paralympic Committee classifiable impairment type impact athlete physiology, health, and performance has not been forthcoming in the literature. This can make it challenging for practitioners to appropriately support para-athletes and implement evidence-based research in their daily practice. Moreover, the lack of a cohesive publication that reviews all classifiable impairment types through a physiological lens can make it challenging for researchers new to the field to gain an understanding of unique physiological challenges facing para-athletes and to appreciate the nuances of how various impairment types differentially impact para-athlete physiology. As such, the purpose of this review is to (1) summarize how International Paralympic Committee classifiable impairments alter the normal physiological responses to exercise; (2) provide an overview of "quick win" physiological interventions targeted toward specific para-athlete populations; (3) discuss unique practical considerations for the para-sport practitioner; (4) discuss research gaps and highlight areas for future research and innovation, and (5) provide suggestions for knowledge translation and knowledge sharing strategies to advance the field of para-sport research and its application by para-sport practitioners.

**Keywords: adapted sports, physical activity, disability, paralympics, exercise**

## INTRODUCTION

Health and exercise performance have traditionally been considered on a continuum in which the athlete, viewed as the epitome of human physiological performance, is at one end and individuals with disability, traditionally medicalized as a condition to be treated, at the other (1). The para-athlete and the Paralympic Games, encompassing the wider Paralympic Movement, have challenged such dogma. While athletes with disabilities were not unheard of (2), it was the Stoke Mandeville Games (established in 1948) and Paralympic Games (retroactively established in 1960) that have most rapidly advanced para-sport and the Movement (3).

The International Paralympic Committee (IPC; established 1989), recognized as the global governing body of the Movement (4), operates under the vision of enabling para-athletes to achieve sporting excellence. In doing so, the IPC has a classification code that governs the process by which athletes are categorized into a number of groups on the basis of common properties (5). The system aims to determine who is eligible to compete at the Paralympic Games, while ensuring that it is not the degree of impairment but sporting excellence that ultimately determines which athlete or team is victorious (6). Presently, there are 10 eligible impairment types: impaired muscle power, impaired passive range of movement, limb deficiency, ataxia (uncoordinated movement), athetosis (involuntary movements), hypertonia (increased muscle tension), short stature, leg length difference, vision impairment, and intellectual impairment (7).

Though research studies dating back to the 1970s and 1980s have documented exercise responses and the unique physiology of "active" individuals with an impairment, it is only in the last 30 years that research into the "para-athlete" has started to emerge as a field of its own. For example, a search of PubMed databases indicates that in 1990 there were 57 articles on "disability sport" and that by 2020 this number had increased more than twenty-fold. Despite the expansion in research and many excellent para-sport research groups globally there still exists a relative paucity of para-sport research and evidence-based practice—particularly in the physiology domain. There is currently no complete overview of how the different classifiable impairment types impact the physiological response to exercise. Therefore, practitioners within the para-sport field may have trouble incorporating data from isolated case studies and/or studies that focus on a single disability type to develop a comprehensive understanding of how various impairments impact the physiological response to exercise. This lack of cohesive understanding can prevent new practitioners from providing appropriate and timely support of para-athletes and may result in researchers having to spend additional time sourcing background material, rather than designing studies to advance the field.

The present review was written as part of a collaborative effort between Own the Podium and members of the Own the Podium Paralympic Professional Development Working Group that is made up of practitioners, athletes, and academics in Canada. The purpose of this review was to provide a comprehensive overview of the physiological considerations that should be taken into account when supporting para-athletes in applied sports performance roles. Nevertheless, the content of this review is structured in such a way that it will benefit researchers and clinicians who want to gain an overview of para-sport physiology as well those wanting to develop interventions to improve the health and performance of para-athletes. Within this review we discuss the neural control of the physiological response to exercise and the current state of research aimed at enhancing exercise performance in athletes with classifiable impairments under the major sub-groups of limb deficiency, cerebral palsy, SCI, and other classifiable neurological impairments. Whilst athletes with visual and intellectual impairments have practical and biomechanical considerations, they are not expected to exhibit an altered physiological response to exercise or energy requirements and hence are omitted from this review for brevity. Finally, we discuss research gaps and areas of interest for future research and innovation.

## NEUROANATOMY OF IMPAIRMENT GROUPS

Given that many of the abovementioned impairments impact neurological function, it follows that the degree of function is highly variable across impairment groups. This can result in some para-athletes exhibiting an unaffected cardiorespiratory, metabolic, and thermoregulatory response to exercise (e.g., visual impairment), and others exhibiting a severely attenuated cardiorespiratory, metabolic, and thermoregulatory response to exercise compared to able-bodied athletes (e.g., cervical SCI). As such, we believe that para-sport researchers and practitioners should familiarize themselves with fundamental neuroanatomy so as to develop appropriate individualized interventions and understand unique practical considerations of the para-athlete. **Figure 1A** provides a general guide to neuroanatomy of the autonomic pathways regulating the thermoregulatory and cardiovascular response to exercise. **Figure 1B** gives an overview of the autonomic and somatic pathways that regulate the ventilatory and skeletal muscle response to exercise. Below we discuss how various impairments may impact the physiological response to exercise in relation to these pathways. It should be noted, however, that there may be a large degree of variability between athletes with the same impairment depending on where the specific lesion (e.g., SCI or cerebral palsy) or injury (e.g., limb deficiency) is located.

## PHYSIOLOGICAL RESPONSE TO EXERCISE ACROSS IMPAIRMENT GROUPS

The cardiac (8), vascular (9), ventilatory (10), metabolic (11), and thermoregulatory (12, 13) response to exercise in able-bodied individuals has been reviewed extensively elsewhere and are only referred to here for comparison. Briefly, at the onset of exercise, the sympathetic nervous system (SNS) (**Figure 1A**) is activated to increase cardiac output and blood pressure to meet the oxygen demands of exercising musculature. Simultaneously, activity of the parasympathetic nervous system (PNS) is gradually withdrawn (14), though present to maximal exercise (14), and increased drive to respiratory muscles (**Figure 1B**) increases tidal volume to facilitate oxygen supply to, and carbon dioxide removal from, the arterial blood. There also exists a complex interplay between peripheral reflexes (i.e., the baroreflex, chemoreflex and exercise pressor reflex) that provide afferent feedback to the brainstem to "fine-tune" cardiorespiratory, metabolic, and thermoregulatory function (15).

### Spinal Cord Injury

The physiological effects of exercise following SCI have been studied extensively and, for further detail, we refer

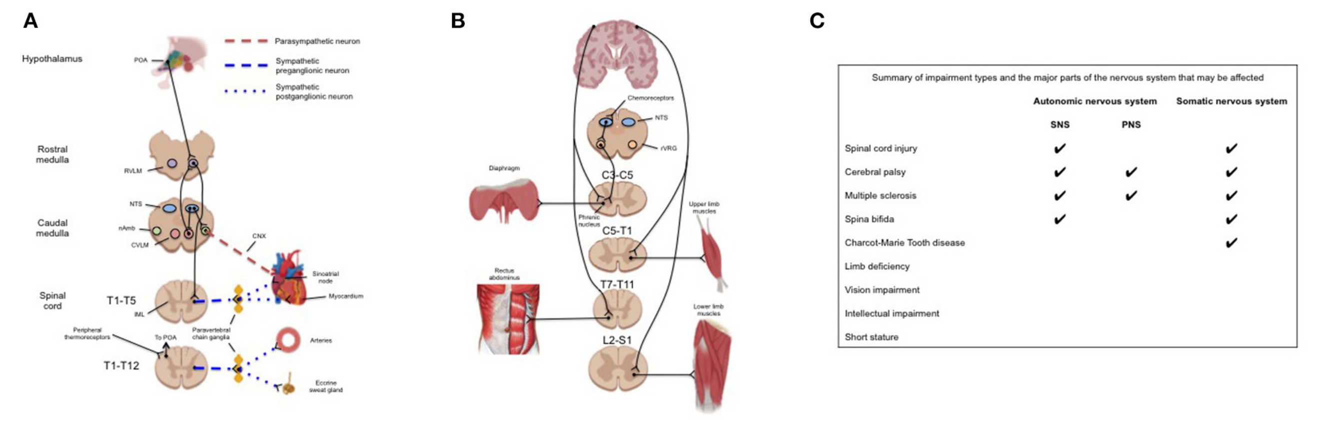


**FIGURE 1 | (A)** Neuroanatomical pathways responsible for the cardiovascular and thermoregulatory response to exercise. Although these responses are integrated within the brainstem the efferent pathways of the sympathetic response pass through the spinal cord and paravertebral chain ganglia. Impairments to these pathways may be at the level of the brainstem (cerebral palsy), spinal cord (spinal cord injury), or peripheral nerves (e.g., multiple sclerosis). **(B)** (Left) Anatomical pathways providing neural drive for the primary muscles of inspiration (diaphragm) and active expiration (rectus abdominus). Pacing of the diaphragm by the rVRG may be overridden by central command and therefore affected by impairments in motor pattern generation (e.g., cerebral palsy); (Right) pathways providing neural drive to upper and lower limb skeletal muscles. **(C)** Summary of common impairment types among para-athletes and which nervous system/s that the impairment may affect. CNX, tenth cranial nerve; CVLM, caudal ventrolateral medulla; IML, intermediolateral cell column; nAmb, nucleus ambiguus; NTS, nucleus tractus solitaries; POA, preoptic area of the anterior hypothalamus; PNS, parasympathetic nervous system; RVLM, rostral ventrolateral medulla; rVRG, rostral ventral respiratory group; SNS, sympathetic nervous system.

readers to several excellent reviews (16–18). In short, the physiological response to exercise following SCI is highly variable and dependent upon the severity and level of the injury. Following a "functionally complete" cervical SCI there is altered cardiovascular and sudomotor function, characterized by a reduced peak heart rate (19), an inability to augment stroke volume (20), exercise-induced hypotension (21), and an impaired sweat response that alters evaporative heat loss (22). Whilst the afferent arms for the three major reflexes are intact, the efferent outflow from the brainstem down the spinal cord does not pass through the injury site, and therefore reduces the ability of these reflexes to modulate descending sympathetic circuitry. Interestingly, our group has found that some athletes who have a "functionally complete high-level SCI" can still present with functional sparing in the descending sympathetic fibers (since these fibers are anatomically distinct from the motor/sensory pathways on which the assessment of functional completeness is based) (23). Across sports, we have found that these athletes will consistently out-perform those who do not have functional sparing of these fibers as they can reach a higher exercising heart rate, stroke volume, cardiac output and oxygen uptake (23). In the respiratory system, although neural drive to the diaphragm remains at least partly intact *via* the phrenic nerve, athletes with complete cervical SCI have severely impaired expiratory function due to the loss of neural drive to expiratory muscles (24), which leads to dynamic hyperinflation during higher intensity exercise (25, 26) that can increase dyspnea and have implications for cardiovascular function.

Among athletes who have sustained a complete SCI at the thoracic level, the physiological response to exercise is highly dependent upon the level of the injury. Athletes with complete high-level SCI (i.e., above sixth thoracic spinal level) will have a loss of the ability to vasoconstrict major splanchnic vascular beds, sweat below the level of the lesion, and are likely to have impaired expiratory function (17). However, they may be able to appropriately elevate their heart rate (23) and increase stroke volume (27) *via* direct cardiac sympathetic excitation during exercise. Athletes with complete lower-level thoracic SCI (i.e., below sixth thoracic spinal level) have intact sympathetic drive to the heart and should be able to vasoconstrict and sweat above the level of their injury. The primary muscle of active expiration, the rectus abdominus, should maintain a degree of neural drive and as such these athletes have greater preservation of expiratory function along with practically normal inspiratory function (17).

With respect to alterations at the muscle, it is now relatively well established that SCI causes a fiber type shift in the inactive limbs toward a Type II (i.e., "fast-twitch") phenotype (28) due to decreases in mitochondrial size and density (29). Despite less favorable conditions for oxygen extraction by the upper limb muscles (30), upper body endurance exercise does appear to result in a number of beneficial adaptations for oxygen off-loading including a muscle fiber type shift toward a greater density of type I fibers (i.e., "slow-twitch") as well as increased capillarisation and glycolytic enzymes (31). For instance, trained athletes with SCI have significantly higher levels of the oxidative enzymes citric synthase and 3-hyrdroxyacyl-CoA dehydrogenase than untrained and able-bodied individuals, with lower activity of the glycolytic enzyme 6-phophofructokinase reflecting a greater dependence on fat oxidation during exercise (31).

The impact of different levels of complete SCI (i.e., injury to descending sympathetic pathways) on peak physiological responses to exercise is summarized in **Table 1**.

**TABLE 1** | Peak physiological responses to arm exercise following complete spinal cord injury relative to able-bodied athletes.

| | Tetraplegia (C5–C8) | High paraplegia (T1–T6) | Low paraplegia (T7–T12) |
|---|---|---|---|
| Heart rate | ↓ (19) | ↓ or = (23) | ↑ (32) |
| Stroke volume | ↓ (20) | ↓ or = (33) | ↓ (32) |
| Cardiac output | ↓ (20) | ↓ or = (33) | ? |
| Blood press | ↓ (21) | ? | ? |
| Catecholamine release | ↓ (34) | ↓ (34) | ↑ (34) |
| Sweat response | ↓ (35) | ↓ (35) | ? |
| Respiratory frequency | = (26) | = | = |
| Tidal volume | ↓ (26) | ↓ | ↓ |
| Minute ventilation | ↓ (26) | ↓ | ↓ |
| Oxygen uptake | ↓ (36) | ↓ (36) | ↓ (32) |

*↓, decreased; =, no change; ↑, increased; ?, unknown.*

## Limb Deficiency

Athletes with limb deficiency, free of other disease or disability, are expected to have an intact central nervous system (CNS) and therefore an unaffected cardiorespiratory and autonomic response to exercise. However, athletes with a limb deficiency (specifically lower limb deficiency performing upper body exercise) can often experience elevated blood lactate concentrations due to a reduced total body mass and high muscle activation in the upper body (37). Due to movement inefficiency, lower limb amputation can increase the energy cost of lower-limb exercise and increase metabolic demand for a given task (38). The limited research available suggests that improving movement efficiency and decreasing the metabolic cost of exercise among lower-limb amputees (38) will likely enhance aerobic exercise capacity.

Athletes with limb deficiency may experience greater thermal strain due to a reduced surface area for evaporative heat loss. Depending on the location of the limb deficiency, and the athletes' use of prostheses, the prostheses may act as an effective insulator and further impair thermoregulation in the remaining limb that is covered by the device. There may also be increased sweat accumulation, and skin breakdown with the use of prosthetics, which may further enhance thermal strain and discomfort. There is evidence that individuals with large surface area of skin grafts may be at higher risk for thermal stress due to the reduction of sweat gland responsiveness and permanent impairment of cutaneous vasodilator capacity of the grafted skin (39).

## Cerebral Palsy

Cerebral palsy is the result of a non-progressive lesion in the developing brain characterized by movement impairments and reduced muscle strength. Among elite level soccer players and cyclists for instance, which are some of the best studied athletes with cerebral palsy, isometric knee extensor strength can be impaired between 31 and 47% (40). Though not in elite athletes, others have shown that voluntary activation can be as low as one-third that of able-bodied individuals (41), suggesting impaired neural recruitment, and that perhaps voluntary strength training may be less effective in this impairment group. In children with cerebral palsy several studies have attributed part of the reduced muscle strength to a type 1 fiber type predominance (41), however we are not aware of such studies in adult or athletic populations. Increased intramuscular fat as well as atrophy and decreased muscle size in the paretic limbs have also been identified as outcomes of cerebral palsy (42).

Athletes with cerebral palsy may present with limited exercise capacity (43) and are at a high risk of musculoskeletal injury, and may additionally experience reduced range of motion, increased muscle stiffness, spasticity, and pain among other medical challenges (44). How the autonomic nervous system functions to meet the physiological response to exercise has not been well-studied (45), however the neural pathways outlined in **Figure 1** are expected to remain intact unless there is damage to the hypothalamus—this is also likely true of para-athletes with movement impairments due to acquired brain injuries. Due to impaired co-ordination of movement and the metabolic cost of exercise, metabolic heat production has been found to be higher in children with cerebral palsy compared to able-bodied children for a given workload (46). Others have suggested that this movement inefficiency, coupled with increased muscle tone, may impede the skeletal muscle pump (47) and limit venous return and left-ventricular stroke volume. In children with cerebral palsy, aerobic exercise training may improve movement efficiency and decrease the metabolic cost of exercise—whether this is true for highly-trained athletes, or whether they have reached a "peak" movement efficiency relative to their impairment is unknown. Finally, lung volumes appear to be smaller in adults with cerebral palsy, however it is not known if this is true among athletes (48).

## Other Classifiable Neurological Impairments

Due to the heterogeneity of how other neurological impairments impact the physiological response to exercise, here we only outline the known physiological effects of select impairments not detailed above.

Multiple sclerosis is an autoimmune disease that degrades the myelin sheath of axons within the CNS (49). The most well-known physiological consequence of multiple sclerosis is impaired thermoregulation and occurs when the lesion impacts thermoregulatory centers such as the hypothalamus (50) and is further compounded by reduced sweat gland output in response to thermal stress (51). Other common symptoms of multiple sclerosis relate to muscle weakness, spasticity and fatigue.

Athletes with spina bifida are often included in studies with athletes with SCI, with the primary difference being that it is a congenital birth defect rather than an acquired injury. Spina bifida is most common in the lumbar or sacral spine, in which case sympathetic function is preserved, but may occur in the cervical spine in rare cases (52). Similar to SCI, the effect of spina bifida on physiological responses to exercise is likely a function of the severity of the impairment (i.e., spina bifida occulta, meningocele, or myelomeningocele).

Neurological conditions that effect peripheral nerves (e.g., Charcot-Marie Tooth disease, Guillain Barre syndrome) primarily effect motor and sensory function, including those responsible for pulmonary function, but have the potential to also cause autonomic neuropathy (53).

## ENERGY AVAILABILITY AND RELATIVE ENERGY DEFICIENCY IN SPORT IN PARA-ATHLETES

Energy availability (EA) represents the amount of energy left over for optimal physiological function after exercise energy expenditure (EEE) is subtracted from energy intake (EI), and corrected for fat free mass (FFM) (54). It is well established that chronic low EA (LEA) results in myriad negative health, psychological and performance outcomes and is the underpinning etiology of Relative Energy Deficiency in Sport (RED-S) (55). However, laboratory and clinical quantification of EA is impressively difficult, and challenged by methodological considerations that introduce risk for significant under- or overestimation of EI (56) and/or EEE (57); although to our knowledge this has primarily been only examined in able bodied athletes. Indeed, we are not aware of studies that have utilized the "gold-standard" approach of double-labeled water to assess total daily energy expenditure (TDEE). Within TDEE, there is significant work to be done to better understand the EEE demands in para-athletes, of which most data is predominantly in wheelchair athletes. However, even our understanding of EEE in wheelchair athletes is limited. For example, our current compendium of energy costs of physical activities for individuals who use manual wheelchairs is now 10 years old and, despite identifying 266 studies, only 11 studies met the inclusion criteria (58). Only four of these studies included para-athletes, of which 91 were male and 6 female. Given the progress of wheelchair technology advances utilized by modern para-athletes, which would potentially impact gross efficiency outcomes of EEE, much more data needs to be developed to accurately estimate EEE. Therefore, the accurate appreciation of EEE in most classifications of para-athletes remains to be elucidated.

Since the assessment of EA is challenging in evaluating LEA, more chronic indicators of LEA tend to be used for RED-S diagnosis in able-bodied athletes (59), and common outcomes include: clinically low hormones involved in the hypothalamic-pituitary-gonadal-adrenal (HPGA)-axis; amenorrhea; clinically low bone mineral density (BMD); restricted eating leading to disordered eating or eating disorders; increased risk of injuries; poor training adaptations and performance outcomes. However a validated RED-S diagnosis tool in able and para-athletes remains to be developed.

Within para-sport, several recent reviews have highlighted that depending on the impairment many para-athletes have significant challenges with optimizing EI, coupled with potentially altered aspects of EEE, EA, and FFM and thus may be especially at risk for LEA and RED-S (60). Despite over 70 papers published since the 1980's on either RED-S or the Female Athlete Triad in able-bodied athletes (55, 61), we are only aware of three recent publications that have assessed the risk of LEA and/or other chronic indicators of RED-S in para-athletes (62, 63). Depending which assessment tool was used, there were indications of ∼73% of elite female athletes with SCI having LEA (62), which was significant more than male athletes with SCI. In another study, 78% of female athletes with SCI were deemed at "risk" for LEA using the LEA in Females Questionnaire (LEAF-Q), although in this study EA assessments were not compromised (63). Taken together and similar to the able-bodied literature, there are considerable discrepancies in RED-S assessment across various tools/parameters, and much more research is required. Among a cohort of male and female track and field para-athletes with cerebral palsy, vision impairment, or limb deficiency >82% had reduced EA and approximately one-third had LEA (64).

The actual prevalence of RED-S may be profound, as a recent survey study in 260 elite para-athletes demonstrated that ∼32% had elevated disordered eating questionnaire scores, ∼45% of premenopausal females had oligomenorrhea/amenorrhea, ∼55% had reported low BMD, but < ∼10% had awareness of the RED-S (65). Indeed, athletes with SCI may be especially at risk. For example, wheelchair athletes often have significant lower body muscle atrophy resulting in lower whole-body FFM, which may artificially elevate EA calculations compared to able-bodied athletes. Additionally, ∼50% of male athletes with SCI have low testosterone (66), due either to altered HPGA-axis outcomes related to SNS dysfunction and/or aspects of LEA. It is important to note that as of yet, no clinical and laboratory normative LEA data has been published in para-athletes linked to adverse clinical outcomes. Therefore, we have to encourage much more research to develop RED-S specific data and assessment tools in these unique para-athlete populations.

## PHYSIOLOGICAL INTERVENTIONS TO OPTIMIZE PERFORMANCE

Elite able-bodied athletes and their support staff constantly strive to gain a competitive advantage within the rules of their sport to enhance performance. Two of the more common physiological performance enhancing strategies used by able-bodied athletes may include altitude training (67) and heat acclimation training (68). Although there are extensive research and review articles on these, and other, interventions for the able-bodied athlete, there is often little to no research on how these interventions may affect elite para-athletes—while some may be beneficial, some may be detrimental to performance and/or health.

In recent years there have been a number of studies that highlight the impacts of respiratory muscle training (69, 70), abdominal binding (71), and heat acclimation (72, 73) in para-athletes (see 23). However, there remains a large gap in the research and applied knowledge for how these protocols may translate to benefit para-athletes in all impairment groups.

**TABLE 2 |** Key physiological interventions specific to para-athletes.

| Key interventions | Existing evidence by athlete group | Considerations for other para-athlete groups that may benefit |
|---|---|---|
| Respiratory muscle training | In able-bodied endurance athletes, IMT appears to benefit performance by delaying diaphragm fatigue, offsetting the respiratory muscle metaboreflex, and attenuating respiratory discomfort (10, 74) | • Among endurance athletes with minimal impairment to cardiorespiratory function (i.e., limb deficiency, intellectual and visual impairment), competing in para-sports with high ventilatory demand, there is likely potential benefit from similar IMT protocols as those applied to able-bodied athletes |
| | In athletes with cervical SCI, pressure threshold IMT (69), and combined IMT and EMT (70) (30 repetitions performed twice daily on 5 days/week) appear to ↑ respiratory muscle strength and aerobic exercise capacity | • No respiratory muscle training program has been assessed in athletes with cerebral palsy. However, it is plausible that strengthening the respiratory muscles could elicit functional and structural adaptations that benefit trunk stability and movement patterns—as has been observed in children with cerebral palsy (75) |
| | | • Combined IMT and EMT has not been examined in athletes with multiple sclerosis or muscular dystrophy, however evidence from the non-athletic population suggests it can reduce self-reported fatigue and the severity of breathlessness, respectively (76, 77) |
| Abdominal binding | In athletes with cervical SCI, can prevent exercise induced hypotension, prevent pooling of blood in the abdomen and acutely ↑ resting cardiac output (78), ↓ exercising lung volumes (25), and enhance both lab- (25) and field-based (71) exercise performance | • The efficacy of abdominal binding among athletes with other impairments has yet to be assessed however, theoretically, it may improve central hemodynamics in athletes with paraplegia who have impaired neural drive to the rectus abdominus (see **Figure 1B**) and may enhance stability among athletes with impaired muscle power or co-ordination of the trunk musculature |
| Heat acclimation | In able-bodied athletes, or para-athletes with an intact SNS: ↓heart rate, ↑cutaneous blood flow and sweat rates, ↓core temperature and improved exercise performance Secondary benefits, may include ↑ plasma volume and ↑stroke volume (68) | • There is potential for athletes with minimal impairment to autonomic pathways, vasomotor and sweat control to benefit from similar physiological responses to heat acclimation as shown in able-bodied literature. Such athletes would likely benefit most from protocols similar to those that have been most effective in able-bodied athletes (79) |
| | | • Some athletes with a SCI and MS have significant challenges in the heat due to poor sweat rates and poor thermoregulation and therefore need enhanced monitoring around optimizing hydration practices. Athletes with MS are especially more sensitive to heat, and need heat mitigating strategies. As such, we suggest that these athletes are closely supervised by a practitioner if undergoing heat acclimation training |
| | In athletes with high risk of thermal strain and/or impaired SNS, potential improvements in heat tolerance, pacing strategies, and ↑ in plasma volume | • Heat acclimation/acclimatization may be recommended to improve performance and heat tolerance for athletes competing in hot-humid environments, sports requiring a high aerobic demand and athletes with a high risk of thermal strain |
| | In athletes with MS, the physiological response and performance benefits of heat acclimation may not outweigh the negative impacts it has on symptoms (50) | • Symptoms including early onset of fatigue associated with MS may be exacerbated by as little as a 0.5°C increase in core temperature in 60–80% of MS patients. In athletes with MS it is important to limit their exposure to the heat, however, more research is needed to better understand if their symptoms and heat tolerance may improve with heat acclimation training |
| Altitude Training | Living and/or training at altitude can enhance aerobic exercise capacity, primarily *via* augmentation of red blood cell count, in elite and sub-elite able-bodied athletes (67) | • Among endurance athletes with minimal impairment to cardiorespiratory function (i.e., limb deficiency, intellectual and visual impairment), competing in para-sports with high aerobic demand (e.g., track event of 5,000 m and greater or their equivalent), there is likely potential benefit from similar altitude training protocols as those proven effective for able-bodied athletes. However, this remains to be established in para-athletes<br>• Given the risks associated with low oxygen availability at altitude and the limited research in para-athletes with oxygen transport limitations we suggest para-athletes are closely supervised by a practitioner if undergoing altitude training |
| | In able-bodied athletes, the use of sildenafil, has been shown to ↑ exercising peak power output and peak oxygen uptake at high altitudes above 4,000 m | • The use of sildenafil would not be recommended prior to competition and athletes who are prescribed sildenafil (commonly used to treat erectile dysfunction in athletes with SCI) should be aware of the potential negative effects on exercise performance |
| | In athletes with SCI, best evidence suggests that sildenafil does not enhance exercise capacity compared to placebo at sea level or altitude (80) | • No major global able-bodied or para-athletes specific competitions are held at altitudes above 4,000 m |

*CP, cerebral palsy; EMT, expiratory muscle training; IMT, inspiratory muscle training; MS, multiple sclerosis; SCI, spinal cord injury; SNS, sympathetic nervous system.*

**TABLE 3 |** Practical and special considerations for supporting para-athletes.

| Special training considerations | Specific sub-factors related to para-athletes | Physiological and applied practical considerations |
|---|---|---|
| Environmental conditions | Medications, sleep deprivation and fitness levels may impact an athlete's heat tolerance | • Medication use and sleeping habits should be monitored when the athlete is traveling or competing in a warm environment. Ensure athletes aerobically fit to tolerate training/competing in the heat |
| | Equipment interactions and sweat hygiene | • Maintain consistent prosthetic hygiene to decrease risk of infections, skin breakdown due to accumulation of sweat or contact with sports equipment especially during hot humid conditions where sweat rates may be higher<br>• Trial equipment and prosthetic fit prior to competing in altered environmental conditions |
| | Early onset of fatigue ↓ in performance in hot or cold environments | • Use of cooling strategies in warm environments (slushies, ice vests, cold water immersion, cold spray, menthol) is recommended for all athletes with impairments (81)<br>• Heat acclimation training to improve heat tolerance (refer to **Table 2**)<br>• Warming strategies in cold environments (warm fluids, extra layers, heated garments). Consider the effects of decrease blood flow, and sensation |
| | Pace awareness and perception of effort are exacerbated in the heat (43, 46) | • Athletes with CP, intellectual impairment or VI who compete without a guide, should practice pacing outcomes in the target weather conditions prior to competition to establish a pre-determined pacing strategy based on the ambient conditions |
| | ↑ Risk of thermal strain; ↓ evaporative or convective heat loss, ↑ metabolic heat, ↓ vasomotor and sweat control | • Athletes with a SCI, limb deficiency, CP, VI, short stature, or other neurological conditions would benefit from individualized internal and external cooling strategies (pre, per and/or post cooling based on the sport)<br>• Consider the cost benefit of the added thermal strain when using additional clothing garments, equipment and wearable devices |
| | Cold environments may impact athletes with muscle stiffness, nerve pain, changes in vasomotor and sudomotor tone | • Ensure the temperature in the gym and training facilities are a neutral temperature. If you are training or competing in cold environments, ensure the athlete has a good warm up and potentially look to pre warming techniques to minimize the cold related symptoms |
| Monitoring | Pace awareness and perception of effort (43, 46) | • In athletes with an intellectual impairment, using RPE scales may not be appropriate<br>• For athletes with intellectual or visual impairment, consider strategic use of a pacer in practice, followed by trialing without the use of a pacer in practice, to mimic race/competition demands |
| | RPE for monitoring | • May consider using a differentiated approach for RPE relative to central (cardiorespiratory), peripheral (blood lactate), and overall (central + peripheral) feeling of effort (82) |
| | Impaired peak heart rates | • In athletes with impaired SNS function, monitor training using heart rates based on the individual athletes peak exercising heart rates |
| | Impaired skin conductivity | • In athletes with skin grafts or neurological conditions it may not be appropriate to use finger or wrist worn heart rate monitors due to decrease skin conductivity |
| | ↓ Blood lactate clearance | • Athletes with a lower limb deficiency performing upper body exercise may have reduce blood lactate clearance due to a decrease in total body mass and increased activation of upper body musculature |
| Training considerations | Consider ADL's (transferring, driving, pushing, cooking, bowel routines etc.) | • ADL's should be a consideration in programming overall training workloads, as these may influence fatigue and readiness for training to a greater extent than able-bodied athletes<br>• In athletes with SCI, lower limb deficiency, and VI the workload completed in training can have a big impact on what the athletes can do for the remainder of their day as well. |
| | ↑ in spasticity following maximal exertion due to an overexcitability of the stretch reflex<br>Sensitivity to the stresses incurred by training sessions with high anaerobic content | • For athletes with hypertonia, ensure appropriate recovery times and balanced training with high intensity/ high anaerobic efforts when planning training phases. May be at risk of increased hypertonia, pain, stiffness, and clonus |
| Travel | Sleep disorders, altered distribution of melatonin and temperature regulation throughout the day | • Jet lag and travel fatigue may be exacerbated in athletes with sleep disorders and athletes with VI due to an already disrupted circadian rhythm. Establish a travel plan, a sleep schedule and periodized training upon arrival pre-event following long haul trips<br>• In athletes with intellectual and visual impairments, and some athletes with a SCI, sleep medications and poor sleep impact on circadian rhythms and optimal hormone regulation (e.g., lowered testosterone, increased cortisol), which can impact on eating behaviors and body composition outcomes |
| | Athletes may dehydrate themselves during travel and may go multiple days without emptying their bowels | • Establish an individualized hydration plan during travel and upon arrival. Consider the athletes bowel routines when planning training schedule upon arrival<br>• Wheelchair users are at increased risk for dehydration, especially when traveling, due to accessibility challenges |

*(Continued)*

**TABLE 3** | Continued

| Special training considerations | Specific sub-factors related to para-athletes | Physiological and applied practical considerations |
|---|---|---|
| | ↑ stiffness or spasticity with long international travel | • Symptoms may be exacerbated with long haul travel for athletes with increased spasticity, decreased range of motion and movement<br>• Promote movement during travel as much as possible, bring their own seat cushions for the plane and encourage weight shift. Focus on mobility and light movement in the first few days of arrival, allow time for the athletes to lay and stretch out upon arrival |
| | Stump volumes change due to the accumulation of fluid if the prosthetic limb is removed in flight | • Awareness and regular stump care. Proper fitting of prosthetics and having alternative training strategies when tissue health is compromised |

*ADL, activity of daily living; CP, cerebral palsy; RPE, rate of perceived exertion; SCI, spinal cord injury; VI, visually impaired.*

It is beyond the scope of this review to fully review these interventions. Nevertheless, some key interventions and practical considerations to working with para-athletes are highlighted in **Tables 2** and **3**, respectively.

## FUTURE AREAS OF RESEARCH AND INNOVATION

The Tokyo 2020 Summer Paralympic Games featured 4,403 athletes from 162 participating countries who competed in a total of 539 events were contested across 22 sports (83). Despite growing interest and participation in Paralympic sport, along with the influx of published articles on disability sport over the last quarter century, there remains a lack of evidence-based physiological interventions to improve performance in the para-athlete, especially those without SCI. This is in part due to the barriers associated with conducting research on small heterogenous cohorts of para-athletes. For example, Stephenson et al. (73) conducted a heat acclimatization intervention in seven elite para-triathletes across five different impairment groups. While conducting such a research study is commendable from a logistical perspective, the heterogeneity of the cohort makes it difficult to form conclusions as to how the intervention may be applied to para-athletes with various impairments. To overcome these limitations, practitioners often rely on knowledge transfer from experience, colleagues, and anecdotal evidence through case study approaches to inform their practice. Current practice is often adapted based on able-bodied research, yet there is little validated data on how many of the interventions, protocols, training methods and assessment tools used by practitioners directly translate to support athletes in all impairment groups or across athletes with varying severities of impairment (i.e., SCI level, hemiplegia vs. diplegia, level of amputation). Furthermore, there is a distinct lack of published data on the physiological demands of para-sports from which practitioners can base physiological interventions and training programs. This includes simple, relatively non-invasive data such as game duration/intensity profiles integrated with heart rate, oxygen consumption, and/or lactate profiles.

Thermoregulation is an area that has been relatively well researched in para-athletes. However, the majority of research focuses on athletes with a SCI due to the known higher risk of thermal strain in this population (22). Therefore, we can only hypothesize as to whether other impairment groups would experience increased thermal strain relative to able-bodied athletes, or would benefit from cooling strategies and/or heat acclimation training. The current literature on heat acclimation is limited in elite para-athletes (72, 73). This not only highlights the difficulty in recruiting large homogeneous samples of para-athletes but emphasizes the need to determine impairment specific physiological responses, adaptations, and performance outcomes to training in the heat for practitioners to individualize training preparation for Paralympic athletes.

Competing in the heat is not the only environmental condition that impacts para-athletes. We believe that further research should also examine the impact air quality, cold environments, and altitude on performance. In particular, altitude training is a common intervention adopted by elite able-bodied endurance athletes (84), but we are unaware of any altitude training interventions in para-athletes whom commonly compete at moderate altitudes. Additionally, the areas of travel and immune function are only starting to emerge in conversation and the scientific literature (85) and warrant further research.

It is understood that elite para-athletes have a high prevalence of injury and illness (86), and is therefore imperative for the integrated support staff to first support the para-athletes health and well-being for them to train and perform at the highest level. There is a plethora of data indicating the negative health, psychological and performance impacts that LEA has on able-bodied athletes (87), however, future research is critical to understanding the prevalence and consequences of LEA and metabolic considerations across all impairment groups. Currently, a challenge for practitioners is the accuracy and validity of assessment tools available for monitoring EA, as all practical methods were developed with no consideration of athletes with an impairment. Even though many para-athletes may be at high risk for developing RED-S, we have no normative data or valid assessment tools to accurately monitor resting metabolic rate, EE, and RED-S in athletes with an impairment.

## CONCLUSIONS

The present review has outlined the neurophysiology of the most common impairment groups and the practical considerations when supporting para-athletes along with performance enhancing interventions. We acknowledge that a highly individualized approach to supporting para-athletes is needed due to the variability not only between but within impairment groups and emphasize the need to further enhance our approach to providing practitioners with evidence-based research. We suggest that this may be achieved by various knowledge translation strategies, including (i) modules to educate current and future practitioners in the para-sport field, (ii) encouraging para-sport practitioners to publish and/or present on unique field observations, and/or (iii) the sharing of data between sport systems including cross collaborative projects between research groups from different nations as well as sports with para-athletes of similar impairment groups.

## AUTHOR CONTRIBUTIONS

All authors provided substantial contributions to the conception and design of the work and drafting the work or revising it critically. Final approval of the version submitted/published and consent for publication has been agreed by all authors.

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# Shoulder Tendon Adaptations Following a Graded Exercise Test to Exhaustion in Highly Trained Wheelchair Rugby Athletes With Different Impairments

*Fransiska Marie Bossuyt* [1,2]*, *Barry S. Mason* [3], *Simon Briley* [3], *Thomas J. O'Brien* [3], *Michael L. Boninger* [4], *Ursina Arnet* [1] *and Victoria Louise Goosey-Tolfrey* [3]

[1] *Shoulder, Health and Mobility Group, Swiss Paraplegic Research, Nottwil, Switzerland,* [2] *Human Performance Laboratory, Faculty of Kinesiology, University of Calgary, Calgary, AB, Canada,* [3] *Peter Harrison Centre for Disability Sport, School of Sport, Exercise, and Health Sciences, Loughborough University, Loughborough, United Kingdom,* [4] *Department of Physical Medicine and Rehabilitation, School of Medicine, University of Pittsburgh, Pittsburgh, PA, United States*

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***Correspondence:**
*Fransiska Marie Bossuyt*
*fransiska.bossuyt@paraplegie.ch*

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**Objective:** This study aimed to identify acute changes in biceps and supraspinatus tendon characteristics before and after a graded exercise test to exhaustion (GXT) in highly trained wheelchair rugby (WR) athletes. A secondary aspect was to define chronic tendon adaptations related to the impairment of the athlete and the occupation of the tendon within the subacromial space (occupation ratio).

**Methods:** Twelve WR athletes with different impairments (age $= 32 \pm 6$ years; body mass $= 67.2 \pm 11.2$ kg; $9.0 \pm 3.6$ years competing) volunteered for this study. Performance Corrected Wheelchair Users Shoulder Pain Index was used to quantify shoulder pain. Quantitative Ultrasound Protocols (QUS) were used to define supraspinatus and biceps tendon thickness, echogenicity, and echogenicity ratio of both dominant and non-dominant shoulder before and after the GXT including $22 \pm 3.1$ min submaximal propulsion and $10.2 \pm 1.7$ min maximal propulsion on a treadmill. Furthermore, the acromio-humeral distance (AHD) defined from ultrasound (US) images was used to calculate the occupation ratios.

**Results:** A mixed-effect multilevel analysis that included shoulder as grouping variable, demonstrated a significant reduction in the echogenicity of the biceps following GXT whilst controlling for impairment [spinal cord injury (SCI) and non-SCI] and the occupation ratio ($\beta = -9.01$, $SE\beta = 2.72$, $p = 0.001$, 95% CI = [−14.34; −3.68]). This points toward fluid inflow into the tendon that may be related to overload and acute inflammation. In addition, persons with a SCI ($n = 8$) had a thicker supraspinatus tendon in comparison to persons with non-SCI ($n = 3$) which may be related to chronic tendon adaptations ($\beta = -0.53$ mm, $SE\beta = 0.26$, $p = 0.038$, 95% CI = [−1.04; −0.03]). Finally, a greater occupation ratio was associated with signs of tendinopathy (i.e., greater biceps and supraspinatus tendon thickness, and lower supraspinatus echogenicity and echogenicity ratio).

**Conclusion:** Acute biceps tendon adaptations in response to the GXT in highly trained WR athletes were evident with chronic adaptations in the supraspinatus tendon being related to the impairment of the athlete. Ultrasound can be used to monitor tendon adaptations in WR athletes for medical diagnosis to assist the scheduling and type of training.

**Keywords: ultrasound, exercise test to exhaustion, para-athlete, supraspinatus tendon, biceps tendon, sports and exercise medicine**

## INTRODUCTION

Wheelchair rugby (WR) is a fast-paced, paralympic sport played by athletes with a variety of health conditions, with impaired trunk and upper limb function (1). Elite WR players push at high speeds with frequent stops and starts during both competition and training (2, 3). With overhead activities such as passing and catching, also a common feature of WR (4), the demands placed on the shoulder and the potential risk of injury and shoulder pain are likely to be elevated (5, 6). Indeed, 9 out of 12 tetraplegic WR-athletes reported some shoulder pain during activity in the past week (7) and 7 out of 8 elite WR players reported pain after exercise (8). However, the extent of the demands during WR and the risk of pain and pathology in WR athletes remains unclear (9, 10).

Musculoskeletal ultrasound (US) has become a popular tool for identifying musculoskeletal pathologies and monitoring tendon health, due to its low cost, ease of use, and non-invasive approach (11–14). Research using US and MRI has indicated that wheelchair users with a spinal cord injury (SCI) experience a number of shoulder pathologies, with tendinopathies or chronic tendon degeneration of the bicipital and supraspinatus tendons amongst the most common (15–17). Supraspinatus and biceps tendinopathy has also been associated with impingement due to a reduction in the sub-acromial space and therefore a greater occupation ratio [i.e., thickness of the tendon relative to the acromio-humeral distance (AHD)] (18), which naturally occurs during overhead and propulsion activities (18–20). Previous research identified differences in the occupation ratio between persons with subacromial impingement syndrome and healthy controls which further underscored the value of not only investigating tendon thickness and AHD separately (21, 22). Subsequently, ample research has utilized US to establish the thickness and structure of the supraspinatus and biceps tendon, as well as the AHD to quantify the subacromial space, in manual wheelchair users with SCI (20, 23–25). However, previous research has primarily focused on SCI wheelchair users and to date, only one study investigated shoulder tendon characteristics in a sample of WR athletes including persons with a tetraplegia ($n = 11$), paraplegia ($n = 21$), and non-SCI ($n = 2$) (26). While it is valuable to investigate homogeneous samples of persons with similar injuries, the lack of research on wheelchair users with non-SCI impairments causes a gap in the literature.

Monitoring tendon adaptations in response to acute loading is needed to better understand the development of tendon degeneration, and ultimately to be able to intervene and prevent injuries. Previous studies have therefore identified acute tendon adaptations pre- and post-fatiguing wheelchair propulsion performed in the users' daily chair (25, 27). More specifically, a 15-min fatigue protocol in combination with treadmill propulsion at different power outputs, and maximum sprint and strength tests, induced an acute reduction in supraspinatus tendon thickness in a population-based sample of 50 wheelchair users with SCI when controlling for fatigue and subject characteristics (25). However, the 15-min fatigue protocol in itself did not induce significant shoulder tendon changes in 60 wheelchair users of which 80% were athletes (27). Furthermore, a graded treadmill-based propulsion test to maximum exhaustion did not induce significant changes in shoulder tendons in 15 wheelchair users (28). Progressing to wheelchair basketball and WR game play, van Drongelen et al. (26) noted a significant decrease in mean echogenicity ratio of the biceps tendon representing potential fluid inflow into the tendon following these sporting activities. That said, the acute changes following the repetitive activities apparent in these sports differed based on the amount of playing time. Moreover, a lower echogenicity ratio was observed both at the onset and following the competitive games in players who reported shoulder pain.

To date, no study has investigated shoulder tendon adaptations following repetitive activity up to maximum exhaustion, when the musculoskeletal system is unstable and particularly prone to tissue adaptations (29), in highly trained WR athletes with different physical impairments. Subsequently, the aims of the current study were (1) to identify acute changes in biceps and supraspinatus tendon characteristics following a graded exercise test to exhaustion (GXT) in highly trained WR athletes, and (2) to define differences in chronic tendon adaptations related to the impairment of the athlete and the occupation ratio. We thereby also investigated a potential association between changes in shoulder tendon characteristics and shoulder pain.

## METHODS

### Participants

Twelve highly trained National level WR players consisting of 11 males and one female player (age = 32 ± 6 years; body mass = 67.2 ± 11.2 kg) provided their informed consent to participate in the current quasi-experimental study with a repeated measures design. The study was approved by the local ethical advisory committee. Participants were grouped according to those who had a tetraplegic (complete lesion level between cervical vertebrae

C5 and C7) SCI ($n = 8$) and those who had a non-spinal impairment (cerebral palsy, critical care polyneuropathy, brachial plexus nerve injury, and Roberts syndrome: non-SCI; $n = 4$). All participants with SCI were 17 years or older when they sustained their injury.

## Experimental Design

The assessments are partly included in the annual monitoring programme of the WR athletes and briefly described below. The additional assessments that are included in the annual monitoring programme (e.g., 30 s Wingate test on a dual roller wheelchair ergometer aimed to determine anaerobic capacity) will be presented elsewhere. Body mass and mass of the daily and rugby chair were obtained to the nearest 0.1 kg with seated balance scales (Seca, Birmingham, UK). Participants completed the Wheelchair Users Shoulder Pain Index (WUSPI), with a performance-corrected version (PC-WUSPI) used to indicate the magnitude of shoulder pain (30, 31). The Upper Extremity Pain Symptom Questionnaire (PSQ) was used as an auxiliary questionnaire to the PC-WUSPI to identify the presence of pain and establish whether shoulder pain was unilateral or bilateral (32). The continuous GXT was performed in participants customized rugby wheelchairs (Rugby chair mass: 17.0 ± 1.4 kg, handrim diameter: 0.54 ± 0.01 m, chamber: 18.1 ± 1.8°) on a motor driven treadmill (HP Cosmos, Traunstein, Germany). Musculoskeletal US examinations were taken to determine (2) the characteristics of the supraspinatus and biceps tendons of both dominant and non-dominant side pre- and post- the GXT, and (3) the AHD pre- the GXT.

## Ultrasound

Two images of the biceps and supraspinatus tendon of both the dominant and non-dominant side were taken in a randomized order at two different time points following previously validated Quantitative Ultrasound Protocols (QUS) (13, 14) with an US device (Legic E9, GE Healthcare, USA). Quantitative Ultrasound Protocols has been used previously before and after fatiguing tasks (25–27) and allows limited error in probe location between measuring time points because of the use of a steel marker taped to the skin that allows to identify the region of interest (ROI). The QUS images were taken before any tasks took place (duration ca. 30 min) (pre-exercise; time point 0) and after the GXT (duration ca. 15 min) (post-exercise; time point ∼ 1 h 45 min). For the longitudinal images of the biceps tendon, participants were seated in their rugby wheelchairs with their arms at 0° abduction and 90° elbow flexion with the palm facing upwards (**Figure 1B**) (13, 14, 25). To take transverse images of the supraspinatus tendon, the participants were asked to externally rotate the shoulder and place the palm flat on the back of the wheelchair (**Figure 1A**) (13, 14, 25). Additionally, three images of the AHD were taken pre-exercise in a seated position with the arms at 0° abduction and 90° elbow flexion with the thumbs facing upwards (**Figure 1C**) (33).

All US images were analyzed in a randomized order by the same examiner requiring 20 min per image (FMB). Using the ROI, tendon images were analyzed to calculate tendon thickness (mean distance between horizontal tendon borders), echogenicity (mean grayscale of the ROI), and echogenicity ratio (echogenicity relative to the mean grayscale of the muscle above the tendon). The shortest distance between the anterior, inferior edge of the acromion and the head of the humerus was used to define AHD. The mean of each repeated variable at the respective time point was used for further analysis. Tendon occupation ratios were calculated as a percentage of the tendon thickness relative to the AHD.

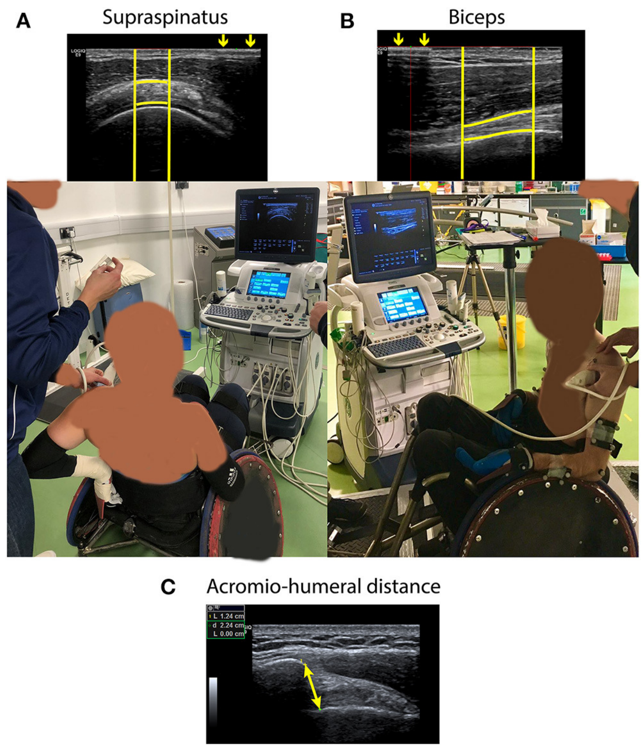


**FIGURE 1 | (A)** Ultrasound image of supraspinatus tendon taken following Quantitative Ultrasound Protocols (QUS) and position in which the image was taken. **(B)** Ultrasound image of biceps tendon taken following QUS and position in which the image was taken. For both **(A)** and **(B)**, arrows demonstrate the interference pattern that resulted from a metal marker taped to the skin from which the bright vertical lines were defined. Bright horizontal lines following the tendon borders are manually defined. **(C)** Ultrasound image of the acromio-humeral distance (i.e., the shortest distance between the anterior inferior edge of the acromion and the humeral head). The image was taken in the same position as the image of the biceps tendon, but with the thumbs facing upwards.

## Graded Exercise Test to Exhaustion

Immediately after the pre-exercise US measures, participants completed a 10 min self-selected warm-up in their own rugby wheelchairs at speeds lower than the subsequent incremental exercise test. Following 5 min passive rest, participants completed a submaximal incremental exercise test. In brief, participants completed 3-min blocks of exercise, where speed was increased by 0.2 m/s ($m.s^{-1}$) for low functioning participants [World WR (1) classification <2.0] or 0.3 $m.s^{-1}$ for higher functioning participants (WWR classification ≥2.0) for determination of speed at blood lactate threshold. The starting speed was individualized according to functional capacity (SCI or non-SCI), WWR classification, and previous test results (where available), with the goal to obtain similar total test durations for all participants (34). Termination of the submaximal test occurred when blood lactate concentration exceeded 4 mmol/l and/or a Rate of Perceived Exertion (RPE) of 17 was reached (34, 35). One investigator (TJO'B) gave all verbal encouragement which included specific quotes such as "Come on, keep pushing," "Keep pushing all the way to the end," "You're doing great, maximum effort," with these quotes kept consistent between participants. Following ∼30 min passive rest, participants completed the GXT with speed increments of 0.1 $ms^{-1}$ every minute to determine maximal oxygen uptake ($VO_{2peak}$) (Metalyzer® 3B, Cortex Biophysik GmbH, Leipzig, Germany) (34). Starting speed for this test was based on visual determination of their blood lactate threshold from the submaximal test. Strong verbal encouragement was given throughout until they could not maintain the speed of the treadmill, which terminated the test. Following completion of the exercise protocol, post-exercise US measurements were completed.

## Statistical Analysis

Statistical analyses were conducted with STATA software (version 14, StatCorp, LP, College Station TX, USA). Subject characteristics between SCI and non-SCI participants were compared using independent sample *t*-tests. The intraclass correlation (ICC) of the repeated US measures (i.e., at each time point of data collection, we collected two images of the biceps tendon and two images of the supraspinatus tendon) were calculated with a two-way random effects model (absolute agreement, random effects: participant ID and measure) to confirm good reliability between measurements at a single time point (36). Ultrasound measures with a poor reliability (ICC ≤ 0.5) would be removed from further analyses (36). Ultrasound data from all shoulders were included into a mixed-effect multilevel analysis to identify the association between dependent variables (tendon characteristic) and time point (pre- or post-exercise; acute adaptations), whilst controlling for impairment (SCI and non-SCI; chronic adaptations) and the occupation ratio. Shoulder (dominant or non-dominant) was included as a grouping variable (random intercept). Normality of the residuals was confirmed with Histogram, Quantile normal plots, and Shapiro-Francia tests. Likelihood-ratio tests after estimation of the unadjusted and adjusted model were used to confirm the significance of the random intercept. Pearson's correlations were used to explore the relationship between shoulder pain (PSQ scores) and tendon characteristics from all shoulders pre- and post-exercise. Correlations were described as negligible (<0.3), low (0.3–0.5), moderate (0.5–0.7), and high (>0.7) (37). Statistical significance was accepted as $p < 0.05$.

# RESULTS

The physical characteristics of participants are presented in **Table 1**. No significant difference existed between SCI and non-SCI except that persons with SCI had a lower $VO_{2peak}$ ($p = 0.04$) and spent less time in the gym ($p = 0.02$).

**TABLE 1** | Characteristics of participants and their wheelchair (whc) stratified by impairment.

| | Total (*n* = 12) | SCI (*n* = 8) | Other (*n* = 4) | *p* |
|---|---|---|---|---|
| Age (years) | 31.8 ± 5.6 | 31.8 ± 6.3 | 31.8 ± 4.8 | |
| Height (cm) | 170 ± 22.1 | 177.3 ± 14.1 | 155.5 ± 30.1 | |
| Body mass (kg) | 66.3 ± 12.1 | 68.4 ± 11.6 | 62.1 ± 13.9 | |
| Years whc use | 14.8 ± 8.3 | 12.6 ± 6.5 | 19.3 ± 10.8 | |
| Years competing | 9.0 ± 3.6 | 9.8 ± 4.0 | 7.5 ± 2.4 | |
| Court hours per week | 8.4 ± 2.8 | 8.1 ± 2.9 | 9 ± 3.2 | |
| Gym hours per week | 4.5 ± 2.3 | 3.4 ± 1.5 | 6.5 ± 2.4 | **0.02*** |
| Other sports e.g., swimming, handbike hours per week | 2.2 ± 1.8 | 1.9 ± 1.6 | 2.8 ± 2.2 | |
| Total training hours per week | 15.1 ± 4.9 | 13.5 ± 3.8 | 18.3 ± 6.6 | |
| Rugby chair mass (kg) | 17.0 ± 1.4 | 16.7 ± 1.3 | 17.6 ± 1.4 | |
| Tire pressure (psi) | 148 ± 36 | 161 ± 38 | 123 ± 36 | 0.07 |
| $VO_2$peak (ml/kg/min) | 25.26 ± 7.88 | 22.08 ± 7.34 | 31.62 ± 4.60 | **0.04*** |

*An * and bold values are marked from the independent sample t-tests with significant p-values (α = 0.05).*

The ICC of the repeated measures ranged between 0.71 and 0.99 representing high correlations for AHD and all investigated tendon characteristics; except for the post-measurements of the supraspinatus echogenicity on the dominant side (ICC = 0.52) representing a moderate correlation.

## Changes in Tendon Characteristics Following Graded Exercise Test to Exhaustion (Acute Adaptations)

The only tendon characteristic to significantly change post-exercise was the echogenicity of the biceps tendon which significantly reduced post-exercise ($\beta = -9.01$, SEβ = 2.72, $p = 0.001$, 95% CI = [−14.34; −3.68]) (**Table 2**; **Figure 2**). More specifically predictive margins of biceps tendon echogenicity changed from 105.14 before GXT (SE = 5.79, 95% CI = [93.79; 116.49]) to 98.11 following GXT (SE = 5.83, 95% CI [86.69; 109.53]) ($p < 0.001$). No further adaptations were observed over time.

## Association Between Tendon Characteristics, Impairment, and Occupation Ratio (Chronic Adaptations)

Persons with a non-SCI had a thinner supraspinatus tendon (β = −0.53 mm, SEβ = 0.26, $p = 0.038$, 95% CI = [−1.04; −0.03]) in comparison to SCI (**Table 2**; **Figure 2**). More specifically, predictive margins for supraspinatus thickness for persons with SCI were 4.19 mm (SE = 0.14, 95% CI = [3.91; 4.47]), while for

**TABLE 2** | Unadjusted characteristics [mean (SD)] of the biceps and supraspinatus tendon pre- and post- a fatiguing bout of exercise, the occupation ratio, and acromio-humeral distance (AHD) in WR players with SCI and Non-SCI.

| | Pre | | Post | | Mixed model | | |
|---|---|---|---|---|---|---|---|
| | SCI | Non-SCI | SCI | Non-SCI | Time | SCI | Interaction |
| | | | | | *P*-value | *P*-value | *P*-value |
| **Acromio-humeral distance (mm)** | 11.9 (2.1) | 10.6 (1.2) | | | | | |
| **Biceps tendon** | | | | | | | |
| Thickness (mm) | 2.7 (0.5) | 3.3 (1.0) | 2.7 (0.4) | 3.6 (1.0) | 0.867 | 0.766 | 0.158 |
| Occupation ratio (%) | 23.1 (3.2) | 32.0 (12.1) | | | | | |
| Echogenicity | 100.2 (19.6) | 111.3 (18.1) | 98.0 (17.3) | 107.5 (16.6) | **0.001*** | 0.335 | 0.222 |
| Echo ratio | 2.3 (0.6) | 1.9 (0.6) | 2.2 (0.6) | 2.0 (0.7) | 0.607 | 0.524 | 0.536 |
| **Supraspinatus tendon** | | | | | | | |
| Thickness (mm) | 4.3 (0.6) | 3.6 (0.5) | 4.2 (0.7) | 3.7 (0.4) | 0.387 | **0.038*** | 0.219 |
| Occupation-ratio (%) | 36.5 (7.05) | 34.8 (6.9) | | | | | |
| Echogenicity | 98.8 (10.4) | 98.2 (9.0) | 96.6 (11.4) | 93.5 (9.2) | 0.323 | 0.643 | 0.504 |
| Echo ratio | 1.2 (0.2) | 1.3 (0.3) | 1.3 (0.2) | 1.3 (0.5) | 0.181 | 0.808 | 0.454 |

*Significant p-values (alpha = 0.05) from the mixed-effects multilevel analysis are marked with bold values and an *.*

persons with non-SCI this was 3.74 mm (SE = 0.20, 95% CI = [3.35; 4.14]) ($p < 0.001$). In addition, it was found that biceps and supraspinatus tendon thickness were positively associated with the occupation ratio (biceps thickness: $\beta = 0.07$, $SE\beta = 0.01$, $p < 0.001$, 95% CI = [0.05; 0.09]; supraspinatus thickness: $\beta = 0.06$, $SE\beta = 0.01$, $p < 0.001$, 95% CI = [0.04; 0.08]). Alternatively, for the supraspinatus, echogenicity ($\beta = -0.78$, $SE\beta = 0.21$, $p < 0.001$, 95% CI = [−1.19; −0.37]), and echogenicity ratio ($\beta = -0.02$, $SE\beta = 0.01$, $p = 0.002$, 95% CI = [−0.03; −0.01]), were negatively associated with the occupation ratio.

## Shoulder Pain

Of the 12 participants, seven reported no shoulder pain (PC-WUSPI mean = 0.07 ± 0.19), whereas five reported shoulder pain (PC-WUSPI mean = 15.5 ± 14.0; range 7.9–40.4). Of these participants, two experienced unilateral and three experienced bilateral shoulder pain. Furthermore, the group of participants with shoulder pain consisted of three persons with SCI (PC-WUSPI mean = 20.5 ± 17.2; range 10.3–40.4), and two persons with a non-SCI (PC-WUSPI mean = 7.8 ± 0.1; range: 7.8–7.9). No relationships between tendon characteristics and shoulder pain were observed pre-exercise (**Table 3**). Post-exercise it was revealed that decreased echogenicity of the supraspinatus tendon was correlated with increased pain.

# DISCUSSION

This novel study demonstrated significant adaptations in tendon characteristics in 12 highly trained WR athletes with different impairments. Acute adaptations were demonstrated in the reduction in the echogenicity of the biceps tendon immediately following the GXT, pointing toward fluid inflow into the tendon (darker tendon). Chronic tendon adaptations are associated with the impairment of the athlete, athletes with SCI presented significantly thicker supraspinatus tendon as compared to athletes with non-SCI. In addition, a greater occupation ratio was positively associated with signs of chronic tendon degeneration. Finally, shoulder pain was only associated to supraspinatus echogenicity following the GXT. Overall, these findings are in line with the high metabolic activity of human tendons (38).

## Acute Tendon Adaptations

The significant acute reduction in the echogenicity of the biceps tendon following the GXT may be related to acute overload and inflammation referred to as reactive tendinopathy which differs from normal tendon adaptation to tensile load (39). With insufficient time to recover, such acute overload, can lead to chronic tendon degeneration or degenerative tendinopathy (39, 40).

In accordance with the present results, an earlier study that employed the same QUS demonstrated a reduction in echogenicity ratio of the biceps tendon in response to a competitive wheelchair basketball or WR game (26). Importantly, however, although the current study reported a reduction in the echogenicity, there was no change in the echogenicity ratio. This suggests that there was a reduction in both the grayscale of the biceps tendon and the muscle above the tendon and may be caused by an overall fluid shift to the arm impacting both muscle and tendon simultaneously. This on its turn, would be related to fluid mobilization rather than inflammation. As a result of the GXT, different changes in the muscle above the tendon may have occurred in the present study due to the potential for more rest in between propulsion bouts, and different movements when compared to actual game play (E.G. turning and ball handling) (26). Nevertheless, both studies support the notion that there are acute adaptations in the biceps tendon grayscale following straining propulsion in WR athletes that could play a role in the development of shoulder pathology and pain in this population.

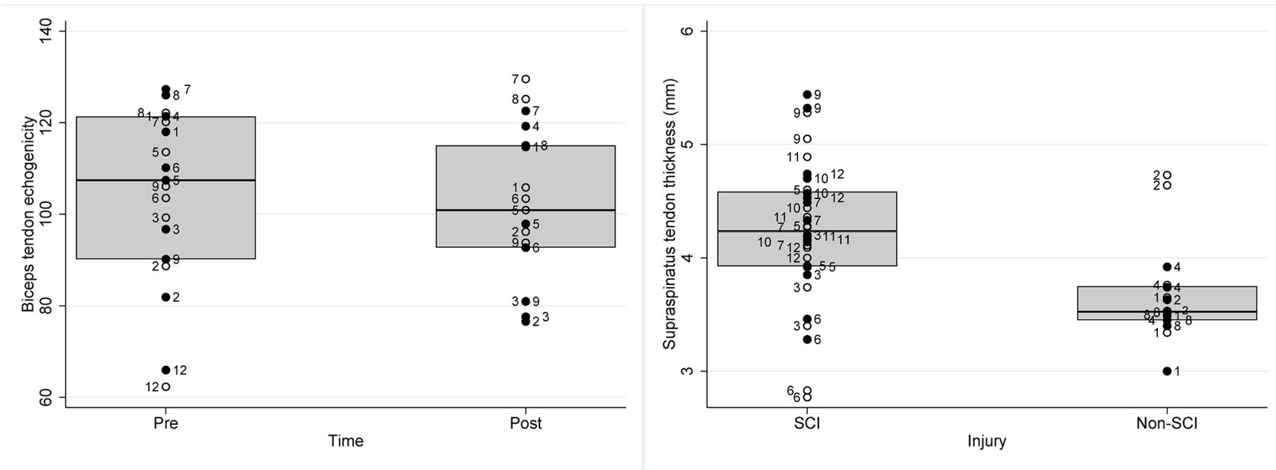


**FIGURE 2 |** Box plots representing raw data of individual wheelchair rugby athletes of (left) biceps tendon echogenicity pre and post a graded exercise test to exhaustion and (right) supraspinatus tendon thickness (mm) in athletes with SCI and non-SCI. Full circles with participant ID on the right side of the circle represents data of the dominant shoulder of the respective athlete, the empty circles with the participant ID on the left side of the circle represents data of the non-dominant shoulder of the respective athlete.

In line with the work of van Drongelen et al. (26), we did not demonstrate an increase in biceps tendon thickness following the GXT which is expected to coincide with acute overload. The current study investigated changes in the biceps and supraspinatus tendon in response to exercise including 22 ± 3.1 min submaximal propulsion and 10.2 ± 1.7 min maximal propulsion, and the study of van Drongelen et al. (26) investigated changes following game play varying between 10 and 70 min depending on the participants time on court. The duration of the activities in both studies may not have been long enough to induce an increase in tendon thickness. Like wheelchair propulsion, swimming is a repetitive sport that places great demands on the shoulder tendon structures while the AHD is reduced. To this effect, an acute increase in supraspinatus tendon thickness has been reported immediately post a high intensity swim training (3.5 km in 2 h) with smaller, but still significant increases in thickness in response to high volume swim trainings (7 km in 2 h) in eight state and national level swim athletes (41). Further research is needed to determine a potential increase in biceps tendon thickness with longer bouts of intense propulsion activity. Nevertheless, findings of this study support the added value of investigating the gray-scale of the tendon, which may be more sensitive to acute changes in reactive tendinopathy, rather than focussing on changes in tendon thickness only.

In contrast to earlier findings (25), this study did not present acute changes in the supraspinatus tendon following repetitive propulsion activity. More specifically, 15-min maximum voluntary propulsion resulted in a significant reduction in supraspinatus tendon thickness (25). A reduction in tendon thickness, a typical response to tensile loading, can be related to alignment of the tendon collagen fibers in the direction of the applied stress (42). A possible explanation for the different results may be related to the higher and complete lesion level of the persons with SCI in the current study [tetraplegia, 100% complete injury; (25): paraplegia, 78% incomplete], and GXT, which is likely to cause greater loads on the shoulder muscles and tendons and subsequently result in a different tendon response. In addition, the small sample size [$n = 12$; (25): $n = 50$] should be acknowledged and differences in wheelchair characteristics of the rugby and daily chair [rugby chair mass: 17.0 ± 1.4 kg with camber: 18.1 ± 1.8° vs. daily chair mass of Bossuyt et al. (25): 14.5 ± 2.1 kg with camber 0°], and subsequent altered position are likely to place different demands on the shoulder tendons. For example, fatiguing propulsion in wheelchair users' daily chair caused greatest signs of neuromuscular fatigue in the pectoralis, deltoideus, and upper trapezius (43) while current results and those of van Drongelen et al. (26) suggest that fatiguing propulsion in the rugby chair places greater demands on the biceps brachii tendon. This underlines the importance of the task-dependency of musculoskeletal loading and subsequent tendon adaptations.

## Chronic Tendon Adaptations

This study demonstrated that WR athletes with SCI had a thicker supraspinatus tendon in comparison to WR athletes with a different impairment. Increased tendon thickness may relate to chronic adaptations that causes tendon hypertrophy and strengthened the tendon by increasing its stiffness (12), or may be caused by chronic inflammation and indicate the presence of pathology (39). Interestingly, persons with SCI spent a significantly lower amount of time in the gym compared to non-SCI. With the lack of trunk function in the SCI group compared to the non-SCI group, it may be plausible that daily tasks (such as propelling and transferring into their wheelchair) may further increase the loads on shoulder tendons in SCI. Thus, despite the reduced gym exposure, hypertrophic adaptations may persist. Interestingly, values for the supraspinatus tendon thickness of

**TABLE 3 |** Association between acromio-humeral distance (AHD), tendon characteristics, and shoulder pain.

| | Pre | | Post | |
|---|---|---|---|---|
| | *r* | *p* | *r* | *p* |
| **AHD (mm)** | 0.311 | 0.139 | | |
| **Biceps tendon** | | | | |
| Thickness (mm) | –0.205 | 0.400 | –0.205 | 0.400 |
| Occ ratio (%) | –0.315 | 0.189 | | |
| Echogenicity | –0.043 | 0.861 | 0.151 | 0.563 |
| Echo ratio | 0.335 | 0.160 | 0.035 | 0.889 |
| **Supraspinatus tendon** | | | | |
| Thickness (mm) | 0.055 | 0.799 | –0.112 | 0.602 |
| Occ ratio (%) | –0.196 | 0.359 | | |
| Echogenicity | –0.279 | 0.186 | **–0.434** | **0.034*** |
| Echo ratio | –0.115 | 0.594 | –0.022 | 0.918 |

*An * and bold values are marked from the Pearson's correlations with significant p-values (α = 0.05).*

persons with a tetraplegia in the current study remain lower as compared to those reported previously in a sample of persons with a paraplegia (25). Persons with SCI also had a significantly lower $\dot{V}O_{2peak}$ further demonstrating differences in functioning between the two groups. Therefore, the previously established differences in volume of activity during rugby games based on functioning of the athletes may play a role in the different tendon adaptations (2).

While the percentage of athletes with shoulder pain in both groups was not markedly different (SCI: 3/8 athletes with pain, non-SCI: 2/4 athletes with pain) the average PC-WUSPI was higher in the SCI group than in the non-SCI group, PC-WUSPI remained below a score of 10. This could also be related to the lack of trunk support in the persons with a tetraplegic SCI thereby increasing loads on the shoulder. However, none of the US measures pre the GXT correlated with pain in the WR athletes which may be due to the small sample size in this study as this reduces the power of the study and increases the risk of type II error. A positive correlation between supraspinatus tendon thickness (defined with QUS) and supraspinatus pathology [Ultrasound Shoulder Pathology Rating Scale (USPRS)] has been established in wheelchair users with SCI (14). However, as far as we are concerned, no previous study compared tendon characteristics in wheelchair athletes with different impairments. In order to better understand the presented chronic adaptations, further imaging is needed to identify potential differences in tendon stiffness, and or inflammatory markers between wheelchair users with SCI and non-SCI impairments. The current findings demonstrate the need for an individualized approach and differentiation between impairments when monitoring tendon adaptations.

A greater occupation ratio for the biceps and supraspinatus was consistently associated with tendon characteristics that have been correlated with increased signs of tendinopathy in wheelchair users with SCI via the USPRS tendon grade (i.e., greater biceps and supraspinatus tendon thickness, and lower supraspinatus echogenicity and echogenicity ratio) (14). Interestingly, the occupation ratio of the supraspinatus remains smaller as compared to able-bodied persons with and without subacromial impingement syndrome (21, 22). It should be considered that the measures used to calculate the occupation ratio (i.e., supraspinatus and biceps tendon thickness and AHD) were taken from different US images with a different position for the supraspinatus tendon thickness. Nevertheless, our results are in line with previous studies that reported a greater occupation ratio in persons with subacromial impingement syndrome vs. healthy controls (21, 22) and support that a smaller space between the tendon and the acromion, or a greater occupation ratio, may be related to signs of chronic tendon degeneration. Therefore, the occupation ratio could be an interesting measure to include in the yearly screening of WR athletes.

## LIMITATIONS AND FUTURE DIRECTIONS

While a strength of this study was that our WR players had similar training histories and measures were taken at the same time-point within their training program, the small sample and heterogeneous nature of their injuries and functional capacities limits the generalizability of our findings. More specifically, the small sample size reduces the power of our study and increases the risk of type II errors. To account for the applied nature of this study, we chose a method that was easy and low-cost so it could be included in the monitoring program of WR athletes in the future. That said, it must be noted that the US images are limited in resolution and only allow two-dimensional measurements. Furthermore, it is important to acknowledge that without prior US imaging experience, it does require time and effort (∼25 h) to become proficient in taking and analyzing images following the QUS. The use of US elastography, a promising tool to define mechanical properties of the tendon including tendon stiffness, could have provided a more comprehensive understanding of the presented tendon adaptations. We are aware that we did not include a matched-control group, yet this may have been difficult due to the aforementioned heterogenous sample. Differences in hydration level could have impacted the US images and our overall results. Although we did not quantify the hydration level of each individual participant, the athletes, who receive educational support in terms of nutrition and hydration, were asked to arrive to the laboratory in a hydrated state. Furthermore, given the design of our study, the athletes acted as their own controls. We feel to advance our current knowledge it would be helpful to include multiple time-points following a rest period to observe changes in tendon adaptations following rest. It could also be of great benefit to include QUS measures to monitor tendon health longitudinally. Such assessments would allow to gain a better understanding of chronic tendon adaptations and asymmetries in WR athletes and the development of chronic degeneration in this population.

## CONCLUSION

There are acute biceps tendon adaptations in response to a GXT in highly trained WR athletes. The presented chronic tendon adaptations are associated with the impairment of the athlete (SCI vs. non-SCI) and the occupation ratio and may play a role in the high prevalence of shoulder problems in this population. Including such assessment methods in screening of wheelchair athletes may provide further insights into the long-term consequences of the reported changes and allow us to better understand and monitor shoulder health as well as to improve injury prevention.

## DATA AVAILABILITY STATEMENT

The datasets presented in this article are not readily available because the participants did not consent to sharing their data when they entered the study. Requests to access the datasets should be directed to fransiska.bossuyt@paraplegie.ch.

## ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Reference number SSEHS-2626 approved by Regulatory, Compliance and Safety Administrator Loughborough University. The patients/participants provided their written informed consent to participate in this study.

## AUTHOR CONTRIBUTIONS

FMB, BSM, SB, UA, and VLG-T initiated this study. FMB, BSM, SB, MLB, UA, and VLG-T contributed to the conception and design of the study. FMB, BSM, SB, and TJO'B performed the data collection and were responsible for the data analyses. FMB and BSM performed statistical analyses and drafted the paper. FMB finalized the paper. All authors interpreted the data, critically revised the paper, read, and approved the final paper.

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**Conflict of Interest:** The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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SYSTEMATIC REVIEW
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# Objective Measurement of Ball-Handling Proficiency in Wheelchair Sports: A Systematic Review

*Viola C. Altmann[1,2]*, Barry S. Mason[1], Tijmen Geurts[3], Sanne A. J. H. van de Camp[3] and Yves C. Vanlandewijck[4,5]*

*[1] Peter Harrison Centre for Disability Sport, Loughborough University, Loughborough, United Kingdom, [2] Klimmendaal, Rehabilitation Centre, Arnhem, Netherlands, [3] Department of Rehabilitation, Donders Institute for Brain, Cognition and Behaviour, Radboud University Medical Centre, Nijmegen, Netherlands, [4] Department of Rehabilitation Sciences, Katholieke Universiteit (KU) Leuven, Leuven, Belgium, [5] Swedish School of Sport and Health Sciences (GIH), Stockholm, Sweden*

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****Correspondence:***
*Viola C. Altmann*
*research@altmannen.nl*
*orcid.org/0000-0002-0671-8115*

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**Background:** In Paralympic sports, classification of athletes based on the impact of impairments on the ability to perform is needed, to prevent a one-sided and predictable outcome of the competition in which the least impaired athlete has the best chance to win. Classification is developing from expert opinion based to evidence based. In wheelchair court sports, there is evidence to support the impact of impairment on wheeled mobility, but not on ball handling. To assess the impact of impairment on the ability to perform ball-handling activities, standardised tests for ball handling are needed.

**Purpose:** To assess if reliable and valid standardised tests for the measurement of ball-handling proficiency in a wheelchair or able-bodied court sports exist; to assist in the development of Evidence-Based Classification (EBC) in wheelchair court sports according to the guidelines of the International Paralympic Committee (IPC).

**Methods:** The review was conducted according to the Meta-Analysis of Observational Studies in Epidemiology (MOOSE) statement. Search terms used were “wheelchair,” “ball,” “ball sports,” “test,” and “performance.” Databases searched were Medline, Embase, PubMed, and Sport Discus. Study quality was assessed using the Strengthening the Reporting of Observational Studies in Epidemiology checklist.

**Results:** Twenty-two articles were included. Foundational Movement Skills in ball-handling proficiency were assessed. Tests for throwing maximal distance showed sufficient reliability and validity. Precision in throwing showed low-to-moderate reliability and conflicting results in validity. Throwing techniques differed between studies. Dribbling the ball showed high reliability, but conflicting results in validity.

