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ORIGINAL ARTICLE

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Force–velocity profiling of elite wheelchair rugby players by manipulating rolling resistance over multiple wheelchair sprints

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Abstract

This study investigated the effect of increased rolling resistance on wheelchair sprint performance and the concomitant force–velocity characteristics. Thirteen wheelchair rugby (WCR) athletes completed five 15 s wheelchair sprints in their own rugby wheelchair on an instrumented dual-roller wheelchair ergometer. The first sprint was performed against a close to overground resistance and in each of the following sprints, the resistance increased with 80% of that resistance. A repeated-measures ANOVA examined differences between sprints. Subsequently, linear regression analyses examined the individual force–velocity relations and then, individual parabolic power output curves were modeled. Increased rolling resistance led to significantly lower velocities (−36%), higher propulsion forces (+150%) and higher power outputs (+83%). These differences were accompanied by a lower push frequency, higher push time, yet a constant recovery time and contact angle. The modeled linear regressions ($R^2 = 0.71 \pm 0.10$) between force and velocity differed a lot in slope and intercept among individual athletes. The peak of the power output parabola (i.e., the optimal velocity) occurred on average at $3.1 \pm 0.6\,\mathrm{m\,s^{-1}}$. These individual force–velocity profiles can be used for training recommendations or technological changes to better exploit power generation capabilities of the WCR athletes' musculoskeletal system.

KEYWORDS

force–velocity, Paralympic sport, power output, propulsion technique, wheelchair Rugby

1 | INTRODUCTION

Wheelchair rugby (WCR) is an indoor team sport for athletes with a disability affecting both arms and legs.[1,2] WCR players are classified into one of seven classes ranging from 0.5 (most impaired) to 3.5 (least impaired). Accelerating faster than your opponent is essential to freely catch the ball, to score points or to successfully defend an opponent.[3] Athletes stop and start more than 200 times during a WCR match and more than 25% of their game activities is high-velocity pushing.[4,5] Wheelchair sprinting is thus an apparent key aspect of a WCR match and should be evaluated and understood accordingly. However, wheelchair sprinting is a challenging upper-body movement. Because of these high velocities, there is little time for the upper-body muscles to contract, to couple the hand with the rotating rim-wheel interface and to transfer the power to the wheelchair. It is currently unknown how much wheelchair sprint power output rugby athletes can produce maximally when hand speed would be lower and how we can optimize training methods and wheelchair design to be able to effectively use this power output.

Isolated active muscles have well-described force generating characteristics that are among others dependent on velocity; high contraction velocities will limit muscle-force production.[6] This inverse relation is derived from the intrinsic properties of the contractile element of the muscle.[6] However, forces exerted during wheelchair propulsion are based on a large number of muscles that exert forces within a complex system of bones, where the leverage of individual muscles vary over the course of the movement.[7] The mechanical properties of muscular systems acting within a multi-joint system are different from isolated muscles and several studies approached the multi-joint force–velocity relation in an approximately linear way.[7] More specifically, a significant linear force–velocity relation is seen in multi-articular cyclic movements such as bicycling,[8–11] arm-cranking[12,13] and handrim wheelchair propulsion.[14]

Power output is defined as the product of force and velocity, and in the case of a linear force–velocity relation, power output and velocity share a parabolic function with the apex of the parabola occurring at a given optimal velocity. To be able to operate around this optimal velocity, bicycles and handcycles have gearing systems, which help to reduce the contraction velocity when velocity of locomotion goes up.[15] In contrast, wheelchair rugby chairs have a fixed gear, with a fixed ratio of the wheel to the handrim diameter.[16] It is currently unknown whether the force–velocity relationship also exists in wheelchair sprinting in a diverse group of WCR athletes. Besides, when such relation is indeed present, it remains questionable at which optimal velocity the maximum power output is produced.

The force–velocity relationship can be explored by applying increased rolling resistances on a wheelchair ergometer during a series of short sprints. Previous research in non-wheelchair using novices already showed that 8 s and 20 s handrim wheelchair sprint tests against higher resistances led to lower wheelchair velocities, higher cycle and push times and concomitant higher mean power outputs, compared to a sprint at a lower resistance.[17–19] However, the able-bodied population studied in these previous studies differed from the WCR population in skill level, impairment and participant characteristics.[20] Furthermore, the wheelchair mechanics and wheelchair-user interface deviated from the typical rugby wheelchair.

