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Objective: To compare upper-limb joint power magnitude and distribution between the shoulder, elbow, and wrist during maximal acceleration (MAC) versus steady-state, self-selected speed (SSS) manual wheelchair propulsion.

Results: Wilcoxon signed-rank testing revealed shoulder power was larger for MAC versus SSS (median peak, 101.5W; interquartile range [IQR], 74.6; median peak, 37.7W; IQR, 22.9; respectively) (P < .01). Elbow and wrist power were unchanged. Peak shoulder power fraction was larger for MAC versus SSS (median peak, .055; IQR, .110 vs peak, .870; IQR, .252) (P < .01). Peak elbow power fraction was smaller for MAC versus SSS (median peak, −.012; IQR, .144 vs peak, .146; IQR, .206) (P < .05). Peak wrist power fraction was smaller for MAC versus SSS (median peak, −.058; IQR, .057 vs peak, −.010; IQR, .150) (P < .05).

Conclusions: Power at the shoulder was larger than at other joints. Peak shoulder joint power and power fraction was larger during MAC versus SSS propulsion. Elbow and wrist power fractions were smaller for MAC versus SSS propulsion. Higher joint power, present under MAC, may predispose manual wheelchair users with spinal cord injury below T1.

Interventions: Not applicable.

Main Outcome Measures: Propulsive joint power magnitude and fractional distribution among upper-limb joints.

Design: Cross-sectional biomechanic study.

Setting: Research university and teaching hospital.

Participants: Volunteer sample of 13 manual wheelchair users with spinal cord injury below T1.

Key Words: Biomechanics; Kinetics; Movement; Rehabilitation; Wheelchairs.

SHOULDER PAIN, CARPAL TUNNEL syndrome, and other upper-limb pathologies are common problems in manual wheelchair users.1,2 They result from high mechanical stress accompanying the propulsion of manual wheelchairs. Overloading injuries of this nature require a biomechanical analysis of the propulsion technique to understand the relationship to injury.3

In the analysis of locomotion, joint power offers insight into the biomechanics of the activity. Joint power has been used in gait analysis to study biomechanics,4,5 as an outcome to evaluate interventions,6,7 and as a reference against which to measure the accuracy of clinical evaluations.8,9 Mechanical power describes the rate of work performed by the muscles. Either generative or absorptive, it is the result of concentric or eccentric muscle contractions, respectively. Applied to joint movement, power is the product of net joint moment and joint angular velocity and thus captures the effects of both net muscle effort and speed of joint rotation.10

Guo et al11 investigated the general characteristics of upper-limb segmental mechanical energy and power flow in healthy male adults propelling a manual wheelchair under steady-state conditions over a laboratory floor. Using a kinematic data collection system and an instrumented pushrim, they determined that at the beginning of the propulsion stroke, power is generated by the upper arm, or transferred from the trunk downward to the forearm and hand. At the end of the propulsion stroke, joint power is transferred upward to the trunk from the forearm and upper arm. They found that work computed via power flow analysis was greater than work computed by the change in mechanical energy, implying that the power supplied to the upper-limb segments is greater than is required. Without elaboration, they speculated that guidelines for the configuration of wheelchairs could be developed to improve propulsion efficiency and reduce the risk of upper-limb injury. Guo et al12 again used power analysis to determine the mechanical cost of wheelchair propulsion and found pushrim diameter was an important factor in wheelchair propulsion mechanics. As such, power analysis may contribute to optimizing wheelchair configuration and design.

Rodgers et al13 examined joint power shifting under fatigue during wheelchair propulsion. Using a mix of nonusers and experienced manual wheelchair users propelling against a resisting ergometer, they showed that with fatigue, joint powers shift from the shoulder to the elbow and wrist. For some kinematic measures, experienced users compensated for fatigue differently than nonusers, for example, in stroke frequency, contact time, and pushrim and joint forces. They
concluded that power shifting may be a necessary strategy for maintaining constant power output under fatigue and may increase the risk of upper-limb injury. Their study showed that power analysis may contribute to understanding and optimizing propulsion strategies employed by wheelchair users and may inform us on the potential for injury.

