# A comparison of glenohumeral joint kinematics and muscle activation during standard and geared manual wheelchair mobility

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## ARTICLE INFO

## ABSTRACT

The high prevalence of upper extremity joint injuries among manual wheelchair users is largely attributed to the high repetitive loading during propulsion. The purpose of this study was to evaluate the effects of using geared wheels for manual wheelchair mobility on shoulder joint biomechanics. Fourteen able-bodied participants performed overground propulsion and ramp ascension using standard and geared manual wheelchair wheels. Spatial temporal parameters, glenohumeral joint kinematics, and shoulder muscle activity were quantified. Findings indicated that regardless of the level of slope, the propulsion speed and stroke distance decreased significantly ($p \ll 0.001$), and the stroke frequency increased significantly ($p \leq 0.025$) during geared manual wheelchair propulsion. The glenohumeral joint ranges of motion in the coronal plane ($p \leq 0.005$) and peak joint angles in the coronal ($p \leq 0.023$) and transverse ($p \leq 0.012$) planes were significantly different between standard and geared wheels usage. Shoulder muscle activity was substantially less using the geared wheels with significant findings in the pectoralis major (level floor, $p \leq 0.008$) and infraspinatus ($p \leq 0.014$) peak muscle activity, and the anterior deltoid ($p \leq 0.014$) and pectoralis major ($p \leq 0.015$) integrated muscle activity. However, the shoulder flexor normalized integrated muscle activity (muscle activity per stroke distance) was not different between the wheels.

## 1. Introduction

An estimated 3.7 million people in the United States use a wheelchair [1]. Approximately 90% of which use hand-rim propelled manual wheelchairs (standard manual wheelchairs) [2]. Manual wheelchairs, which are the most commonly prescribed type of wheelchair, are relatively inexpensive, highly maneuverable on flat surfaces, lightweight, and convenient to transport [3]. However, manual wheelchair mobility has been described as a low efficient and physically straining form of mobility that places manual wheelchair users at high risk of repetitive strain injuries, primarily in the shoulder and wrist [4–6]. This often leads to reduced independence, function, and quality of life [7].

The geared manual wheelchair wheel is an alternative propulsion mechanism that may reduce the biomechanical demands of the upper extremity during propulsion, while maximizing function. Similar to a multi-speed bicycle, geared wheels allow users to choose the option of wheeling in a low gear. Despite this new rehabilitation technology, there is limited scientific evidence surrounding geared manual wheelchair mobility. Finley and Rodgers studied the effects of geared wheels (Magicwheels ®, gear ratio of 2:1) on shoulder pain and function in a longitudinal study with full time manual wheelchair users [8]. Findings in this study indicated the potential for geared manual wheels to reduce shoulder pain. Furthermore, Howarth and colleagues investigated the effects of using geared manual wheelchair wheels (Magicwheels ®, gear ratio of 2:1) on the shoulder joint kinematics in the sagittal plane of motion and muscular demand of able-bodied individuals during ramp ascent [9,10]. They found a significant decrease in peak muscle activity of the shoulder flexors and a significant increase in the integrated muscle activity during ramp ascent [9]. They also reported reduced demands of the abdominal muscles [10], which indicate the potential benefits of geared manual wheelchairs for individuals with compromised activity capacity. The significant increase in the integrated muscle activity during ramp ascent was mainly due to the notable increase in the ramp ascent duration during the geared wheel condition in comparison to the standard wheel condition [9]. This is primarily attributed to the relatively

large gear reduction (2:1) in Magicwheels ® geared wheels. Preliminary results of studies conducted by our group demonstrate that using geared manual wheelchair wheels (IntelliWheels Easy Push[TM], gear ratio of 1.6:1) may be beneficial for demanding tasks, such as ramp ascent and propulsion on carpeted floors [11,12].

To evaluate the biomechanical effects of novel propulsion mechanisms, such as geared wheels, it is warranted to first investigate able-bodied, non-wheelchair users [13,14]. Able-bodied subjects are equally well un-trained and are physically homogenous [13]. To better understand the upper extremity biomechanics during geared manual wheelchair mobility, we evaluated and compared the spatiotemporal parameters, three-dimensional glenohumeral joint kinematics, and shoulder muscle activity during propulsion on level floor and on a ramp using both standard and geared manual wheelchair wheels in able-bodied subjects. We hypothesize that using the geared wheels (gear ratio of 1.6:1) during both overground propulsion and ramp ascension will significantly change the spatiotemporal parameters and glenohumeral joint kinematics and decrease the shoulder muscle activity.

## 2. Methods

### 2.1. Subjects

Fourteen individuals, seven females and seven males, with an average age of $22.5 \pm 3.4$ years (age range from 18 to 29 years) participated in this study. The average weight was $76.1 \pm 18.3$ kg and the average height was $1.76 \pm 0.10$ m. All participants were able-bodied and had no prior wheelchair experience. This study was approved by the University of Wisconsin-Milwaukee Institutional Review Board and consent was obtained from participants.

