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ORIGINAL RESEARCH

Shoulder Pain Is Associated With Rate of Rise and Jerk of the Applied Forces During Wheelchair Propulsion in Individuals With Paraplegic Spinal Cord Injury

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Abstract

Objective: To investigate the association between propulsion biomechanics, including variables that describe smoothness of the applied forces, and shoulder pain in individuals with spinal cord injury (SCI).

Design: Cross-sectional, observational study.

Setting: Non-university research institution.

Participants: Community dwelling, wheelchair dependent participants (N=30) with chronic paraplegia between T2 and L1, with and without shoulder pain (age, 48.6±9.3y; 83% men).

Interventions: Not applicable.

Main Outcome Measures: Rate of rise and jerk of applied forces during wheelchair propulsion. Participants were stratified in groups with low, moderate, and high pain based on their Wheelchair User Shoulder Pain Index score on the day of measurement.

Results: A mixed-effect multilevel analysis showed that wheelchair users in the high pain group propelled with a significantly greater rate of rise and jerk, measures that describe smoothness of the applied forces, compared with individuals with less or no pain, when controlling for all covariables.

Conclusions: Individuals with severe shoulder pain propelled with less smooth strokes compared to individuals with less or no pain. This supports a possible association between shoulder pain and rate of rise and jerk of the applied forces during wheelchair propulsion.

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Shoulder pain is a common problem in the spinal cord injury (SCI) community. Although previous studies have reported a prevalence between 30% and 73%,[1-3] a recent investigation of a population-based sample demonstrated shoulder pain in 36% of individuals living with SCI in Switzerland.[4] Shoulder pain limits activities of daily living (ADL) and wheelchair propulsion,[3,5] and thus has an effect on independence and community integration.[6-8] Furthermore, 46% of individuals with paraplegia experience pain during sleeping hours.[6,7,9] To this effect, pain interferes with quality of life.[10,11] Pain may alter movement patterns and shoulder loads, further resulting in chronic degeneration of shoulder structures and chronic pain.[12]

Shoulder pain is multifactorial and has been related to female sex, having tetraplegia or complete paraplegia, and specific health conditions, such as spasticity and contractures.[4] The most common diagnosis of shoulder pain includes tendon degeneration, which has been positively associated with age, body weight, and

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time since injury (TSI) in wheelchair users with SCI.[13,14] Higher forces applied on the push rim have been associated with injuries at the wrist and shoulder,[15-17] and individuals who experience higher shoulder forces and moments during wheelchair propulsion presented more signs of shoulder pathology.[18] Consequently, high forces and moments should be prevented during manual wheelchair propulsion. Although it is mechanically more effective to propel with a greater propulsive moment (ie, with forces directed more tangential to the push rim and thus with greater fraction of effective force [FEF][19]), this might be accompanied by increased shoulder load, pathology, and physiological demand,[20] which is counterproductive. Interestingly, the direction of the applied forces has also been associated with pathology.[18] Bossuyt et al reported a reduction in total applied force and tangential force in startup propulsion when stressing the musculoskeletal system with a demanding propulsion task, but no change in radial force or FEF.[21] Furthermore, a higher frequency in repetitive tasks is a risk factor for injury and may ultimately result in pain at the shoulder.[22] To lower the risk of injury, it has therefore been recommended to reduce body weight and to propel with long and smooth strokes (long contact time and low frequency).[23,24]

