Master's thesis

NTNU
Norwegian University of Science and Technology
Faculty of Medicine and Health Sciences
Department of Neuromedicine and Movement Science

Veera Turkki

Effect of speed on trunk and upper limb kinematics in manual wheelchair propulsion

Master's thesis in Master of Science in Physical Activity and Health
Supervisor: Anna Cecilia Severin
Co-supervisor: Jørgen Danielsen
May 2022

NTNU
Kunnskap for en bedre verden

Veera Turkki

Effect of speed on trunk and upper limb kinematics in manual wheelchair propulsion

Master's thesis in Master of Science in Physical Activity and Health
Supervisor: Anna Cecilia Severin
Co-supervisor: Jørgen Danielsen
May 2022

Norwegian University of Science and Technology
Faculty of Medicine and Health Sciences
Department of Neuromedicine and Movement Science

NTNU
Norwegian University of
Science and Technology

EFFECT OF SPEED ON TRUNK AND UPPER LIMB KINEMATICS IN MANUAL WHEELCHAIR PROPULSION

BACKGROUND

5 million wheelchair users in Europe[1].

Increasing speed affects timing and patterns of propulsion movements[2].

STUDY

1. 18 novice able-bodied participants
3D marker-based movement analysis
Wheelchair on treadmill
2.5% incline

2. Incremental test:
→ speed increased until max, then reduced for propelling until exhaustion
→ movements captured at 1.11m/s (submax) and max speed

3. Movement analysis:
→ elbow
→ shoulder
→ trunk

FINDINGS AT MAX SPEED

Factors limiting movement production at high speed may relate to technique and movement control, as well as fatigue

TRUNK
More flexion throughout the push
Less time to extend back upright

SHOULDER
More abduction throughout the push
Less extension and external rotation at start of push
Less time to extend before new push

ELBOW
More flexion and less pronation throughout the push
Less time for extension

CONSIDERATIONS

High speed challenges movement production in wheelchair propulsion
→ may affect daily activities and sports

Practitioners should consider movement adaptations at high speed in coaching, health care and rehabilitation

DID YOU KNOW?
Being an experienced wheelchair user doesn't automatically mean a more effective propulsion technique[3].

EFFECT OF SPEED ON TRUNK AND UPPER LIMB KINEMATICS IN MANUAL WHEELCHAIR PROPULSION
MSc thesis in Physical Activity and Health
Human Movement Science
Veera Turkki
Spring 2022

NTNU
Centre for Elite Sports Research – SenTIF
Trondheim - Norway

References

1. Ebels P. Disability in figures https://euobserver.com/disability/1182492012 [updated 22.11.2012. Available from: https://euobserver.com/disability/118249.
2. van Drongelen S, Arnet U, Veeger D, van der Woude LHV. Effect of workload setting on propulsion technique in handrim wheelchair propulsion. Med Eng Phys. 2012;35(3):283-8.
3. Veeger HEJ, Lute EMC, Roeleveld K, van der Woude LHV. Differences in performance between trained and untrained subjects during a 30-s sprint test in a wheelchair ergometer. Eur J Appl Physiol Occup Physiol. 1992;64(2):158-64.

Abstract

Speed has been shown to affect movement patterns in manual wheelchair propulsion, and previously, mainly predetermined or self-selected submaximal speeds have been studied in non-athletes. This study investigated the effects of speed on trunk and upper limb kinematics in propulsion. Eighteen novice, able-bodied individuals (10M, 8F; 33±1yrs) propelled a wheelchair on a treadmill with 2.5% incline during an incremental test. Three-dimensional movements were recorded for 30 seconds at 1.11m/s and at near maximal speed with an 8-camera Qualisys system (120Hz). Start and end of push phase (contact and release) were identified, and maximum and minimum joint angles, range of motion and peak movement velocities between the two speeds were compared by paired t-tests (α=0.05). Cohen's *d* was used to indicate effect size. Right side was analysed, since limb movement asymmetries in the sagittal and frontal planes were small (<6°). From submaximal to maximal speed, cycle length (p<0.018, *d*=0.62) and rate (p<0.001, *d*=1.75) increased, as push time (p<0.001, *d*=-3.00) decreased. Largest changes in movement patterns were seen as increased maximum (p<0.001, *d*=1.82) and minimum (p<0.001, *d*=1.79) trunk flexion. At contact, the shoulder was less extended (p<0.001, *d*=1.22), more abducted (p<0.001, *d*=-1.11), and less externally rotated (p=0.001, *d*=0.99), while the elbow was more flexed (p<0.001, *d*=1.84) and forearm less pronated (p<0.001, *d*=1.74). At release, the shoulder was more abducted (p<0.001, *d*=-1.28) and elbow more flexed (p=0.001, *d*=0.86). Movement velocities increased in all movement planes during push (p<0.05), with the largest increases detected in shoulder internal rotation and trunk flexion. At near maximal speed, there is less time to extend the trunk and shoulders before a new push, challenging propulsion movements. It is likely that factors limiting movement production at high speed relate to technique and movement control, in addition to fatigue.

Introduction

There are approximately 5 million wheelchair users in Europe [1], and it is estimated that ~2% of the world's population require a wheelchair [2]. Manual wheelchair (MWC) propulsion can be described as "walking with one's arms" [3], referring to it being a mode of ambulation demanding considerable effort from the upper extremities. Biomechanical studies on wheelchair propulsion commonly focus on kinematics and kinetics of propulsion movements [e.g., 3, 4-8], and thus help identify different movement patterns and musculoskeletal loads in propulsion, as well as technique and movement timing. Upper limb movements are commonly described, often with a focus on the shoulder, in order to obtain information about healthy / unhealthy movements and to contribute to the work of injury prevention [e.g., 3, 9, 10].

Kinematic studies usually describe movements in conditions with different mechanical demands, like varying speed [e.g., 3], incline [e.g., 6], or resistance [e.g., 11], and in groups with different impairment levels [e.g., 12]. Reported kinematic variables commonly include joint angles and angular displacements, angular velocities, and angular accelerations. In wheelchair propulsion analyses, some studies investigate the whole movement cycle, including both push (propulsion) and recovery phase [e.g., 4, 10]. Other studies may focus on the whole cycle for some variables, for example range of motion, and only the push phase for maximum and minimum joint angles [e.g., 3], thus investigating these joint angles when active propulsion takes place. It is also common to describe the movement timing variables from a whole cycle (like cycle rate) and extract joint angle data from specific points in time, like contact and release [13].

Both two-dimensional (2D) and three-dimensional (3D) MWC kinematics have previously been studied, but recently, many researchers argue for the use of 3D data. For example, both Boninger, Cooper [3] and Newsam, Rao [12] investigated the kinematics of upper extremities in 3D space, emphasizing the arcs of motion in all three planes instead of only the sagittal plane. Although wheelchair propulsion is a bilateral and cyclical action, there have been mixed results from previous studies on movement symmetry, and it seems to be affected by the time period from which it is assessed [14]. Goosey [13] stated that the evaluation of bilateral function may benefit performance enhancement and increase the understanding of injury aetiology. In MWC literature, it is common to use unilateral data [e.g., 10] or average data between sides [e.g., 15], although it has also been stated that the assumption of symmetry may result in errors in detecting different stroke patterns [16]. Therefore, researchers should assess the upper limb propulsion movements for asymmetries before deciding to use only unilateral movement data or to average data between sides.

To help preserve independence and quality of life of wheelchair users, it is important to understand the different techniques used for tasks with increased demands for the upper extremities [17]. Van Drongelen, Arnet [11] argued that different speeds should be considered in analyses of propulsion technique since speed influences the timing variables of propulsion. Similarly, Collinger, Boninger [5] suggested allowing participants to self-select their speed because the way a person propels their wheelchair daily may have a relationship with pathology. Many studies investigating the effects of different speeds on wheelchair kinematics have used either predetermined or self-selected speeds for varying the movement demands [3, 5, 7, 9, 18]. For example, Boninger, Cooper [3], used two set speeds at 1.3m/s and 2.2m/s, while Collinger, Boninger [5] used a self-selected speed (1.09±0.31m/s) as well as set speeds of 0.9m/s and 1.8m/s, the same as Koontz, Cooper [7]. Mason, Vegter [18] used two predetermined speeds of 0.83m/s and 1.67m/s to represent submaximal speeds in wheelchair athletes. Often, changes in speed are also variably combined with an increased incline on a treadmill, or a stationary wheelchair ergometer is used [6, 8, 11]. For example, Van Drongelen, Arnet [11] compared the effect of speed on a treadmill with the effects of resistance produced by incline or a pulley system.

Common findings with increasing speed include increased trunk flexion [e.g., 19, 20], shoulder flexion and ROM [e.g., 3, 7]. However, it is unclear how movement patterns in the arms and trunk change as speed approaches maximal speed, compared to more submaximal and comfortable propulsion speeds in non-athletes. Wheelchair propulsion at these, near maximal speeds likely imposes more challenging performance demands on an individual. From a motor functioning perspective as well, it is interesting to investigate an individual's highest speed and obtain information about the person's performance level and possible movement pattern changes with challenging conditions.