Therefore, the purpose of the current study was to investigate the effect of increased rolling resistance on sprint power output, force production and velocity and to describe the underlying propulsion technique in elite WCR athletes. It was hypothesized that an increased resistance led to lower sprint velocities, subsequently higher generated forces and as a result higher sprint power outputs. An individual linear relation between force and velocity was expected which leads to a parabolic function between velocity and power with the apex of the parabola occurring at the optimal velocity.

2 | MATERIALS AND METHODS

2.1 | Participants

Thirteen experienced WCR players participated in this study (Table 1). Inclusion criteria were to train at an international level for more than 10 h/week and to play WCR for a minimum of 4 years. For this reason, athletes had been advised on the optimization of their WCR chair and had a reproducible acquired technique for wheelchair propulsion. Body mass was assessed with a seated balance scale (Seca 710). Since the included WCR group is already a heterogeneous group in terms of age, classification and disability and WCR is a mixed team sport, we decided to not exclude the female player. The study was approved by Loughborough University Research Ethics Committee, and all participants provided written informed consent

TABLE 1 Participant demographics.

Participant characteristics	*N* or *M* (SD)
Men/women	*12/1*
Age (years)	29 (7)
Body mass (kg)	61 (9)
Wheelchair mass (kg)	18 (1)
Experience in WCR	>4 years
Training hours	>10 h/week
Classification	
0.5	2 (2 SCI)
1.0	1 (SCI)
1.5	2 (1 SCI, 1 LA)
2.0	3 (2 SCI, 1 LA)
2.5	3 (2 SCI, 1 LA)
3.0	2 (1 CP, 1 AMP)

Abbreviations: AMP, Amputee; CP, Cerebal Palsy; LA, Les Autres; SCI, Spinal Cord Injury.

prior to participation. Data from the first sprint at a similar to overground resistance was previously analyzed by Goosey-Tolfrey et al.[21]

2.2 | Wheelchair ergometry

A friction-braked instrumented wheelchair ergometer (VP100H TE, HEF TecmachineR) was used and all participants were tested in their own individualized WCR chair (Figure 1).[22] Rear-wheel tire pressure was set to the player's self-selected pressure, wheel-size ranged from 24 to 25″ and rear-wheel camber from 16 to 20°. The wheelchair ergometer system consisted of two independent rollers that measure linear velocity, torque, and the angle of rotation at 100 Hz. The electromagnetic brake (type ZX) in each roller system had the capability to produce a braking torque of 0–4 Nm. The ergometer was calibrated prior to testing as described by Faupin et al.[23] The rear wheels of the wheelchair were strapped on the center of each roller and the front wheels were strapped down on the frame of the wheelchair ergometer.

To determine the resistance and ensure equal resistance on both sides, each wheelchair-athlete combination was subjected to a deceleration test on the ergometer. For this, we used the method of Theisen et al.[24] The players completed a sprint of four to five maximal pushes from standstill, then leaned forward with their hands on their knees and allowed the rear wheels to freely coast down until the wheels stopped (front castor wheels were not considered, allowing further standardization of the testing set-up). Players were instructed to lean forward as they normally would do in a sprint, dependent on their disability and the use of an abdominal binder. From these data, the individual rolling resistance was determined, proportional to the mass of the participant and chair, and ranged from 6.3 to 14 N (i.e., resistance coefficient = 0.008–0.017) among the athletes. Based on earlier studies on rolling resistance[25] and based on athlete-feedback, we conclude that this resistance was roughly like that of a wooden floor that is used during wheelchair rugby competition. All participants wore their usual gloves with adhesive strapping and some used an abdominal binder as they would have when taking part in a WCR game.

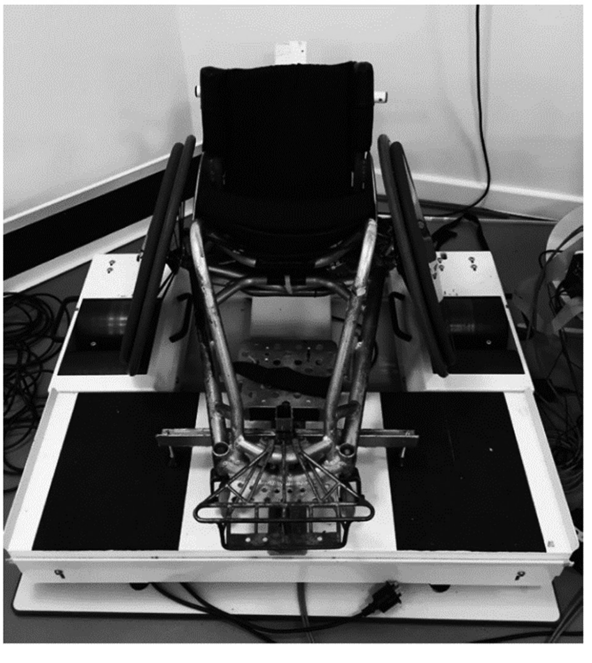


FIGURE 1 Experimental setup (Goosey-Tolfrey et al 2018).