### 2.2. Experimental protocol

A Breezy® Ultra 4 manual wheelchair (Sunrise Medical LLC., Fresno, CA, US) was used with standard wheels and Easy Push (IntelliWheels, Inc., IL) geared wheels for all activities (Table 1). The hand-rim of the Easy Push wheel is connected to the wheel via a planetary gear mechanism with the gear ratio of 1.6:1. The weight of the wheelchair with the geared wheels was 22.8 kg and with standard wheels was 19.2 kg. The width of the wheelchair seat was 0.45 m, and the height of the backrest was 0.4 m. The inertial and geometric specifications of the wheels are listed in Table 1.

Data collection was conducted at the University of Wisconsin-Milwaukee Mobility Lab. Subject-specific measurement were obtained for all participants. The wheelchair's seat height was adjusted close to optimum height by satisfying the following criteria, (1) the elbow was flexed at approximately $120°$ when the hands were on the highest point of hand-rims; and, (2) the footrests clear the ground by five centimeters [15]. The distance between the seat and footrest was adjusted as close as possible to the shank length, therefore the hip and knee joints were approximately $90°$ flexed. Participants were provided with an acclimation period to become familiar with standard and geared wheels over level ground and on a ramp. Participants propelled the manual wheelchair with each type of wheels in a random order, along a 10 m tiled, level floor, and up a 2.5 m ramp with a $4.8°$ slope (Fig. 1; this was below the maximum allowable slope of $5°$ recommended by American

with Disability Act Accessibility (ADA) Guidelines [16]). The conditions were randomized and there were three to five trials for each condition. Participants used a self-selected speed and propulsion pattern for all tasks. Each ramp ascent trial began on the level tile floor with the participant positioned one meter from the ramp base [9].

### 2.3. Data collection

#### 2.3.1. Kinematic data collection

Participants were instrumented with our established upper extremity model consisting of 27 retro-reflective markers on anatomical landmarks of the thorax, clavicles, scapulae, upper arms, forearms, and hands. (Fig. 1) [17]. Four markers were placed on the backrest and one marker on the dominant side wheel axle. Motion data were collected at 120 Hz using a 15-camera Vicon T-series motion capture system (Vicon Motion Systems, Oxford, UK).

#### 2.3.2. Surface electromyography data collection

Shoulder flexion is the dominant movement during the push phase of manual wheelchair propulsion [18,19]. Thus, muscle activity of the primary shoulder flexors (anterior deltoid and pectoralis major) and one of the rotator cuff muscles (infraspinatus) were recorded using surface electromyography (EMG) [20,21]. Delsys Trigno wireless surface electrodes (Delsys, Inc., Natick, MA) with built-in bandpass filter of 20–450 Hz were used to record EMG data of the anterior deltoid, clavicular head of the pectoralis major, and infraspinatus of the dominant arm.

Prior to EMG electrode placement, the skin surface underlying each electrode was scrubbed and cleaned with an alcohol preparatory pad. Electrodes were placed over the anterior deltoid (anterior aspect of the arm, approximately 4 cm below the clavicle), clavicular head of the pectoralis major (approximately 2 cm below the clavicle, midway between the underarm and suprasternal notch), and infraspinatus (parallel and approximately 4 cm below the spine of the scapula) [21]. The Vicon motion capture system was digitally synchronized with the Delsys EMG system. The EMG signals were acquired at a sampling frequency of 2040 Hz using Vicon Nexus software (Vicon Motion Systems, Oxford, UK).