Study aim

The aim of this study was to examine how upper body kinematics of manual wheelchair propulsion changed between a submaximal speed (Baseline) and a speed close to the individuals' maximal speed (Max speed). It was hypothesized that Max speed challenges the ability to produce effective movements since there is less time available to complete active propulsion.

Methods

Data collection was conducted at the movement laboratory at SenTIF (Centre for Elite Sports Research) in Trondheim. Ethical approval was obtained from Norwegian Centre for Research Data (NSD, ID number: 216680), and the data collection was conducted in accordance with the ethical standards by the Norwegian National Committee for Medical and Health research ethics and the Declaration of Helsinki.

Participants

20 able-bodied participants were recruited, for whom the inclusion criteria were: 1) age from 18 to 69 years, 2) no ongoing musculoskeletal disorders in the upper limbs or other physical health

conditions affecting the activity of wheelchair propulsion, 3) no conditions of cognitive impairment affecting the ability to follow the test protocol and instructions. Data of 18 able-bodied participants (10 males, 8 females; 33±1 years; 75.2±11.4 kg) was accepted for analysis (Appendix 2). Due to bad quality data and early failure in the maximal speed test, 2 participants out of the original 20 were excluded. Several wheelchair propulsion studies have investigated able-bodied subjects either as controls, or as the main subjects [10, 21-26]. It has been argued that able-bodied individuals may be easier to recruit, and they may have less within-group variability than wheelchair users with differing impairment levels [10].

Experimental design and data collection

Kinematic data was collected together with physiological data for a larger project. Data collection included propelling a wheelchair on a motorized treadmill (Forcelink Technology, Culemborg, The Netherlands) at 2.5% incline. To ensure participants' safety during the test, the wheelchair was attached to a safety bar moving on rails along the treadmill, allowing smooth forward and backward movements of the wheelchair, but no sideways drifting on the treadmill. The participants performed a 5-minute warm-up on a 0.5% incline with a self-selected speed aiming to produce rate of perceived exertion (RPE) [27] scores of 7-9 on a 6-20 Borg scale at the end of the warm-up. They then propelled for 4 minutes at 3 different submaximal speeds with 2.5% incline as a part of a larger project. After this, an incremental test to voluntary exhaustion was conducted with 2.5% incline. The test started with a speed of 0.83m/s (3km/h), after which the speed was gradually increased by 1km/h per minute for men and by 0.5km/h per minute for women. Once the participants were unable to match the speed of the treadmill, i.e., they reached their Max speed and started to drift backwards on the treadmill, the speed was lowered down to the previous completed stage. Finally, the participants kept propelling until they reached physiological exhaustion and were unable to continue.

During the incremental test, kinematic data was recorded for 30 seconds at each stage, starting 15 seconds into the minute of each speed. Kinematic data for this study was analysed from the recordings at 1.11m/s (Baseline), as well as at the highest recorded speed (Max speed). The 1.11m/s trial was chosen to represent submaximal speed, as this speed is often used in biomechanical investigations [e.g., 4, 11, 19]. Data from the highest speed was accepted for analysis if the participant could keep up the speed without failing (drifting back on the treadmill) for at least 10 recorded seconds, that is minimum 25 seconds after the speed had been increased. Overall, the protocol allowed for at least 17 minutes (5 minutes warm-up at 0.69-0.97 m/s plus 3 x 4 minutes submaximal tests at 0.55-1.38 m/s) of wheelchair propulsion familiarization.

3D kinematics were recorded using an 8-camera Qualisys System. (Oqus Motion Capture System, Qualisys AB, Gothenburg, Sweden), capturing at 120Hz. Low-mass retroreflective markers (14 mm in diameter) were allocated to the following anatomical landmarks: spinous process of the C7, T5, and T12, manubrium, xiphoid process, and bilaterally on the posterior and anterior superior iliac spine, iliac crest, and bilaterally on upper extremities: acromion process, lateral epicondyle of humerus, medial epicondyle of humerus, middle of forearm, lateral and medial wrist, knuckle of 3rd finger (Figure 1). In addition, non-collinear clusters made up of three markers were placed on the lateral surface of the upper arms. To track the movements of the wheelchair, there were 4 retroreflective markers placed, one on the centre of each wheel and one on the rim of each wheel. The individual kinematic reference model for each participant was recorded as a static trial before each testing session, with the participant standing in front of the wheelchair on the treadmill, in the anatomical neutral position. The markers on the posterior superior iliac spines as well as on the medial epicondyles of humerus were removed after a static reference trial, as the rest of the markers stayed in place for the dynamic data collection.

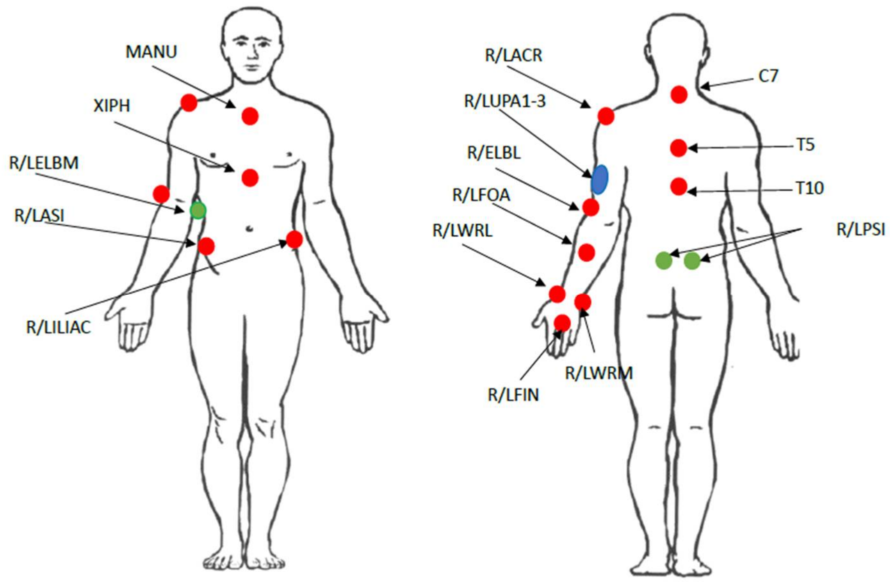


Figure 1. Retroreflective marker placements for the kinematic measurements. Blue indicates the cluster of 3 markers on the upper arm, and green indicates the markers removed after the static trial was performed.

All participants used the same lightweight wheelchair (Küschall K-Serie Attract wheelchair, SB42CM, Invacare, Norway, weight: 13.3 kg). In addition, there were safety stops on the back of the rails, preventing the wheelchair from dropping off the treadmill (Figure 2).

Figure 2. Data collection setup representing the attachment of the wheelchair to the safety bar on rails, and reflective markers. The mouthpiece was used for the cardiorespiratory data collection, which was not used for this project.

Data analysis

Kinematic data (initially processed in Qualisys Track Manager (Qualisys AB, version 2020.3, Gothenburg, Sweden)) was processed in Visual 3D 2021 (C-motion Inc., Germantown, MD, USA). In Visual 3D, the segments were initially defined according to the recommendations of the International Society of Biomechanics [28]. A standard 6dof optimization was used in Visual 3D to reduce skin artefacts, and the marker data was filtered with a 6Hz 4th order Butterworth low-pass filter before calculating the joint angles and velocities. Local coordinate systems were used to define the limb segment movements, and rigid models of the segments were assumed. Trunk movements were defined in relation to the global coordinate system, i.e., the laboratory coordinate system. Shoulder was further defined as a 3 degrees of freedom (dof) joint, and as the movements of the humerus in relation to the trunk. Elbow was defined as a 2 dof joint, and as the movements of the forearm in relation to the humerus. For presentation of joint angles, the XYZ rotation sequence was used. Positive rotations about X indicated flexion, about Y adduction, and about Z internal rotation. For clarity of expression, joint angles are described in anatomical terms, and thus internal-external rotation movements (about Z) are discussed as pronation-supination for the elbow and axial rotation for the trunk, and trunk movements about Y indicate lateral flexion.

Post-processing and analysis were performed in MatLab (The MathWorks, Inc., Natick, MA, USA) and Microsoft Excel (Microsoft 365, Version 2203, Build 16.0.15028.20152, 32-bit). A movement cycle was defined as the time from beginning of one push until the beginning of next (see definition of contact and release angles below). In this study the whole cycle (push and recovery phase) was considered and visually inspected, but the main analysis focused on the push/propulsion phase. This decision was made in order to study the period where active work is produced to propel the wheelchair forward. The outcome variables for the investigation of movement patterns in the upper limbs and trunk during the push phase were maximum and minimum joint angles and range of motion (ROM). Joint velocities are reported as the peak value in each movement plane (Table 3). In addition, the following cycle details were extracted: mean cycle time, maximum cycle time, cycle rate (cycles per minute), cycle length, mean push time and relative push time (% of cycle).

Since this study did not have access to rim forces, kinematic data was used to estimate the moments in time when participants contacted and released the rim of the wheel, indicating the start and end of push phase. The contact -event was defined as the moment in time when the velocity of the marker in the centre of the wheelchair wheel had been at its lowest and started to accelerate forward. The release -event was defined as the moment in time when the lateral wrist marker was at its lowest and most anterior position. Joint angles at these contact and release moments are further discussed as contact and release angles. These points in time are commonly used in kinematic descriptions of propulsion, e.g., by Brown, Knowlton [21] and Davis, Growney [10]. Symmetry between sides was inspected by comparing the joint angles bilaterally, and trunk movements at contact and release were determined based on right-sided data.