In bringing a manual wheelchair up to speed, 2 stages typically exist, an acceleration stage, as the speed is increasing, and a steady-state stage, when the desired speed has been achieved. The relative contribution of either stage to upper-limb injury and pain is unknown. Koontz et al14 showed that forces and moments are higher when starting from a dead stop than when propelling at a steady-state speed, other factors being equal. This would support the view that although of relatively short duration, a rapid acceleration results in higher joint moments, velocities, and powers than does comfortable steady-state propulsion. High joint loading may result in acute injury.

Conversely, during comfortable steady-state propulsion over level smooth surfaces, the joint moments, velocities, and powers would tend to be smaller. Although acute injury may not occur, the long term, repetitive nature of the comfortable steady-state propulsion mode may invite chronic upper-limb injury and pain.15

In actuality, it may be the combined effects of both propulsive modes that lead to upper-limb injury and pain. Should this be the case, it is important to understand the factors involved in both stages of wheelchair propulsion. This should include an understanding of how joint power is redistributed under acceleration because shifting power from large to small joints may increase the risk of injury.16

The literature reveals that joint power analysis enhances our understanding of wheelchair propulsion. In the same way that joint reaction forces are likely responsible, in part, for upper-limb injuries,16 it is reasonable to expect that excessive joint power generation and absorption may be injurious. Excessive joint power may represent an essential, but in and of itself insufficient, contributor to injury. Other factors such as anatomy, limb orientation, and tissue tolerance, may play a mediating role. Whether power analysis proves a clinically useful technique to understand the role propulsion mechanics plays in upper-limb pain and injury remains in question. Nevertheless, joint power analysis may benefit the mechanistic investigation of the origin of upper-limb pain and injury, inform us on ergonomic treatment methods for wheelchair users, and generally lead to improved clinical insight. We hope the novel methods and findings described herein contribute to that goal.

This study’s objective was to compare the magnitude of joint power in the upper limb and its distribution between the shoulder, elbow, and wrist during maximal acceleration (MAC) relative to steady-state, self-selected speed (SSS) manual wheelchair propulsion. Power distribution among the upper-limb joints can be defined either in absolute or relative terms. Given the disparity between the 2 propulsion conditions, a relative measure of power distribution was deemed most appropriate. As used herein, power distribution refers to the fraction of power exhibited at each joint, normalized to the total upper-limb power, that is, the joint power fraction. A joint power fraction of .65 would equate to a concentric power contribution of 65% to the total power, for example. Stated mathematically:

\[
\text{Joint power fraction} = \frac{P_j}{P_T} \tag{1}
\]

where \(P_j\) is power at a specific upper-limb joint and \(P_T\) is the sum of power for all upper-limb joints (ie, total upper-limb power).

The following hypotheses were tested: (1) upper-limb joint power is higher during MAC trials relative to steady-state, SSS trials; and (2) the distribution of joint powers, when expressed on a fractional basis is unchanged when comparing MAC and steady-state, SSS propulsion trials.

Hypothesis 1, included for the sake of completeness and face validity, relies on the finding that forces and moments are higher during start-up.17 Also, from a mechanical energy perspective, power input is required to increase the kinetic energy of the system from rest to achieve steady-state propulsion. During steady-state propulsion, the power requirement is presumably less because it serves only to offset frictional losses. Given the lack of information on power distribution, hypothesis 2 was stated in the form of a null hypothesis without presuming a difference.

METHODS

The following comprised the inclusion and exclusion criteria for the study: complete or incomplete spinal cord injury (SCI) below T1, injury occurred at age 18 years or older, using a manual wheelchair as a primary means of mobility (>40h/wk), between 18 and 65 years of age at the time of the study, greater than 1 year since injury, no history of fractures or dislocations in upper limbs from which user has not fully recovered, no history of arm or shoulder pain as a result of a syrinx, not pregnant at the time of the study, and currently using a wheelchair equipped with quick release axles and 61-cm (24-in) wheels. The last criterion ensured compatibility with the instrumented pushrim (see below). All participants provided their informed consent for the study, which was reviewed and approved by the University of Washington Human Subjects Review Committee.