**Conclusions:** Tests for throwing maximal distance, throwing precision, and dribbling the ball can be used in standardised tests for activity limitation in wheelchair court sports. However, tests need to be adapted and standardised and then reassessed for reliability and validity in athletes with and without arm impairment.

**Keywords: wheelchair, ball handling performance, wheelchair sports, classification, sport specific activities, systematic review**

## INTRODUCTION

The Paralympic Games are the third-largest sporting event in the world and provide an excellent platform to enhance participation and inclusion of persons with impairments in society (1). However, the value and the success of the Paralympic Games would become questionable, and the goal of participation would not be achieved if athletes who win the competition are simply the least impaired athletes. To prevent this, classification systems grouping athletes with impairments with a similar impact on performance in sports have been developed and applied since the start of the Paralympic Games (2). Typically, these classification systems were developed based on expert opinion by volunteer classifiers with a medical background and/or sport-specific expertise (3). The success of an athlete in competition can depend for a significant part on the class in which the athlete is competing. With the increasing professionalism of the Paralympic movement and the Paralympic athletes, a classification system based on expert opinion was no longer sufficient to support the value and success of the Paralympic Games, both for the athletes and society (4).

In 2007, the International Paralympic Committee (IPC) published the IPC Classification Code and International Standards to provide a structure for classification principles for all Paralympic Sports. In this Code, international sports federations were charged with the development of Evidence-Based Classification (EBC) systems through research (5). EBC means that the methods used to allocate sports classes must be based on scientific research, which demonstrates that the aim of classification, to group athletes for competition based on impairment severity with a similar impact on sport-specific performance, is achieved. The development of EBC systems requires four steps: (1) defining eligible impairment types per sport, (2a) developing valid and reliable measures of impairment, (2b) developing valid and reliable measures of determinants of sport-specific performance, and (3) assessing the relationship between impairment and performance determinants to define sports classes. Both the measures of impairment and performance determinants should be highly standardised, objective/instrumented, and ratio scaled where possible (6).

There are three wheelchair team court sports, wheelchair rugby (7), wheelchair basketball (8), and the newly developed sport wheelchair handball (9). The eligible impairment types for these three sports are neuromusculoskeletal impairments (strength, range of motion, coordination, and limb deficiency). Furthermore, these three sports have many commonalities in the activities that determine proficiency in the game. Based on the concept of Fundamental Motor Skills (10), these activities consist of locomotor skills, i.e., wheeled mobility, which is specific for wheelchair sports, and object control, i.e., ball handling, which shows much overlap with able-bodied sports. The term object control/ball handling is elaborated in the model of Foundational Movement Skills and consists of throwing with several techniques, bouncing/dribbling, and catching (11). Despite all commonalities, the classification systems of each of these sports, such as the eligibility criteria, the number of classes, and the criteria that define these sports classes, are completely different (12–14). Of the three classification systems, only the wheelchair rugby classification system is partially evidence-based (15). The evidence that is generated to support wheelchair rugby classification can potentially benefit the development towards EBC in wheelchair basketball and wheelchair handball. So far, wheelchair rugby classification is supported by evidence for trunk impairment in strength, range of movement and coordination (16, 17), and arm strength impairment (18). However, the relationship of these impairments with performance is only determined for wheeled mobility and not for ball handling (16–18). Therefore, in the interest of continuing the development towards EBC for all three wheelchair team court sports, step 2b develops valid and reliable measures of determinants of sport-specific performance, needs to be completed with tests for ball-handling proficiency. The definition we will use for ball handling in this study is based on the Foundational Movement Skills and consists of throwing with several techniques, bouncing/dribbling, and catching. These activities need specifications for wheelchair court sports. In wheelchair court sports, ball handling is restricted to handling a round ball with a size that is suitable for one- and two-handed direct manual ball handling without a device (like a bat or a stick). For throwing, both maximal distance and precision will be included as important aspects for proficiency.

The present study aimed to identify standardised tests for proficiency in ball handling according to the previously mentioned definition in team court sports (both Olympic and Paralympic) from the literature. The second aim was to assess if there is any evidence for the reliability and/or validity of these tests. The third aim was to determine if any or a combination of these tests can serve as a standardised test for ball-handling proficiency for the future development of EBC in wheelchair court sports.

## MATERIALS AND METHODS

This systematic review was conducted and reported according to the consensus statement for the Meta-analysis Of Observational Studies in Epidemiology (MOOSE) (19), because based on the research question, the authors mainly expected to find observational studies. Two researchers performed the article search independently (TG and SC).

### Data Sources

Original articles were searched in Medline (1946–2020), Embase (1974–2020), PubMed (1989–2020), and Sport Discus (1949–2020). The following search terms were used for able-bodied sports: ball sports, performance, test, and arm or trunk. For wheelchair sports, wheelchair, ball, and performance were used as search terms. Search terms were linked with the Boolean AND. The search was extended using the option "related articles" in all databases. First, the title and abstract of the related articles were screened. If the title and the abstract met the inclusion criteria, the article was added to the numbers of identified records. In addition, the grey literature was also explored to ascertain whether any other articles outside the original search matched the criteria.

## Study Selection

Inclusion criteria for articles were (1) the outcome measures were ball handling with a round ball with a size that is suitable for one- and two-handed direct manual ball handling without a device and were presented in objective, quantitative data, (2) assessment for reliability was done, i.e., test-retest or inter-rater reliability and/or assessment of validity was done by the following comparisons: (a) between athletes playing a sport at different competition levels, (b) athletes with differences in age, (c) athletes with different physical characteristics of the arm or trunk (able-bodied sports), or (d) comparisons between participants with different levels of impairment (wheelchair sports), and (3) articles were written in English. Furthermore, for studies about able-bodied sports, (4a) participants were experienced athletes without impairments. Moreover, for wheelchair sports, (4b) the participants were experienced (sport) wheelchair users. To identify eligible articles, two reviewers independently screened the title and the abstract. If one reviewer found an article, both researchers screened for inclusion criteria. If the abstract met the criteria, then both researchers assessed the full text of the article. The article was assessed by a third researcher (VA) when there was a disagreement between the two reviewers. In this case, the third party had the final vote. If an article was found in more than one database, it was only included once.

## Quality Assessment

The methodological quality of each study was assessed independently by two reviewers (TG and SC) using the Strengthening The Reporting of Observational Studies in Epidemiology (STROBE) checklist for reports of observational studies (20). The STROBE checklist has 22 items, which were either scored as present ("1") or absent ("0"). If one of the items had any sub-items, one point was awarded if the study met half or more of these sub-items. When disagreement existed on any item of the STROBE checklist, the same consensus procedure applied for inclusion criteria was used with three authors (TG, SC, and VA). The STROBE recommendations do not provide a guideline for including meaningful studies in a systematic review. However, in a study performed on the methodological quality of observational studies published in high-quality journals, an average of 69% of the STROBE items were reported (21). Consistent with this study and with others using quantitative cut-off scores for observational studies, we decided that a minimum of 15 reported items (69%) indicated "good quality," whereas 14 reported items or less indicated "moderate-to-low quality." Only studies with "good quality" were included in the discussion.

# RESULTS

## Search Results

**Figure 1** shows the number of articles found following each step of the search strategy. After database searching and searching additional sources, the researchers found 301 potentially relevant studies. After assessment for the inclusion criteria based on screening of titles and the abstracts, and if indicated, assessment of the full article, the researchers reached a consensus that 30 articles were eligible for methodological quality assessment. Most articles were excluded because the outcome measures were not based on ball handling as defined in the present study (handling a round ball with a size that is suitable for one- and two-handed direct manual ball handling without a device), or because there was no assessment of validity or reliability.

## Methodological Quality

The results of the quality assessment of the 30 articles that were eligible based on the inclusion criteria are shown in **Table 1**.

## Findings of the Review

Twenty-two articles fulfilled the predetermined minimum of 15 reported items on the STROBE checklist (see **Table 1**) (22–51). They all had a cross-sectional design. Twelve high-quality studies were about able-bodied sports, 11 about handball (23, 29, 31, 34–36, 42–48), and 1 about basketball (32). Two articles (23, 24) were about the same study, of which only one was "high quality." 10 high-quality studies were about wheelchair sports, 9 about wheelchair basketball, (26, 35, 37–40, 46, 49, 51) and 1 about wheelchair rugby (49). Wheelchair athletes had health conditions, such as spinal cord injury, spina bifida, cerebral palsy, neuromuscular conditions, and congenital and acquired amputations, leading to all eligible impairment types for wheelchair rugby, wheelchair basketball, and wheelchair handball (strength impairment, coordination impairment, impaired range of movement, and limb deficiency)**.**

In all but one study, throwing for accuracy and distance using different throwing techniques was assessed. The most frequently used outcome parameters for throwing were ball velocity (12 tests) and throwing distance (13 tests). For accuracy, 12 studies used the number of scores on the target; the target was usually a basketball bucket. Only one study measured throwing accuracy in continuous data, the surplus in centimetres by which a projected target was missed, but this study was rated as moderate-poor quality (24). In 10 studies, (26, 32, 33, 37, 38, 41, 46, 47, 49, 51) running or pushing with the ball, such as the dribbling rules of the game, was assessed. In addition, in four of these studies, picking up the ball from the floor during wheelchair pushing was assessed (33, 37, 38, 49). All these tests involved running or wheeling, and the time for completion of the test was used as the outcome parameter. In only one study catching was included (50).

Test-retest reliability was included in seven studies (37, 38, 40, 43, 45, 48, 50). For outcomes of these studies, see **Table 2**.

Test-retest reliability was high for all tested items of ball proficiency, except for throwing accuracy.

All but one study included a measure for validity. For outcomes of these studies, see **Table 3**.

The groups that were compared were either based on performance (competition divisions, player ranking by experts, or national vs. international players) or on impairment classes. For maximal throwing distance, there was a difference between groups in almost all studies. The only exceptions were national vs. international athletes in one study and impairment classes in one out of five studies. No differences between groups were found in many studies for throwing accuracy and ball handling while

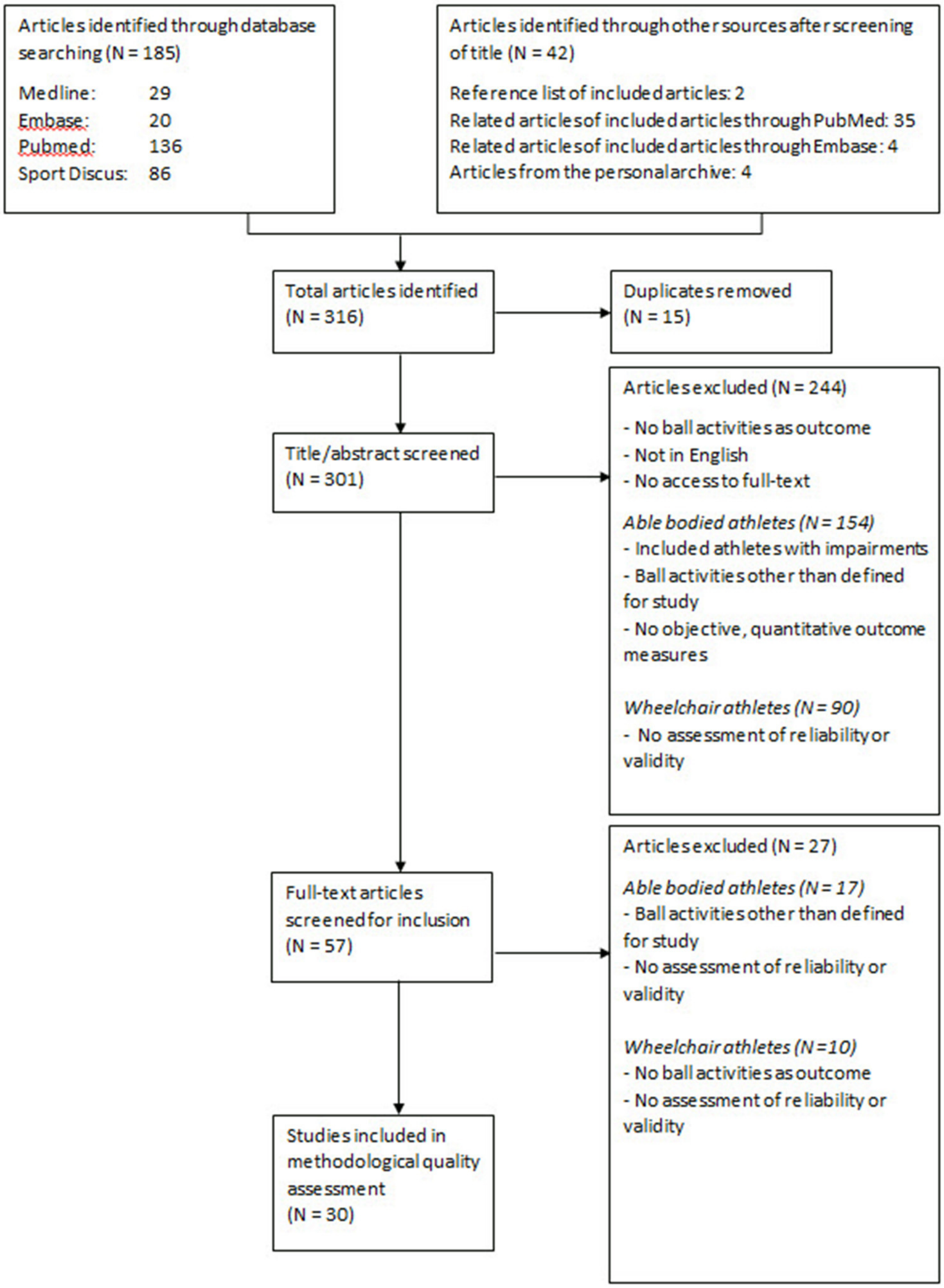


**FIGURE 1 |** Flowchart of the literature search.

running or pushing. There were no differences between groups in one study for catching the ball due to an important ceiling effect.

## DISCUSSION

### Summary of the Evidence

In this systematic review, we synthesised the evidence on the reliability and validity of standardised tests for ball-handling proficiency in court sports, available in the literature. As anticipated, all studies identified were observational studies. In the studies about wheelchair sports, all impairment types that are eligible for wheelchair rugby, wheelchair basketball, and wheelchair handball were included. The evidence indicated that tests for maximal throwing were both reliable (37, 38, 40, 43, 45, 48, 50) and valid (23, 26, 29, 31, 33–37, 39, 42–44, 46, 49) in relation to competition level, anthropometric data, and impairment classes. Tests for throwing accuracy lacked reliability (38, 45, 48), and validity showed conflicting results (26, 38, 50,

**TABLE 1 |** Study participants, interventions, comparisons, and STROBE scores.

| References | Participants | Intervention | Comparison | Total score STROBE | Methodological quality |
|---|---|---|---|---|---|
| Barfield et al. (22) | *WC rugby*<br>10 WRP national team and 10 WRP not national team | Long pass<br>Short pass<br>Slalom with ball | WRP national team, WRP not national team | 13 | Moderate—poor |
| Bayios et al. (23) | *AB handball*<br>15 FD, 12 SD male HP and 15 PE students | Throwing on the spot and with cross-over step and vertical jump shot | FD, SD, PE students | 15 | High |
| Bayios et al. (24) | *AB handball*<br>15 FD, 12 SD, 15 PE students | Ball throw on the spot and with cross-over step | FD, SD, PE students | 13 | Moderate—poor |
| Borges et al. (25) | *WC handball*<br>21 WHP Low point (1.0–1.5) 7 athletes<br>Midpoint (2.0–2.5) 6 athletes<br>High point (3.0–4.0) 9 athletes | Slalom with ball | Low, mid and high point WHP | 14 | Moderate—poor |
| Cavedon et al. (26) | *WC basketball*<br>Class A (0.5) 18 athletes<br>Class B (1.0–1.5) 16 athletes<br>Class C (2.0–2.5) 8 athletes<br>Class D (3.0–4.0) 9 athletes | Maximal pass, pass for accuracy, spot shot, lay-ups, 20 m sprint with ball | FAC 1 vs. 2 vs. 3 vs. 4 | 19 | High |
| Cerrah (27) | *AB soccer*<br>14 male soccer players | Throwing the ball in while standing and running | Isokinetic strength of the arms (shoulder and elbow) and the trunk flexion and extension | 12 | Moderate- poor |
| Costa e Silva (28) | *WC handball*<br>29 WHP Group 1 (1.0–1.5) 6 players<br>Group 2 (2.0–2.5) 8 players<br>Group 3 (3.0–3.5) 6 players<br>Group 4 (4.0–5.0) 9 | Throwing against the wall and catching<br>Slalom with the ball | Group 1, 2, 3 and 4 | 10 | Moderate- poor |
| Debanne et al. (29) | *AB handball*<br>12 high national, 17 high regional and 13 local male HP | Standing overarm throw | No group comparisons | 17 | High |
| Erculj et al. (30) | *AB basketball*<br>23 division A European players<br>25 division B European players | Basketball throw<br>Medicine ball (2 kg) throw | Division A vs. division B players | 8 | Moderate- poor |
| Fieseler et al. (31) | *AB handball*<br>12 FD, 34 TD male HP | Throws with run-up or jump overarm throw with and without precision | FD vs. TD | 20 | High |
| Garcia-Gil et al. (32) | *AB basketball*<br>41 FD female BP from 4 teams in first division national league Spain with varying placements | Dribbling test | 4 FD teams with varying placements | 17 | High |
| Gil et al. (33) | *WC basketball*<br>13 WBP Class 1.0, 1 athlete; class 1.5 1 athlete; class 2.0 3 athletes; class 2.5 1 athlete; class 3.0 2 athletes; class 3.5 2 athletes; class 4.0 2 athletes; class 4.5 1 athlete | Pick-up the ball, maximal pass with basketball and medicine ball (5 kg), 20 m sprint with the ball | Athlete class and injury type (SCI or non-SCI), years of experience in WC and years of experience in WC basketball | 19 | High |
| Gorostiaga et al. (34) | *AB handball*<br>15 FD and 15 SD male HP | Standing throw and 3-step running throw | FD vs. SD<br>And correlation with arm strength and power production | 18 | High |
| Granados et al. (35) | *AB handball*<br>16 FD and 15 SD female HP | Standing throw and 3-step running throw | FD vs. SD | 18 | High |
| Granados et al. (36) | *AB handball*<br>16 national and 14 international female HP | Standing throw and 3-step running throw | National vs. international | 20 | High |
| Granados et al. (37) | *WC basketball*<br>19 FD and TD WBP | Anthropometric and performance values. Ball pick-up, maximal pass with | FD vs. TD | 18 | High |

*(Continued)*

**TABLE 1 |** Continued

| References | Participants | Intervention | Comparison | Total score STROBE | Methodological quality |
|---|---|---|---|---|---|
| | | two arm overhand with basketball and medicine ball (5 kg), 20-m sprint with ball including dribble | | | |
| De Groot et al. (38) | *WC basketball*<br>19 WBP Class 1.5, 2 athletes<br>Class 2.0, 2 athletes<br>Class 2.5, 3 athletes<br>Class 3.5, 1 athlete<br>Class 4.0, 5 athletes<br>Class 4.5, 6 athletes | Pass for accuracy, free-throw shooting, 20 m sprint with the ball, maximal pass, lay-ups, pick-up the ball, spot shot, | Premier league vs. tournament A vs. tournament B. Trial 1 vs. 2. | 18 | High |
| Marszalek et al. (39) | *WC basketball*<br>29 class A (1.0–2.5), 29 athletes<br>class B (3.0–4.5) 32 athletes | Basketball chest pass test | Class A vs. class B | 16 | High |
| Marszalek et al. (40) | *WC basketball*<br>9 WBP | Two handed pass basketball and medicine ball (3 kg) | The first vs. second repetition of the tests | 16 | High |
| Molik et al. (41) | *WC basketball*<br>109 WBP Class 1, 26 athletes<br>Class 2, 25 athletes<br>Class 3, 24 athletes<br>Class 4, 16 athletes<br>Class 4.5, 18 athletes | Two handed chest pass<br>Slalom with the ball | Differences between athlete classes | 12 | Moderate-poor |
| Moss et al. (42) | *AB Handball*<br>47 non-elite, 44 elite and 29 top-elite female youth HP | Standing throw and 3-step running throw | Top-elite, elite and non-elite | 17 | High |
| Ortega et al. (43) | *AB Handball*<br>13 elite, 16 U18 and 16 U16 male HP | 3 step running throw and jump throw | Elite, U18 and U16 | 18 | High |
| Saavedra et al. (44) | *AB Handball*<br>23 A-team, 16 U19, 20 U17 and 21 U15 national team female HP | Standing throw | A-team, U19, U17, U15 | 16 | High |
| Schwesig et al. (45) | *AB Handball*<br>30 male TD HP | Bal throwing with cross-step and throwing time | No group comparisons | 18 | High |
| Tachibana et al. (46) | *WC basketball*<br>Class 1 (1.0–1.5) 7 athletes<br>Class 2 (2.0–2.5) 7 athletes<br>Class 3 (3.0–3.5) 5 athletes<br>Class 4 (4.0–4.5) 8 athletes | Figure of eight with ball, pass for distance in chest-pass, baseball-pass and hook-pass | Wheelchair basketball class | 18 | High |
| Visnapuu et al. (47) | *AB Handball*<br>34 10–11 year, 39 12–13 year, 39 14–15 year and 21 16–17 year old male HP | 30 m dribble test, handball throw from sitting position, passing on speed and precision | 10–11, 12–13, 14–15 and 16–17 years old | 14 | Moderate—poor |
| Wagner et al. (48) | *AB Handball*<br>5 FD, 12 FoD and SiD male HP | Game based performance test including catching and passing ball as fast as possible | 2 tests separated by 7 days | 18 | High |
| Yanci et al. (49) | *WC basketball*<br>14 males, 2 females Category A (class 1.0–2.5) 7 athletes<br>Category B (class 3.0–4.5) 9 athletes | Pick-up the ball, maximal pass, 5 and 20 m sprint with ball | FAC A (1.0 to 2.5) vs. FAC B (3.0 to 4.5) | 16 | High |
| Yilla et al. (50) | *WC rugby*<br>65 WRP with quadriplegia. 60 had spinal cord injuries, 2 poliomyelitis, 1 muscular dystrophy, 1 charcot-marie-tooth syndrome, 1 cerebral palsy | Pass for accuracy, catching, pass for distance | The first vs. second repetition of the tests<br>Players rank, determined by experts | 18 | High |
| Yüksel et al. (51) | *WC basketball*<br>12 FD, 9 SD WBP | Pass for distance, lay-up tests, zone shot test, slalom with ball, pass for accuracy test | FD vs. SD | 16 | High |

*STROBE, strengthening the reporting of observational studies in epidemiology; WC, wheelchair; AB, able bodied; FD, first division; SD, second division; TD, third division; FoD, fourth division; SiD, sixth division; HP, handball players; BP, basketball players; WBP, wheelchair basketball players; WRP, wheelchair rugby players; WHP, wheelchair handball players; FAC, functional ability class; U, under (age limit); PE, Physical Education. Articles with a total STROBE score of ≥ 15 were included in the study.*

**TABLE 2 |** Reliability.

| Ball proficiency item | Outcome parameter | Reliability | Study numbers* |
|---|---|---|---|
| Throwing maximal distance | Throwing velocity (m/s) or distance (m) | High | (**37, 38, 40**, 43, 45, 48,**50**)<br>(**37**) (medicine ball 5 kg)<br>(**40**) (medicine ball 3 kg) |
| Throwing accuracy | Score in basket or goal (n) | Moderate-low | (**38**,45, 48) |
| Ball handling while pushing | Time to complete trajectory (s) | High | (**37, 38, 50**) |
| Catching the ball | Balls caught (n) | High | (**50**) |

**Study numbers with participants without impairments/running sports.*
***Study numbers with participants with impairments/wheelchair sports**.*

51). Besides, the correlation with player ranking as a measure for game performance in wheelchair rugby was low (50). Dribbling or bouncing the ball during running or wheeling showed high reliability (37, 38, 50), but validity showed conflicting results (26, 32, 33, 37, 38, 49, 50). Finally, there was only one study in which catching was included (50). This test showed high reliability but the validity was limited by a large ceiling effect in wheelchair rugby athletes.

Tests for maximal throwing distance showed both adequate reliability and validity for potential use as a measure for sport-specific ball-handling proficiency that can be used in the development for EBC (23, 26, 29, 31, 33–40, 42–46, 48–50). The outcome measure can be distance, (26, 33, 39, 46, 49) which can be measured with limited equipment. However, this test requires a rather large testing area as distances of more than 15 m can be thrown (46). Limitations of room size can be addressed by using a medicine ball (3–5 kg) instead of a normal ball, which reduces the maximal distance to ∼5 m. (37). However, athletes with severe arm and trunk impairment may not be able to throw such a heavy ball. Another good alternative is using throwing velocity as the outcome measure (23, 29, 31, 34–36, 42–44). However, this requires equipment, such as like laser beam emitters and laser beam infrared detectors, (23) a Doppler-radar gun (29, 42, 44), a speed check radar device, (31, 43, 45) or photocells (34–36). However, the objectivity and precision of measuring velocity may be superior to measuring distance, as the distance was measured with a tape measure where the ball was observed to touch the floor instead of with instrumented equipment.

In studies about able-bodied sports, the throwing technique was an overarm, one-handed pass with the dominant arm in either a standing throw, three-step running throw, cross-over-step throw, or a jump throw (23, 29, 31, 34–36, 42–45). In studies about wheelchair sports, all tests were performed standing still, and several throwing techniques were used. Most studies included a chest pass (39, 40, 46) or two-arm overhand pass (37). In several studies, the technique was not specified (26, 33, 38, 49, 50). In only one study, one-handed passes (baseball and hook pass) were assessed (46). Because arm and hand impairment, which is present in all wheelchair rugby players and in part of the wheelchair basketball and wheelchair handball players, can impact the throwing technique, the throwing technique should be standardised, and preferably, both two-handed and one-handed techniques should be included.

Based on the findings in the literature, we advise including maximal throws in the standardised test of ball-handling proficiency for testing of sport-specific performance in wheelchair court sports. Preferably, maximal throws should include standardised techniques for two-handed and one-handed throws. Outcome measurement in ball velocity has advantages over distance in measurement precision and the room needed for the tests.

Throwing distance in court sports is meaningless if the throw does not reach a target. Throwing accuracy is important for successfully passing the ball to other players resulting in a catch in all three wheelchair court sports and for scoring a goal in wheelchair basketball and wheelchair handball (7–9). Throwing accuracy was assessed in several studies (26, 38, 45, 48, 50, 51). In three studies, the number of scores in the bucket was used as an outcome parameter (26, 38, 51), and in two scores in a handball goal (45, 48). Scoring or not scoring is a binary parameter and the difference between scoring and not scoring can be minimal. In only one study, circles around a goal with a maximum score in the middle circle and decreasing scores in the outer circles were used (50). However, this still results in a score on an ordinal scale, where a ratio scale is advised (6). In one study, two-handed throws were used (51), and in two studies, one-handed throws were used (45, 48). In the other studies, the throwing technique was not specified (26, 38, 50). The reliability for throwing accuracy was low in all studies in which this was assessed (38, 45, 48). Perhaps this is due to the binary or ordinal scale that was used, in which a difference of several millimetres in a throw of several metres can make the difference between a score and no score or scoring points. If an interval scale would be used, measuring the surplus from the goal in centimetres of millimetres, a variation of several centimetres or millimetres between throws will result in less difference between measures than "hit" or "no hit," and reliability would increase. Furthermore, scoring as an outcome measure for throwing accuracy, using only one throwing technique is rather limited in comparison to the repeated throws between players and multiple throwing techniques that can be used in a game (7–9). It is striking that the only study in which catching was assessed showed a large ceiling effect (50). Accuracy of a throw plays an important role in catching the ball. However, in the study, the precision of the throw after which the ball needed to be caught was not specified. Maybe the high precision of the throw

**TABLE 3 |** Validity.

| Ball proficiency item | Comparison | Outcome parameter | Difference between groups (yes/no/conflicting) | Study numbers* |
|---|---|---|---|---|
| Throwing maximal distance | Competition divisions | Throwing velocity (m/s) | Yes | (23, 31, 32, 42, 44) |
| | Competition divisions | Throwing distance (m/s) | Yes | (**26, 37, 38**)<br>(**37**) (medicine ball 5 kg) |
| | Strength or anthropometric data | Throwing velocity (m/s) | Yes | (23, 29, 31, 34–36, 42–44) |
| | National- international athletes | Throwing distance (m) | No | (36) |
| | Impairment classes | Throwing distance (m) | Conflicting | (**26, 33, 39, 46, 49**) (yes)<br>(**38**) (no) |
| Throwing accuracy | Divisions | Score in basket or goal (n) | Yes | (**38, 51**) |
| | Impairment classes | Score in basket or goal (n) | Conflicting | (**26**) (yes)<br>(**38**) (no) |
| | Player ranking by experts | Score in goal | No | (**38, 50**) |
| Ball handling while pushing or running | Divisions | Time to complete trajectory (s) | Conflicting | (**38**) (yes)<br>(32,**37, 49**) (no) |
| | Impairment classes | Player ranking by experts | Conflicting | (**33**) (only difference for athletes with spinal cord injury, but not for athletes with other health conditions)<br>(**26**) (no) |
| | Player ranking by experts | Player ranking by experts | Yes | (**50**) |
| Catching the ball | | Balls caught (n) | No | (**50**) (ceiling effect, in which 90% of all athletes achieved maximum score) |

**Study numbers with participants without impairments/running sports.*
***Study numbers with participants with impairments/wheelchair sports*.**

explains the ceiling effect that was found in this test. A more game-specific measure of throwing accuracy may be throwing at a target and the outcome measure is the surplus (proximity to the target in cm), measured in an interval scale. The latter was done in one study (24). However, this research article had moderate-to-low quality and the measurement device, and the methods were not described clearly enough to be repeated. Based on the findings in the literature, methods for throwing accuracy need to be developed and reliability and validity need to be assessed. Throwing at a target using surplus as the outcome parameter in an interval scale seems an interesting option to test throwing accuracy. Similar to tests for throwing distance, tests for throwing accuracy preferably should include standardised techniques for two-handed and one-handed throws. In addition, there may be a relationship between throwing distance and throwing accuracy, in which accuracy decreases if the throwing distance increases to the maximum throwing distance. However, this was not assessed in any of the studies. Tests for throwing accuracy using different percentages of the maximal throwing distance may reveal such a relationship, which is very important for proficiency in ball handling. Furthermore, we advise including tests for catching after standardised throws with more or less velocity and precision, such as can be done by a ball launcher.

Finally, dribbling or bouncing the ball within the game rules, while moving the wheelchair is an activity that contributes to proficiency in wheelchair court sports. Dribbling the ball while running or moving the wheelchair was assessed in several studies (26, 32, 33, 37, 38, 49, 50), of which the ones about wheelchair basketball (26, 33, 37, 38, 49) and wheelchair rugby (50) will be most specific for wheelchair court sports. Reliability was high for picking up the ball (37, 38), 20 m sprint with the ball (37), and manoeuvrability with the ball (50). Assessment for validity showed promising results in several studies about wheelchair sports. Differences were found between impairment classes, but only for athletes with spinal cord injuries (33), and between divisions (38) and player ranking (50). However, in four studies, no differences were found between impairment classes (26, 38, 49) and between divisions (37). Impairment classes in wheelchair basketball are defined by trunk active range of movement (13), and it is known from previous studies that the velocity in pushing a wheelchair largely depends on impairment in arm muscle strength (16, 18). Because the outcome measure was time to complete the test and pushing made up a large part of the time to complete the test, this may have obscured any differences between impairment classes in ball handling during pushing. Performing two tests on the same circuit, one with and one without dribbling and bouncing the ball and then subtracting the results of these tests could minimise the impact of pushing the chair and give more insight into the component of dribbling and bouncing the ball (52).

If athletes with different severities of arm and hand impairment will be included, it is likely that differences in test performance will be found, which will support the validity of tests for dribbling and bounding the ball.

Based on these findings, we advise including standardised tests for dribbling or bouncing the ball while moving the wheelchair in a test battery for ball-handling proficiency. The same circuit should be done with and without the ball handling and the times to complete the test should be subtracted to eliminate the impact of pushing on the outcome from the test. Athletes can push in a straight line or in a circuit and ball handling should include picking up the ball and dribbling. Validity in relation to impairment severity needs additional assessment including athletes with arm impairment.

### Strengths and Limitations

This systematic review has several strengths. First, the strict study protocol using the MOOSE standard for meta-analysis (19) enables replicating the study and extending it in the future if new evidence will become available. Second, several types of bias were considered and minimised. Publication bias was minimised by extending the search to the grey literature. Both bias in selection for study inclusion and bias in the methodological quality assessment were addressed, respectively, by independent literature searches and independent assessment of study quality by two researchers (SC and TG). Finally, bias in results and conclusions was minimised by using the STROBE guideline for methodological assessment of observational studies (20).

There are several limitations that need to be considered when interpreting the results of this literature review. First of all, the studies included had rather small study populations, ranging from 14 to 120 participants in able-bodied sports (27, 42), and 13 to 65 participants in wheelchair sports, (33, 50) which limited the power of each of the studies. This may have obscured differences between groups for the assessment of validity, especially in activities with limited reliability, such as throwing precision. Pooling of the data from several studies to increase the power was not possible, because study populations, throwing techniques, measurement techniques, and outcome measures were different across the studies.

There may have been biases within the studies that were included, based on several issues. In most studies about wheelchair sport, the relationship between the activity and the wheelchair basketball classification was used as a measure for severity of impairment (26, 33, 38, 39, 46, 49). However, these wheelchair basketball classes are not an evidence-based measure of impairment severity (11, 13). Furthermore, several wheelchair basketball classes were grouped for analysis, to increase the number of athletes per group (26, 39, 46, 49). This may have increased the variation of impairment severity within groups, which may have obscured any difference between groups even more. In several studies, the technique for throwing, dribbling, and picking up the ball was not specified (26, 33, 37, 38, 49, 50), and therefore may not have been the same for all participants. This may have limited the variation in the outcome measures for throwing velocity and precision between groups because participants could compensate for their limitations by altering the throwing technique. Furthermore, it was not always clearly described if athletes were allowed to use equipment, such as sticky material on the hands or the ball in handball and gloves in wheelchair rugby. This may also have limited the differences between groups for both throwing velocity and precision. Last but not the least, only studies with experienced athletes were included in this review. However, the levels of experience and training were different across the studies and ranged from recreational athletes training only once a week (26) to elite international athletes competing at the highest level (43). This may have caused considerable variation within groups, obscuring differences between groups. All these forms of bias may have affected the conclusions about the reliability and the validity of the studies. For the development of a standardised test battery for ball-handling proficiency, it will be important to recruit enough optimally trained athletes to participate for sufficient study power. This will be a challenge, because the number of elite wheelchair athletes is limited, and they are spread geographically. However, because aspects of ball-handling proficiency are similar for the three wheelchair team court sports, combining athletes from these sports may help overcome this obstacle. Furthermore, ball activities should be standardised, such as throwing, ball pick-up and dribbling techniques, and the use of equipment.

## CONCLUSIONS

The findings of this review provide valuable information for the development of a standardised test battery for ball activities in team wheelchair court sports. Based on the findings, we advise a test battery, which includes at least all Foundational Movement Skills for ball handling, throwing with several techniques, bouncing/dribbling, and catching. Throwing should include standardised throwing techniques with at least a two-handed and a one-handed throw. Throwing should be assessed for maximal distance and accuracy including the relationship between distance and accuracy. Bouncing/dribbling the ball should include a standardised pushing distance and trajectory, i.e., picking up and dribbling the ball. This activity can be assessed using execution time if the impact of pushing on the test is minimised. Assessment of the test-retest reliability and the validity of the test battery needs to be assessed before this test battery can be used in the steps in the development of EBC that follow steps 1) define eligible impairment types and 2a) develop valid and reliable measures of impairment. These steps are 2b) developing valid and reliable measures of sport-specific performance, and 3) assessing the relationship between impairment and performance to define sports classes.

## DATA AVAILABILITY STATEMENT

The original contributions presented in the study are included in the article/supplementary material, further inquiries can be directed to the corresponding author/s.

## AUTHOR CONTRIBUTIONS

VA, BM, and YV formulated the research question and they established the study design. VA, BM, TG, SC, and YV contributed to the manuscript, discussed the study results, and the relevance with regards to the research questions. TG, SC, and VA performed the check for inclusion criteria and the assessment of the quality of the selected studies. TG and SC performed the literature search. All authors contributed to the article and approved the submitted version.


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# Learning of Wheelchair Racing Propulsion Skills Over Three Weeks of Wheeling Practice on an Instrumented Ergometer in Able-Bodied Novices

*Rick de Klerk*[1†], *Gabriëlle van der Jagt*[1†], *Dirkjan Veeger*[2], *Lucas van der Woude*[1,3*] and *Riemer Vegter*[1,4]

[1] Center for Human Movement Sciences, University Medical Center Groningen, University of Groningen, Groningen, Netherlands, [2] Mechanical, Maritime and Materials Engineering, Delft University of Technology, Delft, Netherlands, [3] Center for Rehabilitation, University Medical Center Groningen, Groningen, Netherlands, [4] Peter Harrison Centre for Disability Sport, School of Sport, Exercise and Health Sciences, Loughborough University, Loughborough, United Kingdom

The acquisition of daily handrim wheelchair propulsion skill as a multi-layered phenomenon has been studied in the past. Wheelchair racing, however, is considerably different from daily handrim wheelchair propulsion in terms of propulsion technique, as well as the underlying equipment and interface. Understanding wheelchair racing skill acquisition is important from a general motor learning and skill acquisition perspective, but also from a performance and injury prevention perspective. The aim of the current lab-based study was 2-fold: to investigate the evolution of racing wheelchair propulsion skill among a sample of novices and to compare them with an experienced wheelchair racer under similar conditions. A convenience sample of 15 able-bodied novices (8 male, 7 female) completed a standardized three-week submaximal uninstructed practice protocol (3 weeks, 3 sessions per week, 3x4 min per session) in a racing wheelchair on an ergometer. Required wheeling velocity was set at 2.78 m/s (10 km/h) and a rolling friction coefficient of 0.011 (resulting in a mean target load of 21W) was used. For comparison, an experienced T54 Paralympic athlete completed one block of the same protocol. Kinetics, kinematics, and physiological data were captured. A mixed effects regression analysis was used to examine the effect of practice for the novices, while controlling for speed. All participants finished the protocol successfully. However, not all participants were able to achieve the target speed during the first few sessions. Statistically significant improvements over time were found for all outcome measures (i.e., lower metabolic strain, longer push and cycle times) with the exception of mean power and torque per push. The athlete used a significantly greater contact angle and showed "better" outcomes on most metabolic and kinetic variables. While the athlete used a semi-circular propulsion technique, most participants used a double looping over



**Edited by:**
*Hua-Fang Liao,
National Taiwan University, Taiwan*

**Reviewed by:**
*Martin Ferguson-Pell,
University of Alberta, Canada
Zhuoying QIU,
China Rehabilitation Research Center/WHO Collaborating Center for Family International Classifications, China*

***Correspondence:**
*Lucas van der Woude
l.h.v.van.der.woude@umcg.nl*

*† These authors have contributed equally to this work and share first authorship*

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technique. Three weeks of uninstructed wheelchair racing practice significantly improved efficiency and skill among a group of novices, in line with previous studies on daily handrim wheelchair propulsion. The comparison with an experienced athlete expectedly showed that there is still a large performance (and knowledge) gap to be conquered.

**Keywords: wheelchair racing, wheelchair athletics, motor learning, propulsion technique, kinematics, physiology, kinetics**

## 1. INTRODUCTION

Wheelchair racing was part of the first international wheelchair sporting competition for people with disabilities in 1952 (1). Since then, wheelchair racing and racing wheelchairs have greatly evolved, the latter now consisting of a long-base three-wheel lightweight configuration with one large wheel in the front and two large rear wheels with relatively small handrims in order to reach and maintain high speeds (2). Races are organized for field and track events and include sprints, middle-long distances, and long distances, including the marathon. Athletes compete in their own class to ensure that athletes with similar impairment race against each other (3). Involvement in sports, such as wheelchair racing after rehabilitation has a positive influence on physical (4) and psychological health and well-being (1, 5). Therefore, it is important that patients with lower-limb impairments get involved in new (adapted) sports, such as wheelchair sports, during, and after rehabilitation. This requires them to learn new propulsion (and game) skills, which is especially thought to be required for wheelchair racing where the wheelchair design and interface require for different postures and propulsion technique. Although there is existing knowledge on skill acquisition during daily wheelchair propulsion (6–8), mechanisms of learning wheelchair racing are still unclear.

To become more proficient in wheelchair racing, an athlete either needs to increase the physical work capacity or become more efficient (1). Experienced wheelchair racing athletes have been studied to gain insight in their propulsion technique and corresponding mechanical efficiency. Compared with regular handrim wheelchair propulsion, athletes use a larger contact angle of approximately 180° and start at 20° past the top-dead center of the handrim (9, 10). Starting further on the handrim allows athletes to be in a more horizontal position in the racing wheelchair, reducing air resistance. Moreover, wheelchair racers use gloves as coupling is infeasible at racing speeds. During racing conditions, as segmental velocities increase, the push is performed as a stroke against the rims. Whereas, during regular handrim wheelchair propulsion one can grab the handrims, making push-pull action possible (11). To increase wheelchair racing performance, athletes need to learn this new movement, requiring different movement patterns and adaptations in both physiology and technique. Yet, little scientific research has focused on the acquisition of wheelchair racing skill thus far.

The acquisition of daily wheelchair propulsion skill has been extensively studied for regular handrim wheelchairs in wheelchair users (12) and (novice) able-bodied participants (6–8). These studies generally examined steady-state submaximal performance at low speeds, using gross mechanical efficiency as the primary outcome measure (13). Experienced participants are said to have a higher mechanical efficiency, meaning they are able to produce the same amount of external power output at a lower energetic cost. This is in line with the framework of Sparrow and Newel (14, 15) and Almåsbakk et al. (16), where cyclic movement patters are thought to emerge through the interaction of different constraints, with metabolic energy as an optimization parameter. The increase in mechanical efficiency can be due to improvements in propulsion skill and/or physiological adaptation (12). A high mechanical efficiency in wheelchair propulsion was linked to increased wheeling proficiency, expressed as greater contact angles and a decreased push frequency, which is especially beneficial as this is thought to improve mobility and reduce risk of injury (17, 18).

A better technique and higher efficiency are also beneficial to racing performance (9, 10) and could reduce injury sensitivity. However, racing and regular handrim wheelchair propulsion skill are distinct and there is no information available on the acquisition of wheelchair racing skill. One challenge specific to wheelchair racing is to maintain extreme high velocities. Smaller sized handrims help to meet the required speeds to some extent, since linear hand speed can be kept lower with smaller rim diameters which was shown to be more efficient and less straining in experienced wheelers (19). Yet, the majority of wheelchair racing performance probably still comes down to underlying coordination and skill. Like regular handrim wheelchair propulsion, wheelchair racing can be approached as a cyclical skill where motor learning can be quantified as a decrease in energy expenditure (8, 14). As such, mechanical efficiency is expected to increase, as mastering this task should result in more optimal kinetic and kinematic solutions (7, 16).

The current study focused on the initial motor learning process of three weeks of lab-based uninstructed wheelchair racing propulsion practice in inexperienced able-bodied participants on a wheelchair ergometer. More specifically, it examines the gross mechanical efficiency as the primary outcome measure for motor learning and the concomitant kinetic and kinematic solutions of the participants. Able-bodied participants were chosen as they are full novice to the task and form a relatively homogeneous group (similar age, lack of wheelchair experience, and no comorbidities), minimizing the inter-individual variation which allows to better isolate the effect of uninstructed learning on the outcomes of the motor learning process (20). Additionally, an experienced athlete performed a similar protocol to provide a reference for skilled wheelchair racing propulsion.

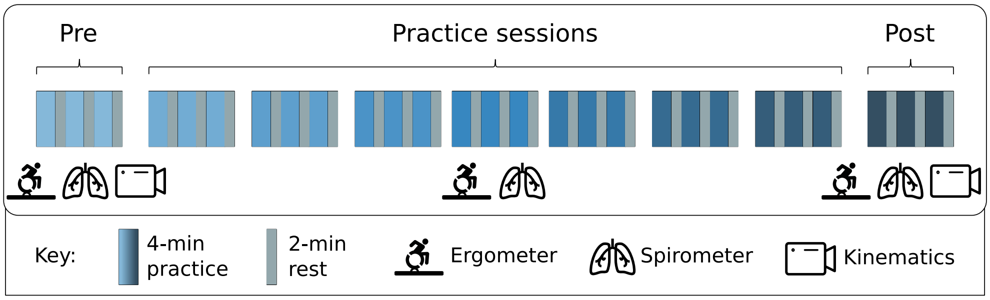


**FIGURE 1 |** Overview of the protocol: participants were tested on 9 occasions spread over three weeks with three blocks of practice each. Data were captured during all sessions, but kinematics were only recorded during the first- and last (pre and post) session.

# 2. MATERIALS AND METHODS

## 2.1. Participants

The current study used a convenience sample of 15 inexperienced able-bodied participants (7 female/8 male, 22.0 (±1.35) years old, 69.3 (±9.87) kg). The sample size was based on previous studies with a similar design in regular handrim wheelchairs (8, 21, 22). Participants were eligible for inclusion if they had no previous experience with wheelchair propulsion and no contraindications for exercise [PARQ, (23)]. Additionally, one high-level T54 middle-distance athlete, was included for comparison (male, 67 kg). All participants provided written informed consent after receiving detailed information about the study. The study was approved by the local ethical committee of the Center for Human Movement Sciences, University Medical Center Groningen, University of Groningen.

## 2.2. Study Design

Participants received a total practice load of 108-min consisting of nine sessions (three sessions per week for three weeks) of 3x4 min of submaximal manual racing wheelchair exercise (**Figure 1**) on an instrumented wheelchair roller ergometer [Lode, Groningen, The Netherlands, (24)]. This practice load was shown to be sufficient to achieve a learning effect in regular handrim wheelchair propulsion (6, 20, 25). They received no advice on propulsion technique prior to the experiment and no feedback during the sessions, resulting in a “natural” learning process (26). Before the start of an exercise block, the sole instruction was to propel at a required speed of 2.78 m/s (10 km/h) and to hit/push the handrim with the soft hand gloves. The required velocity was based on a pilot determining a feasible, yet fast enough, speed for untrained participants. A computer screen in front of the participants provided visual feedback on the actual and target speeds (21).

## 2.3. Equipment

### 2.3.1. Wheelchair

All tests were performed in the same experimental Amasis racing wheelchair (Wolturnes, Nibe, Denmark) with 0.71 m (28-inch) wheels and 0.38 m (15-inch) handrims on the roller ergometer. The wheelchair was not individually accommodated. Participants used soft hand gloves to push the handrim. The athlete performed in his personal racing wheelchair with 28-inch wheels and 37 cm diameter handrim. Tire pressure of the rear wheels was set at 800 kPa (8 bar) before every session.

### 2.3.2. Physiology

Metabolic data were collected using a K5 Cardio-Pulmonary Exercise Testing (CPET) spirometer (COSMED, Rome, Italy) in breath-by-breath mode. Turbine, room air, reference gas, and delay calibrations for the spirometer were performed before each session. Heart rate was measured with a Garmin HRM Dual (Garmin International Inc, Kansas, USA) connected with the CPET. Participants were asked to rate their perceived exertion on a 6-20 Borg scale (27).

### 2.3.3. Kinetics

Force and velocity data were collected with an Esseda (Lode BV, The Netherlands) wheelchair ergometer at 100 Hz [for a technical description see (24)]. The ergometer was calibrated to account for static and dynamic friction before each session. For a demonstration of this process see (28). A rolling friction coefficient of 0.011 was set, resulting in a theoretical power output of 21 W at the mean body weight of the novice participants in this study. The coefficient was based on eight coast-down tests (29) with two athletes at the outdoor athletics track at the Olympic Training Center Papendal. The athlete, originally part of another study, performed at a power output of 28 W.

### 2.3.4. Kinematics

Finally, an active cluster marker was placed on the right-hand glove and tracked by an optoelectronic camera system (Optotrack, Northern Digital, Waterloo, Canada) at 100 Hz. The cluster was used to determine the location of second and fifth metacarpal (M2 and M5) during propulsion.

## 2.4. Analyses

All analyses were performed in Python [The Python Foundation, (30)] using a custom package available on the Python Package Index (31). To examine the motor learning process over time, all blocks were included and the last minute of each block was used, assuming steady-state propulsion. Finally, the mean of the three blocks per session were used for statistical analysis. Pre-processed

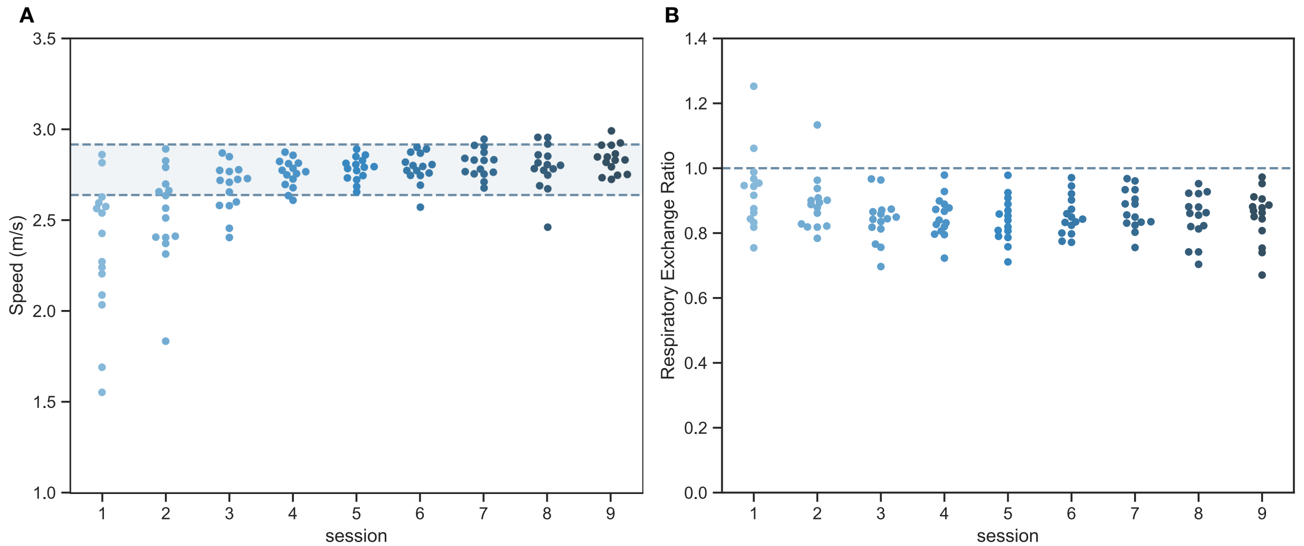


**FIGURE 2 |** Swarmplot of the ability of individual participants to match the target speed ±5% **(A)** and the respiratory exchange ratio (RER) <1.0 **(B)** during each session (n=15).

data are available as a supplementary material file in a comma separated values (.csv) format (32).

### 2.4.1. Physiology

Heart rate, respiratory exchange ratio (RER), and energy expenditure (EE) were obtained from the CPET system. Gross mechanical efficiency (GME) was calculated from the EE and the external power output (PO) obtained from the ergometer:

$$GME(\%) = EE * PO^{-1} * 100 \quad (1)$$

GME for sessions where the mean RER was higher than 1.0 were discarded, which was the case for three samples (**Figure 2**).

### 2.4.2. Kinetics

Kinetic data (force on the roller) from the ergometer were first filtered using a 15 Hz 4th-order zero-phase Butterworth filter. Propulsion technique variables (contact angle, push & cycle time, mean & peak torque and power per push, and work per push) were then determined based on the speed and force data from the ergometer for the left and the right side. Afterwards, the mean of the left and right side was used for further (statistical) analysis.

### 2.4.3. Kinematics

The last fifteen s of the M2 virtual marker location were plotted for each block of the first- and last session. Three raters (GJ, RK, and PW) qualitatively rated the propulsion technique using the definitions of Boninger et al. (33): Arcing (ARC), double looping over propulsion (DLOP), semicircular (SC), and single looping over propulsion (SLOP). Participants using the ARC pattern follow the pushrim closely for a small arc during the push and recovery phase. The DLOP pattern is characterized by the hands starting above the pushrim, then following the handrim, and then going over, crossing, and going under the pushrim during the recovery phase. In the SC pattern the hand dips under the handrim in a circular or elliptic motion and in the SLOP pattern the hand passes over the handrim during the recovery phase (33). The most frequent technique among blocks was identified as the session technique. In the case of a tie, the observed technique of the last block was used, this was done for each rater individually. Finally, the most frequent technique among raters was determined and reported.

## 2.5. Statistics

### 2.5.1. Physiology and Kinetics

A linear mixed effects analysis of the effect of time (session 1–9) was performed using R [R Core Team, (34)] and the lme4 package (35). Time and speed (without interaction term) were included in the model as fixed effects. Speed was added as not all participants were able to achieve the target velocity during the first sessions (**Figure 2**). Separate random intercepts and slopes were added for participants for the effect of time. The final model was defined as:

$$outcome \sim session + speed + (1 + session|subject) \quad (2)$$

There were no obvious deviations in the residual plots with regards to homoscedasticity or normality. P-values were obtained with a likelihood ratio test of the full model vs. a model without the effect of time. Data from the last session were compared with the athlete using a one-sample $t$-test. An alpha of 0.05 was used for all statistical tests.

### 2.5.2. Kinematics

Fleiss' Kappa was calculated to determine the agreement between raters with regards to the propulsion technique and were interpreted based on the suggestions of Landis and Koch (36). A contingency table was produced to describe the development of propulsion technique, but no further

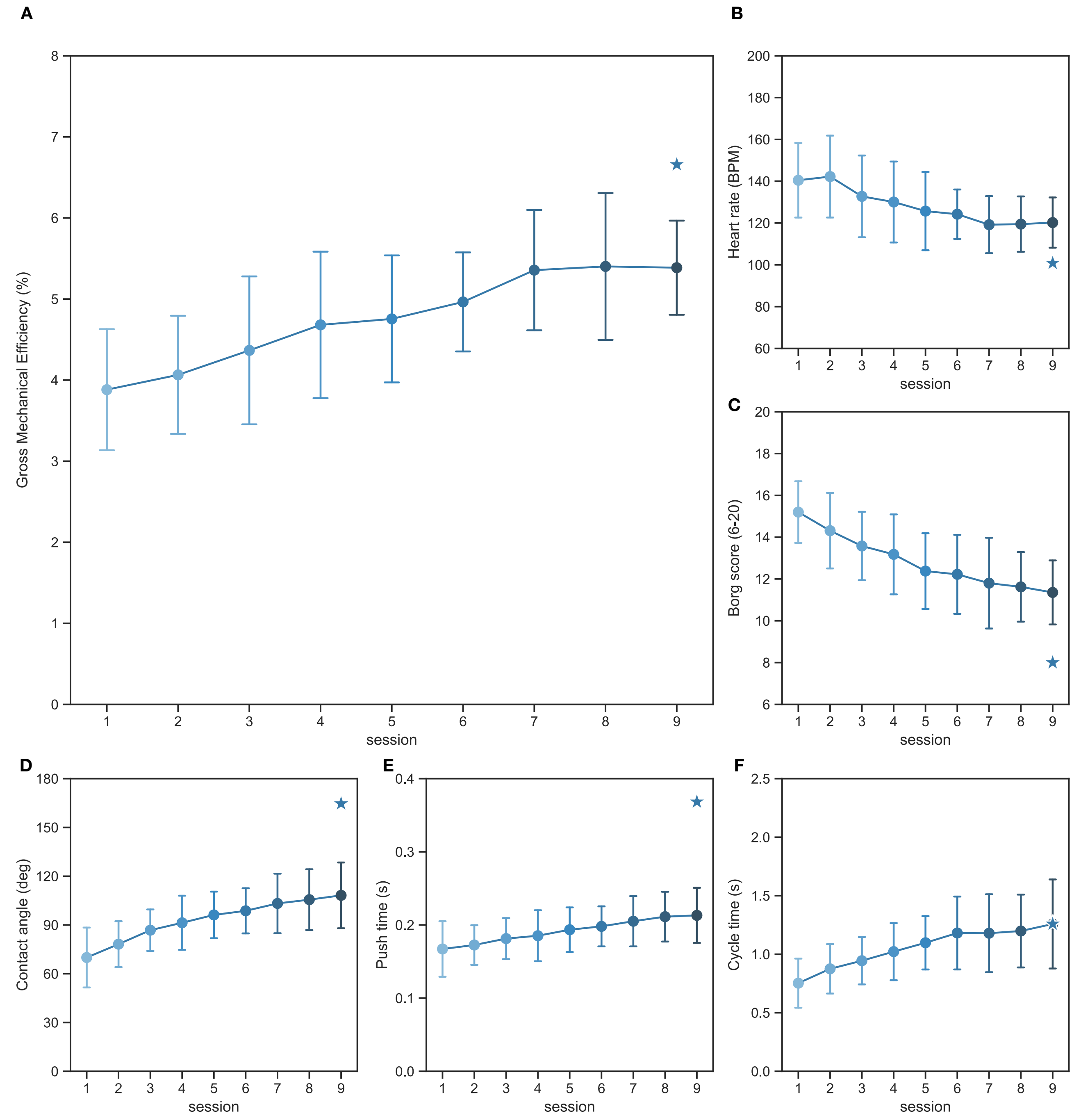


**FIGURE 3 | (A–F)** Outcomes for six metabolic and kinetic parameters aggregated (mean and standard deviations, n=15) by session (★=athlete). All variables shown significantly changed over time. The athlete scored significantly "better" on all parameters shown, except for cycle time **(D)**.

statistical analysis was performed due to the sparsity of the data.

## 3. RESULTS

All participants completed the experiment successfully. Yet, not all novices were able to achieve the desired velocity (±5%) during the first three sessions (**Figure 2**). Resultingly, speed and power output significantly increased between subsequent sessions as participants were increasingly able to achieve the target speed (**Figure 2**). Concomitantly, the average respiratory exchange ratio was higher than 1.0 during the first two sessions for some of the participants (2/15 in session 1 and 1/15 in session 2).

**TABLE 1 |** Outcomes: last minute of each block aggregated by session and reference data of a single wheelchair athlete with mixed effects regression and one-sample *t*-test results (n=15).

| Variable | Session | | | | | | | | | | Statistics | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | 1 | 2 | 3 | 4 | 5 | 6 | 7 | 8 | 9 | A[a] | Speed[c] | Time[c] | $\chi^2$ | p[d] | t[e] | p |
| **Protocol** | | | | | | | | | | | | | | | | |
| Speed | 2.34 (0.39) | 2.53 (0.27) | 2.68 (0.15) | 2.76 (0.08) | 2.78 (0.08) | 2.79 (0.09) | 2.81 (0.08) | 2.79 (0.14) | 2.84 (0.08) | 2.78 | n.a. | 2.45 (0.07) | 13.987 | <0.001 | 2.60 | 0.02[f] |
| Power | 19.7 (4.49) | 21.6 (3.55) | 22.7 (3.15) | 23.6 (3.2) | 23.6 (3.04) | 23.7 (3.03) | 23.8 (2.97) | 23.7 (3.20) | 24.1 (2.97) | 28.4 | n.a. | 0.43 (0.09) | 13.195 | <0.001 | -5.50 | <0.001[f] |
| **Physiological** | | | | | | | | | | | | | | | | |
| RPE (6-20) | 15.2 (1.8) | 14.3 (2.14) | 13.6 (1.85) | 13.2 (2.09) | 12.4 (2.03) | 12.2 (1.98) | 11.8 (2.23) | 11.6 (1.77) | 11.4 (1.60) | 8.0 | -2.04 (0.47) | -0.36 (0.05) | 22.937 | <0.001 | 8.20 | <0.001 |
| HR (BPM) | 141 (19.5) | 142 (20.7) | 132 (20.2) | 130 (20.5) | 126 (19.8) | 124 (12.9) | 119 (14.3) | 120 (14.1) | 120 (12.8) | 101 | 2.36 (4.95) | -3.21 (0.56) | 19.128 | <0.001 | 6.04 | <0.001 |
| EE (W) | 552 (152) | 543 (115) | 544 (134) | 522 (123) | 510 (105) | 482 (72.7) | 451 (71.1) | 445 (61.4) | 450 (58.2) | 426 | 112 (31.4) | -21.7 (3.71) | 19.105 | <0.001 | 1.52 | 0.06 |
| GME (%)[b] | 3.88 (0.78) | 4.06 (0.76) | 4.37 (0.94) | 4.68 (0.93) | 4.75 (0.81) | 4.96 (0.63) | 5.36 (0.77) | 5.4 (0.94) | 5.39 (0.60) | 6.66 | 0.74 (0.32) | 0.18 (0.83) | 18.276 | <0.001 | -8.20 | <0.001 |
| **Kinetics (per push)** | | | | | | | | | | | | | | | | |
| Contact angle (deg) | 70.0 (20.4) | 78.2 (15.0) | 86.8 (13.5) | 91.3 (17.7) | 96.2 (15.7) | 98.7 (14.6) | 103 (18.8) | 106 (19.3) | 108 (20.7) | 165 | 28.7 (4.80) | 3.12 (0.84) | 10.216 | <0.01 | -10.4 | <0.001 |
| Push time (s) | 0.17 (0.04) | 0.17 (0.03) | 0.18 (0.03) | 0.19 (0.04) | 0.19 (0.03) | 0.20 (0.03) | 0.21 (0.04) | 0.21 (0.04) | 0.21 (0.04) | 0.37 | 0.003 (0.001) | 0.006 (0.002) | 8.9818 | <0.01 | -15.4 | <0.001 |
| Cycle time (s) | 0.75 (0.23) | 0.88 (0.22) | 0.94 (0.21) | 1.02 (0.26) | 1.10 (0.25) | 1.18 (0.32) | 1.18 (0.34) | 1.2 (0.32) | 1.26 (0.39) | 1.26 | 0.32 (0.07) | 0.04 (0.01) | 8.8434 | <0.01 | 0.00 | 0.50 |
| Mean torque (Nm) | 6.76 (2.31) | 7.1 (2.39) | 7.15 (2.08) | 7.56 (2.44) | 7.79 (2.48) | 8.07 (2.92) | 7.55 (1.75) | 7.69 (1.85) | 7.73 (1.87) | 6.17 | | | 0.0242 | 0.87 | 3.22 | 0.03 |
| Peak torque (Nm) | 12.4 (3.85) | 12.9 (3.97) | 13.2 (3.92) | 13.9 (4.24) | 14.4 (4.58) | 15.1 (4.93) | 14.6 (3.76) | 15.0 (4.44) | 15.0 (4.10) | 13.0 | 2.35 (0.81) | 0.24 (0.10) | 4.8888 | 0.03 | 1.89 | 0.04 |
| Work (J) | 8.59 (3.73) | 10.2 (3.85) | 11.3 (3.68) | 12.7 (5.21) | 13.7 (5.00) | 14.5 (5.8) | 14.2 (4.71) | 14.9 (4.87) | 15.3 (5.41) | 18.3 | 6.25 (1.01) | 0.49 (0.16) | 7.5477 | <0.01 | -2.07 | 0.03 |
| Mean power (W) | 51.0 (23.7) | 57.2 (22.8) | 60.3 (19.1) | 65.5 (22.1) | 68.0 (22.8) | 70.5 (26.0) | 66.5 (16.0) | 67.2 (17.2) | 68.6 (17.2) | 48.1 | | | 0.0768 | 0.78 | 4.61 | <0.001 |
| Peak power (W) | 93.0 (40.0) | 104 (38.4) | 111 (36.3) | 120 (38.5) | 126 (42.1) | 132 (44.3) | 128 (34.0) | 131 (40.7) | 133 (37.7) | 102 | 54.8 (7.07) | 2.06 (0.85) | 5.2618 | 0.02 | 3.26 | <0.01 |

*a, athlete, single block at higher resistance; b, cases with respiratory exchange ratio (RER) < 1.0; c, unscaled estimates ± standard errors; d, p-value from likelihood ratio test; e, one-sample t-test (df=14); f, two-sided p-value; RPE, respiratory exchange ratio; HR, heart rate; BPM, beats per minute; EE, energy expenditure; GME, gross mechanical efficiency.*

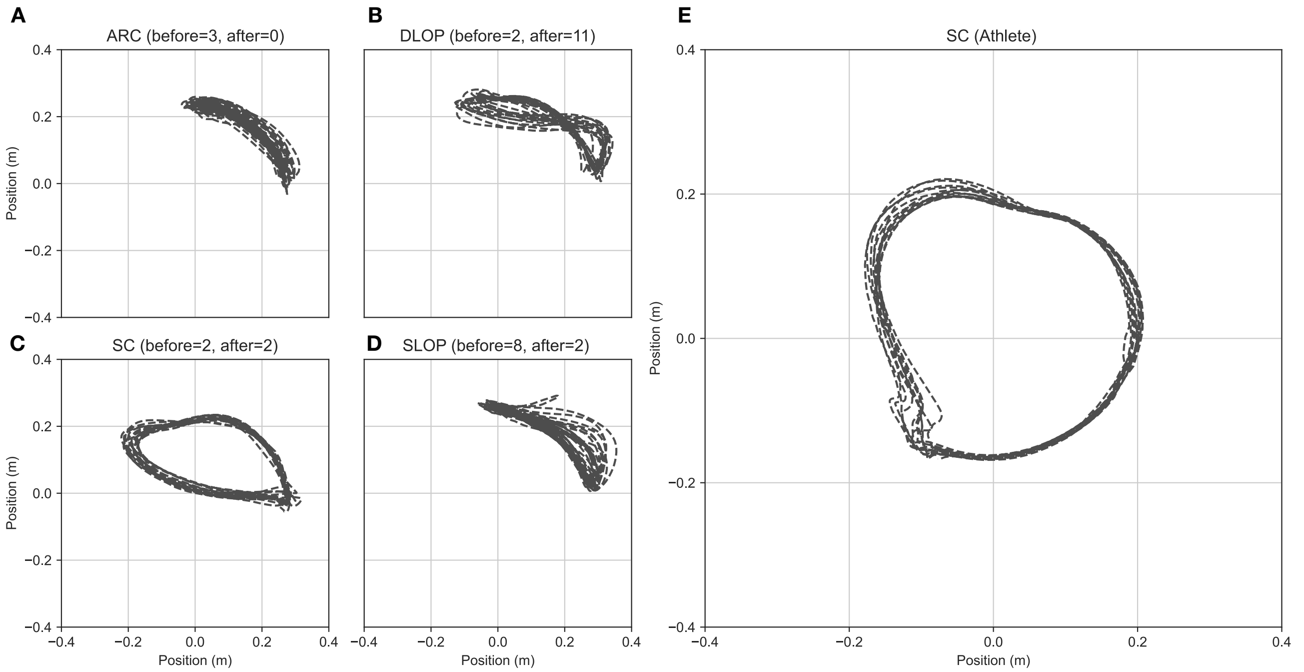


**FIGURE 4 |** Typical kinematics examples of the last fifteen seconds of the M2 virtual marker position displayed by different participants (each subplot shows one): **(A)** arcing, **(B)** double looping over, **(C)** semicircular, **(D)** single looping over propulsion, **(E)** semicircular, athlete. All data are from the last block of the last session with the exception of panel **(A)**.

## 3.1. Physiology and Kinetics

Physiological and kinetic aggregates and statistical outcomes are displayed in **Figure 3** and **Table 1**. A statistically significant improvement (i.e., higher GME, lower metabolic strain, higher push and cycle times) over time was found for all outcome measures with the exception of mean power and mean torque per push. Moreover, the perceived exertion also significantly lowered over time from "hard" to "fairly light." The athlete showed significantly better outcomes (i.e., less straining) on most metabolic and kinetic variables.

## 3.2. Kinematics

Results of the qualitative assessment of propulsion technique during the pre- and post-test are displayed in **Figure 4** and **Table 2**. Agreement among the three raters was "substantial" during the pre-test $\kappa$=.790, $p < 0.001$ and "almost perfect" during the post-test $\kappa$=.813, $p < 0.001$. Most participants started with a SLOP (53%) technique, but the majority gravitated toward a DLOP technique in the post-test (73%). The athlete used an SC propulsion technique.

# 4. DISCUSSION

This is the first study to examine the acquisition of wheelchair racing propulsion skill within the first three weeks of practice of inexperienced able-bodied participants. In general, participants became more proficient in wheelchair propulsion in a racing wheelchair on a wheelchair ergometer, which was reflected in the successful completion of the practice bouts in terms of speed and power output, and the significant improvements in propulsion skill and corresponding reductions in metabolic cost and perceived exertion. However, the novice participants still had a significantly different propulsion technique compared to the professional athlete.

Lower heart rates and energetic cost suggest that the propulsion technique became less strenuous for the inexperienced participants over time, which is corroborated by the decrease in perceived exertion (RPE). While these lower heart rates may have been the result of improved cardiorespiratory fitness, the American College of Sports Medicine (ACSM) states that 150 min of moderate exercise, or 75 min of vigorous exercise per week are required (37). Since these requirements are not met with 108 min of exercise and as energy expenditure also decreased, the lower heart rates were more likely caused by improvement in neuromuscular coordination and thus a reduction in cardiometabolic requirements with improved coordination and skill level (20). Accordingly, gross mechanical efficiency follows an inverse pattern, increasing from 3.9 to 4.5% (+39%). However, this is relatively low compared to other studies in regular handrim wheelchair propulsion (6–8, 12, 20, 21), which is unexpected considering the relatively high power output requirements of wheelchair racing propulsion (22, 38). Experienced wheelchair racing athletes generally have a more efficient propulsion technique (10), as was the case in the current study. Yet, the results of the experienced athlete were not similar to those of experienced wheelchair racing athletes in previous studies (9, 10, 39). However, the speed and power output of the current study (2.78 m/s) were also much lower than those

**TABLE 2 |** Contingency table of propulsion technique during the first and last session n(%).

| | | Before | | | | |
|---|---|---|---|---|---|---|
| | | ARC | DLOP | SC | SLOP | Total |
| After | ARC | 0 (0%) | 0 (0%) | 0 (0%) | 0 (0%) | 0 (0%) |
| | DLOP | 3 (20%) | 2 (13%) | 0 (0%) | 6 (40%) | 11 (73%) |
| | SC | 0 (0%) | 0 (0%) | 2 (13%) | 0 (0%) | 2 (13%) |
| | SLOP | 0 (0%) | 0 (0%) | 0 (0%) | 2 (13%) | 2 (13%) |
| | Total | 3 (20%) | 2 (13%) | 2 (13%) | 8 (53%) | 15 (100%) |

*Arcing (ARC), double looping over propulsion (DLOP), semicircular (SC), and single looping over propulsion (SLOP).*

of previous studies (3.60–7.20 m/s), which could explain the difference in mechanical efficiency (9, 39).

Coordination of wheelchair racing propulsion is complicated due to the use of gloves, a small hand rim and a fast spinning wheel (40). Coupling happens outside of the visual field which makes it harder to start the push with the same hand velocity compared to the wheel velocity (21). As a result of practice, participants were able to increase their contact angle and decrease their push frequency, which is in line with previous studies in regular handrim wheelchairs (6–8, 12, 20, 21) and the longer-slower hypothesis as proposed by Sparrow and Newell (15). The latter states that changes in the timing of movement might be linked to reduced metabolic loads, in line with the increased muscle contraction efficiency at optimum speeds in Hill's muscle model (41). The current study adds to the body of evidence relating control parameters and metabolic expenditure. In contrast to the other parameters, mean power per push did not significantly change. However, using the same mean power on a longer push means that the participants were able to increase the amount of work delivered per push. The wheelchair athlete used an even larger contact angle, resulting in an even longer push time. Even though the athlete performed at a higher external power output, this still allowed for a lower mean and peak power per push.

Only two (13%) participants adopted a semi-circular propulsion pattern which is ubiquitous in competitive wheelchair racing. All other participants used different techniques with the majority (73%) gravitating toward a double looping over propulsion. This propulsion technique is often associated with regular handrim wheelchair propulsion (33, 42). On the other hand, athletes use a near horizontal trunk position during wheelchair racing which limits the available range of motion for the recovery pattern and forces a starting position on the handrim that is beyond top-dead center. As the current study was performed in a lab setting, where no wind or air resistance was present, there is no need to employ a more horizontal position and reduce the exposed surface area. This might have encouraged a different propulsion pattern as the task/environment constraints are different than those of actual wheelchair racing, leading to a different movement solution. However, it is still unclear whether a longer attenuation period may lead to the same kinematic solutions or that the participants are stuck in a local minimum. Finally, while pattern classification is subjective, the inter-rater agreement in this study was high. Nevertheless, some quantitative measures are available and should be further developed to provide a more robust objective method of describing propulsion patterns (42, 43).

Despite piloting beforehand, not all participants seemed able to achieve the desired velocity during the first three sessions. The able-bodied participants were complete novices, whereas regular handrim wheelchair users already have some propulsion skill that could transfer. Wheelchair racing propulsion is a relatively hard task which takes a certain amount of skill to even begin the process of mastery. To borrow terminology from the electronic-sports domain: it has a high skill floor. However, as speed was included in the mixed effects regression model, the statistical outcomes "account" for the effect of speed. The inclusion of one experienced athlete provided information about the reference technique of racing propulsion. Yet, one athlete is not representative for all wheelchair racing athletes across all disciplines. The athlete also performed at a higher external power output than the novice participants, which is known to influence propulsion technique parameters and mechanical efficiency (22, 38, 44). Finally, any potential sex-dependent differences between the athlete and 7/15 novice participants are not accounted for. These specific results should therefore be treated with care. However, for other parameters such as RPE and heart rate the differences found are even more pronounced when considering the higher power output. Finally, it must be noted that all results were obtained on an ergometer and in a small sample of able-bodied participants. The ergometer provides a more constrained, yet standardized, environment than a racing track or other training environments. Moreover, the current ergometer setup only allowed for the examination of straight-line wheelchair propulsion. Previous studies in regular handrim wheelchair propulsion, however, have not found any differences between treadmill/ergometer and overground propulsion practice (20, 45). Whether this is also the case for the more complicated wheelchair racing task is an avenue for future research.

The current study examined the effects of a uninstructed learning setup, to improve our understanding of the learning process of wheelchair racing propulsion. Yet, previous studies in daily handrim wheelchair propulsion have also examined the effects of variable practice (20) and feedback (8). Exploring those setups would be especially interesting since learning processes in sports are generally guided or supervised by trainers or coaches. The effect of feedback or variable practice could therefore provide them with valuable input. Perhaps one of the most essential parts of wheelchair racing is the coupling of the gloved hand with the handrim (40). To provide enough friction between the glove and the handrim, a medio-lateral force is required which reduces the fraction of effective force (46). Therefore, studies that specifically examine this coupling and the influence of sports equipment (i.e., rim and glove type) using 3D kinematics and kinetics are needed. Finally, since the sport is only eligible for athletes with an impairment, this seems crucial for understanding wheelchair racing. As these athletes usually already have some wheelchair experience, but might have a reduced physiological capacity or other impairments that influence the learning process. Therefore, future research should also include experienced wheelchair users that are new to wheelchair racing propulsion.

In short, the current study on motor learning processes found similar results for wheelchair racing and previous research in daily wheelchair propulsion. Similar to previous studies, participants show larger contact angles and a decreased push frequency. Using only uninstructed practice, participants increased their mechanical efficiency by 39% (1.5%-point). A comparison with an experienced athlete showed that both the propulsion pattern, and physiological and kinetic outcomes are still different. The performance gap between the participants and the experienced athletes shows that much can still be learned about the difficult task that is wheelchair racing.

## DATA AVAILABILITY STATEMENT

The datasets presented in this study can be found in online repositories. The names of the repository/repositories and accession number(s) can be found at: https://dataverse.nl/dataset.xhtml?persistentId=doi:10.34894/EBJBMF.

## ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Ethische Commissie Bewegingswetenschappen, University Medical Center Groningen, Groningen, The Netherlands. The patients/participants provided their written informed consent to participate in this study.

## AUTHOR CONTRIBUTIONS

GJ, DV, LW, and RV were involved in the conceptualization of the study, obtaining ethical approval, and writing the data management plan. GJ was responsible for the collection of the data under supervision of RK and RV. Data were pre-processed and analyzed by RK and GJ. RK drafted the final manuscript and all other authors were involved in refining the manuscript. The entire process was supervised by LW and RV.

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# Effect of Haptic Training During Manual Wheelchair Propulsion on Shoulder Joint Reaction Moments

*Rachid Aissaoui[1,2]* and Dany Gagnon[3,4]*

*[1] Laboratoire de Recherche en Imagerie et Orthopédie (LIO), Centre de Recherche du Centre Hospitalier Universitaire de Montréal (CRCHUM), Montreal, QC, Canada, [2] Département de Génie des systèmes, École de technologie supérieure (ETS), Montreal, QC, Canada, [3] School of Rehabilitation, Université de Montréal, Montreal, QC, Canada, [4] Pathokinesiology Laboratory (www.pathokin.ca), Institut universitaire sur la réadaptation en déficience physique de Montréal (IURDPM), Montreal, QC, Canada*

**Background:** Manual wheelchair propulsion remains a very ineffective means of locomotion in terms of energy cost and mechanical efficiency, as more than half of the forces applied to the pushrim do not contribute to move the wheelchair forward. Manual wheelchair propulsion training using the haptic biofeedback has shown an increase in mechanical efficiency at the handrim level. However, no information is available about the impact of this training on the load at the shoulders. We hypothesized that increasing propulsion mechanical efficiency by 10% during propulsion would not yield clinically significant augmentation of the load sustained at the shoulders.

**Methods:** Eighteen long-term manual wheelchair users with a spinal cord injury propelled a manual wheelchair over a wheelchair simulator offering the haptic biofeedback. Participants were asked to propel without the Haptic Biofeedback (HB) and, thereafter, they were subjected to five training blocks BL1–BL5 of 3 min in a random order with the haptic biofeedback targeting a 10% increase in force effectiveness. The training blocs such as BL1, BL2 BL3, BL4, and BL5 correspond, respectively, to a resistant moment of 5, 10, 15, 20, and 25%. Pushrim kinetics, shoulder joint moments, and forces during the propulsive cycle of wheelchair propulsion were assessed for each condition.

**Results:** The tangential force component increases significantly by 74 and 87%, whereas value for the mechanical effective force increases by 9% between the pretraining and training blocks BL3. The haptic biofeedback resulted in a significant increase of the shoulder moments with 1–7 Nm.

**Conclusion:** Increases in shoulder loads were found for the corresponding training blocks but even though the percentage of the increase seems high, the amplitude of the joint moment remains under the values of wheelchair propulsion found in the literature. The use of the HB simulator is considered here as a safe approach to increase mechanical effectiveness. However, the longitudinal impact of this enhancement remains unknown for the impact on the shoulder joint. Future studies will be focused on this impact in terms of shoulder risk injury during manual wheelchair propulsion.

**Keywords: biofeedback, biomechanics, haptic, manual wheelchair, shoulder joint moment, mechanical efficiency**

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****Correspondence:***
*Rachid Aissaoui*
*rachid.aissaoui@etsmtl.ca*

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## INTRODUCTION

Manual wheelchair (MW) propulsion is the primary mean of mobility for individuals that sustained a spinal cord injury (SCI). Although MW propulsion helps those individuals to regain certain independence and maintain or increase societal participation, it remains a very ineffective means of locomotion in terms of energy cost and mechanical efficiency (1, 2). More precisely, it has been shown that almost half of the forces applied to the pushrim do not contribute at moving the MW forward in individuals with a SCI (3–5). Earlier studies looked at the possibility of increasing the force effectiveness (i.e., the tangential component) using training methods such as visual feedback (6, 7). While de Groot et al. (6) found a significant increase in force effectiveness [mechanical effective force (MEF)] between pre- and posttraining in 10 healthy individuals, no significant augmentation was found in the study of Kotajarvi et al. (7) for 18 experienced MW users. The authors suggested that visual feedback of the average force effectiveness value might not be the optimal training strategy to improve force effectiveness during propulsion (7).

Recently, Blouin et al. (8) used an Haptic Biofeedback (HB) simulator developed by Chenier et al. (9) to increase force effectiveness in 18 experienced MW users who sustained a SCI. The authors (8) have shown a significant increase in force effectiveness using the HB. More precisely, the participants were able, on average, to increase force effectiveness by 12–15% bilaterally suggesting an interesting potential as a training tool for MW users. The HB has been previously shown to be an efficient sensory feedback tool to teach movement and force patterns in the rehabilitation of the upper limb in hemiparetic patients (10, 11). In this study, the HB is defined as the ability to our wheelchair simulator to continuously modify the direction of the force applied to the handrim during the propulsion phase in real time. In general, this operation looks as an adaptive process control. Without HB, the user propelled the wheelchair with a specific personalized pattern of propulsion as represented by his initial MEF. From that pattern, a new targeted pattern is artificially created by modifying a portion of the pattern. This forms a closed-loop control in which a resistive moment to the wheel is added or subtracted proportionally to the error signal between the targeted and the initial MEF. Since there is no visual information fed to the subject, but only proprioceptive information, i.e., a gradually resistive moment at 2 kHz, we call this control as the HB (9). In this study, the HB was modulated continuously during the propulsion phase and it takes <10 cycles when the subject senses the difference between his/her own pattern and the one imposed.

Although it seems possible to increase the MEF using an HB simulator, no information regarding the load imposed at the upper limb joints by this increase force effectiveness is yet available. Several authors suggested that propelling a MW with greater force effectiveness would yield to a greater risk exposure for the musculoskeletal structures (6, 12, 13). In the past, an effort was made to gain a better understanding of the relationship between the propulsive force effectiveness and the shoulder joint reaction moment. In fact, two simulation studies have shown that, indeed, a force close to tangential could highly increase the joint moment at the shoulder level during manual wheelchair propulsion (14, 15). More specifically, Bregman et al. (14) have shown almost a 2-fold increase in shoulder joint moments when only using the tangent force component as an input for an inverse dynamic model. However, one study (15) suggested that an improvement in the force effectiveness of around 10% would be possible without yielding significantly higher mechanical demand at the shoulder. Giving the high prevalence of secondary impairments at the shoulder among MW users, it would be interesting to determine the *in-vivo* impact of increasing force effectiveness by 10%, as suggested by Desroches et al. (15) using the HB simulator on the shoulder joint moments (16–19). We hypothesized that increasing force effectiveness by 10% during propulsion would not yield clinically significant augmentation of the load sustained at the shoulders.

## METHODS

### Participants

Eighteen long-term MW users (MWUs) (16 men and 2 women) with a SCI were recruited to participate in this study (**Table 1**). To be included, participants had to have a complete or incomplete SCI [American Spinal Injury Association (ASIA) established a grading system called the ASIA A, B, or C] between C7 and L1 vertebral levels for 3 months or longer, use a manual wheelchair as their primary means of mobility, and be able to perform wheelchair-to-wheelchair transfers independently with or without the use of a transfer board. Participants were excluded from this study, if they had any pressure sores on the buttocks or if they reported any pain that could have hindered their propulsion biomechanics. This study was approved by the research ethics committees of the École de technologie supérieure (ÉTS) and the Center for Interdisciplinary Research in Rehabilitation of Greater Montreal (CRIR).

### Haptic Simulator and Measurements

The experimental tasks were performed using a recently-developed haptic simulator (**Figure 1**) (9). Briefly, this simulator

**TABLE 1** | Participants' characteristics [mean (1 SD)].

| | ***n* = 18** |
|---|---|
| Age | 42.4 |
| (y) | (13.9) |
| Height | 1.73 |
| (m) | (0.20) |
| Weight | 77.4 |
| (kg) | (14.1) |
| Time since injury | 14.8 |
| (y) | (10.1) |
| AIS level | 1 C8, 1 T2, 2 T4, 1 T5, 2 T6, 1 T7, 1 T9, 3 T10, 1 T11, 5 T12 |
| ASIA | 13 A, 2 B, 2 C, 1 D |
| Gender (M/F) | 16/2 |

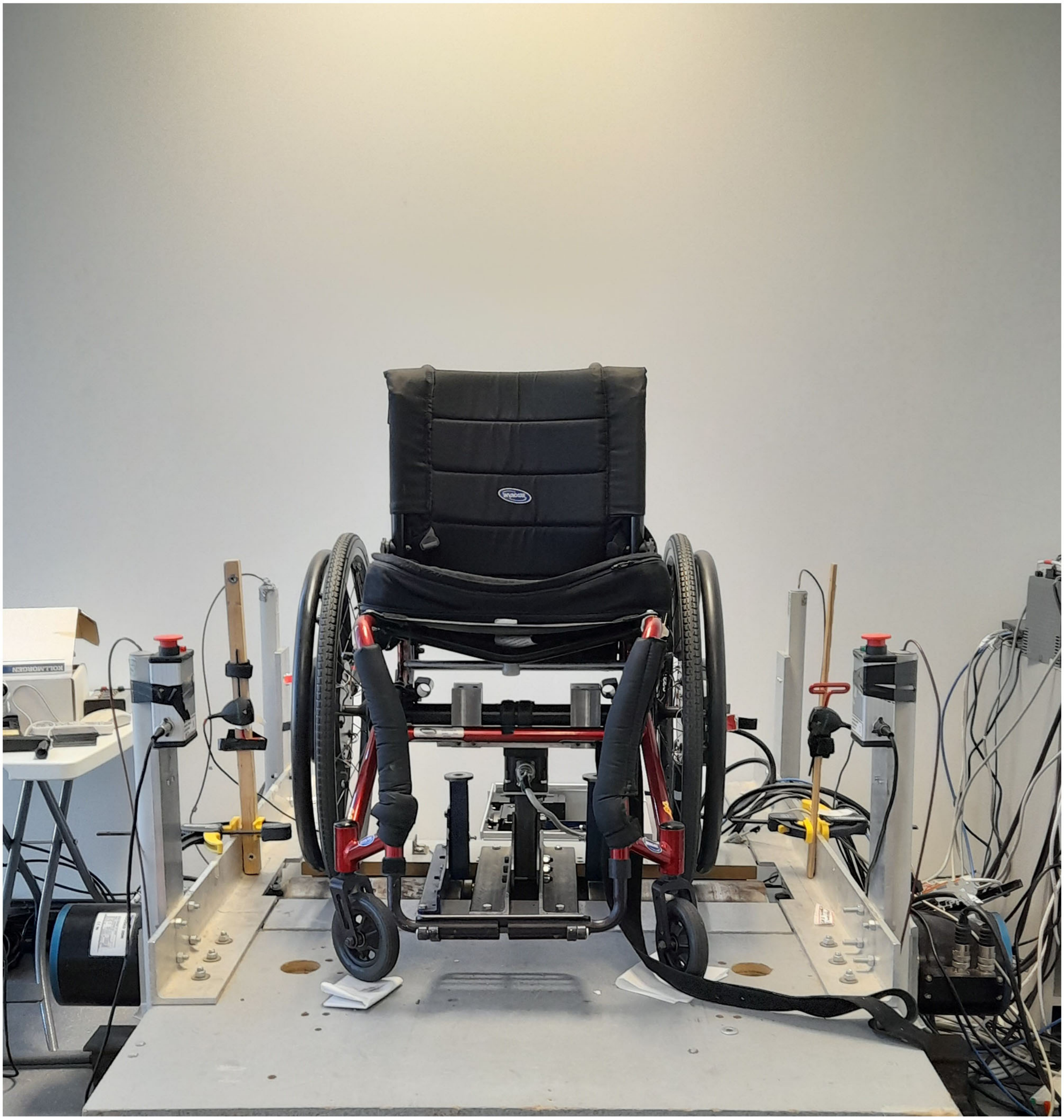

**FIGURE 1 |** Haptic wheelchair simulator.

acquires real-time bilateral three-dimensional forces and moments measured with instrumented wheels (SmartWheel, Three Rivers Holding, LLC) during propulsion. The propulsive moments generated by the user about the wheel hub are used as the input for the dynamic model of a virtual wheelchair. The dynamic model, presented in a study by Chenier et al. (9), estimates the angular velocity of each rear wheel of a virtual wheelchair, which represents propulsion on a ground-level surface. Velocity controllers ensure that the angular velocities of the real wheelchair match those of the virtual wheelchair rear wheels, so that a complete haptic loop is defined between the user and the simulator. Then, based on those forces and moment as well as the desired feedback that will be described later, the haptic feedback is given to the user by two motors under each rear wheel that will induce gradual resistance during the propulsion phase when the user is not following the desired force feedback pattern (8). All the experimental tasks were performed in the *Invacare A4 Ultralight* wheelchair mounted on the simulator. Participants were assisted to transfer from their personal wheelchair to the simulation and their own seat cushion was used. The backrest angle was adjusted as close as possible to that of the participant's personal wheelchair.

The bilateral upper extremity three-dimensional kinematics was recorded with 26 reflective markers captured by six cameras VICON M460 system (Vicon Motion Systems Limited, Oxford, UK) at a sampling frequency of 120 Hz. Markers placement were similar to the one described in a study by Desroches et al. (20). The reflective markers used in this study are the ulnar and radial as well as the second and fifth metatarsal for the hand segment. For the lower arm, a cluster of 3 markers at mid-distance from the wrist and elbow joint, plus the lateral and medial epicondyles. For the humerus segment, a cluster of 3 markers at mid-distance from the elbow to the acromion markers, plus the acromion marker. For the thorax, C7 and T8 markers, plus the jugular notch as well as the sternum end.

Additionally, two reflective markers were added on the surface of each wheel to define each wheel reference systems relative to the global coordinate system. Three-dimensional forces and moments applied at the pushrim by the participants were recorded by the two SmartWheels at a sampling frequency of 240 Hz. Kinematics and kinetic data were acquired and stored on an external computer for further postprocessing.

## Haptic Biofeedback

The determination and application of the HB has been described in a previous study (8). Briefly, the HB was determined in real time and based on the difference between the actual MEF and the targeted optimized MEF ($MEF_T$). MEF is the ratio between the squared tangent force component and the squared total force (2, 21). In the current experiment, the design of the curve pattern of the $MEF_T$ was personalized for each subject. It corresponds to the initial MEF prerecorded pattern of the subject, which was deformed by two linear Gaussian functions in order to increase the maximal value of the original MEF by an amount of 10% (8). During the push phase of the propulsion, the participants perceive an increase of resistance as long as their actual MEF pattern deviates from the desired one $MEF_T$. The intensity of the HB or the resistance felt by the participants was determined relative to their maximum voluntary propulsive moment (MVM) recorded prior to the experiment. Five relative intensities were used in the current experiment: 5, 10, 15, 20, and 25% of the participant's MVM (8).

## Pretraining

To become familiar with the haptic simulator, participants propelled freely for 1 min on the simulator. Then, two trials (named INI) without the HB were conducted at the participant's self-selected velocity for 1 min each. Kinematics and kinetic data were recorded for the last 30 s of each trial. The mean linear velocity reached during each of the two 30-s acquisition periods was calculated. If the mean linear velocity varied by more than 10% between the two trials, a third trial was recorded and trial 1 was discarded.

## Training

Training was divided into five 3-min blocks. 2-minute rest periods were included after each block. Each training blocks were corresponded to an intensity level (BL1–BL5 corresponding to 5–25% with a step of 5%, respectively) and were presented in a random order. The HB was activated 3 s after the beginning of each training block and kept active until the end of the block. Participants were told that they had to push more tangentially on the handrims to increase their MEF. Participants were also instructed to always strive for the lowest resistance possible, which indicated that their actual MEF pattern was coming closer to the target pattern $MEF_T$. Kinematics and kinetics data were acquired during the last 30 s of each training trial. In addition to the HB, the average speed of each block was shown to help participants to match their velocity to the velocity achieved during pretraining.

## Posttraining

After a 2-min rest period, two posttraining trials (POST) without the HB lasting 1 min each were conducted with the same methodology used during pretraining. The only visual feedback provided was the average speed during propulsion.

## Postprocessing

For each of the experimental tasks, three-dimensional trajectories of each kinematic marker were filtered using a 4th order low-pass Butterworth filter with a cutoff frequency at 6 Hz, while pushrim forces and moments were filtered using an 8th order low-pass Butterworth filter with a cutoff frequency of 30 Hz.

## Pushrim Force Measurements

For each of the experimental tasks, the resultant force at the pushrim (Fres) and its tangential component (Ftan) were continuously calculated as well as the MEF bilaterally. The Fres was defined as the vector sum of the three force components measured by the SmartWheel. The Ftan was obtained using the point of force application method (22). The magnitude of the tangential force was estimated by dividing the moment around the medial-lateral axis of the SmartWheel by the handrim radius. The MEF was then obtained as the ratio between the tangent force component squared and the resultant force squared.

## Inverse Dynamics Software

Upper limb net joint moments and forces were estimated using an inverse dynamic method (23). The forces and moments measured by each SmartWheel, upper limb kinematics, and the mass and height of each subject are used as input for the calculation of the shoulder joint reaction forces. The segment coordinate system of the forearm and arm was defined according to the International Society of Biomechanics (ISB) recommendations (24). The segment mass, the position of center of mass, and the inertia tensor of each body segment were estimated by scaling equations based on participants' anthropometry (25). The segment length of the hand and lower and upper arms were measured based on markers. Hand segment length was defined as from the mid-distance between the ulnar and radial markers and the 2nd and 5th metatarsals. The lower arm segment was defined from the mid-distance of the elbow lateral and medial epicondyles and the mid-distance between the ulnar and radial markers. The upper arm was defined from the midepicondyle of the elbow and the center of the glenohumeral joint as defined by statistical equation from the acromion. Also, the gender and the weight of the person were used in the statistical equation in (25) to estimate the location of the center of mass of each segment as well as the moment of inertia around each axis.

The outputs of the calculation were the bilateral net joint forces and moments acting at the shoulder joints and the segment angular velocities in the global coordinate system. The net shoulder joint forces and moments represent the actions exerted by the proximal segment on the distal segment and were expressed in the joint coordinate system (JCS) proposed by (26). Positive shoulder moments were in flexion, adduction, and internal rotation, whereas positive forces were medial, anterior, and proximal.

## Data Analysis

For each trial, the 10 most repeatable push cycles were used (8, 27). The bilateral Fres, Ftan, MEF, and shoulder joint moment and force components were normalized with respect to the push phase in 101 data points and they were subsequently divided into four quartiles: $Q_1 = 0–25\%$, $Q_2 = 25–50\%$, $Q_3 = 50–75\%$, and $Q_4 = 75–100\%$. The analysis was specifically conducted on quartiles $Q_2$ and $Q_3$ because the HB was generally active in this portion of the push phase and also because most of the propulsion effort was provided in the middle of the push. The average of the MEF and each moment components were calculated during $Q_2$ and $Q_3$ for the seven experimental conditions (i.e., INI, BL1-BL5, and POST).

## Statistical Analysis

All the dependent variables (i.e., average of the Fres, Ftan, and MEF) as well as the shoulder moment components during $Q_2$ and $Q_3$ for the flexion/extension, adduction/abduction and internal/external rotation moments, and medial/lateral, anterior/posterior, and proximal/distal force components met the normality criteria (Shapiro–Wilk test, $p > 0.05$). Repeated measures ANOVAs were performed for the dependent variables in order to determine the effect of training intensities with a significance level set at $p < 0.05$. When a significant main effect was found, *post-hoc* analysis using dependent *t*-tests was performed between the pretraining (INI) condition and each of the five training blocks (BL1–BL5) as well as with the posttraining condition (POST). The significance level was adjusted to account for multiple comparisons using the Bonferroni corrections ($p < 0.05/6 = 0.0083$).

# RESULTS

## Pushrim Kinetics

**Table 2** shows the average of the pushrim kinetics parameters. Significant main effects were found for the Fres and Ftan at the pushrim bilaterally during both the $Q_2$ and $Q_3$. *Post-hoc* analysis revealed that all the force components in the INI condition were significantly lower compared to all the training blocks. Significant main effects were found for the average MEF during $Q_2$ and $Q_3$, as subsequent analysis revealed that the MEF in the INI condition was significantly lower compared to BL3, BL4, and BL5 (**Table 2**). In fact, MEF in $Q_2$ varied from 33 and 35% in INI condition to 48 and 50% in BL5, respectively, for the right and left sides. During the $Q_3$ interval, the MEF varied from 52 and 53% in INI condition to reach 61 and 62% in BL5 condition. We can consider here that the participants modify their MEF toward the direction of the MEF target, which corresponds to a 10% increase at the peak value of the MEF. Since the MEF has a pattern that is participant dependent, we show here that our participant learned the new imposed pattern with our simulator. **Figure 2** shows the time-normalized Fres, Ftan, and MEF for a participant that had the lowest MEF and a participant that had the highest MEF at INI and their patterns for all the training blocks.

## Shoulder Joint Moments

Average (1 SD) of each shoulder joint moment component in N.m can be found in **Table 3**. For the adduction/abduction moment component, significant increases during $Q_3$ on the right side and $Q_2$ on the left side were found between INI and BL3 and BL4. For $Q_3$ on the left side, INI was significantly lower than all the training blocks. Significant differences for the internal/external rotation moment component were only found on the left side. More precisely, significant increases were found between INI and BL2–BL5 during $Q_2$. During $Q_3$, INI was significantly lower than all the training blocks. At the right shoulder, significant increases were found for the average flexion/extension moment component between INI and BL2–BL4 during $Q_2$ and BL3 and BL5 during $Q_3$. On the left side, significant augmentation between INI and all the training blocks during $Q_2$ and $Q_3$ were found except for BL1 during $Q_2$.

## Shoulder Joint Forces

**Table 4** shows the average (1 SD) of the shoulder joint force components in N. The anterior/posterior force component significantly increased bilaterally between INI and all the training blocks during $Q_2$ and $Q_3$. For the distal/proximal force component, only a significant increase between INI and BL4 was observed bilaterally during $Q_3$. Significant increases between INI and BL3–BL5 were observed during $Q_3$ at the right shoulder for the average medial/lateral force component. Meanwhile, at the left shoulder, significant higher average medial/lateral force components were found between INI and all the training blocks during $Q_2$ and $Q_3$.

## DISCUSSION

The purpose of this study was to determine the impact of increasing force effectiveness at the pushrim by 10% during actual manual wheelchair (MWC) propulsion in experienced wheelchair users using the HB simulator on the mechanical load sustained at the shoulder. The value of the MEF obtained during INI condition compares well with previous research among individuals with a SCI where the MEF ranged from 21 to 56% (3, 5, 28). In terms of shoulder joint moments, our results are also in line with previous research that showed that the main moment components were in flexion, adduction, and internal rotation (29–31). For the shoulder net joint forces, the highest components were found in the anterior, proximal, and lateral directions that are in agreement with previous studies on individuals with a SCI (29, 30, 32).

### Haptic Biofeedback Intensity Level Influences the Mechanical Load Sustained at the Shoulder

The targeted MEF in this study was based on the earlier hypothesis postulated from a simulation study by Desroches et al. (2008) that stated that an increase of 10% in the MEF effectiveness would not yield a significant augmentation for shoulder loads. In order to reach the 10% target, the HB corresponding to 15% (BL3) had to be applied. This simulation block yielded statistically significant increases in shoulder mechanical loads. This load was found mostly in the sagittal plane (i.e., flexion moment and anterior force component). This confirms previous suggestion made in simulation and analytic studies (12–15). On average, the increases found in the moments and force ranged from 1 to 7 Nm and 5 to 11 N, respectively, which are of small amplitude and probably only have very limited effect on the risk exposure at the shoulders. Vegter et al. (33) reported a net average moment during the propulsion cycle, which varied from 12.4, 16.1, and 15.3 N.m as measured in three periods of 4 min separated by 2 min rests. These data correspond to able-bodied subjects and are slightly higher than the one presented here for our SCI subjects. Frost et al. (34) suggested that repeated tasks performed with force requirements over 10% of the maximal voluntary contraction could increase the risk of shoulder injury. The increases in moments and forces found for the BL3 training block corresponded to 9.1 and 3.8% of their respective moments and force reached during maximal voluntary propulsive moment test prior to the experiment. Thus, the advantages of an increase mechanical efficiency during propulsion outweigh the increase mechanical demand at the shoulders, as it would reduce push frequency and one could suspect that overall less work will have to be performed to cover the same distance (33). The purpose of this study was to investigate the effect of improving the MEF by using haptic feedback onto the shoulder joint moments. The authors are aware of the difficulty to fix a threshold about the joint moment during manual wheelchair propulsion and a risk of injury. It is known in ergonomic studies that risk of injury is either related to the amount of force applied, but also the repetition. In general, a task that demands 30% of maximal force at the joint is considered as fatiguing and constraining task.

The increases in the mechanical load at the shoulder found in this study, although of small amplitude, might have partly resulted in application of the external forces. Because of the nature of the HB that is to give feedback associated using force application, it is not possible to dissociate the increased force requirements in order to achieve the desired movement pattern. However, it is possible to suspect that over a longer period of time (i.e., longer training), the participants might develop the proper motor pattern that would avoid the increase resistance at the wheel and yield higher propulsion efficiency without the increase mechanical loads (35). Future studies should focus on the adaptation yielded from a longer training program that might give insight into proper future training regimen.

**TABLE 2 |** Average (1 SD) bilateral resultant force at the pushrim (Fres), tangential (Ftan) force component in Newtons, and mechanical effective force (MEF) during $Q_2$ and $Q_3$.

| | | | INI | BL1 | BL2 | BL3 | BL4 | BL5 | POST |
|---|---|---|---|---|---|---|---|---|---|
| FRES (N) | R | $Q_2$ | 23.12 (6.77)$^{12345}$ | **26.57 (8.35)$^{345}$** | **34.00 (16.16)$^{45}$** | **36.81 (11.90)$^{45}$** | **42.50 (13.73)** | **43.39 (16.80)** | 23.28 (7.84) |
| | | $Q_3$ | 31.36 (7.99)$^{12345}$ | **38.29 (9.63)$^{2345}$** | **50.04 (16.39)$^{45}$** | **53.90 (17.92)$^{45}$** | **60.58 (17.27)** | **60.35 (16.78)** | 32.32 (8.95) |
| | L | $Q_2$ | 22.15 (5.70)$^{12345}$ | **26.81 (8.01)$^{2345}$** | **32.80 (13.81)$^{45}$** | **35.07 (12.37)$^{5}$** | **40.65 (13.26)** | **43.26 (16.36)** | 22.94 (6.91) |
| | | $Q_3$ | 29.38 (6.05)$^{12345}$ | **36.77 (8.95)$^{2345}$** | **46.49 (13.50)$^{45}$** | **50.99 (16.75)$^{45}$** | **57.33 (17.30)** | **59.88 (16.45)** | 31.00 (7.98) |
| FTAN (N) | R | $Q_2$ | 41.62 (7.16)$^{12345}$ | **46.50 (9.82)$^{2345}$** | **54.61 (13.88)$^{45}$** | **57.18 (15.00)$^{45}$** | **63.25 (16.21)** | **62.99 (15.82)** | 42.53 (8.08) |
| | | $Q_3$ | 44.52 (10.70)$^{12345}$ | **53.50 (13.88)$^{2345}$** | **66.31 (20.32)$^{45}$** | **70.63 (25.00)$^{5}$** | **77.53 (23.71)** | **77.84 (19.82)** | 46.31 (12.72) |
| | L | $Q_2$ | 38.52 (7.33)$^{12345}$ | **44.78 (8.55)$^{2345}$** | **52.38 (11.87)$^{5}$** | **54.34 (14.35)$^{5}$** | **59.18 (15.33)** | **61.12 (15.38)** | 40.27 (7.18) |
| | | $Q_3$ | 41.66 (9.67)$^{12345}$ | **51.26 (14.33)$^{2345}$** | **63.41 (20.03)$^{45}$** | **68.30 (26.09)$^{5}$** | **74.11 (24.59)$^{5}$** | **78.04 (21.99)** | 44.56 (13.25) |
| MEF | R | $Q_2$ | 0.33 (0.13)$^{345}$ | **0.34 (0.12)$^{345}$** | **0.39 (0.19)** | **0.43 (0.14)** | **0.46 (0.15)** | **0.48 (0.18)** | 0.32 (0.15) |
| | | $Q_3$ | 0.52 (0.13)$^{345}$ | **0.54 (0.13)$^{345}$** | **0.59 (0.16)** | **0.61 (0.14)** | **0.63 (0.12)** | **0.61 (0.14)** | 0.51 (0.15) |
| | L | $Q_2$ | 0.35 (0.12)$^{345}$ | **0.37 (0.13)$^{345}$** | **0.40 (0.19)$^{5}$** | **0.43 (0.16)** | **0.48 (0.14)** | **0.50 (0.19)** | 0.34 (0.14) |
| | | $Q_3$ | 0.53 (0.14)$^{345}$ | **0.55 (0.14)$^{345}$** | **0.57 (0.17)$^{45}$** | **0.60 (0.16)** | **0.63 (0.15)** | **0.62 (0.16)** | 0.53 (0.16) |

*Bold characters denote significant main effect for training intensities ($p < 0.05$).*
$^{x}$*significant difference found with the training block (BLX) ($p < 0.0083$).*

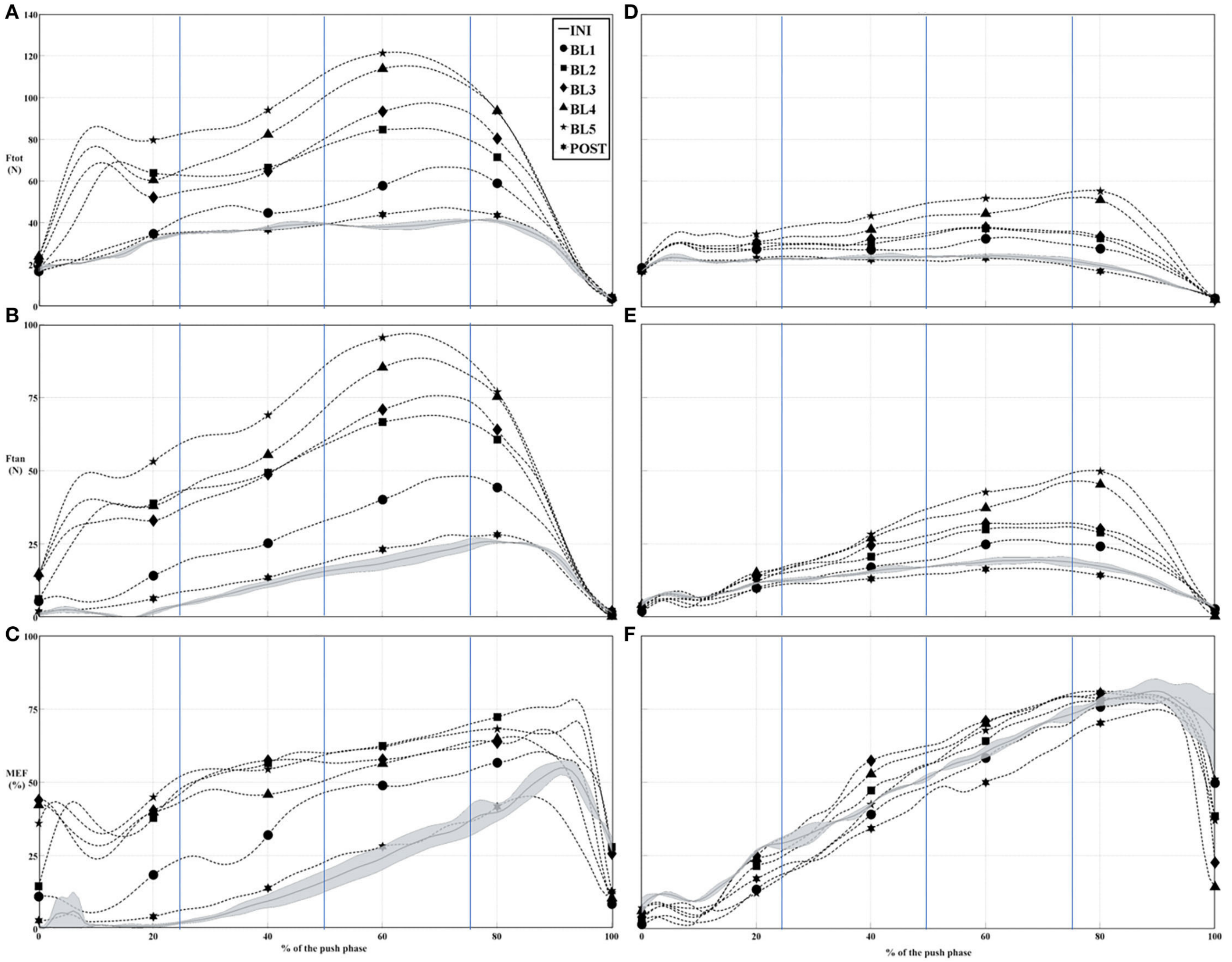

FIGURE 2 | Normalized time series of the Ftot, Ftan, and mechanical effective force (MEF) for a participant that was initially inefficient **(A–C)** and a participant that had initially high MEF **(D–F)**. Vertical lines indicate each quartile separation, i.e., 25, 50, and 75%.

## Haptic Biofeedback as a Training Tool for Wheelchair Propulsion to Increase the Mechanical Efficiency

The premise behind the use of the HB is that it provides the motor system with additional proprioceptive and somatosensory cues to enhance motor planning (35). These additional cues might yield specific neural adaptations based on the desired imposed movement (36) and have a better potential for long-term residual effect when used as a training method, even more so if combined with visual feedback (10, 35, 36). These adaptations or the changes elicited when subjected to the HB might be more evident when the participants are either novice to the task or have a poor initial performance (37). As highlighted in **Figure 2**, a participant that was initially inefficient (i.e., poor performer; MEF = 20%) seems to benefit greatly from the HB training, whereas a participant that had initially an efficient propulsion (i.e., MEF = 50%) did not modified his response to the HB training. Thus, this suggest that the training should be adapted to each individual and future studies should focus on investigating which parameters would be more beneficial in order to optimize propulsion performance.

In this study, different blocs of haptic feedback BL1–BL5 were investigated during wheelchair propulsion. The last bloc BL5 induced a high resistance. The general idea in this study was to prove that the continuous modification of the MEF was possible, since the direction of the resultant force tended to follow the targeted direction. The targeted directions have been arbitrarily set by adding 10% to the initial MEF peak of each participant. It happens in this study that during training with BL3 block, the measured MEF was close to the arbitrarily imposed MEE target. To find out the reason of this behavior, we suggested to base this study to the general organizational principle of control. In fact, van Ingen Schenau et al. (38) have shown that many tasks have a conflicting effect in terms of orientation of reaction forces and the distribution of net joint moments either in the upper limb (push and pull) or lower limb (cycling). They have attributed a special

**TABLE 3 |** Average (1 SD) bilateral shoulder joint moment components in N.m during $Q_2$ and $Q_3$.

| | | | INI | BL1 | BL2 | BL3 | BL4 | BL5 | POST |
|---|---|---|---|---|---|---|---|---|---|
| ADD(+)/ABD(–) | R | $Q_2$ | 1.32 (1.24) | 1.30 (1.09)$^{4}$ | 1.87 (1.76) | 1.95 (1.66) | 1.74 (1.40) | 1.96 (1.57) | 1.36 (0.98) |
| | | $Q_3$ | 1.94 (1.49)$^{34}$ | **2.17 (1.51)$^{4}$** | **2.53 (1.46)** | **2.84 (1.81)** | **3.24 (1.69)** | **3.16 (1.81)** | 2.11 (1.33) |
| | L | $Q_2$ | 1.59 (1.13)$^{345}$ | **2.23 (1.56)$^{4}$** | **2.91 (2.72)** | **3.35 (2.79)** | **3.58 (2.71)** | **3.65 (2.42)** | 1.92 (1.22) |
| | | $Q_3$ | 2.84 (1.83)$^{12345}$ | **3.99 (2.34)$^{345}$** | **5.03 (2.83)$^{45}$** | **6.23 (3.97)** | **6.48 (3.39)** | **6.49 (3.77)** | 3.47 (1.97) |
| INT (+)/EXT (–) ROTATION | R | $Q_2$ | 4.89 (2.27) | 5.01 (2.06) | 5.18 (2.90) | 5.87 (3.20) | 6.19 (2.65) | 6.01 (2.97) | 4.88 (2.28) |
| | | $Q_3$ | 4.10 (2.12) | 4.40 (1.91) | 4.60 (2.35) | 4.76 (2.22) | 5.15 (1.92) | 5.44 (2.75) | 4.07 (1.94) |
| | L | $Q_2$ | 7.59 (4.54)$^{345}$ | **9.36 (4.48)** | **9.88 (4.73)** | **11.93 (6.18)** | **12.79 (7.12)** | **13.03 (9.69)** | 8.01 (3.04) |
| | | $Q_3$ | 6.35 (3.47)$^{12345}$ | **8.10 (3.44)$^{345}$** | **9.14 (4.01)$^{45}$** | **10.79 (4.45)** | **11.34 (5.21)** | **11.96 (5.10)** | 6.99 (2.81) |
| FLEX (+)/EXT (–) | R | $Q_2$ | 11.60 (3.92)$^{2345}$ | **12.66 (4.07)$^{345}$** | **13.89 (4.22)$^{5}$** | **15.38 (4.63)** | **16.72 (5.72)** | **16.40 (5.01)** | 11.62 (3.79) |
| | | $Q_3$ | 8.41 (3.84)$^{35}$ | **8.98 (3.77)$^{5}$** | **10.06 (4.47)** | **10.23 (3.30)** | **10.97 (5.30)** | **12.05 (5.02)** | 8.32 (3.24) |
| | L | $Q_2$ | 12.84 (4.97)$^{12345}$ | **15.68 (5.08)$^{345}$** | **17.01 (5.55)$^{45}$** | **19.77 (6.65)** | **21.61 (7.58)** | **21.76 (8.99)** | 13.66 (3.89) |
| | | $Q_3$ | 9.49 (4.49)$^{2345}$ | **11.47 (4.92)$^{345}$** | **13.00 (5.87)$^{45}$** | **14.88 (4.96)** | **15.70 (6.41)** | **16.62 (6.55)** | 10.23 (4.03) |

*Bold characters denote significant main effect for training intensities ($p < 0.05$).*
$^{x}$*significant difference found with the training block (BLX) ($p < 0.0083$).*

**TABLE 4 |** Average (1 SD) bilateral shoulder joint force components in N.m during $Q_2$ and $Q_3$.

| | | | INI | BL1 | BL2 | BL3 | BL4 | BL5 | POST |
|---|---|---|---|---|---|---|---|---|---|
| ANT(+)/POST(–) | R | $Q_2$ | 26.21 (8.07)$^{12345}$ | **31.08 (8.58)$^{2345}$** | **37.69 (10.42)$^{45}$** | **42.09 (12.95)$^{4}$** | **47.48 (13.57)** | **44.91 (12.35)** | 27.59 (8.69) |
| | | $Q_3$ | 27.53 (8.90)$^{12345}$ | **34.38 (10.83)$^{2345}$** | **42.09 (16.88)$^{45}$** | **45.56 (17.70)$^{5}$** | **49.44 (16.39)** | **51.55 (14.29)** | 29.10 (11.27) |
| | L | $Q_2$ | 24.13 (9.70)$^{12345}$ | **29.54 (9.78)$^{345}$** | **34.00 (11.16)$^{45}$** | **38.55 (14.57)** | **43.00 (15.75)** | **41.84 (14.91)** | 25.83 (8.98) |
| | | $Q_3$ | 25.06 (7.62)$^{12345}$ | **32.28 (10.40)$^{2345}$** | **39.14 (15.27)$^{45}$** | **43.98 (17.68)5** | **46.90 (15.44)** | **50.38 (16.98)** | 27.53 (9.71) |
| PROX (+)/DIST (–) | R | $Q_2$ | 26.15 (7.12) | 26.18 (7.10) | 21.90 (8.23) | 24.25 (7.91) | 24.07 (10.02) | 23.60 (11.83) | 26.17 (8.37) |
| | | $Q_3$ | 12.75 (7.48)$^{4}$ | 11.85 (7.78)$^{24}$ | 8.59 (6.11) | 10.05 (7.52) | 7.64 (5.12) | 10.27 (11.68) | 12.48 (9.16) |
| | L | $Q_2$ | 26.45 (7.92) | 27.53 (8.77) | 24.83 (11.49) | 26.91 (9.88) | 27.08 (11.46) | 25.65 (13.61) | 27.23 (8.82) |
| | | $Q_3$ | 15.41 (8.90)$^{4}$ | **16.51 (9.70)** | **13.50 (11.64)** | **12.49 (10.35)** | **12.35 (8.05)** | **13.67 (8.19)** | 15.32 (10.62) |
| LAT (+)/MED (–) | R | $Q_2$ | 7.12 (3.75) | 6.96 (3.14) | 6.63 (3.31) | 6.66 (3.15) | 6.61 (3.97) | 6.76 (3.53) | 6.93 (3.12) |
| | | $Q_3$ | −5.44 (4.73)$^{345}$ | **−7.05 (3.86)$^{345}$** | **−9.51 (5.59)** | **−10.67 (5.63)** | **−12.36 (6.86)** | **−11.36 (7.55)** | −6.66 (3.91) |
| | L | $Q_2$ | 7.45 (4.66) | **6.38 (4.65)** | **6.52 (4.81)** | **7.44 (5.60)** | **7.50 (6.06)** | **7.40 (5.33)** | 6.20 (5.49) |
| | | $Q_3$ | −9.20 (6.15)$^{12345}$ | **−12.20 (6.88)$^{345}$** | **−16.31 (10.23)$^{4}$** | **−22.22 (17.33)** | **−23.14 (15.67)** | **−23.62 (19.76)** | −10.24 (5.53) |

*Bold characters denote significant main effect for training intensities ($p < 0.05$).*
$^{x}$*significant difference found with the training block (BLX) ($p < 0.0083$).*

role to biarticular muscles as responsible for the direction of the reaction forces, whereas the work done by this reaction force will be mainly realized by monoarticular muscles. It will be interesting in the future to test this hypothesis with either muscular activity measurement or musculoskeletal modeling approach.

In earlier study, Blouin et al. (8) have shown that some subjects keep their new MEF pattern slightly higher that the pretraining pattern [see Figure 8 in (8)] into the posteffect condition. More precisely, 7 subjects rise their MEF with respect to the initial one, whereas 11 subjects lower their MEF during the posteffect condition. It is known that learning consolidation necessitates many training periods during weeks. Unfortunately, with the data of this study, it is not possible to predict the consolidation of the new MEF pattern and future study will help to see if longitudinal training can improve the original MEF pattern.

## Limitations

This study has a few limitations. The proposed training with the HB was tested on 18 participants with a SCI, which limits the generalization of this study to the other manual wheelchair users. In addition, although the parameters of the simulator were adjusted for each participant, the propulsion training on the simulator was still conducted using a single standard wheelchair for all the participants. Hence, the participants may have been less adapted to propelling a wheelchair that was not theirs. Future studies should focus on the adaptation yielded from a longer training program that might give insight into proper future training regimen. Moreover, in the inverse dynamic model, we do not consider all the shoulder girdle joints and possible contribution of clavicle and scapula motions to glenohumeral loading.

## CONCLUSION

Increases in shoulder loads were found for the corresponding training blocks but even though the percentage of the increase seems high, the amplitude of the joint moment remains under the values of wheelchair propulsion found in the literature. The use of a haptic feedback (HB) simulator is considered here as a safe approach to increase mechanical effectiveness. However, the longitudinal impact of this enhancement remains unknown for the impact on the shoulder joint. Future studies will be focused on this impact in terms of shoulder risk injury during manual wheelchair propulsion.

## DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

## ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Ecole de technologie supérieure Centre of Interdisciplinary Research in Rehabilitation of Greater Montreal. The patients/participants provided their written informed consent to participate in this study.

## AUTHOR CONTRIBUTIONS

RA was responsible for the collect of data, computational analysis, and writing. DG was responsible for experimental design and statistical analysis. All authors contributed to the article and approved the submitted version.

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REVIEW
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# A Scoping Review on Shoulder Injuries of Wheelchair Tennis Players: Potential Risk-Factors and Musculoskeletal Adaptations

*Laura Mayrhuber[1,2†], Thomas Rietveld[1*†], Wiebe de Vries[2], Lucas H. V. van der Woude[1,3,4], Sonja de Groot[5,6] and Riemer J. K. Vegter[1,3]*

*[1] Center for Human Movement Sciences, University Medical Center Groningen, University of Groningen, Groningen, Netherlands, [2] Swiss Paraplegic Research, Nottwil, Switzerland, [3] School of Sport Exercise & Health Sciences, Peter Harrison Centre for Disability Sport, Loughborough University, Loughborough, United Kingdom, [4] Center for Rehabilitation, University Medical Center Groningen, University of Groningen, Groningen, Netherlands, [5] Amsterdam Rehabilitation Research Center Reade, Amsterdam, Netherlands, [6] Department of Human Movement Sciences, Faculty of Behavioral and Movement Sciences, Vrije Universiteit, Amsterdam, Netherlands*

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**Edited by:**
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***Correspondence:**
*Thomas Rietveld*
*t.rietveld@umcg.nl*

*† These authors have contributed equally to this work and share first authorship*

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Wheelchair tennis players are prone to develop shoulder injuries, due to the combination of wheelchair propulsion, overhead activities and daily wheelchair activities. A methodical literature search was conducted to identify articles on shoulder complaints in wheelchair tennis, wheelchair sports and tennis. The aims were to identify (1) type of shoulder complaints; (2) possible risk factors for the development of shoulder injuries; (3) musculoskeletal adaptations in the shoulder joint in wheelchair tennis players. Fifteen papers were included in this review, five on wheelchair tennis, three on wheelchair sports and seven on tennis. Type of shoulder complaints were acromioclavicular pathology, osteoarthritic changes, joint effusion and rotator cuff tears. Possible risk factors for the development of shoulder injuries in wheelchair tennis are overhead movements, repetitive activation of the anterior muscle chain and internal rotators, as well as a higher spinal cord injury level. Muscular imbalance with higher values for the internal rotators, increase in external range of motion, decrease in internal range of motion and reduced total arc of motion were the most common proposed musculoskeletal adaptations due to an unbalanced load. These presented risk factors and musculoskeletal adaptations might help researchers, coaches and wheelchair tennis players to prevent shoulder injuries.

**Keywords: wheelchairs, shoulder injuries, physical activity, wheelchair tennis, adapted sports**

## INTRODUCTION

Wheelchair sports participation, like wheelchair tennis, is growing in popularity and is a great opportunity for people with disabilities to get physically active (1). Wheelchair users have an elevated risk to develop various diseases due to a restricted mobility and often sedentary lifestyle, therefore, exercise is crucial to maintain health (2–4). Even though sports participation in wheelchair sports has a broad range of positive effects, it also leads to an increase in stressors on the shoulder complex in addition to the loading from daily activities (5, 6). The prevalence of shoulder problems in wheelchair athletes is reported to have a broad range, i.e., from 16% (7) up to 76% (8). This is similar to able-bodied tennis, in which the shoulder is the most common area of injury of the upper extremity (9). Shoulder pain is prevalent in 24% of the elite tennis players (12–19 years old) (10).

Becoming wheelchair dependent changes the role of the shoulder complex, from providing a great range of motion (ROM) to perform small and detailed movements, into the main source of power for mobility in daily life (11). The motion sequence of wheelchair propulsion itself puts relatively low internal joint forces on the shoulder during regular wheelchair propulsion (12, 13). However, the high frequency of performing the movement in addition to the high shoulder load during specific daily activities, such as transfers in and out of the wheelchair, result in a high exposure to the shoulder joint (12). Changes in the role of the shoulder complex, which require an increased force generation of the upper extremity might lead to imbalances of the muscular system and impact the positioning of the scapula in respect to the humerus as well as both in respect to the thorax (14). Altered conditions in the shoulder joint favor an impingement within the subacromial space and a greater abrasion of the joint (6, 15).

Wheelchair tennis is the most popular adapted racket sport but it involves a high incidence of shoulder complaints (16–19). In wheelchair tennis, the tennis racket is an additional constraint during propulsion of the wheelchair since it interferes with the hand/rim interaction (20). With the racket in one hand, which leads to unilateral power losses because of the more difficult coupling to the hand rim, greater forces need to be produced to maintain balanced power production at both sides (21, 22). As in able-bodied tennis, wheelchair tennis players have a repetitive activation of the anterior muscle chain, due to the unidirectional movements of the strokes. Furthermore, a seated position, as is the case in wheelchair tennis, leads to a modified force generation, as well as changes in shoulder alignment and trunk rotations (17, 23, 24). The core stability and sitting position in the wheelchair have a great impact on the shoulder mechanics and, therefore, on the force generation in the serve and ground strokes (23).

Wheelchair dependence and overhead activities in combination with high training intensities increase the already heavy strain on the shoulder and might be a possible risk factor for overuse injuries in wheelchair tennis athletes (22, 25). Injuries to the upper extremity or overuse symptoms not only negatively affect sport performance but also have a tremendous impact on body functions, activity, and participation in daily life (11). Therefore, it is highly important to identify possible causes and aggravating factors and avoid shoulder injuries in wheelchair tennis. The aims of this review are to: (1) identify type of shoulder complaints; (2) potential risk factors for the development of shoulder injuries in wheelchair tennis; (3) investigate potential musculoskeletal adaptations causing shoulder complaints in the shoulder joint in wheelchair tennis. Given the small number of wheelchair tennis papers, an overview will be given from a wheelchair tennis perspective, as well as a broader view from a wheelchair sports and able-bodied tennis perspective. Due to the recency of written reviews by Heyward et al. (22) on shoulder injuries in wheelchair sports and by Kekelekis et al. (26) on shoulder injuries in able-bodied tennis, these two papers were taken as central papers in the respective parts of the current review and extended with additional papers.

# MATERIALS AND METHODS

## Search Strategy

A methodical search strategy was conducted in October 2020 using the PRISMA checklist (**Supplementary Table 1**) for Scoping reviews by two independent researchers (LM, TR) to identify relevant published articles on the topic of shoulder complaints in (i) wheelchair tennis, (ii) wheelchair sport and (iii) tennis. In case of discrepancies between authors, articles were discussed between the two researchers. PubMed and Web of Science were used to search for relevant articles. The PubMed search strategy shown below was adapted for the second database Web of Science.

(1) ("Wheelchairs"[Mesh])

(2) ("Sports"[Mesh])

(3) ("Tennis"[Mesh])

(4) ("Shoulder Joint"[Mesh] OR "Upper Extremity"[Mesh] OR "Shoulder"[Mesh] OR "Scapula"[Mesh] OR "Rotator Cuff"[Mesh])

(5) ("Muscle Strength"[Mesh] OR "Pain"[Mesh] OR "Musculoskeletal Pain"[Mesh] OR "Chronic Pain"[Mesh] OR "Shoulder Pain"[Mesh] OR "Wounds and Injuries"[Mesh] OR "Athletic Injuries"[Mesh] OR "Rotator Cuff Injuries"[Mesh] OR "Tendon Injuries"[Mesh] OR "Stress Disorders, Post-Traumatic"[Mesh] OR "Arm Injuries"[Mesh] OR "Shoulder Impingement Syndrome"[Mesh] OR "Shoulder Injuries"[Mesh] OR "Bursitis"[Mesh] OR "Rotator Cuff Tear Arthropathy"[Mesh] OR "Risk"[Mesh] OR "Risk Factors"[Mesh] OR "Health Risk Behaviors"[Mesh] OR "Pathology"[Mesh] OR "Syndrome"[Mesh] OR cause*[tiab] OR mechanism*[tiab] OR complaint*[tiab] OR discomfort*[tiab])

Search string – Wheelchair tennis: (1), (3), (4) and (5)

Search string – Wheelchair sports: (1), (2), (4) and (5)

Search string – Tennis: (3), (4) and (5)

Articles from the database search were first checked for duplicates, secondly the titles and abstracts were screened. Thirdly, the full text of the remaining articles was assessed and included if criteria were met.

### Inclusion Criteria

Articles in the English language that incorporated some type of shoulder complaint or assessment either in wheelchair sports, tennis or a combination of the two.

### Exclusion Criteria

Papers from all categories (wheelchair tennis, wheelchair sports, tennis) were excluded if they had a treatment/ intervention program, an assessment was evaluated/tested and when it was an epidemiological study. For the able-bodied tennis and wheelchair sport papers, articles were also excluded when pain in the shoulder joint was not reported. This was not an exclusion criterion for the wheelchair tennis papers, due to the scarcity of available literature.

## Data Extraction and Quality Assessment

Quality assessment was also performed by two independent researchers (LM, TR) for all included articles and was performed with a checklist of Webster et al. (27), adapted by Heyward et al.

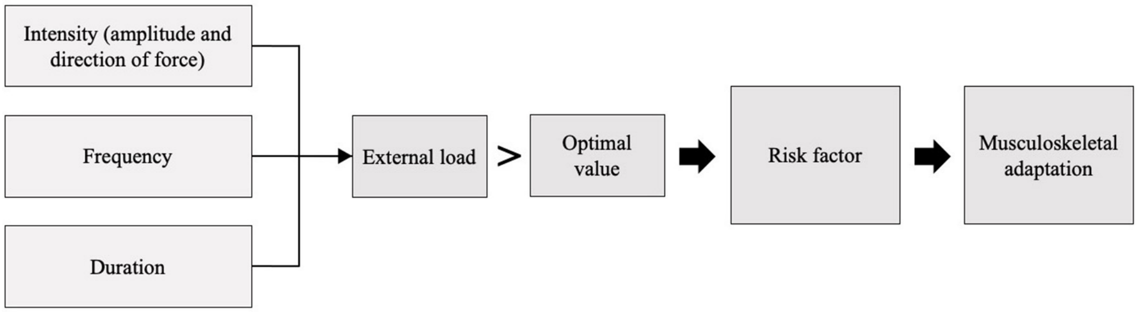


**FIGURE 1 |** Conceptual model of risk factors and musculoskeletal adaptations for shoulder injuries, based on the model of Hoozemans et al. (28).

(22). This checklist was chosen because there is no standardized checklist available for this type of study. For each question a score of 1 was given for an "adequate" or "yes" response, a score of 0.5 was given for a "partial" or "limited" response; and a score of 0 was awarded for a "no", "not stated" or "inadequate" response. A maximum score of 8 was possible. There were no minimum criteria set due to the limited number of papers that were included in the study.

## Definitions of Risk Factors and Musculoskeletal Adaptations

Risk factors for complaints in this review were defined based on Hoozemans et al. (28) in which "external load" was defined using three factors: intensity, frequency and duration (**Figure 1**). The risk for complaints occurs if the value of one of these three factors or the combination of the factors deviates from their optimal value (28, 29). Musculoskeletal adaptions are caused by the risk factors and lead to unfavorable biomechanical conditions in the shoulder complex. An example of a risk factor could be an increased internal rotation balance ratio, due to greater activation of the anterior muscles and repetitive movements. The musculoskeletal adaptation that occurs could be a muscular imbalance. Due to the limited research in the topic, statistically proven risk factors as well as proposed risk factors were included in this review.

# RESULTS

A flow chart of the selection process is shown in **Figure 2**. Five papers were included regarding wheelchair tennis. For wheelchair sports, an interpretation of 13 papers of the review of Heyward et al. (22) will be given, with an additional three papers selected for the current review. For the tennis papers, an interpretation of 23 papers of the review of Kekelekis et al. (26) will be given, with an additional seven papers for the current review.

In 12 of the 15 included articles shoulder problems or a history of shoulder problems were reported, of which eight included clinical testing of the shoulder complaint. A wide variety in screening of indicators for shoulder complaints were reported. Radiographic analysis was used in three articles (18, 30, 31), a strength test in six (30, 32–36), a kinematic analysis in three (5, 24, 37), a kinetic analysis in two (38, 39), the Wheelchair User's Shoulder Pain Index (WUSPI) in two (5, 31), visual analogue scale (VAS) in two (33, 34), range of motion (ROM) measurement in four (32–34, 40) and the scapular resting position (25), perceived function (25), serve speed (34) and post impact ball velocity (39) in one of the articles.

## Quality of the Evidence

The results of this review should be viewed with consideration to the level of evidence (**Supplementary Table 2**). The quality of the articles in the review of Heyward et al. (22) were checked using the same checklist and ranged from low (3) to good (7), with a mean of around 4.5. Especially inclusion/exclusion criteria, reliability and validity were poorly described across papers. The quality of the articles in the review of Kekelekis et al. (26) were checked using a different, extensive checklist by Downs and Black (41). Due to the great number of subcategories and their specificity, the overall score average was low to moderate, with especially the internal validity lacking.

Of all 15 included papers in the current review, eight described the participants characteristics adequately (5, 18, 24, 31–34, 36). Six papers fully described inclusion and exclusion criteria (5, 25, 30, 33, 34, 40) and 13 described the limitations of the study (5, 18, 25, 30–35, 37–40). The key variables, pain, strength and injuries were measured adequately in seven of the 15 papers (5, 18, 24, 32, 34, 37, 39). Overall, the validity and reliability of the used assessments had limited description. In six of the included studies the reliability (5, 24, 30, 32, 33, 40) and in seven the validity (5, 24, 30, 31, 33, 39, 40) were adequately described. Only two papers (25, 32) adequately discussed the external validity of the results.

## Type of Shoulder Complaints

### Wheelchair Tennis

An overview of the included papers can be seen in **Table 1**. In the wheelchair tennis papers, two (5, 18) of the five papers reported shoulder complaints by the participants. Causes of complaints in wheelchair tennis were acromioclavicular pathology in the dominant shoulder, osteoarthritic changes, joint effusion and rotator cuff tears (18) in the dominant as well as the nondominant

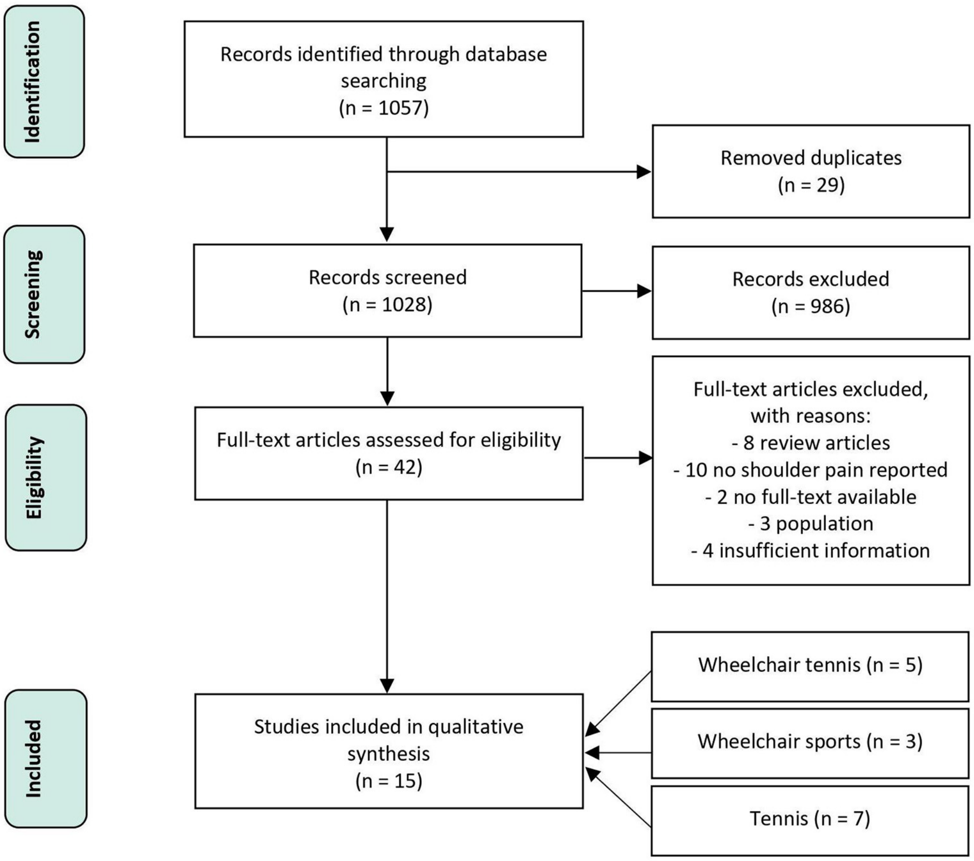


**FIGURE 2 |** Flow chart describing the selection process included.

shoulder, most commonly in the supraspinatus tendon. The paper of Warner et al. (5) reported two participants with previously experienced pain due to shoulder impingement and one participant with subacromial pain syndrome.

### A Broader View From Wheelchair Sports and Able-Bodied Tennis

Pain was reported as the most frequent shoulder complaint in the review of Heyward et al. (22) and the three additional selected papers (25, 31, 37) (**Table 2**). Other shoulder problems included rotator cuff tears, rotator cuff impingement, acromion-clavicular and bicep tendon pathology, subdeltoid and subacromial effusion, as well as non-specific shoulder issues (22). In one of the additional included papers (31) tendinopathy and bursitis were listed as other shoulder complaints. A history of shoulder problems in the selected able-bodied tennis papers and the review of Kekelekis et al. (26) were tendinosis (30), general pain (33, 34, 40, 42), rotator cuff tears or tendinopathy (38, 39, 42, 43), osteoarthritic changes (44) and labral lesion or tears (38, 42, 43).

## Risk Factors

The interpretation of the possible relationships between risk factors and musculoskeletal adaptions are schematically represented using the previous defined model of Hoozemans (**Figure 3**). Due to the low number of articles in wheelchair tennis describing risk factors and musculoskeletal adaptations, a broader view from wheelchair sports and able-bodied tennis is presented as well. Firstly, the risk factors will be described, secondly the musculoskeletal adaptation. These summarizing results will be further interpreted in the discussion part.

### Wheelchair Tennis

A proposed risk factor for shoulder problems in wheelchair tennis, especially in the dominant shoulder, is overuse, caused by wheelchair propulsion, transfers in and out of the wheelchair and playing tennis (18). The repetitive impingement positioning during the play can lead to rotator cuff tears, especially the supraspinatus muscle, and high compressive forces on the acromioclavicular joint (18). The risk for overuse increases when the internal rotation balance ratio is higher compared to the normal range (35). Another proposed risk factor is the level of spinal cord injury (SCI) (36). Higher values of torque and power for internal and external muscles were observed in athletes with a low level SCI (T11-L3) in comparison with athletes with a higher level SCI (T5-T8) (36). The level of lesion does not necessarily influence the rotator balance ratio in the shoulder by the activation of internal rotator muscles but by the participation of the external rotators (36). Age, training time per day, duration of wheelchair usage and wheelchair tennis career did not present as risk factors (18).

**TABLE 1 |** Overview of articles describing type of shoulder complaints, proposed risk factors and musculoskeletal adaptations in wheelchair tennis.

| References | QAS (0–8) | Sport (N) | Disability types | M/F | Age (mean) | Cases shoulder pain/injury | Type of complaint | Objective measure | Clinical testing | Activity level | Sport activity/TSI (years) | Proposed risk factor | Musculo skeletal adaptation |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| Bernard et al. (36) | 3 | WRa/WT (21), ABT (15) | 12 high lesions, 9 low lesions | 36/0 | 27 | X | X | Strength test | X | X | X/13 | Level of SCI | Muscular imbalance |
| Jeon et al. (18) | 5 | WT (33) | Paraplegic | 26/7 | 36 | 23 | Pain, AC pathology, rotator cuff tears, biceps tendon pathology, sub-acromial/ deltoid effusion | Radiographic analysis | Yes | 4–7 h/day | 5–15/6–20 | Overuse, repetitive impingement positioning | Scapula dyskinesis |
| Moon et al. (35) | 2.5 | WT (12) | 10 SCI, 1 amputee, 1 other | X | 33 | X | X | Strength test | X | X | 7/X | X | Muscular imbalance |
| Reid et al. (24) | 3.5 | WT (2) | 1 L1, 1 Incomplete T10 SCI | 2/0 | X | X | X | Kinematic analysis | X | X | X | Reduced shoulder joint loading | X |
| Warner et al. (5) | 7.5 | WT (11) | X | 8/3 | 27 | 1 | Previously experienced pain | Kinematic analysis, WUSPI | Yes | 18 h/week | X / 15 | X | Scapula posterior tilt & external rotation |

*QAS, quality assessment score; TSI, time since injury; AC, acromio-clavicular; WT, wheelchair tennis; WRa, wheelchair racing; ABT, able-bodied tennis; SCI, spinal cord injury; WUSPI, wheelchair user shoulder pain index.*

**TABLE 2** | Overview of articles describing type of shoulder complaints, proposed risk factors and musculoskeletal adaptations in wheelchair sports.

| References | QAS (0-8) | Sport (N) | Disability types | M/F | Age (mean) | Cases shoulder pain/injury | Type of complaint | Objective measure (s) | Clinical testing | Activity level | Sport activity/TSI (years) | Proposed risk factor | Musculo skeletal adaptation |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| Aytar et al. (25) | 5 | Amputee soccer, WB WTT (63) | 29 amputees, 10 poliomyelitis, 4 spina bifida, 6 SCI, 14 others | 55/8 | 24 | X | General pain in the shoulder | Scapular resting position, pain, perceived function | X | X | 6 months /X | X | Abnormal scapula resting position |
| Mason et al. (37) | 4 | WR (10) | 6 HP & 4 LP players | X | 34 | 5 | General pain in the shoulder | Kinematic analysis | X | X | X/14 | X | Rotated scapula |
| You et al. (31) | 4.5 | WTT (19), WAR (16) | 31 SCI, 3 amputees | 24/11 | 47 | X | Tendinopathy, bursitis | WUSPI, Radiographic analysis | Yes | 24,8 h/week | 15/25 | Overuse, high torques on shoulder | X |

*QAS, quality assessment score; TSI, time since injury; WB, wheelchair basketball; WR, wheelchair rugby; WTT, wheelchair table tennis, WAR, wheelchair archery; SCI, spinal cord injury; WUSPI, wheelchair user shoulder pain index; HP, High-point; LP, Low-point.*

### A Broader View From Wheelchair Sports and Able-Bodied Tennis

Participating in wheelchair sports bears several risk factors for shoulder problems, which are multifactorial (22). Proposed risk factors were overuse of the shoulder muscles and less trunk control. Performing overhead sports in a wheelchair increases the risk for rotator cuff tears due to overhead motion and the recurrent microtrauma (8). Repetitive shoulder movements with a great internal rotation torque can lead to the occurrence of a mechanical shoulder impingement and are a proposed risk factor for shoulder overuse injuries (31). No associations were observed between age, wheelchair usage duration, training load and the amount of pain due to shoulder problems (31).

Prolonged tennis exposure was identified as the most common proposed risk factor for able-bodied tennis players in the review of Kekelekis et al. (26), due to its negative effect on muscle performance, serve maximal angular velocities and joint kinetics (26, 45). Furthermore, skill level and technique of the player were identified as risk factors. Less shoulder joint load with a lower risk for the development of injuries was observed in professional tennis players (43). Additional risk factors were a prolonged abduction during the external rotation phase of the serve (46) and scapula dyskinesia (47). Proposed risk factors were a stiffer racket (48), racket with a higher polar moment of inertia (49) and previous injuries (46, 50).

The additional selected papers in this review showed that repetitive overhead movements (42) and overuse due to rigorous training schedules (32) seem to be related to shoulder injuries (**Table 3**). During the serve and smash, the dominant arm is in an abducted position with full external rotation and extension which leads to structural lesions of the rotator cuff and superior labral lesions (42). The overhead motion causes repetitive microtraumas to the capsule and a posterior capsule tightness in tennis players with shoulder pain can be observed (33). Additionally, serve variations like the waiter's serve (39) as well as improper techniques (38) are risk factors for the development of overuse injuries, since alterations in timing of trunk and shoulder rotation in the serve can lead to higher shoulder joint loads. Lastly, a strong upper trapezius in both sides (32) was listed as a proposed risk factor for the development of shoulder injuries in tennis players.

## Musculoskeletal Adaptations

### Wheelchair Tennis

In wheelchair tennis, the supposed musculoskeletal adaptations in shoulder problems are multifactorial. Three of the five papers (18, 35, 36) mentioned a muscular imbalance as alteration in the shoulder girdle, but only one (18) connected it with the occurrence of shoulder problems. Two papers (35, 36) described a muscular imbalance with a higher extension than flexion strength and higher values for internal than external rotator muscles, especially on the dominant side. Differences between the dominant and non-dominant side for scapula

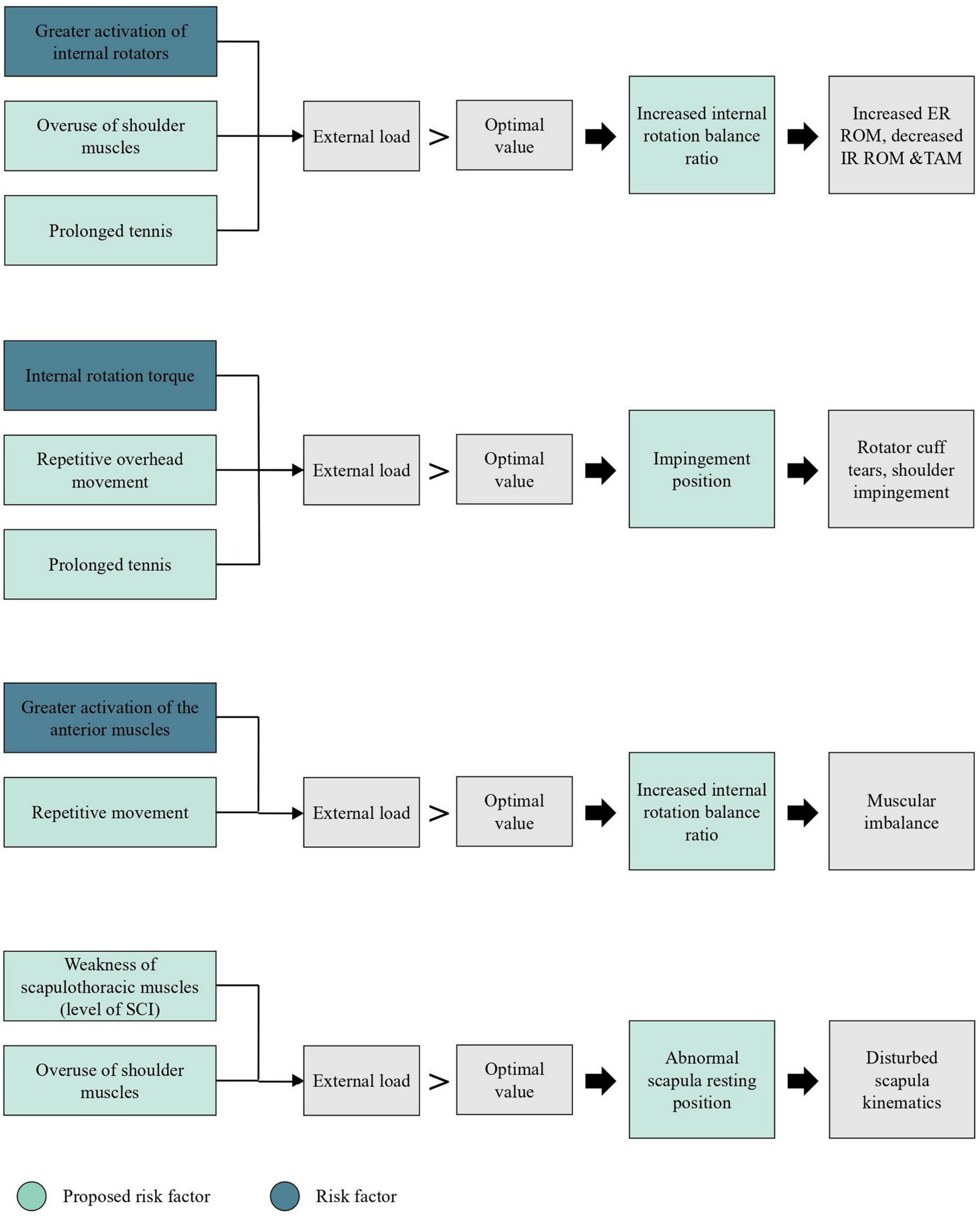


**FIGURE 3 |** Schematic representation of (potential) risk factors and musculoskeletal adaptations for shoulder injuries in wheelchair tennis, based on the conceptual model of Hoozemans et al. (28). ER = external rotation, IR = internal rotation, ROM = range of motion, TAM = total arc of motion, SCI = spinal cord injury.

posterior tilt were observed, with a more posteriorly tilted scapula on the dominant side (5). The upwardly rotated scapula of the dominant arm in wheelchair tennis players was higher compared to able-bodied participants with shoulder impingement (5).

### A Broader View From Wheelchair Sports and Able-Bodied Tennis

Musculoskeletal adaptations associated with shoulder pain in wheelchair sports were difficult to identify. In the review of Heyward et al. (22) it was suggested that shoulder pain was

**TABLE 3 |** Overview of articles describing type of shoulder complaints, proposed risk factors and musculoskeletal adaptations in able-bodied tennis.

| References | QAS (0–8) | Sport (n) | M/F | Age (mean) | Cases shoulder pain/injury | Type of complaint | Objective Measure | Clinical testing | Activity level | Sport activity/TSI (years) | Proposed risk factor | Musculo skeletal adaptation |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| Gillet et al. (32) | 6.5 | ABT (91) | 91/0 | 11 | 30 | History of shoulder problems | Strength test, ROM | Yes | 11h/week | 6/None | X | Muscular imbalance, increased GH ROM |
| Johansson et al. (30) | 6.5 | ABT (35) | 15/20 | 17 | X | Tendinosis | Radiographic analysis, strength test | Yes | 12–20 h/week | X/None | X | Larger infraspinatus & teres minor |
| Marcondes et al. (33) | 8 | ABT (49) | 49/0 | 26 | 27 | Pain in the shoulder | VAS, ROM, strength test | Yes | 8–12 h/week | 8/None | ER strength deficit | Posterior capsule tightness, IR deficit, ER gain |
| Martin et al. (38) | 4 | ABT (20) | 20/0 | 25 | 6 | SLAP lesion, RC tendinopathy, labral tears | Kinetic values, post impact ball velocity | X | X | X | Timing trunk/shoulder rotation in serve, lower ball velocity, high joint kinetics* | X |
| Moreno-Perez et al. (40) | 6.5 | ABT (47) | 43/0 | 23 | 19 | History of shoulder pain | ROM | X | X | 16/None | X | Decreased GH IR & TAM |
| Moreno-Pérez et al. (34) | 5 | ABT (58) | 58/0 | 21 | 20 | History of shoulder pain | ROM, serve speed, strength test, VAS | Yes | 17 h/week | 13/None | X | Muscular imbalance, increased ER ROM, reduced IR ROM |
| Touzard et al. (39) | 4.5 | ABT (18) | 18/0 | 14 | 17 | Shoulder tendinopathy | Kinetic analysis, post-impact ball velocity, | Yes | X | X | Waiters serve posture, higher upper limb kinetics* | X |

*QAS, quality assessment score; TSI, time since injury; ABT, able-bodied tennis; GH, Glenohumeral; ROM, range of motion; ER, external rotation; IR, internal rotation; TAM, Total arc of motion; VAS, visual analogue scale; SLAP, superior labral tear from anterior to posterior; RC, rotator cuff. *Statistically proven risk factors.*

connected to weaknesses in the internal/external rotation, as well as adduction of the shoulder. In one of the additional included papers (25) it was discussed that a weakness of scapula thoracic muscles due to participation in wheelchair sports potentially leads to an abnormal positioning of the scapula. As a consequence, disturbances in the scapula humeral rhythm and general shoulder dysfunction might be observed (25). Scapula position in bilateral shoulder pain in symptomatic individuals had less upward rotation than symptomatic individuals with unilateral pain (37). During the push phase, the scapula moves towards a more internally, upwardly rotated and less anterior position. During the recovery phase the scapula maintained an upward rotated position (37).