Statistical analysis

Data was visually inspected for consistency and quality, and QQ plots of variable residuals were used to assess that the data was approximately normally distributed. Possible differences between Baseline and Max speed, as well as between left and right side (asymmetry) were analysed with paired t-tests. Results are presented as mean values and standard deviations, with 95% confidence intervals and p-values (with $\alpha=0.05$). The effect size is presented with Cohen's *d,* and as defined by Cohen [29], with a cut-off point of $d=0.5$ indicating medium effect. All statistical testing was performed with SPSS (IBM SPSS Statistics for Windows, Version 27.0. Armonk, NY: IBM Corp).

Results

Movement data from the Baseline and Max speed trials was analysed from the incremental speed test (mean max speed 1.9±0.4m/s, range 1.3 - 2.5m/s, Appendix 2). RPE scores after the incremental speed test and propelling with a reduced speed until exhaustion were 17.4±1.6. After reaching their highest speed, most participants were able to continue propelling with the reduced speed for some time (64s ±40s), as seen in Table 1.

Table 1. RPE scores from the incremental speed test after propelling with a reduced speed until exhaustion, and time from speed reduction to end of the entire test.

Participant no.	RPE	Seconds*
1	19	**
2	19	80
3	17	20
4	20	20
5	17	80
6	19	30
7	18	80
8	18	40
9	15	30
10	16	50
11	17	40
12	14	180
13	18	45
15	18	37
16	15	65
17	17	95
18	18	104
19	18	90
n=18		
Mean	17.4	64
Sd	1.6	40
Max	20	180
Min	14	20

*Approximate times **Max speed of participant 1 was the same as reduced speed / no speed reduction made.

Effects of speed

Movement asymmetries are presented in Appendix 3. All sagittal and frontal plane asymmetries (max, min, ROM values) were below 6° at both speeds. Despite a few larger joint asymmetry values in shoulder internal/external rotation (10-19°), the bilateral movements were considered approximately symmetrical, and for the rest of this study, data from the right side (the dominant side of all participants) was analysed. Table 2 presents the effects of the two speeds on cycle details. All variables differed between Baseline and Max speeds, with main effects seen as reduced cycle time and increased cycle rate and cycle length.

Table 2. The effects of Max speed on propulsion cycle details. Values presented as mean (standard deviations).

	Baseline	**Max speed**	**95% CI**	**Effect size (*d*)**	**p-value**
Cycle time (s)	1.4 (0.3)	1.0 (0.1)	-0.6 - -0.3	**-1.67**	<0.001
Max cycle time (s)	1.6 (0.3)	1.1 (0.2)	-0.6 - -0.3	**-1.58**	<0.001
Cycle rate (cy/min)	44.0 (8.8)	62.6 (9.2)	13.3 - 23.8	**1.75**	<0.001
Cycle length (m)	1.6 (0.3)	1.8 (0.4)	0.1 - 0.5	**0.62**	0.018
Relative push time (% of cycle)	32.8 (3.3)	27.8 (4.1)	-6.8 - -3.1	**-1.37**	<0.001
Push time (s)	0.5 (0.1)	0.3 (0.1)	-0.2 - -0.2	**-3.00**	<0.001

Table 3. Effect of Max speed on joint angles in the upper extremities and trunk during push phase. Values presented as mean (standard deviations).

	Baseline	Max speed	95% CI	Effect size (*d*)	p-value
TRUNK					
Max flexion (°)	26.2 (12.4)	47.4 (12.9)	15.2 - 27.2	**1.82**	<0.001
Min flexion (°)	12.6 (8.3)	30.4 (11.9)	12.6 - 22.8	**1.79**	<0.001
ROM flx-ext (°)	13.6 (6.5)	17.1 (4.2)	0.8 - 6.1	**0.68**	0.013
Max lateral flexion (°)	1.4 (1.8)	1.8 (2.0)	-0.1 - 0.9	0.39	0.013
Min lateral flexion (°)	0.4 (1.7)	0.8 (1.7)	-0.1 - 0.9	0.45	0.083
ROM lateral flexion (°)	1.0 (0.6)	1.0 (0.8)	-0.3 - 0.2	-0.09	0.715
Max axial rotation (°)	2.1 (4.2)	2.7 (4.4)	-0.5 - 1.7	0.28	0.269
Min axial rotation (°)	0.9 (4.0)	0.7 (4.0)	-1.1 - 0.7	-0.12	0.641
ROM rotation (°)	1.2 (0.5)	2.0 (0.9)	0.3 - 1.3	**0.81**	0.004
Sagittal plane velocity (deg/s)	57.5 (23.4)	129.7 (31.3)	86.6 - 57.8	**2.48**	<0.001
Frontal plane velocity (deg/s)	7.1 (3.6)	12.6 (7.9)	1.8 - 9.2	**0.74**	0.006
Transverse plane velocity (deg/s)	-8.3 (4.2)	-13.1 (6.7)	-7.9 - -1.8	**-0.78**	0.004
SHOULDER					
Max flexion (°)	11.3 (8.0)	12.4 (7.9)	-0.2 - 2.5	0.43	0.096
Min flexion (°)	-47.0 (7.2)	-38.6 (6.5)	4.9 - 12.0	**1.22**	<0.001
ROM flx-ext (°)	58.3 (9.3)	51.0 (9.2)	-11.2 - -3.4	**-0.96**	0.001
Max adduction (°)	-22.7 (5.0)	-27.7 (4.9)	-7.6 - -2.5	**-1.01**	0.001
Min adduction (°)	-34.4 (6.0)	-38.6 (5.6)	-6.0 - -2.4	**-1.17**	<0.001
ROM add-abd (°)	11.7 (5.3)	10.9 (3.6)	-3.2 - 1.5	-0.19	0.456
Max internal rotation (°)	7.9 (9.5)	11.5 (13.7)	-0.4 - 7.7	0.46	0.075
Min internal rotation (°)	-31.8 (10.2)	-26.2 (11.7)	2.8 - 8.4	**1.04**	0.001
ROM rotation (°)	39.7 (9.1)	37.7 (8.8)	-5.4 - 1.4	-0.30	0.237
Sagittal plane velocity (deg/s)	194.6 (52.6)	230.8 (71.6)	4.8 - 67.5	**0.57**	0.026
Frontal plane velocity (deg/s)	83.2 (29.8)	126.9 (50.8)	20.1 - 67.3	**0.92**	0.001
Transverse plane velocity (deg/s)	163.8 (38.3)	279.8 (97.3)	69.1 - 162.9	**1.23**	<0.001
ELBOW					
Max flexion (°)	91.4 (6.5)	95.3 (7.1)	1.0 - 6.7	**0.69**	0.011
Min flexion (°)	45.2 (7.5)	50.1 (7.8)	1.9 - 7.7	**0.86**	0.003
ROM flx-ext (°)	46.2 (6.0)	45.2 (9.1)	-4.2 - 2.3	-0.15	0.546
Max pronation (°)	109.9 (14.0)	104.7 (12.1)	-10.2 - -0.2	**-0.54**	0.042
Min pronation (°)	77.0 (15.1)	72.0 (13.3)	-7.5 - -2.5	**-1.04**	0.001
ROM pro-sup (°)	32.9 (9.2)	32.7 (10.8)	-4.6 - 4.2	-0.02	0.928
Sagittal plane velocity (deg/s)	-297.7 (78.0)	-383.2 (133.4)	-137.2 - -33.7	**-0.82**	0.003
Transverse plane velocity (deg/s)	-134.8 (56.4)	-214.8 (71.7)	-130.4 - -29.8	**-0.79**	0.004

Negative shoulder adduction values represent abduction, and negative internal rotation values represent external rotation. Bolded Cohen's *d* values are d≥0.5 or d≤-0.5.

Table 4. Joint angles at contact and release at Baseline and at Max speed.

Joint angle (°)	Baseline	Max speed	95% CI	Effect size (*d*)	p-value
Contact angle					
TRUNK					
Flexion	12.7 (8.2)	30.4 (11.9)	12.5 - 22.8	**1.77**	<0.001
Lateral flexion	0.9 (1.7)	1.1 (1.8)	-0.2 - 0.7	0.25	0.314
Axial rotation	1.4 (4.1)	1.1 (4.2)	-1.3 - 0.7	-0.15	0.538
SHOULDER					
Flexion	-47.0 (7.2)	-38.6 (6.5)	4.9 - 12.0	**1.22**	<0.001
Adduction	-23.1 (5.5)	-29.1 (6.0)	-8.8 - -3.2	**-1.11**	<0.001
Internal rotation	-31.7 (10.4)	-26.2 (11.7)	2.6 - 8.3	**0.99**	0.001
ELBOW					
Flexion	80.0 (9.0)	91.8 (8.4)	8.5 - 15.0	**1.84**	<0.001
Pronation	93.3 (20.4)	81.1 (17.7)	-15.8 - -8.6	**-1.74**	<0.001
Release angle					
TRUNK					
Flexion	23.7 (12.9)	45.3 (13.3)	15.4 - 27.6	**1.81**	<0.001
Lateral flexion	1.0 (1.9)	1.3 (2.0)	-0.4 - 0.8	0.20	0.426
Axial rotation	1.4 (3.9)	1.7 (4.1)	-0.8 - 1.4	0.14	0.574
SHOULDER					
Flexion	11.3 (8.0)	12.4 (7.9)	-0.2 - 2.5	0.43	0.095
Adduction	-26.9 (5.4)	-30.7 (4.4)	-5.4 - -2.3	**-1.28**	<0.001
Internal rotation	7.9 (9.5)	11.5 (13.7)	-0.4 - 7.7	0.46	0.075
ELBOW					
Flexion	45.3 (7.5)	50.1 (7.8)	1.9 - 7.7	**0.86**	0.003
Pronation	108.9 (14.8)	104.7 (12.0)	-9.3 - 0.7	-0.44	0.090

Bolded Cohen's d values are $d \geq 0.5$ or $d \leq -0.5$.