Before the experimental trials, maximum voluntary isometric contractions (MVIC) were performed for each muscle using a BTE PrimusRS (BTE, Hanover, MD). During the MVIC trials, the participant sat on the BTE chair with the trunk in an upright position and trunk and legs secured to the chair with Velcro straps. Three MVIC trials were performed for each muscle, with contraction durations of three seconds. Verbal encouragement was provided during each trial by the investigator. The MVIC of the anterior deltoid was tested with the upper arm next to the trunk with $45°$ of shoulder flexion while resistance was applied posteriorly against the distal end of humerus during isometric shoulder flexion. The pectoralis major muscle was tested while the upper arm was horizontal and midway between anterior and lateral directions, and the forearm was flexed $90°$ at the elbow. Resistance was applied laterally against the elbow during isometric shoulder horizontal flexion [22]. The MVIC of the infraspinatus was tested with the upper arm next to the trunk with the forearm horizontal and flexed $90°$ at the elbow. Resistance was applied medially against the wrist during isometric shoulder external rotation. The peak EMG of each trial was calculated and the mean peak EMG magnitude of three trials was used as the MVIC of each muscle.
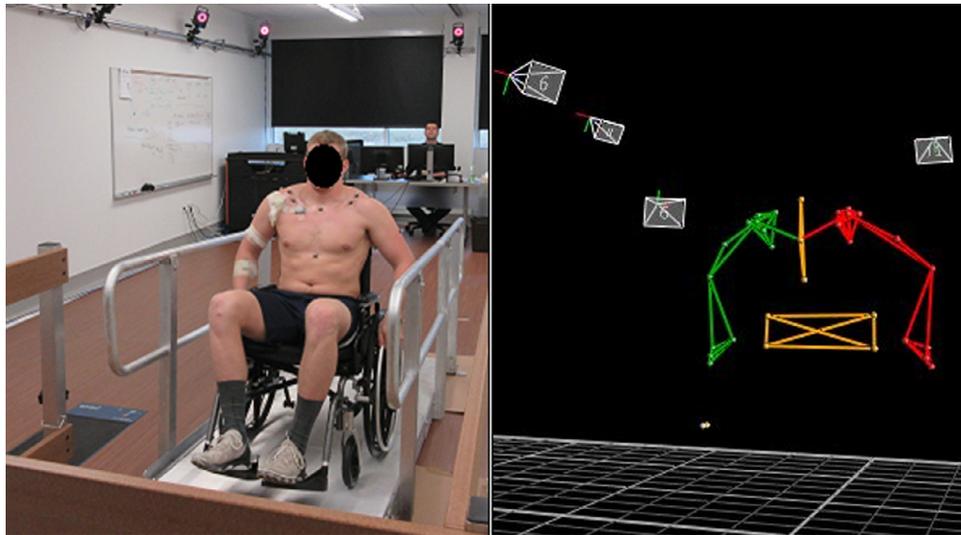
### 2.4. Data processing

#### 2.4.1. Kinematics

To identify the push and recovery phases of each propulsion stroke, sagittal plane kinematics of the third metacarpal marker

**Table 1**
Specifications of the standard and geared wheels.

| Wheel type | Weight (kg) | Wheel width (cm) | Wheel size (cm) |
|---|---|---|---|
| Standard | 2.6 | 6.0 | 61.0 |
| Geared | 4.4 | 7.7 | 61.0 |

**Fig. 1.** A subject ascending a ramp using a manual wheelchair, instrumented with motion capture markers and surface EMG electrodes (left); and corresponding Vicon rendering of the custom upper extremity biomechanical model (right).

was used. The three critical points in time were the (1) initial hand contact, (2) hand release, and (3) second hand contact [23]. These were visually inspected and identified using the method of Chow et al. [23]. The push phase was defined as the initial hand contact to hand release, and a complete stroke cycle was defined from the initial hand contact to the second hand contact. Temporal parameters examined in this study were stroke time, push phase (expressed as percentage of stroke time), and stroke frequency (one second divided by the stroke time). Stroke distance was computed as the distance between the position of the wheel center in the sagittal plane at the time of the initial and second hand contacts. Stroke speed was calculated by dividing the stroke distance by the stroke time. The normalized stroke frequency was obtained by dividing the stroke frequency by the stroke distance. In most cases the fourth stroke cycle (steady state condition) was analyzed during overground propulsion [24]. If that was not possible the third stroke cycle (semi-steady state) was used for analysis. The first full stroke cycle that occurred after the wheelchair castors passed the midpoint of the ramp was used for analysis of ramp ascent [23]. For each subject the average profile over three stroke cycles (one for each trial) was used in statistical analysis.

Kinematic data were processed using Vicon Nexus software. The marker trajectories were filtered using a Woltring filter with a mean squared error of 20 [25]. Body segment parameters were estimated using equations developed for an adult population [26]. Our inverse dynamics model was used to calculate the three-dimensional (3-D) movements of the glenohumeral joint [17]. The kinematic metrics of interest, including 3-D ranges of motion (ROMs) and joint angles of the glenohumeral joint, were calculated using our upper extremity inverse dynamics model developed in MATLAB software (MathWorks, Inc.) [17].

### 2.4.2. Electromyography

Methods previously described by Beres and Harkema were used to define the EMG burst duration after rectification of the EMG signals [27]. The detection threshold was defined as the mean of the baseline plus four standard deviations. The onset of an EMG burst was then defined as the time when the signal amplitude rose above the detection threshold and remained above it for at least 30 ms [27]. The end of an EMG burst was defined as the time when the signal amplitude fell below the detection threshold and remained below it for at least 50 ms [27]. Visual examination was also performed.

The raw EMG signals were full wave rectified and processed using a root mean square (RMS) algorithm with a time averaging period of 25 ms. The RMS envelope of the EMG data recorded for each muscle during wheelchair propulsion trials were normalized as a percentage of the MVIC obtained from each respective muscle from the mean RMS envelope of the MVIC trials. The normalized EMG signals were used to determine the peak EMG during the burst duration and the integrated EMG over each stroke cycle using MATLAB software. The normalized integrated muscle activity was calculated by dividing the integrated muscle activity by the distance traveled per stroke cycle.