Although forces applied to the push rim have been associated with shoulder loads and pathology, to date, there is limited evidence of an association between the forces applied on the push rim and shoulder pain itself.[18,25] The study by Walford et al is the only study to date to find biomechanical predictors for shoulder pain, including weaker shoulder adductor muscles, higher positive shoulder joint work, and less trunk flexion.[26] As a limitation, they stated that pain was represented as a binary outcome, as different levels of pain may more accurately describe how biomechanical measures are associated with pain. Traditionally, a push cycle is segmented into 2 phases. However, according to Kwarciak et al, the push cycle does not consist of only propulsion and recovery, but can be divided into a more granular structure.[27] Based on total force and negative and positive axle moments, the push cycle can be segmented into initial contact (IC), propulsion (PROP), release (REL), and recovery. Investigating the entire push phase by including IC, PROP, and REL can provide further insights into differences in force application between individuals with different levels of shoulder pain. Finally, based on ergonomic insights in which a strong relation between *frequency of tasks* and risk of injury was found,[28,29] it is also argued that the *variability in change offorce application* causes musculoskeletal strain and could be related to pain. Some previous studies described the smoothness of force application during the propulsive phase using the maximum rate of rise (ROR) of force.[22,23] However, ROR only describes the steepness of the first peak in the force signal of a push cycle. Jerk has been used to describe smoothness of motion as the rate of change (derivative) of acceleration of a movement.[30] Here, acceleration is the second and jerk is the third derivative of position. Similarly, rate of change of force (also called jerk, but as a measure related to the derivative of applied force) can be used as a measure of smoothness of force exertion over a complete push cycle in wheelchair propulsion. Jerk as such has not been extensively investigated in this population, and it has not been associated with shoulder pain.

Investigating how individuals with different levels of shoulder pain apply forces throughout the entire push phase of wheelchair propulsion may offer additional insight into the development of shoulder pain in wheelchair users with SCI. Therefore, the goal of this study was to investigate the association between propulsion biomechanics, including ROR and jerk of the applied forces, and shoulder pain in individuals with SCI. It is hypothesized that higher levels of shoulder pain will be associated with ROR and jerk of the applied forces.

List of abbreviations:

ADL	**activities of daily living**
FEF	**fraction of effective force**
HPG	**high pain group**
IC	**initial contact**
LPG	**low pain group**
MPG	**moderate pain group**
NRS	**numeric rating scale**
PC-WUSPI	**performance-corrected Wheelchair User Shoulder Pain Index**
PROP	**propulsion**
REL	**release**
ROR	**rate of rise**
SCI	**spinal cord injury**
TSI	**time since injury**
WUSPI	**Wheelchair User Shoulder Pain Index**

Methods

Study design and study population

This study had a cross-sectional, observational design. Ethics approval was obtained from the Ethics Committee of Northwest and Central Switzerland (project ID: 2015-192). Suitable participants were selected based on the Swiss Spinal Cord Injury Cohort Study community survey (Ethics Committee of Northwest and Central Switzerland project ID: 1008). The inclusion criteria were (1) age between 16 and 85 years, (2) chronic paraplegia between T2 and L1, and (3) wheelchair dependency for ADL. The exclusion criteria were (1) major trauma of the upper extremity in the past year, (2) history of shoulder surgery, and (3) ferromagnetic implants or a cardiac pacemaker. These criteria were required to allow magnetic resonance imaging, the results of which are not described in this article. Participants were requested to complete a pain questionnaire (ie, numeric rating scale [NRS], with 0 indicating no pain and 10 indicating the worst shoulder pain ever experienced for each side). The study information and the NRS were sent to individuals who met the inclusion criteria (n=137). Of the individuals who were willing to participate (n=58), a convenience sample of 30 participants was selected, including 15 participants who scored more than 3 on the NRS for at least 1 of the shoulders and 15 participants who scored less than 3. This threshold of 3 on the NRS was selected based on discussions with experts in the field to assure a threshold that was both specific and sensitive to identify individuals with shoulder pain. However, results from the Wheelchair User Shoulder Pain Index (WUSPI) were used to determine shoulder pain on the day of the measurement itself.[6]