The effects of Max speed on joint angles and peak joint velocities in the arm and trunk during push phase are presented in Table 3. Joint angles at contact and release angles are presented in Table 4. Joint excursions during push are presented in Figures 3, 5 and 6, and of the whole cycle (push and recovery phase) in Appendix 1. The largest changes in trunk movements between the two speeds were seen in the sagittal plane (Figures 3 and 4). The trunk was more flexed throughout the push, with increasing flexion towards release at both speeds. ROM in the sagittal plane increased slightly (by 3.5°) with Max speed. Differences in maximum trunk lateral flexion (Baseline: 1.4°±1.8, Max: 1.8°±2.0) and axial rotation (Baseline: 2.1°±4.2, Max: 2.7°±4.4) were small between the two speeds. Peak velocities in the trunk increased in all three movement planes during push phase, with the highest velocity reached during flexion at both speeds (Table 3).

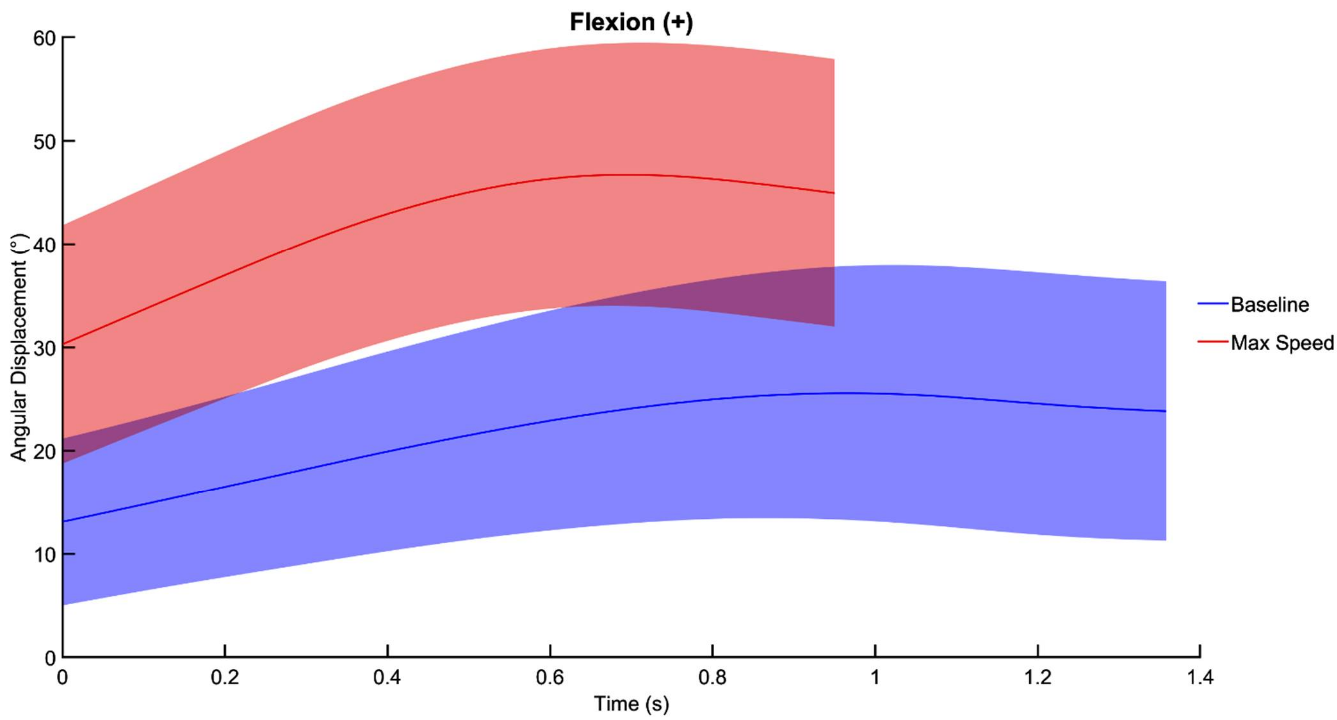


Figure 3. Flexion-extension movements of the trunk with two different speeds. The graph displays the group mean push phase with standard deviation.

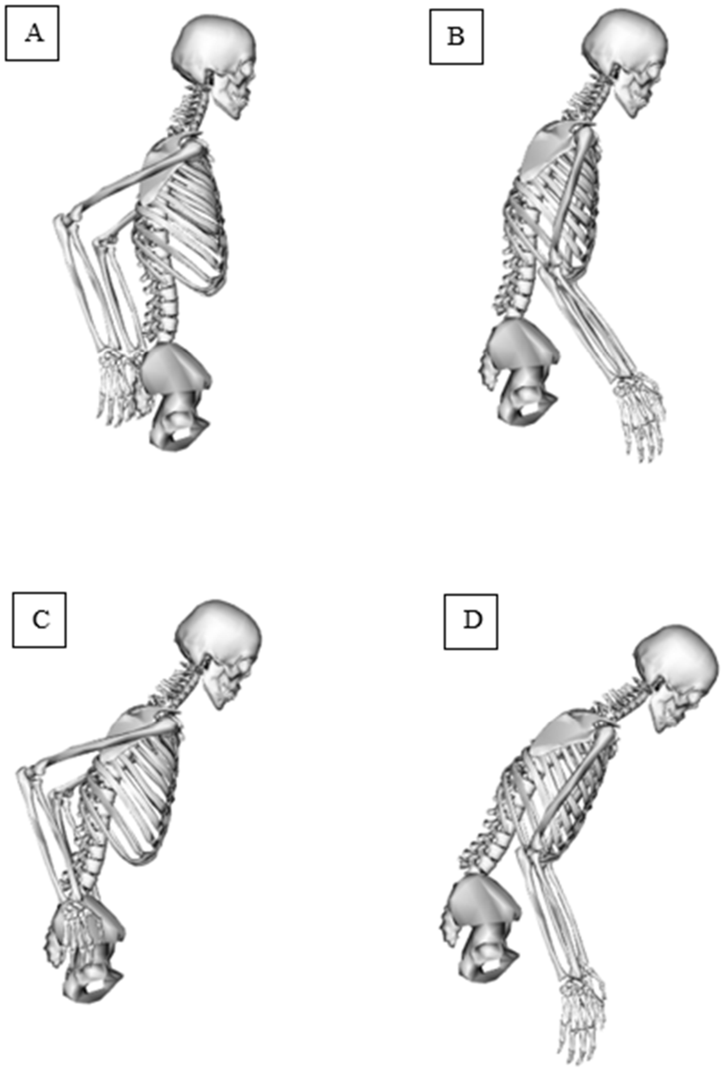


Figure 4. Upper body positions at contact (A, C) and release (B, D) at Baseline (upper row) and at Max speed (lower row).

Max speed produced differences in shoulder movement patterns on all planes, as presented in Figure 5. The largest differences were detected in the sagittal plane, with slightly smaller differences in the frontal and transverse planes. While there was less shoulder extension at contact with Max speed, the release angles in the sagittal plane were similar (Table 4), so the overall sagittal plane ROM in the shoulder was smaller at Max speed than at Baseline (Table 3). At contact, the shoulder was more abducted and less externally rotated with Max speed compared to Baseline. This increased abduction was present throughout the push until release. Shoulder adduction-abduction ROM did not change with speed. Shoulder internal rotation did not increase, but external rotation decreased. However, the changes in rotation movements were relatively small, and the slight change in rotational ROM in the shoulder did not reach significance. Peak velocities in the shoulder increased in all three movement planes, as displayed in Table 3, with the highest difference detected in internal rotation.