### 2.5. Data analysis

The subject means and standard deviations of the spatiotemporal parameters, glenohumeral joint kinematics, and peak and integrated muscle activity were computed for each wheel type (standard and geared) and slope (level floor and ramp) for three trials. A two-way analysis of variance (2 wheel types × 2 slopes) with repeated measures was used for statistical analysis. If there was no significant interaction between the factors (wheel type and slope), the main effects of the wheel type factor (regardless of the level of slope) were investigated. When there was a significant interaction between factors, simple effects of the wheel type factor within each level of slope were examined using a paired-samples $t$-test. All statistical analyses were completed with SPSS software (IBM Corporation) using a general linear model with repeated measures and paired sample t-tests (significance level = 0.05).

## 3. Results

### 3.1. Spatiotemporal parameters

The statistical analysis results indicated that there was a significant interaction between the wheel type and slope for the stroke distance ($F = 6.65$, $p = 0.023$), speed ($F = 20.26$, $p = 0.001$) and normalized stroke frequency ($F = 16.79$, $p = 0.001$; Table 2). Analysis of the simple effects indicated that when using geared wheels on level floor, there was a 21% decrease in the stroke distance ($t = 7.30$, $p \ll 0.001$) and a 20% decrease in propulsion speed ($t = 9.16$, $p \ll 0.001$), compared to standard wheel use. Additionally, there was a 26% decrease in stroke distance ($t = 7.28$, $p \ll 0.001$)

**Table 2**
Group mean wheelchair propulsion spatiotemporal parameters (mean ± 1 SD) for each experimental condition (standard and geared wheels on a level floor and on a ramp).

| Metric | Terrain | Standard wheels Mean ± SD | Geared wheels Mean ± SD | p |
|---|---|---|---|---|
| Speed (m/s)[#] | Level floor | 1.10 ± 0.16** | 0.88 ± 0.14** | ≪0.001 |
| | Ramp | 0.61 ± 0.12** | 0.50 ± 0.08** | ≪0.001 |
| Stroke distance (m)[#] | Level floor | 1.20 ± 0.23** | 0.94 ± 0.22** | ≪0.001 |
| | Ramp | 0.68 ± 0.14** | 0.51 ± 0.09** | ≪0.001 |
| Stroke cycle time (s)[*] | Level floor | 1.10 ± 0.19 | 1.07 ± 0.18 | 0.038 |
| | Ramp | 1.13 ± 0.21 | 1.02 ± 0.13 | |
| Push phase (%)[*] | Level floor | 47.79 ± 5.28 | 41.07 ± 4.36 | ≪0.001 |
| | Ramp | 62.51 ± 3.92 | 55.55 ± 3.41 | |
| Stroke cycle frequency (Hz)[*] | Level floor | 0.93 ± 0.14 | 0.95 ± 0.14 | 0.025 |
| | Ramp | 0.91 ± 0.15 | 1.00 ± 0.12 | |
| Normalized stroke cycle frequency (1/ms)[#] | Level floor | 0.81 ± 0.24** | 1.09 ± 0.35** | ≪0.001 |
| | Ramp | 1.42 ± 0.48** | 2.05 ± 0.54** | ≪0.001 |

[#] Statistically significant interaction between wheel type and slope, $p < 0.05$.
[*] Statistically significant difference, $p < 0.05$ (wheel type main effect).
[**] Statistically significant difference, $p < 0.05$ (wheel type simple effects for level floor and ramp).
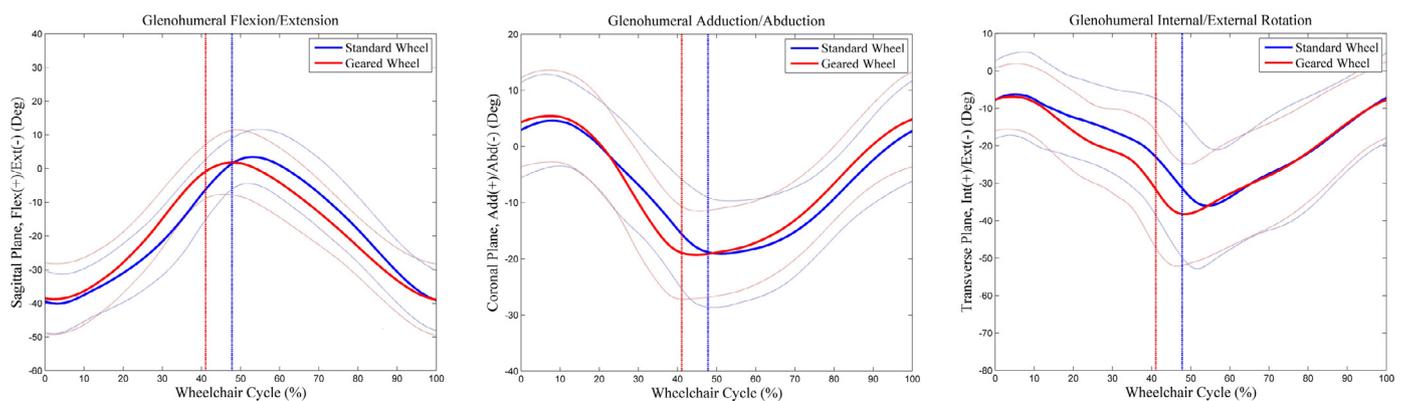