We collected 3-dimensional upper-limb kinematic data using a Qualisys® passive-marker motion capture system. Spherical retro-reflective markers were placed on the following landmarks and captured at 120Hz: temporomandibular joint (bilateral), C7 and T2 vertebrae, sternal notch, along with unilateral markers on the nondominant side: acromion, elbow lateral epicondyle, olecranon, radial, and ulnar styloids and third metacarpal joint. In addition, the hub and a spoke of the SmartWheel® on the nondominant side were marked. The protocol followed in this study is part of a larger study investigating the relationship between wheelchair propulsion biomechanics and upper-limb pain and injury. We studied the nondominant side because it is less likely to be injured from repetitive use other than wheelchair propulsion. Wheelchair propulsion biomechanics are generally similar between sides; Boninger et al17,18 found that left and right side kinetic and kinematic data are correlated.

The triaxial pushrim forces and moments applied on the nondominant side were captured using a SmartWheel at 240Hz (fig 1). This is the refined and commercialized version of the instrumented wheelchair pushrim developed and tested by Asato19 and VanSickle20 and colleagues. The pushrim was either anodized metal or vinyl-coated, according to the style in daily use by the participant. Only 1 SmartWheel was available; it was installed on the participant’s own wheelchair, on the user’s nondominant side. A standard wheel with a matching pushrim style was installed on the dominant side. Both wheels featured solid tire inserts irrespective of the tire style in daily use by the participant. Kinematic data capture was synchronized using the SmartWheel data capture system as a trigger source.

Participants propelled with their wheelchair secured, via an adjustable strap and rail, to the platform of a passive, dual-roller system (see fig 1); each roller acted independently and supported 1 rear wheelchair wheel. A visual feedback system
provided speed and directional information (to ensure straight-line propulsion), to the participant and investigators. This was accomplished by using a tachometer-generator for each roller and a laptop computer with analog-to-digital capability, running custom LabView® software.

We asked participants to propel under 2 conditions: (1) at their SSS and (2) under an MAC propulsion condition. This protocol has the effect of varying the level of effort between 2 extremes with no attempt to prescribe a fixed speed between subjects. We reasoned that holding speed constant at 2 extremes may not be truly reflective of the level of effort for different people; for some people the highest speed may not have been achievable because size, strength, experience level, etc, vary widely among users. In other words, a high fixed speed requiring great effort for one person may not represent a significant effort for another person and any conclusions regarding the distribution of joint power under these circumstances may be clouded.

Participants were permitted to acclimate to the roller system prior to data collection. Data were captured for a minimum of 20 and a maximum of 40 seconds for each condition. For the SSS trial, data capture was performed such that at a minimum, the last 20 seconds of the trial were at a steady-state pace in the judgment of an investigator monitoring the visual speed feedback system. For the maximal acceleration trial, the participant was instructed, via the computer’s display, to accelerate maximally (from a stop) for 5 seconds, brake and rest for 5 seconds, accelerate maximally again for 5 seconds, and then coast for the remainder of the trial. The MAC trial was an attempt to elicit maximum propulsive power output over a short time interval. The wheel speed during the MAC trial varied continuously over the trial and featured both the start-up speed from a dead stop and the participant’s maximum limiting speed over the acceleration period. Although the speed was variable, the intent of the protocol was that the participant’s propulsion effort was maximal.