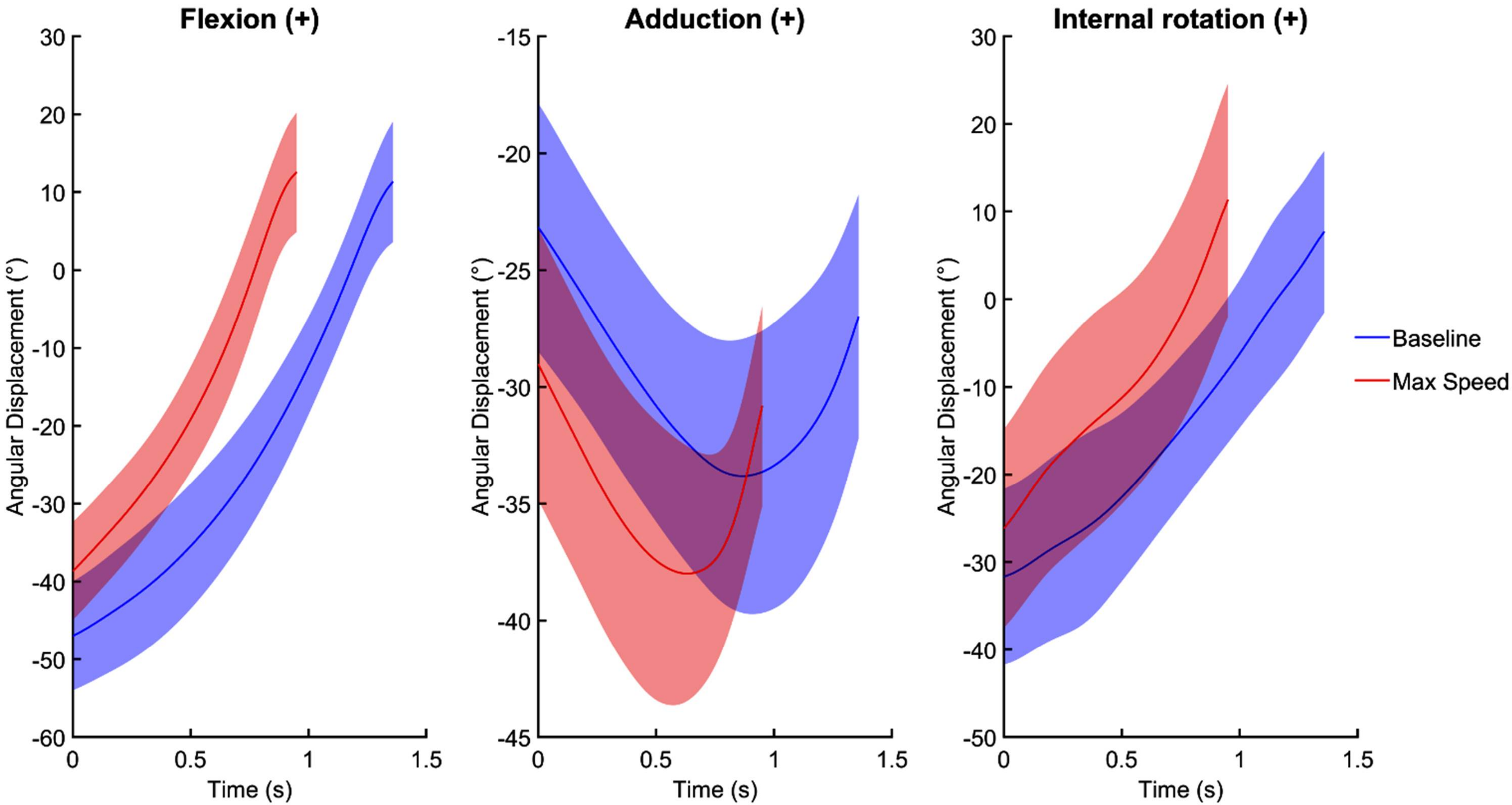


Figure 5. Flexion-extension, adduction-abduction, and internal-external rotation movements of the shoulder at the two speeds. The graph displays the group mean push phase with standard deviation.

Figure 6 presents elbow flexion-extension movements and forearm pronation-supination movements at Baseline and at Max speed. At Max speed, the elbow was more flexed and the forearm less pronated throughout the push phase. ROM did not change significantly in either of these movement planes. Peak joint velocities increased with Max speed and are reported of extension and supination movements in Table 3.

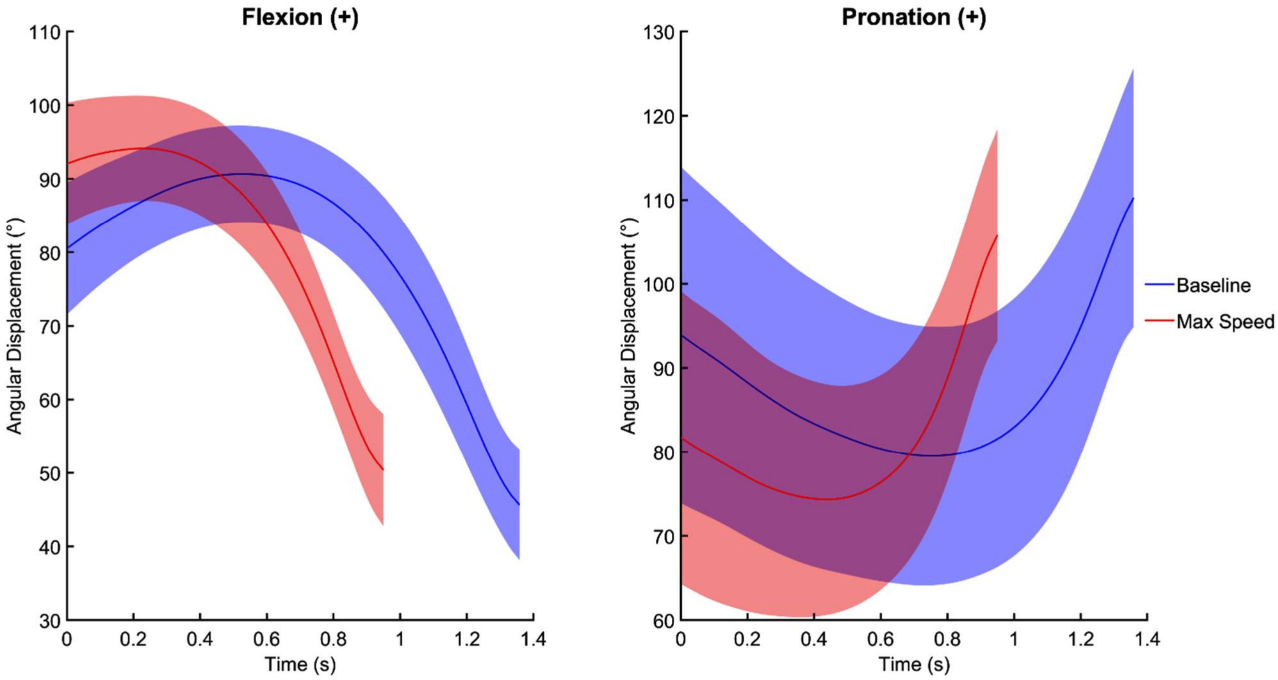


Figure 6. Flexion-extension movements of the elbow and pronation-supination movements of the forearm. The graph displays the group mean push phase with standard deviation.