Muscular imbalance in the shoulder joint was the most frequent proposed musculoskeletal adaptations in shoulder problems in able-bodied tennis players (30, 32, 34). The studies describe an unbalanced ratio between internal and external rotators in tennis players, especially in the dominant arm. Increases in internal rotators strength are favored due to the demand during tennis strokes (30, 32, 34). A deficit in external rotation strength in the dominant arm in tennis players with shoulder pain has been observed in the study of Marcondes et al. (33). With an imbalance of the muscular system, a change of ROM often takes place, which can be associated with shoulder problems. Four papers (32–34, 40) describe an increase in external ROM, a decrease in internal ROM and a reduced total arc of motion (TAM) in the glenohumeral joint of the dominant arm of tennis players with a history of shoulder pain. The TAM, is defined as the sum of internal rotation ROM and external rotation ROM (32).

## DISCUSSION

The aim of the current review was to identify type of shoulder complaints and potential risk factors for the development of shoulder injuries in wheelchair tennis and investigate potential musculoskeletal adaptations in the shoulder joint in wheelchair tennis players. In the course of this review, risk factors and musculoskeletal adaptations in wheelchair tennis, wheelchair sports and able-bodied tennis were presented (**Figure 3**). There was a scarcity of literature in all three areas, but by connecting available literature, implications for future research and practice were derived.

Overhead activity with the shoulder joint in an impingement position was proposed as a risk factor for shoulder problems in wheelchair tennis (18), wheelchair sports with an overhead movement (8) and able-bodied tennis (42). Overhead activities, like the service or smash in tennis, repeatedly decrease the subacromial cavity by an elevation of the upper arm and lead to an impingement position (42). The supraspinatus tendon passes laterally beneath the cover of the acromion and the bursa subacromialis in the subacromial cavity, therefore it can be damaged due to the repetitive mechanical impingement (51). That could explain the high prevalence of supraspinatus pathology and bursitis in the dominant arm in athletes performing overhead activities (8, 18, 31, 42).

In tennis players with a history of shoulder problems, a reduced glenohumeral TAM was observed (32). Tennis players appear to evolve an increase in external ROM due to osseus alterations, a decrease in internal ROM due to stiffening of the posterior capsule and a loss of TAM in the dominant arm (32, 34). A loss of internal ROM and TAM in the dominant arm compared to the non-dominant arm is a common adaptation in shoulder injuries (32, 34). The rotator cuff muscles have to compensate for the integrity of the shoulder if the ROM and flexibility increases which could then lead to an overuse of the rotator cuff muscles (32, 34).

The combination of being wheelchair-bound and being an overhead athlete can cause alterations in the position of the shoulder joint and scapula which leads to unfavorable biomechanical conditions in the shoulder complex. Wheelchair tennis consists of short intermittent sprints, that demand a constant acceleration and deceleration with changes in direction, as well as the generation of powerful serves and groundstrokes (11, 22). Due to the seated position and lower ball velocities during the serve, wheelchair tennis players reported less load on the shoulder compared to able-bodied tennis players (24). Wheelchair propulsion as well as playing tennis lead to an unbalanced ratio in the dominant arm in tennis players between internal/external rotators due to a high demand of internal rotators during strokes (34, 36). Comparing wheelchair tennis players with able-bodied tennis players, even higher values for internal rotation were observed (36), which suggests a greater muscular imbalance in wheelchair tennis players.

A higher risk of muscular imbalance and shoulder problems seems to occur in wheelchair tennis athletes who have a higher level of SCI and, as a consequence, less trunk control (36). This is in line with the findings of Heyward et al. (22) in which wheelchair athletes with low trunk control had more shoulder complaints compared to athletes with high trunk control. The lack of muscular control and stabilization in the trunk limits the power generation in the kinematic chain (7). Therefore, the upper body has to compensate for the lack of power, which can overload the shoulder joint and increases the stress on the joint (22). In addition, Bernard et al. (36) suggest that a higher level of SCI influences the internal and external rotator ratio by the preferential development of flexor, internal rotator, and adductor muscles. A muscular imbalance oftentimes alters the scapula position to a more upward and internal rotated position (37). In this abnormal position, the impingement within the subacromial space in the shoulder joint is favored and a greater abrasion of the joint occurs, which is suggested to be one of the reasons for shoulder injuries (5, 35).

Given the above-stated factors, wheelchair tennis players are expected to be prone to develop a muscular imbalance which leads to alterations in the joint positioning. This is supported by a study of Aytar et al. (25) that showed that a high percentage of abnormal scapular resting positions was prevalent in wheelchair sports players, which was associated with pain as well as bad perceived shoulder function. In contrary to this hypothesis, Warner et al. (5) reported that the scapula was more posterior tilted and externally rotated on the dominant than the non-dominant side and only one of the wheelchair tennis players reported pain. Postural abnormalities of the scapula, with a protraction of the scapula are associated with decreasing the subacromial space and the prevalence of shoulder impingement (52). The absence of shoulder pain might be related to the posterior tilt of the scapula. A reduced upward rotation, external rotation and posterior tilt of the scapula are increasing the sub-acromial space, which leads to less abrasion in the shoulder joint (52). The connection between an upwardly rotated scapula and a higher prevalence of pain, was also described by Warner et al. (5). It was suggested that an absence of shoulder pain occurred due to a posterior tilted and externally rotated scapula in the dominant arm. The low prevalence of shoulder pain reported in this sample may be explained by a protective benefit due to a specific training program or sports participation, that prevents a protraction and internal rotation of the scapula (5).

### Future Research

Further research should be directed toward more specific wheelchair tennis research focused on the load of the shoulder, risk factors and musculoskeletal adaptations. Shoulder load was never assessed in wheelchair tennis, only the influence of the racket and a different hand rim were investigated (20, 21, 53, 54). First the influence of the racket on shoulder load should be investigated, afterwards wheelchair tennis players with and without shoulder complaints could be compared to identify differences. Further investigation of identified risk factors and musculoskeletal adaptations in the course of this review, such as muscular imbalance and alterations in ROM, can give valuable insight for the development of preventive training and exercise programs for wheelchair tennis players.

### Limitations

Overall, the lack of publications and research in the wheelchair tennis field brought a limited number of papers out of the literature search that investigated shoulder joint injuries in wheelchair tennis. Due to the lack of high-quality literature on wheelchair tennis to be included in this review, it was necessary to combine it with papers about shoulder complaints in other wheelchair sports and able-bodied tennis. This review is a first attempt to gain insight into potential risk factors for shoulder injuries in wheelchair tennis and their musculoskeletal adaptations by comparing and connecting the available information with outcomes of tennis and other wheelchair sports papers.

Additionally, it is important to mention that the included articles about wheelchair sports in general had a relatively low number of participants, which is a common problem in wheelchair sport literature (55). Several papers did not directly investigate risk factors and musculoskeletal adaptations but proposed multiple potential reasons based on their findings, which makes wheelchair tennis focused research even more important. Furthermore, the studies oftentimes did not specify which type of shoulder complaint was the cause and differentiated in the objective measurement tools, which made it challenging to compare the outcomes and draw conclusions.

## CONCLUSION

Risk factors and musculoskeletal adaptations in wheelchair tennis can only be described from a broader wheelchair sports and tennis perspective. Possible risk factors for the development of shoulder injuries in wheelchair tennis are overhead movements, repetitive activation of the anterior muscle chain and internal rotators, as well as a higher SCI level. Muscular imbalance with higher values for the internal rotators, increase in external ROM, decreased internal ROM and reduced TAM were the most common proposed musculoskeletal adaptations due to an unbalanced load. In the future, these risk factors and musculoskeletal adaptations should be investigated in a more wheelchair tennis focused research.

## AUTHOR CONTRIBUTIONS

LM, TR, and RV: conceptualization, investigation, and methodology. LM and TR: formal analysis and writing—original draft. RV, LvdW, SdG, and WdV: supervision. LM, TR, RV, LvdW, SdG, and WdV: writing—review & editing. All authors contributed to the article and approved the submitted version.

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## SUPPLEMENTARY MATERIAL

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# Effect of Fatiguing Wheelchair Propulsion and Weight Relief Lifts on Subacromial Space in Wheelchair Users

*Ursina Arnet[1]*, Michael L. Boninger[2], Ann Cools[3] and Fransiska M. Bossuyt[1,4]*

[1] *Shoulder Health and Mobility Group, Swiss Paraplegic Research, Nottwil, Switzerland,* [2] *Department of Physical Medicine and Rehabilitation, School of Medicine, University of Pittsburgh, Pittsburgh, PA, United States,* [3] *Department of Rehabilitation Sciences and Physiotherapy, University of Ghent, Ghent, Belgium,* [4] *Human Performance Laboratory, Faculty of Kinesiology, University of Calgary, Calgary, AB, Canada*

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***Edited by:***
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*Centre d'Etudes et de Recherche sur l'Appareillage des Handicapés, France*

****Correspondence:***
*Ursina Arnet*
*ursina.arnet@paraplegie.ch*

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**Objective:** This study aimed to identify targets of intervention for reducing shoulder pain in wheelchair users with spinal cord injury (SCI) by (1) examining changes in subacromial space [acromiohumeral distance (AHD) and occupation ratio (OccRatio)] with fatiguing wheelchair propulsion, and different loading conditions [unloaded position vs. weight relief lifts (WRL)]; (2) associating these changes with wheelchair user capacity, as well as (3) identifying subject characteristics associated with subacromial space, such as sex, lesion level, time since injury, body mass index and impaired shoulder range of motion.

**Methods:** Fifty manual wheelchair users with SCI [11 females, age = 50.5 (9.7) years, time since injury = 26.2 (11.4) years] participated in this quasi-experimental one-group pretest-posttest study. Ultrasound images were used to define AHD during an unloaded position, and during personal and instructed WRL before and after fatiguing wheelchair propulsion. Furthermore, supraspinatus and biceps thickness defined from ultrasound images were used to calculate OccRatios. Wheelchair user capacity was quantified as functional strength (maximum resultant force reached during maximum isometric forward push) and anaerobic work capacity (highest power output reached during 15-m sprint test). Multilevel mixed-effects linear regression analyses controlling for between subject variability and covariables were performed to address the research questions.

**Results:** AHD was significantly smaller during personal WRL ($p < 0.001$) and instructed WRL ($p = 0.009$, AHD both 11.5 mm) compared to the unloaded position (11.9 mm). A higher wheelchair user capacity (higher anaerobic work capacity) reduced the impact of WRL on AHD decrease. The fatiguing wheelchair propulsion had no effect on AHD ($p = 0.570$) and on OccRatio of supraspinatus ($p = 0.404$) and biceps ($p = 0.448$). Subject characteristics related to a larger subacromial space were lower lesion level, shorter time since injury, impaired external rotation, a lower body mass index and a higher anaerobic work capacity.

**Conclusion:** This study showed a significant reduction in AHD during WRL with no effect of fatiguing wheelchair propulsion on the subacromial space in wheelchair users with SCI. A higher anaerobic work capacity was beneficial in stabilizing the shoulder during WRL. Our findings may assist clinicians in designing a shoulder injury prevention program.

**Keywords: acromiohumeral distance, occupation ratio, subacromial pain syndrome, impingement, spinal cord injury, fatigue, rotator cuff, shoulder pain**

## INTRODUCTION

Wheelchair users with spinal cord injury (SCI) face high demands on the upper extremity during ambulation, transfers, weight relief lifts (WRL) and numerous other activities of daily living. Especially the shoulder is at high risk for injury and pain. A recent review study reported a pooled prevalence of 44% of shoulder pain in wheelchair users (1).

Pathologies of the rotator cuff have been recognized as one of the main causes of shoulder pain in a general population (2). In manual wheelchair users, rotator cuff disorders are highly present, with supraspinatus tendons most often affected (84-100%) (3, 4). Also, pathologies of the biceps tendons are commonly detected (67-80%) (3, 4). Both tendons pass through the subacromial space and might be compressed due to narrowing of the available space between the humerus and the coracoacromial arch of the scapula. This may result in inflammation, chronic tendon degeneration and/or tendon rupture. Thus, narrowing of the subacromial space is hypothesized as one possible extrinsic mechanism that contributes to shoulder pain (5). Acromiohumeral distance (AHD), which is the shortest linear distance between the most inferior aspect of the acromion and the adjacent humeral head, is a good indicator of the size of the subacromial space and has previously been used to quantify the risk for subacromial pain syndrome (6). The occupation ratio (OccRatio) is defined as the percentage of AHD that is occupied by the tendon (7). This ratio might be even more informative regarding risk for subacromial pain syndrome than absolute distance, since tendon thickness can also change. Both parameters, AHD and OccRatio, can be measured by ultrasound with reliable and consistent results. For AHD, good to excellent intraclass correlation coefficients (ICC) of 0.85-0.98 for intra-rater reliability and 0.88-0.94 for inter-rater reliability were reported (8–10). Regarding OccRatio, ICC values of 0.88-0.92 for intra-rater reliability and 0.79 for inter-rater reliability were reported by BaGcier et al. (8).

The daily demands on the wheelchair users' shoulder may influence OccRatio and therefore the risk for shoulder complaints. The high load acting on the shoulder during weight lifting tasks, such as transfers or WRL for pressure injury prevention might reduce AHD due to cranial humerus migration into the subacromial space (11). The movement of the scapula with respect to the humeral head might further reduce the available subacromial space during these tasks (11). Furthering the risk, the repetitiveness of wheelchair propulsion might fatigue the rotator cuff muscles and change their tendon properties (12) as well as their capability to stabilize the shoulder joint. A better wheelchair user capacity, e. g. higher anaerobic work capacity or functional shoulder muscle strength, might enable a better shoulder stabilization and thus reduce the risk for subacromial pain syndrome (13).

With this study we aimed to identify targets of intervention for reducing shoulder pain in wheelchair users with SCI. The goal of the study was (1) to examine changes in subacromial space (AHD and OccRatio) with fatigue due to wheelchair propulsion, and different loading conditions (unloaded position vs. WRL); (2) to associate these changes with wheelchair user capacity (functional strength and anaerobic work capacity), as well as (3) to identify subject characteristics associated with subacromial space, such as sex, lesion level, time since injury, body mass index (BMI) and impaired shoulder range of motion (ROM). We hypothesized that there will be a significant decrease in AHD and OccRatio due to fatiguing wheelchair propulsion and different loading conditions, and that greater changes will be observed in wheelchair users with a lower capacity.

## MATERIALS AND METHODS

### Study Design and Participants

The study has a quasi-experimental one-group pretest-posttest design (ClinicalTrials.gov, identifier: NCT03153033). Parts of the data collected for this study were published elsewhere (12, 14).

A sample of 50 participants was recruited from the population-based Swiss Spinal Cord Injury Cohort study (SwiSCI) database (15). Inclusion criteria of the study were (1) nonprogressive traumatic or non-traumatic SCI, (2) diagnosed neurological lesion level at T2 or below, (3) at least 1 year post discharge from rehabilitation, (4) between 18 and 65 years old, (5) daily use of a pushrim wheelchair and no required support for propelling for more than 100 m, and (6) quick-release axle to remove wheels from the wheelchair in order to attach a measurement wheel during the later experiment. Exclusion criteria were (1) receiving palliative care, (2) SCI due to congenital conditions, persons with neurodegenerative disorders, or Guillain–Barré syndrome, (3) upper-extremity pain that limits the ability to propel a wheelchair, (4) history of shoulder, elbow, or wrist fractures/dislocations that are still causing symptoms, and (5) history of cardiopulmonary problems that could be exacerbated by strenuous physical activity. In a first step, eligible participants were selected in the SwiSCI database fulfilling inclusion criteria 1, 2, 3, and 4. Subsequently, an information letter including a description of the study, all intended measurements and requirements, as well as a short questionnaire to verify the remaining inclusion and exclusion criteria was sent to the eligible participants. With this procedure a sample size of 50 participants was reached.

Prior to data collection, ethical approval was obtained from the Ethikkommision Nordwest-und Zentralschweiz and all participants read and signed the informed consent.

## Procedure

Participants were invited for one testing session at the biomechanical laboratory of Swiss Paraplegic Research. They were instructed to avoid strenuous exercises 48 h prior to testing.

Several measurements were conducted before and after standardized wheelchair propulsion on a treadmill and a fatiguing intervention of 15 min. The fatiguing intervention was a figure-8 protocol, consisting of three 4-min intervals of maximum voluntary wheelchair propulsion including right and left turns, start and stops, separated by 90 s of rest (total duration of 15 min, **Figure 1**). For that, two cones were placed 18 m apart on a concrete floor and the participants started in the middle of the cones. They were instructed to propel after the start signal as fast as possible toward the first cone, make a right turn around the cone and stop at the starting point. Immediately after a full stop they propelled with a left turn at maximum speed around the second cone and stopped again in at the starting point. This figure-8 was repeated as often as possible within 4 min. Instructions given during fatiguing interventions were standardized. The protocol has been used before in combination with ultrasound examinations (16).

## Data Collection and Analysis

### Subject Characteristics

After introduction of the study and signing the informed consent, participants were asked to self-report socio-demographic variables (age, sex, and height), characteristics of the injury (traumatic or non-traumatic etiology, date of injury, completeness of the injury, and neurological lesion level). Weight was collected with a wheelchair scale by subtracting the weight of the wheelchair from the total weight.

### Range of Motion

Passive shoulder range of motion was measured prior to the fatiguing intervention with a goniometer while sitting in the wheelchair. Shoulder range of motion was classified as impaired when meeting the following criteria: anteroflexion $< 170^{\circ}$, external rotation $< 50^{\circ}$ or abduction $< 170^{\circ}$.

### Wheelchair User Capacity

The wheelchair user capacity tests were performed prior and after the fatiguing intervention. Wheelchair user capacity tests consisted of functional strength test and anaerobic work capacity tests. During the capacity tests, 3-dimensional forces and moments applied to the pushrim were collected at 240 Hz with the SmartWheel (Three Rivers Holdings, Inc, Mesa, AZ) fitted to the non-dominant side of the participants' personal wheelchair. The non-dominant side was chosen as this project aims to investigate the shoulder most predominantly affected by wheelchair propulsion and less by other activities of daily living, such as overhead reaching, lifting objects, etc. A dummy wheel with an equal tire as the SmartWheel was attached to the contralateral side.

To evaluate functional strength, participants performed three times a 5-s maximum isometric forward push with hands on top of the pushrim and wheelchair attached from behind to restrict forward movement (17). Functional strength was defined as the maximum resultant force reached during the three maximum isometric forward pushes.

To determine anaerobic work capacity a 15 meter overground wheelchair sprint was completed prior to the fatiguing intervention (17). The outcome was the peak power output measured during the sprint test.

### Ultrasound: AHD, Tendon Thickness and Occupation Ratio

Ultrasound images of the supraspinatus tendon and the subacromial space of the non-dominant shoulder were taken before any propulsion activity and following the fatiguing intervention. A single examiner (FMB) took all ultrasound images in a randomized order (NextGen Logiq TM e R90.2, GE Healthcare, USA). Image field depth was set at 4 cm and gain was set at 60 dB. To allow for repeated measurements before and after the propulsion tasks with limited error in variation of probe location, a steel reference marker was taped to the skin.

For quantifying AHD in an unloaded position, three images were taken during 90° elbow flexion with the thumb facing upward (**Figure 2**). Furthermore, three images of the AHD were taken during WRL without any instructions given (personal WRL), and during instructed WRL, where participants were asked to depress and retract the shoulders (**Figure 2**). AHD was defined as the shortest distance between the anterior inferior edge of the acromion and the most superior aspect of the humerus (10). The average measure of the three repeated measures was always used. All blinded ultrasound images were analyzed in randomized order by a single examiner (FMB) using Matlab R2016b custom programs (Mathworks, Inc., Natick, MA, USA).

For quantifying supraspinatus tendon thickness, two transverse images were taken in a seated position with the palm placed on the lower back, the shoulder extended, and the elbow flexed posteriorly (**Figure 2**). For the tendon of the long head of the biceps brachii, two longitudinal images were taken in a seated position with 90° elbow flexion and the hand palm facing upward while resting on a cushion (**Figure 2**). The region of interest of each ultrasound image was defined from the interference pattern at the top of the images, created from the steel reference markers attached to the skin. Within the region of interest, tendon thickness was measured as the mean distance between top and bottom border of the tendon and the average of the two repeated measures was used.

The occupation ratio expresses the tendon thickness relative to the available subacromial space. Occupation ratio of the supraspinatus and biceps tendon was calculated as the percentage of the mean tendon thickness relative to the mean AHD (7). Occupation ratio was only calculated for the unloaded position since tendon thickness was not measured during WRL.

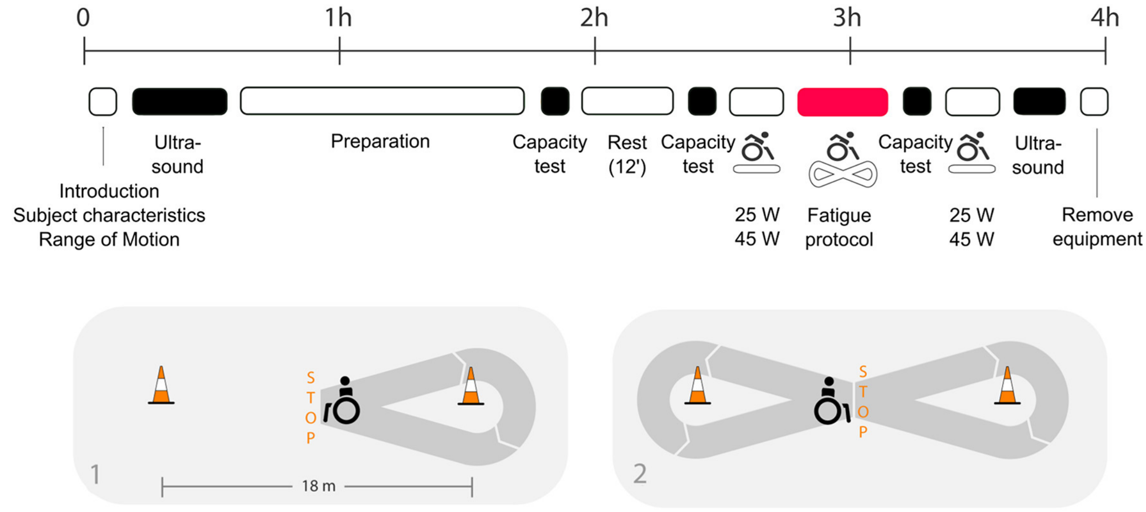


**FIGURE 1 |** Figure adjusted from (12): timeline of the assessments taken in the biomechanical laboratory including (1) introduction, self-reported subject characteristics and measurements of the shoulder range of motion, (2) ultrasound exams (pre and post fatigue), (3) preparation phase including a test to define individual drag force and familiarization with treadmill propulsion and fatigue protocol, (4) wheelchair user capacity test (capacity test): three maximum push tests and a maximum 15 m overground sprint test (pre and post fatigue), (5) passive rest phase, (6) manual wheelchair (MWC) propulsion at two different conditions (25 and 45 W, pre and post fatigue), and (7) fatigue protocol: overground wheelchair propulsion along an eight-shaped course. The detailed course of the fatigue protocol is presented below the timeline.

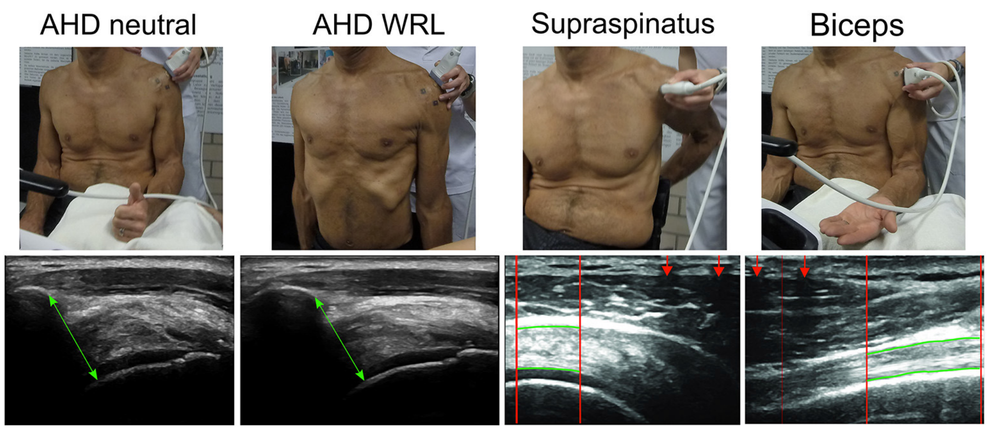


**FIGURE 2 |** Ultrasound measurements: position and example of measurement. Each type of measurements represents a different example participant and does not relate to the person of the position images. AHD neutral: acromiohumeral distance images in the neutral position were taken in a seated position with 90° elbow flexion with the thumb facing upward. AHD WRL: acromiohumeral distance images during WRL. Supraspinatus: transverse ultrasound images of supraspinatus tendon were taken in a seated position with the palm placed on the lower back, shoulder extended, and the elbow flexed posteriorly. Biceps: longitudinal images of long head of biceps brachii tendon were taken in a seated position with 90° elbow flexion and the hand palm facing upward. The region of interest (ROI) is presented between the red vertical lines with the upper border and lower border of the tendons marked with the manually identified red horizontal lines. Lines are marked thicker as compared with the actual analysis for visualization reasons. ROI is selected based on the interference pattern that resulted from a metal marker taped to the skin (assigned with red arrows).

## Statistical Analysis

Statistical analyses were conducted with STATA software (version 16.1, StatCorp, LP, College Station TX, USA). Multilevel mixed-effects linear regression analyses controlling for between subject variability and covariables were performed to:

1) identify the association between the dependent variable AHD and different loading conditions (neural position, personal WRL and instructed WRL) before and after fatigue. Interactions between wheelchair user capacity (functional strength, anaerobic work capacity) with time and loading conditions were included.

2) identify the association between the dependent variable OccRatio of the supraspinatus and biceps tendon before and after fatigue, including interactions with wheelchair user capacity.

Covariables included known risk factors for shoulder pain, such as subject characteristics (sex, lesion level, body mass index (kg/m$^2$), years since injury), impaired shoulder range of motion [in anteroflexion (<170°), external rotation (<50°), and abduction (<170°)], and wheelchair user capacity. If a significant difference ($\alpha = 0.05$) was found between time points or loading conditions, pairwise comparisons with Bonferroni corrections were used to evaluate differences.

# RESULTS

Subject and lesion characteristics of the 50 participants [mean age 50.5 (SD 9.7) years, 11 females, 39 males] are listed in **Table 1**. Mean time since injury was 26.2 (SD 11.4) years and the majority of the participants had a complete lesion (78%). Range of motion in anteroflexion was most often impaired (in 84% of participants), followed by abduction (48%) and external rotation (24%). Regarding wheelchair user capacity, a mean functional strength of 221 N (SD 49 N) was reached during the isometric forward push and a mean power output of 84 W (SD 32 W) was measured during the sprint test.

## Acromiohumeral Distance

AHD values measured pre- and post-fatigue, as well as during different positions (neutral, personal WRL, and instructed WRL) can be found in **Table 2**. AHD was smaller during WRLs compared to the unloaded position. When controlling for all covariables, AHD was significantly larger during the unloaded position [mean 11.9 mm, 95% confidence interval (CI) 11.3-12.5 mm] compared to the personal WRL (mean 11.5 mm, CI 10.9-12.1 mm, $p < 0.001$) and instructed WRL (mean 11.5 mm, CI 10.9-12.1 mm, $p = 0.009$).

No effect of the fatiguing wheelchair propulsion on AHD was found (**Table 2**). When controlling for all covariables, AHD pre-fatigue (mean 11.6 mm, CI 11.0-12.2 mm) was not significantly different than AHD post-fatigue (mean 11.7 mm, CI 11.1-12.3 mm, $p = 0.570$).

There was a significant interaction effect of position and anaerobic work capacity ($p < 0.001$). Participants who reached a lower power output during the sprint test (low anaerobic work capacity) had reduced AHDs during the WRLs compared to the unloaded position. In participants with a higher anaerobic work capacity there was no difference in AHD between unloaded position and WRLs (**Figure 3**).

There were significant associations of AHD and lesion level, as well as impaired ROM (**Table 3**). Participants with lower lesion levels (L1-L2) had a significantly larger AHD (mean 15.2 mm, CI 13.5-16.8mm) than participants with higher lesion levels (T7-T12: mean 11.0 mm, CI 10.0-12.1 mm, T2-T6: mean 10.9 mm, CI 9.7-12.0 mm, both $p < 0.001$). Participants with an impaired ROM in external rotation (<50°) had a larger mean AHD of 13.2 mm (CI 11.9-14.9 mm) compared to participants with no impairments of external rotation ROM (mean 11.1 mm, CI 10.4-11.8, $p = 0.008$). There were no significant associations with any other included subject characteristics (**Table 3**; **Figure 4**).

## Occupation Ratio

OccRatios of the supraspinatus and biceps measured pre- and post-fatigue can be found in **Table 2**. The fatiguing wheelchair propulsion had no effect on the OccRatio of the supraspinatus and biceps. When controlling for all covariables, OccRatio of the supraspinatus was not significantly different pre-fatigue (mean 48.1%, CI 45.2-51.0%) compared to post-fatigue (mean 46.5%, CI 43.6-49.0%, $p = 0.404$). The same accounts for the OccRatio of the biceps, where mean pre-fatigue values of 38% (CI 34.6-41.5) and post-fatigue values of 37.8% (CI 34.4-41.2% $p = 0.448$) were found.

Regarding supraspinatus, participants with a shorter time since injury had lower OccRatio ($p = 0.025$, **Figure 5**). When external rotation ROM was impaired (<50°), participants had a lower supraspinatus OccRatio (mean 39.5%, CI 46.5-53.2%) compared to unimpaired ROM (mean 49.8%, CI 46.5-53.2%, $p = 0.005$, **Table 3**). There were no other significant associations of supraspinatus OccRatio with the analyzed subject characteristics (**Table 3**; **Figure 5**).

OccRatio of the biceps was significantly smaller in participants with a lower BMI ($p < 0.001$, **Figure 5**), as well as in participants with a higher sprint peak power output ($p = 0.006$, **Figure 5**). When external rotation ROM was impaired (<50°), participants had a lower biceps OccRatio (mean 30.2%, CI 23.0-37.5%) compared to unimpaired ROM (mean 40.5%, CI 36.5-44.4% $p = 0.019$, **Table 3**). No other significant associations were found for biceps OccRatio (**Table 3**; **Figure 5**).

# DISCUSSION

This study found a significant reduction in AHD during WRL compared to the unloaded position in 50 wheelchair users with SCI. This was mainly found in participants who reached a lower power output during the sprint test (low anaerobic work capacity). However, fatiguing wheelchair propulsion had no effect on subacromial space since neither AHD nor OccRatio of the supraspinatus and biceps tendon were changed after this intervention. Subject characteristics associated with a larger subacromial space were: lower lesion levels, shorter time since injury, a lower BMI, impaired external rotation ROM and a higher anaerobic work capacity.

## Subacromial Space and Associated Factors

AHD is considered as a good indicator of the size of the total subacromial space (9). In the studied population of wheelchair users with SCI, we found a mean AHD of 11.8 mm during the unloaded position, when the elbow was 90° flexed and the lower arm was supported. These values are slightly higher than previously reported values of 9.4 mm (18) to ∼11 mm (9) in the same population of wheelchair users with SCI. Subacromial space quantified by AHD is an external factor that has been commonly investigated in patients with subacromial pain syndrome. However, no clear association between AHD values in resting position and subacromial pain syndrome was found in previous studies (6, 7, 19, 20).

OccRatio gives more detailed information on the available subacromial space than AHD by taking tendon thickness into account. When analyzing the subacromial space in relation

**TABLE 1 |** Subject characteristics, lesion characteristics, and wheelchair user capacity [% or mean (SD)] for the total sample and stratified by lesion level.

| | Total | Lesion level | | |
|---|---|---|---|---|
| | | **T2-T6** | **T7-T12** | **L1-L2** |
| | **(*n* = 50)** | **(*n* = 20)** | **(*n* = 22)** | **(*n* = 8)** |
| Sex (% male) | 78 | 95 | 68 | 63 |
| Age (years) | 50.5 (9.7) | 48.4 (10.4) | 50.5 (9.6) | 56.0 (6.7) |
| Weight (kg) | 72.4 (13.3) | 73.4 (12.6) | 69.6 (13.0) | 77.4 (15.7) |
| BMI (kg/m$^2$) | 24.0 (4.4) | 23.6 (4.1) | 23.1 (3.7) | 27.5 (5.4) |
| Time since injury (years) | 26.2 (11.4) | 27.2 (11.3) | 24.9 (11.4) | 27.3 (13.1) |
| Lesion completeness (% complete) | 78 | 90 | 77 | 50 |
| ROM anteroflexion (% impaired) | 84 | 100 | 73 | 75 |
| ROM abduction (% impaired) | 48 | 55 | 46 | 38 |
| ROM exorotation (% impaired) | 24 | 30 | 27 | 0 |
| FrMaxpush (N) | 221 (49) | 229 (43) | 214 (54) | 222 (51) |
| Sprint peak power output (W) | 84 (32) | 76 (26) | 91 (37) | 83 (31) |

*BMI, body mass index; ROM, range of motion; FrMaxpush, maximum resultant force reached during the three maximum isometric forward pushes.*

**TABLE 2 |** Unadjusted values [mean (SD)] of the dependent variables acromio-humeral distance (AHD) and occupation ratio (OccRatio) of supraspinatus and biceps tendon pre- and post-fatiguing wheelchair propulsion and during different positions: neutral, personal weight relief (pWRL) and instructed weight relief (iWRL).

| **Time** | | **Pre** | | | **Post** | | | **Mixed Model p values** | |
|---|---|---|---|---|---|---|---|---|---|
| **Position** | ***n*** | **Neutral** | **pWRL** | **iWRL** | **Neutral** | **pWRL** | **iWRL** | **time** | **Position** |
| AHD (mm) | 50 | 11.8 (2.8) | 11.5 (2.5) | 11.5 (2.6) | 12.0 (3.0) | 11.6 (2.7) | 11.6 (2.7) | 0.570 | <0.001$^a$, 0.009$^b$, 0.112$^c$ ♦ |
| OccRatio supraspinatus (%) | 50 | 47.3 (12.2) | | | 45.7 (14.5) | | | 0.404 | |
| OccRatio biceps (%) | 50 | 37.5 (17.9) | | | 37.2 (15.2) | | | 0.448 | |

*Pairwise comparisons: $^a$neutral vs. pWRL, $^b$neutral vs. iWRL, $^c$pWRL vs. iWRL.*
*Significant interactions (alpha = 0.05) from the mixed-effects multilevel analysis: ♦ = interaction position x capacity (sprint).*

to the space occupied by the tendons, we found a mean OccRatio of 47.3% for supraspinatus and 37.5% for biceps. These values are higher than previously reported OccRatios of 36.5% (supraspinatus) and 23.1% (biceps) in wheelchair users with SCI (21), but lower than OccRatio of the supraspinatus reported in asymptomatic able bodied individuals [53.5% (22), 56.4% (20)]. A lower OccRatio is seen as beneficial since less space is occupied by the tendon and more space is potentially available. In this line, higher OccRatios have been found in previous studies in persons with subacromial pain syndrome (7, 20, 22). These findings suggest that tendon thickness in relation to AHD should be considered when analyzing the risk for subacromial pain syndrome.

The present study found several subject characteristics associated to the size of the subacromial space. Participants with a shorter time since injury had a lower supraspinatus OccRatio. This indicates that with longer time in the wheelchair, and with more cumulated load on the shoulder, either supraspinatus tendon might increase or AHD decreases. Since AHD was not associated with time since injury in the present study, this change may be related to an adaption of the supraspinatus tendon over time as a response to chronic overload. This statement is supported by findings of Malanga et al. who found thicker supraspinatus tendons on the dominant side of baseball pitchers when comparing to the non-dominant side (23). Clinical practice guidelines recommend selective strengthening and stretching exercises for rotator cuff muscles in manual wheelchair users. The changes in the OccRatio further support this recommendation as such exercises may prevent pathology in the rotator cuff.

Participants with an impaired external rotation ROM had a lower OccRatio of supraspinatus and biceps, as well as a larger AHD. This finding points toward a mediating effect of the external rotators. Leong et al. (24) also reported a mediating effect of the external rotators since individuals with greater strength in external rotation presented larger AHD. Whether these findings are related or how muscular imbalance affects the subacromial space should be examined in future studies.

Finally, a lower BMI and a higher anaerobic work capacity was associated with a smaller biceps OccRatio. This supports the general recommendation that a reduced body weight and higher capacity is beneficial for the weight bearing shoulder (25). The effect of training has been evaluated in a previous study where individuals with subacromial pain syndrome participated in a rehabilitation program including strengthening of the rotator cuff and trunk muscles and endurance training. Savoie et al. found a significantly increased AHD after the rehabilitation program (13). This is a further indication that increasing wheelchair user capacity through training reduces the risk for subacromial pain syndrome.

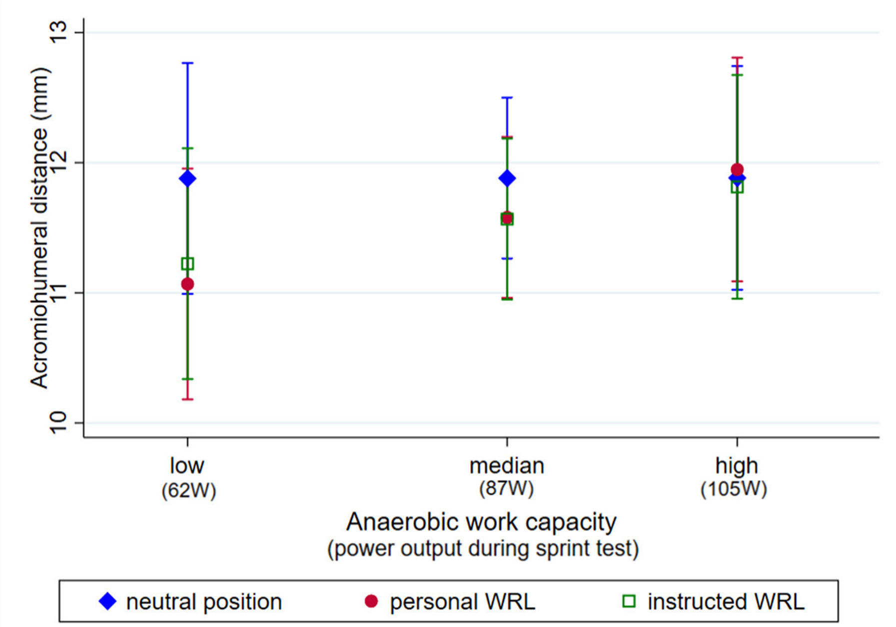


**FIGURE 3 |** Predictive margins with 95% confidence interval of acromiohumeral distance (AHD, mm) measured during different loading positions (unloaded position, personal WRL, and instructed WRL) in participants with a low (62 W, 25% percentile), median (87%) and high (105, 25% percentile) anaerobic work capacity.

**TABLE 3 |** Predictive margins with 95% confidence interval (95% CI) of acromiohumeral distance (AHD) and occupation ratio (OccRatio) of supraspinatus and biceps for categorical covariables sex, lesion level, shoulder range of motion in anteroflexion (AF), external rotation (ER) and abduction (ABD): predictive margins with 95% confidence intervals.

| | | AHD | | | OccRatio supraspinatus | | | OccRatio biceps | | |
|---|---|---|---|---|---|---|---|---|---|---|
| | | **Mean** | **95% CI** | ***P*** | **Mean** | **95% CI** | ***P*** | **Mean** | **95% CI** | ***P*** |
| Sex | Female | 10.7 | 8.9-12.5 | 0.257 | 49.3 | 40.9-57.6 | 0.626 | 34.9 | 25.0-44.8 | 0.527 |
| | Male | 11.9 | 11.2-12.7 | | 46.7 | 43.1-50.4 | | 38.8 | 34.5-43.2 | |
| Lesion | T2-T6 | 10.9 | 9.7-12.0 | 1.000[a] | 48.4 | 43.0-53.8 | 1.000[a] | 39.9 | 33.5-46.3 | 1.000[a] |
| level | T7-T12 | 11.0 | 10.0-12.1 | <0.001[b] | 49.7 | 44.7-54.6 | 0.127[b] | 39.6 | 33.7-45.4 | 0.192[b] |
| | L1-L2 | 15.2 | 13.5-16.8 | <0.001[c] | 38.1 | 30.1-46.1 | 0.057[c] | 28.8 | 19.3-38.3 | 0.192[c] |
| ROM | <170° | 11.6 | 10.9-12.3 | 0.885 | 48.1 | 44.8-51.4 | 0.365 | 36.7 | 26.0-47.3 | 0.811 |
| AF | >170° | 11.8 | 9.9-13.7 | | 43.4 | 34.4-52.3 | | 38.1 | 34.3-42.0 | |
| ROM | <170° | 11.7 | 10.7-12.6 | 0.981 | 44.9 | 40.4-49.4 | 0.174 | 38.5 | 33.6-43.5 | 0.747 |
| ABD | >170° | 11.6 | 10.8-12.5 | | 49.4 | 45.2-53.6 | | 37.2 | 32.0-42.5 | |
| ROM | <50° | 13.2 | 11.9-14.9 | 0.008 | 39.5 | 46.5-53.2 | 0.005 | 30.2 | 23.0-37.5 | 0.019 |
| ER | >50° | 11.1 | 10.4-11.8 | | 49.8 | 46.5-53.2 | | 40.5 | 36.5-44.4 | |

*Pairwise comparisons: [a]T2-T6 vs. T7-T12, [b]T2-T6 vs. L1-L2, [c]T7-T12 vs. L1-L2.*

## Effect of WRL on Subacromial Space

A temporary narrowing of the subacromial space due to high load or due to movement patterns of the shoulder structures is generally seen as a risk factor for compression of the soft tissue under the acromioclavicular arch and inflammation (26). Previous kinematic studies on the orientation of the scapula and humerus identified WRL as an activity of daily life of wheelchair users, where the risk for narrowing of the subacromial space is high (11, 27, 28). During a WRL, glenohumeral external rotation is decreased and the scapula is anteriorly tilted and internally rotated. This reduces the subacromial space, and in combination with the large superior forces at the shoulder (29), places the shoulder of the wheelchair user at high risk for compression of the structures in the subacromial space (11).

The present study found significantly decreased AHD during WRL, which indeed points to a risk for shoulder injury. Whether participants performed WRL in their own style (no instruction given) or whether they followed the instructions to ensure optimal shoulder position (depressed and retracted the shoulders) did not result in a significant difference in AHD.

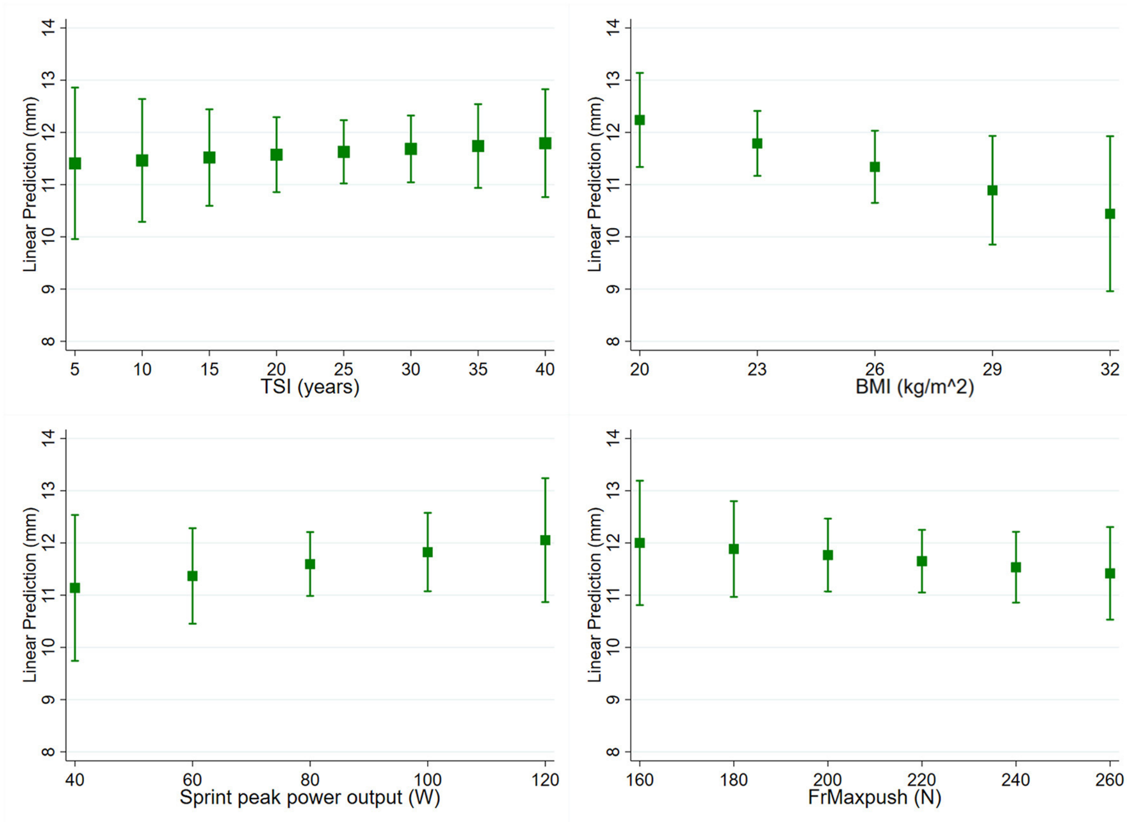


**FIGURE 4 |** Predictive margins with 95% confidence interval of acromiohumeral distance (AHD, mm) for continuous covariables time since injury [(TSI), $p = 0.727$], body mass index [(BMI), $p = 0.083$], sprint peak power output ($p = 0.941$) and resultant force reached during maximum isometric forward pushes [(FrMaxpush), $p = 0.418$].

Similar reductions of AHD were found in previous studies (26). This reduction in AHD during WRL strengthens the current notion to avoid weight relief maneuvers that place high, superiorly directed forces on the arm. Whenever possible, alternative techniques for pressure relief like forward or side leans should be used (25).

A remarkable interaction effect was found for position (unloaded vs. WRL) and anaerobic work capacity. Participants with a low anaerobic work capacity (lower power output reached during the sprint test) presented the above-mentioned reduction in AHD between unloaded position vs. WRL. Participants with a higher anaerobic work capacity, however, could maintain their AHD also during WRL. These results highlight the importance of anaerobic work capacity in shoulder function in the context of WRL. A well-planned preventive training program that safely increases wheelchair user capacity may reduce shoulder complaints (30).

To our knowledge, no study analyzed OccRatio during WRL. Mozingo et al. took however tendon thickness of infraspinatus, subscapularis and supraspinatus into account and defined risk scores based on fluoroscopy images to estimate mechanical impingement risk (5). Their results showed only minimal to no impingement risk during pressure relief lifts. Despite these findings, the authors advised wheelchair users to perform side leans for pressure relief and pressure injury prevention instead of WRL to reduce loading of the shoulder.

## Effect of Fatigue on Subacromial Space

Fatigue of the muscles stabilizing the shoulder joint may reduce the subacromial space and increase stress on the tendons within the space. There are two fatigue-based mechanisms proposed to cause narrowing of the subacromial space: superior migration of the humeral head with respect to the glenoid and alteration of the movement of the acromion with respect to the humeral head due to fatigue (31). A simulation study including empirically generated fatigue data has shown that the subacromial space was affected by fatigue and that superior humeral migration was the dominant fatigue-related mechanism associated with shoulder injury risk (31).

Our intervention study, however, did not show an effect of fatigue on either AHD nor OccRatio. Also, no interaction with wheelchair user capacity was found. This suggests that daily wheelchair propulsion, as simulated in this study by the fatiguing intervention, does not contribute to temporary changes in subacromial space. Similar findings have been reported by Lin et al. who found in general no changes in subacromial

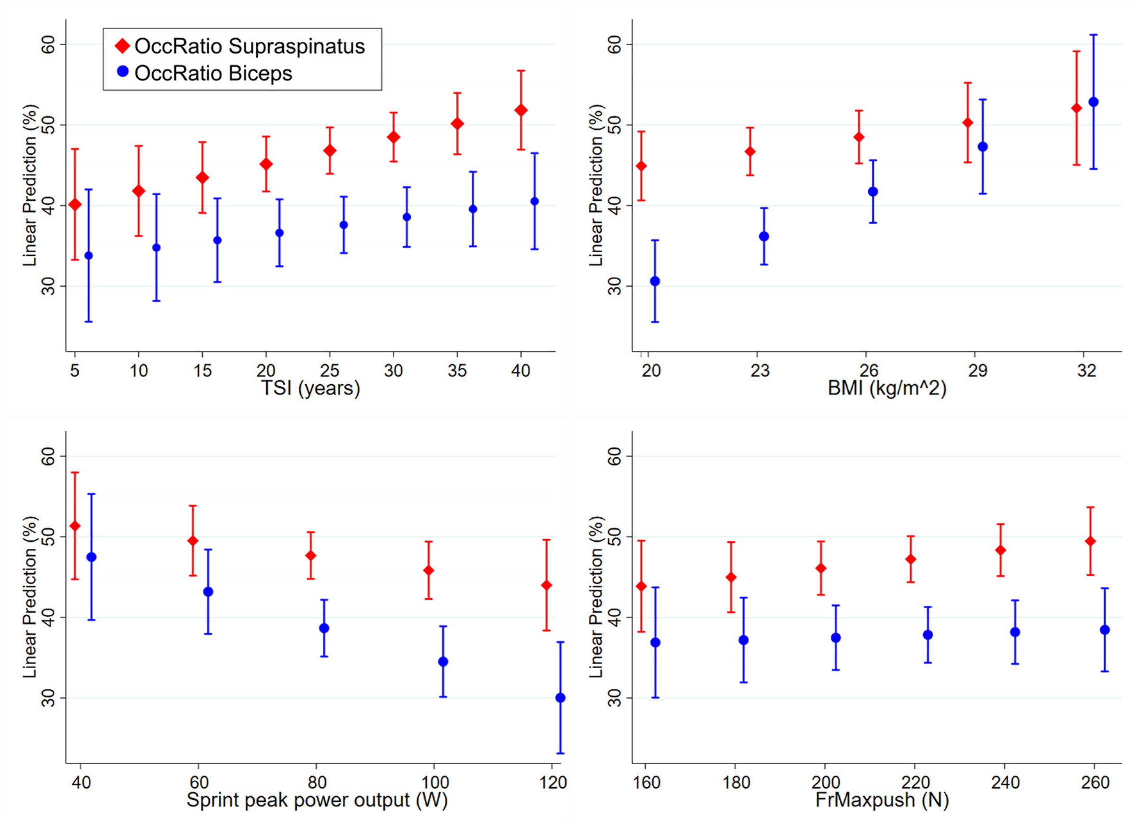


**FIGURE 5 |** Predictive margins with 95% confidence interval of occupation ratio (OccRatio, %) of supraspinatus and biceps for continuous covariables time since injury [(TSI), supraspinatus $p = 0.025$, biceps $p = 0.280$], body mass index [(BMI), supraspinatus $p = 0.144$, biceps $p < 0.001$], sprint peak power output (supraspinatus $p = 0.288$, biceps $p = 0.006$) and resultant force reached during maximum isometric forward pushes [(FrMaxpush), supraspinatus $p = 0.251$, biceps $p = 0.637$).

space after performing repetitive WRL and shoulder external rotations (26). Participants with greater levels of shoulder pain, however, showed a greater percentage narrowing of AHD. The present study excluded participants with upper-extremity pain that limits the ability to propel a wheelchair. This might be an explanation why no narrowing of the subacromial space was found.

## Study Limitations

The intervention of fatiguing wheelchair propulsion used in this study was chosen to simulate everyday load acting on the shoulder of a wheelchair user. The fatiguing protocol included maximum voluntary overground propulsion, starting, stopping and turning. Other demanding tasks for the shoulder, such as transfers, WRL and lifting heavy objects were not included. For future studies on the effect of fatigue resulting from everyday life activities, these additional tasks could be included in the fatigue protocol as long as they can be performed in a safe way. However, the used protocol is expected to be more demanding than everyday life activities since it requires maximum voluntary propulsion.

Regarding measures to quantify subacromial space it should be considered that ultrasound images only allow for two-dimensional measurements and that the measures used to calculate OccRatio in this study (AHD and tendon thickness) were taken from different ultrasound images and with different arm positions of the participants (**Figure 2**). This has been done similarly in previous studies quantifying OccRatio (7–9). Unfortunately, the arm position used to measure thickness of the supraspinatus makes it impossible to quantify this thickness and thus supraspinatus OccRatio during WRL. Since OccRatio is more informative on the available subacromial space and thus on the shoulder injury risk, the quantification of OccRatio during WRL would be an interesting venue for the future if technology and analysis software allow.

While we excluded individuals with upper-extremity pain that limits the ability to propel a wheelchair, participants may still have had pain. Future studies should look at the impact of pain on the measures collected in this study.

# CONCLUSIONS AND IMPLICATIONS

This study showed a significant reduction of the AHD during WRL compared to the unloaded position in wheelchair users with SCI. A higher anaerobic work capacity reduced the impact of WRL on AHD decrease and was thus beneficial in stabilizing the shoulder. Fatiguing wheelchair propulsion had no effect on the subacromial space. Subject characteristics related to a larger subacromial space were lower lesion level, shorter time since injury, impaired ROM in external rotation, a lower BMI and a higher anaerobic work capacity. Preventive fitness training to increase wheelchair user capacity, alternative modes for pressure relief and lowering BMI are suggested interventions to lower the risk for subacromial pain syndrome in wheelchair users with SCI. These findings may assist clinicians in designing injury prevention programs.

## DATA AVAILABILITY STATEMENT

The datasets generated and/or analyzed during the current study are available from the corresponding author on reasonable request.

## ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Ethikkommision Nordwest-und Zentralschweiz, Switzerland. The patients/participants provided their written informed consent to participate in this study. Written informed consent was obtained from the individual(s) for the publication of any potentially identifiable images or data included in this article.

## AUTHOR CONTRIBUTIONS

UA, FB, and MB initiated the study. UA, FB, MB, and AC contributed to the conception and design of the study. FB performed the data collection. UA was responsible for all analyses, drafting, and finalization of the paper. All authors critically revised the paper and have read and approved the final paper.

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# Characterizing the Thermal Demands and Mobility Performance During International Wheelchair Rugby Competition

*Erica H. Gavel[1,2]*, Melissa A. Lacroix[2], Vicky L. Goosey-Tolfrey[3] and Heather M. Logan-Sprenger[1,2,4]*

[1] Faculty of Science, Ontario Tech University, Oshawa, ON, Canada, [2] Canadian Sport Institute Ontario, Toronto, ON, Canada, [3] School of Sport, Exercise and Health Sciences, Peter Harrison Centre for Disability Sport, Loughborough University, Loughborough, United Kingdom, [4] Faculty of Health Science, Ontario Tech University, Oshawa, ON, Canada

**Objective:** To determine the thermoregulatory responses and mobility performance of wheelchair rugby (WCR) players during international competition.

**Methods:** Eleven male National Team WCR players volunteered for the study. Testing occurred during a four game series against international competition (temp 24.7 ± 0.7°C, relative humidity 50.1 ± 3.6%), with movement time (MT) and gastrointestinal temperature ($T_{gi}$) recorded continuously.

**Results:** The mean maximal $T_{gi}$ was 38.6 ± 0.6°C (37.9–39.7) and did not significantly differ among Low-Class, Mid-Class, and High-Class athletes ($p > 0.05$). Moreover, there was a strong and significant relationship between minutes (min) played per quarter of the game and change in $T_{gi}$ ($r = 0.36$, $p = 0.01$). Athletes moved a total of 27:43 ± 9:40 min:seconds (s), spent a total of 15:02 ± 8.23 min:s in Zone 1 (53.5%), 8:19 ± 3:20 min:s in Zone 2 (31.7%), and 5:59 ± 1:51 min:s in Zone 3 (21.3%). There were no differences among classification in total movement time ($p = 0.169$) or for speed in Zone 1, Zone 2, or Zone 3 ($p > 0.05$). The relationship between peak forward speed and total movement time was strong ($p = 0.021$, $r = 0.68$).

**Conclusion:** This study demonstrated that the time spent in absolute movement zones is not classification dependent, the change in core temperature is related to movement time per quarter. Furthermore, peak speeds obtained on-court were linked to overall movement time which suggests athletes should warm-up before going on court.

**Keywords: wheelchair rugby, thermoregulation, wheelchair mobility, Paralympic sport, performance**



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*Katarzyna Domaszewska, Poznan University of Physical Education, Poland*

****Correspondence:***
*Erica H. Gavel*
*erica.gavel@ontariotechu.ca*

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## INTRODUCTION

Wheelchair rugby (WCR) is an intermittent contact sport (1) which spends most time at low speeds (<50% of mean peak speed) (2, 3). In WCR, there are a total of seven classifications ranging from, 0.5 to 3.5 in order of greatest impairment to least impaired (i.e., 0.5, cervical spinal cord injury (cSCI); 3.5, quad-amputee). During an official World Wheelchair Rugby (4) game which consists of four 8-min quarters with stopped time, the total number of points on court permitted at any time is 8.0 ($n = 4$ players) (4). With that, the demands of the game and activity profiles can be dependent upon classification (5, 6), e.g., Rhodes et al. (2), clearly showed that WWR Class 0.5 athletes significantly traveled less and had lower peak speeds relative to Class 3–3.5, 881 ± 137 m vs. 1153 ± 172 m and 3.0 vs. 3.8 m/s, respectively.

While WCR is competed indoors in a temperature-controlled gymnasium (18–20°C), it has been demonstrated in athletes with a cSCI (7, 8) and without a SCI, that exercising in thermoneutral environments may induce severe heat strain (9–11). For example, work by Griggs et al. (9) reported that WCR athletes with a cSCI reached a core temperature of 39.3 ± 0.5°C, whereas athletes without a SCI reached 38.8 ± 0.3°C. Moreover, in other wheelchair sports Logan-Sprenger and McNaughton (10), noted two athletes from the Canadian Senior Women's National wheelchair basketball team without a SCI were attaining (>39.3°C) which were sustained for 10–25 mins over multiple games. As such, while level of injury can influence ones' ability to regulate body temperature, people should also be mindful that athletes without SCI's may also experience heat strain.

During exercise, metabolic energy is converted to one of two forms, mechanical energy to perform external work (20–30%), or thermal energy to produce heat (70–80%) (12). That said, if heat production exceeds heat dissipation, one will experience increases in core body temperature and decrements in physiological performance (13, 14). For example, Forsyth et al. (15) found that when the metabolic heat production was similar among athletes with cSCI, paraplegia, and non-SCI, those with cSCI displayed the greatest compromised sweat response. Moreover, similar outcomes were observed in work by Griggs et al. (9) where the level of injury was correlated with changes in core temperature and heat storage.

Although sport practitioners and coaches should be mindful that the level of SCI can be correlated to heat storage and increases in core temperature, one should also consider the influence overall playing time has on heat storage and potential detriments in performance. For example, although the work by Griggs et al. (16) demonstrated that athletes with a cSCI reported higher core temperatures than non-SCI athletes, this work was conducted in a practice setting where the athletes played the entire simulated game. Extending this work but within wheelchair basketball, Logan-Sprenger and McNaughton (10) gathered data during international competitive game play to which, the will-to-win was high, providing high ecological validity. As such, the intensity of play along with coach substitutions influencing playing time may be an important difference between a simulated game and a vital game for international ranking. As of late, no study has characterized mobility performance and examined the influence these demands may have on core body temperature, thermoperception, and peak speed during international WCR competition. Thus, the purpose of this study was to (1) characterize mobility performance, (2) describe the thermoregulatory responses, and (3) evaluate the physiological and thermoperception of international level WCR players during a World Wheelchair Rugby event against an international Top 10 ranking competitor.

# METHODS

## Subjects

Eleven ($n = 11$; cSCI = 10, quad amputee = 1) elite WCR players volunteered to participate in the study (see **Table 1**). The research team tested 3–4 athletes per game with each athlete being monitored once. As such, data collection occurred over four separate games to ensure all 11 athletes were tested. All participants were male and members of the Canadian National WCR Team. Typically, the team had three athletes with a cSCI, and one quad amputee on the floor at the same time. Each player had played at the international level for 10.5 ± 6.5 yrs and trained an average of 10–15 h per week. Participants were informed of the experimental protocol before written informed consent was obtained. All procedures were approved by the Ontario Tech University Ethics Committee (file #16414) and conformed to the principles defined in the Declaration of Helsinki.

## Design

This was an observational study assessing wheelchair mobility performance, gastrointestinal temperature ($T_{gi}$), heart rate (HR), ratings of perceived exertion (RPE, Borg 6–20), thermal sensation (TS), and thermal comfort (TC) during a four game series against international competition at the 2019 Japan World Wheelchair Rugby Challenge.

## Methodology

Upon waking on game day athletes were asked to provide a mid-stream urine sample for measurement of urine specific gravity (USG) (17) using a handheld refractometer (Atago–PEN-PRO, Geneq, Montreal QC) (17). Additionally, athletes were instructed to swallow an ingestible thermistor (e-Celcius; Bodycap; Herouville Saint-Clair, France) a minimum of 6 h before the game to measure gastrointestinal temperature (18, 19).

Athletes arrived at the gymnasium at their usual pregame time (∼2 h prior to the game); 1 game took place at 12:00 (noon) while the other 3 games started at 18:00. The athlete's water bottle(s) was labeled with the athletes' name and weighed prior to and upon completion of the game to determine total fluid intake throughout the game. Three athletes had a slushie drink following warm-up and before the start of the game and used a water spray during the match (Low-Class, $n = 1$; Mid-Class, $n = 2$ ). $T_{gi}$ was continuously measured and recorded throughout the game from the start of warm-up until the end of the game and downloaded intermittently. HR was collected throughout using a downloadable Polar® OH1 heart sensor (Polar, CAN). Each athlete's chair was equipped with an inertial measurement unit (IMU) (Shimmer, Cambridge MA) positioned on the axels and center of chair frame (20) to measure peak speed, real-time speed, and time spent in different speed zones which was recorded throughout the game. Speed zones were categorized as: Zone 1 = 0–1 m/s, Zone 2 = 1–2 m/s, Zone 3 => 2 m/s. Moreover, playing time and movement time were also recorded. Playing time was characterized as time defined by the "game clock" without the inclusion of stoppages, whereas movement time was the total time the athlete was on-court. For classification and positional analysis, athletes were grouped as Low-Class to

**TABLE 1 |** Descriptive characteristics of the male participants (mean ± SD).

| | IWRF class | Level of injury (LOI) | Completeness of lesion | Pre-USG | Total PT (mins:secs) | Δ Tc (°C) | Mean Tc (°C) | Peak Tc (°C) | Mean HR (bpm) | Peak HR (bpm) |
|---|---|---|---|---|---|---|---|---|---|---|
| **SCI** | | | | | | | | | | |
| | 0.5 | C6 | Complete | 1.001 | 17:38 | 0.6 | 38.5 | 38.9 | 83 | 91 |
| | 1.0 | C5 | Complete | 1.008 | 19:27 | 1.4 | 38.7 | 39.0 | 106 | 124 |
| | 1.0 | C6 | Incomplete | 1.010 | 9:30 | 1.4 | 37.5 | 37.9 | 89 | 115 |
| | 1.5 | C7 | Complete | 1.010 | 5:46 | 1.4 | 37.9 | 38.3 | 96 | 130 |
| | 2.0 | C6 | Incomplete | 1.006 | 7:42 | 1.1 | 37.7 | 38.1 | 86 | 119 |
| | 2.0 | C6 | Complete | 1.013 | 10:42 | 1.4 | 38.2 | 38.6 | 95 | 110 |
| | 2.0 | C6 | Incomplete | 1.005 | 11:50 | 2.3 | 38.8 | 39.7 | 110 | 135 |
| | 2.0 | C6 | Incomplete | 1.006 | 12:21 | 0.9 | 38.0 | 38.2 | 94 | 104 |
| | 3.0 | C6 | Incomplete | 1.017 | 18:09 | 0.6 | 38.0 | 38.3 | 120 | 156 |
| | 3.0 | C7 | Complete | 1.014 | 10:59 | 0.9 | 37.7 | 38.1 | 98 | 116 |
| | 3.5 | QA | NA | 1.013 | 23:25 | 0.7 | 38.5 | 38.8 | 134 | 176 |
| ***Mean*** | | | | 1.010 | 12:38 | 1.2 | 38.2 | 38.6 | 99 | 123 |
| ***SD*** | | | | 0.003 | 6:40 | 0.5 | 0.5 | 0.7 | 10 | 16 |

*LOI, level of injury; QA, quad-amputee; BM, body mass; USG, urine specific gravity; PT, playing time;* **Δ** *Tc, change in core temperature from start to end of game; Mean Tc, core temperature from start to end of game; HR, heart rate.*

**TABLE 2 |** WC kinematics throughout the WC game.

| | IWRF class | Injury | Peak speed (m/s) | Movement time (mins) | Total distance (m) | Zone 1 (m) | Zone 2 (m) | Zone 3 (m) |
|---|---|---|---|---|---|---|---|---|
| | 0.5 | C6 | 2.59 | 26.1 | 3177.2 | 652.1 (19.8%) | 1713.4 (51.8%) | 0 (0%) |
| | 1.0 | C5 | 3.4 | 43.0 | 2484.9 | 423.9 (17.1%) | 1139.7 (45.9%) | 887.4 (4.0%) |
| | 1.0 | C6 | 3.6 | 37.4 | 1249.6 | 1079.7 (86.4%) | 153.4 (12.3%) | 16.6 (1.3%) |
| | 1.5 | C7 | 3.5 | 22.0 | 1449.0 | 212.3 (14.7%) | 653.4 (45.1%) | 548.6 (37.9%) |
| | 2.0 | C6 | 4.5 | 34.7 | 1905.8 | 377.3 (19.8%) | 992.4 (52.1%) | 534.8 (28.1%) |
| | 2.0 | C6 | 3.1 | 27.3 | 1541.2 | 330.6 (21.5%) | 771.8 (50.1%) | 407.3 (26.4%) |
| | 2.0 | C6 | 2.8 | 26.4 | 1884.5 | 266.4 (14.4%) | 759.5 (40.3%) | 774.0 (41.1%) |
| | 2.0 | C6 | 4.1 | 43.4 | 1316.2 | 212.0 (16.1%) | 653.0 (49.6%) | 411.3 (31.2%) |
| | 3.0 | C7 | 3.4 | 15.6 | 1055.6 | 180.1 (17.1%) | 543.8 (51.5%) | 320.2 (30.3%) |
| | 3.0 | C6 | 4.0 | 21.4 | 795.4 | 264.4 (18.2%) | 715.5 (49.4%) | 469.9 (32.4%) |
| | 3.5 | QA | 5.3 | 37.5 | 3436.8 | 322.9 (9.4%) | 1144.0 (33.3%) | 1379.6 (40.1%) |
| ***Mean*** | | | 3.8 | 27.4 | 1857.0 | 384.4 (23.4%) | 814.7 (44.3%) | 466.3 (25.6%) |
| ***SD*** | | | 0.77 | 9.4 | 877.1 | 269.1 (21.2%) | 419.9 (12.3%) | 383.8 (13.9%) |

*Zone 1, 0-1 m/s; Zone 2, 1-2 m/s; Zone 3, > 2 m/s; m, meters; s, seconds; %, percent of total distance spent in each velocity zone.*

(0.5–1.5), Mid-Class (2.0–2.5), and High-Class (3.0–3.5) (4). RPE (21), TS (22), and TC (22) were collected upon substitution or at the completion of each quarter.

## Statistical Analysis

All data was tested for normality of distribution and displayed as the mean and standard deviation. Differences between quarters were analyzed using a one-way ANOVA, and time verses group was tested using a mixed-ANOVA, to detect singular differences, a Tukey's honestly significant difference (HSD) *post-hoc* was performed. A Student's paired *t-test* was used to compare singular parameter differences where appropriate. Categorical data was tested using the Krusal-Wallis test, to detect singular differences a Dunn's *post-hoc* test was performed. A Wilcoxon Signed Ranks Test was detected to test differences in singular categorical data. Statistical significance was accepted at $p < 0.05$. Correlations between variables were assessed using a Pearson's correlation analysis. Exact $p$-values, Cohen's D, and 95% confidence intervals are presented to show magnitude of effect. The magnitude of effect was classed as trivial (<0.2), small (0.2-0) moderate (0.6–1.2), large (1.2–2.0), and very large (≥2.0) (23).

## RESULTS

### Ambient Conditions

The gymnasium temperature (°C) and relative humidity (RH) (%) was similar between games and remained stable within games (pre 24.6 ± 0.8°C, RH 51.4 ± 2.4%; post, 24.4 ± 0.1°C, RH 50.4 ± 3.0%, $p$>0.05).

### Morning Urine Specific Gravity and Fluid Intake

All athletes were hydrated on game day (**Table 1**) with a mean USG of 1.010 ± 0.004 (1.005–1.015). On average, athletes consumed 1073 ± 840 milliliters (ml) (112–1,669 ml) of fluid per game.

### Game Playing Time

There was large variability in playing time among athletes (2:58–23:25 min:s). On average, athletes played a total of 14:04 ± 5:49 min:s (**Table 1**). There were no differences in playing time among Low-Class, Mid-Class, and High-Class players ($p$ = 0.169), and three athletes played >50% (>16 min) of the game (Low-Class: 19:27; High-Class: 18:09 min:s; High-Class: 23:25 min:s).

### Movement Time and Time Spent in Velocity Zones

Excluding warm-up, athletes moved a total of 27:43 ± 9:40 min:s (12:10–43:10) over the course of the game (**Table 2**). On average, athletes spent a total of 15:02 ± 8.23 min:s in Zone 1 (53.5%), 8:19 ± 3:20 min:s in Zone 2 (31.7%), and 5:59 ± 1:51 min:s in Zone 3 (21.3%) (**Figure 1**). Movement time among Low-Class, Mid-Class, and High-Class did not significantly differ ($p$ = 0.169) with athletes playing a mean of 31:27 ± 9:27, 25:48 ± 10:22, 25:18 ± 11:27 min:s, respectively.

There was no difference among classification in total movement time ($p$ = 0.169) or for Zone 1 ($p$ = 0.601), Zone 2 ($p$ = 0.173), or Zone 3 ($p$ = 0.222) (**Figure 2**). On average, time spent in Zone 1 was significantly greater than Zone 2 ($p$ = 0.0154, 95% CI 74.4 to 757.4, ES = 1.09) and Zone 3 ($p$ = 0.0002, 95% CI 298.7 to 981.4, ES = 1.53), whereas Zone 2 and Zone 3 did not significantly differ ($p$ = 0.253, 95% CI −117.2 to 565.9, ES = 0.87).

There was a significant difference in cumulative distance between Zone 1 and Zone 2 (Zone 1, 384.4 ± 269.1 vs. Zone 2, 814.7 ± 419.9 m, $p$ = 0.025, 95% CI −812.2 to −48.3, ES =1.22), while Zone 1 vs. Zone 3 (Zone 1, 384.4 ± 269.1 vs. Zone 3, 466.3 ± 383.8 m, $p$ = 0.858, 95% CI −463.8 to 300.1, ES = 0.25) and Zone 2 vs. Zone 3 did not differ (Zone 2, 814.7 ± 419.9 vs. Zone 3, 466.3 ± 383.8 m, $p$ = 0.079, 95% CI −33.55 to 730.44, ES = 0.87) (**Table 2**).

Peak speed in the game did not significantly differ between classification (Low-Class, 3.24 ± 0.45 m/s; Mid-Class, 4.44 ± 1.14 m/s; High-Class, 4.22 ± 0.94 m/s; $p$ = 0.373). The mean peak forward speed throughout games was 3.58 ± 0.9 m/s. There were no significant differences in peak speed between quarter 1, quarter 2, quarter 3, and quarter 4 ($p$ = 0.338). Furthermore, there were no significant differences in time spent in Zone 1, Zone 2, and Zone 3 between quarters ($p$ > 0.05).

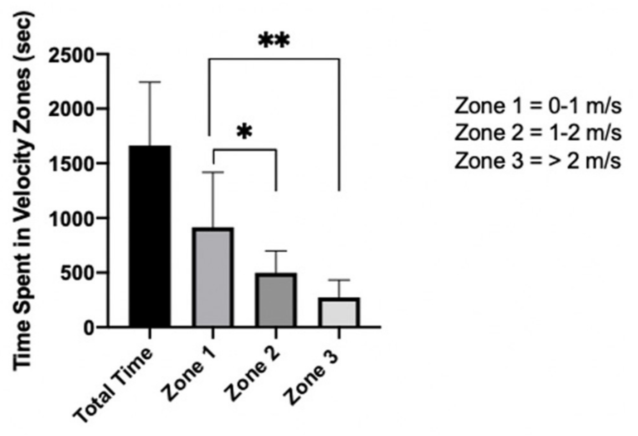


**FIGURE 1 |** Mean movement time [seconds (sec)] spent in each velocity zone (zone 1–3) throughout a wheelchair rugby game. (*) denotes significance between Zone 1 and Zone 2, whereas (**) denotes significance between Zone 1 and Zone 3.

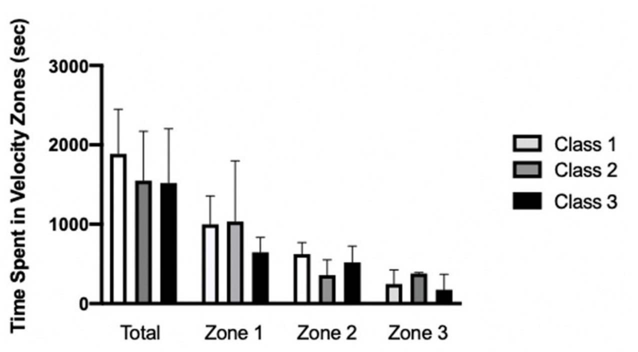


**FIGURE 2 |** Mean time spent in different velocity zones (Zones 1–3) during a wheelchair rugby game and movement time for Class 1 ($n$ = 4), 2 ($n$ = 4), and 3 ($n$ = 3). $p$ > 0.05.