Data collection

Upon arrival in the biomechanical laboratory, participants signed the informed consent, after which they completed the WUSPI questionnaire to determine shoulder pain.[6] The WUSPI questionnaire includes 15 questions about self-reported pain in the past week for 15 different ADLs. Responses are scaled from a minimum of 0 (no pain) to a maximum of 10 (worst pain ever experienced). During the measurement session, participants were asked to propel their personal wheelchair on a motorized treadmill (belt area, 250×120 cm).[a] Participants started propulsion in the middle of the belt, were not secured with elastic bands or sliding supports, and could propel freely within the physical constraints of the belt and its speed. For security reasons, an examiner who could immediately interfere in case of any risk was positioned next to the participant on the treadmill. A SmartWheel (24 inch)[b] was fitted on the personal wheelchair of the participant to capture forces and moments in the 3 global reference planes (Fx, Fy, Fz, Mx, My, Mz). The SmartWheel was fitted on the side where shoulder pain was highest according to the NRS. If the participants did not report pain, the SmartWheel was fitted on the dominant side. The original wheelchair setup of the wheelchair user was maintained to retain their natural propulsion technique. Before familiarization, rolling resistance was calculated via a drag test.[31] Rolling resistance was used to calculate the weight needed to regulate the external power output (ie, resistance) induced by a pulley system.[32,33] Participants performed 3 predefined speed-resistance combinations (trials) in a randomized order: 1.11 m/s, 20W; 1.11 m/s, 30W; and 1.11 m/s, 40W. These values were presumed to be feasible, as they were used in previous studies with individuals with paraplegia,[34] and resulted in similar push rim applied forces and timing characteristics as over ground propulsion at 1.11 m/s.[35] Data were collected from the last 30 seconds of a trial, after a constant propulsion speed had been reached.

Data analysis

Shoulder pain was defined by the performance-corrected WUSPI (PC-WUSPI) score, ranging from 0 to 150.[9] Based on the 50th and 75th percentile of the PC-WUSPI scores of the investigated sample (15.5 and 39, respectively), participants were divided in 3 groups that represented a low pain group (LPG; PC-WUSPI, <15.5), a moderate pain group (MPG; PC-WUSPI, 15.5-39), and a high pain group (HPG; PC-WUSPI, ≥39).

The 3 trials at 1.11 m/s and 20, 30, and 40 W were used for data analyses of the push rim applied forces and timing characteristics. Raw SmartWheel data were collected at 240 Hz and filtered through a fourth order Butterworth filter with a cutoff frequency of 10 Hz.[36] Concurrently, the data were segmented into push cycles according to Kwarciak et al.[27] Each cycle consisted of

Table 1 Outcome variables

Outcome Variable	Description
Push frequency, pushes/min	Number of pushes per minute
Push angle, degrees	Angle of the entire push phase
$Ftotal_{max}$, N	Average of all maximal values of total force in 3 dimensions identified per cycle
$Ftan_{max}$, N	Average of all maximal values of force tangential to the push rim identified per cycle (propulsive torque divided by rim-radius, as only kinetic data are available)
Fz_{max}, N	Average of all maximal values of force perpendicular to the wheel identified per cycle
Mean fraction of effective force, FEF_{mean}, %	$FEF = F^2_{tan}/F^2_{total}$ continuous over cycle, from which the mean value of the push phase per cycle was extracted, averaged over cycles per trial
Jerk	Monotonous increasing jerk: value of change of force as derived by Hogan and Sternard[34]; natural logarithm of the area under the curve of squared derivate of force. $Jerk = \log \int (d(Force)/dt)\hat{2}$
ROR of force	ROR = maximum of derivate of total force
Moment, Nm	Absolute value of the 3 components of moments as measured by the SmartWheel
Spatiotemporal wheelchair propulsion characteristics	Following Kwarciak et al[25]: Hand contact: when Ftotal is above a threshold of 2 N, recovery below threshold. Within hand contact: Propulsion when torque around wheel axle is propulsive greater than 2 Nm Initial contact and release when torque is not propulsive and lower than the threshold of 2 Nm Thresholds based on noise levels of system used
Cycle time, s	Average of cycle time identified per cycle
Contact time, s	Average of contact time identified per cycle
Recovery time, s	Average of recovery time identified per cycle
Relative contact time, %	Contact time relative to cycle time
Relative recovery time, %	Recovery time relative to cycle time
Relative initial contact time, %	Coupling time relative to contact time
Relative propulsion time, %	Propulsion time relative to contact time
Relative release time, %	Decoupling time relative to contact time

Abbreviation: N, Newton.

IC, PROP, REL, and recovery. Hand contact was defined as total applied force exceeding a threshold of 2 Newtons. Within hand contact, IC and REL have a braking torque around the axle, below the threshold, and PROP a propulsive torque above a threshold of 2 Nm. Thresholds were defined based on the noise level of the measurement wheel. All complete pushes recorded during the 30 seconds of propulsion were used for the analyses. Propulsion biomechanics were calculated using a custom MATLAB program,[c] normalized to cycle duration and averaged over all cycles measured during the trial to obtain a participant and condition specific representation. Table 1 shows all calculated outcome variables.