2.3 | Test protocol

Following the deceleration test, athletes could propel their wheelchair for 5 min at a self-chosen velocity for familiarization with the ergometer. Following a rest period of 3 min, the test protocol started.

Five all-out 15 s sprints from stationary start on the wheelchair ergometer were performed, with 3 min rest in between. Duration of 15 s was chosen to ensure that at least 28 m was covered in the first sprint by all participants; 28 m represents the length of the playing court. The first sprint was performed on the initially determined resistance (=100%, specified as S100). For each of the following four sprints, resistance increased with 80% of the initial value (i.e., S100) which resulted in a total of five resistances, referred to as S100, S180, S260, S340, and S420. To illustrate, the rolling resistance of one athlete was determined at 10.5 N (=S100). The following sprints were performed at 18.9 N (S180, +80% of S100), at 27.3 N (S260, +160% of S100), and so on. Since rolling resistance varied between players, increments in percentages were preferred over absolute increments of resistances. These percentages were determined during a pilot experiment to range up to the maximal feasible resistance by the most impaired WCR athlete, that is, he was still able to accelerate the wheels and perform a sprint test. Verbal encouragement was provided at each sprint and no feedback about their sprint performance was given. No randomization between resistances was applied, since the primary aim of the initial overall experiment was different. The overall aim was to induce fatigue with the sprints on an increased resistance and compare two sprints at a resistance of 100%, one conducted before and one directly after the sprints with increased resistances. However, no significant differences between the pre and post-test sprints at the resistance of S100 were found, which meant that fatigue apparently had a limited influence among this group of well-trained WCR players (unpublished data, Table S1).

2.4 | Data analysis

All analyses were performed in custom-made Python routines (Work lab package, DOI 10.5281/zenodo.3268671), and consistent with previous literature.[21,26,27] Torque and velocity were recorded separately for the left and right wheels and data were filtered with a recursive second-order Butterworth filter (cut-off frequency 10 Hz).[28] The tangential force (F) at each side was calculated from the measured torque (M) and wheel radius (r_w) (equation 1). The power output (PO) at each side was calculated from the tangential force and wheel velocity (v_w) (equation 2).

$$F\ [N] = M\ [\mathrm{Nm}] * r_w^{-1}\ [\mathrm{m}] \quad (1)$$

$$\mathrm{PO}\ [\mathrm{W}] = F\ [\mathrm{N}] * v_w\ \left[\mathrm{ms}^{-1}\right] \quad (2)$$

Sprint performance was defined over the whole 15 s sprint. Push-related outcomes were defined per push and then averaged over all pushes in that sprint. PO, F, and work over the entire sprint and per push are the sum of the left and right arm, other outcomes are the average of left and right.

2.4.1 | Sprint performance

Distance [m] was defined as the entire distance covered in 15 s. Mean power output (PO_{mean} [W]), mean force (F_{mean} [N]) and mean velocity (v_{mean} [ms^{-1}]) were calculated as the average over the entire sprint. Maximal velocity (v_{max} [ms^{-1}]) was defined as the highest one sample velocity peak during the sprint. Total work [J] was defined as PO integrated over the wheel rotation angle.

2.4.2 | Push-related outcomes

One push was defined as each period of continuous positive power that the hand exerted on the handrim with a positive minimum of at least 1 W. Parameters were first determined from the unilateral PO signal and then summed or averaged between left and right. Then, parameters were averaged over all pushes. PO_{peak} [W], F_{peak} [N], and v_{peak} [ms^{-1}] were defined as the one sample peak in a push. Work per push [J] was the delivered work during a push. Push time [s] was the time of one push, recovery time [s] the time between the end of a push and the start of the next push. Frequency [Hz] was the number of pushes divided by the entire sprint time (i.e., 15 s). The angle of rotation [°] was calculated by integrating the angular velocity and the contact angle [°] was defined as the angle at the end of a push minus the angle at the start.