After collection, we adapted the 240-Hz kinetic data to match the 120-Hz sampling frequency of the kinematic data by using every other frame. The synchronized 3-dimensional kinematic and kinetic data were forward and reverse filtered using a second-order digital Butterworth filter with a 5-Hz low-pass cutoff for each stage. Filter parameters are similar to those described by Winter10 and Cooper et al.21

A link-segment model of the upper-limb on the nondominant side was generated using anthropometric parameters published by Winter10 (based on subjects without physical disabilities). A separate anthropometric model was created for each participant based on their weight and measured anatomic landmarks along with the published parameters to compute the limbs’ inertial characteristics. Joint moments were calculated based on the segmental accelerations from kinematic measurements, external loads measured using the SmartWheel and inertial characteristics and segmental gravitational forces based on anthropometric data. The point of force application was approximated by the location of the third metacarpal joint marker. Although Cooper et al22 observed that the point of force application was in general not collocated with any anatomic landmark, Sabick et al23 concluded that kinematic methods of estimating point of force application were the most stable. Joint centers of rotation for the shoulder, elbow, and wrist were approximated by the location of the acromion, lateral epicondyle, and the mid-point of the radial- and ulnar-styloid markers, respectively. Segmental angular velocities were calculated and joint powers were computed as the scalar product of the joint moment and joint angular velocity vectors.

Because the contribution of each joint to the total upper-limb power output likely varied over the propulsion cycle, we used the time of peak exertion, given by the time of maximum total upper-limb power output. The algebraic sum of shoulder, elbow, and wrist powers gives the total net upper-limb power; it includes any negative power that may be present. In addition, the timing of the half-power points on either side of the peak total upper-limb power (ie, the points at which the total power is 50% of the peak) would reveal if the contribution of each joint changes prior, and subsequent to, the peak. Also, these 3 points (pre-peak, peak, post-peak) provided a conveniently defined boundary encompassing the time of greatest total propulsive effort. Note that the peak powers of each joint individually may occur at different times from the peak of the total upper-limb power. Investigating the joint power fraction under these circumstances leads to difficulties in interpretation and was therefore not performed.

To focus the analysis on the start-up acceleration strokes during the MAC trials, we evaluated the first 4 strokes of each of the 2 acceleration sets. Koontz et al24 showed that the start-up moments after the first 4 strokes are statistically similar. Because they are likely dissimilar, the joint power and power fraction for the first 4 strokes, for each of 2 successive attempts, were averaged to characterize the MAC trials. For the SSS trials, the last 8 full propulsion strokes were evaluated. Thus, for each condition, 8 strokes were averaged to estimate the power outputs for the shoulder, elbow, and wrist at the time of peak total and half-power points. Also, the power outputs for the shoulder, elbow, and wrist at the time of peak and half-

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**Fig 1. Reflective markers applied to volunteer, SmartWheel, and passive dual-roller system.**
Power points were used to determine the joint power fraction (see equation 1). This served as a relative measure of each joint’s contribution to the total power output.

To evaluate the statistical significance of any data trends, we applied the Wilcoxon signed-rank test for paired data\textsuperscript{24} using SPSS.\textsuperscript{24} The modest sample size dictated the use of a nonparametric test, because power data normality could not be ensured. Pairing the data effectively compared changes within each subject. Significance levels of $P$ less than .01 and $P$ less than .05 were applied. A 1-tailed test was applied to test hypothesis 1 and a 2-tailed test was applied to test hypothesis 2. In addition, to test the significance of power differences within each condition and between the upper-limb joints, the Friedman test\textsuperscript{25} was used because the data across joints are related.

**RESULTS**

The volunteer participant sample consisted of 4 women and 9 men, ranging from 32 to 61 years of age (mean, 46.5±10.3y). Injury levels ranged from T3 to L1, with time since injury ranging from 2 to 33 years (mean, 15±9.4y). Participant weight ranged from 49 to 88kg (mean, 65.4±13.1kg) and height ranged from 1.5 to 1.93m (mean, 1.73±0.12m). All 13 participants were right-hand dominant and were therefore instrumented on the left side.

For reference purposes, sample power versus time curves (for a person) for each joint during an SSS trial are given in figures 2A through 2C. The corresponding power versus time curves for an MAC trial are shown in figures 3A through 3C. These graphs display several cycles of propulsion and include a step function that indicates when a propulsive moment exceeding $0.25\text{Nm}$ was applied to the SmartWheel, effectively indicating when contact on the pushrim occurred. In addition, the total upper-limb power and the individual joint power are plotted. Positive power values denote concentric contraction and negative values denote eccentric contraction.