**Fig. 2.** Glenohumeral joint kinematics in sagittal, coronal and transverse planes of motion. Group mean profiles (solid line) +/- one standard deviation (dotted line) of 14 subjects (dominant side) for the standard and geared manual wheelchairs during propulsion on level floor are depicted. The vertical dash-dot lines indicate the transition from the end of the push phase to the start of the recovery phase for each wheel type.
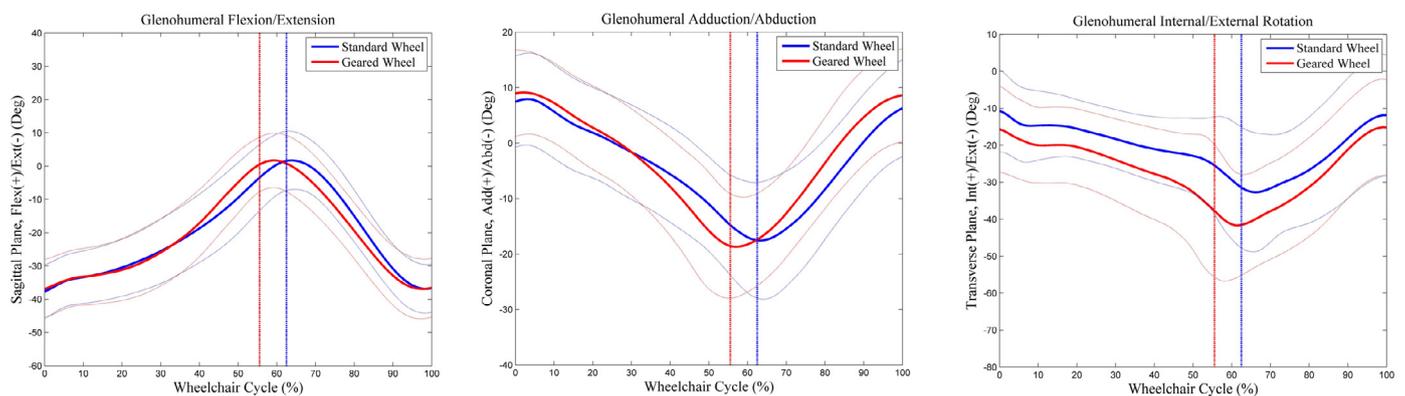


**Fig. 3.** Glenohumeral joint kinematics in sagittal, coronal and transverse planes of motion. Group mean profiles (solid line) +/- one standard deviation (dotted line) of 14 subjects (dominant side) for the standard and geared manual wheelchairs during propulsion on a ramp are depicted. The vertical dash-dot lines indicate the transition from the end of the push phase to the start of the recovery phase for each wheel type.

and an 18% decrease in propulsion speed ($t = 5.54$, $p \ll 0.001$), during ramp ascent with geared wheels, compared to standard wheels. Using geared wheels increased the normalized stroke frequency by 33% ($t = -4.68$, $p \ll 0.001$) on level floor and by 43% ($t = -6.93$, $p \ll 0.001$) on the ramp (Table 2).

The interaction between the wheel type and slope was not statistically significant for the stroke cycle time, frequency and push phase. Analysis of the main effects indicated that regardless of the slope level, geared manual wheelchair propulsion decreased the stroke cycle time by 6% ($F = 5.35$, $p = 0.038$) and push phase by 12% ($F = 128.25$, $p \ll 0.001$), and increased the stroke cycle frequency by 6% ($F = 6.36$, $p = 0.025$) (Table 2).

### 3.2. Kinematics

Mean glenohumeral joint kinematics in each plane of motion for all subjects for each wheel type and slope level are depicted (Figs. 2 and 3). The glenohumeral joint trajectories in each plane

**Table 3**

Group mean glenohumeral joint ranges of motion (mean $\pm 1$ SD) during the standard and geared manual wheelchair propulsion on level floor and on a ramp (degrees).

| Plane | Terrain | Standard wheels Mean ± SD | Geared wheels Mean ± SD | p |
|---|---|---|---|---|
| Sagittal | Level floor | 43.19 ± 9.42 | 44.67 ± 9.25 | 0.727 |
| | Ramp | 41.77 ± 6.51 | 41.33 ± 7.10 | |
| Coronal* | Level floor | 25.56 ± 10.55 | 27.44 ± 9.36 | 0.005 |
| | Ramp | 27.62 ± 10.28 | 30.26 ± 9.51 | |
| Transverse | Level floor | 35.625 ± 9.63 | 38.34 ± 7.69 | 0.129 |
| | Ramp | 30.38 ± 8.26 | 33.68 ± 6.26 | |

* Statistically significant difference, $p < 0.05$ (wheel type main effect, regardless of level of slope).

of motion had similar morphology for all tasks. Wheel type did not alter the general trend. In the sagittal plane, the glenohumeral joint started in the push phase from an extended position and exhibited flexion for 40–60% of wheelchair cycle on level-floor and 55–70% of wheelchair cycle on the ramp. In the coronal plane, the glenohumeral joint started from an adducted position in push phase and exhibited abduction for 40–50% of wheelchair cycle on level-floor and 50–65% of wheelchair cycle on the ramp. In the transverse plane, the glenohumeral joint exhibited external rotation for 45–60% of wheelchair cycle on level-floor and 55–70% of wheelchair cycle on the ramp.