2.5 | Statistical analyses

The Shapiro–Wilk test showed that data were distributed normally. Subsequently, a repeated measures ANOVA was used to examine the differences in outcomes over the five sprints. A post-hoc test with Bonferroni correction was used when a main effect was observed. Significance for the repeated measures ANOVA was set at $p < 0.05$ and by use of the Bonferroni correction the significance for the post-hoc t-test between any two different sprints was $p < 0.01$.[29] The Mauchly's test of sphericity was used to investigate whether the assumption of sphericity had been violated. When this was the case, a Greenhouse–Geisser correction was used.[30] Effect size was calculated using partial eta-squared (η^2) and interpreted as small (≥0.01), medium (≥0.06) or large (≥0.14).[31]

To address the individual relation between force and velocity, linear regression procedures were performed for each participant between F_{peak} per push and v_{peak} per push over all pushes in each of the five sprints. To exclude the initial acceleration phase, the first push was removed from data analysis. To gain insights in the complete force–velocity spectrum, these linear regression lines were extrapolated and the parabolic power-velocity curve was calculated from this linear line (power = force × velocity). All data were analyzed using SPSS version 26 (SPSS Inc.).

3 | RESULTS

All participants successfully completed the sprint test protocol. Results from the five sprints and the repeated measures ANOVA are shown in Table 2. A typical individual example of sprinting on a low and high resistance for one athlete is visualized in Figure 2.

3.1 | Sprint performance

The covered distance for each sprint significantly decreased by an average of 35% with increasing resistance (from S100 to S420). PO_{mean} and F_{mean} significantly increased from S100 to S420 with an average of respectively 83% and 150% (Table 2). v_{mean} and v_{max} significantly decreased from S100 to S420 with an average of respectively 36% and 29% (Table 2). Total work significantly increased from S100 to S420 with an average of 86% (Table 2). All post-hoc tests for the sprint performance outcomes were significant and large effect sizes (≥0.14) were observed

TABLE 2 Mean (standard deviation) of the sprint performance and push-related outcomes during the five 15 s sprints and the relative difference between the first and fifth sprint (Δ100–420 (%)). Additionally, the *p*-value and the effect size of the repeated measures ANOVA are shown.

Sprint performance	**S100**	**S180**	**S260**	**S340**	**S420**	**Δ100-420 (%)**	**ANOVA results**	
							p-res	η^2
Resistance coefficient	0.014 (0.003)	0.025 (0.005)	0.036 (0.008)	0.047 (0.01)	0.059 (0.012)	+ 320 (0)	<0.01*	0.96
Distance [*m*]	61 (10)	54 (10)	48 (9)	44 (8)	40 (8)	- 35 (10)	<0.01*	0.85
PO_{mean} [W]	70 (20)	91 (26)	106 (31)	118 (35)	128 (38)	+ 83 (23)	<0.01*	0.88
F_{mean} [N]	19 (3)	27 (4)	34 (6)	41 (7)	49 (9)	+ 150 (26)	<0.01*	0.95
v_{mean} [ms^{-1}]	4.2 (0.7)	3.7 (0.6)	3.2 (0.6)	2.9 (0.6)	2.6 (0.5)	- 36 (8)	<0.01*	0.91
v_{max} [ms^{-1}]	4.7 (0.8)	4.2 (0.7)	3.8 (0.7)	3.6 (0.7)	3.3 (0.6)	- 29 (7)	<0.01*	0.89
Total work [J]	1037 (278)	1367 (410)	1614 (479)	1793 (533)	1934 (574)	+ 86 (24)	<0.01*	0.86
Push-related outcomes								
PO_{peak} per push [W]	377 (139)	456 (189)	484 (208)	496 (214)	489 (198)	+ 31 (23)	<0.01+	0.49
F_{peak} per push [N]	90 (25)	120 (36)	142 (42)	159 (48)	172 (49)	+ 93 (25)	<0.01+	0.85
v_{peak} per push [ms^{-1}]	4.4 (0.7)	3.9 (0.7)	3.6 (0.6)	3.3 (0.6)	3.0 (0.5)	- 30 (8)	<0.01*	0.89
Work per push [J]	29 (7)	40 (10)	49 (13)	56 (14)	64 (16)	+ 122 (36)	<0.01*	0.90
Push time [*s*]	0.12 (0.03)	0.14 (0.03)	0.15 (0.03)	0.17 (0.03)	0.19 (0.04)	+ 62 (24)	<0.01*	0.86
Recovery time [*s*]	0.30 (0.04)	0.31 (0.04)	0.31 (0.05)	0.31 (0.04)	0.32 (0.04)	+ 7 (12)	0.16	0.14
Push frequency [Hz]	2.4 (0.3)	2.3 (0.2)	2.2 (0.3)	2.1 (0.2)	2.0 (0.2)	- 17 (7)	<0.01+	0.74
Contact angle [°]	84 (18)	88 (15)	88 (15)	87 (14)	89 (14)	+ 8 (18)	0.42	0.07

Note: + Represents a significant main effect, but not for (all) post-hoc differences.