Wheelchair speed during the SSS condition averaged ± standard deviation $1.29\pm0.41\text{m/s}$. The peak wheelchair speed attained on the fourth stroke of the MAC condition averaged $2.07\pm0.43\text{m/s}$. Descriptive statistics for the joint-specific power output at the time of peak total upper-limb power under SSS and MAC conditions are given in table 1. Wilcoxon signed-rank statistical testing revealed that, at a $P$ less than .01 significance level, median shoulder power was greater under MAC conditions ($102\text{W}$) than under SSS conditions ($38\text{W}$). At the time of peak total upper-extremity power, power at the elbow and wrist did not differ significantly for the propulsion conditions. At the time of peak total upper-limb power, median joint power was greater ($P<.001$) for the shoulder (SSS, 37.7W; MAC, 101.5W) than for the elbow (SSS, 3.2W; MAC, −1.8W) and wrist (SSS, −0.3W; MAC, −5.0W), for both SSS and MAC conditions, according to the Friedman test.

The descriptive statistics for joint power fractions under SSS and MAC conditions are displayed in table 2. Wilcoxon signed-rank statistical testing revealed that, at a $P$ less than .05 significance level, median shoulder power was greater ($P<.001$) for the shoulder (SSS, 37.7W; MAC, 101.5W) than for the elbow (SSS, 3.2W; MAC, −1.8W) and wrist (SSS, −0.3W; MAC, −5.0W), for both SSS and MAC conditions, according to the Friedman test.
signed-rank statistical testing revealed that median peak shoulder power fraction was higher ($P<.01$) for MAC versus SSS (1.055 vs .870). Median pre-peak shoulder power fraction was also higher ($P<.01$) for MAC versus SSS (.722 vs .489). Median peak elbow power fraction was lower ($P<.05$) for MAC versus SSS (-.12 vs .146). Median pre-peak elbow power fraction was also lower ($P<.05$) for MAC versus SSS (.369 vs .495). Median peak wrist power fraction was lower ($P<.05$) for MAC versus SSS (-.058 vs -.010). Median pre-peak wrist, and post-peak comparisons for all 3 joints, did not differ significantly between SSS and MAC conditions. Peak power fraction was greater ($P<.001$) for the shoulder than for the elbow and wrist, for both SSS and MAC conditions, according to the Friedman test.

Graphical comparisons of the joint power fractions at the time of peak net upper-limb power output for the shoulder, elbow, and wrist for SSS and MAC conditions are shown in figure 4. Likewise, pre-peak graphical comparisons are shown in figure 5.

**DISCUSSION**

Only at the shoulder did the data support hypothesis 1 that peak joint power is higher during acceleration than during steady-state propulsion; the elbow and wrist joints showed no consistent change (see table 1). Examination of the sample joint power data (see figs 2A–C, 3A–C) reveals an asymmetry of the total upper-limb joint power, with greater concentric power (positive) relative to eccentric power (negative). Furthermore, the concentric power during the propulsion phase exceeds the concentric power during the recovery phase. Both findings are consistent with the participant working against the external load imposed by the passive dual-roller system due to rolling friction, similar to what might be expected during overground propulsion.

During the propulsion phase, the shoulder dominates the power output behavior between the half-power points, relative to the other joints, for both propulsion conditions. This was confirmed statistically by the Friedman test to compare power levels between joints at the time of peak total upper-limb power output. The finding that the shoulder power was greater during