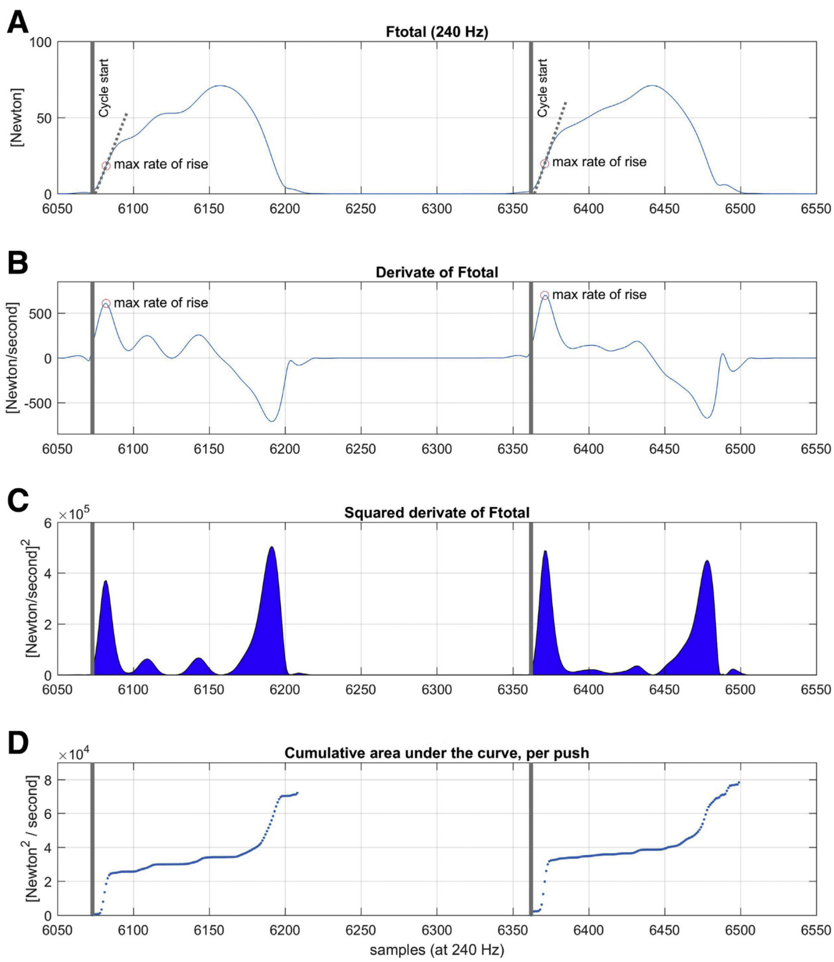

Fig 1 Definition of ROR and jerk. (A) Typical example of Ftotal applied at the rim. Vertical grey lines indicate the start of a push. (B) Derivate of Ftotal and the maximum per push, the ROR. This ROR is also indicated in (A) as the steepest rising flank of the signal. (C) Square of the derivate of Ftotal, the signal in (B), and its area under the curve (shaded). (D) Cumulative area under the curve (integral) of the signal in (C). The logarithm of the highest value of this monotonous increasing jerk is the total area under the curve per push and is used for further analysis.

Table 2 Participant characteristics: mean (standard deviation)

Group	Total No.	Sex	Height, cm	Weight, kg	BMI, kg/m^2	Age, y	Bonferroni Corrected *P* Value	Time Since Injury, y	Bonferroni Corrected *P* Value
LPG	15	12 men, 3 women	176.1±5.9	67.1±13.1	21.6±3.8	46.4±10.8		20.4±11.6	
MPG	7	7 men	179± 6.8	67.7±10.5	21.1±3.0	48.5±3.0		26.9±7.6	*
HPG	8	6 men, 2 women	173.9±8.9	65.4±11.1	21.6±3.4	52.6±10.1	*	27.2±12.4	*

NOTE. Data are presented as means ± SD.
Abbreviation: BMI, body mass index.
* Significant difference of the respective pain group in reference to the LPG after Bonferroni corrections for multiple comparisons.