*Notes a significant post-hoc difference between each pair of the five sprints.

(Table 2). These results can also be seen in Figure 2 (left). Power output, force, and area under the power curve (work) were lower in S100 compared to S420. Vice versa, velocity was higher in S100 compared to S420.

Work per push significantly increased from S100 to S420 with an average of 122%, all post-hoc tests for the sprint pairs were significant (Table 2). Push time significantly increased from S100 to S420 with an average of 62% and post-hoc tests were significant for all sprint pairs (Table 2). Push frequency significantly decreased from S100 to S420 with an average of 17% (Table 2). Post-hoc testing showed that S100 differed from all other sprints and additionally, S180 differed from S340 and S420. In contrast, recovery time and contact angle did not change with increased resistances (Table 2). For all significant main effects, large effect sizes were observed. These results are illustrated in the two subsequent pushes in Figure 2 (right). The solid lines (S100) of the power output showed a lower area under the curve and were less wide compared to the dotted lines (S100). The time between two subsequent pushes remained constant.

3.2 | Push-related outcomes

PO_{peak} and F_{peak} per push increased significantly with an average of 31% and 93% from S100 to S240 respectively (Table 2). Post-hoc testing showed that the PO_{peak} per push differed between S100 and the other four sprints. F_{max} per push showed significant post-hoc effects for all sprint pairs, except for sprint pair S340 and S420. From S100 to S420, v_{peak} per push significantly increased with an average of 30% and all post-hoc tests among sprint pairs were significant (Table 2). For all three variables, large effect sizes were observed. These results can also be seen in the two subsequent pushes in Figure 2 (right). The solid lines (S100) of the power output and force were lower compared to the dotted lines (S420). Vice versa, the solid line (S100) of velocity was higher compared to the dotted line (S420).

3.3 | Linear regression results

The individual results of the linear regression procedures for all athletes are shown in Table 3 and Figure A1. The linear regression lines resulted in an average explained variance (R^2) of 0.71 (0.10) and both the intercept and coefficient of the linear regression were significant for all athletes ($p < 0.05$). On average, the apex of the power output parabola was at 534 ± 211 W, occurring at an optimal velocity of 3.1 ± 0.6 ms^{-1}.

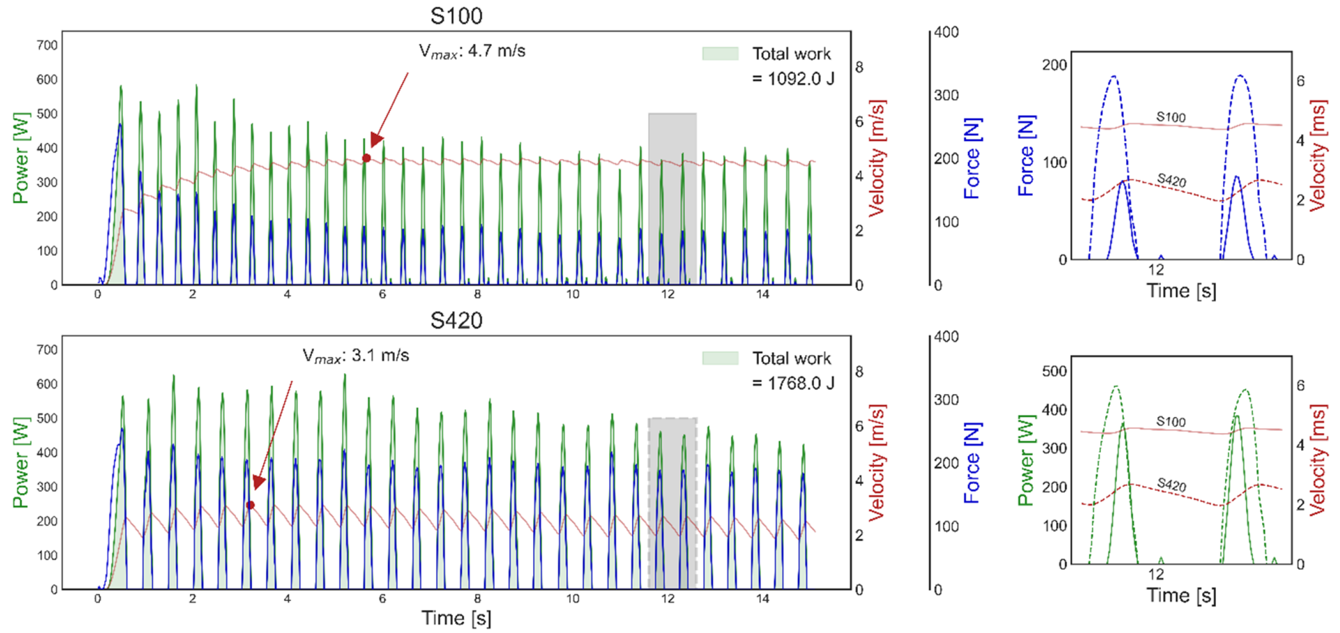
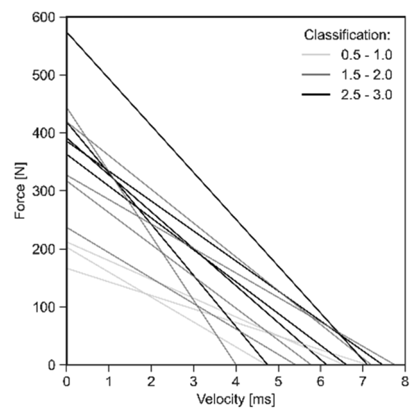