**Table 1: Joint Powers at Time of Peak Upper-Extremity Power During SSS and MAC Conditions**

<table>
<thead>
<tr>
<th>Joint</th>
<th>SSS Median</th>
<th>MAC Median</th>
<th>SSS IQR</th>
<th>MAC IQR</th>
<th>SSS Min</th>
<th>MAC Min</th>
<th>SSS Max</th>
<th>MAC Max</th>
</tr>
</thead>
<tbody>
<tr>
<td>Shoulder</td>
<td>37.7*</td>
<td>101.5*</td>
<td>22.9</td>
<td>74.6</td>
<td>11.0</td>
<td>26.9</td>
<td>104.7</td>
<td>178.8</td>
</tr>
<tr>
<td>Elbow</td>
<td>3.2</td>
<td>-1.8</td>
<td>6.3</td>
<td>8.7</td>
<td>-3.7</td>
<td>-13.6</td>
<td>10.9</td>
<td>23.8</td>
</tr>
<tr>
<td>Wrist</td>
<td>-0.3</td>
<td>-5.0</td>
<td>8.0</td>
<td>4.6</td>
<td>-8.1</td>
<td>-33.2</td>
<td>6.7</td>
<td>23.9</td>
</tr>
</tbody>
</table>

*P<.01 (Wilcoxon signed-rank test).

NOTE. Values are median, interquartile range (IQR), minimum (Min), and maximum (Max). Values in watts. N=13.

Fig 3. (A) Power versus time for the shoulder joint (solid) and total upper-limb power (dashed) under acceleration. Step function indicates when a propulsion moment is applied to the SmartWheel. (B) Power versus time for the elbow joint (solid) and total upper-limb power (dashed) under acceleration. Step function indicates when a propulsion moment is applied to the SmartWheel. (C) Power versus time for the wrist joint (solid) and total upper-limb power (dashed) under acceleration. Step function indicates when a propulsion moment is applied to the SmartWheel.
the MAC condition reflects the dominant role of the shoulder in generating concentric power under circumstances of maximal propulsive effort. By contrast, the elbow and wrist joints are minor contributors to concentric power output under both propulsive conditions and they show no significant difference when comparing SSS to MAC propulsion. The elbow power curves indicate a tendency for the elbow to act in an eccentric capacity during the later half of the propulsion stroke.

During the recovery phase, the shoulder again dominates power generation and absorption. This is likely the result of the

## Table 2: Joint Power Fractions at Time of Peak, Pre-Peak, and Post-Peak Upper-Extremity Power During the SSS and MAC Conditions

<table>
<thead>
<tr>
<th>Value</th>
<th>Peak Shoulder Power Fraction</th>
<th>Peak Elbow Power Fraction</th>
<th>Peak Wrist Power Fraction</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>SSS</td>
<td>MAC</td>
<td>SSS</td>
</tr>
<tr>
<td>Median</td>
<td>0.870*</td>
<td>1.055*</td>
<td>0.146*</td>
</tr>
<tr>
<td>IQR</td>
<td>0.252</td>
<td>0.110</td>
<td>0.206</td>
</tr>
<tr>
<td>Min</td>
<td>0.68</td>
<td>0.64</td>
<td>−0.04</td>
</tr>
<tr>
<td>Max</td>
<td>1.13</td>
<td>1.15</td>
<td>0.29</td>
</tr>
<tr>
<td>Pre-Peak</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Median</td>
<td>0.489*</td>
<td>0.722*</td>
<td>0.495†</td>
</tr>
<tr>
<td>IQR</td>
<td>0.442</td>
<td>0.351</td>
<td>0.501</td>
</tr>
<tr>
<td>Min</td>
<td>0.05</td>
<td>0.42</td>
<td>0.03</td>
</tr>
<tr>
<td>Max</td>
<td>0.93</td>
<td>1.08</td>
<td>0.93</td>
</tr>
<tr>
<td>Post-Peak</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Median</td>
<td>1.158</td>
<td>1.013</td>
<td>−0.078</td>
</tr>
<tr>
<td>IQR</td>
<td>0.713</td>
<td>0.562</td>
<td>0.636</td>
</tr>
<tr>
<td>Min</td>
<td>0.26</td>
<td>0.24</td>
<td>−0.51</td>
</tr>
<tr>
<td>Max</td>
<td>1.63</td>
<td>1.88</td>
<td>0.77</td>
</tr>
</tbody>
</table>

NOTE. Values are median, IQR, minimum, and maximum. N=13.