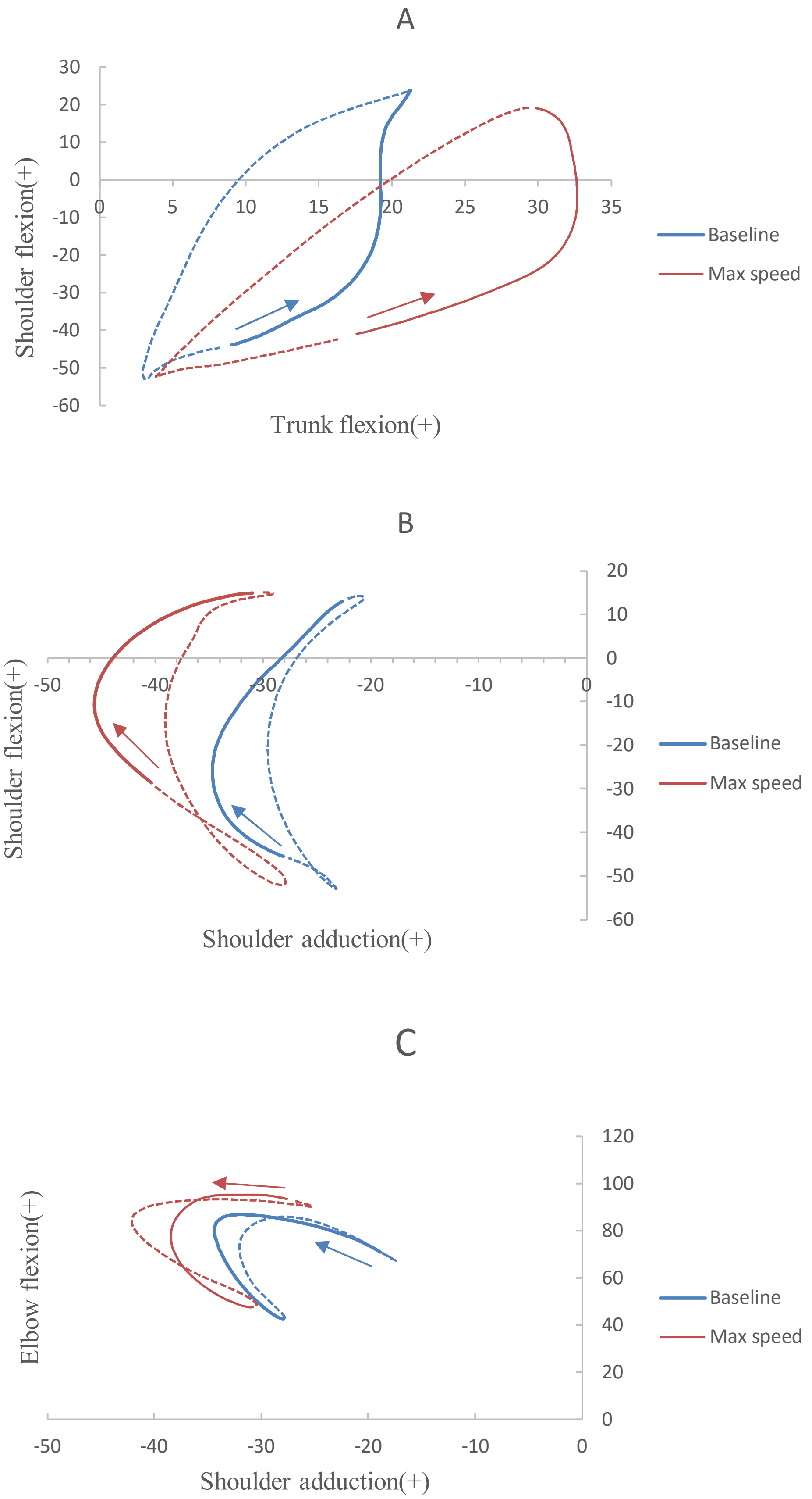


Figure 7, A-C: Angle-angle plots displaying A) shoulder and trunk sagittal plane movements, B) shoulder sagittal and frontal plane movements, and C) elbow sagittal and shoulder frontal plane movements. Movements displayed in degrees. Arrows represent movement direction in the push phase, and the dashed lines represent the recovery phase.

The inter-joint coordination (shown for one representative participant, Figure 7) shows that the coordination patterns between the two movements were affected by the increased speed from Baseline to Max speed. The general patterns of the movements between the two speeds remained similar, while often one of the movements occurred with larger excursions. Participants typically showed increases in trunk flexion, shoulder abduction, as well as elbow flexion. As it is rare to see a single optimal solution to a movement problem due to individual differences [30], the movements of a representative individual were selected to display these coordination patterns.

Discussion

This study aimed to investigate how upper body kinematics of manual wheelchair propulsion differed between Baseline and Max speed. This was done by having a range of able-bodied participants perform an incremental speed test on a treadmill. Key findings support the hypothesis that maintaining wheelchair propulsion at Max speed becomes a challenge to novice, able-bodied individuals, perhaps due to having less time to complete the propulsion movements. As the speed increased, the participants demonstrated increased trunk flexion, shoulder abduction, and elbow flexion, while shoulder extension decreased. These results indicate challenges in movement production as individuals approach their maximal speed in manual wheelchair propulsion. This information may help practitioners and coaches working with wheelchair users to identify different movement patterns with increasing speed, and to help individuals adapt their propulsion movements with challenging conditions.

Decreased push time and increased cycle rate found in this study were expected results and are supported by previous investigations on the effects of speed [e.g., 4, 7, 17]. Avoiding high cycle rates and preferring longer and smoother strokes are advisable in wheelchair propulsion, as rapid loading of the pushrim has been linked to e.g., increasing risk of median nerve injury [31]. In addition, high stroke rate may excessively overload the shoulder due to the repetitiveness of the task, which has been reported to increase the risk of injury [32]. However, maintaining low cycle rates becomes challenging with high speeds, suggesting that practitioners should instruct wheelchair users proper technique, as well as ways to avoid excessive amounts of propelling with high rates. To which extent an individual's inexperience in wheelchair propulsion affects the levels of stroke rate is a matter of deeper research. As an example, an investigation by Briley, Vegter [4] indicated that athletic wheelchair users produce lower rates of propulsion than non-athletic wheelchair users. However, since all participants in that study were accustomed wheelchair users, it remains unknown whether novice wheelchair users, or also able-bodied individuals, respond in the same way. It is possible that they may not be able to produce as long and smooth strokes with high speeds as the more experienced wheelchair users, which could lead to higher cycle rates and possible increases in injury risk.

Over the full movement cycle (push and recovery phase), the trunk, shoulder and elbow joints displayed different joint excursions at Max speed compared to Baseline. Further, clear increases in movement variability were detected after release, likely due to the participants being inexperienced in manual wheelchair propulsion. When focusing further on the push phase, we observed increased trunk flexion at contact and continuously until release. This may be an indication that the participants struggled to keep up with the movements and experienced a lack of time to retain a fully upright posture before the next cycle began. As a result, the whole cycle occurred with the trunk leaning more forward, affecting the upper limb movements as well. For example, the reduced shoulder extension at contact was possibly also a consequence on insufficient time for resetting the posture before the new cycle. With wheelchair users, this increased flexion of the trunk at higher speeds is a common finding [e.g., 19, 20, 33]. As for shoulder movements in the sagittal plane, maximum shoulder flexion did not change in our study, and flexion-extension ROM even decreased. In addition, there were also significant increases in shoulder flexion-extension velocities. However, increases in maximal shoulder flexion [7], and shoulder flexion-extension ROM with higher speeds [3] have been reported previously with wheelchair users. The differing results between our study and some previous ones may be due to novice, able-bodied individuals having more challenges in finding a proper technique with little practice and maintaining upper limb movements at high speeds as there is less time to complete a propulsion cycle.

Further, in our study, propulsion movements in the shoulder occurred in an increasingly abducted position with near maximal speed. In contrast however, Boninger, Cooper [3] found that subjects propelled the wheelchair with arms increasingly adducted, i.e., closer to the trunk with increasing speed. In addition, the shoulder abduction appeared to be minimal at contact in our study, as Koontz, Cooper [7] reported it being minimal at the end of push (release). This difference is perhaps due to the participants in that study being experienced individuals who used manual wheelchair propulsion as a primary mode for mobility, and thus demonstrated better technique. For inexperienced individuals not familiar with good propulsion techniques, increased abduction could be a compensation mechanism for the insufficient time available to reset the posture. In addition, with the trunk more flexed throughout the cycle, it is possible that the participants adapted a more abducted shoulder to rely better on the involvement of the trunk and perhaps as an adaptation to the arms getting tired as well.

Our study showed a pattern of increasing internal rotation in the shoulder towards release, or rather a reduction of external rotation, which is in agreement with Briley, Vegter [4] and Collinger, Boninger [5]. However in contrast, some studies have shown shoulder internal rotation decreasing during push phase [3, 10], and being minimal at the end of push (release) [7]. It is noteworthy that in our study, there were large variations in rotation movements of the shoulder between participants. Although no significant changes in maximal internal rotation were detected with Max speed, the velocity of the movement increased significantly, reflecting challenging conditions for shoulder rotations.

At Max speed, increases in elbow flexion were observed throughout the push, most prominently at contact. However, elbow flexion-extension ROM did not change in our study, which is in line with e.g., Boninger, Cooper [3]. It is likely that the more forward-leaning posture in the trunk necessitated increased elbow flexion throughout the push, in order to be able to grab the pushrim and produce propulsion. Also, lack of time for movement production may have made extending the elbow before a new push more challenging at Max speed. Further, decreased forearm pronation at Max speed, with pronation-supination ROM staying unaffected, likely indicate that changes in the movements of proximal joints affect the forearm positions during push phase as well.

It is likely that both neuromuscular and respiratory fatigue played a role in determining Max speed. However, the relatively long time periods that the participants were able to continue propelling after speed reduction, suggest that not only respiratory fatigue limited movement production, but also some aspects of technique and movement control. Deeper analysis into the different techniques adapted by novices at near maximal speed wheelchair propulsion is a matter for further research. In addition, it would be interesting to investigate whether testing the individuals' maximal speed would produce different results if the speed increases were done more quickly and with the participants less fatigued.

Strengths, limitations, and future considerations

As there are some aspects limiting the representability and comparability of the presented results, there are several strengths as well. Firstly, all participants used the same wheelchair, of which the seat height or other measures were not adjusted individually. This could have affected movement patterns, especially at Max speed. Secondly, the wheelchair was, for safety reasons, attached to safety rails allowing only forward-backward movements of the wheelchair, removing the natural sideways steering movements. However, the rails can be seen as a strength as well. They allowed more stable conditions for the clinical testing, thus enabling better investigation and comparison of selected variables, despite lacking some representativeness of natural wheelchair propulsion surroundings.

Further, the timing of contact and release moments of the push phase could not be detected as accurately as would have been with force sensors on the rim of the wheel. This makes direct comparisons with previous literature challenging, as many kinematic studies have used wheelchair pushrim sensors for accurate detection of contact time, [e.g., 3, 5, 7, 16]. Overall, varying study setups and outcome variable definitions in previous literature produce challenges for comparisons. It also varies among studies whether whole cycle movements have been analysed, or if push and recovery phases are discussed separately.