FIGURE 2 *Left*: Typical example of sprinting on a low (S100, upper graph) versus high (S420, lower graph) resistance (classification athlete = 2.5). Power output is shown in green, work as the area under the power curve, velocity in red and force in blue. *Right*: a direct push-by-push comparison is shown from each full sprint; two pushes around 12 s in the sprint are displayed in these two smaller plots: solid lines represent S100, dotted lines S420.

TABLE 3 Individual linear regression outcomes. Every row displays an individual athlete by classification. Optimal velocity (Opt v) is the calculated theoretical velocity and power at Opt v represents the peak of the power output parabola.

Classification	Regression[a]	R^2	Opt v [ms^{-1}]	Power at opt v [W]	Individual force–velocity curves
0.5 (Figure A1)	$y = 167 - 23x$	0.66	3.6	297	
0.5	$y = 203 - 43x$	0.88	2.3	240	
1	$y = 212 - 32x$	0.76	3.3	348	
1.5	$y = 237 - 44x$	0.72	2.7	321	
1.5	$y = 317 - 55x$	0.84	2.9	459	
2	$y = 419 - 58x$	0.53	3.6	754	
2 (Figure A1)	$y = 444 - 111x$	0.71	2.0	445	
2	$y = 327 - 42x$	0.61	3.9	635	
2.5 (Figure A1)	$y = 386 - 52x$	0.68	3.8	720	
2.5	$y = 419 - 88x$	0.56	2.4	498	
2.5	$y = 363 - 55x$	0.76	3.3	601	
3	$y = 391 - 64x$	0.68	3.1	601	
3	$y = 574 - 81x$	0.82	3.5	1020	
Mean (SD)		0.71 (0.10)	3.1 (0.6)	534 (211)	**FIGURE A1**: Individual force–velocity curves.

[a] $y = F_{max}$ per push, $x = v_{max}$ per push.

Three typical examples are visualized in Figure 3. For five athletes, the apex of the power output curve was within the measured values (similar to left athlete in Figure 3), for three athletes in the extrapolated values (similar to middle athlete in Figure 3) and for five athletes, the apex just touched the actually measured data (similar to right athlete in Figure 3). All individual figures can be found in Figure S1.

4 | DISCUSSION

The current study explored the effects of increased rolling resistance on sprint power output, force production, velocity, and underlying propulsion technique in elite WCR players. As expected, the v_{mean} over the entire sprint decreased, which coincided with an increase in F_{mean}. Consequently, the PO_{mean} increased with a higher