*P<.01; †P<.05 (Wilcoxon signed-rank test).

![Fig 4](https://example.com/fig4.png)

**Fig 4.** Joint power fractions (median and quartiles) at time of upper-limb peak power for SSS and MAC trials.

![Fig 5](https://example.com/fig5.png)

**Fig 5.** Joint power fractions (median and quartiles) at time of upper-limb pre-peak power for SSS and MAC trials.
larger inertial load applied to the shoulder in accelerating and decelerating the upper limb in comparison to the elbow and wrist, as well as a larger angular velocity likely to be present at the shoulder. In the MAC trial example, the elbow power is more pronounced during recovery than in the SSS trial, possibly a reflection of a more vigorous retraction of the arm in the maximal acceleration trial.

Comparisons of the SSS joint power data (typified by figs 2A–C) with power flow data of Guo et al11 reveals similar outcomes, despite differences in the methodology between the 2 studies. The Guo study computed segmental power flows (upper arm, forearm, hand), whereas the current study computed joint powers (shoulder, elbow, wrist). Also, the Guo study relied on nonwheelchair users propelling over level ground, rather than on experienced users with SCI, propelling on a passive dual-roller system. Nevertheless, it may be revealing to compare the peak upper-limb power during propulsion for each method. The mean total upper-limb power flow, obtained by Guo, revealed an initial positive peak of about 22W during early propulsion followed by a nearly continuous decrease in power flow until the onset of the recovery phase. The current study found a somewhat larger mean peak upper-limb power (40.6W) during the SSS propulsion phase, as determined by the sum of individual mean joint powers in table 1. This may be reflective of differences due to propelling on a passive, dual-roller system by experienced wheelchair users in contrast to nonusers rolling over ground. The type of tire used by Guo was not specified and differences in tires may also play a role.

Shoulder joint power fraction was greater for the MAC condition relative to the SSS condition, whereas elbow and wrist power fractions were less, contrary to hypothesis 2, which posited no change in power fraction (see table 2, figs 4, 5). Rodgers et al13 observed a shift in power from the shoulder to the elbow and wrist under conditions of fatigue. In their mix of both wheelchair users and nonusers working against an ergometer, they found that the shoulder provided 61%, the elbow 15%, and the wrist 24% of the total power during nonfatiguing propulsion. Under fatigue, the joint power fraction was 50%, 19%, and 31% for the shoulder, elbow, and wrist, respectively. The wrist power fraction observed in the present study appeared to be considerably lower (see table 2, figs 4, 5) than that observed by Rodgers. It is possible that differences in the load applied by their ergometer relative to our roller system may explain the wrist power difference.

The power fraction at the shoulder was greater than at the elbow and wrist for both SSS and MAC conditions, as determined by the Friedman test comparing power fractions across joints. Comparison of joint power fractions between the SSS and MAC conditions revealed that the shoulder joint generated a significantly larger fraction of the power during the MAC trials. This behavior parallels that of the shoulder power and reinforces the finding that the shoulder dominates during the propulsion phase, especially during acceleration. This was in contradistinction to the elbow and wrist, which exhibited a numerically lower peak power fraction under the MAC condition. Contrary to the shoulder and elbow, the wrist tended to show power absorption under both conditions, more so under the MAC condition, as reflected by a negative value of the power fraction.

Pre-peak power fractions for the shoulder and elbow followed the same pattern as the peak power fractions, but were statistically weaker. No significant difference in pre-peak power fraction was observed for the wrist. The post-peak power fraction did not show a systematic change for any joint. This supports the rationale that the pre- and post-peak regions effectively bracket the behavior of interest, namely, the time of peak propulsive effort when systematic differences between conditions and joints may be at their greatest.