The interaction between the wheel type and slope factors was statistically significant for the maximum glenohumeral joint angle in the transverse plane (minimum external rotation angle, $F = 10.26$, $p = 0.007$) (Table 4). The interaction between the factors was not statistically significant for the rest of glenohumeral joint ROM and glenohumeral joint angles (Tables 3 and 4). Analysis of the simple effects indicated that during ramp ascension using the geared wheels significantly decreased the maximum glenohumeral joint angle in the transverse plane ($t = 3.98$, $p = 0.002$). Analysis of the main effects indicated that regardless of the slope level, the minimum glenohumeral joint angle in the transverse plane (maximum external rotation) increased significantly ($F = 8.47$, $p = 0.012$) during geared manual wheelchair propulsion (Table 4). The wheel condition was not significantly effective on the glenohumeral joint ROM in the transverse plane (Table 3). Analysis of the main effects indicated that regardless of the slope level, using the geared wheels significantly increased the glenohumeral joint ROM ($F = 11.68$, $p = 0.005$) and the maximum glenohumeral joint angle (peak adduction; $F = 6.64$., $p = 0.023$) in the coronal plane (Tables 3 and 4). Using geared wheels did not significantly affect the glenohumeral joint ROM and peak glenohumeral joint angles in the sagittal plane (Tables 3 and 4).

### 3.3. Shoulder muscle activity

There was a statistically significant interaction between the wheel type and slope factors for the pectoralis major peak EMG ($F = 5.01$, $p = 0.043$) (Fig. 4). Analysis of the simple effects indicated that when using geared wheels on level floor, the peak muscle activity of pectoralis major decreased by 39% ($t = 3.09$, $p = 0.08$) in comparison to the standard wheel type. The pectoralis major peak activity decreased notably during geared manual wheelchair propulsion on the ramp, but was not statistically significant ($t = 1.19$, $p = 0.25$). The interaction between the wheel type and slope was not significant for the anterior deltoid and infraspinatus peak EMG. Analysis of the main effects indicated that regardless of the slope level, the peak muscle activity of infraspinatus decreased by 20% ($F = 8.00$, $p = 0.014$) during geared manual wheelchair propulsion in comparison to standard manual wheelchair propulsion (Fig. 4). The anterior deltoid peak EMG was not significantly different ($F = 2.75$, $p = 0.17$) between the geared and standard manual wheelchair propulsions (Fig. 4).
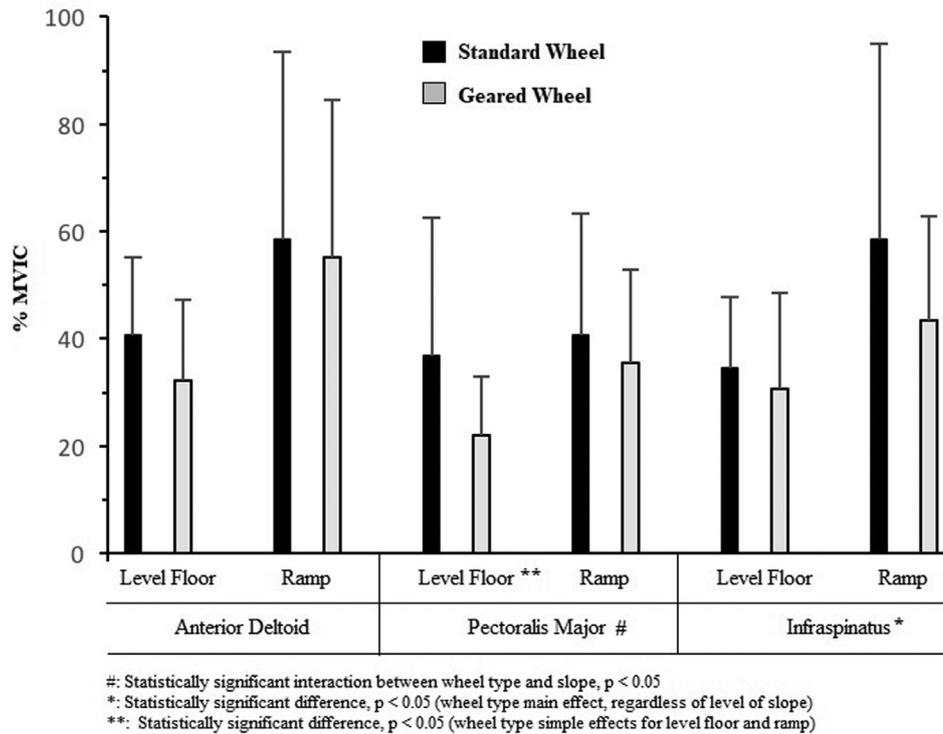
The interaction between the wheel type and slope was not significant for the integrated EMG of anterior deltoid, pectoralis major, and infraspinatus. Analysis of the main effects indicated that regardless of the slope level, the integrated muscle activity of anterior deltoid and pectoralis major decreased by 20% ($F = 7.97$, $p = 0.014$), and 35% ($F = 7.87$, $p = 0.015$), respectively, during geared manual wheelchair propulsion (Table 5). The infraspinatus integrated muscle activity decreased notably (14%), but was not statistically significant ($F = 4.51$, $p = 0.053$; Table 5).