The participants being able-bodied and inexperienced in wheelchair propulsion may be seen as a limitation to the representativeness of manual wheelchair propulsion biomechanics. As the participants were not accustomed to the activity, it is assumed that their maximal speeds would be lower than those of experienced wheelchair users, limited by insufficient technique. However, in investigations of wheelchair propulsion biomechanics, it is important to recognize the value of having as subjects both individuals who use a manual wheelchair in their everyday life, as well as able-bodied persons. In fact, inexperience in wheelchair propulsion does not automatically mean a less effective technique [26], and having able-bodied individuals as participants may be an asset in wheelchair propulsion research. The learning curve in propulsion movements is shown to be quite steep, and both novice and able-bodied individuals seem to improve in propulsion efficiency and technique relatively fast, already with a short practice period [34]. This ability to improve is important in preventing pain and injuries from developing due to bad technique. Thus, regardless of the subjects being experienced wheelchair users or able-bodied individuals, it should be considered that individual differences in technique exist, relating to experience level, motor learning processes [34] as well as differing neurological impairment levels [10]. Future studies comparing groups with different levels of experience in manual wheelchair propulsion should investigate these issues further.

A strength in our study is the use of individual near maximal speeds, as this enabled the investigation of manual wheelchair propulsion relative to the individual performance level. It has been recognised before that in order to detect significant changes in movement patterns, it is essential to have a high enough speed. For example, Veeger, van der Woude [19] found changes in cycle time and push time with increasing speed, but not in push angles (the difference between contact and release angles), and they hypothesized that the fastest speed they used (1.39m/s) may have been too slow for changes to occur. In the future, it would be interesting to compare the results of able-bodied individuals and accustomed wheelchair users and thus evaluate the effect of practice and experience. In addition, investigating the kinetics and kinematics of maximal speed wheelchair propulsion together would allow a more complete evaluation of the effects of maximal speed on the musculoskeletal system.

Conclusion

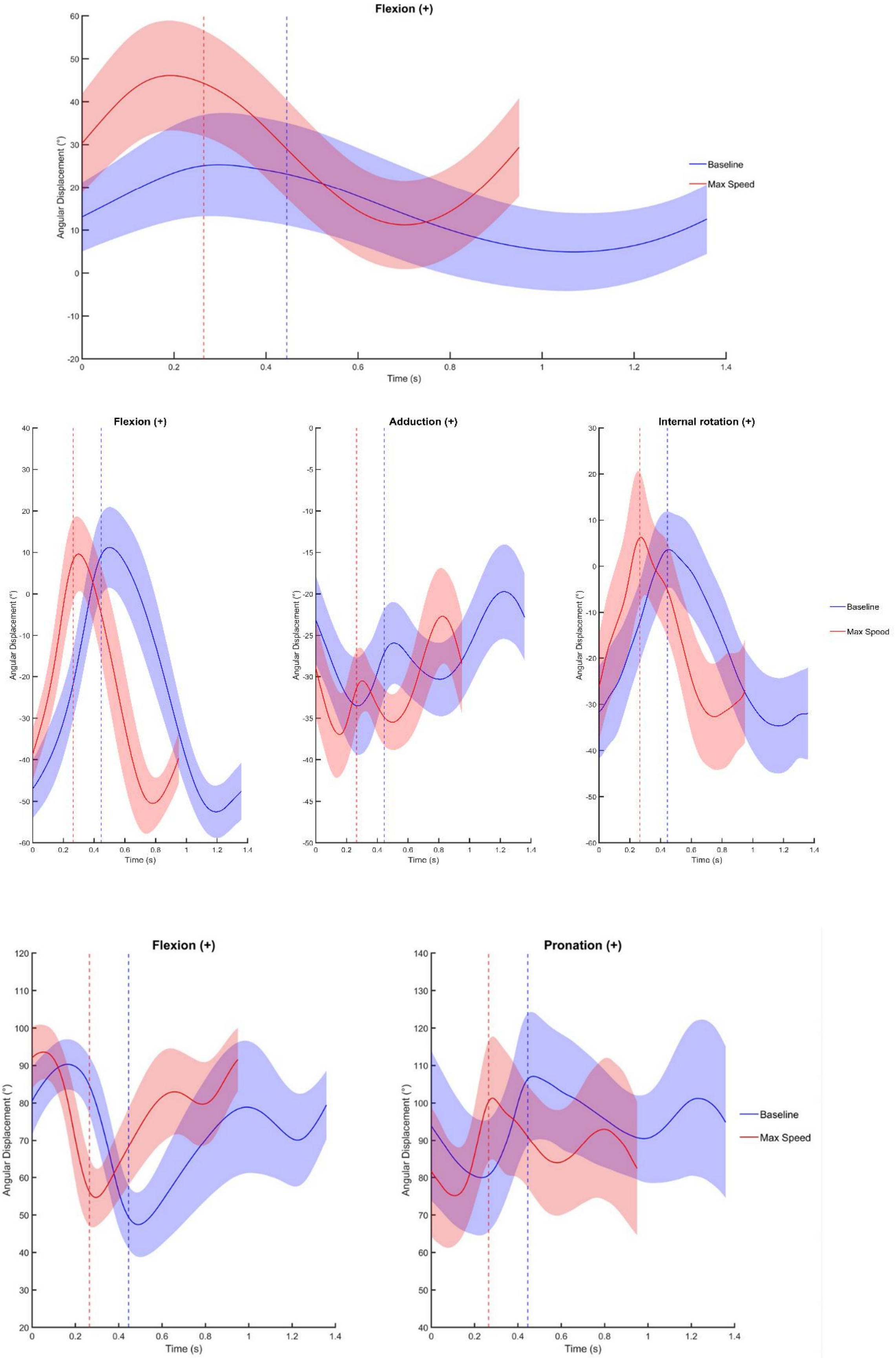
Speed affects the ways in which able-bodied individuals propel a wheelchair. It seems that lack of time for movement completion is a considerable factor in producing upper limb and trunk movements at high speeds. As hypothesized, nearly all participants were able to continue propelling for a considerable period after the slight reduction of speed. This indicates that technique and/or aspects of movement control, rather than solely physiological fatigue play a key role in limiting speed during manual wheelchair propulsion. With the trunk more flexed and the shoulders more abducted at near maximal speed, there are challenges in maintaining arm movements, and more effort is needed from the trunk. Moreover, high speed affects the joint angles at contact and release, resulting in more challenging arm positions at the start and end of the push phase. The results of this study can offer new perspectives to professionals working with both experienced and novice wheelchair users. Knowing how movements may change with near maximal speeds can help in coaching, as well as in rehabilitation and preventive work in health care. Future research is needed to investigate how these changes affect structural loading of the musculoskeletal system, and whether experienced wheelchair users change their movements differently to able-bodied individuals.

References

1. Ebels, P. *Disability in figures*. 2012 22.11.2012 [cited 2021 31.10.2021]; Available from: https://euobserver.com/disability/118249.
2. *Wheelchair Needs in the World*. 2021 [cited 2022 12.1.2022]; Available from: https://www.wheelchairfoundation.org/fth/analysis-of-wheelchair-need/.
3. Boninger, M.L., et al., *Shoulder and elbow motion during two speeds of wheelchair propulsion : a description using a local coordinate system.* Spinal Cord, 1998. **36**(6): p. 418-426.
4. Briley, S.J., et al., *Propulsion biomechanics do not differ between athletic and nonathletic manual wheelchair users in their daily wheelchairs.* J Biomech, 2020. **104**: p. 109725-109725.
5. Collinger, J.L.B.S., et al., *Shoulder Biomechanics During the Push Phase of Wheelchair Propulsion: A Multisite Study of Persons With Paraplegia.* Arch Phys Med Rehabil, 2008. **89**(4): p. 667-676.
6. Gagnon, D., et al., *Trunk and Shoulder Kinematic and Kinetic and Electromyographic Adaptations to Slope Increase during Motorized Treadmill Propulsion among Manual Wheelchair Users with a Spinal Cord Injury.* Biomed Res Int, 2015. **2015**: p. 636319-15.
7. Koontz, A.M., et al., *Shoulder kinematics and kinetics during two speeds of wheelchair propulsion.* J Rehabil Res Dev, 2002. **39**(6): p. 635-649.
8. Slowik, J.S., et al., *The influence of speed and grade on wheelchair propulsion hand pattern.* Clin Biomech (Bristol, Avon), 2015. **30**(9): p. 927-932.
9. Bossuyt, F.M., et al., *Compensation strategies in response to fatiguing propulsion in wheelchair users: Implications for shoulder injury risk.* Am J Phys Med Rehabil, 2019. **99**(2): p. 91-98.
10. Davis, J.L., et al., *Three-dimensional kinematics of the shoulder complex during wheelchair propulsion : A technical report.* J Rehabil Res Dev, 1998. **35**(1): p. 61-72.
11. Van Drongelen, S., et al., *Effect of workload setting on propulsion technique in handrim wheelchair propulsion.* Med Eng Phys, 2012. **35**(3): p. 283-288.
12. Newsam, C.J., et al., *Three dimensional upper extremity motion during manual wheelchair propulsion in men with different levels of spinal cord injury.* Gait Posture, 1999. **10**(3): p. 223-232.
13. Goosey, V.L., *Symmetry of the elbow kinematics during racing wheelchair propulsion.* Ergonomics, 1998. **41**(12): p. 1810-1820.
14. Chénier, F., J. Malbequi, and D.H. Gagnon, *Proposing a new index to quantify instantaneous symmetry during manual wheelchair propulsion.* J Biomech, 2016. **51**: p. 137-141.
15. Boninger, M.L., et al., *Manual wheelchair pushrim biomechanics and axle position.* Archives of physical medicine and rehabilitation, 2000. **81**(5): p. 608-613.
16. Boninger, M.L., et al., *Propulsion patterns and pushrim biomechanics in manual wheelchair propulsion.* Arch Phys Med Rehabil, 2002. **83**(5): p. 718-723.
17. Russell, I.M., et al., *Modifications in Wheelchair Propulsion Technique with Speed.* Front Bioeng Biotechnol, 2015. **3**: p. 171-171.
18. Mason, B.S., et al., *Bilateral scapular kinematics, asymmetries and shoulder pain in wheelchair athletes.* Gait Posture, 2018. **65**: p. 151-156.
19. Veeger, D., L.H. van der Woude, and R.H. Rozendal, *Wheelchair propulsion technique at different speeds.* Scand J Rehabil Med, 1989. **21**(4): p. 197-203.
20. Tai Wang, Y., et al., *Three-dimensional kinematics of wheelchair propulsion across racing speeds.* Adapted physical activity quarterly, 1995. **12**(1): p. 78-89.
21. Brown, D.D., et al., *Physiological and biomechanical differences between wheelchair-dependent and able-bodied subjects during wheelchair ergometry.* Eur J Appl Physiol Occup Physiol, 1990. **60**(3): p. 179-182.
22. Leving, M.T., et al., *Changes in propulsion technique and shoulder complex loading following low-intensity wheelchair practice in novices.* PLoS One, 2018. **13**(11): p. e0207291-e0207291.
23. MacGillivray, M.K., et al., *Effects of Motor Skill-Based Training on Wheelchair Propulsion Biomechanics in Older Adults: A Randomized Controlled Trial.* Arch Phys Med Rehabil, 2020. **101**(1): p. 1-10.
24. Rodgers, M.M., et al., *Upper-limb fatigue-related joint power shifts in experienced wheelchair users and nonwheelchair users.* J Rehabil Res Dev, 2003. **40**(1): p. 27-37.
25. van der Woude, L.H., et al., *Optimum cycle frequencies in hand-rim wheelchair propulsion. Wheelchair propulsion technique.* Eur J Appl Physiol Occup Physiol, 1989. **58**(6): p. 625-32.