Because jerk is a rather new variable in this field, it is described in more detail. Jerk has been used in describing smoothness of motion as rate of change of acceleration of a movement. Similarly, rate of change (derivate) of force can be used as a measure of smoothness of force exertion over a complete push cycle in wheelchair propulsion. The area under the curve from the squared derivate of total force, on a logarithmic scale, was calculated per cycle and averaged over cycles to have a single value per participant and condition. This definition of monotonously increasing jerk is derived from the work by Hogan and Sternad,[37] described in table 1 and visualized in figure 1.

Statistical analysis

One-way analysis of variance was used to compare subject characteristics and demographic variables between shoulder pain groups. When a significant difference between pain groups was found, Bonferroni post-hoc tests were performed. A mixed-effect multilevel analysis was used to identify the association between the dependent variables (propulsion biomechanics) and group of shoulder pain when controlling for sex, body weight, TSI, age, and height. These covariables were selected based on their association with shoulder loads or shoulder pain, as described in the introduction. The 3 different trials (20 W, 30 W, and 40 W at 1.11 m/s) were used as a grouping variable to account for between-subject variability. Bonferroni post-hoc tests were performed when a significant difference was found between pain groups. All statistical analyses were performed with Stata, version 14.[d]

Results

The characteristics of the 30 participating paraplegic manual wheelchair users are presented by shoulder pain group in table 2. Grouping based on level of shoulder pain resulted in 15 individuals in the LPG, 7 in the MPG, and 8 in the HPG. Owing to technical failure of the SmartWheel, 6 trials (1 trial each for 6 participants) were unusable. Significant differences in subject characteristics between groups are presented in table 2. More specifically, individuals in the HPG were significantly older compared with the LPG. Furthermore, individuals in the HPG and MPG had significantly greater TSI compared with the LPG.

ROR and jerk were significantly associated with shoulder pain group. ROR was significantly higher in the HPG compared with both the LPG and MPG, and jerk was significantly higher in the HPG compared with only the LPG group. The actual data for the variables are summarized in table 3, and significant findings are presented in table 4.

No significant difference was found among shoulder pain groups for spatiotemporal wheelchair propulsion characteristics, push frequency, maxima of forces, FEF, moments, and push angle between the defined groups. Figure 2 shows the group mean of the normalized Ftotal applied on the push rim for all 3 power levels and corresponding ROR.

Furthermore, it was found that women had a significantly smaller push angle ($\beta = -18.18$; SD, 8.86; $P = .04$) and lower contact time ($\beta = -0.09$; SD, 0.04; $P = .04$), but a greater mean FEF ($\beta = 0.19$; SD, 0.09; $P = .03$) than men. Heavier participants had significantly higher ROR ($\beta = 9.17$; SD, 3.78; $P = .02$) and spent more time on coupling ($\beta = 0.11$; SD, 0.04; $P = .01$). Taller participants spent less time on coupling ($\beta = -0.13$; SD, 0.06; $P = .02$). There were no other significant associations.

Discussion

This study demonstrated an association between the level of shoulder pain in individuals with paraplegia with ROR and jerk, 2 aspects of propulsion biomechanics. The higher ROR in the HPG shows that individuals with severe shoulder pain reach the peak force during each push within a shorter time. Higher ROR has

Table 3 Unadjusted mean ± SD of dependent variables that presented significant associations for each pain group

Variable	Group	Mean ± SD
Push angle, deg	LPG	104.1 ± 16.3
	MPG	110.6 ± 14.7
	HPG	99.8 ± 11.2
Jerk, N^2/s	LPG	11.62 ± 0.45
	MPG	11.84 ± 0.44
	HPG	11.91 ± 0.42
ROR of force, N/s	LPG	751.3 ± 205.3
	MPG	738.9 ± 173.2
	HPG	984.6 ± 283.3
Absolute contact time, s	LPG	0.51 ± 0.08
	MPG	0.54 ± 0.07
	HPG	0.48 ± 0.06
Relative initial contact time (%) − coupling time relative to contact time	LPG	3.00 ± 2.67
	MPG	3.32 ± 2.66
	HPG	3.77 ± 2.18
FEF mean, N	LPG	0.51 ± 0.15
	MPG	0.42 ± 0.13
	HPG	0.56 ± 0.16