The shoulder power fraction data (see figs 4, 5) revealed that in some cases the power fraction exceeded 1.0. This implied that a power absorbing, eccentric contraction occurred at other joints. This was confirmed because the elbow and wrist exhibited a concurrent negative power fraction in those cases. A study by van der Linden et al26 showed that negative power could be generated under certain conditions of force and joint orientations in conjunction with the kinematic constraints imposed by the trajectory of the hand. At the time of peak total upper-limb power output, the shoulder joint power fraction exceeded 1.0, and the elbow and wrist power fractions were negative, in more cases under MAC than under SSS conditions. In effect it appears that the shoulder concentric contraction was working against the wrist and elbow eccentric contractions. Thus, it can be argued that for many of the participants, propulsion was less effective under maximal acceleration than it could have been.

There appeared to be a shift in power toward the shoulder and away from the elbow during the MAC trials. This power shift may expose the shoulder to greater risk of injury and/or pain during acceleration. Three factors present under MAC conditions may contribute to greater injury risk, namely, higher power and power fraction at the shoulder and the higher likelihood of exhibiting simultaneous eccentric and concentric contractions over the upper limb. The higher level, and fraction, of power being shifted to the shoulder during acceleration implies that the muscles at the shoulder have the excess power generating capacity to perform the MAC task that the muscles of the more distal joints may lack. It may also imply that the shoulder is at greater risk for acute injury during maximal acceleration than the other joints. Exhibiting excess concentric contraction at the shoulder while simultaneously exhibiting eccentric contractions at the distal joints may extend the risk of injury and and/or pain in those persons using this propulsion strategy. Correlations of a user’s injury and pain status with their joint power fraction pattern under acceleration may facilitate investigation of shoulder pain and may provide direction for wheelchair user training to minimize pain and injury. In relation to chronic injury, it may be revealing to examine the joint power fraction patterns during SSS conditions, for persons who have not developed chronic pain and injury, to serve as a training model or as an indication of when an intervention may be warranted.

The study methodology could not account for co-contractions of the upper-limb joints and thus permitted only net power calculations. The alternatives invoking electromyography-based muscle modeling to estimate co-contraction forces introduce other assumptions and limitations, however. In addition, the anthropometric parameters used in modeling were based on published values for persons without physical disability; however, the statistical testing relied on pairing of data within a subject (ie, differential changes); this would presumably minimize the associated error.

To leverage these findings and methods to improve the clinical care of manual wheelchair users, future research should investigate: (1) the association between upper-limb joint power and the development of upper-extremity pain or injury (eg, Are persons with greater shoulder power fraction at a higher risk for injury?) and (2) the sensitivity of this metric to detect differences in wheelchair modifications (eg, pushrim type, seat/back adjustments, postural supports), training interventions (eg, Do power fractions change with training?), and user characteristics (eg, How does shoulder strength influence the joint’s power fraction?).
CONCLUSIONS

This study’s objective was to compare the magnitude of joint power in the upper limb and its distribution between the shoulder, elbow, and wrist during maximal acceleration relative to steady-state, self-selected speed manual wheelchair propulsion. At the time of peak total upper-limb power output, the shoulder played the dominant role, relative to the elbow and wrist, for both propulsion conditions. Also, shoulder joint power was greater during maximum acceleration, relative to self-selected steady-state propulsion, whereas the wrist and elbow joints showed no consistent change. Shoulder joint power fraction was greater during maximum acceleration relative to self-selected steady-state propulsion, but elbow and wrist power fractions tended to be smaller. Some participants showed simultaneous concentric contractions at the shoulder and eccentric contractions at the elbow and wrist during the maximal acceleration trials, indicating reduced propulsion effectiveness. Because they are higher, joint powers and power fractions present under maximal acceleration conditions may predispose manual wheelchair users to injury, especially at the shoulder. Further study is required to better understand how joint powers and power fractions relate to the development of upper-limb pain or injury in manual wheelchair users and how these findings may be applied clinically.

References


Suppliers

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b. Three Rivers Holdings LLC, 1826 W Broadway Rd, Ste 43, Mesa, AZ 85202.
c. National Instruments Corp, 11500 N Mopac Expwy, Austin, TX 78759-3504.
d. SPSS Inc, 233 S Wacker Dr, 11th Fl, Chicago, IL 60606-6307.

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