26. Veeger, H.E.J., et al., *Differences in performance between trained and untrained subjects during a 30-s sprint test in a wheelchair ergometer.* Eur J Appl Physiol Occup Physiol, 1992. **64**(2): p. 158-164.
27. Borg, G.A., *Psychophysical bases of perceived exertion.* Med Sci Sports Exerc, 1982. **14**(5): p. 377-81.
28. Wu, G., et al., *ISB recommendation on definitions of joint coordinate systems of various joints for the reporting of human joint motion—Part II: shoulder, elbow, wrist and hand.* J Biomech, 2005. **38**(5): p. 981-992.
29. Cohen, J., *Statistical Power Analysis for the Behavioral Sciences*. 2013: Academic Press.
30. Sidaway, B., G. Heise, and T. Zohdi, *Quantifying the variability of angle-angle plots.* Journal of Human Movement Studies, 1995. **29**: p. 181-197.
31. *Preservation of Upper Limb Function Following Spinal Cord Injury: A Clinical Practice Guideline for Health-Care Professionals.* J Spinal Cord Med, 2005. **28**(5): p. 434-470.
32. Linaker, C.H. and K. Walker-Bone, *Shoulder disorders and occupation.* Best Pract Res Clin Rheumatol, 2015. **29**(3): p. 405-423.
33. Julien, M.C., et al., *Trunk and neck kinematics during overground manual wheelchair propulsion in persons with tetraplegia.* Disabil Rehabil Assist Technol, 2014. **9**(3): p. 213-218.
34. Vegter, R.J.K., et al., *Initial Skill Acquisition of Handrim Wheelchair Propulsion: A New Perspective.* IEEE Trans Neural Syst Rehabil Eng, 2014. **22**(1): p. 104-113.

Appendix 1



Effect of speed on joint angles during the entire propulsion cycle. The graphs display the group mean cycle with standard deviation. The dashed vertical lines represent the end of push phase and beginning of recovery phase at the different speeds.

Appendix 2

Participant details and their maximal speeds reached in the incremental test.

Participant	Gender	Age	Body weight	Height	Right/left-handed	Max speed m/s
1	Male	35	92.2	1.87	Right	2.2
2	Male	42	78.2	1.88	Right	2.2
3	Male	30	68.4	1.74	Right	1.9
4	Female	30	61.6	1.75	Right	1.5
5	Male	60	95.2	1.86	Right	1.4
6	Male	26	82.4	1.80	Right	2.5
7	Male	25	81	1.84	Right	1.9
8	Male	26	94	1.83	Right	1.9
9	Female	26	77.5	1.72	Right	2.1
10	Male	29	87.1	1.96	Right	2.5
11	Female	30	80.5	1.69	Right	1.5
12	Female	30	57	1.68	Right	1.8
13	Female	53	66	1.62	Right	1.8
15	Female	27	59.5	1.57	Right	1.8
16	Female	29	70	1.68	Right	1.3
17	Male	25	78.5	1.78	Right	1.9
18	Male	42	76	1.79	Right	1.7
19	Female	30	65	1.66	Right	1.5
n=18						
Mean		33.1	76.1	1.76		1.9
Sd		9.9	11.7	0.10		0.4
Max		60.0	95.2	1.96		2.5
Min		25.0	57.0	1.57		1.3

Appendix 3

Symmetry of joint angles in the push phase of propulsion at Baseline (1.11m/s) and at Max speed.

Joint angle (ᴼ)	Right	Left	95% CI	Effect size (*d*)	p-value
SHOULDER **Baseline**					
Max flexion	11.3 (8.0)	9.6 (7.0)	-1.2 - 4.5	0.30	0.229
Min flexion	-47.0 (7.2)	-45.5 (5.5)	-3.5 - 0.4	-0.41	0.113
Flx - ext ROM	58.3 (9.3)	55.1 (8.1)	1.4 - 5.0	**0.89**	0.002
Max adduction	-22.7 (5.0)	-23.2 (5.9)	-2.0 - 3.0	0.11	0.653
Min adduction	-34.4 (6.0)	-34.1 (6.3)	-2.6 - 2.0	-0.07	0.786
Add - abd ROM	11.7 (5.3)	10.9 (2.7)	-0.9 - 2.5	0.25	0.310
Max internal rotation	7.9 (9.5)	18.3 (7.5)	-16.3 - -4.5	**-0.91**	0.002
Min internal rotation	-38.1 (10.2)	-19.0 (10.6)	-18.0 - -7.7	**-1.29**	0.000
Int - ext rotation ROM	39.7 (9.1)	37.3 (8.8)	-1.2 - 6.1	0.34	0.175
Max speed					
Max flexion	12.4 (7.9)	11.6 (6.2)	-2.0 - 3.6	0.14	0.559
Min flexion	-38.6 (6.5)	-35.0 (7.5)	-6.6 - -0.6	**-0.61**	0.023
Flx - ext ROM	51.0 (9.2)	46.6 (7.3)	2.2 - 6.6	**1.01**	0.001
Max adduction	-27.7 (4.9)	-25.3 (4.9)	-5.0 - 0.2	-0.47	0.071
Min adduction	-38.6 (5.6)	-36.7 (4.9)	-4.5 - 0.7	-0.37	0.142
Add - abd ROM	10.9 (3.6)	11.4 (3.3)	-1.9 - 0.9	-0.19	0.454
Max internal rotation	11.5 (13.7)	20.1 (7.5)	-15.8 - -1.5	**-0.62**	0.021
Min internal rotation	-26.2 (11.7)	-13.3 (9.3)	-18.5 - -7.4	**-1.20**	0.000
Int - ext rotation ROM	37.7 (8.2)	33.4 (6.7)	0.2 - 8.4	**0.54**	0.040
ELBOW **Baseline**					
Max flexion	91.4 (6.5)	90.5 (6.0)	-1.4 - 3.1	0.20	0.420
Min flexion	45.2 (7.5)	49.2 (7.7)	-7.6 - -0.2	**-0.54**	0.041
Flx - ext ROM	46.2 (6.0)	41.4 (8.4)	1.8 - 7.8	**0.83**	0.004
Max pronation	109.9 (14.0)	101.1 (17.5)	0.2 - 17.3	**0.53**	0.044
Min pronation	77.0 (15.1)	75.2 (14.6)	-5.6 - 9.2	0.12	0.615
Pro - sup ROM	32.9 (9.2)	25.9 (10.5)	3.3 - 10.7	**0.97**	0.001
Max speed					
Max flexion	95.3 (7.1)	92.5 (7.7)	0.6 - 4.9	**0.66**	0.015
Min flexion	50.1 (7.8)	52.7 (7.2)	-5.5 - 0.1	-0.49	0.061
Flx - ext ROM	45.2 (9.1)	39.8 (8.7)	2.9 - 8.0	**1.10**	0.000
Max pronation	104.7 (12.1)	95.3 (15.4)	1.5 - 17.3	**0.61**	0.022
Min pronation	72.0 (13.3)	70.9 (14.5)	-6.9 - 9.0	0.07	0.779
Pro - sup ROM	32.7 (10.8)	24.4 (8.8)	3.8 - 12.8	**0.95**	0.001

Bolded Cohen's *d* values ≥0.5 or ≤-0.5.