The interaction between the wheel type and slope was not significant for the normalized integrated EMG of anterior deltoid, pectoralis major, and infraspinatus. Analysis of the main effects indicated that regardless of the slope level, the infraspinatus normalized integrated muscle activity increased by 13% ($F = 5.24$, $p = 0.039$) during geared manual wheelchair propulsion compared to standard manual wheelchair propulsion (Table 6). Changes in anterior deltoid and pectoralis major normalized integrated muscle activity during standard and geared manual wheelchair propulsion were not statistically significant (Table 6).

## 4. Discussion

We successfully characterized the spatiotemporal parameters, glenohumeral joint kinematics and shoulder muscle activity during manual wheelchair propulsion on level floor and up a ramp in 14 able-bodied individuals using geared and standard wheels.

Use of the geared wheels resulted in reduced stroke distance, speed, stroke time, and push time, and increased stroke

**Table 4**

Group mean peak glenohumeral joint kinematics (mean $\pm 1$ SD) during the standard and geared manual wheelchair propulsion on level floor and on a ramp (degrees).

| Plane | Terrain | Terrain | Standard wheels Mean ± SD | Geared wheels Mean ± SD | p |
|---|---|---|---|---|---|
| Sagittal | Max flexion | Level floor | 2.42 ± 7.79 | 2.84 ± 8.94 | 0.858 |
| | | Ramp | 3.21 ± 8.59 | 3.13 ± 7.81 | |
| | Max extension | Level floor | 40.77 ± 8.96 | 41.83 ± 8.24 | 0.598 |
| | | Ramp | 38.56 ± 8.06 | 38.19 ± 8.45 | |
| Coronal | Max adduction* | Level floor | 5.72 ± 9.55 | 6.70 ± 7.75 | 0.023 |
| | | Ramp | 8.87 ± 8.40 | 10.20 ± 7.78 | |
| | Max abduction | Level floor | 19.84 ± 12.84 | 20.74 ± 8.40 | 0.078 |
| | | Ramp | 18.74 ± 10.53 | 20.05 ± 9.44 | |
| Transverse | Min external rotation# | Level floor | 4.21 ± 11.61 | 4.19 ± 10.04 | 0.983 |
| | | Ramp | 7.39 ± 10.82** | 12.79 ± 12.25** | 0.002 |
| | Max external rotation* | Level floor | 39.47 ± 14.93 | 42.53 ± 12.37 | 0.012 |
| | | Ramp | 37.77 ± 14.64 | 46.47 ± 16.02 | |

# Statistically significant interaction between wheel type and slope, $p < 0.05$.
* Statistically significant difference, $p < 0.05$ (wheel type main effect, regardless of level of slope).
** Statistically significant difference, $p < 0.05$ (wheel type simple effects for level floor and ramp).

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*: Statistically significant difference, p < 0.05 (wheel type main effect, regardless of level of slope)
**: Statistically significant difference, p < 0.05 (wheel type simple effects for level floor and ramp)

**Fig. 4.** Group mean peak muscle activity of the anterior deltoid, pectoralis major, and infraspinatus during one stroke cycle of standard and geared manual wheelchair propulsion on level floor and on ramp. Error bars indicate the standard deviations as reported as percent of maximum voluntary isometric contraction (%MVIC).

**Table 5**
Group mean integrated shoulder muscle activity (mean ± 1 SD) during one stroke cycle of the standard and geared manual wheelchair propulsion on level floor and on a ramp, reported as percent of maximum voluntary isometric contraction (%MVIC).

| Muscle | Terrain | Standard wheels Mean ± SD | Geared wheels Mean ± SD | p |
|---|---|---|---|---|
| Anterior deltoid* | Level floor | 179.68 ± 76.34 | 150.55 ± 92.29 | 0.014 |
| | Ramp | 339.19 ± 223.70 | 267.42 ± 145.01 | |
| Pectoralis major* | Level floor | 120.61 ± 79.28 | 72.45 ± 32.53 | 0.015 |
| | Ramp | 246.41 ± 167.53 | 167.31 ± 98.24 | |
| Infraspinatus | Level floor | 200.43 ± 74.42 | 192.34 ± 99.82 | 0.053 |
| | Ramp | 342.13 ± 175.03 | 276.96 ± 110.31 | |

* Statistically significant difference, $p < 0.05$ (wheel type main effect).