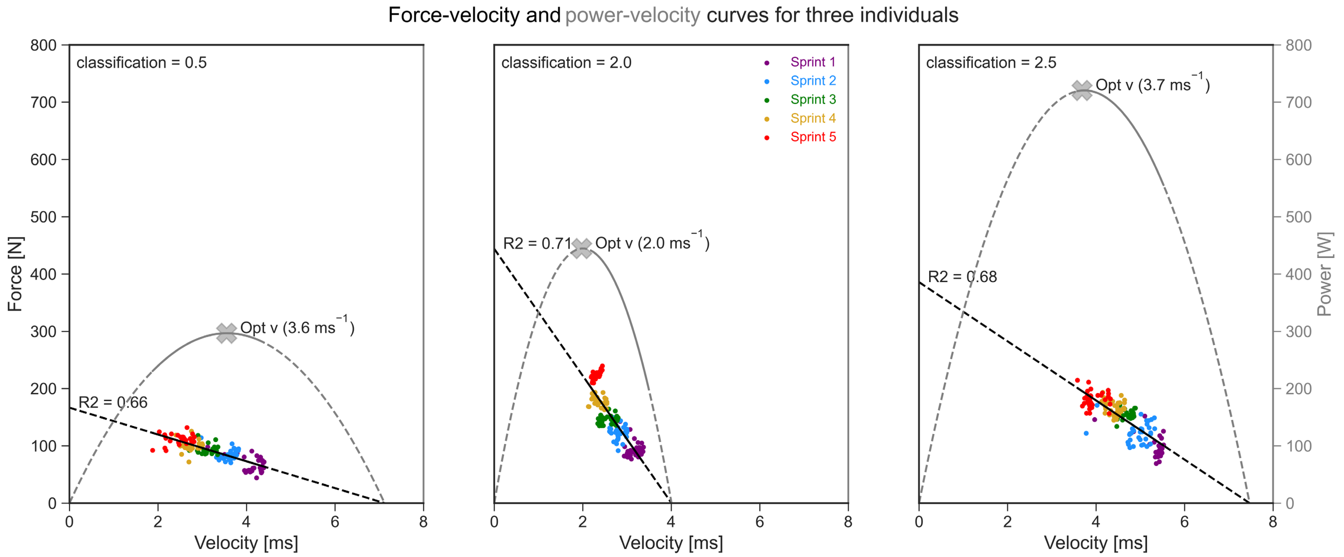


FIGURE 3 The individual linear force–velocity line (in black) and parabolic power-velocity (in gray) curve for three athletes. The scatterplots show the F_{max} per push versus the V_{max} per push for all individual pushes per sprint, excluding the first push. The linear regression line was plotted in a solid black line and extrapolated with a dotted black line. The gray line shows the power output parabola: the solid line was calculated from the solid black line and the dotted gray line was calculated from the dotted black line.

resistance. These differences were performed with a decreased push frequency and increased push time, while recovery time and contact angle remained constant over the sprints. The linear force–velocity relation was modeled for every athlete ($R^2 = 0.71 \pm 0.10$) and differed a lot in slope and intercept among individual athletes. These individual force–velocity profiles can be used for training recommendations or technological changes to allow WCR athletes to better exploit the power generation capabilities of the musculoskeletal system.

The PO_{mean} consistently increased with every sprint at a higher resistance but showed no flattening-off in the range of the five sprints. In line with our hypotheses, an increase in resistance led to a total decrease of 35% in v_{mean} and a total increase of 150% in F_{mean}. Taken together, this led to an increase of 83% in PO_{mean} from S100 to S420. In contrast with our hypothesis, the PO_{mean} showed no flattening-off at higher resistances. Hintzy et al. and Veeger et al. conducted a comparable experimental study with increased resistances in able-bodied participants and reported a decreasing velocity, increasing force and an increasing power output. However, and different from our study, the mean power output in their studies was highest in the second highest resistance condition.[17,18] The v_{mean} in the sprint against the highest resistance of the current study (i.e., S420) was $2.6 \pm 0.5\,\mathrm{ms}^{-1}$, while these two previous studies reported mean velocities below $2\,\mathrm{ms}^{-1}$ at the highest resistance.[17,18] As already said, able-bodied participants are difficult to compare with WCR athletes, but it seems that the applied resistance in the context of our study was not high enough to reach their maximal power output.[20]

The increase in power output can be explained by longer push times (+62%) and similar contact angles that allowed for more time to couple the hand to the rim-wheel interface, to contract the upper-body musculature, and transfer forces to the wheelchair. These timing effects are similar to results from previous studies[17,18] and are predominantly caused by changes in the angular velocity of the rim-wheel interface.[32]

The linear force–velocity relation was found to be significant among all athletes but not all athletes reached the optimal velocity. Since wheelchair sprinting is a complex multi-joint movement and athletes employ different individualized wheelchairs under the constraints of different impairments, the linear force–velocity relation showed a lot of variation in slope and intercept (Table 3). Other studies that investigated the force–velocity relation in multi-joint movements, such as cycling[8–11] and arm-cranking,[12,13] found the same approximately linear relation and corresponding parabolic power-velocity curves. However, the current study only visualized part of the force–velocity-power spectrum (see Figure 3: solid lines were measured, dotted lines were extrapolated). Future research should confirm whether the power output parabola can be modeled more precisely, in all athletes, using a wider array resistance-based wheelchair sprint tests in wheelchair athletes.