### Thermoregulatory Responses Across the Game

Mean $T_{gi}$ was significantly greater from the start (following warm-up) to end of game (start, 37.9 ± 0.3°C vs. finish, 38.6 ± 0.7°C, $p$ = 0.009). Mean $T_{gi}$ significantly differed between quarter 1, quarter 2, quarter 3, and quarter 4 ($p$ < 0.05). The mean rise of $T_{gi}$ was 1.2 ± 0.5°C, and there was no significant rise in $T_{gi}$ between quarter 1, quarter 2, quarter 3, and quarter 4 ($p$ = 0.766). The mean maximal $T_{gi}$ was 38.6 ± 0.6°C (37.9–39.7). $T_{gi}$ did not significantly differ among Low-Class, Mid-Class, and High-Class athletes throughout the game ($p$ > 0.05,

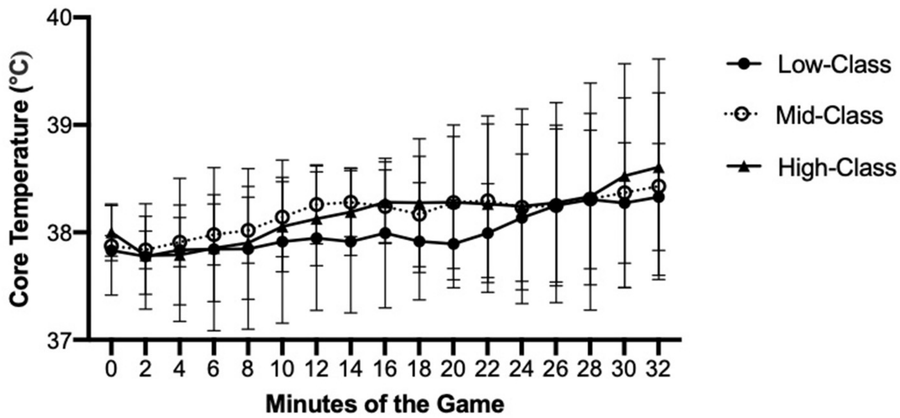


**FIGURE 3 |** Mean gastrointestinal temperature among Low-Class, Mid-Class, and High-Class athletes across a WCR game.

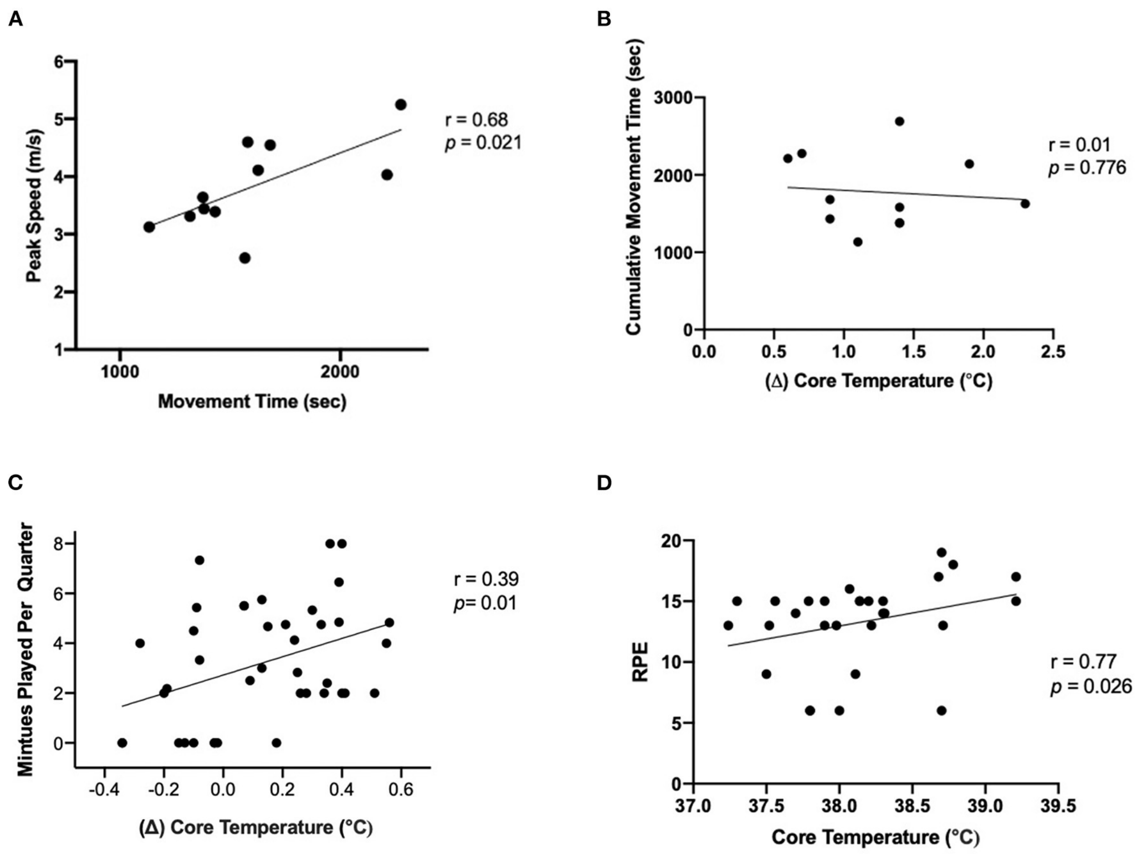


**FIGURE 4 | (A)** correlation between peak speed (m/sec) and movement time, **(B)** correlation of mean (Δ) of gastrointestinal temperature and cumulative movement time (sec), **(C)** correlation of mean change (Δ) of gastrointestinal temperature (°C) and playing time per Q (mins), and **(D)** correlation of gastrointestinal temperature (°C) and RPE (6–20).

**Figure 3**). Furthermore, there were no significant differences between athletes with complete and incomplete SCIs ($p > 0.05$). Two of the 11 athletes (Low-Class; Mid-Class reached a $T_{gi}$ of >39°C and at the start of quarter 4 sustained it for 14:02 ± 1:06 min:s (13:15; 14:48) with a mean movement time of 30:10 ± 4:08 min:s (27:15; 33:05) and playing time of 14:00 ± 2:59 min:s. (19:27; 11:50).

## Thermal Sensation, Thermal Comfort, Ratings of Perceived Exertion

The median TS for the game was 6 (1). TS did not significantly differ from the start to end of game [start, 6 (2) vs. end, 5 (2), $p = 0.930$]. The median TS did not significantly differ between quarters ($p = 0.210$) or between classifications ($p = 0.178$). As well, median TC was 2 (1). TC did not significantly differ from start to end of game [start, 2 (2) vs. end, 3 (2), $p = 0.586$]. TC did not significantly differ between quarters ($p = 0.750$) or among classification ($p = 0.359$). The median RPE was 15 (2). RPE significantly increased from the start to end of game [start, 13.0 (8) vs. end, 15.0 (2.5), $p = 0.023$]. The median RPE did not significantly differ between quarters ($p = 0.875$) or between classification ($p = 0.951$).

## Correlations

There was a significant relationship between total movement time and time spent in Zone 1 ($p = 0.002$, $r = 0.85$); however, the relationship between total movement time and Zone 2 ($p = 0.315$, $r = 0.35$), and Zone 3 ($p = 0.51$, $r = 0.24$) was not strong. Moreover, the relationship between peak forward speed and total movement time per quarter was strong ($p = 0.021$, $r = 0.68$, **Figure 4A**).

There was not a significant relationship between total playing time and change in $T_{gi}$ ($p = 0.641$, $r = 0.41$), change in $T_{gi}$ and time spent in Zone 1 ($p = 0.236$, $r = 0.19$), Zone 2 ($p = 0.112$, $r = 0.26$), and Zone 3 ($p = 0.134$, $r = 0.24$), and peak $T_{gi}$ and total movement time ($p = 0.245$, $r = 0.438$, **Figure 4B**). However, there was a strong relationship between movement time and change in $T_{gi}$ per quarter ($p = 0.014$, $r = 0.39$, **Figure 4C**). There was a strong relationship between $T_{gi}$ and RPE ($p = 0.026$, $r = 0.77$, **Figure 4D)** while the relationship between $T_{gi}$ and TC ($p = 0.341$, $r = -0.15$) and TS ($p = 0.61$, $r = -0.08$) was not significant.

# DISCUSSION

This is the first study in WCR to demonstrate that differences in core temperature may not be related to classification or physiological function, but with movement time per quarter. Furthermore, the results of this study also exhibited, (1) the mean movement time was $28.3 \pm 8.5$ min and did not significantly differ among classification, (2) athletes spent most of their time at low speeds (Zone 1= 0–1 m/s) during the game, (3) there was a positive relationship between movement time and peak speed ($r = 0.68$, $p = 0.021$), (4) there were no significant differences among core body temperature and classification, and (5) TS and TC did not change over the course of the game.

## Wheelchair Mobility Profiles

The present study supported previous findings suggesting that athletes spend most time in low intensity zones; 53.5% in Zone 1, 31.7% in Zone 2, and 21.3% in Zone 3 (1, 11, 12). That said, contrary to Rhodes et al. (1), we did not display any significant differences between classification groups, which may be due to a smaller sample size and competitor discrepancies (24) (i.e., Zonal Championships vs. World Wheelchair Rugby Challenge). Across the game, we found no differences in mean peak speed between quarters which was also noted by Rhodes et al. (1). Interestingly, there was a strong correlation between movement time and peak speed, which warrants further investigation suggesting that warming-up courtside prior to being substituted into the game should be encouraged allowing athletes to achieve high wheelchair speeds when they are not the starting players. This is a topic of recent study (24) and needs to be considered alongside individual thermoregulatory responses.

## Thermoregulatory Responses

Notably, the present study demonstrated a positive relationship between playing time per quarter and change in core temperature with no differences in core temperature between classification, which aligns with wheelchair basketball work done by Logan-Sprenger and McNaughton (10). In the study, the researchers demonstrated no core temperature difference between classification, instead they observed that the change in core temperature was dependant on minutes played, which is like the present study showcasing that the change in core temperature was dependent on minutes moved per quarter. Although, it is well established that sudomotor activity plays a pertinent role in thermoregulation (25), both the present study and the one by Logan-Sprenger and McNaughton (10) demonstrate that in team sports such as WC basketball and rugby, the increase in core temperature may be more dependent on playing time rather than muscular function and the inability to dissipate heat; however, more research is needed to better understand the relationship (26).

While the present study saw ∼1.2°C change of core temperature over the course of the game, there were no differences among classification and only two athletes who reached >39°C and sustained it for ∼14 min. This data, however, contrasts with work published by Griggs et al. (9) who demonstrated significant differences between classification. In the study by Griggs et al. (9), the authors reported that lower class players displayed an increase of 1.6°C and upper-class players showed an increase of 0.7°C. In contrast to our methodology, Griggs et al. (9) collected data during an intra-squad game where the athletes played the entire game under simulated game conditions. Given that the current study was against an international competitor at a World Wheelchair Rugby event, there was high variability in playing time (CV = 36%) as the team was playing to win. As such, it is hard to compare $T_{gi}$ responses between studies given that playing time and competition level differed (27).

## Thermo-Perception

The current study demonstrated no correlation between core temperature and thermal sensation or comfort. Furthermore, there were no differences in thermal sensation or comfort from the start to end of the game, between quarters, or among classification groups, which is in line with work by Webborn et al. (28) and Griggs et al. (25). In the study by Webborn et al. (28), participants completed an intermittent sprint protocol with pre-cooling, per-cooling, and no cooling. In the study, there was no significant relationship between core temperature changes and thermal sensation. Furthermore, similar results were exhibited by Griggs et al. (25) who analyzed the influence relative humidity had on heat storage for athletes with a SCI. In the study, there was no significant relationship between increases in core temperature and changes in thermal perception.

Thermo-physiological models have been developed to assess the influence of environmental conditions on the human body (29, 30); however, no model has been validated with the inclusion of exercise in athletes with a SCI or an impairment. For example, work by Flouris and Cheung (31) has demonstrated that thermoperception can vary for the same stimulus of skin and core body temperature during exercise in a hot environment. Furthermore, research by Nicotra and Ellaway (32) suggests that the level and completeness of the SCI can influence heat and cold thresholds for a similar stimulus, whereas work by Griggs et al. (25) reported no difference between paraplegic and tetraplegic in similar environmental conditions. Moreover, work by Webborn et al. (28) demonstrated that thermal sensation did not correlate to changes in core temperature or total time exercising.

## LIMITATIONS

Although all athletes but one had a SCI, it's worth highlighting that we selected to include a player with quadruple amputation in the data analyses. Close inspection of our findings indicates that removal of the amputee would not impact the results too much as the athletes were grouped on a positional basis verses physiological function. Furthermore, their thermoregulatory responses were within 1 SD of the group responses. Secondly, given that the data was collected over a series of four separate games, one could suggest that certain games could be lower intensity than others. However, we observed no difference in RPE from one game to the next.

## CONCLUSION

To our knowledge, no study has evaluated the relationship between activity profiles, thermoregulatory responses, and thermo-perception during international WCR match play. First off, this study demonstrated that the time spent in absolute movement zones is not classification dependent, the change in core temperature is related to movement time per quarter. Secondly, peak speeds obtained on-court were linked to overall movement time which suggests a mid-event warm-up could be beneficial. Finally, like previous work, thermo-perception models should be taken with caution when working with athletes with a SCI.

## DATA AVAILABILITY STATEMENT

The original contributions presented in the study are included in the article/supplementary materials, further inquiries can be directed to the corresponding author.

## ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Ontario Tech University Research Ethics Board. The patients/participants provided their written informed consent to participate in this study.

## AUTHOR CONTRIBUTIONS

All authors provided substantial contributions to the conception, study design, and the drafting of the work, or revising it critically. Final approval of the version submitted/published and consent for publication has been agreed by all authors.

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SYSTEMATIC REVIEW
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# How Was Studied the Effect of Manual Wheelchair Configuration on Propulsion Biomechanics: A Systematic Review on Methodologies

*Capucine Fritsch[1,2]*, Yoann Poulet[1], Joseph Bascou[1], Patricia Thoreux[3,4] and Christophe Sauret[1,2]*

[1] Centre d'Études et de Recherche sur l'Appareillage des Handicapés, Institution Nationale des Invalides, Paris, France, [2] Arts et Métiers Institute of Technology, Université Sorbonne Paris Nord, IBHGC – Institut de Biomécanique Humane Georges Charpak, HESAM Université, 151 Bd de l'Hôpital, Paris, France, [3] Hôpital Hôtel-Dieu, AP-HP, Paris, France, [4] Université Sorbonne Paris Nord, Arts et Métiers Institute of Technology, IBHGC – Institut de Biomécanique Humane Georges Charpak, HESAM Université, 151 Bd de l'Hôpital, Paris, France

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Mediterranean Neurological Institute Neuromed (IRCCS), Italy

***Correspondence:**
Capucine Fritsch
capucine.fritsch@invalides.fr

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**Background:** For both sports and everyday use, finding the optimal manual wheelchair (MWC) configuration can improve a user's propulsion biomechanics. Many studies have already investigated the effect of changes in MWC configuration but comparing their results is challenging due to the differences in experimental methodologies between articles.

**Purpose:** The present systematic review aims at offering an in-depth analysis of the methodologies used to study the impact of MWC configuration on propulsion biomechanics, and ultimately providing the community with recommendations for future research.

**Methods:** The reviewing process followed the Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) flowchart on two databases (Scopus and PubMed) in March 2022.

**Results:** Forty-five articles were included, and the results highlighted the multiplicity of methodologies regarding different experimental aspects, including propulsion environment, experimental task, or measurement systems, for example. More importantly, descriptions of MWC configurations and their modifications differed significantly between studies and led to a lack of critical information in many cases.

**Discussion:** Studying the effect of MWC configuration on propulsion requires recommendations that must be clarified: (1) the formalism chosen to describe MWC configuration (absolute or relative) should be consistent with the type of study conducted and should be documented enough to allow for switching to the other formalism; (2) the tested MWC characteristics and initial configuration, allowing the reproduction or comparison in future studies, should be properly reported; (3) the bias induced by the experimental situation on the measured data must be considered when drawing conclusions and therefore experimental conditions such as propulsion speed or the effect of the instrumentation should be reported.

**Conclusion:** Overall, future studies will need standardization to be able to follow the listed recommendations, both to describe MWC configuration and mechanical properties in a clear way and to choose the experimental conditions best suited to their objectives.

**Keywords: manual wheelchair, configuration, settings, methodology, experiment, kinematics, kinetics, PRISMA**

## INTRODUCTION

Manual wheelchairs (MWC) allow disabled people with impaired lower limb function to regain autonomy and physical mobility. However, if the MWC is not properly adapted to its user, propulsion can become exhausting, and could favor the appearance of musculoskeletal disorders either through an increase in shoulder net joint moment (1), a decrease in mechanical efficiency (2), or through ranges of motion closer to articular limits (3). Therefore, providing the user with an optimally fitted MWC is crucial and requires in-depth studies of the effect of MWC characteristics on propulsion. Among the different characteristics of a MWC, it is possible to distinguish dimensional (e.g., seat and backrest width and depth) and positional (e.g., camber, backrest, and seat angles) characteristics defining the MWC configuration (i.e., its geometry); and the resulting mechanical properties (e.g., mass, position of the center of mass (CoM), or rolling resistance, etc.). Sometimes, the literature also refers to the word "settings" which is used in the present article as the selected value for a given dimensional or positional characteristic. Besides, various scientific approaches can be implemented to study the effect of MWC characteristics on propulsion, such as physiology, biomechanics, and even human and social sciences. Among the different approaches, biomechanics (i.e., kinematics and kinetics) is particularly well-suited because it relies on physical quantities and measurement systems allowing to obtain instantaneous values, unlike physiological measurements (4).

Tackling the issue of identifying the optimal MWC configuration is challenging because the numerous characteristics involved and the multiple tasks that constitute MWC propulsion (e.g., slope, cross-slope, turning, curbs etc.) result in too many conditions to be tested by a single subject. Hence, researchers tend to isolate a single MWC characteristic in their studies and to focus on one task, generally straight-line propulsion on flat ground. To date, numerous articles have already attempted to quantify the effect of a MWC characteristic on propulsion biomechanics, with a growing interest over the past 20 years. This has led to some authors providing literature reviews with special emphasis on daily (5) or sport displacements (6–8). If some trends could be drawn, contradictory results were also obtained, which could be attributed to differences in methodologies and lack of standardization (7). Indeed, comparing the results of various studies requires dealing with similar experimental conditions (i.e., power output, speed, or metabolic power) (9); and similar MWC configurations to ensure the results portability. Also, some reserves could be expressed on the different studies due to the difficulty in isolating a change in a single setting (10) and by the effective control of the power output across the different tested conditions (11), in particular due to changes in rolling resistance (12, 13), and also due to the alteration of MWC stability (14–16).

Given the diversity of methodologies highlighted by previous authors, the purpose of this systematic review is to identify and report the multiplicity of methodologies used by the literature while studying the effect of MWC configuration on propulsion biomechanics, both for sports and everyday uses, with particular emphasis on experimental task, experimental environment, propulsion speed, MWC configuration reporting, number of configuration under study, measurement systems, MWC used, and participants. Lastly, the discussion strives to provide readers with guidance regarding experiments to be performed, literature analysis, and suggestions for future works.

## METHODS

The present review aimed at identifying and analyzing the methodologies used in studies that dealt with MWC configurations and their impact on propulsion biomechanics. The review was conducted following the Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) 2020 updated guidance (17). PubMed and Scopus databases were individually searched for relevant articles, regardless of their publication date. The request, initially emitted on December 2020 and updated on March 2022 to add the articles that were published since, focused on retrieving all the articles considering the impact of MWC configuration on propulsion biomechanics and was worded as follows:

*(wheelchair[Title]) AND ((setting*[Title/Abstract]) OR (configuration*[Title/Abstract]) OR (design[Title]) OR (propert*[Title/Abstract]) OR (characteri*[Title/Abstract]) OR (seat[Title/Abstract]) OR (backrest[Title/Abstract]) OR (camber[Title/Abstract]) OR (wheel[Title/Abstract]) OR (pushrim[Title/Abstract]) OR (handrim[Title/Abstract]) OR (footrest[Title/Abstract]) OR (fork[Title/Abstract]) OR (caster[Title/Abstract]) OR (interface[Title/Abstract]) OR (gear[Title/Abstract]) OR (profile[Title/Abstract]) OR (form[Title/Abstract]) OR (tube[Title/Abstract])) NOT (electric*[Title]).*

Two authors took part in the screening process (Y. P., C. F.), following the PRISMA flowchart and independently managing half of the records. After duplicate removal, the remaining articles were first screened by title, then by abstract, and finally by full text. Inclusion criterion was studies covering the effect of MWC configuration on propulsion biomechanics from an experimental point of view. In contrast, exclusion criteria were the following: articles involving electric or power assisted wheelchairs; articles about sit-to-stand, reclining, stair-climbing and children-sized wheelchairs; articles involving other propulsion system than manual handrim or no propulsion at all; articles only studying physiological parameters; and articles that were not original studies or not written in English. When in doubt, records were identified and kept in a separate list so that the two authors could reach an agreement. Ultimately, 45 articles were retained and were sorted in main categories according to the characteristics they focused on. For the analysis, the same two authors collected the methodology described in each article, with special attention given to the description of the wheelchair configuration, the subjects' wheelchair experience, and the experimental tasks and devices.

The PRISMA Checklist is appended to this article (**Supplementary Material**).

# RESULTS

The compilation on both databases resulted in 3,698 references. After duplicate removal, 2,775 references remained. The screening through the title filter resulted in 160 references. After reading the abstracts, 68 articles were selected, and finally, 44 articles were considered after the full texts were read. An additional relevant article, not identified through the screening process but found in the bibliography of another article, was included in this review (6). This approach is summarized in **Figure 1**.

The table which led to the redaction of the results section is appended to this article (**Supplementary Material**).

## General Results

Over the 45 articles remaining from the screening process, 24 articles recruited exclusively MWC users, 17 articles exclusively able-bodied (AB) volunteers, and four articles included both MWC users and AB subjects (**Table 1**).

The results were organized per characteristics and gathered into three major parts:

- the characteristics related to the wheels ($n = 17$)
- the characteristics related to seating ($n = 9$)
- the characteristics describing the vertical and horizontal positions of the seat with respect to the rear wheels ($n =2\ 0$)

In total, three articles studied characteristics included in two of the sections listed above.

## Wheel-Related Articles

Seventeen articles studied wheel-related characteristics focusing either on the rear wheel camber angle ($n = 8$), on the rear wheel ($n = 3$) or handrim ($n = 3$) diameters, on handrim shapes ($n = 2$) or on tire pressure and type ($n = 1$).

### Camber

Camber is by far the most studied wheel-related characteristic. The 8 articles focusing on camber angle can be organized following their experimental conditions: overground ($n = 3$), on a roller ergometer ($n = 3$) or on a treadmill ($n = 2$).

#### *Overground*

**Methods:** Three studies investigated the effect of camber angle through overground experiments, involving novice AB subjects (38, 39), or highly trained MWC basketball and tennis athletes (52). As one article is sports oriented and the others focus on daily life displacements, the range of studied camber angles varied noticeably between studies: from 15° to 24° (52), and from 0° to 15° (38, 39). On average, 3 different camber angles (minimum 2, maximum 4) were tested per study. Experiments were carried out using the same MWC without adjustment to the participant (38, 39) or using the same MWC but with seat height adjusted to the athlete's personal MWC by copying their elbow angle when hands were at the handrim top dead center (52).

Wheelchair configurations were described with varying levels of detail among the articles. MWC brand and model were always provided, along with rear wheel, caster and handrim diameters, or seat width, depth, and height.

All the articles specified taking care of preventing "toe-in toe-out" (i.e., alignment of the wheels in the transverse plane) for each camber angle condition. The impact of varying camber on the MWC configuration was controlled and standardized by maintaining the top-to-top rear wheel width constant either at 48 cm (52), or at 40% of the user's arm span (38).

Experiments consisted of a straight-line propulsion over 4-meters long displacements (with 3–4 propulsion cycles before and after the measurement area) at 1 m/s (controlled by measuring the time to complete the 4 meters) (38, 39), and of a combination of a 20 m sprint, linear mobility, and maneuverability drills at maximal speed (52).

**Materials:** Experiments were monitored with motion-capture systems (60 Hz) and 6-components wheel dynamometers (hereafter referred to as "instrumented wheels") (38, 39), using force plates to compute friction coefficients (39). Others implemented the velocometer device (63) to assess MWC speed (52).

**Parameters of interest:** Parameters of interest included spatiotemporal parameters (time to perform the task, MWC mean and peak velocities, number or frequency of pushes, start, release, and total push angles, stroke patterns, acceleration over the first 2 or 3 pushes) ($n = 2$), kinematics (trunk, shoulder, elbow and wrist peak joint angle and ranges of motion (RoM)) ($n = 1$) and kinetics (maximum power output, external mechanical work, mean rolling resistance coefficient) ($n = 1$).

**Results:** Results showed that the time required to perform a task increased with increasing the camber angle from 15 to 24°, leading to a deterioration of overground sprint and mobility drills performances (52). Similarly, the release angle and the trunk RoM were shown to increase with increasing camber angle (38). The last authors also found a trend of change in the stroke pattern toward a single looping over propulsion (SLOP) pattern (64). Finally, both rolling resistance and total power output were also found to increase with camber (39), explained by the modification of the wheel-ground contact surface.

#### *Roller Ergometer*

**Methods:** Three studies used commercially available roller ergometers and were all sports oriented (21, 35, 36). They

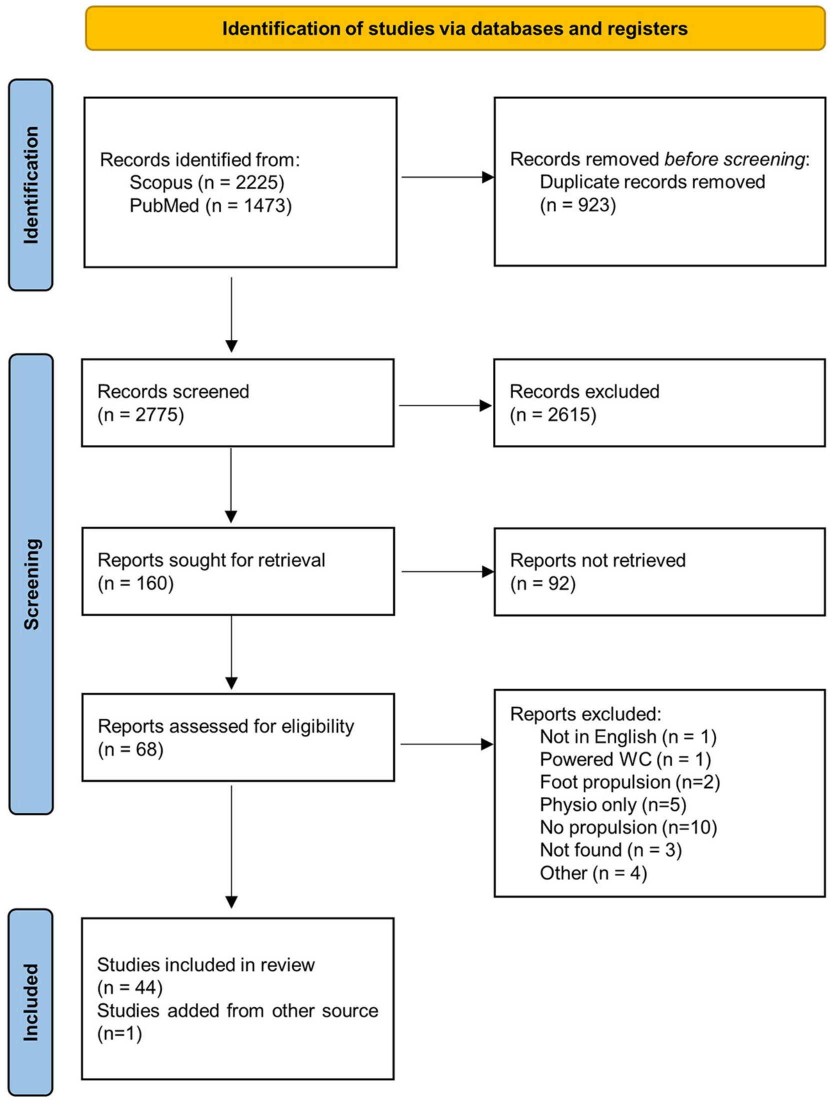



**FIGURE 1 |** PRISMA 2020 flow diagram.

involved experienced MWC users such as MWC basketball and rugby athletes. Camber varied from 9° to 22°, with 3 different camber angles per study. Experiments were systematically performed using the same MWC for all subjects.

The initial MWC configuration was defined through brand, model, weight, overall length, seat and backrest angles, backrest height, seat depth and width, rear-wheel diameter and tire pressure in two studies (35, 36), whereas it was limited to weight, rear-wheel diameter, seat depth and height in the remaining article (21).

One study adjusted the initial configuration to every subject by mimicking the participants' own MWC configuration (21), but the resulting configurations were not reported. For the two other articles, the top-to-top rear wheel width was maintained constant (48 cm) between configurations (35, 36). Finally, two out of three articles specified that a special care was taken to avoid "toe-in toe-out" between configurations.

For all studies, participants were asked to propel at maximal speed for 8 s.

**Parameters of interest:** Parameters of interest included spatiotemporal (MWC average speed, push time, recovery time) ($n$ = 2) and kinetic (residual torque, power output) ($n$ =2 ) parameters.

**Results:** Temporal parameters showed conflicting results; one article reported an increase of the push time with camber angle (35) whereas another one did not report any change (21). Higher camber angles were associated with a decrease of the recovery time (21) and of the MWC speed (35). This last result can be explained by the increase of rolling resistance (35, 36) due to the type of contact between the wheels and the rollers.

**TABLE 1** | Number and type of participants to each study.

| Number of subjects | Able-bodied | | Spinal cord injury (SCI) | Older people | Athletes* |
|---|---|---|---|---|---|
| | Novice | Experienced | | | |
| 1 | (18, 19) | | | | (20) |
| 2–5 | | | | | (21, 22) |
| 6–10 | (23–28) **(29)** **(30)** **(31)** (32, 33) | | **(29)** **(30)** **(31)** (34) | | (35–37) |
| 11–20 | (38–43) **(44)**** | (45) | (46–49) | (50, 51) | (52–56) |
| 21 + | (57) | | (58, 59) **(44)**** | (60) | (61, 62) |

*Bold, same study involving different cohorts; *Athletes cohorts described by pathology or classification in the articles, not reported in this table; **Lin and Sprigle (44): cohort of SCI subjects with one subject with ataxia.*

Finally, Faupin et al. (36) showed that reorienting the rollers perpendicular to the wheel plane allowed for a more realistic setup by maintaining residual torque closer to overground or treadmill conditions.

### *Treadmill*

**Methods:** Two articles resorted to treadmill experiments to investigate the effect of camber on propulsion. One article studied daily life speeds and involved AB subjects (23), whereas the other article was sports oriented and recruited highly trained basketball and tennis wheelchair athletes (53). Camber angles were between 0° and 9° for the first study and between 15° and 24° for the second one. In each study, the same MWC was used for all participants.

Both articles described the initial MWC configuration through brand, model, weight, handrim or rear wheel diameter, tire brand or pressure and seat height. Seat height was adapted to the participants through the elbow angle when subjects were seated in the MWC with their back resting on the backrest and their hands placed at the handrim top dead center. For Veeger et al. (23), the elbow angle was fixed at 120° for all participants whereas Mason et al. (53) reproduced the angle of participants sitting in their own MWC. When varying configurations, Mason et al. (53) specified maintaining the participant's elbow angle constant and a 48 cm top-to-top rear wheel width across every tested configuration. Mason et al. also specified that "toe-in toe-out" was controlled between configurations, while Veeger et al. (23) took special care in maintaining an equal rolling resistance between configurations.

Both experiments were performed on the same commercially available motor-driven treadmill. For each configuration, subjects were asked to propel for 12 min with increasing speed every 3 min (0.56, 0.83, 1.11, 1.39 m/s) (23), or to propel for 4 min at 2.2 m/s on a 0.7% gradient slope (53).

**Materials:** Both experiments were monitored with video cameras coupled with an optoelectronic motion capture system (53) and EMG electrodes (23).

**Parameters of interest:** Parameters of interest included spatiotemporal parameters (contact and release angles, push and recovery times) ($n = 2$), kinematics (shoulder flexion/extension and abduction/adduction, elbow and trunk flexion, shoulder, elbow, and wrist angular velocities) ($n = 2$), kinetics (rolling resistance, power output) ($n = 2$) and muscular activity (upper limb and trunk muscles activation) ($n = 1$).

**Results:** One study reported differences in push time, push angle and shoulder abduction between 3° and 6° camber angles (23) whereas the other study did not report such results but an increase in shoulder, elbow and trunk RoM in the sagittal plane (53). Finally, one study reported a decrease in rolling resistance with camber (23) whereas the other study found an increase (53).

## Rear-Wheel Diameter

**Methods:** Rear wheel diameter was studied in three articles involving experienced MWC basketball athletes. Experiments were either performed on the athlete's personal MWC (61), whose configuration was not reported, or on a MWC provided by the authors which was the same for all subjects (54, 55). In that case, the brand, model, gear ratio (i.e., handrim radius divided by wheel radius), camber, and tire pressure were reported. Furthermore, seat height was adapted to each participant through the reproduction of the elbow angle they have in their personal basketball MWC. This elbow angle and the gear ratio were maintained constant for every rear wheel diameter. The authors reported that they were not able to maintain the top-to-top rear wheel width constant between configurations due to camber. Besides, the authors did not report if the change in rear wheel size was associated with an adaptation of both the seat angle and the inclination of the caster fork axle with respect to the MWC frame, while the latter is crucial for turning maneuvers.

Regarding the experimental conditions, one study was carried out on a treadmill at 2.2 m/s (55) whereas the other two studies consisted of overground mobility tests performed at maximal speed such as a 20 m sprint, a linear mobility drill requiring multiple successive forward and backward propulsions and an agility drill composed of sharp turns (54) or the Wheelchair Mobility Performance test composed of 15 mobility exercises such as a 12-meter sprint and a rotation, with and without handling a ball (61).

**Materials:** Measurements involved video cameras (55, 61), a velocometer (54), and an instrumented wheel with additional weight around the hub of the opposing wheel to counterbalance its weight and inertia (55).

**Parameters of interest:** Parameters of interest included spatiotemporal parameters (time to perform the task, stroke frequency, push time, push angle, acceleration over 2 and 3 pushes, peak velocity) ($n$ = 3), upper limb kinematics (joints angular displacement at contact and release instants: shoulder flexion and abduction, elbow and trunk flexion, wrist extension) ($n = 1$) and handrim kinetics (resultant and tangential forces, fraction of effective force (FEF), mechanical work, and power) ($n = 1$).

**Results:** Results showed that larger rear-wheel diameter improved sprinting performances without negatively influencing initial acceleration or maneuverability performances (55, 61). If push time, stroke frequency and upper body joint kinematics were not found to be altered by rear wheel diameter, push angle was reported to increase with wheel diameter (55). Larger rear wheel diameters were also associated with smaller handrim total force and larger tangential component.

### Handrim Diameter

**Methods and Materials:** Three studies focused on the effect of handrim diameter, either involving novice AB subjects using the same MWC (24, 25) or focusing on a single wheelchair racing athlete in his personal racing MWC (20). Handrim diameters ranging from 34 to 37 cm were studied for the racing MWC, while larger handrim diameters, from 32 to 54 cm, were studied in the two other articles (24, 25). In all studies, the handrim diameter varied while keeping the rear wheel diameter constant, inducing an alteration of gear ratio for each configuration.

Description of the MWC characteristics was done through rear wheel diameter and seat depth, width, and height with respect to the ground for one study (25), whereas only the brand and model was given for another article (20) and no information at all were reported in the third one (24).

The experiments on racing MWC took place on a 400 m long athletics track, on which the subject performed laps and 5 min bouts of propulsion, both with 200 m head starts at speeds ranging between 12 and 24 km/h (20). A 500 Hz camera mounted on the MWC allowed for the definition of the propulsion cycle parameters. Participants to the two other studies were asked to propel overground at self-selected speeds over 25 m (25) or for 5 propulsion cycles (24) in a motion-capture equipped runway. One study used pressure sensors, placed inside gloves (25), and the other used an instrumented wheel (24).

**Parameters of interest:** Parameters of interest included spatiotemporal parameters (stroke frequency, push time) ($n$ = 1), upper limb kinematics (shoulder and elbow flexion/extension RoM, shoulder adduction/abduction and rotation RoM) ($n$ = 1), and upper limb kinetics (hand pressure, mechanical power, power flow) ($n = 2$).

**Results:** Results on spatiotemporal parameters in racing MWC showed that smaller handrims resulted in longer push time and lower push frequency (20). In standard overground propulsion, even if the speed was self-selected, no differences in speed were observed between the different handrim diameters. However, larger handrim diameter was found to be associated with larger shoulder and elbow RoM and larger hand contact forces and pressure (25) and related to greater work and total mechanical energy in upper extremity segments during propulsion (24).

### Handrim Shape

**Methods:** Two articles studied handrim shape, involving novice AB subjects and using the same MWC for all subjects (26, 57). The articles compared the use of conventional 18- and 20-mm diameter cylindrical metallic handrim either to an oval shape section handrim or to an ergonomically shaped handrim.

The first study (26) used a custom built simulator described by handrim diameter, camber, backrest height and seat height and angle, asking participants to perform two submaximal exercice tests. For the second article, each handrim was mounted on a separate set of wheels with different tire type and pressure, which were reported along with backrest height and seat width and depth to describe the MWC configuration. The experiments consisted of 8-shape displacements at comfort speed.

**Materials:** Measurements were carried out using the custom-built simulator or a Grip VersaTek Wireless System placed inside gloves allowing for the measurement of hand pressure (57).

**Parameters of interest:** Parameters of interest included spatiotemporal parameters (cycle time, push frequency, push angle) ($n = 1$), handrim kinetics ($n = 1$) and upper limb kinetics (hand pressure) ($n = 1$).

**Results:** No significant effect of handrim shape was observed on spatiotemporal propulsion parameters or in power output (26). The contoured handrim design was related to reduced levels of contact pressure on most hand regions, however it concentrated a high level of pressure on the medial phalanges, preventing the authors from recommending this feature (57).

### Tire Type and Pressure

**Methods:** One article focused on the effect of tire type and pressure on physical strain and propulsion technique (40). Experiments were conducted on novice AB subjects. Two tire types (pneumatic and solid) and four pneumatic tire pressure conditions [100, 75, 50, and 25% of the recommended pressure (i.e., 6 bars in that case)] were evaluated. A configuration with extra mass (5 and 10 kg) added on the rear wheel axle was examined both for 100% pressure pneumatic tires and for solid tires. All the participants used the same MWC, defined by brand, model, total mass, rear wheel diameter, camber and seat and backrest angles with respect to the horizontal and frontal planes, respectively. The experiments consisted of 4-min bouts of propulsion at 1.11 m/s on a level treadmill.

**Materials and parameters of interest:** The MWC was equipped with 2 instrumented wheels, allowing for the measurement of spatiotemporal (push time, cycle time, push frequency and push angle) and kinetic (total and tangential forces, propelling torque, FEF and power) parameters.

**Results:** Results showed that lower tire pressure resulted in smaller cycle time and push angle, as well as in significantly lower FEF and higher power output. Solid tires were also found to increase power output. Additional mass did not have a significant impact on propulsion technique (timing and force application) or power output, although a trend of increasing power output with solid tires was observed.

## Seating-Related Articles

Nine articles studied seating-related characteristics, with focus on seat and backrest angles ($n = 6$), backrest height ($n = 1$), and footrest positioning ($n = 2$ ).

### Seat and Backrest Angles

**Methods:** Six articles focused on seat and backrest angles. Two articles included the effect of both angles on propulsion with elderly people (50, 51) and four articles focused only on the effect of seat angle either on spine curvature and scapular kinematics during propulsion in users with spinal cord injury (SCI) (58), on mobility and propulsion kinematics in elite MWC rugby players (37), on seating ergonomics and mobility efficiency in SCI users (46), or on the position of the MWC-user's CoM during propulsion of one AB user (18). Overall, seat angles were studied in the range of 0° to 14°; and backrest angles were studied between 95° and 105°. All articles defined seat and backrest angles with respect to the horizontal plane. However, one article, using a specific platform, tilted the entire MWC during its experiments (18). Except for the characteristics of interest (i.e., seat and backrest angles), the descriptions of the MWC configurations were scarce in all the retrieved studies.

Experiments were conducted at comfort speed (∼1 m/s) using a custom-built ergometer (18, 50, 51), the participants' own MWC on a roller ergometer (58), or the same MWC for all participants, either using the same predefined configuration for all participants on a treadmill (46) or mimicking each participant's own MWC configuration during maximal speed overground mobility tests (37).

**Materials:** Experiments were monitored with optoelectronic motion capture systems (50, 51, 58), video cameras (37, 46), or inertial measurement units (IMU) (37). An instrumented wheel was used in two studies, adjusting the second wheel inertial properties by adding weight to it (50, 51).

**Parameters of interest:** Measurements included spatiotemporal data (push frequency, contact, release and total push angles, time to perform the task) ($n = 4$), kinematics (shoulder rotations, global CoM displacement) ($n = 2$) and kinetics (handrim forces, FEF, shoulder net joint moments, power output, mechanical efficiency) ($n = 3$).

**Results:** Results showed that push angle increased with increasing seat (46, 51) and backrest angles (51). During sprint and agility tasks, reduced seat angle was found to reduce the time required to perform the task (37). Regarding kinematics, seat angle did not alter glenohumeral rotation, but higher inclination resulted in higher scapulothoracic internal rotation (58). Regarding kinetics, FEF was improved with increasing seat and backrest angles (50) without affecting peak and mean shoulder net joint moment in elderly people (51).

### Backrest Height

**Methods:** One article focused on backrest height (59). Experiments were conducted in SCI users with injury from T8 to L5 vertebrae. Two backrest heights were tested, described as a fixed height of 40.6 cm for the highest condition and subject-specific (50% of the user's trunk length) for the lowest. All the participants used an identical MWC, provided into two seat widths to accommodate various body sizes but the MWCs characteristics were not reported. Only backrest height varied between tested configurations. The experiment consisted of four 30-s propulsions at 0.9 m/s on a treadmill, with varying slope inclination (0–3°).

**Materials and parameters of interest:** An optoelectronic motion capture system was used, coupled with two instrumented wheels allowing for the determination of spatiotemporal (push time and push angle), kinematic (shoulder peak extension and shoulder flexion/extension RoM), and kinetic (mechanical effective force) parameters.

**Results:** Results showed a smaller push time and push angle and a higher push frequency with the higher backrest, which also resulted in smaller shoulder extension angles at the beginning of push phase and smaller shoulder flexion/extension RoM. The mechanical effective force was not found to be altered by seat height.

### Footrest Positioning

**Methods:** Two articles studied the impact of footrest positioning on MWC propulsion in AB participants. One examined the effect of footrest angle, defined through knee flexion, on MWC turning maneuver (27), while the second studied the effect of footrest height, defined through hip flexion, on MWC linear acceleration during a straight-line displacement (45). The first article examined fully extended (0° knee flexion) and fully flexed (120° knee flexion) positions during angular velocity tests, requiring the participants to rotate the MWC over 900° (2.5 full turns) as fast as possible (27). Prior to angular velocity tests, the MWC-user's CoM, overall length, rolling and turning resistances and yaw mass moment of inertia (MoI) were determined. For the second article, three hip flexion angles (0°, 45°, and 90°) were tested, and the participants were asked to propel at maximal speed on a custom-built roller ergometer for 20 s.

Both articles used the same MWC for all participants without adjustments. The MWC characteristics were described through brand, model, seat width and depth, backrest height and rear wheel diameter for both articles, plus seat angle and rear wheel camber (45) or rear-wheel axle plate position, MWC-user's CoM, overall length, rolling and turning resistances and yaw mass MoI (27). During the experiments, the modifications in footrest positioning (height or angle) did not impact any other MWC geometrical characteristics.

**Materials and parameters of interest:** Both experiments were monitored by video cameras, allowing for the determination of spatiotemporal (task time, peak velocity and acceleration during the first 2 s from standstill, covered distance after 1, 2, and 3 s from standstill) ($n = 2$), and kinematic (trunk flexion/extension positions) ($n = 1$) parameters.

**Results:** Results showed that fully flexed knee position resulted in a greater angular velocity, a more rearward position of the CoM and thus a decrease in rolling and turning resistances. Results also showed an improvement of the covered distance during the first 3 s when thighs were parallel to the floor and a reduced capacity to accelerate was noted with hip completely flexed (thighs on the trunk with vertical trunk). Regarding trunk kinematics, flexion/extension average position was altered but the trunk only actively participated in the first push in the condition with thighs parallel to the floor.

## Seat Vertical and Horizontal Positions

Twenty articles studied the position of the seat relative to the wheels, with focus on the vertical position (seat height) ($n = 6$), on the horizontal position (seat fore-aft position) ($n = 6$), or on both ($n = 8$).

### Seat Vertical Position (Seat Height)

The 14 articles studying seat height can be organized following their experimental propulsion conditions: overground ($n = 5$), on a treadmill ($n = 2$), on a roller ergometer ($n = 4$) or on a stationary wheelchair simulator ($n = 3$).

#### *Overground*

**Methods:** Articles studying the effect of seat height on the biomechanics of overground propulsion involved either AB participants (41) or experienced MWC users (37, 47, 56, 62). Three articles were sports oriented: two focused on MWC basketball (56, 62) and one on MWC rugby (37). For these studies, participants either performed the Wheelchair Mobility Performance test gathering 15 sport-specific tasks (56, 62) or a combination of 5 m sprints, Illinois Agility Test, and a specific "skill" test (37). Other experiments consisted of overground propulsion at a self-selected speed (47) or of maximal speed propulsion over a 3 m long ramp with a 1:12 slope (41). Two articles (56, 62) used the athlete's own MWC and moved the seat up and down by 7.5% of its initial position. Two other studies used a custom-made adjustable MWC and modified seat height either by plus and minus 15 mm (37) or using four pre-selected heights covering 10 cm from the lowest to the highest position (47). The last article tested four seat heights defined from elbow flexion angle (0°, 30°, 60°, and 90°) using the same MWC for all participants (41).

MWC configurations were either not described (41, 56, 62) or described through seat depth, angle, and tire pressure (37) or through brand, model, rear and front wheel diameter and type, handrims diameter, seat width and depth and camber angle (47).

While varying seat position, one article specified keeping constant "all other configuration parameters" (37), and two other studies mentioned "preserving other chair ratios" and modifying backrest and footplate heights by the same amount as seat height (56, 62). One article provided the seat angles associated with the highest and lowest tested seat heights (47).

**Materials:** Experiments were monitored using an optoelectronic motion capture system combined with instrumented wheels (47), EMG electrodes (41), IMUs (37, 56) and video cameras (37, 62).

**Parameters of interest:** Parameters of interest included spatiotemporal parameters (time to perform the task, push frequency, push and recovery times, distance traveled per stroke, contact, release, and total push angles, MWC peak or average forward and rotational speed and acceleration) ($n = 4$), kinematics (elbow flexion angle) ($n = 1$), kinetics (axial, tangential and radial handrim forces, FEF, peak propelling torque) ($n = 1$) and muscular activity of the upper limbs ($n = 1$).

**Results:** Results showed an increase of both push time and push angle with lower seat heights (47). Regarding task time, contradictory results were obtained with either a decrease (56, 62) or an increase (37) with lower seat positions. Increasing seat height was also found to decrease the elbow flexion angle when the hand is at the handrim top dead center (47). Regarding handrim forces, lower seat positions were found to increase peak radial and axial forces but were not found to impact the tangential component, the mean FEF or the peak propelling torque (47). Finally, higher activation levels of the pectoralis major and of the triceps muscles were associated with lower seat positions (41).

#### *Treadmill*

**Methods:** Two articles used treadmill experiments, involving AB subjects propelling for 12 min with increasing speed every 3 min (0.56, 0.83, 1.11, 1.39 m/s) (28) or SCI MWC users propelling for 6 min at 1 m/s (46).

The first study (28) used the same solid-frame basketball MWC for all participants, with an adapted wood seat allowing for seat height modifications independently from seat fore-aft position. The MWC initial configuration was described through brand, weight, caster and rear wheel diameters, handrim diameter, tire pressure and camber angle; and four different seat heights were investigated, defined through the elbow extension angle when the hand is placed at the handrim top dead center (100°, 120°, 140°, and 160°). The second study (46) also used a single MWC for all participants adapted to fit each subject by modifying seat width and backrest angle. The MWC configuration was described by brand and model and two seating positions were investigated, defined through the seat angle (5° and 12°), with a difference in seat height of 55 mm.

**Materials:** Measurements involved video cameras for both articles and EMG electrodes (28).

**Parameters of interest:** Parameters of interest included spatiotemporal parameters (push angle, push frequency, contact and release angles) ($n = 2$), kinematics (trunk elbow and shoulder flexion/extension) ($n = 1$) and muscular activity (left arm and trunk muscles) ($n = 1$).

**Results:** Results showed a decrease in push angle and push frequency with increasing seat height in both articles. Regarding kinematics, higher seat heights resulted in a decrease of elbow flexion and shoulder extension and abduction, while elbow extension and trunk flexion were increased (28). Finally, a shorter activation period was found with a higher seat for upper-limb muscles, except for the triceps, which exhibited a longer activity (28).

#### *Roller Ergometer*

**Methods:** Over the four articles that used a roller ergometer, two included both AB participants and MWC users for comparison during daily locomotion (29, 30) and two focused on sports MWC, involving experienced MWC athletes. The latter studied rugby with propulsions at maximal speed for the equivalent of a 14 m sprint (21) and racing with propulsions at 60% of the participant's maximal speed for three 90-s trials (22). For daily locomotion, participants were asked to propel at a self-selected speed for 15 propulsion cycles (29, 30).

All the articles used a single adjustable MWC for all their participants. The initial MWC configuration was either described through weight, rear-wheel diameter, seat depth and height (21); through brand, camber, seat and seat-to-backrest angles (22); or not described at all (29, 30). Overall, the configuration was specified to be controlled and/or maintained constant while changing the seat height. Regarding the number of tested configurations, two articles tested three seat heights (44, 47, 50 cm; distances taken between the ground and the back of the seat) (29, 30); one article investigated the participant's usual MWC seat height and two other heights (3 and 6 cm above the usual seat height) (21) and the last article tested two positions which were defined by the seat position at which the user's distal phalanges were aligned with the lowest portion of the handrims and 10 percent of the subject's arm length higher (22).

**Materials:** Experiments were performed on custom-built roller ergometers for all the articles, except for one that used a commercially available ergometer (21). Measurement systems included an optoelectronic motion capture system (29, 30), video cameras (22), and built-in sensors within the ergometer (21). Two articles included EMG electrodes on the participant's upper limbs (22, 29).

**Parameters of interest:** Parameters of interest included spatiotemporal parameters (cycle, push and recovery times, push frequency, push, contact and release angles, mean MWC velocity and push phase acceleration) ($n = 3$), kinematics (trunk, shoulder, elbow and wrist RoM, trunk, arm and hand angular velocities and accelerations, shoulder, elbow and wrist joints velocities) ($n = 2$) and muscular activity of the upper limb muscles ($n = 2$).

**Results:** Results showed a decrease in cycle time (21, 30), an increase in push phase acceleration (21), and a decrease in upper limb RoMs (30) while increasing seat height. Regarding muscle activity, contradictory results were found with seat height associated both with an increase in upper-limb muscle activation during push phase (29), but also with a decrease in muscle activation over the whole cycle, including both push and recovery phases (22).

#### *Stationary Wheelchair Simulator*

**Methods and materials:** Three articles used stationary simulators to investigate the effect of seat height. Studies involved either AB participants (42), SCI subjects (48) or both (31). Participants were asked to propel at daily life speeds of 3 km/h with power output at 7.5 W for all participants (31), or between 0.42 and 0.83 m/s and with individual power output ranging from 5.5 to 14 W (48), or to perform maximum isometric exercises during 6 s per configuration (42).

The stationary wheelchair simulators were either described through camber, seat and seat-to-backrest angles, wheel and handrim diameters, size of the rim tube, top-to-top rear wheel width and seat fore-aft position (48); through seat fore-aft position and handrim diameter (31), or through handrim radius (42). In one article, the simulator was adapted to every participant by aligning the subject's acromion vertically above the wheel axle (48).

Regarding the investigated seat heights, all three studies tested seat heights defined through the elbow flexion angle when hands are at the handrim top dead centers. One included two seat heights with values of 90° and 100°, corresponding to an average difference between seat heights of 3.3 cm (31). Another compared eight seat heights with steps of 10° from 70° to 140° (48). The last study investigated nine configurations, defined from both shoulder and elbow angles to define both vertical and fore-aft position of the seat, (shoulder: from 30° to 70° with 10° increments; elbow: from 65° to 100° with 5° increments) (42). When altering seat height, one article mentioned keeping all the other settings constant (48). The seat and the wheels were separated in the other two custom-made ergometers (31, 42), allowing for the independent modification of seat height.

**Parameters of interest:** Parameters of interest included spatiotemporal parameters (cycle, push, and recovery times) ($n = 1$), kinematics (trunk, shoulder, and elbow flexion/extension RoMs) ($n = 1$), kinetics (FEF, average and peak propelling torque) ($n = 1$) and upper limb muscles activation ($n = 1$).

**Results:** Despite the small difference (3.3 cm) between the seat heights tested, probably explaining the lack of differences observed in propulsion temporal characteristics (cycle, push, and recovery times), the lower seat position resulted in higher upper-limbs RoM in Hughes et al. (31). Regarding kinetics, FEF was found to increase with seat height (48) contrary to the average and peak torques during isometric measurement (42). Similarly, seat height was not found to influence anterior deltoid and triceps muscles activities (42).

### Seat Horizontal Position (Seat Fore-aft Position)

Fourteen articles focused on seat fore-aft position and are presented below according to their experimental conditions: overground ($n = 6$), on a roller ergometer ($n = 6$), or on a stationary wheelchair ergometer ($n = 2$).

#### *Overground*

**Methods:** Six articles used overground propulsion and involved experienced MWC users (37, 47), novice AB participants (19, 32, 43) or older people with no information on their MWC experience (60). One article focused on MWC rugby athletes (37). Participants were asked to propel at a comfortable self-selected speed either for at least 4 cycles on a linear path on 4 different surfaces (60), for 10 m (32), for 20 m (47) or for 30 cycles (43); to perform a combination of 5 m sprints, Illinois agility test and a specific "skill" test (37), or a combination of a 15 m straight line sprint and a slalom course (19).

The six studies used a single MWC for all their participants. The initial configuration was either described through MWC brand, rear-wheel diameter, tire type, handrim diameter, front wheel type, seat width and depth and camber angle (47); through brand, mass, seat width, depth, height and inclination, cushion thickness and type, side guards material, rear and caster wheel types and diameters, handrim material and camber (60); through brand and seat height (32); through fore-aft seat position (19); through seat height, angle, and tire pressure (37) or not described

(43). All articles except one (43) used an adjustable MWC, but only three reported a MWC adaptation to each participant (37, 47, 60), and Kotajarvi et al. (47) specified this adaptation to be the reproduction of the participant's backrest and footrest heights.

In all studies, changes in seat fore-aft position were defined by the difference with respect to an initial configuration, but only one study (19) also provided the actual fore-aft position of the seat with respect to the rear wheel center. The total amplitude of variation varied between 3 cm (37) and 8 cm (47, 60). Four studies specified that some or all the other settings were maintained constant and/or controlled between configurations (32, 37, 47, 60).

**Materials:** Regarding measurement devices, the experiments used optoelectronic motion capture systems (32, 47), video cameras (37), instrumented wheels (47, 60), IMUs or accelerometers (37, 43), and EMG electrodes (19, 32).

**Parameters of interest:** The parameters of interest included spatiotemporal parameters (average speed, stroke time and frequency, stroke distance, push and recovery times and contact and release angles) ($n = 3$), kinematics ($n = 1$) (trunk, shoulder, elbow, and wrist ROMs), kinetics (total handrim force and, radial and tangential components, propelling torque, FEF) ($n = 2$) and upper limb muscles activations ($n = 3$).

**Results:** Results showed that the fore-aft seat position did not impact push frequency and stroke distance (60), but a more forward seat position was found to improve skill test performances with sports MWC (37). Regarding kinematics, a forward position of the seat was found to increase the RoM of all the upper limb joints (32). As for kinetics, a rearward seat position was associated with lower total and tangential handrim forces (60) and with reduced upper limb muscle activity in daily life (32) and for sport (19).

### *Roller Ergometer*

**Methods:** Six articles used a roller ergometer to study the effect of seat horizontal position, all involving experienced MWC users, and two also including AB participants for comparison (29, 30). Sport propulsion was considered in two articles, with focus on rugby with propulsions at maximal speed for the equivalent of a 14 m sprint (21) and racing with propulsions at 60% of the participant's maximal speed for three 90-s trials (22). For daily life locomotion, participants were asked to propel at self-selected speed for 15 propulsion cycles (29, 30) or at comfort speed for two trials of graded propulsion at 8% incline (34, 49).

Most studies used a single adjustable MWC for all their participants, except for two studies that used two similar adjustable MWCs with different seat width to cover all anthropometric differences (34, 49). Overall, four studies added custom-made adjusting systems on a MWC (29, 30, 34, 49). Three studies used individual initial configurations, either based on the reproduction of the participant's own MWC (21) which were not reported or based on the vertical alignment of the subject's shoulder with the rear-wheel center (34, 49). The initial MWC configuration was either described through weight, rear-wheel diameter, seat depth and height (21); through brand, wheel camber, seat and seat-to-backrest angles (22); through brand, seat width and camber (34); through brand and seat width (49); or not described (29, 30).

Seat fore-aft position was identified by the horizontal distance between the rear-wheel axle and the back of the seat, either as an absolute value (22, 29, 30), or relative to its initial position (21, 34, 49).

Regarding the number of tested configurations, the studies investigated between 2 and 4 fore-aft positions of the seat with the total amplitude of variation going from 6 to 10 cm. Three articles specified controlling and/or maintaining the MWC configuration constant while changing the seat fore-aft position (21, 22, 30).

**Materials:** Regarding experimental conditions, studies were performed on a commercially available ergometer (21), on custom-built roller ergometers (22, 29, 30), or on a roller ergometer with removable flywheels (34, 49). Measurement systems included optoelectronic motion capture systems (29, 30, 34, 49), video cameras (22), instrumented wheels (34, 49), and surface (22, 29) or fine-wire EMG electrodes (34, 49).

**Parameters of interest:** The parameters of interest included spatiotemporal parameters (cycle, push, and recovery times; push frequency; push, contact and release angles; MWC average speed and push phase MWC mean acceleration) ($n = 5$), kinematics (trunk, shoulder, elbow, and wrist RoM; and trunk, arm and hand angular velocities and accelerations) ($n = 2$), kinetics (propelling torque and power; shoulder net joint force) ($n = 1$) and upper limb muscle activation ($n = 4$).

**Results:** Results showed an increase of the push angle (21, 30), an increase of upper limb joint RoM (30), and a decrease of peak elbow extension velocity (22) with a more backward seat position. The net shoulder force direction was also impacted by the seat fore-aft position (49). Finally, EMG outputs showed that the combination of backward and low seat position was associated with the lowest muscle activation level (pectoralis major and anterior deltoid muscles) (22, 34). However, other authors found contradictory results with a higher muscle activation for a backward position of the seat (29).

### *Stationary Wheelchair Simulator*

**Methods:** Two articles used stationary MWC simulators (31, 42) to study the effect of seat fore-aft position on propulsion, involving either AB subjects (42) or both AB and SCI subjects (31); either performing maximum isometric pushes (42), or propelling in a straight line at 3 km/h with a power output of 7.5 W (31).

In both articles, the fore-aft position was modified through the position of the handrim hub relative to the back of the seat. The other MWC characteristics were not described. One study aligned the backrest with the hub as an initial position, and then defined two other tested positions with respect to the user's arm length (31). The second article considered 9 positions by altering both the vertical and fore-aft rear wheel positions, defined through shoulder and elbow flexion angles (42).

**Materials:** In addition to the measurements provided by the simulators, rotary potentiometers (31) and EMG electrodes were used (42).

**Parameters of interest:** The parameters of interest were kinematics (upper limb joint RoM) ($n = 1$), kinetics (peak and average torque, force vector) ($n = 1$) and muscular activity ($n = 1$).

**Results:** Results showed greater elbow and shoulder RoM in the frontal and transverse planes for frontward seat positions whereas shoulder RoM in the sagittal plane was greater for rearward seat positions (31). Isometric torque increased for a rearward seat position and the upper limb muscles seemed to be recruited differently between the handrim positions (42).

# DISCUSSION

Numerous articles revolving around MWC configuration and its impact on propulsion biomechanics were published in the last 40 years. From this consideration, the present review aimed at identifying and reporting the multiplicity of methodologies used in the literature to investigate the effect of MWC configurations on propulsion biomechanics, both for sports and everyday use. In doing so, this review highlighted issues in the methods implemented to study MWC configuration that are discussed below.

## Standardizing the Description of MWC Configuration

### An Intelligible Description of MWC Configuration

The first issue raised by this review is the lack of essential details in the description of MWC configurations despite it being crucial to ensure results portability to clinical or sports fields and to allow the comparison and aggregation of results between studies. Indeed, some articles reported information limited to MWC brand and model, which does not provide information about MWC characteristics. Therefore, it requires the reader to make tedious research on manufacturer commercial and technical booklets, which also limits the comparison of studies. Similarly, reporting tire type does not provide the reader with intelligible information; reporting rolling and steering resistances would be more informative. However, the level of essential details that are required also depends on the experimental propulsion condition (i.e., overground, treadmill, roller ergometers or stationary wheelchair ergometer) and the studied propulsion task. For instance, when studying turning, assessment of both the MWC CoM location and yaw mass MoI are crucial, which is not the case when studying straight line propulsion. Also, reporting the MWC mass when propelling on a roller ergometer is not relevant because the only useful information is the rolling resistance resulting from the load applied by the rear wheels on the rollers.

Hence, it seems necessary to standardize MWC configuration description, which would facilitate the comparison of results between studies and the reproduction of similar experimental conditions. It is also critical to ensure the efficient integration of results to clinical and sports fields for the benefit of MWC users. The following list displays the MWC parameters that should be systematically reported for an intelligible description of the MWC characteristics directly or indirectly linked to the MWC configuration:

- Dimensional parameters: rear wheel, caster and handrim diameters; seat width and depth; backrest width and height; rear and front wheel track; wheelbase; caster trail; footrest length.
- Positional parameters: rear wheel camber; backrest and seat angles; back of the seat fore-aft position with respect to the rear wheel axle; back seat height with respect to the ground; footrest position and orientation; fork axis angle.
- Mechanical parameters: inertial parameters (mass, CoM, yaw mass MoI); rolling and steering resistances.

However, obtaining all those parameters is not straightforward and determining MWC positional, dimensional, and mechanical characteristics is time consuming (12, 13, 65–69). The development of material and computer tools that allow a quick and easy determination of MWC characteristics would also favor their more systematic reporting in future publications. Additionally, reporting all these details will take a noticeable writing space in papers where word limits encourage not to report such level of information. Sharing additional data as Supplementary Material, for instance, would allow to overcome this issue.

### Description of Configuration Changes

Another challenge is the standardization of how the configuration is altered between tested configurations during the experiments. Firstly, some articles only reported the range of variation in the characteristic of interest without reporting the actual initial setting and the whole description of the MWC configuration, thus preventing reproducing their experiments. Secondly, most of MWC geometrical characteristics are interdependent (70), and one must be careful, when modifying a MWC dimensional or positional characteristic, to consider its impact on the others. Indeed, modifying one characteristic could require several adjustments to keep the rest of the configuration constant (e.g., the modification of rear wheel camber implies a variation in at least nine parameters of the MWC) (10). However, it might be impossible for many commercial MWCs. In that case, impacted settings should be monitored and reported.

Also, there can be an ambiguity between modifying a geometrical characteristic and modifying the mechanical system that allows this change. For instance, altering the seat angle could necessitate several mechanical changes if the seat height is expected to be maintained constant. Most of the articles indicated that the other settings were either "controlled" or "maintained constant," without providing a clear overview on what was actually unchanged and how it impacted the results.

Therefore, researchers are encouraged to select a MWC with adjustment modalities that do not generate other setting changes than the one under study. When not possible, a careful examination of the interdependent characteristics and how to correct them to maintain the rest of the configuration constant is necessary. When performed, authors are encouraged to specify that they have checked all the characteristics of the MWC for each configuration. Regarding mechanical properties, changes in MWC configuration result in changes in MWC-user's CoM position and yaw mass MoI; and in rolling and turning resistances. The researchers are thus incited either to try to compensate or, at least, to assess their impact on the provided results.

### "Absolute" vs. "Relative" Formalisms

Beyond the fact that, in the literature, multiple designations can sometimes refer to the same geometrical characteristic (e.g., seat fore-aft position and rear wheel axle fore-aft position), **Table 2** illustrates that two formalisms are commonly used to describe MWC configurations and their changes. Firstly, MWC characteristics can be expressed as dimensional measurements such as distances between points or angles between planes, as defined by the international standard ISO 7176-7 or in usual MWC provider datasheet. This formalism will be referred to as "absolute." Differently, MWC parameters can be defined according to the user's anthropometric parameters, the most frequent example being seat height defined from the user's elbow extension angle when hands are placed at the handrim top dead center (28). This formalism will be referred to as "relative" hereafter.

The "absolute" formalism has the merit of being self-explanatory as it echoes the measurements of manufacturers and occupational therapists. Still, some of the recommendations provided by the ISO standards are not practical to implement in a clinical or even research context (e.g., seating and wheel dimensions measured with a specific dummy in the MWC seat) and therefore are not always followed (71), leading to different measurements for the same characteristic (e.g., seat depth taken as seat upholstery depth or as the distance between the front of the seat upholstery and the intersection between the seat and the backrest).

The "relative" formalism can impose closer joint configurations between subjects, mitigating the effect of anthropometric differences and therefore of inter-individual variations on the studied outcome parameters (48), which is relevant to study the performance of the musculoskeletal system. Yet, by doing so, the absolute differences between the configurations are different for each participant, leading to non-uniform variations in the MWC mechanical parameters such as stability or rolling resistance among participants.

Both formalisms have their pros and cons, but the lack of consensus over which formalism to use with respect to the aim of the study combined with the almost systematic absence of the necessary data to switch from one formalism to the other complicates the comparison of results across articles.

Hence, the authors suggest the community provide a consensual way to describe MWC configurations, which would depend on the purpose of the study and would involve absolute or relative descriptions. Future articles are also encouraged to provide data allowing for the conversion from one formalism to the other.

## Importance of Methods and Experimental Setups

The multiplicity of methodologies used in the literature to study the effect of MCW configurations on propulsion biomechanics is explained by the fact that they each have their own advantages and drawbacks. The following paragraphs tackle the main methodological aspects:

### Experimental Environments

When studying MWC propulsion, the first important methodological choice is the "experimental environment." It can vary from free overground propulsion to propulsion on a wheelchair treadmill, on a roller ergometer or on a stationary wheelchair simulator. Each condition has its advantages and disadvantages, summarized in **Table 3**. For instance, overground propulsion appears to be the most ecologically valid testing environment, offering infinite trajectory possibilities, but it reaches its limitations when trying to monitor propulsion biomechanics and to control power output between configurations. Conversely, treadmills and roller ergometers are suitable for instrumentation, but a familiarization period is needed for the user and only straight-line propulsion can be simulated. Studies on treadmills must also prevent the subject from falling using a security system which impacts measurements.

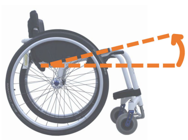
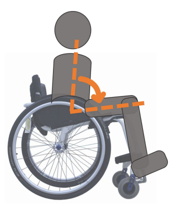
Comparisons of different experimental environments (e.g., overground, treadmill, roller ergometer, stationary wheelchair simulator) were previously performed by numerous authors (72–75) who agreed that the different experimental environments could be considered similar for the study of MWC propulsion. However, they only considered straight line displacement at steady state speeds. Moreover, it was already shown that MWC configuration affects the fore-aft stability during overground propulsion (16) and that roller ergometers or stationary wheelchair simulators prevent such a phenomenon from occurring. Hence, recommendations about which experimental environment should be used depending on the purpose of the study would provide more adapted results when studying the effect of MWC configuration.

### Control of Propulsion Speed and/or Power Output

Among the articles included in this review, most do not report the actual resistance or power output. Also, many asked the participants to choose "comfort" or "self-selected" propulsion speeds without reporting the actual speed. The impossibility to assess power output prevents from making synthesis by compiling results of different studies because speed and power output can critically influence the effect of a change in MWC characteristics on propulsion biomechanics (16). Therefore, both speed and power output should always be documented. However, accurate assessment of power output is not always straightforward and special care is necessary for its quantification (9, 76).

In addition, this assessment needs to be done for every configuration tested by a participant because both resistances (rolling and steering) and MoI are affected by changes in MWC configuration. Depending on the objective of the study, it would be necessary either to maintain power output between configurations or to report the resulting change in power output due to a change in MWC configuration. Indeed, if the goal focuses on performance of the musculoskeletal system resulting from changes in joints configuration induced by a change in MWC configuration, it would be necessary to maintain the power output between configurations. Because altering the velocity is known to affect propulsion biomechanics (77, 78), the rolling resistance needs to be adapted, that is necessary for experiment

**TABLE 2 |** List of the different geometrical characteristics of a MWC and definitions used throughout the literature.

| MWC geometrical characteristics | Representation(s)* | | Type | Different definitions in literature |
|---|---|---|---|---|
| | Absolute | Relative | | |
| Wheel camber |  | | *Positional* | • Angle of the main wheels in relation to the vertical |
| Wheels/ handrim size |  | | *Dimensional* | • Diameter of the rear-wheels only (Change in the gear ratio)<br>• Diameter of the handrim only (Change in the gear ratio)<br>• Diameter of both the rear-wheels and the handrim (No change in the gear ratio) |
| Seat angle |  | | *Positional* | • Seat angle from the horizontal plane<br>• System tilt angle (seat and backrest tilt)<br>• Seat dump |
| Backrest angle |  | | *Positional* | • Seat-to-backrest angle<br>• Angle between backrest and the horizontal or vertical plane |
| Backrest height |  |  | *Dimensional* | • Distance between the back of the seat and the top of the backrest<br>• Backrest placed at a specific trunk height |
| Footrest positioning |  |  | *Positional* | • Hip flexion angle<br>• Knee flexion/extension angle |
| Seat height |  |  | *Positional* | • Vertical distance between the floor and the back of the seat<br>• Vertical distance between the rear-wheel axle and the back of the seat<br>• Elbow flexion/extension angle<br>• Elbow and shoulder flexion/extension angles<br>• Difference in height at the top of the head<br>• Padding thickness |
| Seat fore-aft position / Rear wheel axle fore-aft position |  |  | *Positional* | • Fore-aft position of the seat (Rear-wheel axle horizontal position) relative to the rear-wheel hub (resp. to the seat) (absolute or relative to anthropometric features)<br>• Elbow and shoulder flexion/extension angles<br>• Backrest thickness<br>• Seat depth |

**Examples arbitrarily chosen by the authors.*

**TABLE 3 |** Experimental conditions advantages and disadvantages for manual wheelchair (MWC) propulsion evaluation.

| Experimental environment | Advantages | Disadvantages |
|---|---|---|
| Overground | - Most ecologically valid ("realistic") testing environment (requires trajectory and stability management while propelling)<br>- All movements are possible<br>- Can fit any MWC (including the user's own MWC) | - Changes are limited by the used MWC<br>- Instrumented wheels change the MWC inertial properties, influencing propulsive torque<br>- Difficult to control the velocity or the power output |
| Treadmill | - Can fit any MWC (including the user's own MWC)<br>- Need to control trajectory and stability<br>- Physiological and kinetics results close to overground propulsion<br>- Control of speed and power-output<br>- Effect of trunk motion taken into account | - Changes are limited by the used MWC<br>- No acceleration or sprint testing<br>- No turning, asymmetric propulsion<br>- Cross-slopes conditions difficult to safely reproduce<br>- Familiarization period needed<br>- Security system impacts measurements |
| Roller ergometer | - Can fit any MWC (including the user's own MWC)<br>- Physiological and kinetics results closest to overground propulsion regarding other ergometers<br>- Control of resistance/power output with certain ergometers | - Changes are limited by the used MWC<br>- Straight line propulsion simulation only (except for separated rollers ergometers with visual feedback)<br>- Trunk motion has no impact on MWC velocity and stability |
| Stationary wheelchair simulator | - Easy to adapt to every participant<br>- Any setting can be varied independently<br>- Adjustable resistance<br>- Can be easy to change the settings without interdependence with others setting | - Straightforward propulsion simulation only (except for haptic controlled ergometer and visual feedback)<br>- Trunk motion has no impact on MWC velocity and stability |

with propulsion overground, on a treadmill and on roller ergometers. Through all the studies included in this systematic review, there is only one study that performed such adaptation in power output (23). However, if the objective of the study is actual displacement performance, change in power output should not be compensated for, but should still be assessed and reported.

Moreover, regarding speed, it has already been shown that participants' self-selected speed on a treadmill is lower than their speed overground (79). Also, the usual speed studied in the literature ($1\,\text{m.s}^{-1}$) is above the average daily propulsion speed of MWC users ($0.5–0.8\,\text{m.s}^{-1}$) (80), but this is a consequence of averaging speed over short displacements from standstill to full stop. Other tasks than steady-state propulsion, while more representative of daily propulsion, are however left out when studying the effect of MWC configurations, likely due to the experimental environment.

## Measurement Systems

Along with the various experimental environments, a wide variety of measurement systems were used, from optoelectronic motion capture systems to IMUs, EMGs, pressure sheets, video cameras, force plates and instrumented wheels; each coming with its own pros and cons. For instance, IMUs allow to overcome the spatial restriction imposed by optoelectronic motion capture systems, enabling field measurements, but are less accurate to assess body orientation (81). Ideally, beyond their level of accuracy, measurement systems would not noticeably impact the subject's propulsion and the MWC characteristics. However, it is necessarily the case for some measurement systems such as instrumented wheels which modify wheel and MWC mass and mass moments of inertia (82, 83). Yet, the interest of measuring one parameter could be higher than the limitation induced by the measurement system. When using such a system, its expected impact on the results should be discussed in the study.

## MWC Used

Another important methodological choice is the MWC used for experimentation, which can either be the participant's own MWC or the same MWC for all participants, with adaptations to each participant or not. Using the same MWC for all participants standardizes some variables and makes the experiments easier to carry out and the results easier to interpret. Using each subject's personal MWC would be more realistic but would generate variations on MWC configurations and thus on power output. In that case, a precise description of each MWC initial configuration should be provided for this choice to be relevant.

## Participants

Choosing to study participants in their own MWC implies the recruitment of experienced MWC users for the experiments, which is generally associated with recruitment difficulties. Despite these difficulties, over 60% of the articles included in this review involved experienced MWC users (**Table 1**). The other articles involved novice AB subjects. The impact of studying MWC propulsion with novice AB participants has already been investigated multiple times, showing differences in power output (84), mechanical efficiency (85), energy expenditure (86), upper limb muscle recruitment (29) and kinematics (30). Based on these findings, generalization of results obtained on AB subjects to MWC users should be cautiously done. Despite numerous studies on the training of novice AB subjects (14, 87–91) showing significant improvements in propulsion technique, no article has yet been published on the amount of training necessary to achieve propulsion parameters (i.e., stroke pattern, timings, joint kinematics, forces, etc.) like those of experienced MWC users. One must be careful about the fact that the fatigue onset does not emerge at the same time for experienced and inexperienced users, and therefore propulsion time must be adapted when developing an experimental protocol. Additionally, AB subjects' morphology can be different from impaired users. However, despite these notable differences between MWC users and AB subjects, it remains possible that the conclusions on the effect of a MWC adjustment obtained in AB subjects remain valid for MWC users.

An alternative to recruiting AB subjects to compensate for the difficulty of recruiting MWC users is to enroll MWC athletes instead. Indeed, despite differences in their physical abilities, a recent study found that athletic users, that are generally easier to recruit for experiments, could be considered equivalent to non-athletic users when studying kinematic and kinetic parameters during daily propulsion (92).

It should be noted that MWC users are often considered as a homogeneous population despite being composed of a wide variety of people (spinal cord injury, multiple sclerosis, cerebral palsy, lower limb amputees, elderly people, etc.). This variability within the same group should be considered in studies, either by including diverse participants or by replicating the experiments on multiple cohorts.

### Experimental Task

Obviously, the experimental task plays a major role in the comparison of results. Despite the recent recommendations that biomechanical research should concentrate on initiating movement, maneuvering MWC and stopping to be more representative of the actual use of a MWC in a natural environment (80, 93, 94), researchers still tend to focus on studying straight-line propulsion at steady-state speed (61% of the studies). This is less of a concern for sports-oriented studies which tend to implement multiple tasks involving different speeds in their experimental protocols. However, in the latter case, the trend of developing specific tests in each study could make comparison and literature synthesis difficult.