Table 4 Results of mixed-effect multilevel analysis to identify the association between the dependent variables (propulsion biomechanics) and group of shoulder pain when controlling for sex, body weight, TSI, age, and height

Variable	Wald χ^2	*P* Value, χ^2		β	SE β	*P* Value	Bonferroni Corrected
Push angle, deg	9.64	.21	LPG				
			MPG	0.57	6.49	.93	
			HPG	−5.93	5.94	.32	
			Sex	−18.18	8.86	.04*	
			Weight	−0.36	0.30	.22	
			TSI	0.37	0.27	.17	
			Age	0.04	0.31	.90	
			Height	0.28	0.42	.50	
Jerk, N^2/s	17.40	.02†	LPG				
			MPG	0.27	0.15	.07	
			HPG	0.40	0.13	.00*	† LPG
			Sex	−0.08	0.20	.70	
			Weight	0.01	0.01	.06	
			TSI	−0.01	0.01	.35	
			Height	−0.00	0.01	.91	
			Age	−0.01	0.01	.37	
ROR, N/s	24.99	.00†	LPG				
			MPG	41.07	82.78	.62	
			HPG	284.13	75.48	.00*	† LPG, †MPG
			Sex	−39.27	113.08	.73	
			Weight	9.17	3.78	.02*	
			TSI	−5.05	3.40	.14	
			Height	−6.83	5.34	.20	
			Age	−1.38	3.94	.73	
Contact time, s	9.90	.19	LPG				
			MPG	0.01	0.03	.81	
			HPG	−0.03	0.03	.30	
			Sex	−0.09	0.04	.04*	
			Weight	−0.00	0.00	.22	
			TSI	0.00	0.00	.20	
			Height	0.00	0.00	.50	
			Age	0.00	0.00	.89	
Coupling time relative to contact time, %	22.97	.00	LPG				
			MPG	0.50	0.86	.56	
			HPG	0.45	0.79	.57	
			Sex	−0.58	1.18	.62	
			Weight	0.11	0.04	.01*	
			TSI	−0.02	0.04	.53	
			Height	−0.13	0.06	.02*	
			Age	0.07	0.04	.11	
FEF mean, N	9.56	.21	LPG				
			MPG	-0.03	0.06	.58	
			HPG	0.06	0.06	.28	
			Sex	0.19	0.09	.03*	
			Weight	0.00	0.00	.30	
			TSI	−0.00	0.00	.93	
			Height	0.00	0.00	.76	
			Age	−0.00	0.00	.15	

* Significant association.
† Significant difference of the respective pain group in reference to the LPG after Bonferroni corrections for multiple comparisons.

been related to shoulder pathology and may further affect the development of pathology in individuals who already experience severe pain.[18] The higher jerk in the HPG further demonstrates greater variations in force throughout the push phase in those with severe shoulder pain. Given that both ROR and jerk are an indication of smoothness, it can be suggested that participants in the HPG propel with less smooth stokes compared with individuals with less pain. The lack of differences in maximal total forces

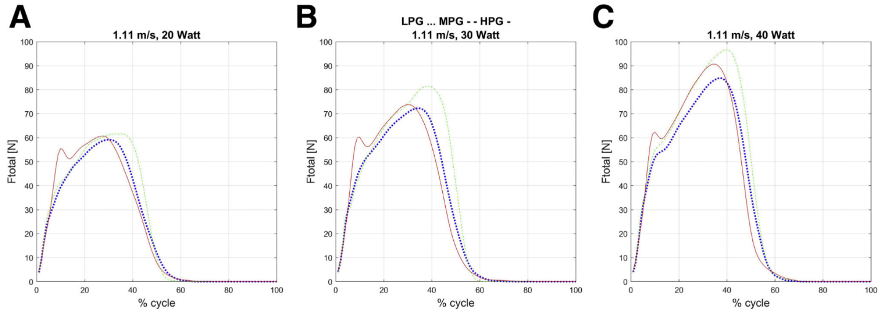

Fig 2 Ftotal for the 3 pain groups. Ftotal, cycle normalized to 101 points, for trial (A) 1.11 m/s, 20W; (B) 1.11 m/s, 30W; and (C) 1.11 m/s, 40W for the LPG, MPG, and HPG. Data represents mean values per group.

suggests that the main difference is not how much force is applied to the push rim, but rather the speed and variation with which this force is produced.