Maximal power output experiments are preferably performed around the individual's optimal velocity.[33] Previous wheelchair-specific maximal power output experiments were conducted at a high resistance, in order to keep the maximal velocity for every individual below 3 ms^{-1}.[34,35] While looking at the individual variation of the optimal velocity (range: 2.0–4.0 ms^{-1}, Table 3), one shared guideline might not result in the maximal power output for every individual. Athletes with a coordination impairment (e.g., CP) might experience hand-speed related problems at a lower velocity, compared to athletes with non-coordination impairments. The sample size of the current study did not allow us to compare these two groups. Before elaborating on ways to individualize maximal power output tests, future research should explore whether experiments performed around an individual's optimal velocity indeed results in the highest power outputs.

4.1 | Practical implications

Athletes with a steep force–velocity curve can produce much more power at an increased resistance and are clearly limited by a high hand speed. These athletes might benefit from training their upper-body coordination at high velocities, achieved by propelling downhill or with strong tailwind.[36–38] On the other hand, athletes with a flatter force–velocity curve could improve more from training their force generating capacities. For instance, 8 weeks of bench press training, where similar muscles are used as in wheelchair propulsion showed that the 10 m wheelchair sprint performance increased.[39] Making this more wheelchair-specific by for example uphill sprinting or with additional weights on the chairs,[36] it is hypothesized that wheelchair sprint performance will improve even more.

Force–velocity profiling can also help identifying differences between two wheelchair configurations. For example, differences in force–velocity profiles have been examined between a seated and non-seated positions in cycling.[40] When translating to wheelchair rugby, where athletes often hit the wheel-rim interface instead of grabbing the handrim, the wheelchair configuration (e.g., a larger wheel size for a reduction of hand speeds) can be altered in order to better exploit the power generation capabilities of the musculoskeletal system.

4.2 | Limitations and future research

The current study did not use a counterbalanced study protocol for rolling resistances due to a different primary aim of the initial study. This potentially could have induced fatigue in the last sprint at the highest resistance. However, there were no differences between the pre and post-test at the resistance of S100 (Table S1). Besides, if loads were applied in a counterbalanced way, and we, therefore, would rule out a potential systematic confounding effect of fatigue, the results would even be more pronounced since force and power output increased with increased resistances (regardless of potential fatigue). Thus, the protocol appears valid to answer the research question.

The rolling resistance was based on an individual deceleration test on the rollers of the wheelchair ergometer.[24] However, due to small differences in the position of the athlete-wheelchair combination (e.g., left–right alignment, type of wheelchair) on the roller ergometer, small errors could occur and the initial resistance was not exactly the same between participants. Resistance increased with a fixed percentage and while the current study only looked at within-subjects effects, this had a limited influence on the results. However, researchers and practitioners often make a distinction between High-point ($HP \geq 2.0$) and Low-point ($LP \leq 1.5$) players.[1,3,21] HP players achieve higher power sprint power outputs and also reach higher mean and peak speeds during training and competition.[3,21] Future research should address the role of disability or classification on sprint performance within the current experimental context.

5 | PERSPECTIVES

The instrumented wheelchair ergometer enabled the exploration of increased rolling resistances during wheelchair sprints. Higher resistances resulted in higher power outputs, as a result of a lower velocity but a substantially higher force production. This could be explained by a lower frequency, higher push time and a similar recovery time and contact angle, which allows muscles to operate at a more optimal velocity (given their intrinsic force–velocity properties) and which could facilitate coordination and muscle recruitment. Individual force–velocity profiles can be established and used for training recommendations or technological changes to allow WCR athletes to better exploit the power generation capabilities of the musculoskeletal system.

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CONFLICT OF INTEREST STATEMENT

The authors declare that they have no conflict of interest.

DATA AVAILABILITY STATEMENT

The data that support the findings of this study are available on request from the corresponding author. The data are not publicly available due to privacy or ethical restrictions.

ETHICS STATEMENT

The study was approved by Loughborough University Research Ethics Committee.

PATIENT CONSENT STATEMENT

All participants provided written informed consent prior to participation.

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