## Number of MWC Characteristics Investigated

The next challenge to consider is the number of MWC characteristics investigated. Because geometrical characteristics might not have independent effects on outcome parameters, conclusions drawn from experiments performed using a given initial configuration might differ when another initial configuration is used. In other words, it means that the cross-effect of geometrical characteristics should be considered and that future studies should vary multiple geometrical characteristics and interpret the results accordingly. However, as displayed in **Table 4**, most articles studied a single MWC characteristic and a large majority studied either one or two MWC characteristics (respectively 60% and 88% of the studies). Indeed, increasing the number of investigated characteristics impacts the number of configurations to test which could compromise results due to subject fatigue or weariness. This bias can be reduced through order randomization of the tested configurations, which most of the studies did (i.e., 88% of the studies). Additionally, one must consider the amount of time necessary for one participant to adapt to a new configuration, which also impacts the total duration of the experiment.

**TABLE 4 |** Total number of configurations tested, and MWC characteristics investigated per article reviewed*.

| Total number of tested configurations | Number of MWC characteristics investigated | | | |
|---|---|---|---|---|
| | 1 | 2 | 3 | 4 |
| 1 | | (44) | | |
| 2 | (20, 26, 27, 34, 39, 49, 57–59) | (46) | | |
| 3 | (24, 25, 32, 35, 36, 38, 43, 45, 54, 55) | | | |
| 4 | (19, 23, 28, 41, 52, 53) | (60) | | |
| 6 | | (22, 31) | (56, 62) | |
| 8 | (48) | | (40) | |
| 9 | | (42, 47, 50, 51) | | (21, 37) |
| 12 | | (29, 30) | | |
| 27 | | (18) | | |

*Articles comparing distinct MWCs rather than a single MWC with distinct settings were not included in the table (n = 2).*

Promising techniques exist today to overcome the issue of testing multiple characteristics simultaneously such as fractional factorial experimental design or numerical simulations.

### Factorial Experimental Design

Because the number of investigated configurations increases exponentially with the number and the range of settings under study when using full factorial experimental design, some authors proposed to use fractional factorial designs, allowing for proper extrapolation of the results from a minimal number of configurations. Two articles listed in this review (21, 37) used Taguchi's methods (95) to reduce the number of configurations to test from 81 to 9, while varying simultaneously 4 settings. It must be noted that one hypothesis of Taguchi's experimental design is that input variables should be independent or have known simultaneous effect on the outcome parameters. This hypothesis was a major concern in both articles and remains unverified.

Therefore, further studies should first consider studying setting interactions to define those that can be neglected. Then, future studies could rely on experimental design to expand knowledge on MWC.

### Numerical Simulations

Another solution to avoid experimental limitations is to resort to numerical simulation. Some studies already embraced this approach based on simplified 2-D wheelchair propulsion models (14, 96–99), or through 3-D musculoskeletal simulations (100). Still, all these techniques rely on experimental data to feed the model.

Recently, fully-predictive simulation relying on optimal control theory was implemented to study MWC propulsion (101) and the technique was used to study the effect of seat position during sport propulsion on roller ergometer

(102), drawing meaningful perspectives. Contrary to the other previous numerical techniques, fully-predictive optimal control simulation does not require experimental data. However, these simulations are still relying on simplified 2-D models due to computational cost, and their application is limited to straight line propulsion on ergometer where they represent the model that needs to be implemented.

Hence, despite the unquestionable interest of numerical techniques to limit or to dispense with subjects' participation in experiments, a substantial workload remains. In particular, further work should focus on the validation of numerical techniques and the inclusion of subjects' variability to represent the various physical capacities of MWC users.

## LIMITATIONS

Through the methodological process described in the "Methods" section, it remains possible that the current review is still not exhaustive and that some articles are missing. In particular, articles not written in English were excluded and could have brought broader knowledge. However, the authors think this would neither alter the analysis done on methodology nor the recommendations that were made for future studies.

The authors also acknowledge that the focus of the present review on MWC propulsion does not allow to draw conclusions on the effects of MWC configuration in the MWC user daily life, as stability, accessibility, compatibility with accommodation arrangement, etc. should also be considered. However, most of the recommendations made here to study propulsion would remain valid for these other aspects.

Another limitation of this review is the focus on experimental methodology, which does not include biomechanical models and data processing choices, such as angle sequences or even coordinate system in which forces and moments are expressed (103–105). Standardization efforts are also needed on these aspects.

Despite these limitations, this review provides the scientific community with perspectives to coordinate research teams especially through consensual standardization and assistance for methodological choices depending on the aim of the study.

A quality assessment of the articles was not considered relevant in this review as the goal was to identify the different methodological choices necessary to study the effect of MWC configurations on propulsion and not to evaluate results from the different articles relative to their methodologies.

## CONCLUSION

The 45 articles reviewed in this article were designed to understand the impact of MWC configuration on propulsion biomechanics, a goal that is still not fully accomplished today. To achieve a global understanding of the relationship between MWC configuration and propulsion biomechanics, it is crucial to evaluate the impact of each MWC characteristic, in the wider range possible, on each outcome parameter studied, and for each experimental task (e.g., straight-line propulsion, turns, curbs, slope, cross-slope). Such a huge amount of work could only be done through collaboration between research teams on a global scale. However, this work needs standardization and recommendations beforehand, to avoid the pitfalls caused by using unsuitable methodologies (mainly due to limitations of lab facilities). Indeed, because each equipment is more adapted to certain study objectives than others, future recommendations could assist researchers in adapting their research goal to their available equipment. A standardization effort in reporting MWC configuration should also be done earlier on.

## DATA AVAILABILITY STATEMENT

The original contributions presented in the study are included in the article/**Supplementary Material**, further inquiries can be directed to the corresponding author.

## AUTHOR CONTRIBUTIONS

CF and YP: conceptualization, methodology, screening, analysis, and writing—original draft. JB and PT: supervision and writing—review and editing. CS: conceptualization, methodology, analysis, supervision, and writing—review and editing. All authors contributed to the article and approved the submitted version.

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## SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fresc.2022.863113/full#supplementary-material

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BRIEF RESEARCH REPORT
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# Impact of Sprinting and Dribbling on Shoulder Joint and Pushrim Kinetics in Wheelchair Basketball Athletes

***Félix Chénier*** [1,2]****, Ilona Alberca*** [3]***, Dany H. Gagnon*** [2,4] ***and Arnaud Faupin*** [3]

*[1] Mobility and Adaptive Sports Research Lab, Department of Physical Activity Science, Université du Québec à Montréal (UQAM), Montreal, QC, Canada, [2] Centre for Interdisciplinary Research in Rehabilitation of Greater Montreal (CRIR), Montreal, QC, Canada, [3] Université de Toulon, Impact de l'Activité Physique sur la Santé (UR IAPS n°201723207F), Campus de La Garde, Toulon, France, [4] School of Rehabilitation, Université de Montréal, Faculty of Medicine, Montreal, QC, Canada*

**Background:** While wheelchair basketball is one of the most popular Paralympic sports, it eventually causes shoulder problems and pain in many athletes. However, shoulder kinetics has never been assessed during propulsion in wheelchair basketball. This study analyzes the impact of sprinting and dribbling on pushrim and shoulder kinetics in terms of external forces and net muscular moments.

**Methods:** A group of 10 experienced wheelchair basketball athletes with various classifications performed four, 9-m sprints on a basketball court using classic synchronous propulsion, and four sprints while dribbling forward. Pushrim and shoulder kinetics were calculated by inverse dynamics, using a motion capture device and instrumented wheels.

**Findings:** Sprinting was associated to peak shoulder load from 13 to 346% higher than in previous studies on standard wheelchair propulsion in most force/moment components. Compared to sprinting without a ball, dribbling reduced the speed, the peak external forces in the anterior and medial direction at the shoulder, and the peak net shoulder moment of internal rotation.

**Interpretation:** The high shoulder load calculated during both sprinting and dribbling should be considered during training sessions to avoid overloading the shoulder. Dribbling generally reduced the shoulder load, which suggests that propelling while dribbling does not put the shoulder at more risk of musculoskeletal disorders than sprinting.

**Keywords: wheelchair sports, adaptive sports, performance, shoulder dynamics, biomechanics**



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*Riemer J. K. Vegter, University Medical Center Groningen, Netherlands*

***Reviewed by:***
*Rienk Van Der Slikke, The Hague University of Applied Sciences, Netherlands*
*Ciro Winckler, Federal University of São Paulo, Brazil*
*Jeffery Wade Rankin, Rancho Los Amigos National Rehabilitation Center, United States*

****Correspondence:***
*Félix Chénier*
*chenier.felix@uqam.ca*

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## INTRODUCTION

Adaptive sports offer many important benefits for people with disorders and disabilities, such as decreasing the risk of cardiovascular disease, improving general health and enhancing quality of life (1). Among the various adaptive sports available, wheelchair basketball (WB) is one of the most popular and is the most advanced in terms of organization, standardization and training quality (2–4). Each player is assigned a classification according to their functional ability. In Canada, where classification closely follows the International Wheelchair Basketball Federation (IWBF) but also allows able-bodied athletes to play, this classification ranges from 1 point (players with the least ability) to 4.5 points (minimal to no impairment).

This sport, which is very similar to its abled-bodied counterpart, contains intermittent phases of high intensity combining wheelchair maneuvers and ball handling. However, it is possible that such high intensity may be detrimental to the athletes' musculoskeletal integrity. In everyday mobility, propelling a standard manual wheelchair (MW) is considered in itself a high intensity activity and causes musculoskeletal disorders (MSD) in half of all MW users, especially at the shoulder (5–7).

It is still unclear whether playing WB puts the musculoskeletal system at higher risk compared to standard wheelchair propulsion. Finley and Rogers (6) found no difference in the occurrence of shoulder pain between athletic and non-athletic MW users. Wheelchair sports could have a protective effect by delaying the onset of symptoms (8), but these observations contradict those of Mateus (9) where 17 of 25 participants who reported pain during the last year attributed their pain to WB. Furthermore, Akbar et al. (10) reported that 76% of athletes who perform overhead sports such as WB have rotator cuff impairments, compared to 25% in non-athletes.

Nevertheless, elite wheelchair athletes are subject to high shoulder injury rates (11). In WB, the most reported disorders are rotator cuff impingement or a tear, biceps tendinopathy, and acromioclavicular joint pathology (12, 13). While it is unclear if these injuries are due to overhead movements, to wheelchair maneuvering, or (probably) to a combination of both, Mercer et al. (14) found in a previous study on the propulsion of standard MW, that specific components of shoulder load are associated to shoulder disorders:

1) increased external glenohumeral forces in posterior and lateral directions, and increased internal moments in flexion and adduction, are related to a higher prevalence of coracoacromial ligament edema and/or thickening that may lead to subacromial impingement and rotator cuff tear;
2) increased external glenohumeral forces in superior direction and increased internal moment in external rotation, are related to increased signs of symptomatic shoulder pathology.

To date, measurement of shoulder joint kinetics in WB athletes has been performed only in non-ecological conditions such as isokinetic testing (15). Therefore, the aim of this exploratory work is to measure the shoulder kinetics in WB athletes during the propulsion of a sports wheelchair in ecological conditions, and to compare these measurements to previous measurements in standard MW propulsion. This work focuses on two mobility aspects of WB: sprinting and dribbling. We assessed the impact of these tasks on both pushrim and shoulder kinetics, and more precisely: on the three components of the pushrim forces (tangential, radial, and mediolateral) and the propulsive moment, to obtain insight on the efficiency of the applied force during the complete push phase, on shoulder dynamics, to evaluate the effect of sprinting and dribbling in relation to the association between shoulder load and MSD described by Mercer et al. (14).

We hypothesized that shoulder load would be higher in sports wheelchair sprinting than in previous studies on standard MW propulsion. Moreover, in light of our previous results (16) where dribbling reduced the mean propulsive moments compared to sprinting, we hypothesized that dribbling would generally decrease the shoulder load.

# METHODS

## Participants

Ten WB athletes participated in this experiment. To be included, athletes could not have a current or recent (≤3 months) injury or pain that could interfere with their ability to carry out the tasks. The experimental protocol was approved by the Institutional Research Ethics Committee of Université du Québec à Montréal (UQAM) (certificate #CIEREH 2879_e_2018). This work is based on the same data as presented in Chénier et al. (16), except that participant #9 in the first study was replaced by participant #4 due to a problem with the motion capture device. Participant demographics are provided in **Table 1**, where participants are ordered by classification and by years of experience in WB.

## Tasks

After a personal 5-min warm-up, every participant performed 9-m sprints at maximal speed in a straight line from a stopped position on a wooden basketball court. Participants were asked to propel synchronously, with both arms pushing at the same time, in two conditions:

1) Classic Propulsion (CP), during four sprints, without a ball.
2) Dribble Propulsion (DP), during four sprints, where they were instructed to forward dribble. After two acceleration pushes, they had to push the ball forward, give one push on the wheels, recover the ball on the rebound, then place the ball on their knees, as described in Chénier et al. (16). They were asked to repeat this sequence until they had completed the 9-m distance.

A total of eight sprints was recorded: 2 conditions × 2 sides (right/left) × 2 repetitions. The order of the sprints was randomized, and participants were allowed to rest for a self-selected duration between trials.

## Instrumentation

### Kinetics

Participants used their own sports wheelchair equipped bilaterally with two instrumented wheels (SmartWheel). A wheel size of 25 or 26 inches was selected based on the participant's wheelchair. The instrumented wheels measured the propulsion forces and moments in 3D around the wheel hubs at 240 Hz. These wheels have a weight and moment of inertia of approximately 4.9 kg and 0.15 kg·m$^2$ (17). To limit the added resistance due to their increased weight, the SmartWheels' standard solid tires were switched to inflatable tires and fully inflated to 110 PSI.

**TABLE 1** | Participant demographics.

| Participant | Sex (M/F) | Age (years) | Dominant limb (R, L) | Disorder | Height (m) | Weight Weight (kg) | BMI (kg/m$^2$) | Experience (years) | Classification (1.0–4.5) | Belt/ strap | Wheel size (in) | Wheel camber (deg) |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| 1 | F | 31 | R | SCI T6-A | 1.60 | 61 | 23.8 | 3 | 1.0 | torso thigh | 25 | 20 |
| 2 | M | 60 | R | SCI D6-D7-A | 1.83 | 71 | 21.2 | 6 | 1.0 | torso thigh | 26 | 16 |
| 3 | M | 29 | L | CP | 1.68 | 60 | 21.3 | 10 | 1.0 | thigh | 25 | 19 |
| 4 | M | 40 | R | SCI T12-A | 1.75 | 66 | 21.6 | 12 | 1.0 | torso thigh | 25 | 22 |
| 5 | M | 34 | R | SCI T7-A | 1.50 | 73 | 32.4 | 10 | 1.5 | thigh | 26 | 18 |
| 6 | M | 33 | R | SCI T10-A | 1.76 | 95 | 30.7 | 1.5 | 2.0 | thigh | 26 | 19 |
| 7 | M | 32 | R | MD | 1.73 | 52 | 17.4 | 6 | 2.0 | thigh legs | 25 | 22 |
| 8 | M | 23 | R | SD | 1.63 | 58 | 21.8 | 11 | 2.0 | thigh | 26 | 17 |
| 9 | F | 30 | R | ND | 1.61 | 62 | 23.9 | 3 | 4.5 | thigh | 26 | 20 |
| 10 | M | 24 | R | ND | 1.78 | 78 | 24.6 | 16 | 4.5 | thigh | 26 | 20 |
| Mean (SD) | 7×M 3×F | 33.6 (10.5) | 9×R 1×L | / | 1.69 (0.10) | 67.6 (12.3) | 23.9 (4.5) | 7.9 (4.7) | 2.1 (1.4) | / | 4×25in 6×26in | 19.3 (1.9) |

*SCI, Spinal Cord Injury; CP, Cerebral Palsy; MD, Muscular Dystrophy; SD, Spastic Dysplasia; ND, Non-disabled.*

### Kinematics

An optoelectronic system consisting of 14 cameras (Prime13, Optitrack) was used to measure the participants' kinematics unilaterally. The cameras were arranged to build an acquisition volume that covered the entire sprint. The following landmarks were recorded at 120 Hz: second metacarpal distal heads, center of the hand, ulnar and radial styloid processes, lateral and medial elbow epicondyles, acromion, C7, T12, and both rear wheel centers. Landmarks that could not be followed directly due to occlusion (e.g., rear wheel center of the opposite side, medial elbow epicondyle) were reconstructed using rigid clusters of three to four markers affixed on the wheelchair, arms, and forearms. The position of the second metacarpal distal heads was not measured directly but was calculated using the styloid processes and hand markers. The rear wheel camber was measured using static kinematic acquisitions where different points of the wheels were probed and expressed relative to the wheelchair.

## Data Processing

### Kinetics

The dynamic offsets in the measured pushrim forces and moments due to the wheel camber were canceled as described in Chénier et al. (18). Synchronization between kinetics and kinematics was done at the beginning of each recording, by gently impacting the instrumented pushrim with a stick instrumented with a reflective marker. This impact was identified as a simultaneous event in both instruments: as a force spike in the kinetic data, and as a sudden stop of the marker's motion in the kinematic data.

### Kinematics

Marker positions were filtered at 10 Hz using a second-order, no-lag Butterworth filter. The definition of the coordinate system is provided in **Figure 1A**. The forearm and humerus coordinate systems were defined following the recommendations of the International Society of Biomechanics (ISB) (19), using both elbow epicondyles and both styloid processes, and approximating the glenohumeral joint by the acromion. Because of the flexed position adopted by some participants, the thorax could only be defined by markers in the back (T12 and C7). Since propulsion was synchronous, we considered that the thorax was not axially rotated, and therefore the y axis of the thorax was defined as the line from T12 to C7, and the yz plane of the thorax was defined by its y axis and the wheelchair's mediolateral axis. The coordinate systems of the wheel hubs were defined at the hub centers with their y and z axes inclined according to the wheel camber. All left side recordings were mirrored across the wheelchair mediolateral axis, and all subsequent data processing was considered right sided.

### Inverse Dynamics

A generic inverse dynamics method composed of four segments (wheel, forearm+hand, arm, thorax) was used to iteratively calculate the shoulder joint kinetics from the wheel's hub to the second metacarpal distal head, then to the elbow center and finally to the shoulder joint (20). Inertial characteristics (mass, moments of inertia) were personalized based on each

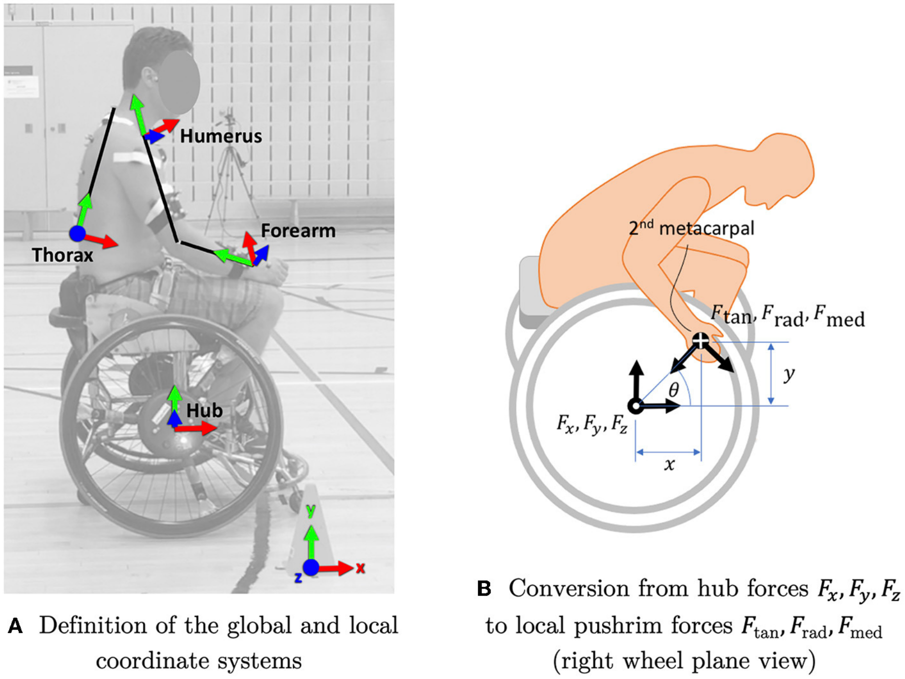


**FIGURE 1 | (A,B)** Coordinate system definitions.

participant's mass, sex and segments length, using inertial data compiled by Winter [(21), chap. 4. Anthropometry].

## Push Selection

For all conditions, pushes 1 and 2 were considered to be transitional and were discarded from the analysis. For the CP condition, all pushes after push 3 (included) were analyzed. For the DP condition, pushes performed while the ball was in the air, and deemed valid as described in section 3.2 were analyzed.

## Outcome Variables

Speed was defined as the speed reached at the end of the fourth push and was calculated based on the wheel angles, using a 131-point, first-order derivative Savitzky-Golay filter (22).

The total pushrim force and the three pushrim force components were calculated by converting the pushrim forces $F_x$, $F_y$ and $F_z$ and moment $M_z$ (measured in the non-rotating hub coordinate system), to the point of force application estimated by the position of the second metacarpal distal head as shown in **Figure 1B**:

- Total pushrim force: $F_{tot} = \sqrt{F_x^2 + F_y^2 + F_z^2}$
- Tangential pushrim force: $F_{\tan} = F_x \sin\theta - F_y \cos\theta$
- Radial pushrim force: $F_{rad} = -F_x \cos\theta - F_y \sin\theta$
- Medial pushrim force: $F_{med} = -F_z$
- Propulsive moment: $M_{prop} = M_z$

where $\theta$ is an angle in the wheel plane, between a horizontal line and a line from the wheel center to the projected second metacarpal distal head.

Shoulder forces and moments were expressed in the thorax coordinate system. The reported forces are external, i.e., a superior shoulder force means that the external reaction force pushes the humeral head upward relative to the thorax. The reported moments are internal and relate to the net muscular action at the shoulder joint.

For each analyzed push, the following outcome variables were calculated:

- Pushrim kinetics: peak values of $F_{tot}$, $F_{\tan}$, $-F_{\tan}$, $F_{rad}$, $-F_{rad}$, $F_{med}$, $-F_{med}$, $M_{prop}$ and $-M_{prop}$;
- External shoulder forces: peak values of anterior, posterior, superior, inferior, lateral and medial forces.
- Internal shoulder moments: peak values of flexion, extension, adduction, abduction, internal rotation, and external rotation moments.

## Statistical Analysis

For each outcome variable, data normality of the difference between both conditions was verified using a Shapiro-Wilk test with $\alpha = 0.05$. For data where normality was confirmed, parametric tests (paired $t$-tests) with $\alpha = 0.05$ were used to test for the mean difference between both propulsion conditions. Due to the exploratory nature of this work, significance thresholds were not corrected for multiple comparisons. The effect size was reported for every comparison using:

$$d = \frac{\text{mean}\,(x_{DP}) - \text{mean}\,(x_{CP})}{\text{s.d.}\,(x_{CP})}$$

and was interpreted using Cohen's recommendation: small ($d = 0.2$), moderate ($d = 0.5$) and large ($d = 0.8$) (23). For data that fail the Shapiro-Wilk normality test, non-parametric tests (Wilcoxon's signed rank tests) were used instead, and the effect size was calculated using the rank-biserial correlation.

For each condition, we also plotted typical profiles for the pushrim forces, shoulder forces and shoulder moments during the push. To reduce both the intra-participant and inter-participant variability, each assessed variable $x$ was first time-normalized from −25 to 125% of the push, and then amplitude-normalized using:

$$\text{normalized}\left(\mathrm{x}\left(\%\text{push}\right)\right) = \mathrm{x}\left(\%\text{push}\right) \times \frac{\overline{\left(\mathrm{A}_{\mathrm{p-p}}\right)}}{\mathrm{A}_{\mathrm{p-p}}}$$

where $A_{\mathrm{p-p}}$ is the peak-to-peak amplitude of $x$ for a given push cycle, and $\overline{A_{\mathrm{p-p}}}$ is the averaged $A_{\mathrm{p-p}}$ over every push of a given condition (CP, DP).

All calculations were performed with Python/SciPy using Kinetics Toolkit (24). Statistics were calculated using JASP 0.14.1.

### Comparison to Other Studies

The calculated shoulder kinetics were compared to results from 10 studies from 2001 and up that used a similar method (inverse dynamics with rigid bodies) to calculate the shoulder load during standard MW propulsion. To avoid comparing too different conditions, we only included results from non-elderly wheelchair users, without upper-limb impairment, who propelled a real wheelchair (as opposed to an integrated, custom ergometer). This resulted in a total of 10 studies, in which the participants propelled on rollers, treadmills or an ascending ramp at speeds from 0.8 to 2.2 m/s (14, 25–32).

# RESULTS

## Outcome Variables

**Table 2** shows the outcome variables and their comparison between both conditions. Individual results are also available as graphs in **Supplementary Material**.

### Pushrim Kinetics

In both CP and DP, the tangential and inward radial forces were the two most important force components. A braking moment and a negative tangential force were observed. Dribbling had no effect on the propulsive components of the pushrim kinetics (i.e., the peak tangential force and peak propulsive moment). However, dribbling generally reduced the peak negative tangential force in 9 of the 10 participants ($-5.6\ N, -27\%,\ p = 0.01,\ d < -0.8$). Dribbling also mainly reduced the non-propulsive force components: the peak lateral force decreased in 9 participants ($-4.9\ N,\ -44\%,\ p < 0.01,\ d < -0.8$), and the peak inward force decreased in 8 participants ($-26.6\ N,\ -17\%,\ p = 0.04,\ d = -0.77$), although at the expense of an increase in peak outward force in 7 participants ($+5.6\ N, +35\%,\ p = 0.01,\ d > 0.8$).

### Shoulder Forces

In the following sections, each main force/moment component is reported and compared to its maximal counterpart from the 10 studies indicated in section 3.4.7. The main external shoulder force was in the posterior direction (172 N), which is 87% higher than the same component measured by Kloosterman et al. (28) with 11 wheelchair users who propelled at 0.9 m/s on a treadmill (92 N). The second highest shoulder force component was in the anterior directions (118 N), which is 136% higher than the same component measured by Gil-Agudo et al. (26) with 16 wheelchair users who propelled at 1.1 m/s on a treadmill (50 N). Compared to sprinting, dribbling reduced the peak anterior force in 9 participants ($-27.4\ N,\ -23\%,\ p < 0.01,\ d < -0.8$), and the peak medial force 9 participants ($-18.6\ N,\ -30\%,\ p < 0.01,\ d < -0.8$).

### Shoulder Moments

The main net joint moment was in flexion (65 Nm), which is 64% higher than the same component measured by Sabick et al. (33) with 16 wheelchair users who propelled on a 20:1 ascending ramp (40 Nm). The second main moments were both in adduction and external rotation. Adduction (41 Nm) was 31% higher than the same component measured by Koontz et al. (29) with 27 individual with SCI who propelled at 1.8 m/s on rollers (21 Nm). External rotation (41 Nm) was 101% higher than the same component measured by Collinger et al. (25) in a multisite study with 61 wheelchair users who propelled at 1.8 m/s on rollers (21 Nm). Compared to sprinting, the main effect of dribbling on shoulder moments was in the transverse and sagittal planes. Dribbling reduced the peak internal rotation moment in 7 participants ($-4.74\ Nm,\ -20\%,\ p = 0.05,\ d = -073$).

## Kinetic Profiles

**Figure 2** shows the typical profile for the pushrim forces, shoulder forces and shoulder moments from −25 to 125% of the push. Both conditions have similar profiles. At the shoulder, external forces in posterior, inferior and lateral direction, and net moments in flexion, adduction and external rotation were observed during the push. The inverse was observed after releasing the pushrims. In DP, we observed a decreased inward radial pushrim force, which peaks at about 25% of the push in the CP condition compared to a plateau between 30 and 65% of the push in the DP condition. We also observed a decreased anterior shoulder force and a decreased shoulder moment of internal rotation during the transition from push to recovery, at about 110% of the push. Finally, we observed a decreased slope for each pushrim force component, a delayed anterior shoulder force and a delayed shoulder moment of flexion.

# DISCUSSION

The aim of this work was to assess the effect of sprinting and dribbling using a sports wheelchair on the different components of the pushrim and shoulder kinetics. Compared to previous studies on standard MW propulsion, the shoulder load is much higher, independently of the CP or DP condition. Obviously, difference in speeds between these studies and ours most

**TABLE 2 |** Comparison of outcome measures between both conditions.

| | CP | | DP | | Diff | | *p* | d | np | Standard MW[a]. |
|---|---|---|---|---|---|---|---|---|---|---|
| Speed (m/s) | 2.57 | (0.32) | 2.39 | (0.31) | −0.18 | (0.16) | **0.007** | **−1.11** | | |
| **Peak pushrim kinetics** | | | | | | | | | | |
| Total force $F_{tot}$ (*N*) | 215.8 | (46.7) | 202.0 | (44.3) | −13.7 | (24.2) | 0.11 | **−0.57** | | |
| Forward tangential force $F_{tan}$ (*N*) | 146.9 | (41.9) | 140.1 | (32.5) | −6.7 | (19.1) | 0.30 | −0.35 | | |
| Negative tangential force $-F_{tan}$ (*N*) | 21.0 | (7.4) | 15.5 | (6.4) | −5.6 | (4.9) | **0.01** | **−0.86** | | |
| Inward radial force $F_{rad}$ (*N*) | 160.3 | (37.1) | 133.7 | (49.3) | −26.6 | (34.5) | **0.04** | **−0.77** | | |
| Outward radial force $-F_{rad}$ (*N*) | 15.8 | (11.6) | 21.5 | (11.6) | 5.6 | (5.2) | **0.01** | **1.09** | | |
| Medial force $F_{med}$ (*N*) | 86.4 | (35.7) | 90.1 | (32.7) | 3.7 | (10.8) | 0.31 | 0.34 | | |
| Lateral force $-F_{med}$ (*N*) | 11.2 | (7.7) | 6.3 | (4.6) | −4.9 | (4.2) | **0.005** | **−1.17** | | |
| Propulsion moment $M_{prop}$ (Nm) | 37.2 | (9.5) | 35.80 | (8.35) | −1.42 | (4.40) | 0.70 | −0.16 | * | |
| Braking moment $-M_{prop}$ (Nm) | 5.1 | (1.7) | 4.25 | (1.52) | −0.82 | (1.38) | 0.08 | **−0.64** | * | |
| **Peak shoulder forces** | | | | | | | | | | |
| Anterior (*N*) | 118.4 | (24.9) | 91.0 | (27.9) | −27.4 | (21.8) | **0.003** | **−1.25** | | 5–50 |
| Posterior (*N*) | 171.8 | (42.4) | 157.6 | (35.3) | −14.2 | (27.4) | 0.16 | **−0.53** | | 27–92 |
| Superior (*N*) | 60.1 | (19.1) | 64.3 | (15.4) | 4.2 | (14.2) | 0.38 | 0.30 | | −16–108 |
| Inferior (*N*) | 79.8 | (22.8) | 88.1 | (44.4) | 8.3 | (32.7) | 0.45 | 0.25 | | −33–58 |
| Lateral (*N*) | 66.6 | (30.3) | 61.5 | (28.3) | −5.1 | (11.0) | 0.18 | **−0.46** | | 7–50 |
| Medial (*N*) | 66.5 | (23.4) | 48.0 | (20.7) | −18.6 | (11.7) | **<0.001** | **−1.59** | | 7–15 |
| **Peak shoulder moments** | | | | | | | | | | |
| Flexion (Nm) | 65.3 | (17.8) | 58.78 | (13.4) | −6.49 | (9.8) | 0.07 | **−0.66** | | 6–40 |
| Extension (Nm) | 31.0 | (8.5) | 24.61 | (11.4) | −6.40 | (15.2) | 0.22 | −0.42 | | 5–17 |
| Adduction (Nm) | 40.6 | (13.0) | 40.72 | (15.6) | 0.11 | (6.2) | 0.56 | 0.24 | | 0–31 |
| Abduction (Nm) | 30.3 | (12.8) | 21.03 | (9.8) | −9.24 | (10.8) | 0.85 | 0.00 | | 0–12 |
| Internal rotation (Nm) | 23.8 | (11.7) | 19.02 | (12.9) | −4.74 | (6.5) | **0.05** | **−0.73** | | 0–21 |
| External rotation (Nm) | 41.3 | (17.3) | 38.83 | (15.0) | −2.48 | (6.8) | 0.28 | −0.36 | | 0–21 |

*Parentheses, standard deviation; d, effect size; np, non-parametric test.*
*Bold and underlined p-values indicate* ***p < 0.05*** *and* ***p < 0.01****.*
*Bold and underlined d-values indicate* ***moderate*** *(0.5) to* ***large*** *(>0.8) effect sizes.*
**Non-parametric test.*
[a] *Peak shoulder kinetics ranges from previous studies on standard MW propulsion on treadmill or rollers from 0.8 to 2.2 m/s (14, 25–33).*

probably account for these large differences. However, while the participants in our study propelled only 0.4 m/s faster than in Mulroy et al. (31), with 2.6 m/s compared to 2.2 m/s, the posterior shoulder force was 130% higher (172 vs. 75 Nm) and the flexion moment was 170% higher (65 vs. 24 Nm). Apart from the superior and inferior shoulder force that varies a lot between studies, every shoulder force of moment component was 13% to 346% higher than its highest counterpart in every other study. In addition to the wheelchair's geometry and user's position that are different between standard and sport wheelchairs, these large differences in shoulder kinetics may be explained by two reasons. The first reason is that every of these other studies were performed during continuous propulsion on rollers or treadmill, whereas our study was conducted on the ground. While our conditions were more ecologically valid, the athletes did not completely reach their maximal velocity after only two pushes; the remaining acceleration requires higher propulsion moments. The second reason relates to the limbs' inertia. Since the speed was higher in our study, the joint forces required for accelerating and decelerating the limbs were also higher. The effect of these inertial components can be seen in **Figure 2**, where immediately after the push, no force is applied on the wheel, but important shoulder forces and moments can still be observed, especially in the anterior shoulder force and in the shoulder moments of flexion, adduction and external rotation.

The values found in this work are generally high and may be worrisome. For example, the peak posterior shoulder force was 172 N, compared to 42 N in Mercer et al. (14) who correlated such high values to an increased risk of coracoacromial ligament disorder. Moreover, the peak shoulder moment of external rotation was 41 N, compared to 9 N in Mercer et al. (14) who correlated such high values to symptomatic shoulder pathology. This raises a flag on the intensity of propelling in WB compared to everyday propulsion. However, in WB, half of the game time is spent coasting or resting, and a rather small percentage of the time is performed sprinting (9%) or dribbling (<1%) (34). Thus, we believe that the causes of shoulder disorders could not only be associated with sprinting or dribbling, but most probably to a combination of tasks such as accelerating, challenging/handling the ball, and sprinting. However, this high load should be considered when planning training sessions to avoid overloading the shoulder.

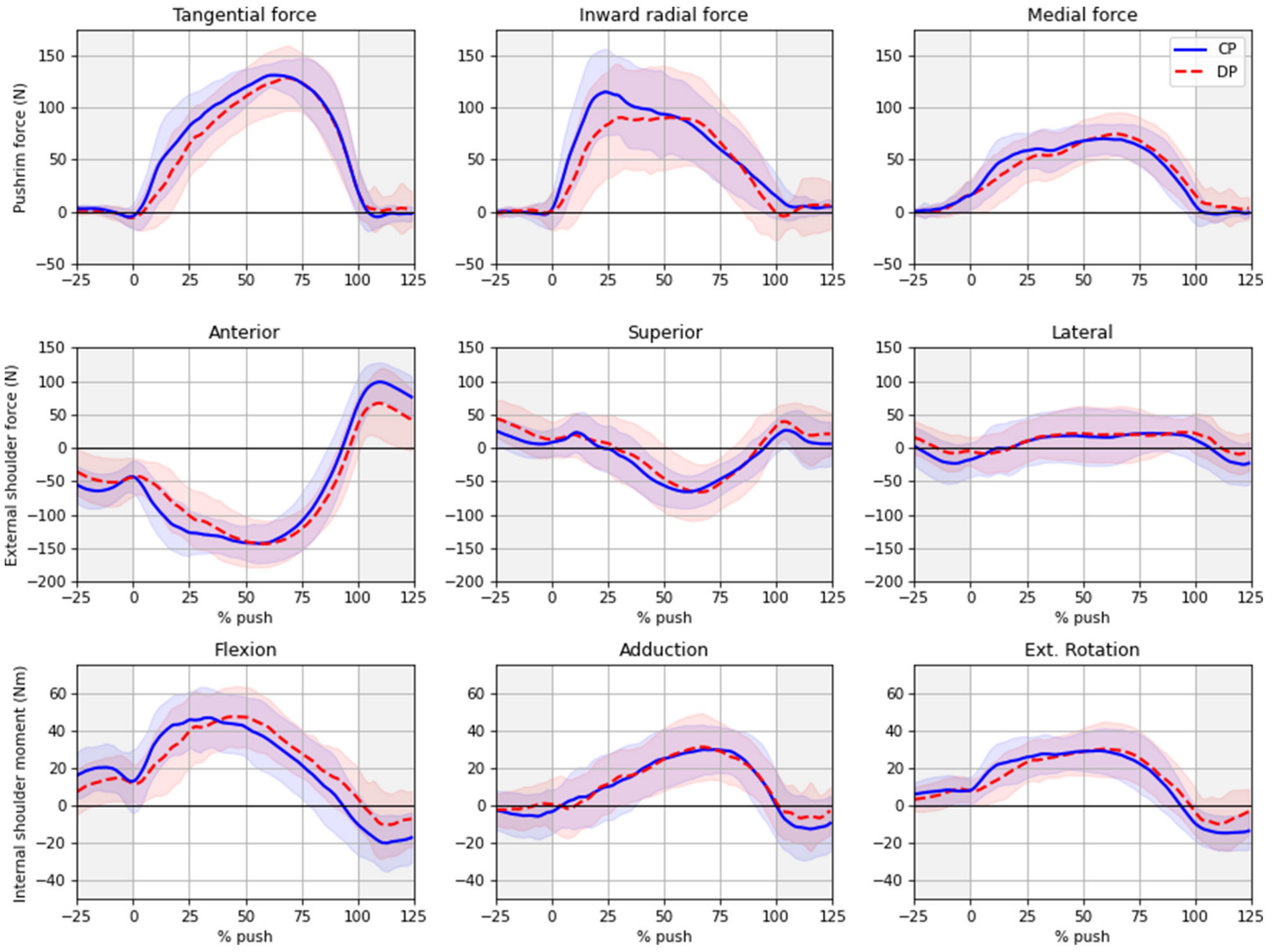


**FIGURE 2 |** Kinetic profiles during CP and DP conditions.

When comparing CP and DP, dribbling reduced every peak force value except the positive tangential and medial forces. Dribbling also reduced the peak negative tangential forces. This combined reduction in peak force components is viewed as a beneficial change in terms of push efficiency. However, these differences may be attributable to the reduced speed observed during dribbling: similar relationships between speed and pushrim kinetics have been observed in a study by Kwarciak et al. (35) where 54 participants with paraplegia who propelled their own wheelchair on rollers increased the amplitude and the number of occurrences of negative moments as speed increased. In terms of shoulder kinetics, we expected that dribbling would decrease the shoulder load. We indeed observed a reduction in the peak posterior shoulder force, such a component being associated with coracoacromial ligament edema or thickening in standard MW propulsion (14). Since dribbling was not associated with other specific kinetic components related to shoulder disorders, this suggests that propelling while dribbling may be less detrimental to the shoulder joint than sprinting.

In this work, we chose to refer to the shoulder moments in the thorax reference frame to be consistent with Mercer et al. (14). However, special care must be taken in interpreting the results in this reference frame, especially shoulder moments of internal/external rotation. When the arms are not elevated, the reported moments of rotation in either the thorax or humeral reference frame are similar because the humeral and thorax longitudinal axes are nearly coincident. However, when the arm is more elevated like it is in sports wheelchair propulsion, the reported shoulder rotation moments may include significant crosstalk (moments from other axes). For example, for a 90-degree abduction, a moment reported in the thorax reference frame as an external rotation would be better understood as a moment of horizontal abduction. This example highlights the difficulty of comparing shoulder load between tasks that are kinematically different. Currently, there is no consensus on the best way to report shoulder kinetics. Some authors (including those of this work) reported shoulder kinetics in the thorax reference frame (14). Others reported the forces in the thorax frame but the moments in the humeral frame (25), while others used four components instead of three, with three standard anatomical axes associated with the thorax reference frame, and an additional axis (the humerus longitudinal axis) to express humeral rotation moments (29, 36). Research is still needed to define what axes are the best axes to report shoulder kinetics as a function of the studied task.

Among the limits of the study, we note the limited number of participants and their variety of disorders and classifications. However, since the observed differences between both conditions and between previous literature were generally large, we believe that this work allowed much needed insight to be gained on the impact of wheelchair propulsion in WB on shoulder load. Another limitation is the evaluation of the associated risks of shoulder disorders using the work of Mercer et al. (14) who assessed these risks for standard MW propulsion on rollers, not for sports wheelchair propulsion on a basketball court. It is therefore important to consider these comparisons as indicative and not as a direct relationship between propulsion and specific shoulder disorders. Finally, using SmartWheel instrumented wheels increased the rolling resistance and wheelchair inertia due to their added weight. We however limited this effect by using fully inflatable tires instead of the standard solid Smart Wheel tires.

As highlighted in this work, the differences between sprinting and dribbling on shoulder load seem much lower than the differences between everyday propulsion and sports propulsion. Consequently, we believe that including dribbling sessions in addition to sprinting sessions during training should not be riskier for the shoulder, which supports our previous conclusions based on spatiotemporal and generic pushrim kinetic parameters (16). Future work should reproduce a similar analysis to other tasks found in WB, such as accelerating, changing direction, and challenging and handling the ball, which would increase our understanding of the risks of MSD associated with WB.

## DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

## ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Université du Québec à Montréal's Comité institutionnel d'éthique de la recherche avec des êtres humains (CIEREH). The patients/participants provided their written informed consent to participate in this study.

## AUTHOR CONTRIBUTIONS

FC contributed to the study design, to the data processing and is the main author of the manuscript. IA contributed to data processing. DG and AF contributed to the study design. All authors contributed to manuscript revision, read, and approved the submitted version.

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## SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fresc.2022.863093/full#supplementary-material

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# Influence of Wheelchair Type on Kinematic Parameters in Wheelchair Rugby

*Sadate Bakatchina[1]*, Thierry Weissland[2], Florian Brassart[1], Ilona Alberca[1], Opale Vigie[1], Didier Pradon[3] and Arnaud Faupin[1]*

[1] Laboratory Physical Activity Impact on Health (IAPS), University of Toulon, Toulon, France, [2] Laboratory of Material to System Integration (IMS), University of Bordeaux, Pessac, France, [3] Pole Parasport - ISPC Synergies, CHU Raymond Poincaré, APHP, Garches, France

**Introduction:** In wheelchair rugby, players use either an offensive or defensive wheelchair depending on their field position and level of impairment. Performance of wheelchair rugby players is related to several parameters, however it is currently unclear if differences in performance are related to wheelchair type or no: the effect of wheelchair type on performance variables has not been evaluated. The aim of this study was to compare offensive and defensive wheelchairs on performance variables during a straight-line sprint.

**Methods:** Thirteen able-bodied people performed two 20 m sprint trials: one with an offensive and one with a defensive wheelchair. Data were collected using inertial measurement units fixed on the wheelchair. Peak wheelchair velocities and left-right asymmetries in peak wheel velocities were measured during the acceleration and constant peak velocity phases. Sprint time, cycle frequency, and mean and maximum velocity were calculated over the entire sprint.

**Results:** The peak velocities of the first 2 pushes (acceleration phase) were significantly higher with the defensive than the offensive wheelchair ($p < 0.04$ and $p < 0.02$). Mean and maximum sprint velocity were significantly higher ($p < 0.03$ and $p < 0.04$, respectively) with the defensive wheelchair. Cycle frequency and asymmetry did not differ between wheelchairs.

**Conclusion:** Performance was higher with the defensive than the offensive wheelchair, suggesting that the frequent finding that the higher performance of offensive as compared to defensive players is not related to the use of an offensive wheelchair.

**Keywords: wheelchair rugby, sprint, peak velocity, asymmetry, inertial measurement unit**



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*Dirkjan Veeger, Delft University of Technology, Netherlands*

***Reviewed by:***
*Marc Klimstra, University of Victoria, Canada*
*Rowie Johanna Francisca Janssen, University of Groningen, Netherlands*

****Correspondence:***
*Sadate Bakatchina*
*sadate.bakatchina@univ-tln.fr*

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## INTRODUCTION

Wheelchair rugby is a high-performance team sport which was included in the Paralympic program in 2000. Wheelchair rugby players have different types of disabilities (IWRF International wheelchail rugby federation, 2021) that may result from conditions such as spinal cord injury, amputation, polio, cerebral palsy, peripheral neuropathy, or congenital limb deficiency (Gee et al., 2018; Bakatchina et al., 2021a; IWRF International wheelchail rugby federation, 2021). For training and during matches, players are often grouped according to their level of impairment: high point (HP), mid-point (MP) and low point (LP) (IWRF International wheelchail rugby federation, 2021). LP players have a low level of physical ability whereas HP players have a high level of physical ability (IWRF International wheelchail rugby federation, 2021). MP players have intermediate level of physical ability. Studies have classified wheelchair rugby players into two groups: LP and HP (Goosey-Tolfrey et al., 2018) or three groups: LP, MP and HP (Usma-Alvarez et al., 2014; Rhodes et al., 2015a; Haydon et al., 2016, 2018a). Others classified players according to the type of wheelchair used during the game: offensive and defensive players (Bakatchina et al., 2021a). The wheelchairs used during matches have been designed for use by players with different levels of impairment. Offensive wheelchairs (OW) have a front bumper to prevent other wheelchairs from hooking them during the game; defensive wheelchairs (DW) have a bumper that allows them to hook and hold other wheelchairs. OWs are shorter and heavier than DWs (Haydon et al., 2016), consequently OW and DW can be differentiated by the mass distribution. LP players use DW and HP players use OW; MP players can use either type, depending on the coach's strategy.

Comparison of wheelchair rugby players using OW or DW found that those who used an OW achieved higher peak velocities during the acceleration and constant peak velocity phases than those who used a DW (Bakatchina et al., 2021a). However, cycle frequency, which is an indicator of injury risk (Boninger et al., 1999), was higher in players using an OW than those who used a DW (Bakatchina et al., 2021a). According to Boninger et al. (1999), gesture repetition such as cycle frequency during manual wheelchair propulsion would more expose the wheelchair users to risks of injury to their upper limbs. The literature indicated that performance in wheelchair rugby players is related to several parameters such as: players' classification (Sarro et al., 2010; Rhodes et al., 2015a,b; Goosey-Tolfrey et al., 2018), training hours (Furmaniuk et al., 2010; Berzen and Shayke Hutzler, 2012) experience in wheelchair using, gender and age. In addition, the performance during wheelchair manual propulsion is related to the rolling resistances which are the forces that oppose wheelchair displacement causing wheelchair deceleration (Sauret et al., 2009). Thus, wheelchair velocity decreases during wheelchair deceleration (Sauret et al., 2009) impacting player's performance in terms of sprint time during straight-line sprint.

HP players are faster and achieve higher peak power and peak velocity compared to LP players during a 15 s sprint on an instrumented ergometer (Goosey-Tolfrey et al., 2018). However, HP players have higher left-right asymmetry in peak wheel velocity (Goosey-Tolfrey et al., 2018). During matches, HP players achieve higher velocities than LP and MP players (Rhodes et al., 2015a,b) and they spend more time performing high-intensity activities and cover higher distances during the game (Rhodes et al., 2015b). Furthermore, the rate of decrease in velocity between the first and second halves of the match is lower in HP than LP players (Sarro et al., 2010). Wheelchair configuration parameters influence performance, for example camber angle, seat height, seat depth and wheel diameter (Faupin et al., 2004; Mason et al., 2011, 2012, 2013). Larger camber angle is associated with higher power generation (Faupin et al., 2004; Mason et al., 2011) and lower velocities during straight-line wheelchair propulsion (Faupin et al., 2004). Large diameter wheels increased 20 m sprint time and maximum velocity compared to small diameter wheels (Mason et al., 2012).

During a wheelchair rugby game, the ability of players to sprint, pivot, and brake while dribbling or holding the ball are key performance variables. During counter-attacks, players must sprint in a straight line. This important ability can be evaluated using the straight-line sprint test (Gee et al., 2018; Haydon et al., 2018b; Bakatchina et al., 2021a). Performance on the test can be evaluated by measuring kinematic variables such as velocities, accelerations and cycle frequencies (Gee et al., 2018; Bakatchina et al., 2021a). Analysis of these variables during the acceleration and constant peak velocity phases (Haydon et al., 2018b; Bakatchina et al., 2021a) is useful when determining the attributes of a wheelchair. Only Bakatchina and collaborators evaluated peak velocities during the acceleration and constant peak velocity phases of a 20 m straight-line sprint on the field; they found that players using an OW achieved higher peak velocities compared to players using a DW. To our knowledge, no study has investigated the specific influence of wheelchair types (OW or DW) during the acceleration and constant peak velocity phases of a 20 m straight-line sprint on the field in wheelchair rugby. However, it is unclear if the difference in performance was related to the wheelchair type or no. Given that the wheelchair is one of the most important parameters of performance in wheelchair sport (Goosey-Tolfrey, 2010), it is important to analyze the impact of the type of wheelchair on kinematic performance variables. This will serve both to optimize wheelchairs and to guide coaches in their allocation of different wheelchair types to different players. To evaluate the specific effects of OW and DW on performance parameters, the inclusion of able-bodied people is important

**TABLE 1 |** Individual anthropometric characteristics: gender, age, mass, and height.

| | Gender | Age (years old) | Mass (kg) | Height (cm) |
|---|---|---|---|---|
| AB1 | M | 20 | 67 | 173 |
| AB2 | M | 20 | 72 | 179 |
| AB3 | M | 21 | 69 | 175 |
| AB4 | F | 20 | 59 | 163 |
| AB5 | F | 22 | 55 | 162 |
| AB6 | F | 23 | 68 | 165 |
| AB7 | F | 21 | 52 | 163 |
| AB8 | M | 23 | 70 | 185 |
| AB9 | F | 21 | 66 | 172 |
| AB10 | M | 20 | 77 | 176 |
| AB11 | F | 21 | 63 | 170 |
| AB12 | F | 22 | 80 | 181 |
| AB13 | F | 21 | 50 | 164 |
| M (Q1; Q3) | 8F; 5M | 21 (20; 22) | 67 (59; 70) | 172 (164; 176) |

*M (median), Q1 (first quartile) and Q3 (third quartile).*

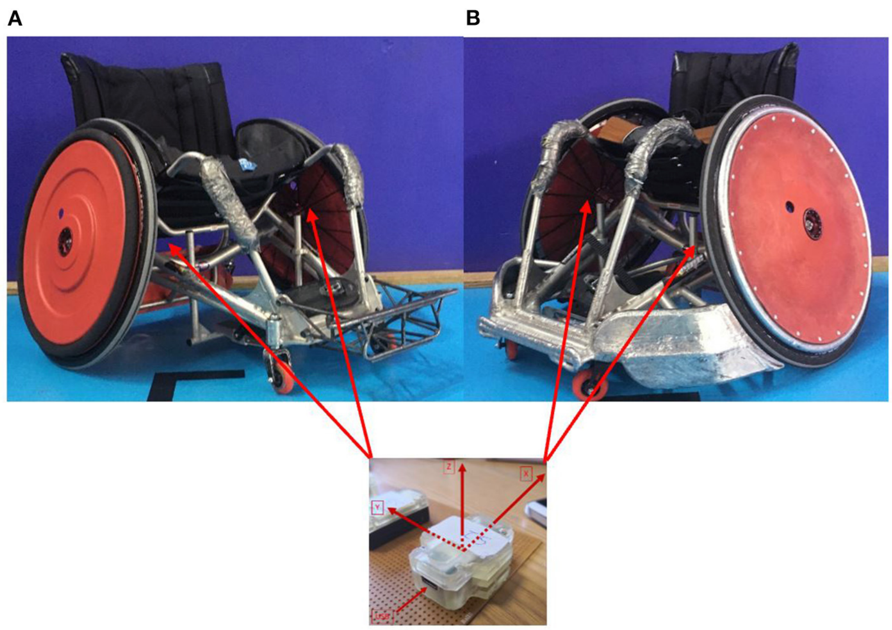


**FIGURE 1 |** Location of the inertial measurement units on the wheels. **(A)** Defensive wheelchair. **(B)** Offensive wheelchair (2).

**TABLE 2 |** Wheelchairs characteristics.

| | Mass (kg) | Seat angle (°) | Camber angle (°) | Frame length (m) | seat length (m) | Wheel diameter (inch) |
|---|---|---|---|---|---|---|
| OW | 21.8 | 22.4 | 18 | 0.53 | 0.38 | 25 |
| DW | 20.7 | 26.4 | 18 | 0.68 | 0.36 | 25 |

*OW, Offensive wheelchair; DW, Defensive wheelchair.*

because they are not yet used to a DW or an OW, so they are unbiased participants.

Consequently, the aim of this study was to compare kinematic variables between OW and DW during the acceleration and constant peak velocity phases of a 20 m straight-line sprint, using IMUs. We hypothesized that: (i) peak velocities during the acceleration and constant peak velocity phases would be higher with the OW, (ii) asymmetry and cycle frequency across the whole sprint would be higher with the OW, exposing the user at risk of injury, and (iii) The rolling resistance would be greater with the DW.

## MATERIALS AND METHODS

### Participants

A total of 13 able-bodied adults (7 females and 6 males) (**Table 1**) trained in wheelchair propulsion (see below) were included. None had experienced any upper limb injuries or pain within 6 months preceding the study. All participants were informed of the purpose of the study and any risks that may arise during the test; they all provided informed consent. We chose to perform this study in able-bodied people because we wished to evaluate the specific effects of wheelchair type without the confounding factor of disability; furthermore, studies have shown that trained able-bodied people provide consistent results in experiments using manual wheelchairs (van der Woude et al., 2003; Faupin et al., 2008). The study was approved by the National Ethics Committee for Research in the Physical Activity and Sports Sciences (CERSTAPS N° 2018-16-07-26).

### Wheelchairs

According to Haydon et al. (2016), there are two typical wheelchairs: OW & DW. All participants included in current study used one typical DW (**Figure 1A**) and one typical OW (**Figure 1B**). The OW weighed 21.8 kg, had a camber angle of 18° and 25-inch wheels (**Table 2**). The DW weighed 20.7 kg, had an 18° camber and 25-inch wheels (**Table 2**). We measured frame length, seat length and seat angle (**Table 2**) according to Haydon et al. (2016). We checked the function of the front casters and rear wheels of each wheelchair before testing.

### Inertial Measurement Unit (IMU)

IMUs are composed of a gyrometer, an accelerometer and a magnetometer which, respectively, allow the measurement of rotational velocity (Usma-Alvarez et al., 2011; van der Slikke et al., 2016; Bakatchina et al., 2021a,b), acceleration (Usma-Alvarez et al., 2011; van der Slikke et al., 2016; Haydon et al., 2018b) and orientation with respect to the magnetic north. We used 2 IMU: 128 Hz, 3 × 3 (accelerometer, gyrometer, magnetometer, and Bluetooth module, WheelPerf System, AtoutNovation, France) (**Figure 1**) and synchronized them with

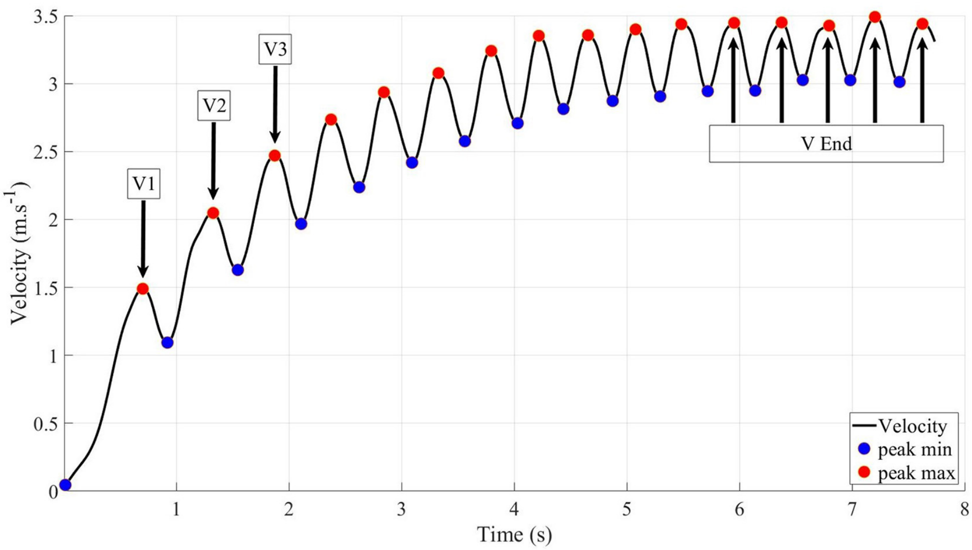


**FIGURE 2 |** The first three peak velocities (V1, V2, and V3) on the acceleration phase and five peak velocities (V End) on the constant peak velocity phase (2).

a tablet computer using Bluetooth version 4.0 technology as described by Bakatchina et al. (2021b).

## Test Protocol

Prior to the test, participants underwent five 2 h training sessions in wheelchair propulsion. We followed the description by Alberca et al. (2021), thus the training included: forward, backward and slalom propulsion over 5, 10, and 20 m. At the beginning of each training session, participants performed a 5–10 min warm-up consisting of forward and backward propulsion and repeated sprints over 20 m. After the warm-up, the participants practiced propelling the wheelchair in a straight line (forward and backward) and around a slalom course at different speeds over 5, 10, and 20 m using both types of wheelchairs. Just before the test, they warmed up for 8–10 min as described by Bakatchina et al. (2021a). They then performed one maximum velocity 20 m sprint with OW and one maximum velocity 20 m sprint with DW, recovery time between both sprints was 10 min. A standing start was used (participants started 20 cm from the starting line). No instructions were given regarding trunk movement during the sprint. Participants sprinted up to the finish line and slowed after crossing the line. The tests were performed in a sports hall on parquet flooring. The same OW and DW were used by all participants and the order of the wheelchairs was randomized.

Rolling resistance tests were then performed with a 20 kg mass placed on the front (first condition) and the rear (second condition) of each wheelchair type seat as described by Bascou et al. (2019). For each condition (forwards and backwards), six trials of deceleration test were performed with each wheelchair type over 5 m. Trials were performed by the experimenter who pushed the wheelchair and stopped it manually on 5 m. Each deceleration test was performed as reported by Bascou and collaborators: "(1) 2 s static phase on a departure mark fixed on the ground, (2) clean manual push to accelerate the manual wheelchair between two 1 m-separated marks, (3) deceleration while verifying the straightness of the manual wheelchair path, (4) clean manual stop between two ending marks, (5) 2 s static phase" (Bascou et al., 2019).

## Data Processing

We placed one IMU on each rear wheel (**Figure 1**) as described by Bakatchina et al. (2021a). They were positioned between two spokes near the hub and aligned vertically with respect to the horizontal axis of the wheel plane, with the z-axis perpendicular to the vertical axis of the wheel plane. We calculated the rotational velocity of the wheel around the z-axis as described by Fuss (2012) using the gyrometer data. To remove random noise, we used a Butterworth filter (fourth-order zero lag: low-pass-filtered) (Cooper et al., 2002; Bergamini et al., 2015) with a cut-off frequency of 8 Hz (Bakatchina et al., 2021a).

We used the finder function of the Matlab program to identify the minimum and maximum peaks on the rotational velocity curve as described by Bakatchina et al. (2021a). Kinematic parameters were calculated during the acceleration and constant peak velocity phases. The acceleration phase was defined as the first 3 pushes and the constant peak velocity phase as the last five pushes (Bakatchina et al., 2021a) (**Figure 2**).

We calculated the following performance variables according to Bakatchina et al. (2021a): the peak velocity of each of the first 3 pushes (V1, V2, and V3), the mean velocity of the last 5 pushes (Vend) during the constant peak velocity phase, sprint time, and mean and maximum velocity over the whole sprint. Cycle frequency (F) was defined as the number of cycles per minute and asymmetry (Asy) as the difference between the peak velocities of the right and the left wheels (Equation 1) (Goosey-Tolfrey et al., 2018; Bakatchina et al., 2021a).

$$Asy = \frac{|\text{V dh} - \text{V non} - \text{dh}|}{\text{V non} - \text{dh}} \times 100\ (\%) \tag{1}$$

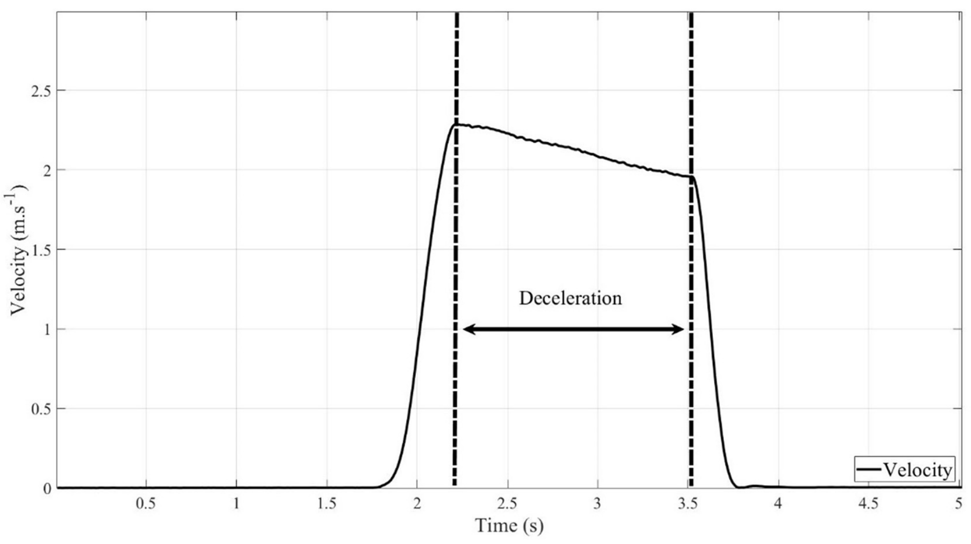


**FIGURE 3 |** Deceleration on the velocity curve.

where Asy: asymmetry; V dh: peak velocity of the dominant hand; V non-dh peak velocity of the non-dominant hand (2). Dominant hand was the hand that achieved higher peak velocity and non-dominant hand achieved lower velocity.

During propulsion, drag force (DF) is composed of rolling resistance forces (RRF), air resistance forces (ARF), gravitational forces (GF), and internal frictional forces (IFF) (Equation a) (van der Woude et al., 2001; Rietveld et al., 2021). ARF, GR and IFF are negligible (Equation b) as indicated by Rietveld et al. (2021). We calculated deceleration values (**Figure 3**) by deriving the linear velocity (c) of the wheels. We then calculated rolling resistance according to Equation (d).

$$\mathrm{DF} = \mathrm{RRF} + \mathrm{ARF} + \mathrm{GR} + \mathrm{IFF} \qquad (a)$$

$$\mathrm{DF} = \mathrm{RRF} = \mathrm{m.a} \qquad (b)$$

$$\mathrm{a} = \frac{\mathrm{d(v)}}{\mathrm{t}} \qquad (c)$$

$$\mathrm{RRF} = |\mathrm{m.a}| \qquad (d)$$

where m: mass of the wheelchair and the 20 kg additional masses; a: deceleration value; v: linear velocity.

Sprint times were also collected using cell gates (Brower Timing Systems, WITTY.GATE). The cell gates were placed at the start and finish lines connected to an electronic timer allowing to display the time of each sprint after.

## Statistical Analyses

We used velocity data from the (Bakatchina et al., 2021a) and G*Power 3.1 software to determine the minimum number of participants required for this study. This minimum number found is 8, which is less than the number of participants included in our study. Because the distribution of the variables was not normal according to the Lilliefors normality test, we calculated medians (M) and first (Q1) and third (Q3) quartiles. We compared variables between the OW and DW using the Wilcoxon test. We calculated effect sizes for all variables: low ($r < 0.3$), medium ($0.3 < r < 0.5$), and large ($r \geq 0.5$). STATISTICA version 7.1 was used for all statistical analyses and $p < 0.05$ was considered as statistically significant.

# RESULTS

Sprint times were faster with DW than OW (**Table 3**). Significant differences were found in terms of velocities between both wheelchairs (**Figure 4**). Peak velocity values V1 and V2 were higher with DW (**Table 3**). Mean (Vav) and maximum velocity were significantly higher with DW (**Table 3**). The values of V3 and Vend did not differ significantly between wheelchairs (**Table 3**). The magnitude of these effects was either medium or large (range from 0.44 to 0.88).

In addition, neither cycle frequency nor asymmetry during the acceleration and constant peak velocity phases differed between the wheelchairs (**Table 3**).

Rolling resistance values differed significantly between OW and DW for each condition. For both the first condition (additional mass placed on the front of the seat) and second condition (additional mass placed on the rear of the seat), rolling resistance was significantly higher with DW compared to the OW, with large effect sizes (**Table 3**).

# DISCUSSION

Performance in wheelchair rugby players is related to several such as: players level of impairment, experience in a wheelchair using, training hours, physical and technical capacities, gender, and age. Currently, it is unclear if differences in performance between rugby wheelchair players could be related to their wheelchair type or no. Consequently, we investigated the influence of an

**TABLE 3 |** M (median), Q1 (first quartile) and Q3 (third quartile) of velocities (m.s$^{-1}$); time (s); asymmetry (%): relative difference in velocity between the left and right side; F (cycle.min$^{-1}$): Cycle Frequency and RR (N): rolling resistance.

| | | OW M (Q1; Q3) | DW M (Q1; Q3) | *p* | *r* |
|---|---|---|---|---|---|
| Velocity | V1 | 1.83 (1.75; 1.98) | 2.00 (1.82; 2.09) | 0.04* | 0.55 |
| (m.s-1) | V2 | 2.36 (2.17; 2.57) | 2.59 (2.33; 2.69) | 0.02* | 0.63 |
| | V3 | 2.72 (2.48; 2.89) | 2.94 (2.80; 3.15) | (0.11) NS | 0.44 |
| | Vend | 3.75 (3.43; 4.04) | 3.91 (3.56; 4.21) | (0.07) NS | 0.49 |
| | Vmax | 3.84 (3.60; 4.30) | 4.09 (3.67; 4.37) | 0.03* | 0.59 |
| | Vmean | 2.70 (2.42; 2.82) | 2.74 (2.59; 3.06) | 0.04* | 0.55 |
| Time (s) | T | 7.42 (7.12; 8.26) | 7.31 (6.57; 7.72) | 0.04* | 0.56 |
| Asymetries (%) | Asy1 | 4 (3; 5) | 6 (3; 8) | (0.27) NS | 0.30 |
| | Asy2 | 3 (1; 7) | 3 (2; 6) | (0.80) NS | 0.07 |
| | Asy3 | 5 (3; 6) | 4 (1; 6) | (0.13) NS | 0.42 |
| | Asy end | 3 (3; 4) | 2 (2; 3) | (0.08) NS | 0.47 |
| Cycle frequency (cycle.min$^{-1}$) | F | 94.74 (85.80; 105.83) | 94.32 (84.04; 101.83) | (0.88) NS | 0.16 |
| Rolling resistance (*N*) | RR front | 6.26 (5.35; 7.05) | 9.34 (8.17; 9.74) | 0.002** | 0.88 |
| | RR rear | 5.57 (5.05; 6.27) | 7.88 (6.64;8.5) | 0.004** | 0.81 |

*DW, defensive wheelchair; OW, offensive wheelchair. Significant differences (*p < 0.05, **p < 0.01). No significant difference (NS). Effect size (r).*

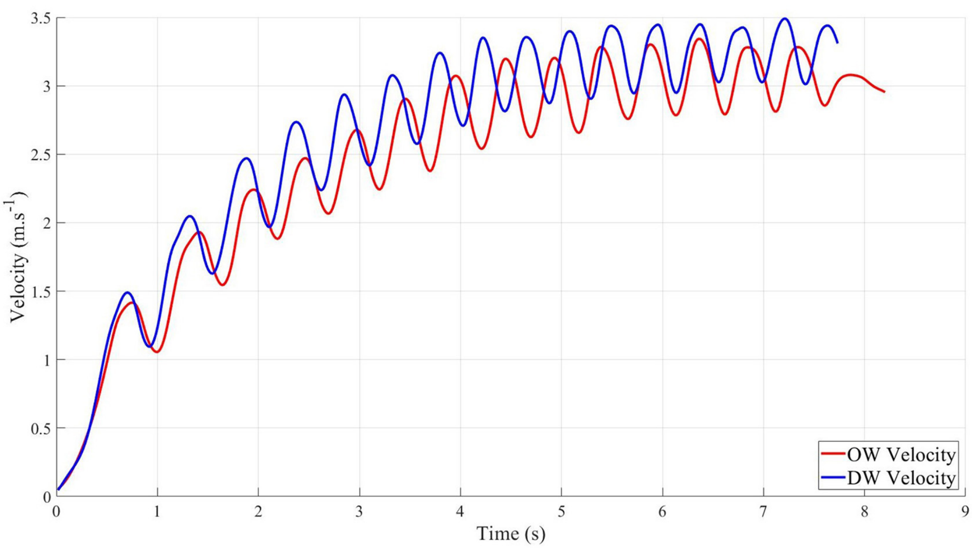


**FIGURE 4 |** Example of velocity curves of offensive wheelchair (OW) and defensive wheelchair developed by typical participant during 20 m sprint.

OW and a DW on kinematic variables during a straight-line sprint. To our knowledge, this is the first study to compare the impact of wheelchair type on kinematic variables in wheelchair rugby. Our results showed that all kinematic performance variables were higher for the DW than the OW. In addition, no difference in terms of asymmetry was found between both wheelchairs. However, the results of this study confirmed our third hypothesis that the rolling resistance would be greater with the DW.

Peak velocities during the first 2 pushes (i.e., acceleration phase) were significantly higher with the DW than the OW, with a large effect size ($r \geq 0.55$). These results are not similar to those of Bakatchina et al. (2021a) who compared players using a DW with players using an OW and found that players using an OW developed higher peak velocities than the players using a DW during the acceleration phase. The mean and maximal velocity during the 20 m sprint were also higher and sprint time was shorter with the DW, with large effect sizes ($r \geq 0.55$). These differences between our results and those of Bakatchina et al. (2021a) can be attributed by the fact that the participants in our study are able-bodied people while those in Bakatchina et al. (2021a) were people with impairments.

The difference in terms of first and second peak velocities, mean velocity, maximal velocity and sprint time found between the OW and the DW in the current study could be related to maneuverability, stability and steering during wheelchair propulsion. OW was handy allowing the user to prevent other wheelchairs from hooking it in practice. In addition, the OW was shorter than the DW (Haydon et al., 2016), which could cause more maneuverability leading instability with OW. According to Tomlinson (2000) who studied the managing maneuverability and rear stability of adjustable manual wheelchairs, they showed that the stability decreases as maneuverability improves. The instability of OW would cause a high variability in kinetic or kinematic variables between both sides of wheelchair as described by Vegter et al. (2013) and Soltau et al. (2015) who compared simultaneous results of two wheels attached to the different sides of the wheelchair. This kinetic or kinematic variables variability between both sides of the wheelchair would cause a steering movement (Wieczorek and Kukla, 2020) which would prevent OW to run in a straight line (De Groot et al., 2002; Soltau et al., 2015) causing a decrease in the performance variables. Steering movement corrections by OW user during a straight-line sprint would lead to increase energy cost (Vegter et al., 2014; Soltau et al., 2015), and causing a decrease performance in terms of mean velocities and sprint time.

Differences in performance between OW and DW could also be related to the user's position in the manual wheelchair. The performance of manual wheelchair players is also related to the wheelchair user position relative to the main axle position (Brubaker, 1986). Thomas et al. (2018) indicated that reclining the wheelchair seat relative to the horizontal axis increased stability in a wheelchair user. According to Haydon et al. (2016), the seat angle of the DW used by LP players was significantly higher than the seat angle of the OW used by HP players. In the current study the seat angle of the DW was slightly higher compared to the OW (26.4° for DW and 22.4° for OW). Consequently, participants would be more stable with the DW during propulsion causing a better sprint time and a high development of peak velocities.

Asymmetry is considered to be related to both decreased performance and increased risk of injury (Vegter et al., 2013; Gagnon et al., 2016). Comparison of asymmetry during straight-line sprinting is important as this is a component of matches. We found no difference in asymmetry between the OW and the DW during either the acceleration or the constant peak velocity phases. The asymmetry values were similar to those reported by Bakatchina et al. (2021a) in a comparison of players using an OW and players using a DW. Cycle frequency is also a key determinant of propulsion injury risk (Boninger et al., 1999). However, our study showed no significant difference in cycle frequency between OW and DW. This contrasts with the findings of Goosey-Tolfrey et al. (2018) who found a higher cycle frequency in HP players than LP players. Cycle frequency may be related to the level of impairment, which would explain the between-group difference in the (Goosey-Tolfrey et al., 2018) study, and the lack of difference in the present study of able-bodied individuals.

Rolling resistance values differed significantly between both wheelchair types for each condition; they were higher with DW than OW in both conditions (when the 20 kg additional mass was placed in front or rear for both wheelchair types). These higher rolling resistance values with DW could be related to the frame length of DW which was higher compared to the OW (Haydon et al., 2016), resulting in a more distribution of DW mass on the front casters. According to Rémy N de et al. (2003) and Sauret et al. (2013), when the mass distribution of the wheelchair-user system is higher on the front of the wheelchair, rolling resistance during wheelchair propulsion is increased.

## PERSPECTIVES

In wheelchair rugby, the choice of the wheelchair type (DW or OW) is related to several factors such as: players' physical capacity or coach's strategies. The current results indicate performance in wheelchair rugby could be related to the wheelchair type. For example, MP players may use either an OW or a DW during the game, therefore some wheelchair rugby clubs have two types of wheelchairs for each MP so that the coach can change the role of the MP between seasons or at half-time. Consequently, coaches and MP players could optimize the choice of wheelchair type improving players performance and coach's strategy. Performance is also partly related to the functional capacity of the abdominal muscles (Vanlandewijck et al., 2010); we believe it would be pertinent to review the configuration of the OW to decrease steering movement in HP players during straight-line sprint. In addition, the ability to accelerate and pivot whilst maintaining control of the ball are also key performance variables in wheelchair rugby. Consequently, future studies should compare the performance of these wheelchairs during pivoting tasks such as the 8 test.

## CONCLUSIONS

Wheelchair configuration is considered as a key performance variable in wheelchair rugby; few studies have evaluated interactions between the user and the wheelchair. The results of our study suggest that wheelchair type influences performance in wheelchair rugby. Mean and maximal velocity and peak velocity during the acceleration phase were higher with the DW than the OW. Sprint time was also faster with the DW. Cycle frequency and asymmetry, which are risk parameters for injury and indicators of high-performance parameters in HP players, do not appear to be influenced by wheelchair type. These results should provide guidance to coaches in the choice of wheelchair type for MP players.

## DATA AVAILABILITY STATEMENT

The raw data presented in the study are not readily available because the participants did not consent to sharing their data when they entered the study. To access the raw data, please contact us at: sadate.bakatchina@univ-tln.fr.

## ETHICS STATEMENT

The studies involving human participants were reviewed and approved by National Ethics Committee for Research in the Physical Activity and Sports Sciences (CERSTAPS No 2018-16-07-26). The patients/participants provided their written informed consent to participate in this study. Written informed consent was obtained from the individual(s) for the publication of any potentially identifiable images or data included in this article.

## AUTHOR CONTRIBUTIONS

SB contributed to conception, design, organized the database and of the study, performed the statistical analysis and wrote the first draft of the manuscript. TW, FB, IA, OV, DP, and AF wrote sections of the manuscript. All authors contributed to manuscript revision, read, and approved the submitted version.