The current findings are in line with guidelines that aim to preserve upper limb functioning and suggest avoiding the application of high forces and reduce shoulder loads by applying long and smooth strokes.[22] These guidelines were based on previous associations between forces applied on the push rim and shoulder and wrist pathology.[15,18,24,26] This is the first study to find an association of shoulder pain in paraplegia with ROR (a commonly used variable for wheelchair propulsion characteristics) and jerk (a relative new variable that quantifies smoothness of the applied forces during the entire push phase). This stresses that smoothness is an important aspect of these guidelines in relation to individuals who experience severe levels of pain. Interestingly, when controlling for multiple comparisons, there were no differences between the LPG and MPG. However, there was a significant difference in ROR between the HPG and both the MPG and LPG. This supports the importance of not only comparing individuals with or without pain, but also differentiating between multiple pain levels. This may explain the different findings of Rice et al,[38] who did not find a difference in mean ROR of force between individuals with or without shoulder pain. The use of ROR and jerk, which can be measured fairly easy with a variety of instrumentation (instrumented wheels, ergometer), could aid clinicians to give directed feedback to alter propulsion biomechanics in wheelchair users, especially in those with shoulder pain.

Furthermore, this study demonstrated that different levels of shoulder pain were not associated with push angle or contact time. However, these variables were associated with sex, suggesting that sex may have a stronger association with temporal propulsion variables compared with pain itself. More specifically, women propelled with a smaller push angle and lower contact time, which may be owing to structural differences between men and women, as described by Schultz et al.[39] This is in line with the previously discussed guidelines and associates with studies reporting that women have more odds of having shoulder pain.[4,10] Interestingly, women also propelled with higher FEF, which further questions the role of FEF in shoulder loading and whether a higher or an optimal FEF should be the aim. Additionally, heavier participants had a higher ROR and spent more time on coupling, which may implicate less efficient propulsion biomechanics. In contrast, taller participants spent less time on coupling, which could be the result of structural differences and may implicate more efficient propulsion biomechanics.

Study limitations

The first limitation of this study is its cross-sectional design, which limits any sort of causal inference. Secondly, there was no information on the wheelchair or its setup, both of which are known to influence wheelchair propulsion kinetics[22] and may mediate the reported associations between propulsion biomechanics and sex, height, and weight. Finally, propulsion biomechanics were investigated during treadmill propulsion. Although previous studies reported similar propulsion biomechanics as during over ground propulsion,[35] differences may be present and treadmill propulsion does not resemble daily life situations. For example, some individuals may need a longer adaption period to treadmill propulsion, which may have affected propulsion biomechanics.

Conclusion

This cross-sectional observational study is the first to provide indications that individuals with severe shoulder pain (PC-WUSPI, $\geq$39) use less smooth strokes during wheelchair propulsion compared with individuals with less or no pain. These propulsion biomechanics have been associated with shoulder pathology and support the need of wheelchair training programs, especially in individuals who experience severe shoulder pain. Interestingly, this study demonstrated the value of a rather new variable (ie, jerk) to describe the smoothness of the entire push phase, which can be used to provide personal feedback and as one of many parameters to identify effectiveness of wheelchair training programs. Longitudinal research is needed to investigate causal inferences between propulsion biomechanics and the development of shoulder pain. Such studies should include measures of smoothness of applied force such as jerk and ROR and may also investigate the effect of the wheelchair and its setup.

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Suppliers

a. Bonte Technology B.V.
b. SmartWheel, 24 inch; Three Rivers Holdings, Inc.
c. MATLAB; MathWorks, Inc.
d. Stata, version 14; StataCorp LP.

Keywords

Rehabilitation; Shoulder pain; Spinal cord injuries; Wheelchairs

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