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# Impact of Holding a Badminton Racket on Spatio-Temporal and Kinetic Parameters During Manual Wheelchair Propulsion

*Ilona Alberca[1]*, Félix Chénier[2,3], Marjolaine Astier[1,4], Marion Combet[4], Sadate Bakatchina[1], Florian Brassart[1], Jean-Marc Vallier[1], Didier Pradon[5], Bruno Watier[6] and Arnaud Faupin[1]*

*[1] IAPS, Université de Toulon, La Garde, France, [2] Mobility and Adaptive Sports Research Lab, Department of Physical Activity Sciences, Université du Québec à Montréal, Montreal, QC, Canada, [3] Centre for Interdisciplinary Research in Rehabilitation of Greater Montreal, Institut Universitaire sur la Réadaptation en Déficience Physique de Montréal, Montreal, QC, Canada, [4] Université de Toulon, LAMHESS, EA 6312, La Garde, France, [5] Pole Parasport - ISPC Synergies, Hôpital Raymond-Poincaré, Garches, France, [6] LAAS-CNRS, Université de Toulouse, CNRS, UPS, Toulouse, France*

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*Riemer J. K. Vegter, University Medical Center Groningen, Netherlands*

**Reviewed by:**
*Jolanta Marszalek, Józef Piłsudski University of Physical Education in Warsaw, Poland*
*Ursina Arnet, Swiss Paraplegic Research, Switzerland*

***Correspondence:**
*Ilona Alberca*
*ilona.alberca@univ-tln.fr*

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**Introduction:** Para badminton entered the Paralympic world for the first time with the 2021 Paralympic Games in Tokyo. The particularity of this sport lies in the handling of the wheelchair and the racket simultaneously. To the best of our knowledge, and considering the youthfulness of this sport, it appears that no study has looked at the impact of the badminton racket on the kinetic and spatiotemporal parameters. Therefore, the aim of our study was to investigate the impact of the badminton racket on the amplitude of kinetic and spatiotemporal parameters of wheelchair propulsion, considered as propulsion effectiveness and risk of injury criteria. We hypothesized that holding a badminton racket while propelling the wheelchair modifies the kinetics and temporal parameters of the athlete's propulsion due to the difficulty to hold the handrim, therefore decreasing propulsion effectiveness and increasing risk of injury.

**Materials and Methods:** For six 90-min sessions, 16 able-bodied individuals were introduced to badminton. No injuries hindered their propulsion. They had to propel with and without a racket held on the dominant side along a 20 m straight line at a constant velocity of 5 km/h. They all used the same sports wheelchair equipped with two instrumented wheels (SmartWheel).

**Results:** Participants increased their maximal total force and force rate of rise but decreased their fraction of effective force with their dominant hand compared to the non-dominant hand when using a racket. In addition, they decreased their fraction of effective force, push time, cycle time, and push angle, and increased their maximal propulsive moment, maximal total force, and force rate of rise when comparing the same dominant hand with and without the racket.

**Discussion:** Using a badminton racket modifies the athlete's force application in a way that is generally related to lower propulsion effectiveness and a higher risk for injury. Indeed, it seems that propulsion with a racket prevents from correctly grabbing the handrim.

**Keywords: biomechanics, wheelchair, risk of injury, propulsion effectiveness, Para badminton**

## INTRODUCTION

Para badminton is a young sport as it was first played in the 1990s when several German athletes became interested in adapting the rules of classical badminton for the people with disabilities. It entered the Paralympic world for the first time with the 2021 Paralympic Games in Tokyo.

Small-court wheelchair sports, such as Para badminton, are described as intermittent aerobic activities that are interspersed with brief periods of high-intensity work (Coutts, 1992; Bloxham et al., 2001; Goosey-Tolfrey et al., 2006; Roy et al., 2006; Mota and Almeida, 2020). The nature of the discipline requires athletes to perform rotations, abrupt forward and backward movements, and short sprints. The different shots performed by the players such as the release, the smash, or the drive require high-intensity efforts (Yüksel, 2018a,b). Like wheelchair tennis, the originality of this sport lies in the handling of the wheelchair while holding and using a racket. To the best of our knowledge, no study has investigated the impact of the badminton racket on propulsion effectiveness and risk of injury. However, the wheelchair tennis has been the subject of more studies, some of which focusing on the impact of the racket on kinetic and temporal parameters of the propulsion. These studies have shown that:

- Maximal velocity is reduced on the first three pushes with a racket (Goosey-Tolfrey and Moss, 2005).
- Power loss and power output generation are higher with the racket due to the longer time needed to couple the hand with the racket to the rim (de Groot et al., 2017).
- The arm holding the tennis racket has to withstand higher forces when propelling the wheelchair in sprints, compared to the arm without the racket (de Groot et al., 2017).

Taken together, these findings in wheelchair tennis suggest that the use of the racket induces adjustment in the mechanical spatiotemporal parameters of the athletes related to a decrease in propulsion effectiveness (Goosey-Tolfrey and Moss, 2005; de Groot et al., 2017). Likewise, an increase in the forces carried by an upper limb is associated with an increased risk of injury (Boninger et al., 2005).

The area of interest here is Para badminton, which remains largely unstudied in the scientific literature. However, wheelchair tennis and Para badminton are being the two disciplines close to each other; we can assume that in badminton also, the racket could have a negative impact on the propulsion effectiveness and the injury risk of the athletes. Propulsion effectiveness and injury risk are related to several kinetic and spatio-temporal parameters such as total force, propulsive moment, force rate of rise, fraction of effective force, power, push and cycle time, and push angle (Boninger et al., 2000, 2005; Chow et al., 2001; de Groot et al., 2002, 2008; Goosey-Tolfrey and Moss, 2005; Koopman et al., 2016). Comprehensive analysis including forces developed by the athletes would allow calculating parameters related to the propulsion effectiveness and the risk of injury. Therefore, the aim of our study is to investigate the impact of holding a badminton racket on the kinetic and spatiotemporal parameters of wheelchair propulsion. Specifically, we would like to analyze the impact of the badminton racket during wheelchair propulsion on maximal total force, maximal propulsive moment, rate of rise, fraction of effective force, maximal power output, push and cycle time and push angle. Those are essential parameters that can impact propulsion effectiveness, defined here as the ability to reach and maintain a given velocity, and risk of injury. Based on results in wheelchair tennis we hypothesized that wheelchair propulsion while holding a badminton racket modifies the kinetics and temporal parameters of the athlete's propulsion due to the difficulty to hold the handrim, therefore decreasing propulsion effectiveness and increasing risk of injury (Goosey-Tolfrey and Moss, 2005; Sindall et al., 2013; de Groot et al., 2017).

## MATERIALS AND METHODS

### Study Design

The design of our study focused on the comparison of the measured parameters according to two conditions: propulsion without holding a badminton racket and propulsion while holding a badminton racket. In order to make this comparison and after a 5-min wheelchair warm-up, participants had to propel along with a 20-meter straight line at a constant velocity of 1.4 m/s (5 km/h) using a regular sound signal in a sports complex. They started the test at a standstill. Markers were placed at regular intervals along the 20-meter straight line. Each time the signal sounded; the participant had to be at the next markers, and so on for each marker until the end of the 20 meters. The participant had to propel continuously without braking or accelerating abruptly. To get used to the sound system, the participants were allowed to practice the course prior to the registration of the trial. Two passages were made in a randomized order: with and without a badminton racket. The racket was the same for all participants (Yonex Astrox Smash Navy Blue, 73 g) and was held on the dominant side. Because the test was submaximal, a 1-min recovery time was implemented between each trial.

### Setting

The tests done in this study were performed at the University of Toulon (La Garde, France) on November 21, 2018. The experimental protocol was approved by the Comité d'Ethique pour les Recherches en STAPS (CERSTAPS) from Conseil National des Universités de France [certificate #CERSTAPS 2018-16-07-26] filed on June 6, 2018 and accepted on July 7, 2018. Participants were recruited starting in September, 2018.

### Participants

Our study included 16 able-bodied sports students. Our exclusion criteria were injury or pain that could interfere with wheelchair propulsion. We used a statistical power test to determine the sample size needed for the study. The article by de Groot et al. (2017) was used as a reference for this test. Thus, for a statistical power of 0.95, the calculation of statistical power gave us an average of 8 participants for the statistical tests we wished to perform on our measures. Based on this average, 16 participants were included in the study. Statistical power was calculated using G*Power software (G* Power, 2020; g-power.apponic.com). All participants were introduced to wheelchair maneuverability and

**TABLE 1 |** Participants' characteristics.

| Participant | Gender | Age (years) | Height (cm) | Body mass (kg) | BMI (kg/m$^2$) | Dominant hand |
|---|---|---|---|---|---|---|
| S1 | Man | 42 | 180 | 75 | 23.2 | R |
| S2 | Man | 27 | 179 | 65 | 20.3 | R |
| S3 | Woman | 20 | 165 | 60 | 22.0 | R |
| S4 | Man | 22 | 175 | 95 | 31.0 | R |
| S5 | Man | 21 | 180 | 75 | 23.2 | R |
| S6 | Man | 21 | 179 | 75 | 23.4 | R |
| S7 | Man | 21 | 171 | 64 | 21.9 | R |
| S8 | Man | 20 | 174 | 61 | 20.2 | R |
| S9 | Woman | 21 | 169 | 52 | 18.2 | R |
| S10 | Woman | 24 | 172 | 59 | 19.9 | L |
| S11 | Woman | 19 | 161 | 50 | 19.3 | R |
| S12 | Man | 19 | 176 | 77 | 24.9 | L |
| S13 | Woman | 20 | 170 | 63 | 21.8 | R |
| S14 | Woman | 22 | 163 | 62 | 23.3 | L |
| S15 | Man | 19 | 175 | 95 | 31.0 | R |
| S16 | Man | 22 | 175 | 63 | 20.6 | R |
| Mean(SD) | | 22.5(5.6) | 172.8(5.9) | 68.2(13.1) | 22.8(3.7) | |

*With SD, standard deviation; BMI, Body Mass Index.*

**TABLE 2 |** Description and equations for the outcome measures.

| Parameters | Description | Equations |
|---|---|---|
| | **Pushrim kinetics** | |
| Maximal total force (Ftot$_{peak}$) [N] | Sum of the maximal forces in the 3 planes of space applied to the handrim for each push | $max(sqrt(Fx^2 + Fy^2 + Fz^2))$ |
| Maximal propulsive moment (Mz$_{peak}$) [Nm] | Maximal propelling moment applied to the handrim for each push | Calculation carried out by the SmartWheel software |
| Rate of rise (RoR) [N.s$^{-1}$] | Rate of rise in maximal total force for each push | $\frac{dFtot_{max}}{dt}$ |
| Fraction of Effective Force (FEF) [%] | Percentage of forces useful for propulsion | $abs\left(\frac{Ftan}{Ftot}\right) x\ 100$ |
| Maximal power output (PO$_{peak}$) [W] | Maximal power output developed by the participant to the handrim for each push | $peak[\theta \times Mz]$ |
| Angular impulse (AI) [Nm.s] | Gain of propulsive moment during one push | $Mz_{mean}\ x\ PT$ |
| | **Temporal parameters** | |
| Push time (PT) [s] | Contact time between hand and wheelchair handrim | $t_{end}(i) - t_{start}(i)$ |
| Cycle time (CT) [s] | Time between the start of first push and next push for each push | $t_{start2}(i) - t_{start1}(i)$ |
| Push angle (PA) [°] | Wheel angle course during push time | Calculation carried out by the SmartWheel software |

*With Fx, horizontal force; Fy, vertical force; Fz, mediolateral force; r, wheel radius; start, start of a push; end, end of a push; t, time (s); v, wheel velocity; i, push considered.*

Para badminton during 6 practice sessions of 90 min. They were novices in wheelchair handling and wheelchair propulsion. These practice sessions are part of their school curriculum in Sciences et Techniques des Activités Physiques et Sportives (STAPS). Characteristics of all participants are presented in **Table 1**.

## Data Measurement

Participants used a single multi-sport wheelchair with a wheel size of 26 inches and a camber angle of 18 degrees, which is similar to chairs used in Para badminton. The chair was equipped bilaterally with two instrumented wheels (SMARTWheel. 2013 edition, Outfront LCC). Measurement tools such as instrumented wheels allow to measure parametersin conditions close to the original discipline and without impeding propulsion. These wheels have a weight and moment of inertia of ∼4.9 and 0.15 kg·m$^2$ (Sprigle et al., 2016). With these tools, we can measure the wheel angle $\theta$, forces $F_x$, $F_y$, $F_z$($F_y$ is the force applied up and down on the pushrim; $F_x$ is force applied laterally; $F_z$ is the force out of the plane of the wheel *SmartWheel 2008*[1] p. 46. *Users Guide*, 2014) and force moments $M_x$, $M_y$, $M_z$applied on each handrim for all sessions at 240 Hz. Dynamic kinetic offsets were canceled using a method described in Chénier et al. (2017)

[1]SmartWheel 2008 Users Guide (2014).

because the recorded kinetics may include dynamic offsets that affect the accuracy of the measurements. Wheelchair velocity was calculated from wheel angles using a 131-point first-order Savitzky-Golay derivative filter (Chénier et al., 2015).

All pushes recorded by the instrumented wheels were segmented. A 30 N threshold selected experimentally based on the recorded dataset helped us to make this segmentation. This automated segmentation was manually checked for each of the push for each trial to correct any errors. For each run, the first two and last pushes were excluded and considered as transitional pushes.

## Variables

The parameters presented in **Table 2** were calculated and averaged over all the selected pushes in a bilateral manner. Thus, we obtained kinetic and spatiotemporal data for the dominant and non-dominant hand of each participant.

All data processing and calculations were performed using Python/SciPy and the Kinetics Toolkit library (Chénier, 2021).

## Statistical Methods

A total of 10 variables were calculated. The means and standard deviation of these variables were calculated per condition and per limb separately. All data were analyzed using SPPS version 20 (SPSS Inc., Chicago, Illinois USA).

The Shapiro–Wilk test showed that all outcomes' measures were not normally distributed. Thus, the statistical analyses were performed on the log-transformed data. A repeated measures ANOVA was then performed (with two within factors: with racket vs. without racket; dominant vs. non-dominant hand) to look at the existing differences between dominant and non-dominant hand according to the with-or-without-racket condition. A Mauchly sphericity test was performed to check if the sphericity hypothesis was violated. This was the case for all the calculated variables. A Greenhouse-Geisser correction was applied. A Bonferroni adjustment was made for multiple comparisons with $p = 0.05$. For each significant difference, the effect size $\eta_p^2$ was calculated using the following equation:

$$\eta_p^2 = \frac{SS_{effect}}{SS_{effect} + SS_{error}} \quad (1)$$

*With $\eta_p^2$: partial eta-squared of the considered variable; $SS_{effect}$: effect sums of squares of the considered variable; $SS_{error}$: error sums of squares of the considered variable.*

Effect size was interpreted according to Cohen (1988): small ($\eta_p^2 = 0.01$), medium ($\eta_p^2 = 0.06$), and large ($\eta_p^2 = 0.14$).

We also performed a paired student test to compare the parameters of the same hand with and without a racket on the log-transformed data. Statistical significance was set at $p < 0.05$. For each significant difference, the effect size d was calculated using the following equation:

$$d = \frac{mean(X_0) - mean(X_1)}{s.d.(X_0)} \quad (2)$$

**TABLE 3** | Kinetic and spatiotemporal parameters in 20-meter wheelchair straight propulsion according to the condition (with racket, without racket) and upper limb (dominant, non-dominant).

| | With racket | | | Without racket | | | ANOVA | | | | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | D | ND | | D | ND | | Condition effect | | | Side effect | | | Interaction (Condition × Side) | | |
| | Mean (SD) | Mean (SD) | D | Mean (SD) | Mean (SD) | d | F | p | $\eta^2$ | F | p | $\eta^2$ | F | p | $\eta^2$ |
| $P_{peak}$ [W] | 112.53 (63.74) | 105.00 (51.43) | 0.130 | 104.78 (65.87) | 95.98 (55.43) | 0.145 | **4.879** | **0.028** | **0.025** | **6.298** | **0.013** | **0.032** | 0.252 | 0.616 | 0.001 |
| $Mz_{peak}$ [N.s] | 22.24 (8.21) | 21.61 (8.76) | 0.074 | 20.55 (10.53) | 19.56 (10.22) | 0.095 | **7.049** | **0.009** | **0.036** | **5.680** | **0.018** | **0.029** | 0.343 | 0.559 | 0.002 |
| $Ftot_{peak}$ [N] | **117.77 (45.36)** | **86.53 (40.75)*** | **0.725** | **86.53 (38.65)** | **73.75 (38.81)*** | **0.330** | **32.738** | **<0.001** | **0.148** | **123.513** | **<0.001** | **0.395** | **12.211** | **<0.001** | **0.061** |
| FEF [%] | **29.36 (6.93)** | **40.85 (14.15)*** | **1,031** | **35.82 (10.94)** | **41.14 (13.14)*** | **0.440** | **33.888** | **<0.001** | **0.152** | **124.709** | **<0.001** | **0.398** | **25.260** | **<0.001** | **0.118** |
| Ror [N/s] | 587.42 (305.96) | 388.64 (228.52) | 0.736 | 388.65 (210.90) | 289.02 (181.49) | 0.506 | **43.815** | **<0.001** | **0.188** | **118.824** | **<0.001** | **0.386** | 1.765 | 0.186 | 0.009 |
| AI [Nm.s] | 4.01 (2.00) | 4.00 (2.19) | 0.005 | 4.25 (2.41) | 4.10 (2.44) | 0.062 | 0.102 | 0.750 | 0.001 | 0.286 | 0.594 | 0.002 | 1.191 | 0.276 | 0.006 |
| PT [s] | 0.34 (0.10) | 0.34 (0.08) | 0 | 0.36 (0.07) | 0.37 (0.07) | 0.143 | **12.254** | **<0.001** | **0.061** | 0.003 | 0.955 | 0.000 | 0.004 | 0.950 | 0.000 |
| CT [s] | 1.13 (0.43) | 1.11 (0.39) | 0.049 | 1.29 (0.45) | 1.30 (0.45) | 0.022 | **12.797** | **<0.001** | **0.063** | 0.394 | 0.531 | 0.002 | 0.685 | 0.409 | 0.004 |
| PA [°] | 84.68 (30.47) | 83.22 (19.44) | 0.057 | 90.98 (19.60) | 91.06 (19.23) | 0.004 | **10.886** | **0.001** | **0.054** | 0.196 | 0.659 | 0.001 | 0.043 | 0.836 | 0.000 |

*With racket, racket held in the dominant hand; D, dominant hand; ND, non-dominant hand; Condition, with or without racket; Side, dominant or non-dominant hand; SD, standard deviation.*
**Significant difference in Post hoc pair wise comparisons with Bonferroni adjustment (dominant vs. non-dominant hand) with $p < 0.001$.*
*F, result of the ANOVA; p, p-value fixed at 0.05; d, effect size for the significant difference in Post hoc pair wise comparisons with Bonferroni adjustment (dominant vs. non-dominant hand); $\eta^2$, effect size for the significant difference in the ANOVA. Bold values indicate the significant values.*

**TABLE 4 |** Kinetic and spatiotemporal parameters in a 20-meter wheelchair straight propulsion of the same dominant hand with and without racket.

| | Dominant hand | | *T*-test | | |
|---|---|---|---|---|---|
| | With racket | Without racket | With racket × Without racket | | |
| | Mean (SD) | Mean (SD) | t | *P* | d |
| $P_{peak}$ [W] | 112.53 (63.74) | 104.78 (65.87) | **1.867** | **0.032** | **0.120** |
| $Mz_{peak}$ [N.s] | 22.24 (8.21) | 20.55 (10.53) | **2.356** | **0.010** | **0.179** |
| $Ftot_{peak}$ [N] | 117.77 (45.36) | 86.53 (38.65) | **7.530** | **<0.001** | **0.741** |
| FEF [%] | 29.36 (6.93) | 35.82 (10.94) | **8.197** | **<0.001** | **0.705** |
| Ror [N/s] | 587.42 (305.96) | 388.65 (210.90) | **7.597** | **<0.001** | **0.756** |
| AI [Nm.s] | 4.01 (2.00) | 4.25 (2.41) | 0.330 | 0.371 | 0.108 |
| PT [s] | 0.34 (0.10) | 0.36 (0.07) | **3.086** | **0.001** | **0.231** |
| CT [s] | 1.13 (0.43) | 1.29 (0.45) | **3.134** | **<0.001** | **0.363** |
| PA [°] | 84.68 (30.47) | 90.98 (19.60) | **2.555** | **0.006** | **0.246** |

*With racket, racket held in the dominant hand; SD, standard deviation; t, results of the t-test; d, effect size for the significant difference; p, p-value fixed at 0.05; ANOVA. Bold values indicate the significant values.*

*With X: studied parameter, 0: data without racket or dominant hand according to the statistical analysis and 1: data with racket or non-dominant hand according to the statistical analysis.*
Effect size was interpreted according to (Cohen, 1988): small ($d = 0.2$), moderate ($d = 0.5$), and large ($d = 0.8$) (Cohen, 1988).

## RESULTS

We checked the average velocity of the participants to ensure that the constant velocity requirement was met. The participants reached a mean velocity of 1.44 m/s during the runs with racket and 1.42 m/s during the runs without racket, which corresponds to the imposed velocity.

### Bilateral Analysis

The results of the bilateral analysis are presented in **Table 3**. When comparing the results of both hands with and without a racket, an effect of the racket was found for all parameters except AI. Indeed, with racket $P_{peak}$ ($p = 0.028$) and $Mz_{peak}$ ($p = 0.009$) increase slightly and $Ftot_{peak}$ ($p < 0.001$) and Ror ($p < 0.001$) increase largely. On the contrary, FEF ($p < 0.001$) decreases largely, PA ($p = 0.001$) decreases slightly and PT ($p < 0.001$) and CT ($p < 0.001$) decrease moderately in condition with racket. Significant differences between dominant and non-dominant hand regardless of the condition were noted for $P_{peak}$ ($p = 0.013$), $Mz_{peak}$ ($p = 0.018$), $Ftot_{peak}$ ($p < 0.001$), FEF ($p < 0.001$), and Ror ($p < 0.001$). $P_{peak}$ and $Mz_{peak}$ are slightly higher on the dominant hand and $Ftot_{peak}$ and Ror are largely higher on the dominant side. Conversely, FEF is largely lower on the dominant side compared to the non-dominant side. Finally, an interaction between the condition and the side considered was found for $Ftot_{peak}$ ($p < 0.001$) and FEF ($p < 0.001$).

### Unilateral Analysis

The results of the unilateral analysis of the data are presented in **Table 4**. When we compare the same dominant hand with and without racket, we note that $P_{peak}$ ($p = 0.032$) and $Mz_{peak}$ ($p = 0.010$) are slightly higher and $Ftot_{peak}$ ($p < 0.001$) and Ror ($p < 0.001$) are largely higher with racket. While FEF ($p < 0.001$) is largely lower and PT ($p = 0.001$), CT ($p < 0.001$), and PA ($p = 0.006$) are slightly lower with racket compared to the passage without racket.

## DISCUSSION

The design analyzing the impact of holding a badminton racket conducted in this article is, to our knowledge, the first of his kind in Para badminton. The objective of this article was to study the impact of the badminton racket on the amplitude of kinetic and spatiotemporal parameters of wheelchair propulsion. We hypothesized that wheelchair propulsion while holding a badminton racket modifies the kinetics of the athlete's propulsion. This hypothesis has been verified. Indeed, the use of the racket induces a negative impact on propulsion effectiveness when comparing the same hand with and without racket (fraction of effective force, push time, and push angle) and the dominant hand with racket vs. non-dominant hand (fraction of effective force). Although athletes can maintain the imposed constant overall velocity, their propulsion effectiveness is impacted. However, we must mention that only one propulsion effectiveness parameter (fraction of effective force) is impacted by the racket in the bilateral analysis of the data and that maximal propulsive moment increases slightly in the dominant hand with the racket compared to the same hand without the racket, which is positively related to better propulsion effectiveness. Moreover, the use of a badminton racket also seems to increase parameters related to risk of injury when comparing the dominant and non-dominant hand (maximal total force and rate of rise) and the same hand with and without the racket (maximal total force, rate of rise increased, and cycle time).

The increase in the maximal propulsive moment in the dominant hand during racket propulsion is accompanied by a moderate decrease in the fraction of the effective force, the push time, and the push angle. These parameters are related to propulsion effectiveness and our results appear to be consistent with a decrease in participant propulsion effectiveness. It is possible that the difficulty to grab the handrim of the wheelchair with the racket explains these results. Indeed, participant weakly increases their propulsive moment with the racket but with less continuity as evidenced by the push time and the push angle. Therefore, the proportion of forces that is useful for propulsion decreases. It seems that the wheelchair user makes shorter and reduced movements. For push time, de Groot et al. (2017) also looked at it in tennis and their study showed a decrease in push time and push angle, or contact angle as it is written in their study, with a tennis racket. These results are like ours although we do not deal with the same adapted sport. The decreases observed for these two parameters in the study of de Groot et al. (2017), are greater than those of our study. Indeed, the push time and the push angle decrease, respectively, by 18 and 20% in the study of de Groot et al. (2017) while in our study they decrease only by 5 and 8%. These differences may be due to the properties of the rackets. Indeed, a tennis racket is heavier and has a wider handle than a badminton racket. As a result, we can assume that the impact of a tennis racket is greater than that of a badminton racket. Moreover, we must remember that our study was carried out on able-bodied players. They therefore benefit from abdominal capacities that may be absent in people with disabilities. In addition, they have fewer skills than the Para badminton players.

The use of the racket appears to cause an increase in maximal total force when we look at the results of both hand with and without racket and the same dominant hand with and without a racket, resulting in a moderate increase in the rate of rise in the hand carrying the racket. It is possible to assume that the use of the racket hinders participants and prevents them from properly catching the handrim. They will then compensate for this lack of grip by applying more force on the handrim. In addition, the cycle time decreases when using the racket. For the same propulsion velocity, the participant made more and faster pushes, therefore increasing propulsion frequency. These sets of changes are considered to be risk factors for injury (Boninger et al., 2005). This result may be of particular interest for the coaches. Indeed, knowing that the use of the racket can increase the risk of injury, coaches can propose adapted sessions such as longer rest periods or specific active recoveries.

The results of the ANOVA show the existence of significant differences between dominant and non-dominant hand regardless of the propulsion condition. $P_{peak}$ and $Mz_{peak}$ are slightly higher on the dominant hand and $Ftot_{peak}$ and Ror are largely higher on the dominant side. FEF is largely lower on the dominant side compared to the non-dominant side. These differences indicate the existence of an asymmetry between dominant and non-dominant hand for these parameters. Indeed, it seems that participants apply greater forces and powers on the dominant side than on the non-dominant side. These sets of changes are considered to be risk factors for injury (Boninger et al., 2005). Similarly, they appear to slightly produce more force useful for propulsion on the dominant side without this increasing their FEF. This indicates that the participant increases more forces not useful for the propulsion of the wheelchair, which is related to less propulsive effectiveness. It is possible that the participants' sport practice besides the study induced this asymmetry. Indeed, it is the case of asymmetrical sports practices such as racket sports that develop more muscle strength on the side of the limb carrying the racket. Several authors have also shown that one arm is specialized in a task compared to the second arm (Bagesteiro and Sainburg, 2002, 2003; Sainburg and Wang, 2002; Wang and Sainburg, 2003, 2004; Haaland, 2004; Sainburg and Schaefer, 2004; Schaefer et al., 2007). It is possible that the dominant limb is specialized in force production, unlike the non-dominant limb, which would explain this asymmetry.

We believe that the main limitation of this study concerns the group of participants. Indeed, our experiment was conducted on a population of able-bodied participants not experienced in wheelchair propulsion. The study on able-bodied participants provides homogeneous groups (Rice et al., 2010). However, for people who use manual wheelchairs daily, such as individuals with a paraplegia or tetraplegia, abdominal and trunk capabilities may be reduced due to the severity of the disability. Moreover, even though the participants were trained in Para badminton, badminton players have better racket handling technique than not experienced able bodied participants. The propulsion technique will differ from a novice participant to an expert in Para badminton. This influences propulsion, therefore inducing that our results will not be completely transferable to a population of people with disabilities. In addition to this limitation, we also studied wheelchair propulsion at constant velocity with and without a racket. However, this discipline mainly requires players to perform short sprints forward and backward. Our study being one of the first to look at the impact of the badminton racket on propulsion, we chose to carry out the tests in submaximal condition. This allows us to make a general assessment before being able to study the impact of the badminton racket in various conditions, to be sure that our results are the consequence of the addition of a condition (here the racket). Finally, the use of instrumented wheels increased the weight of the wheels, which may increase the rolling resistance of the wheelchair and its inertia. However, we believe that our results remain valid since the measurements are taken under the same conditions: we use two instrumented wheels that increase the rolling resistance in the same way on each side.

The objective of this study was to analyze the impact of the badminton racket on the kinetic parameters of wheelchair propulsion. We have highlighted that its use agrees with a modification of the kinetics of the participants related to a decrease of the propulsive effectiveness and an increase of the risks of injuries. To complete this analysis and to better understand the impact of the racket, future studies should be conducted under conditions encountered in playing Para badminton, such as consecutive forward and backward propulsion tests that approximate the movements encountered during practice. Moreover, an interesting aspect would also be to work on the comparison of the different possibilities of holding the badminton racket during propulsion. In the field of Para tennis, Koopman et al. (2016), have already been interested in testing different racket holding techniques. We could do the same in the field of Para badminton to complete the analysis of the impact of the racket on propulsion. Finally, proposing new handrim designs could be a solution to the difficulties encountered during propulsion with a badminton racket.

## DATA AVAILABILITY STATEMENT

The original contributions presented in the study are included in the article/supplementary materials, further inquiries can be directed to the corresponding authors.

## ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Comité d'Ethique pour la Recherche en Sciences et Techniques des Activités Physiques et Sportives. The patients/participants provided their written informed consent to participate in this study. Written informed consent was obtained from the individual(s) for the publication of any potentially identifiable images or data included in this article.

## AUTHOR CONTRIBUTIONS

IA: data curation, formal analysis, and writing—original draft. FC: data curation and writing—review and editing. MA and MC: funding acquisition and methodology. SB and FB: funding acquisition and writing—review and editing. J-MV: project administration and validation. DP: data curation. AF: methodology, project administration, and writing—review and editing. All authors contributed to the article and approved the submitted version.

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CASE REPORT
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# Case Report: Training Monitoring and Performance Development of a Triathlete With Spinal Cord Injury and Chronic Myeloid Leukemia During a Paralympic Cycle

*Oliver J. Quittmann[1,2]*, Benjamin Lenatz[1], Patrick Bartsch[3], Frauke Lenatz[1], Tina Foitschik[1] and Thomas Abel[1,2]*

[1] Department IV: Movement Rehabilitation, Neuromechanics and Paralympic Sport, Institute of Movement and Neurosciences, German Sport University Cologne, Cologne, Germany, [2] European Research Group in Disability Sport (ERGiDS), Bonn, Germany, [3] Gesundheitszentrum Gektis, Radevormwald, Germany

**Introduction:** Paratriathlon allows competition for athletes with various physical impairments. The wheelchair category stands out from other paratriathlon categories, since competing in swimming, handcycling, and wheelchair racing entails substantial demands on the upper extremity. Therefore, knowledge about exercise testing and training is needed to improve performance and avoid overuse injuries. We described the training monitoring and performance development throughout a Paralympic cycle of an elite triathlete with spinal cord injury (SCI) and a recent diagnosis of chronic myeloid leukemia (CML).

**Case Presentation/Methods:** A 30-year-old wheelchair athlete with 10-years experience in wheelchair basketball contacted us for guidance regarding testing and training in paratriathlon. Laboratory and field tests were modified from protocols used for testing non-disabled athletes to examine their physical abilities. In handcycling, incremental tests were used to monitor performance development by means of lactate threshold ($P_{OBLA}$) and define heart rate-based training zones. All-out sprint tests were applied to calculate maximal lactate accumulation rate ($\dot{V}La_{max}$) as a measure of glycolytic capabilities in all disciplines. From 2017 to 2020, training was monitored to quantify training load (TL) and training intensity distribution (TID).

**Results:** From 2016 to 2019, the athlete was ranked within the top ten at the European and World Championships. From 2017 to 2019, annual TL increased from 414 to 604 h and demonstrated a shift in TID from 77-17-6% to 88-8-4%. In this period, $P_{OBLA}$ increased from 101 to 158 W and $\dot{V}La_{max}$ decreased from 0.56 to 0.36 mmol·l$^{-1}$·s$^{-1}$. TL was highest during training camps. In 2020, after he received his CML diagnosis, TL, TID, and $P_{OBLA}$ were 317 h, 94-5-1%, and 108 W, respectively.

**Discussion:** TL and TID demonstrated similar values when compared with previous studies in para-swimming and long-distance paratriathlon, respectively. In contrast, relative TL during training camps exceeded those described in the literature and was accompanied by physical stress. Increased volumes at low intensity are assumed

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Christof A. Leicht,
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Ben Thomas Stephenson,
Loughborough University,
United Kingdom
Julia Kathrin Baumgart,
Norwegian University of Science and Technology, Norway

***Correspondence:**
Oliver J. Quittmann
o.quittmann@dshs-koeln.de

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to increase $P_{OBLA}$ and decrease $\dot{V}La_{max}$ over time. CML treatment and side effects drastically decreased TL, intensity, and performance, which ultimately hindered a qualification for Tokyo 2020/21. In conclusion, there is a need for careful training prescription and monitoring in wheelchair triathletes to improve performance and avoid non-functional overreaching.

**Keywords: spinal cord injury (SCI), chronic myeloid leukemia (CML), case report, paratriathlon, training, TRIMP, sRPE, performance**

## INTRODUCTION

Paratriathlon is an endurance sport for people with a physical impairment that made its Paralympic debut in 2016 and is increasingly featured in newspaper articles (1). The athletes compete in various sports classes covering the ambulant/standing, visually impaired, and wheelchair (PTWC) categories (2). Within the PTWC, there are two sports classes for most (PTWC1) and least (PTWC2) impaired wheelchair users. During the competition, the PTWC1 starts with a time advantage before PTWC2 athletes (3). Although the ambulant/standing and visually impaired categories appear to be rather similar to conventional triathlon, the locomotion in PTWC is substantially different. Since these athletes purely rely on their upper extremities, handcycling and wheelchair racing are used as equivalent to leg cycling and running, respectively.

Previous case studies already reported training characteristics in para-swimming (4), handcycling (5, 6), wheelchair racing (7), and long-distance amputee paratriathlon (8). Training characteristics and performance development of a paratriathlon long-distance world champion (with unilateral below-the-knee amputation) were described over a period of 19 months (8). Mean training volumes were found to be lower when compared with non-disabled Olympic-distance triathletes and attained values of $8 \pm 3$, $6 \pm 4$, and $2 \pm 1$ h/w in swimming, cycling, and running, respectively (8). However, training practices may have changed from 2011/2012 (when the data of this case study were recorded) to the present and do not necessarily apply to triathletes with spinal cord injury (SCI).

Traumatic SCIs are defined as damage to the spinal cord due to a mechanical trauma "that temporarily or permanently causes changes in its function" (9). Depending on the level and (in)completeness of the lesion, afferent and efferent neurons as well as autonomic function are affected to a certain extent. Although paraplegia indicates that two limbs are affected (predominantly damage to thoracic, lumbar, and sacral regions), tetraplegia refers to impairment in all four limbs. The incidence of traumatic SCI was found to differ among age groups (10) and regions and is ~1 case per 100,000 individuals in Germany (11). The major causes of traumatic SCIs are accidents in motor vehicles (11). As a treatment of traumatic SCIs, surgical decompression in an early state and neuroprotective and/or regenerative strategies in the follow-up may help to reduce symptoms and side effects (12). Besides, habitual exercise was highlighted as "an effective countermeasure for addressing physical deconditioning after SCI" (13). However, since wheelchair athletes purely rely on their upper extremities, a high prevalence of upper extremity injuries was highlighted (14).

Chronic myeloid leukemia (CML) is a myeloproliferative neoplasm that accounts for approximately 15% of newly diagnosed cases of leukemia in adults (15). The incidence of CML is stated to be 1–2 cases per 100,000 adults with a mortality of 1–2% (15). The genetic origin of CML is assumed to be a fusion oncogene (BCR-ABL1) on the so-called *"Philadelphia chromosome"* (22q11.2) (15). The first-line treatment of CML in the chronic phase is different types of tyrosine kinase inhibitors (TKIs) that lead to a normal life expectancy for most patients (16). It was recently shown that the majority of patients with CML receiving TKI therapy experience severe fatigue that causes an increased need for sleep, a reduction of physical activity, and consequently an impaired quality of life (17). However, alternative treatments to TKIs indicated promising results in terms of molecular response and side effects (18). Besides a stable deep molecular response, CML therapy aims for treatment-free remission (16). While moderate exercise was found to be a promising tool in the treatment of acute myeloid leukemia (19), findings regarding the interaction of CML and exercise are still lacking.

This case report addresses several research gaps. First, longitudinal studies on the training and development of Paralympic athletes are generally sparse—especially over several years. Second, evidence on how paratriathletes prepare for the Paralympic Games is lacking. Third, the PTWC category demonstrates the highest difference from conventional triathlon when compared with other categories and as such requires the implementation of modified tools in exercise testing and training. Finally, to the best of our knowledge, there is no study that analyzed the acute reaction to and side effects of CML and its treatment on performance, training, and wellbeing in a highly trained athlete.

## CASE DESCRIPTION

In November 2014, a 30-year-old male wheelchair athlete (ID: BL) with SCI classified as ASIA C (20) contacted our university and asked for support regarding his athletic orientation toward

**Abbreviations:** BLC, blood lactate concentration (mmol·$l^{-1}$); CML, chronic myeloid leukemia; CSS, critical swim speed; CV, critical velocity; $P_{OBLA}$, power according to a lactate concentration of 4 mmol·$l^{-1}$; PTWC, wheelchair category in paratriathlon; sRPE, session ratings of perceived exertion (scale from 1 to 10); TKI, tyrosine kinase inhibitors; TLI, Total load index [sRPE × training duration (min)]; $TRIMP_3$, training impulse (based on three-zone model); $TRIMP_5$, training impulse (based on five-zone model); $\dot{V}La_{max}$, maximal lactate accumulation rate (mmol·$l^{-1}$·$s^{-1}$); $\dot{V}O_{2max}$, maximal oxygen uptake (ml·$min^{-1}$·$kg^{-1}$); $\dot{V}O_{2peak}$, peak oxygen uptake (ml·$min^{-1}$·$kg^{-1}$).

paratriathlon (**Figure 1**). BL had participated in professional wheelchair basketball for a decade and had already finished several triathlons including national championships. He wanted to have guidance in sport-specific training and testing to achieve his ultimate goal: participating in the Paralympic Games in either Rio de Janeiro (2016) or Tokyo (2020). The athlete gave written informed consent to take part in this study. Standardized guidelines for reporting were used (refer to **Supplementary Material 1**).

From 1990 to 2003, BL participated in motocross races. In March 2003, at the age of 18 years, he was involved in a quad bike accident that caused lumbar spine compression (ICD10-S32.01), kidney contusion (ICD10-S37.01), and incomplete paraplegia (ICD10-S34.71) with neurogenic bladder dysfunction (ICD10-N31.1) that required the permanent use of a wheelchair (ICD10-Z99.3). Immediate surgery stabilized the spine by internal fixation of the thoracolumbar junction. BL received physiotherapy, ergotherapy, and medical care during acute rehabilitation up to June 2004. He started wheelchair basketball soon after rehabilitation and played for several teams at the national level. Regular medical check-ups remained unsuspicious. In August 2013, he participated in his first sprint-distance paratriathlon.

# METHODS

Due to the unique demands (focus on an upper extremity) of PTWC, we modified laboratory and field tests that are commonly performed by non-disabled athletes to assess BL's physical abilities in swimming, handcycling, and wheelchair racing. Depending on his performance in paratriathlon events (discipline-specific ranking and split times) and the outcomes of the exercise tests, the training prescription focused on disciplines and/or physiological parameters that seemed to provide particular performance gains. As such, handcycling was identified as the discipline with the highest potential for improvement. Moreover, the applied training concept was oriented toward similar case studies and the international literature that highlighted the value of high training volumes performed at low intensity (5, 6, 8, 21). In our study, low intensity is oriented toward maximal fat oxidation rate ($Fat_{max}$) (22). Training camps of a 2-weeks duration were performed 2–3 times a year during the preparation period (typically between October and March). Procedures to systematically quantify his training in every discipline were applied (23).

## Exercise Testing

Prior to any testing, the participant received a medical check-up following the guidelines of the European Society of Cardiology, which includes the individuals' own medical, family, and personal history, a physical examination, and a resting electrocardiogram (24). The frequency of exercise tests increased from once per year in 2015–2016 (performed in May) to 4 times per year in 2018–2019 (performed at the beginning of the preparation and competition period as well as preceding the training camps). Also, the number of procedures increased over the years and finally required one testing day for swimming, handcycling, and wheelchair racing, respectively. Test days were separated by 1–2 rest days with (at most) low-intensity training at ($Fat_{max}$). Besides rather common procedures in exercise testing which target maximal oxygen uptake ($\dot{V}O_{2max}$) and/or lactate threshold (25), we developed procedures to determine maximal lactate accumulation rate ($\dot{V}La_{max}$) as a measure of the glycolytic metabolism. These procedures demonstrated sufficient reliability, were associated to physical performance in handcycling (26, 27) and running (28, 29), and were modified for this case study accordingly. $\dot{V}La_{max}$ is derived from short sprint tests and calculated by dividing the increase in postexercise blood lactate concentration (BLC) by the assumed lactic period (in this study 3 s) (30, 31). Discipline-specific procedures to determine $\dot{V}La_{max}$ are described below.

The high number of tests was physically and mentally challenging for BL and led to a trade-off between testing and training. Testing required access to various sports facilities and was dependent on weather conditions and temperature for wheelchair racing field tests. For tests including exhaustion in the laboratory, a medical doctor was on on-call duty which complicated scheduling. Procedures focused on aerobic as well as anaerobic parameters to create a holistic physiological profile of the athlete in all disciplines (30, 32). Furthermore, we wanted to supplement the ease of field testing with the standardization and reliability of lab tests.

In swimming, critical swim speed (CSS) was determined by performing 200-m and 400-m time trials in a 50-m pool (33). Later, testing was expanded by an initial 25-m sprint test and a closing 750-m time trial. The 750-m trial was requested and consequently performed by the athlete since this is the swimming distance in standard paratriathlon events and approximates his performance in competition. Immediately before and after the time trials, blood samples were collected from the earlobe and analyzed using an enzymatic-amperometric sensor chip system (Biosen C-Line, EKF-diagnostic GmbH, Barleben, Germany) to assess the net lactate production. Postexercise BLC of the 25-m sprint test was recorded every minute for 10 min to estimate $\dot{V}La_{max}$.

In handcycling, an incremental test on an ergometer (Cyclus 2, RMB electronic automation GmbH, Leipzig, Germany) was performed to determine the power corresponding to a BLC of 4 $mmol \cdot l^{-1}$ ($P_{OBLA}$) (25) and the peak oxygen uptake ($VO_{2peak}$) which was measured by a spirograph (ZAN 600, nSpire Health, Inc., Longmont, CO, United States), as these parameters are significantly associated with handcycling performance (34–36). The incremental test started with an initial load of 20 W and increased intensity by 20 W every 5 min until the athlete attained subjective exhaustion (26). Furthermore, an isokinetic 15-s all-out sprint test was performed on the same ergometer to determine $\dot{V}La_{max}$ (26, 27). Since $\dot{V}O_{2peak}$ may depend on the used protocol (37, 38), an additional ramp test (80 W, 5 W, 15 s) was performed on some occasions to determine $\dot{V}O_{2max}$, which "is defined as the highest rate at which oxygen can be taken up and utilized by the body during severe exercise" (39).

In wheelchair racing, an initial 110-m sprint test on an outdoor track was performed to calculate $\dot{V}La_{max}$ analogously to previous studies in running (28). Later, time trials over 1,500 and

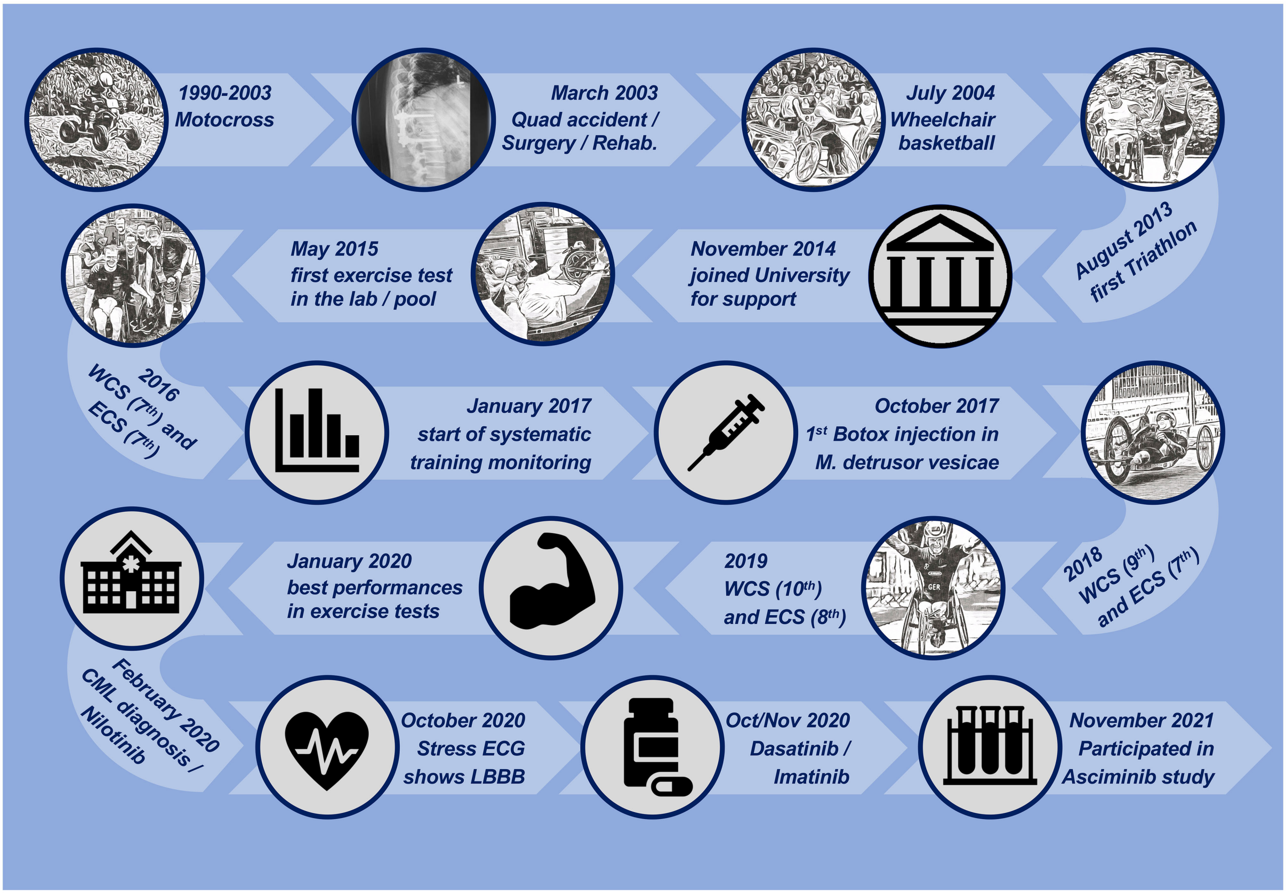


**FIGURE 1 |** Timeline of the athlete's personal history and milestones during the exposure. CML, chronic myeloid leukemia; ECG, electrocardiogram; ECS, european championship; SCI, spinal cord injury; LBBB, left bundle branch block; WCS, world championship.

3,000 m were applied to determine performance, critical velocity (CV), and immediate post-exercise BLC. These trials were used to determine discipline-specific performance and use CV as an indicator of the high-intensity domain (40).

## Training Monitoring

The training was quantified by methods already applied in conventional triathlon (23). External (e.g., time, velocity, power, and cadence) and internal (e.g., heart rate) training measures were recorded by a sports watch or bike computer that was (even in swimming) connected with a heart rate monitor (Forerunner 920XT, Edge 20, HRM-Tri and HRM-Swim, Garmin International, Inc., Olathe, KS, United States). Although all these measures were used to schedule the training, heart rate was found to be most suitable for quantifying the training and comparing between the disciplines. Discipline-specific heart rate intensity zones (T5-T1) were determined as percentages of maximum heart rate with thresholds of 93, 85, 75, and 60%, respectively. These thresholds were found to fit well with training zones from physiological exercise testing (21) and attain stable results. Despite differences in power over time, the heart rate corresponding to a BLC of 2 and 4 mmol·l$^{-1}$ was always 131 $\pm$ 1 and 158 $\pm$ 1 bpm which corresponded to ~70 and ~80% maximal heart rate, respectively. Although the maximal heart rate was 179 bpm in swimming, handcycling and wheelchair racing attained values up to 188 bpm. Training load (TL) was quantified by several parameters to assess their comparability. The training impulse of a five-zone model (TRIMP$_5$) was calculated by multiplying the minutes spent in each zone by their identifier (23). For a three-zone model (TRIMP$_3$), the highest (T4–T5) and lowest zones (T1–T2) were combined and multiplied analogously by 1–3 (41). The polarization index was calculated according to the literature (42). Besides these scientific procedures of quantification, the athlete's sensation of acute fatigue ("heavy arms") was subjectively recorded. This type of sensation indicates that a typical feeling of soreness following training is exceeded and may affect the following training sessions.

As a subjective measure of TL, session ratings of perceived exertion (sRPE) were recorded (43). The total load index (TLI) was calculated by multiplying training duration (min) and sRPE. Training sessions were synchronized *via* the Garmin-Connect-App and entered in an EXCEL spreadsheet to calculate TL for a whole Paralympic cycle (2017–2020). Data are expressed as a sum of 4-weeks blocks.

## Strength Training and Physiotherapy

Additional strength training and physiotherapy were assumed to be crucial for meeting the high demands on the upper extremities in PTWC, improving performance, and minimizing the risk of overuse injuries (14, 44, 45). Every strength training session was preceded by a movement preparation that included stretching and activating exercises for the upper extremity and trunk. Additionally, exercises targeting the external shoulder rotators were performed with elastic bands to improve stability and avoid muscular imbalance (46).

Stationary strength training was performed on automatically guided and software-controlled devices (Milon Industries GmbH, Emersacker, Germany) that monitored the eccentric and concentric loads of rowing, bench press, trunk flexion/extension, and pull-down exercises. Concentric failure was attained after a desired number of reps (±2) that decreased during the preparation period (starting annually in October/November). After 4–8 weeks of 2 × 20 reps (30 s rest) and 3–6 weeks of 2 × 12 reps (45 s rest), a high-intensity block of 4 × 6 reps (70 s rest) was applied for 2 weeks. Maintenance (moderate) training once a week was applied during the rest of the year. Since *M. deltoideus*, *M. biceps* brachii, and *M. trapezius* are highly activated in handcycling and are assumed to be major contributors to tonicity/discomfort (47), preventive manual therapy was applied 1–2× per week.

## Changes in the Intervention

Training contents and periodization were largely influenced by the athlete's work duration, the access to sports facilities (e.g., swimming pool), the short-term announcements of paratriathlon starting lists, and perceived discomfort/fatigue. Therefore, flexibility and trade-offs were common practices during the intervention. In June/July 2017, severe physical complaints caused by neurogenic bladder dysfunction led to a mandatory break in training. Consequently, *M. detrusor vesicae* were inhibited by annual injections of Botulinum toxin (Botox®). During a training camp in August 2019, BL attained a stress fracture of two of his ribs (ICD 10-S22.42, 6th and 7th), which resulted in a reduced TL for several weeks and rescheduling of international paratriathlon events. In 2020, the coronavirus pandemic caused several restrictions like the first lockdown (in Germany from April to May) or the closure of sports facilities that significantly affected psychological variables in amateur and recreational athletes (48). Although this also applied to BL, CML diagnosis and treatment overshadowed pandemic effects. Accordingly, vigorous training especially at high intensity had to be avoided for several months.

## Chronic Myeloid Leukemia

The most crucial challenge of this exposure was caused in February 2020. A training camp on Lanzarote had to be canceled after a few days due to spontaneous and sustained nausea, discomfort, and remarkably reduced physical performance. Initial white blood cell differentiation showed elevated myelocytes (9.0%), metamyelocytes (3%), and promyelocytes (2%). On 14 February, BCR-ABL1 transcripts (Type e13a2) of 56.8% confirmed CML (ICD10-C92.1). BL started treatment right away with the second generation TKI Nilotinib (Tasigna, 150 mg, 2-0-2). In the following months, BL started experiencing thoracic pain—especially during exercise. In October 2020, an electrocardiogram during incremental handcycling exercise demonstrated an exercise-induced left bundle branch block (ICD10-I44.7). Since this was interpreted as a side effect of Nilotinib, BL continued treatment with Dasatinib (Sprycel, 100 mg, 1-0-0). However, due to intense headache and vertigo, treatment was continued with Imatinib (Glivec, 400 mg, 1-0-0) and prescribed from November 2020 onward. In July 2021, the Imatinib dosage was reduced to 300 mg due to gastrointestinal complaints and increasing anemia (Hemoglobin toward 13 g/day). In November 2021, BL started participating in a clinical study that examines the effects of Asciminib medication (40 mg/day, 1-0-1) on BCR-ABL1 development and side effects in patients who have previously been treated with ≥2 ATP-binding site TKIs. The development of BCR-ABL1 levels over time is illustrated in **Supplementary Material 2**.

# RESULTS

## Physical Exercise Tests

In accordance with his ranking at international paratriathlon events (**Figure 1**), exercise tests demonstrated an increase in physical performance. In handcycling, $P_{\mathrm{OBLA}}$ increased from 101 W in 2017 to 158 W in 2020 (**Figure 2A**). In this period, $\dot{V}La_{max}$ decreased from 0.56 to 0.36 $mmol{\cdot}l^{-1}{\cdot}s^{-1}$. $\dot{V}O_{2peak}$ showed the highest annual values following the preparation period and tended to decrease during the competition (March to July/August) and the transition period (August/September to October). **Figure 2B** illustrates performance development in swimming. From 2015 to 2020, CSS increased from 1:34 to 1:27 $min{\cdot}100\ m^{-1}$. From 2018 to 2020, the sport-specific 750-m pace improved from 1:47 to 1:41 $min{\cdot}100\ m^{-1}$ while the sprint pace demonstrated an annual pattern. In wheelchair racing, CV increased from 2:55 $min{\cdot}km^{-1}$ in June 2018 to 2:19 in July 2019 (**Figure 2C**). Overall, postexercise BLC was found to be highest in swimming and lowest in wheelchair racing.

## TL and Training Intensity Distribution

Annual training duration increased from 414 h in 2017 to 604 h in 2019 (**Table 1**). In this period, the proportion of handcycling and wheelchair racing increased, while swimming was maintained and strength training was decreased. BL reduced his office work by 50% in 2018 and started full-time training in 2019. The distribution of TL between disciplines was similar for all measures. In 2020, the annual training duration decreased to 317 h and demonstrated a relatively high proportion of handcycling and strength training and a considerably low amount of swimming. **Figure 3A** illustrates overall TL and time spent in various intensity zones from 2017 to 2020. The least variation in overall TL occurred in 2019. TLI distribution among disciplines over time is illustrated in **Figure 3B**. Periods of increased training volume seemed to primarily result from handcycling exercise, whereas the strength TL was high during preparation and low during the competition period. During training camps, daily TLI and

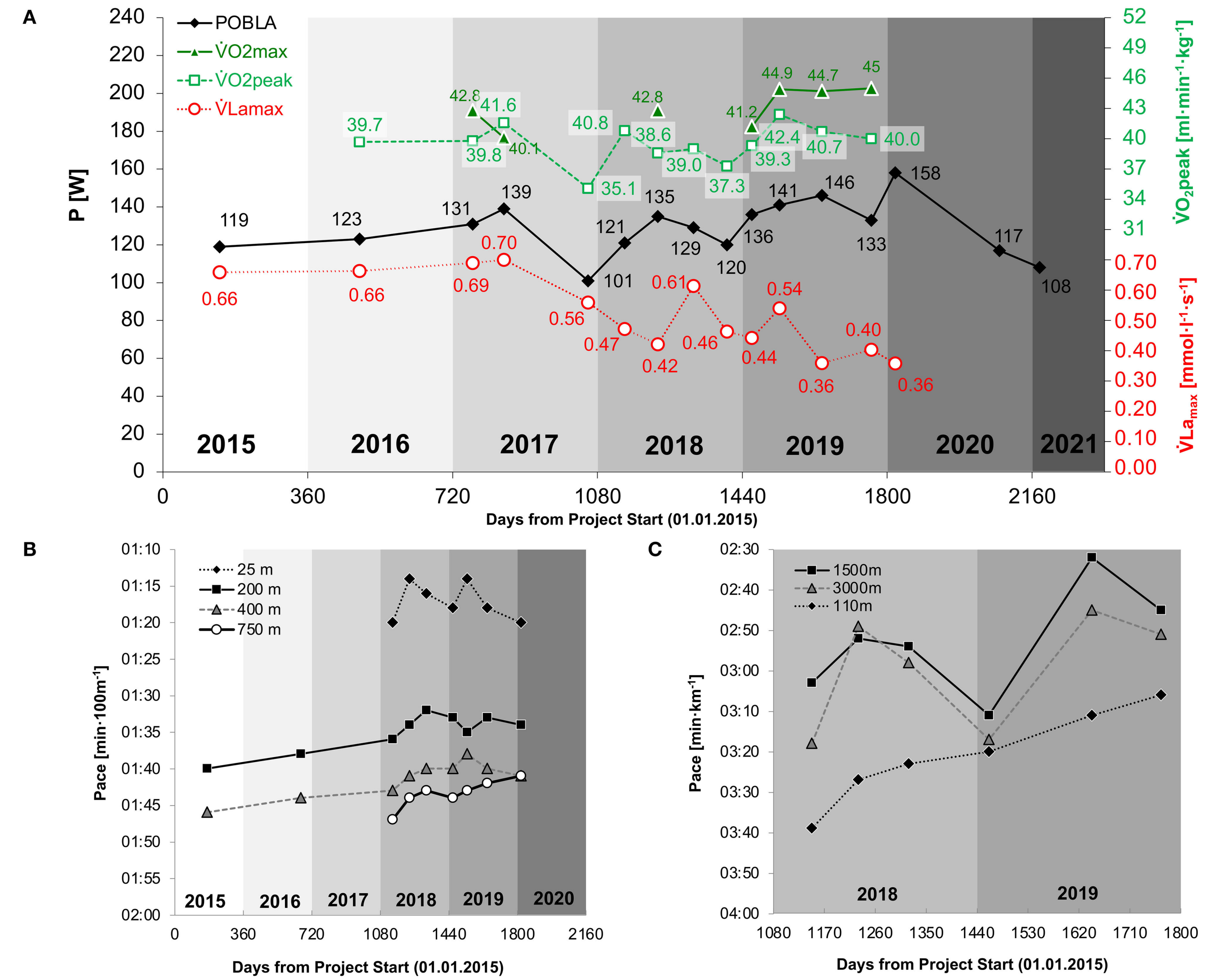


**FIGURE 2 |** Performance development in handcycling **(A)**, swimming **(B)**, and wheelchair racing **(C)** $\dot{V}La_{max}$, maximal lactate accumulation rate (mmol·l$^{-1}$·s$^{-1}$); $\dot{V}O_{2max}$, maximal oxygen uptake (ml·min$^{-1}$·kg$^{-1}$); $\dot{V}O_{2peak}$, peak oxygen uptake (ml·min$^{-1}$·kg$^{-1}$).

TRIMP$_3$ were found to be around 2,000 and 350, respectively, and were separated by rest days on a 2:1 to 3:1 basis. An example training camp from February 2019 is illustrated in **Supplementary Material 3**. Weekly training duration during the camps attained values of 20–30 h with an overall TID of ∼ (84-13-3%). TLI, TRIMP$_3$, and TRIMP$_5$ demonstrated high correlations on a weekly ($R^2$ = 92-98%) and monthly ($R^2$ = 90–97%) basis, with the highest correlation between TRIMP$_3$ and TRIMP$_5$ (refer to **Supplementary Material 4**). Sensations of acute fatigue ("heavy arms") were frequently reported (∼once per month) and were highest during a high-intensity block periodization (December 2018).

The TID showed a pyramidal periodization and a shift toward high volume training from 2017 (77-17-6%) to 2019 (88-8-4%) (refer to **Supplementary Material 5**). In 2020, overall TID was found to be 94-5-1%. The highest percentage of high-intensity was found in wheelchair racing (∼10%), followed by swimming/handcycling (∼5%) and strength training (<0.5%). Training intensity of stationary strength training demonstrated an increase during preparation periods as well as over the years (**Supplementary Material 6**).

## BCL-ABL1 Transcripts

From an initial value of 56.8 in February 2020, BCL-ABL1 transcripts decreased substantially after 41, 91, and 248 days to values of 5.65, 0.0416, and 0.008, respectively (**Supplementary Material 2**). Reduced BCL-ABL1 transcripts imply a reduced tumor load indicating that a patient is positively responding to the applied therapy.

**TABLE 1 |** Yearly and discipline-specific training load (TL) during the Olympiad.

| Parameter | Discipline | 2017 | 2018 | 2019 | 2020 |
|---|---|---|---|---|---|
| Duration (h) | Overall | 414 | 557 | 604 | 317 |
| | Swimming | 83 (20%) | 110 (20%) | 136 (22%) | 24 (8%) |
| | Handcycling | 161 (39%) | 258 (46%) | 271 (45%) | 182 (57%) |
| | Wheelchair racing | 54 (13%) | 82 (15%) | 116 (19%) | 46 (15%) |
| | Strength training | 91 (22%) | 97 (17%) | 79 (13%) | 59 (19%) |
| $TRIMP_3$ | Overall | 32,279 | 40,744 | 42,289 | 20,406 |
| | Swimming | 6,832 (21%) | 8,670 (21%) | 9,650 (23%) | 1,556 (8%) |
| | Handcycling | 12,266 (38%) | 18,238 (45%) | 18,830 (45%) | 11,200 (55%) |
| | Wheelchair racing | 4,715 (15%) | 6,643 (16%) | 8,909 (21%) | 3,215 (16%) |
| | Strength training | 5,573 (17%) | 5,898 (14%) | 4,783 (11%) | 3,526 (17%) |
| $TRIMP_5$ | Overall | 44,731 | 59,604 | 67,487 | 33,337 |
| | Swimming | 9,778 (22%) | 13,021 (22%) | 15,382 (23%) | 2,567 (8%) |
| | Handcycling | 17,486 (39%) | 28,103 (47%) | 31,910 (47%) | 20,064 (60%) |
| | Wheelchair racing | 6,896 (15%) | 10,092 (17%) | 14,820 (22%) | 5,626 (17%) |
| | Strength training | 6,229 (14%) | 6,445 (11%) | 5,200 (8%) | 3,716 (11%) |
| TLI | Overall | 107,083 | 188,383 | 198,002 | 82,423 |
| | Swimming | 22,951 (21%) | 39,678 (21%) | 43,059 (22%) | 6,247 (8%) |
| | Handcycling | 46,408 (43%) | 94,084 (50%) | 93,294 (47%) | 48,028 (58%) |
| | Wheelchair racing | 13,133 (12%) | 28,486 (15%) | 39,910 (20%) | 13,355 (16%) |
| | Strength training | 24,593 (23%) | 26,348 (14%) | 21,749 (11%) | 14,794 (18%) |

## DISCUSSION

This case report represents one of the most extensive descriptions of complex exercise testing and long-term training monitoring in Paralympic sports. We demonstrated that the physical abilities of a paratriathlete in the wheelchair category improved with increasing TL and reductions in office work duration. Training intensity distribution (TID) showed a pyramidal periodization and shift toward low-intensity training. However POBLA increased and $\dot{V}La_{max}$ decreased over the years, $\dot{V}O_{2peak}$ indicated an annual pattern and attained the highest values following the preparation period. With CML diagnosis and its treatment, TL (especially with respect to high intensities) and physical abilities substantially decreased. This hindered the athlete from competing in international events and ultimately from qualifying for the Paralympic Games in Tokyo.

A similar description was recently provided for a female paraswimmer during a Paralympic cycle (4). Although annual training hours and their development over the years were similar to this case study, the TID of the paraswimmer demonstrated an even higher percentage of low-intensity training. This is influenced by the fact that the authors used a session goal approach to determine TID, whereas a time-in-zone approach was used in this case study (49). When compared with previous studies in handcycling (5, 6), the performance gains were lower, and TID was less polarized in this study. However, these authors used time in power zones rather than heart rate zones to calculate TID, which was shown to differ between methods (50)—especially during high-intensity sessions (51). Future studies need to examine these differences in handcycling.

Training load was quantified by means of separate internal measures (TLI and TRIMP) in order to ensure comparability and standardization between paratriathlon disciplines. The high correlation between these parameters indicates an overall agreement between objective and subjective measures of TL. However, we noticed that the discrepancy between subjective and objective TL (**Supplementary Material 3**) coincided with acute fatigue (“heavy arms”), especially when subjective TL was substantially higher when compared with measures of objective TL. Thus, considering both types of loads and their discrepancy could be helpful for carefully monitoring daily TL and preventing acute fatigue and non-functional overreaching, especially during training camps. Despite the assumptions of previous research (52), recent findings indicate that ratios between subjective and objective measures (sRPE:TRIMP) do not provide additional information to monitor fatigue in cyclists (53). Furthermore, it was shown that TLI is not associated with alterations in physical capacity, whereas the time attained in heart rate zone 2 of 3 significantly correlated with improvements in $\dot{V}O_{2peak}$ during the preparation for the HandbikeBattle (54). This is in line with the initially high $\dot{V}O_{2peak}$ values (relative to his development) of BL (when a lot of zone 2 training was performed) and the fact that $\dot{V}O_{2peak}$ only slightly increased with high-volume low-intensity training (2018–2019).

A similar TID as in this case study was reported for a male long-distance paratriathlete with below-the-knee-amputation (8). The increase in low-intensity percentages over time might be due to the mere increase in training volume and/or the fact that the BL became increasingly patient about performing his training sessions in the prescribed training zones. However,

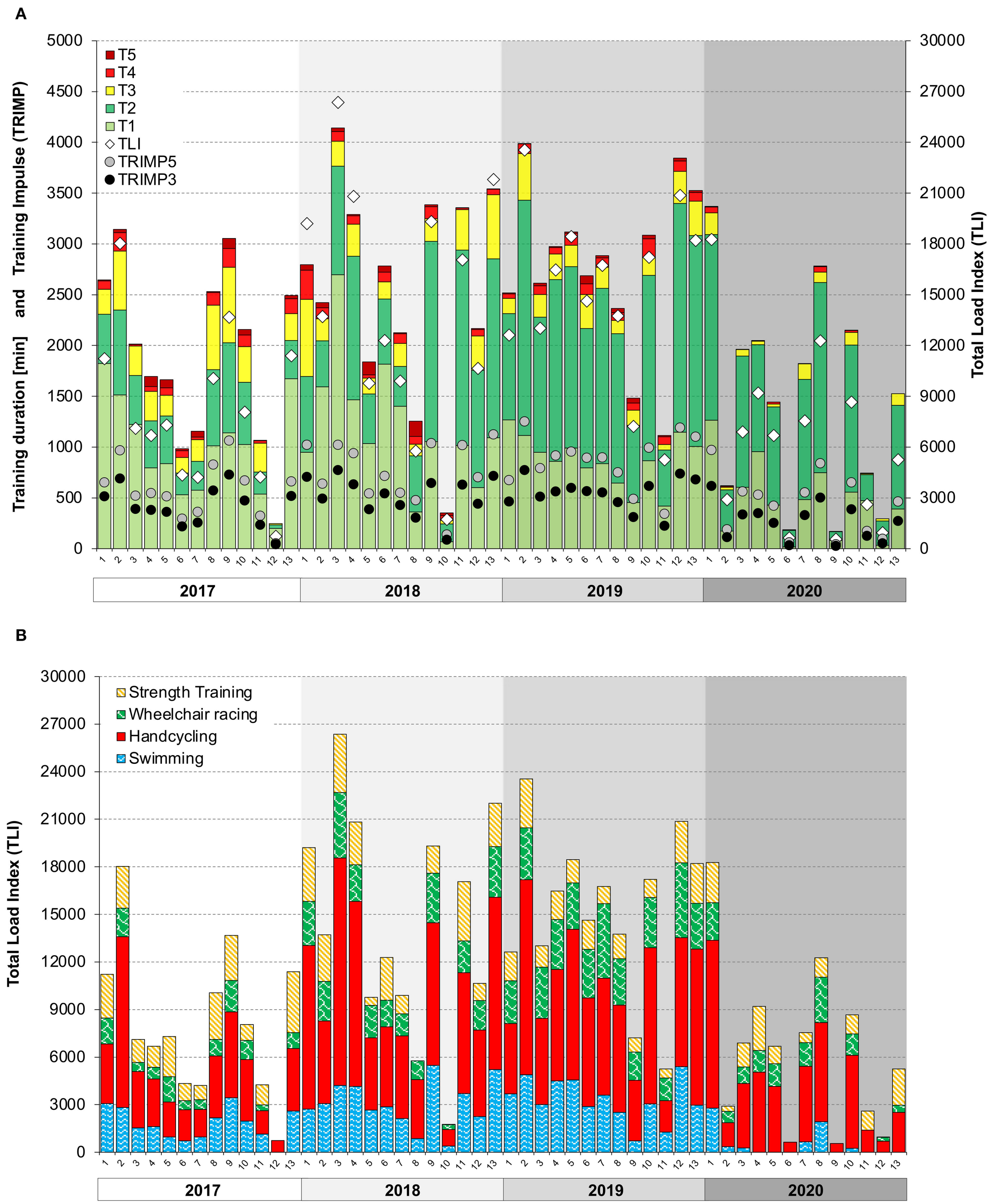


**FIGURE 3 |** Training monitoring from 2017 to 2020. **(A)** Total training duration in every exercise zone (T1–T5) over time in 4-week blocks (13 for each year). TRIMP and total load index (TLI) over time in 4-week blocks (13 for each year). ($TRIMP_3$), training impulse based on a three-zone model; $TRIMP_5$, training impulse based on a five-zone model; TLI, total load index; sRPE × training duration [min]. **(B)** TLI in every discipline over time in 4-week blocks (13 for each year). TLI, total load index; sRPE × training duration (min).

this might also indicate that the training prescription of BL is overly focused on high-volume rather than high-intensity training. In fact, we tried to apply high-intensity training in a block periodization, which has been shown to be an adequate training strategy to improve $\dot{V}O_{2max}$ (55). However, the athlete did not tolerate more than three high-intensity interval sessions in a week due to acute fatigue. This might be due to the fact that wheelchair triathletes purely rely on their upper extremities during training and activities of daily living. Therefore, the overall higher load applied on the upper extremities increases the risk for acute fatigue, which is less severe in conventional triathletes. We experienced that the duration at a high intensity (heart rate zones 4 and 5) was higher and more easily triggered by performing wheelchair racing rather than handcycling. However, this discipline comes along with a substantially higher shoulder load (56). Due to the BL's medical complaints, the fluctuations of TL between training blocks were higher compared with previous studies (4, 6, 8), which affected training consistency, which is observed in *"full-time, year-round athletes"* (57).

In our case study, reducing employment increased the athlete's amount of available time and energy which allowed for more (intense) training. This increase in training duration is accompanied by an enhanced training adaptation as documented by the exercise tests. This increase in training and recovery duration is facilitated by corresponding sponsorship that was not constantly available for this athlete. Therefore, at first, training camps were tightly scheduled due to restricted training time. Although previous studies reported an average training volume during training camps of 137 ± 33% of preceding (regular) TL in the absence of acute fatigue or excessive stress (58), BL experienced a considerably higher load during the initial training camps (>200%) when he frequently reached his physical limits. From 2018 onward, we provided a less extensive and more flexible schedule in the following camps, performed daily (objective and subjective) monitoring, and implemented more (relaxing) rest days that substantially improved the feasibility, recovery, and mood. In future projects, the latter might be complemented by the Profile of Mood States Questionnaire that was successfully applied to wheelchair marathoners (7) and elite paratriathletes (58). To prevent acute fatigue and improve training quality during training camps, suitable strategies of micro-periodization were established. For example, combining high-intensity wheelchair racing in the morning followed by low-intensity handcycling was appropriate to properly exercise and recover on intensified days. Rest days consisted of an easy swim session and moderate strength training.

The characteristic pattern of $VO_{2peak}$ development within each season (with the highest values observed after the preparation period) is similar to those reported in a world-class middle-distance runner (59). We assumed that the observed pattern shows a typical build-up, followed by a plateau and subsequent decline in performance during detraining. However, the athlete's paratriathlon performance in the competition was less affected over the respective years in terms of overall and split times. Jones demonstrated that world-class endurance performance can improve over several years despite a decrease/stagnation of $\dot{V}O_{2max}$ as long as submaximal parameters improve (60, 61). In our case, improvements in maximal fat oxidation and/or movement economy might be the reason for the less severe decline in sport-specific performance. Especially during long rides, an improved "durability" in terms of an improved tolerance and less severe increase in heart rate during prolonged exercise were observed over the years which might be due to the high volumes of low-intensity training (62).

According to previous simulation approaches of glycolysis and oxidative phosphorylation, $\dot{V}La_{max}$ and $\dot{V}O_{2max}$ interact to determine lactate threshold (32). In simple terms, it is assumed that net lactate production results from the difference between the rate of lactate formation (as a percentage usage of $\dot{V}La_{max}$) and the rate of lactate removal (which is assumed to be proportional to oxygen uptake). As such, maximal lactate steady state demonstrates the highest equilibrium of lactate formation and removal (net lactate production = 0) (63). This relationship is indicated by following the development of $P_{OBLA}$ in **Figure 2A**. If we assume that $P_{OBLA}$ is improved by an increase in $\dot{V}O_{2peak}$ and/or decrease in $\dot{V}La_{max}$ (and *vice versa*), we can qualitatively estimate the alterations in $P_{OBLA}$. For example, $P_{OBLA}$ decreased from 135 to 129 W despite a constant $VO_{2peak}$ (probably) due to a huge increase in $\dot{V}La_{max}$ immediately after a training camp. This is in line with a previous study highlighting that $\dot{V}La_{max}$ significantly decreased after only 2 weeks of sprint interval training in trained cyclists (64). However, research on $\dot{V}La_{max}$ adaptations is generally sparse. The reduced $\dot{V}La_{max}$ values in our study are in accordance with previous research in ultra-endurance cycling, demonstrating a decrease in $\dot{V}La_{max}$ during a prolonged period of high-volume low-intensity training (65). Since $\dot{V}La_{max}$ was found to be increased by various forms of resistance exercise (66), an intensified fine-tuning of strength and sport-specific training contents was applied. Our preliminary findings demonstrated that $\dot{V}La_{max}$ is affected by exercise modality, highest in swimming and lowest in wheelchair racing, which might be due to the usage of muscle mass. Although the reliability of $\dot{V}La_{max}$ has been sufficiently assessed in handcycling (26, 27), future studies need to examine $\dot{V}La_{max}$ in swimming and wheelchair racing.

There are several limitations of this case report that need to be mentioned. Given the high number of contextual variables (e.g., medical, logistical, nutritional, and social), training was frequently adapted to the acute circumstances, which makes it challenging to highlight causations. In PTWC triathlon, the need for various materials (handcycling and racing wheelchair), dependence on barrier-free facilities, and the pure focus on upper extremity locomotion demonstrate substantial constraints that the athletes have to overcome in order to be competitive. Especially as far as side effects of CML and its treatment are concerned, it is likely that experienced fatigue and reduced physical activity interact to ultimately decrease performance (17). Furthermore, TID is purely based on heart rate zones and as such hardly comparable with studies using session goals or time in power zone approaches.

In conclusion, this case report illustrates the training monitoring and performance development of a triathlete with SCI and CML during a Paralympic cycle. We demonstrated the need for careful training prescription in PTWC triathletes to improve performance in the absence of acute fatigue, overuse injuries, and non-functional overreaching. We encouraged athletes and coaches to refrain from overly extensive and/or intense training schedules and recommend the application of objective and subjective monitoring tools. This is stressed by the high demands on the upper extremities in wheelchair triathletes, who require special support, sponsors, and training prescriptions.

## ATHLETE PERSPECTIVE

"In my opinion, having a great team of coaches and supporters was essential for gaining the last boost during the highs and lows of this journey. The same applies to those who gave medical assistance, which was highly important—especially in my case. Health complaints (e.g., bladder infections) kept me from performing in training and competitions, which was also mentally challenging. Accepting CML diagnosis and its consequences for fulfilling my goal took some time, even though my SCI-background helped to cope. I would like to share my experiences with sports and diseases to inspire others in the future."

## DATA AVAILABILITY STATEMENT

The original contributions presented in the study are included in the article/**Supplementary Material**, further inquiries can be directed to the corresponding author.

## ETHICS STATEMENT

Ethical review and approval was not required for the study on human participants in accordance with the local legislation and institutional requirements. The patients/participants provided their written informed consent to participate in this study. Written informed consent was obtained from the individual(s) for the publication of any potentially identifiable images or data included in this article.

## AUTHOR CONTRIBUTIONS

OJQ planned the triathlon-specific training, participated in training camps, collected and analyzed the data, and drafted the manuscript. BL volunteered in this study, performed the training and exercise tests, and provided insights into his perspective. FL and PB were handlers at competitions and participated in training camps. FL provided logistical planning. PB planned and coached strength and conditioning contents and advised BL on aspects of nutrition and supplements. TF provided medical check-ups and supervised medical aspects of the project. All authors reviewed, edited, and approved the manuscript.

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## SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fresc.2022.867089/full#supplementary-material

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# Development of Tests for Arm Coordination Impairment in Paralympic Classification

*Viola C. Altmann*[1,2,3*], *Nadine Hendriks*[1,4], *Eline A. Lammens*[1] *and Mariska Janssen*[1,4]

*[1] Expertise Centrum, Klimmendaal Rehabilitation Centre, Arnhem, Netherlands, [2] Peter Harrison Centre of Disability Sport, School of Sport, Exercise and Health Sciences, Loughborough University, Loughborough, United Kingdom, [3] World Wheelchair Rugby, Sheffield, United Kingdom, [4] Department of Rehabilitation, Radboud University Medical Center, Donders Institute for Brain, Cognition and Behaviour, Nijmegen, Netherlands*

**Background:** In Paralympic sport, classification of impairment with the ability to detect misrepresentation of abilities is mandatory. In wheelchair rugby, there is currently no objective method to classify arm coordination impairment. In previous research, sufficient correlation between the spiral test (ST) and activity in wheelchair rugby was found in athletes with coordination impairment. However, the ST depends on maximum voluntary effort.

**Purpose:** To assess if the ST is an objective test for arm coordination impairment, in which maximum voluntary effort can be distinguished from intentional misrepresentation. The aims of this study were to (1) assess the test-retest reliability of the ST and (2) assess if Fitts's law is applicable to the ST.

**Methods:** Nineteen volunteers without impairments performed two sessions with three STs per arm. The STs were projected and measured on a tablet and had three different indices of difficulty based on differences in spiral width. The time to complete the spiral was measured and a penalty time was added for each time the borderline of the spiral was touched (3 s) or crossed (5 s).

**Results:** Test-retest reliability was assessed using a Bland-Altman analysis and showed limits of agreement that were wider than the margins of 2SD from the group mean. Repeated measurement correlation coefficients between the index of difficulty according to Fitts's law and the movement time were $> 0.95$ ($p$-value $< 0.001$) for both test and retest. A *post-hoc* optimisation of penalty times revealed an optimum penalty time of 2.0 s for the dominant arm and 2.5 for the non-dominant arm for any contact with the margins of the spiral.

**Conclusions:** The ST has sufficient test-retest reliability and Fitts's law is applicable. Therefore, it is a promising option for classification of arm coordination impairment with the option to distinguish intentional misrepresentation from maximum voluntary effort.

**Keywords: wheelchair, coordination, impairment test, wheelchair sports, Paralympic**

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****Correspondence:***
*Viola C. Altmann*
*research@altmannen.nl*
*orcid.org/0000-0002-0671-8115*

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# INTRODUCTION

The Paralympic Games were founded in Great Britain after the Second World War as a sport event with the goal to enhance participation of wounded veterans in society. Over the years, sports for veterans became an international event and were connected to the Olympic Games in 1956 (1). Nowadays, the Paralympic Games are the world's third largest sports event, with athletes competing from all over the world. In 2016, broadcasting of the Paralympic Games was covered in 154 countries, with 4.1 billion people watching (2). In the early days, patients were competing against other patients with the same health condition, like Spinal Cord Injury (SCI) or amputations, in only a few sports (1). Nowadays in the Paralympic summer games, over 4,000 professional athletes compete in 22 sports and earn their daily living by it (2).

To guarantee an attractive and fair competition, the best athlete should win, and not the one who is the least impaired, and there should be enough athletes to compete against. To achieve this, athletes compete in categories (classes) in which the impact of impairment on the ability to perform should be similar. The process that leads to categorizing athletes is called classification. The aim of classification is that winning or losing the competition is based on training, motivation, talent and skills rather than the severity of impairments (3). To determine the optimal class for each athlete, testing of impairment is mandatory. However, there is a risk that athletes will try to misrepresent their abilities, to try to compete in a class with athletes with more severe impairments than their own. This is called Intentional Misrepresentation (IM). In an impairment test for classification, it should be possible to distinguish IM from Maximum Voluntary Effort (MVE). Currently, most classification systems are based on expert opinion of experienced classifiers. However, with the increasing professionalism of Paralympic sports, the International Paralympic Committee stated that classification should develop toward Evidence-Based Classification, in which the number and the borderlines of the classes per sport should be supported by empirical data (3, 4).

One of the Paralympic Sports is Wheelchair Rugby (WR). It was developed by and for athletes with tetraplegia due to spinal cord injury (SCI) in 1977. Since 2000, WR is a full medal sport in the Paralympic Games. There are more than forty countries that actively participate in WR, or who are developing WR programmes within their nation. WR was developed for athletes with SCI, but athletes with other health conditions, such as neuromuscular diseases, cerebral palsy (CP) and limb deficiencies are also allowed to compete. WR as a Paralympic sport is a mixed gender team sport with the age of elite players varying on average between 20 and 35 years. In classification in WR, athletes can be grouped in one of the seven classes: 0.5, 1.0, 1.5, 2.0, 2.5, 3.0, and 3.5 points, based on arm and trunk impairment. During competition, four athletes are on the court and the total points for one team cannot exceed eight points (5). At this moment, the impairment tests to allocate scores for arm impairment are based on muscle strength, because WR was developed for athletes with SCI. However, the number of WR athletes with coordination impairment is expected to be much higher in the future since the incidence of CP is 39–150 times as high as the incidence of SCI (6, 7). In contrast to classification of arm strength impairment and trunk impairment (including all neuromusculoskeletal impairment types), which are largely evidence based (8–13), there is very limited evidence to support the classification of arm coordination impairment (14).

None of the current classification systems in Paralympic Sports include objective, Evidence Based impairment tests for arm coordination impairment (15, 16). The spiral test (ST) is a multilevel, parsimonious tests that may be suitable for classifying arm coordination impairment in WR. In previous research, a moderate-strong correlation between the ST and activities in WR was found. Furthermore, athletes with coordination impairment could be distinguished from individuals without impairments, so that minimum impairment criteria for eligibility of arm coordination impairment could be established (14). However, the results of the ST depend on MVE and so far, it has not been assessed if IM can be distinguished from MVE in the ST. IM The next step that needs to be taken in the development of Evidence Based Classification is to assess if this distinction can be made in the ST.

In general, a method for differentiating between MVE and IM must satisfy two main criteria: (a) sufficient test-retest reliability and (b) detectable differences between the results achieved under the presence and absence of MVE. The test-retest reliability of the ST has not been assessed so far (14). To detect differences between the presence and absence of MVE, Fitts's law is a promising option (17, 18). Coordination affects both movement accuracy and movement speed. Faster movements are less accurate and higher accuracy is achieved at lower speeds (19). In Fitts's law the relationship between the movement accuracy and precision is reflected in the index of difficulty (ID). Movement time and different IDs show a significant linear relationship in tests with movements between two targets in which several target sizes and target widths result in different IDs (20). If a movement time is significantly different from that line in a minimum of three tests, this is a sign that there was no MVE in one of the tests (17). At this moment, it is not clear if Fitts's law is also applicable to the ST that was used in previous classification research in WR.

In this study we further investigated the ST as a possible objective test for arm coordination impairment in WR athletes. To elaborate the options for distinguishing MVE from IM, the aims of this study were to (1) assess the test-retest reliability of the ST and (2) assess if Fitts's law is applicable to the ST.

# MATERIALS AND METHODS

## Participants

A convenience sample of nineteen adults without impairments in the same age range as athletes in Paralympic sports, (mean age of 27 years; range: 19–33), participated in this cross-sectional study. Eighty-nine percent were male ($n = 17$) and eleven percent were female ($n = 2$) similar to the ratio of men and women in wheelchair rugby (21). We selected this population with similar age and gender to WR athletes, because in previous studies there appeared to be an impact of age and gender

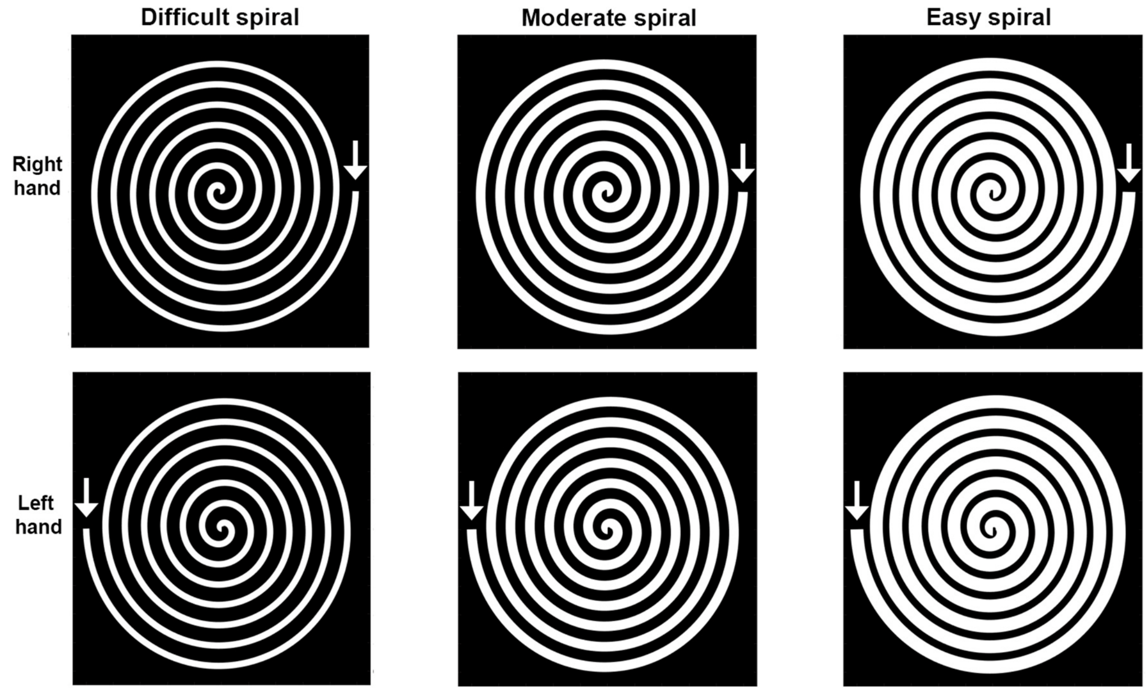


**FIGURE 1 |** Spiral tests (ST) with different index of difficulty for the left and the right hand. Difficult (spiral width 3.5 mm), Moderate (spiral width 5.3 mm), and Easy (spiral width 7.1 mm).

on coordination (22). Three of the nineteen participants were left dominant, one participant was ambidextrous and fifteen participants were right dominant. Because an ambidextrous person is expected to use his right arm more than his left arm in a society with a majority of right handed persons, the ambidextrous participant's data were analyzed as right dominant. All participants gave written informed consent prior to participating, and the study was performed in accordance with the Declaration of Helsinki (2013) developed by The World Medical Association (32). The study has been assessed by the Medical Ethical Committee of the Netherlands, region Arnhem and Nijmegen, (registration number 2021-13107) and received local approval of the scientific committee of Klimmendaal rehabilitation center.

## Spiral Test

All participants performed six spiral tests (STs) with a pen, three spirals with different levels of difficulty with the dominant arm and three spirals with the same, different levels of difficulty with the non-dominant arm. The level of difficulty was determined by the widths of the spirals, i.e., 3.528, 5.291, and 7.056 mm. Each spiral had seven turns with a length of 2,204,771 mm, resulting in three different IDs, i.e., difficult: 901.6, moderate: 601.1 and easy: 450.8. For each version there was a right-handed and a mirrored left-handed spiral. The right-handed spiral was completed clock-wise and the left handed counter clock-wise, see **Figure 1**. Spiral lengths were calculated using the function "arclength.m" in Matlab, which calculates the length of a path based on its x and y coordinates. IDs were calculated using Formula 1, where A is the length of the spiral and W is the spiral width (23).

$$Formula\ 1 : ID\infty = \frac{A}{W \ln 2} \tag{1}$$

This formula differs a little for the original formula for Fitts's law which was developed for a movement with only a fixed start and finish target, but a free movement trajectory between the start and the finish target. In the ST, participants had to stay the entire task with the pen within the white spiral. So besides the start and the finish, the whole movement trajectory was fixed. **Figure 1** shows an example of the right- and left-handed spirals. Real size spirals are available in **Supplementary Material 1**.

The STs were performed on a digitized graphic tablet (Wacom Cintiq 16, model nr: DTK1660K0B, 2019) (24). Calibration of the pen was performed before the start of every measurement. Participants were asked to draw a line within the spiral as quickly and accurately as possible from the arrow to the center (**Figure 1**). The primary outcome measure for this test was the total time in which the spiral was completed, indicated as movement time. A 3 s penalty was added for each time the borderline between the spiral and the black area was touched with the pen, and 5 s penalty was added for each time the pen was in the black area (22). However, the spirals we used had a different lay-out with a white trajectory on a black background, compared to the original research in which the windings were in between two black lines on a white paper. Touching the lines was similar in the two lay-outs. However, crossing the line to end up in another winding was possible in the original test, but was unlikely in the lay-out

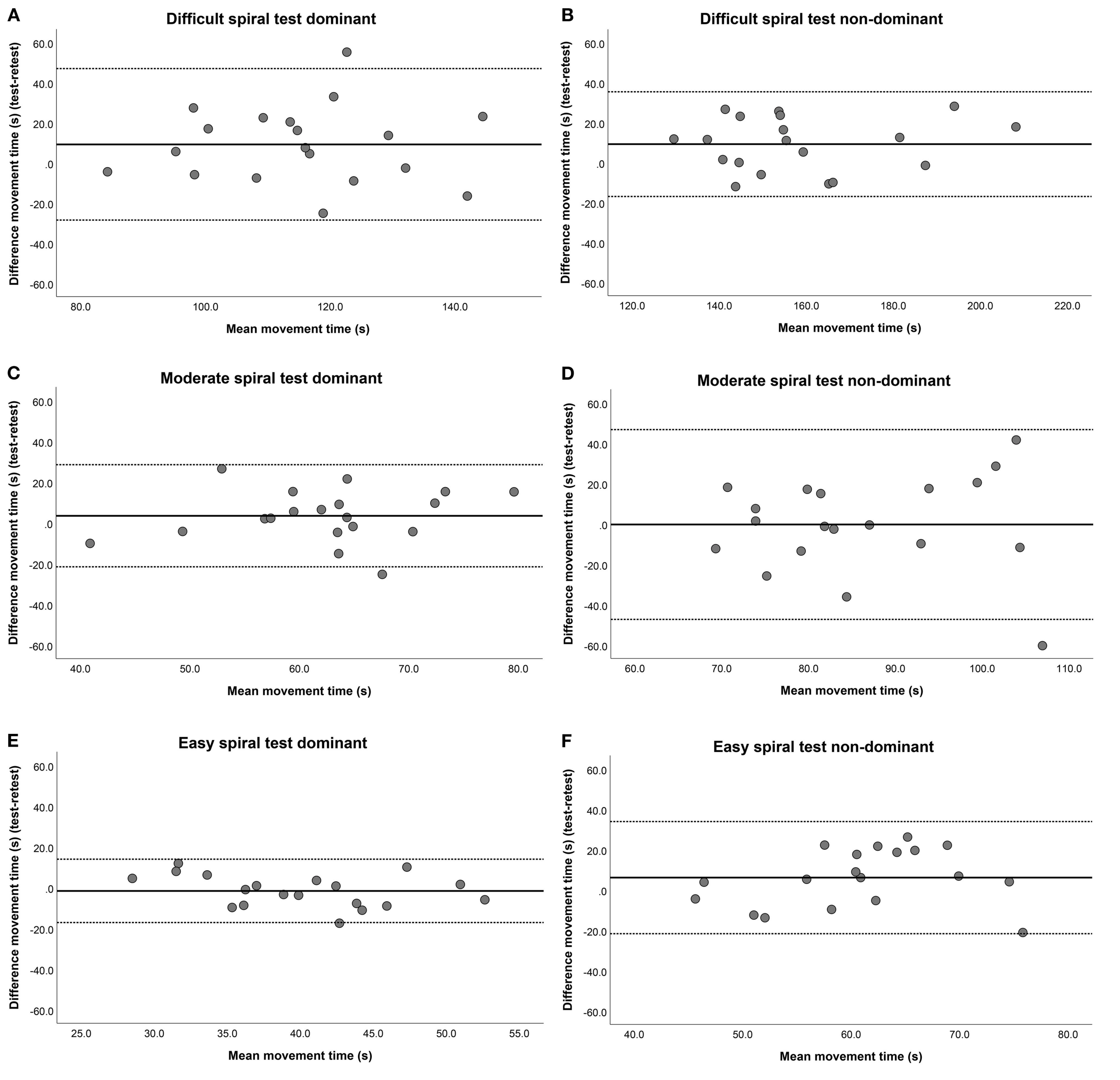


**FIGURE 2 |** Bland-Altman plots for test-retest reliability. **(A)** ST difficult dominant side, **(B)** ST difficult non-dominant side, **(C)** ST moderate dominant side, **(D)** ST moderate non-dominant side, **(E)** ST easy dominant side and **(F)** ST easy non-dominant side. The horizontal lines represent the mean difference (Bias) and two standard deviations, i.e., LOA (dashed lines) of the differences between the movement time (s) at test and the movement time (s) at retest.

we used. Therefore, we decided to use the original penalty times, but to also perform an optimisation of the penalty times for this new lay-out used in the present research.

## Test Protocol

All participants were seated in an everyday wheelchair without armrests with the brakes on while testing (Summit Benelux BV, Deventer, the Netherlands). The tablet was positioned on a height adjustable table, so the shoulder was in a neutral position and the elbow was in 90° flexion. Participants performed the three STs per arm in one session per day. The same STs were repeated on another day, 1–2 weeks apart. The order of the ST was randomized per arm and per day, so the order could be different for each arm and on each testing day. STs were recorded with a video camera and execution time was measured with a stopwatch in s.

## Data Analysis

Test-retest reliability was assessed using a Bland-Altman analysis to determine mean bias, limits of agreement (LOA) and 95%-confidence intervals (CI) of the STs. Bland-Altman analysis was used as it gives insight in both reliability and agreement of the ST which is important when the ST is going to be used in paralympic classification (25). No standard cut-off values for sufficient reliability exist in the literature for the difference between test and retest values (26). In previous research, it was possible to distinguish athletes with coordination impairment from volunteers without impairment, with a test accuracy of 93.5% using 2 standard deviations from the mean (14). Therefore, in this study STs were deemed reliable, if differences between test and retest times (i.e., limits of agreement) were within two standard deviations from the group mean. In addition, linear regression analyses were performed to examine if there was proportional bias in the data.

To test if Fitts's law was applicable to the data, we calculated repeated measure correlation coefficients (Rmcorr) between ID and movement time (27, 28).

There did not appear to be significant restriction of range or gross violations of normality based on Shapiro-Wilk test for normality and inspection of the histograms. A *post-hoc* optimisation of the penalty times was done, using a range of penalty times from 0 s to 3 s with 0.5 s intervals for both touching and crossing the lines. For this *post-hoc* optimization of we recalculated the Rmcorr for the data corrected for different penalty times for both test and retest. The penalty time with the highest Rmcorr was selected as the optimal penalty time.

## RESULTS

**Figure 2** shows the Bland-Altman plots of all STs. On average the large spirals have the narrowest LOA. **Table 1** shows the average times of test and retest per spiral, the difference between test and retest and the LOA. Most spiral tests showed no significant fixed bias, except for the difficult spiral with the non-dominant arm. That spiral had a fixed bias of 9.2 s. Furthermore, none of the spiral tests showed proportional bias. Finally, for most spiral tests the LOA were wider than the margins of 2SD of the group mean, indicating sufficient test-retest reliability, except for the most difficult spiral test with the non-dominant arm. For that test the LOA were smaller than 2SD of the group mean **Table 2** shows the absolute movement time, nr. of penalties and corrected movement time for test and retest per condition. Although, on average the differences in movement time and the nr. of penalties between test and retest were small. Individually, there could large differences in (corrected) movement times and nr. of penalties, which can be seen in the outliers displayed in **Figures 2A,C,D**.

Regarding the applicability of Fitts law to the spiral test, the Rmcorr between ID and movement time was 0.97 ($p$-value < 0.001) for the test at the dominant side and 0.96 ($p$-value < 0.001) for the retest at the dominant side. For the non-dominant side Rmcor was 0.95 ($p$-value < 0.001) for the test and 0.95 ($p$-value < 0.001) for the retest. Indicating that Fitts' law is applicable to the data. **Figure 3** shows the relationship between ID and movement time for the dominant and non-dominant side. Both individual data and the average are shown.

**Table 3** shows the *post-hoc* optimisation of the penalty times, which showed the best fit to Fitts's law for 2.5 s for the dominant

**TABLE 1** | Bland-Altman analysis for test-retest reliability.

| ST | Side | *N*-valid | Test Mean (Sd) | Retest Mean (SD) | Group mean (test + retest)/2 Mean (SD) | Mean difference (LOA) | 95% CI | Fixed bias | Regression | *p*-value | Proportinal bias | LOA < 2 SD group mean |
|---|---|---|---|---|---|---|---|---|---|---|---|---|
| Difficult | Dominant | 19 | 119.8 (18.6) | 110.6 (18.3) | 115.2 (18.8) | 9.2 (−28.5 to 47.0) | (−0.1 to 18.5) | No | 0.018 | 0.940 | No | Yes |
| Difficult | Non-dominant | 19 | 163.1 (22.5) | 153.9 (21.2) | 158.5 (22.1) | 9.2 (−17.0 to 35.4) | (2.8 to 15.6) | Yes | 0.103 | 0.676 | No | No |
| Average | Dominant | 19 | 64.1 (11.7) | 60.7 (10.1) | 62.4 (10.9) | 3.4 (−21.6 to 28.5) | (−2.7 to 9.6) | No | 0.159 | 0.516 | No | Yes |
| Average | Non-dominant | 19 | 86.3 (17.3) | 86.6 (16.9) | 86.4 (16.9) | −0.3 (−47.3 to 46.7) | (−11.9 to 11.2) | No | 0.028 | 0.910 | No | Yes |
| Easy | Dominant | 19 | 39.2 (6.4) | 40.8 (8.8) | 40.0 (7.7) | −1.7 (−17.2 to 13.9) | (−5.5 to 2.2) | No | −0.349 | 0.143 | No | Yes |
| Easy | Non-dominant | 19 | 64.0 (12.1) | 57.9 (9.8) | 60.9 (11.3) | 6.1 (−21.7 to 33.9) | (−0.7 to 13.0) | No | 0.207 | 0.395 | No | Yes |

**TABLE 2 |** Absolute movement time, nr. of penalties and corrected movement time for test and retest per condition.

| | | | Test | | | | Retest | | | |
|---|---|---|---|---|---|---|---|---|---|---|
| ST | Side | *N*-valid | Movement time<br>Mean (SD) | Nr. Penalties line touched (3 s penalty)<br>Mean (SD) | Nr. Penalties line crossed (5 s penalty)<br>Mean (SD) | Corrected movement time<br>Mean (SD) | Movement time<br>Mean (SD) | Nr. Penalties line touched (3 s penalty)<br>Mean (SD) | Nr. Penalties line crossed (5 s penalty)<br>Mean (SD) | Corrected movement time<br>Mean (SD) |
| Difficult | Dominant | 19 | 53.2 (18.9) | 15 (6) | 4 (4) | 119.8 (18.6) | 47.9 (13.7) | 15 (4) | 4 (3) | 110.6 (18.3) |
| Difficult | Non-dominant | 19 | 61.5 (24.2) | 16 (5) | 10 (5) | 163.1 (22.5) | 57.2 (18.4) | 18 (5) | 9 (4) | 153.9 (21.2) |
| Average | Dominant | 19 | 37.8 (12.2) | 6 (4) | 2 (1) | 64.1 (11.7) | 36.4 (9.2) | 6 (2) | 1 (2) | 60.7 (10.1) |
| Average | Non-dominant | 19 | 46.3 (17.4) | 6 (4) | 4 (3) | 86.3 (17.3) | 42.8 (11.9) | 7 (3) | 5 (3) | 86.6 (16.9) |
| Easy | Dominant | 19 | 30.1 (7.8) | 2 (1) | 1 (1) | 39.2 (6.4) | 30.4 (7.5) | 2 (2) | 1 (1) | 40.8 (8.8) |
| Easy | Non-dominant | 19 | 36.3 (10.9) | 5 (2) | 3 (2) | 64.0 (12.1) | 35.6 (8.0) | 4 (3) | 2 (2) | 57.9 (9.8) |

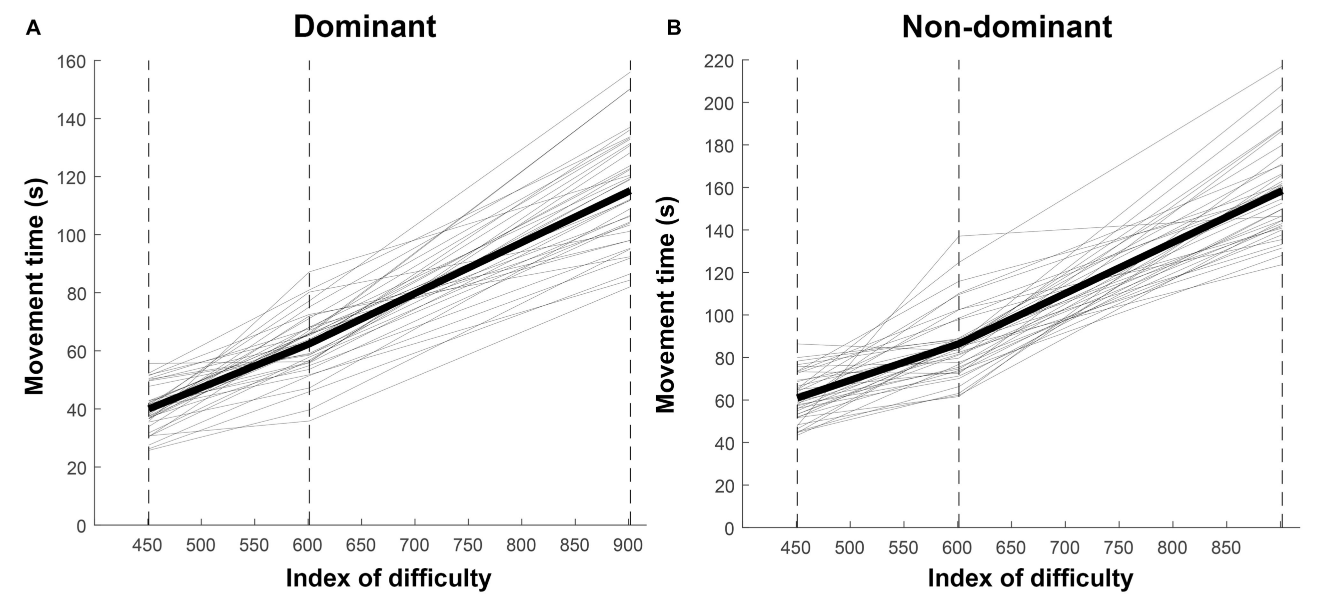


**FIGURE 3 |** Line graph of movement time relative to index of difficulty. **(A)** dominant side and **(B)** non-dominant side. The thin lines show the participants individual results and the thick line shows the average over all participants. The dashed lines show the *x*-position of the measured indices of difficulty.

arm and 2 s for the non-dominant arm for both touching and crossing the black area with the pen.

## DISCUSSION

In the present study, we assessed if the ST is an objective test that can be used in classification of arm coordination impairment in WR in which MVE can be distinguished from IM. To make this distinction, the ST should have sufficient test-retest reliability and Fitts's law should apply to a minimum of three spirals with different IDs. We found sufficient test-retest reliability, using a Bland-Altman analysis with a cut-off in the limits of agreement of 2 SD from the group mean between the two individual attempts for each spiral width. Only the difficult spiral test with the non-dominant arm did not meet this criterion, which was caused by the large variation in number of penalties (SD for combined 3 and 5 s. penalties more than 9) that resulted in a large variation in movement times between two attempts (SD = 22,5 s. for the test and 21.2 s for the retest). Furthermore, Fitts's law could be applied if three spirals with different IDs were used. Optimisation of the penalty times that were added to the movement time in case the borderlines of the spiral were hit or crossed, resulted in a better fit to Fitts's law.

Although the test-retest reliability was sufficient, there was a tendency for a larger variation between the two attempts if the ID was higher, except for the small spiral width (highest ID) for the non-dominant arm. This was reflected in more variation (i.e., larger limits of agreement) between the two attempts in the small spiral width (higher ID), than in the spiral with the large spiral width (lower ID). The exception for the ST with the highest ID in the non-dominant arm was caused by a combination of a slow movement and a high number of penalties, which resulted in some trade-off in the total movement time between the test and the retest. In athletes with coordination

**TABLE 3** | *Post-hoc* analysis for penalty time optimization.

| Penalty time (s) | Test Rmcorr | Retest Rmcorr |
|---|---|---|
| **Dominant** | | |
| 0 | 0.83 | 0.85 |
| 0.5 | 0.92 | 0.92 |
| 1 | 0.96 | 0.95 |
| 1.5 | 0.97 | 0.96 |
| 2 | 0.98 | 0.96 |
| 2.5 | 0.98* | 0.96 |
| 3 | 0.98 | 0.96* |
| Original[†] | 0.97 | 0.96 |
| **Non-dominant** | | |
| 0 | 0.86 | 0.85 |
| 0.5 | 0.92 | 0.92 |
| 1 | 0.95 | 0.94 |
| 1.5 | 0.95 | 0.95 |
| 2 | 0.98* | 0.96* |
| 2.5 | 0.96 | 0.95 |
| 3 | 0.95 | 0.96 |
| Original[†] | 0.95 | 0.95 |

[†] *Original penalty time of 3 s for touching the line and 5 s for crossing the line.*
[*] *Highest Rmcorr between index of difficulty and movement time.*

impairment, we expect a longer movement time and more penalties, which could potentially result in a decrease of the test-retest reliability which is not acceptable. However, all spirals used had a very high ID (450.8–901.6) compared to previous research in arm coordination impairment using Fitts's law (3–5) in which tapping tests were used (17). So there is more than enough room to decrease the ID by decreasing the number of windings, which will increase the test-retest reliability, so it will also be acceptable in athletes with coordination impairment. Therefore, we advise to use spirals with fewer windings than the current seven turns to increase test-retest reliability. The optimum number of windings still needs to be determined in additional research.

In the spirals with the highest IDs, the longer movement time and the variation in movement time was not only caused by slower movements, but also by more often touching or crossing the black area resulting in more penalty time. This may be a sign that the penalty time used in the original research of the ST is too long (22). This was one of the reasons we performed a *post-hoc* optimisation of the penalty times. There was also a difference in variation between the two attempts in the dominant vs. the non-dominant arm, in which it took generally longer to complete the spiral with the non-dominant arm and there was more variation in movement time with the non-dominant arm. Again, the longer movement times were for a large part determined by the penalty times that were the same for the dominant and the non-dominant arm. In previous research, the final position accuracy of the movement in a reaching task was similar in the dominant and the non-dominant arm. However, the movement trajectory was different, with a longer trajectory for the non-dominant arm (29). In the ST in which the trajectory is restricted, this can result in a longer movement time and/or more penalty time for the non-dominant arm than for the dominant arm. This was a second reason to optimize the penalty times and to consider a difference in penalty time between the dominant and the non-dominant arm. The final reason for *post-hoc* optimisation of the penalty times was the difference in lay-out of the ST used in the present research, compared to the ST in the original research. In the present research, the difference in penalty time between touching and crossing the line of the spiral width seemed less relevant, because crossing the black area with the pen to end up in the next winding did not occur. Based on the *post-hoc* optimisation of the penalty times, we found the highest Rmcorr with a penalty time of 2.5 s for the dominant arm and 2.0 s for the non-dominant arm for any contact with the black surface. In addition to lowering the number of spiral widths, we advise to optimize the penalty time into one penalty time for any contact with the black surface, but separately for the arms, 2.5 s for the dominant arm and 2.0 s for the non-dominant arm.

The applicability of Fitts's law for the ST is promising to detect IM. But to be a valid method for differentiating between MVE and IM two main criteria must be met: (1) there must be significant differences between the results achieved under MVE and IM conditions; and (2) there must be acceptable sensitivity and specificity (17). The penalties for IM during classification are severe, ranging from banning from the competition where the IM occurred to a lifetime ban for all Paralympic sports. Besides, there are potential substantial ethical and legal consequences for labeling an athlete as a cheat (30). Therefore, maximum specificity for detecting IM is crucial, to avoid false accusations. However, sensitivity must be high enough, to discourage athletes to attempt IM (31). A threshold for deviation from the line of Fitts's law to label the test result as IM with close to perfect specificity and optimal sensitivity still needs to be determined.

## Strengths and Limitations

The strength of this study is that the participants were volunteers with the same age and gender as WR athletes. This match was chosen, because there is an impact of age and gender on coordination (22). Because of the match, the study results can be used as a reference/normal values for future research in athletes with coordination impairment. Another strength is that the ST was performed on a tablet instead of on paper like in previous research (14). It will be easier to make more than the minimum of three STs with different IDs to increase the precision of the application of Fitts's law, which can enhance the sensitivity and the specificity for IM.

More difficult versions of the ST with longer movement times, resulted in more variation of the MT. We anticipate that athletes with arm coordination impairment will need longer MT to complete even easier versions of the ST. In future research in the ST in athletes with arm coordination impairments the optimal ID (spiral length/number of windings and spiral width) need to be determined for maximum reliability and the applicability of Fitts's law. In addition, we would like to collect objective tracking data from the pen and tablet (i.e., movement time, x- and y-coordinates and pen pressure) for better accuracy of the movement time and to determine if these parameters could give additional insight in intentional misrepresentation.

The present study is only focussing on developing optimal tests for arm coordination impairment. If the spiral test is an optimal and parsimonious test for arm coordination impairment, assessment in athletes with an underlying health condition that leads to coordination impairment will be the next step. This research should include the assessment of the relationship between test outcomes and performance in standardized activities that determine proficiency in WR. Only after finalizing these additional steps, evidence-based classification can be achieved.

## CONCLUSIONS

The ST is a parsimonious test that provides an objective, reliable, compound measure for coordination impairment at all joint levels of the arm. Furthermore, it is a feasible test that requires minimum equipment (14). The current research provides supporting evidence that IM may also be detected successfully using the ST. These features are promising for future use in classification of arm coordination impairment in Paralympic sports such as WR.

## DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

## ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Medical Ethical Committee of the Netherlands, region Arnhem and Nijmegen. The patients/participants provided their written informed consent to participate in this study.

## AUTHOR CONTRIBUTIONS

VA, NH, EL, and MJ formulated the research question and they established the study design and discussed the study results and contributed to the manuscript. NH, MJ, and VA performed the measurements and the data analysis. All authors contributed to the article and approved the submitted version.

## SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fresc.2022.865133/full#supplementary-material

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# Prediction of Propulsion Kinematics and Performance in Wheelchair Rugby

*David S. Haydon[1,2]*, Ross A. Pinder[2,3], Paul N. Grimshaw[2,4], William S. P. Robertson[2] and Connor J. M. Holdback[2,3]*

[1] South Australian Sports Institute, Kidman Park, SA, Australia, [2] Faculty of Sciences, Engineering, and Technology, University of Adelaide, Adelaide, SA, Australia, [3] Paralympic Innovation, Paralympics Australia, Adelaide, SA, Australia, [4] College of Health and Life Sciences, Hamad Bin Khalifa University, Doha, Qatar

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***Correspondence:**
*David S. Haydon*
*david.haydon@sa.gov.au*

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Prediction of propulsion kinematics and performance in wheelchair sports has the potential to improve capabilities of individual wheelchair prescription while minimizing testing requirements. While propulsion predictions have been developed for daily propulsion, these have not been extended for maximal effort in wheelchair sports. A two step-approach to predicting the effects of changing set-up in wheelchair rugby was developed, consisting of: (One) predicting propulsion kinematics during a 5 m sprint by adapting an existing linkage model; and (Two) applying partial least-squares regression to wheelchair set-up, propulsion kinematics, and performance. Eight elite wheelchair rugby players completed 5 m sprints in nine wheelchair set-ups while varying seat height, seat depth, seat angle, and tire pressure. Propulsion kinematics (contact and release angles) and performance (sprint time) were measured during each sprint and used for training and assessment for both models. Results were assessed through comparison of predicted and experimental propulsion kinematics (degree differences) for Step One and performance times (seconds differences) for Step Two. Kinematic measures, in particular contact angles, were identified with mean prediction errors less than 5 degrees for 43 of 48 predictions. Performance predictions were found to reflect on-court trends for some players, while others showed weaker prediction accuracy. More detailed modeling approaches that can account for individual athlete activity limitations would likely result in improved accuracy in propulsion and performance predictions across a range of wheelchair sports. Although this would come at an increased cost, developments would provide opportunities for more suitable set-ups earlier in an athlete's career, increasing performance and reducing injury risk.

**Keywords: paralympic sport, wheelchair propulsion, wheelchair configuration, regression, modeling**

## 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 (Mason et al., 2013), optimizing parameters in isolation (Vanlandewijck et al., 2011; Mason et al., 2012, 2015), or requiring substantial amounts of testing (Usma-Alvarez et al., 2014; Haydon et al., 2019). These issues stem from difficulties in: monitoring on-court performance, where inertial measurement units (IMUs) only recently provide a reliable solution (Pansiot et al., 2011; van der Slikke et al., 2015, 2016; Shepherd et al., 2016); the substantial cost associated with wheelchair purchase (often $5,000–$10,000 USD); adjusting wheelchair set-ups on current wheelchairs; and optimization that varies for individual players, where a greater focus on individual impairments could potentially improve the ability to quickly achieve near optimal set-ups.

In wheelchair rugby (WCR), players are assigned point classification scores ranging from 0.5 to 3.5 points depending on their sport specific activity limitation (the ability to perform key tasks within the sport with regards to their impairment) where a lower score indicates greater limitation (International Paralympic Committee, 2020). The classification process considers trunk, arm, and hand function [where "function" includes strength, range of motion and co-ordination (Haydon et al., 2018a)] and hence players with varying impairment types [i.e., impaired muscle power—potentially due to spinal cord injuries (SCI)—or limb deficiencies which can be congenital or due to amputation] can be assigned the same classification scores. Optimizing wheelchair set-up based on either classification or impairment type is therefore not viable (Haydon et al., 2019; International Wheelchair Rugby Federation, 2021). Hence methods are needed that can provide detailed quantitative (and individualized) insights into the effects of set-up parameters on performance factors, while also minimizing the amount of time and effort of on-court testing.

Ideally, on-court testing would be used for optimizing wheelchair configurations, where athletes can be tested under conditions that are representative of competition demands as far as practically possible (Goosey-Tolfrey and Leicht, 2013). This testing can reveal significant differences in performance for set-up parameters such as wheel camber angle (Mason et al., 2012), seat angle and depth (Haydon et al., 2019), and even glove type (Mason et al., 2009). Small changes to some parameters can have substantial impacts on performance and on-court results, with the difference between executing or missing blocks on opposition dependent on just centimeters of position (Haydon et al., 2018a). However, despite the use of improved sensor technology and algorithms (combined with high-speed video) to identify key features of performance (Haydon et al., 2018a), on-court assessments remain difficult. This is due to the number of possible set-up parameters and combinations, with each of these having various effects on acceleration, agility, and ball-handling (Mason et al., 2010, 2013). Achieving a balance across the range of set-up parameters (seat height, seat angle, etc.) and performance measures (acceleration, agility, etc.) becomes even more difficult when considering the trade-off for various parameters on performance, as well as the interaction between various parameters (Mason et al., 2010, 2013). To address this problem, a substantial time commitment is required from athletes and coaches for both testing and results interpretation (van der Slikke et al., 2016; Haydon et al., 2019) which also has limitations due to skill adaptation and preferences of athletes based on their previous experiences (Haydon et al., 2018b, 2019). Further developments are therefore desired in maximizing efficiency in optimizing wheelchair set-ups at an individual level; propulsion modeling provides a potential method to achieve this.

Most current wheelchair propulsion modeling approaches have focused on musculoskeletal models attempting to quantify shoulder loads in daily propulsion to assess or reduce the likelihood of shoulder injuries (Morrow et al., 2010; Rankin et al., 2012; Slowik and Neptune, 2013; Hybois et al., 2018; Lewis et al., 2018). 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, creating valid individual representations of anthropometrics and muscular function is also an extensive process (Dembia et al., 2020; McErlain-Naylor et al., 2021). 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 [the vertical distance from the rear of the seat to main (rear) wheel axle] and seat depth (often referred to as fore-aft position, the horizontal distance from rear of the seat to main (rear) wheel axle) during daily propulsion (Richter, 2001; Leary et al., 2012). It should be noted that these terms are used as they were clearly understood by the coach and participants involved, as well as aligning with previous literature, but in some cases do not conform to ISO standards (Waugh and Crane, 2013)—care should be taken to interpret these measures correctly. This method appears to be a more realistic solution for optimizing wheelchair set-up for performance due to the reduced time requirements and ease of adjusting for individual players. However, assessing the relationship between kinematics and on-court performance measure is difficult, particularly when this relationship with performance varies across players (Fletcher et al., 2021).

The development of regression approaches, such as partial least squares (PLS), provide a potential method for quantifying the relationship between wheelchair set-up, propulsion kinematics, and performance. These regression approaches consider several 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. Regression approaches attempt to find a relationship between the predictor variables and the predicted variable by minimizing the error across all conditions (Schumann et al., 2013). The PLS approach does this by linking the variability of predictors with the response through a simultaneous decomposition of all variables (Schumann et al., 2013). Such approaches have been used across a range of areas, including the design of running shoes and emotional reaction of consumers (Shieh and Yeh, 2013), pelvic shape prediction (Schumann et al., 2013), determination of sport rock climbing performance (Mermier et al., 2000), and technique analysis in sports (Federolf et al., 2014; Gløersen et al., 2018).

The aim of the current study was to investigate the ability of a PLS approach to predict sprint performance based on individual wheelchair set-up and predicted propulsion approaches. A subsequent aim was to assess the prediction accuracies of propulsion kinematics of a linkage model in comparison to measured propulsion kinematics. To achieve these aims, a linkage model was implemented to predict alterations in propulsion kinematics with changing wheelchair set-up for elite WCR players, and then use a PLS approach to predict the effect of these alterations on sprint performance. This work is intended as an exploration to determine if there is scope to expand research in this area rather than a validation of this approach. Using this simplified model (in comparison with musculoskeletal modeling), it was expected that prediction of propulsion kinematics with changing wheelchair set-up would be successful, and subsequently be able to infer performance measures (sprint time). The ability to predict performance measures is expected to link closely with the ability to predict propulsion kinematics.

# METHOD

## Participants

Eight elite WCR players were recruited and provided informed, written consent before completing testing. All players were members of the Australian WCR team, were classified by the International Wheelchair Rugby Federation (IWRF) and completed testing in an adjustable wheelchair using 25-inch wheels. Individual player details are summarized in **Table 1**.

## Testing

Testing consisted of an orthogonal design approach using an adjustable wheelchair. Orthogonal design is a robust design approach that reduces the time and cost associated with optimizing parameters in real-world applications (Mori and Tsai, 2011). Using an orthogonal array reduces the number of tests required by systematically varying the combinations of parameters and levels while maintaining the ability to identify the effects of specific parameter levels. After testing has been completed, level averages from each parameter (e.g., reduced seat height) are compared against the grand average to determine the effect of each parameter level (Mori and Tsai, 2011). This approach allowed for the variation of four set-up parameters (seat height, seat depth, seat angle, and tire pressure) at three levels (player's current level, an increase, and a decrease) using an L9 orthogonal array (9 total set-ups). Seat height and seat depth used the definitions described above, while seat angle as the sagittal angle of the seat above the horizontal—these are shown in **Figure 1**. Seat height and seat depth were adjusted by ±15 mm, seat angle by ±5 degrees, and tire pressure by ±15 psi. An example of the orthogonal design is provided in **Supplementary Material**. Players completed a warm-up and familiarization process in each set-up before completing two sprints while monitoring performance measures and propulsion kinematics. The 5 m sprint which was conducted from standstill in the athlete's own time with sprint time recorded using laser timing gates (SpeedLight, Swift Performance). All testing (including the athlete's current set-up) was performed in an adjustable wheelchair (mass of 14 kg), with the athlete using their own wheels and gloves, and strapping was consistent across trials. The set-ups were tested in a randomized order, including a set-up that replicated the players typical set-up. For more details on testing implementation and analysis, see Haydon et al. (2016).

Propulsion kinematics (contact and release angles) and performance time for the 5 m sprints, along with the set-up information, were monitored for the first three strokes due to their importance on WCR performance (West et al., 2014). Angles projected in the sagittal plane were calculated from digital footage (120 Hz, Go Pro Hero 3+, California, U.S.) that was analyzed as part of a custom Matlab (Mathworks, 2017b) script. The points of contact and release were identified by acceleration spikes from inertial measurement units (IMUs) located on each wheel (recording at 500 Hz, IMeasureU, NZ). The IMUs were secured to the outside of the disc wheel using tape in a location that avoided any interference during the stroke, with this resulting in the distance from the axle varying for each player. When these acceleration spikes were selected, the corresponding GoPro video frame (and ±2 frames either side) were prompted, with the user then visually confirming the contact or release moment. The propulsion kinematics were then measured in the sagittal plane view by selecting: (i) the center of the wheel as a reference point, (ii) a point directly superior to this in the digital video frame to attain the vertical direction, and (iii) the position of the hand on the pushrim/wheel. Hand landmarks differed between athletes due to limb impairments and variations in propulsion technique. Additionally, as seat angle has previously been linked to trunk motion (Vanlandewijck et al., 2011), trunk angles at contact and release for each of the first three strokes were investigated for the various seat angle levels. Trunk angle was determined using a similar method by selecting: (a) an approximate hip position (identification of hip position varied across players due to wheelchair design resulting in occlusion; landmarks specific to each player were used) as a local reference point, (b) a point directly superior to this in the digital video frame, and (c) the acromion to determine the trunk angle. Refer to **Figure 1** for model representation of the hip and acromion positions. Note, the flexed trunk position was defined as a positive trunk angle. These results were then used as the input for each player's trunk angle in the linkage model, depending on the seat angle level. The intra- and inter-evaluator reliability of kinematic analysis was assessed across 20 trials by the lead researcher 2 weeks after initial analysis, and by an additional researcher, with good-to-moderate results of 2.6–9.7% technical error of measurement across all variables (Duthie et al., 2003).

## Modeling

Performance predictions for various wheelchair set-ups from on-court testing results occurs in two main steps: (One) predicting propulsion changes when altering wheelchair set-up, and (Two) predicting performance for inputs of wheelchair set-up and propulsion kinematics. Step Two relies on propulsion prediction inputs from Step One 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 2** and is detailed in the following sections.

**TABLE 1 |** Player information, including impairment, classification, and experience information. Contact Prediction Method refers to whether these players required an alteration to the equations for calculating their kinematics (see Section Propulsion Prediction).

| 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 |

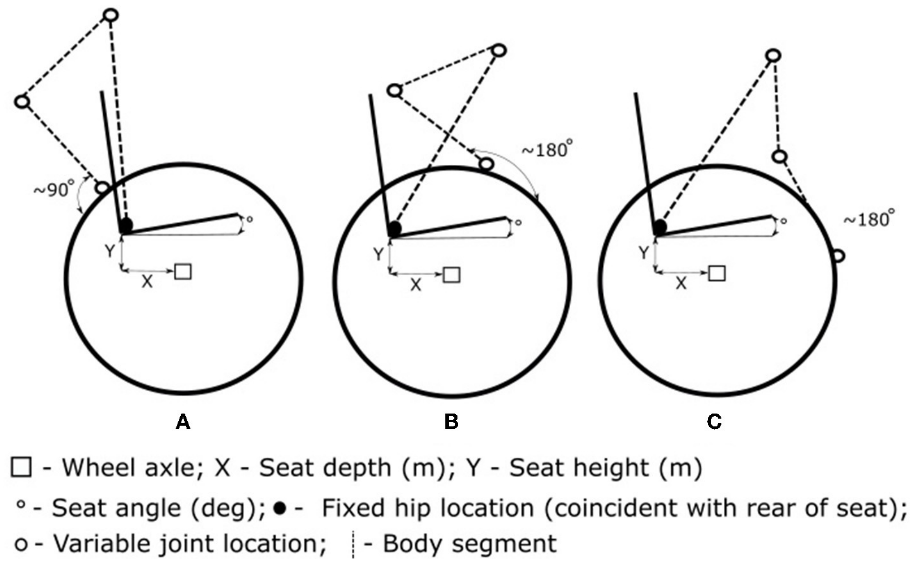


**FIGURE 1 |** The propulsion model consisted of a trunk, upper arm, and forearm segments with a fixed hip position and variable seat height [the vertical distance from the rear of the seat to main (rear) wheel axle], seat depth [often referred to as fore-aft position, the horizontal distance from rear of the seat to main (rear) wheel axle], and seat angle (angle of the seat above the horizontal). Contact angle estimation varied between the previous assumption of the forearm being perpendicular to the wheel tangent at contact **(A)**, and an altered propulsion method where the forearm is close to parallel with the wheel tangent **(B)** at contact (Leary et al., 2012) for the added assumption. Release angle **(C)** is also presented for comparison with the contact positions, with assumption that release occurs when the forearm is parallel to the wheel tangent when the trunk is in its most flexed position. The propulsion kinematic angles (contact and release) are measured with respect to the location about top dead center of the wheel. The hip position visually presented here does not intend to represent the actual hip position for athletes in wheelchair rugby, with the model assuming that hip location is coincident with the rear corner of the seat*.

## Propulsion Prediction

A sub-maximal linkage model (Richter, 2001; Leary et al., 2012) was adapted that calculated hand contact and release angles [relative to top dead center (TDC) location, with in-front of TDC positive and behind TDC negative] based on individual anthropometrics and chair set-up. In advancing previous models to predict maximal effort propulsion (Vanlandewijck et al., 2011), the model included an additional trunk segment with trunk angular rotation (flexion) occurring about the hip position, which was assumed to be coincident with the rear of the seat and subsequently changed with seat depth and seat height. The equations for contact and release were derived to use shoulder position based on the trunk angle at hand contact and release rather than a fixed shoulder position. Trunk angular velocity (i.e., rate of progression from trunk angle at contact to trunk angle at release) was assumed to be constant throughout the stroke phase. The assumption of contact occurring when the forearm is perpendicular to the tangent of the wheel (Leary et al., 2012) was not valid for some players due to various propulsion techniques as seen in **Figure 1**. Players with greater trunk range of motion (in this participant group, some players with limb impairments) generally utilized an approach with a greater proportion of "push" [see Haydon et al., 2018a, where "push" is the phase of the stroke that occurs during elbow extension (Vanlandewijck et al., 2001)]. 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 1** as Altered), a 90-degree addition was included for the prediction of the contact

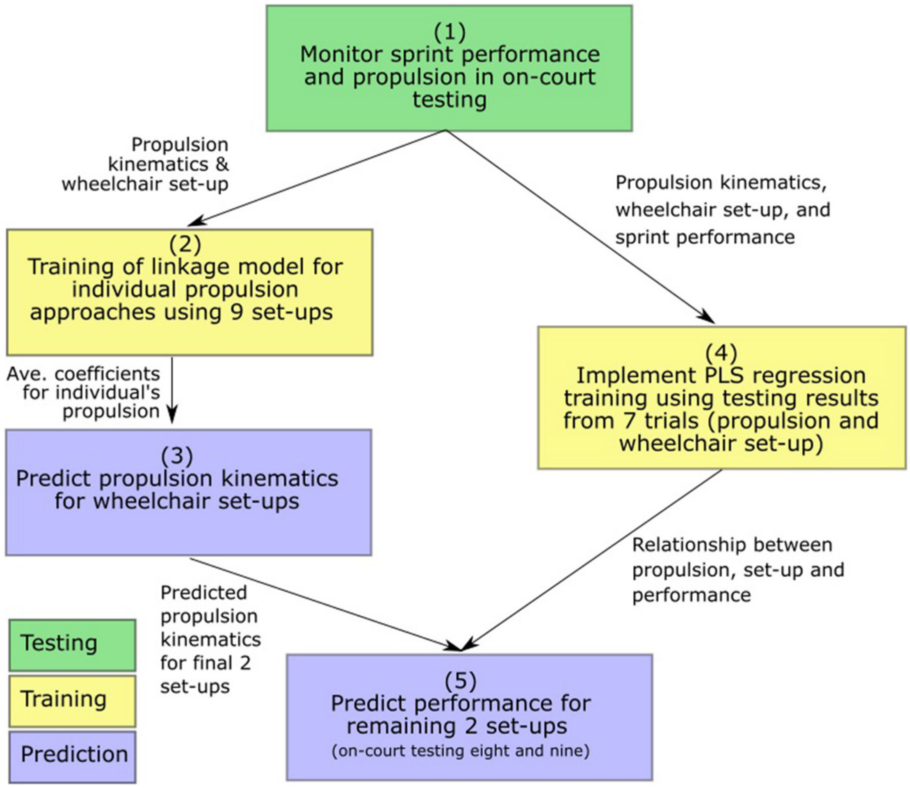


**FIGURE 2 |** Outline of the procedure from on-court testing to performance prediction.

angle (Equation 1).

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

Where $\beta$ is a contact coefficient varied from $-0.5$ to 1.5 [a coefficient of 1 means the assumption of hand contact (perpendicular/parallel) is true; discussed in more detail below]; $\theta_c$ is the hand 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 $\theta_{TI}$ is the initial trunk angle. Anthropometric measures were completed with the support of a physiotherapist familiar with the athletes and adapted to suit the needs of each individual as per their impairment. This enabled the prediction of contact and release angles based on an individual player'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 and hence the trunk angles were linked with corresponding seat angle measures from testing.

The contact coefficient accounts for variation from the assumption that contact occurs when the forearm segment is perpendicular (or parallel for some players), with the coefficient being 1 when the assumption is true. This allows individual propulsion approaches to be accounted for within the overarching assumptions. During analysis of the nine set-ups tested, a contact coefficient was determined (to two decimal places) for an individual for each of the first three strokes that minimized the error between measured and predicted angles from the above equation. 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 was used to determine release angle coefficient for each of the three strokes using the prediction equation from previous work (Leary et al., 2012), with release angle defined as when the forearm is parallel to the wheel tangent and the trunk at most flexed position; this differs to the altered contact angle as the shoulder is now at the most forward position (due to trunk flexion). 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 (Haydon et al., 2018a). Despite the potential asymmetry present in WCR propulsion (Goosey-Tolfrey et al., 2018), this process combined left and right propulsion kinematics to reduce the impact of any outliers in coefficient calculations. The use of a single coefficient for each stroke also assumes that a player would not substantially alter their propulsion technique across wheelchair set-ups.

## Performance Prediction

The experimental data was analyzed using a Partial Least Squares (PLS) regression. These included thirteen input variables: seat height, seat depth, seat angle, tire pressure, contact angles for the first three strokes, releases angle for the first three strokes, and the push angles for the first three strokes. The predicted variable was the sprint time. These regression approaches were trained independently in Matlab (using the *plsregress* function, Mathworks, 2017b), with the first seven of the nine set-ups from experimental testing used to train the prediction methods (within typical training-test ratios of 70–30% and 80–20%). The number of PLS components typically used in the function was set at five based on assessments of explained variance, with selection made once explained variance appeared to plateau. The number of components was adapted for each athlete depending on these results (an exemplar plot of PLS components and explained variance for Player 4 is provided in **Supplementary Material**, as well as the number of components and explained variances for all athletes). The performance of the prediction method was then assessed using the final two set-ups from experimental testing for each athlete. While 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 rather than measured kinematics to ensure a true prediction from set-up to performance. The method of progression from on-court testing to performance prediction is outlined in **Figure 2**.

## Statistics

Mean (and standard deviations) were calculated for the difference between experimental and modeling kinematic results for each player ($n = 8$) and each stroke ($n = 3$), resulting in 24 strokes for contact and release. The ability to predict propulsion kinematics and performance was typically investigated at an individual level, with results focusing on obvious trends within these. To support this, Welch's $t$-test (for unequal variance) using an alpha of 0.05 before a Bonferroni correction (alpha adjusted to 0.008 due to six comparisons—contact angles compared with other contact angles, release angles compared with other release angles) were completed across contact and release angle differences for each stroke.

For modeling assessment, no statistical analysis was completed due to the small sample size (only two comparisons for each player) and interest in how the modeling performed at an individual level. Assessments were made from the magnitude and direction of difference between experimental and modeling performance measures.

# RESULTS

## Propulsion Prediction

For each player, kinematic data was recorded for the first three strokes with two successful trials per player (eight participants, hence 24 mean stroke results). The kinematics for each stroke were calculated and summary statistics determined for the differences between measured and predicted contact and release angles (**Figure 3**). Mean values suggest contact angles could be predicted with differences less than 0.5° for 18 of 24 (75%) contacts. However, the maximum differences between a measured and predicted contact angle varied by greater than 10° for 9 of 24 (37.5%) of these contacts. There were no significant differences in contact angle prediction between the three strokes. Furthermore, mean release angle prediction differences increase during later strokes after the sprint start, with significantly less error between experimental and modeling release angle prediction for stroke one compared with strokes two (release angle one: $0.05 \pm 5.29°$; release angle two: $-3.07 \pm 4.80°$; $p < 0.001$) and three (release angle three: $-4.55 \pm 5.46°$; $p < 0.001$). Maximum differences were also greater for the release angles compared with contact angles for the majority of players. There were no obvious trends for contact predictions when considering the altered contact equation (Players 1, 3, and 5 as noted in **Table 1**) compared with the contact prediction from previous linkage models. Specific experimental and modeling propulsion kinematic results are provided in **Supplementary Material**.

## Performance Prediction

Sprint time predictions were calculated using chair set-up parameters and predicted propulsion angles as inputs to the PLS regression approach. Comparisons with actual (measured) sprint time for the two set-ups that were not included during training of the regression model are presented in **Figure 4**. Mean (±SD) sprint performance prediction error for both set-ups across all players was 0.04 (±0.25) seconds; with a minimum difference of 0.01 s (Player 4) and a maximum of 0.87 s (Player 7, set-up 2). All players, excluding Players 5 and 7, had average prediction errors of less than 0.1 s.

# DISCUSSION

Modeling of wheelchair propulsion has the potential to minimize 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 regression. On-court testing captured propulsion kinematics and performance across nine set-ups using an adjustable wheelchair. Propulsion prediction equations were developed using all nine on-court testing set-ups to allow for contact and release angles for the first three strokes to be predicted for any wheelchair set-up for an individual player. A PLS regression approach was trained using seven on-court testing set-ups, leaving two for assessment of the performance prediction method. For these final two set-ups, the propulsion prediction equations were applied to provide "predicted" rather than on-court kinematics, with the PLS regression model then producing a performance time that could be compared with on-court results.

Mean values for contact angle predictions were typically similar to on-court testing results (75% within 0.5°); however, maximum differences for each player can vary substantially with the mean results impacted by the casual summation of positive and negative errors. These large differences likely occur due to the assumption that a player will attempt to employ the same propulsion technique regardless of their wheelchair set-up—evident by using an average coefficient from all nine set-ups. This assumption was a limitation of this work as players can be expected to adapt their propulsion to the specific set-up; however,

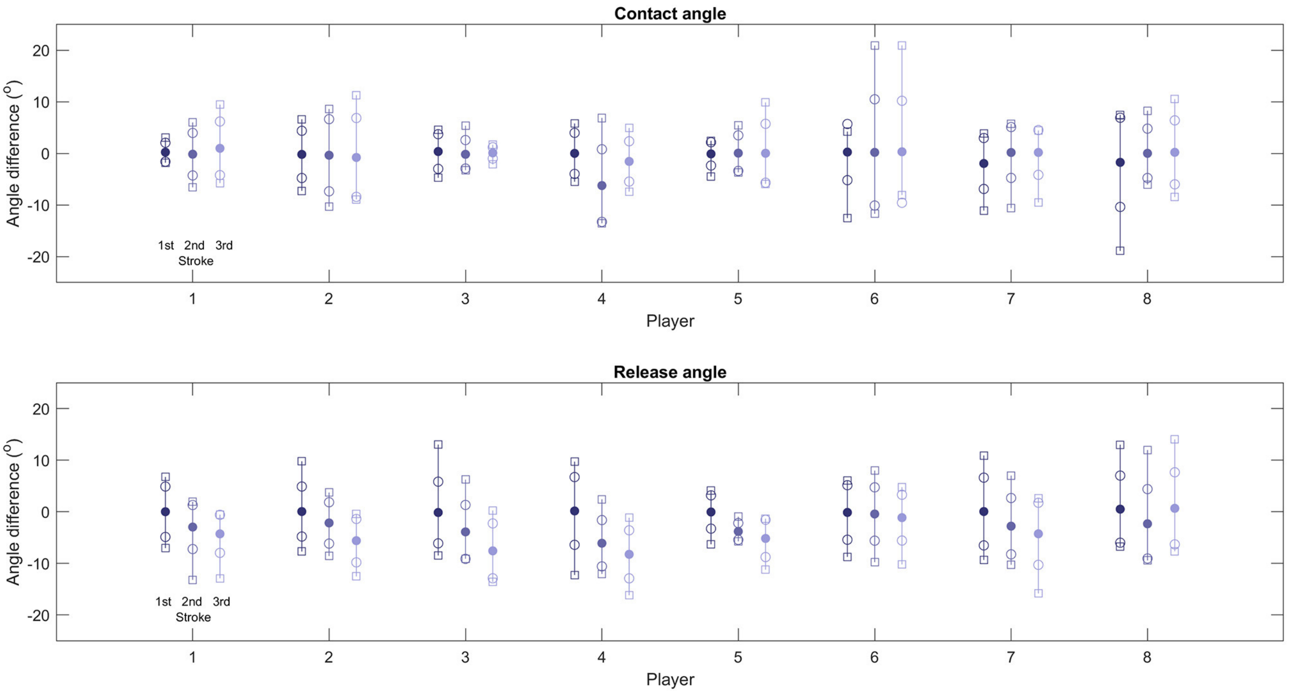


**FIGURE 3 |** Contact and release angle prediction differences from testing results. The first three strokes for each player are presented on individual bars, with each bar containing the mean difference (filled circle), the standard deviation (open circle), and minimum and maximum differences from testing results (open squares).

it was unclear how players would adapt their technique, hence the use of the average coefficient. Changes in propulsion approach between set-ups were therefore not accounted for which likely resulted in the large differences. Mean prediction error for release angle increases after the first stroke following a sprint start for most of the players (release angle one: $0.05 \pm 5.29°$; release angle two: $-3.07 \pm 4.80°$; release angle three: $-4.55 \pm 5.46°$; $p < 0.01$ when comparing release angle one with release angle two and three). For stroke one, mean prediction error is less than 0.51° for all players and less than 0.16° for 7 of 8 players. However, for the third stroke, only 2 players had an absolute mean prediction error less than 4.27°, with a maximum error 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 $+105°$; Haydon et al., 2018a). 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. For example, Player 8 had the smallest error for release angle estimation for the third stroke, and this player displayed the smallest release angles. This is potentially due to the variations in the coefficient value having less of an effect on the magnitude of the error—although this case does not confirm the hypothesis across the wider group. Both experimental and modeled kinematics were considered in the sagittal plane only, which is a simplification of the real-world behavior.

Regression prediction results varied between players; predicted results matched on-court testing results for some players (Vanlandewijck et al., 2011; Mason et al., 2013; Haydon et al., 2019) but were inconsistent for others (Players 2, 3, 5, 7). Player 4's results display the most potential for continued use of this approach. Despite large differences in on-court performance time in set-ups eight and nine, these changes in performance are predicted within 0.01 s by the model. 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. Player 4 regression was able to explain a high percentage of the variance, hence the ability of the model to predict performance. However, this is one case out of eight from testing; this alone does not support continued use of this approach. Although predicted performance times for Players 1 and 6 do not 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 the regression model was poor in predicting changes in performance despite supposedly showing a high percentage of explained variance based on training for the first seven set-ups. The PLS regression approach predicted improved performance for Player 2's set-up nine, but decreased performance was evident in on-court testing.

Similarly, Player 3 had similar performances in set-ups eight and nine, but regression predictions expected performance to vary by 0.13 s. Player 5 prediction did not align with performance times (average prediction error across two set-ups of 0.24 s), with performance underestimated substantially—although a slight change in prediction and on-court performance is evident. As above, this is likely due to changing relationships between wheelchair set-up and propulsion, which is emphasized for this athlete due to their lack of experience in comparison with other players. Player 7 predicted results showed minimal relationship with on-court results. Both predictions substantially overestimated the performance time, with set-up nine out by 0.87 s. For the on-court performance times for Player 7 (∼2.3 s), this amount of error is clearly unacceptable. 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, such as individual activity limitation, however this is often difficult to achieve in practice. While propulsion prediction shows potential for some athletes, developing regression relationships that translate to on-court performance is difficult due to changing propulsion techniques. Increasing data capture would allow for stronger relationships to be determined, improving this capability, however this would require substantial time and effort. An activity that allows for simpler data capture and has clearer translation to performance measures may provide a valuable tool to further investigate the capabilities of this approach.

This wheelchair performance assessment 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 (Richter, 2001; Leary et al., 2012). Assessment of maximal effort propulsion from standstill in WCR requires consideration of trunk motion—due to trunk motion accompanying force generation (Vanlandewijck et al., 2011)—and player specific approaches due to the substantial variations in activity limitation across classifications (Haydon et al., 2018a). The PLS regression approach 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. For this to be effective, both propulsion kinematics and performance times should be considered and be able to consistently identify small, meaningful changes. After completing on-court testing, this modeling approach can be implemented to identify set-ups of further 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 prior to making chair modifications which can be expensive in both cost and time commitment (Haydon et al., 2019). This improves upon current implemented approaches, where small adjustments to wheelchair

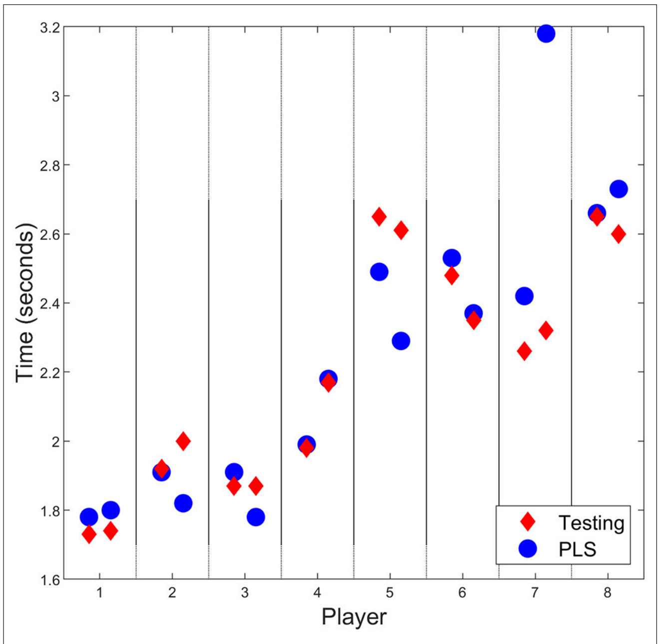


**FIGURE 4 |** Comparison of sprint times from testing and the regression approach for all players. Most predictions follow the testing data closely, with largest differences seen in Players 5 and 7.

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 (Fletcher et al., 2021).

The linkage model used in this study was investigated as it simplifies the model of an individual (particularly with a focus on 2D kinematics, rather than more realistic 3D kinematics), resulting in a reduction in time for development and processing. The model presented in this study is an adaption to a previous model that has been successfully used to link measured kinematics with propulsion measures (Richter, 2001; Leary et al., 2012). This adaption has been added to account for atypical variations in technique exhibited by the athletes, particularly those with trunk function who are able to lean forward and "push" on the wheel/pushrim during maximal effort propulsion. While an important adaption to include for these athletes, the added features of the new model should be considered a minor addition to the original model. The results show that whilst this linkage model approach might be appropriate for some cases, it is unlikely to be suitable for all athletes; some are likely to require more detailed models that greater reflect their activity limitation (McErlain-Naylor et al., 2021). This may be the case for athletes with greater activity limitation (lower classification scores), who are less able to adapt technique to wheelchair set-up. Due to limitations in wheelchair design, lower classification scores were under-represented in this work as the majority used a smaller wheel size than was possible with the adjustable wheelchair (lower classifications often use 24-inch wheels (Goosey-Tolfrey et al., 2018) compared with 25-inch wheels for the adjustable wheelchair). 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 (Lewis et al., 2018), with the ability to develop and customize musculoskeletal models improving rapidly (Dembia et al., 2020). Individual customization of the musculoskeletal models would require further processing time and more detailed on-court testing assessment including motion capture and electromyography, which is more suited to elite level athletes initially. By constraining joint ranges of motion and adapting the level of muscle activation for an individual [both of which are possible through software such as OpenSim (SimTK, 2021)], more realistic propulsion approaches can be determined for a range of set-ups. This would likely result in greater accuracy when attempting to predict performance, particularly as the user can define specific cost functions for performance and optimize for these. However, musculoskeletal models currently find it difficult to independently scale individual body segments which would limit their applicability to amputees. The selection of modeling approach should therefore consider the ability to accurately measure and replicate individual capabilities (Lewis, 2018; McErlain-Naylor et al., 2021) as well as time restraints around any prescription approach.

An additional benefit of the modeling approaches outlined is they may allow for the reduction of experience related effects on performance. Athletes may have developed a propulsion technique that is either (i) not in fact optimal for maximal sprint performance, or (ii) is highly specific to maximizing their sprint performance in their current chair set-up (Haydon et al., 2018b). A small amount of on-court testing (i.e., a familiarization period in each testing set-up prior to data capture) is unlikely to promote adaptation to a new set-up quickly enough to get a true indication of likely performance once the athlete has adapted to the new set-up. Alternatively, changes to chair set-up may perturb the propulsion coordination and increase movement variability for a short period, again impacting the testing results (Fletcher et al., 2021). Modeling, when accounting for athlete activity limitation, could remove this concern and give a greater prediction of final performance outcomes should an alternative (i.e., predicted optimal) propulsion technique be considered. Specialists in motor control and learning (i.e., skill acquisition specialists) would then be best placed to support coaches and athletes with targeted technical (learning) interventions.

Currently, this approach requires 2- to 3-h of on-court testing with various set-ups for each individual in order to measure propulsion approaches and performance. With further progression of this method, there is the possibility to markedly reduce the amount of on-court testing required, particularly if musculoskeletal models can be developed. 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 and improve reliability, international collaborations are recommended to increase the pool of elite wheelchair sport athletes.

## CONCLUSION

The process of wheelchair prescription is currently a time-consuming process that relies heavily on player and coach experience. This study presents a method to predict propulsion kinematics based on changing wheelchair set-ups for maximal effort sprinting. To account for maximal propulsion, an equation to predict contact angle while accounting for trunk motion was developed, improving on previous methods. Regression approaches (such as PLS) can be trained using on-court testing results, and then applied with propulsion predictions to estimate sprinting performance for WCR. Results for propulsion prediction found that the assumption of a consistent propulsion approach by using an adapted linkage model may not be appropriate, particularly for release angles. Improved understanding of wheelchair prescription impact on propulsion kinematics will support further development of accurate predictions. This scoping project suggests that while the linkage model prediction of propulsion kinematics may be suitable for some athletes, others may require more detailed models (e.g., musculoskeletal) that more accurately reflect their function. Regression approaches were inconsistent in their ability to accurately predict performance changes. Player 4's performance was predicted almost exactly despite the large variations (relative to other players) present in sprint time across set-ups eight and nine, likely due to the consistent relationship between wheelchair set-up, propulsion kinematics, and performance. However, other results were unable to achieve the same accuracy, with the expected cause being the inconsistent propulsion predictions and regression relationships. Substantial further work is required in this area to improve the process of wheelchair prescription for performance, with a greater understanding of these relationships likely to have a substantial impact on wheelchair prescription and subsequent performance.

## DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

## ETHICS STATEMENT

The studies involving human participants were reviewed and approved by University of Adelaide Ethics Committee. The participants provided their written informed consent to participate in this study. Written informed consent was obtained from the individual(s) for the publication of any potentially identifiable images or data included in this article.

## AUTHOR CONTRIBUTIONS

DH, RP, PG, and WR contributed to study design and data collection. DH completed primary analysis and wrote the first draft of the manuscript, with RP, PG, WR, and CH all contributing to analysis, review and writing sections of the manuscript. All authors have reviewed the manuscript and approved its submission.

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## SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fspor.2022.856934/full#supplementary-material

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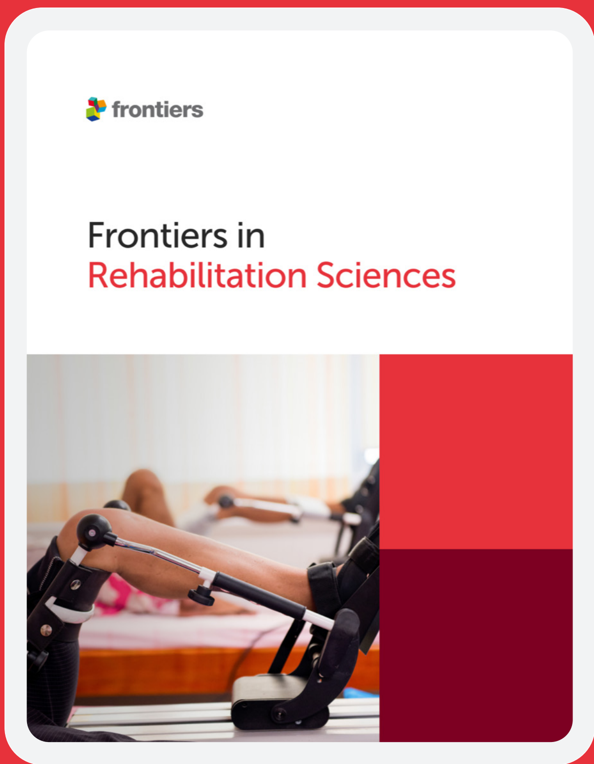


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