**Table 6**
Group mean normalized integrated shoulder muscle activity (mean ± 1 SD) during the standard and geared manual wheelchair propulsion on level floor and on a ramp, reported as percent of maximum voluntary isometric contraction per meter (%MVIC/m).

| Muscle | Terrain | Standard wheels Mean ± SD | Geared wheels Mean ± SD | p |
|---|---|---|---|---|
| Anterior deltoid | Level floor | 150.74 ± 60.44 | 161.34 ± 92.81 | 0.421 |
| | Ramp | 500.44 ± 311.67 | 534.82 ± 277.13 | |
| Pectoralis major | Level floor | 96.73 ± 55.45 | 76.93 ± 32.73 | 0.309 |
| | Ramp | 357.15 ± 243.19 | 337.30 ± 206.11 | |
| Infraspinatus* | Level floor | 171.08 ± 71.42 | 211.07 ± 114.35 | 0.039 |
| | Ramp | 511.96 ± 252.54 | 559.28 ± 228.13 | |

* Statistically significant difference, $p < 0.05$ (wheel type main effect).

frequency when compared to the use of the standard manual wheelchair wheels. The spatiotemporal parameters including propulsion speed, stroke time, stroke distance, push time, and stroke frequency during standard manual wheelchair use on level floor were similar to previously reported values (speed = 1.02–1.29 m/s, stroke time = 1.08–1.17 s, stroke distance = 1.20–1.36 m, push phase = 42.1%, and stroke frequency = 0.9–0.96 Hz) [23,28]. The propulsion speed, stroke distance, and stroke cycle frequency during the ramp task were less than previously

reported values, while the stroke time and push phase were greater (speed = 1.22 m/s, stroke time = 0.98 s, stroke distance = 1.19 m, push phase = 50.6%, and stroke frequency = 1.04 Hz) [23]. Possible reasons for the differences could be due to the participants in our study being able-bodied, as well as the wheelchair setup and configuration. The heavier weight of the geared wheels compared to standard wheels was due to the extra mass of the gear mechanism. Because of the symmetrical distribution of the gear mechanism extra mass around the wheel axle, the extra mass

in the geared wheels did not affect the biomechanical demands of manual wheelchair propulsion. Previous studies have also shown that extra mass (5–10 kg) did not lead to a significant increase in power output, and physical strain, or changes in propulsion technique [29].

The significantly decreased propulsion speed during the geared wheel use is primarily attributed to the reduction in the stroke distance, a direct effect of the gear reduction (1.6:1) in the geared wheel and a trade-off for the significant decrease in the peak shoulder muscle activity. Stroke time decreased significantly (stroke frequency increased significantly) during geared wheel use; the geared wheel had a greater effect during ramp propulsion. The normalized stroke frequency substantially increased during geared wheelchair wheel use similar to the results reported in the literature [9]. Using the geared wheels resulted in significantly reduced push phase which was different from Howarth et al. in which there was no significant difference between the geared and standard conditions for the push phase [9]. The effects of using the geared wheelchair wheels on normalized stroke frequency and push phase was significant and should be taken into consideration for manual wheelchair prescription. The scientific evidence previously outlined for recommendations related to wheelchair propulsion emphasize a minimization of cadence or push frequency [30]. Increasing push frequency has been linked to increased risk of secondary upper extremity injuries in manual wheelchair users [4,30]. Long propulsive strokes (push phase) has been also recommended to minimize cadence and peak forces [30].

Glenohumeral joint kinematic data were similar to previously reported values for wheelchair biomechanics [31]. The wheel condition did not significantly affect the glenohumeral joint kinematics in the sagittal plane. The glenohumeral joint ROM in the sagittal plane increased notably during ramp ascent in comparison to level floor propulsion for both standard and geared wheels. This might be because of increased trunk forward flexion during ramp ascent [23]. The significant increase in the glenohumeral joint ROM in the coronal plane could be due to the larger wheel width of the geared wheels in comparison to standard wheels (Table 1). Increased glenohumeral joint internal rotation has been associated with a reduction in the subacromial space and impingement [31] which is a main contributor to the development of the shoulder pain in manual wheelchair users [32]. Therefore, the significant increase in glenohumeral joint external rotation during geared manual wheelchair mobility which we observed could be beneficial for manual wheelchair users.

Using the geared manual wheelchair resulted in substantial reduction in the peak muscle activity of the primary shoulder flexors (pectoralis major, anterior deltoid), and infraspinatus compared to standard wheels. Similar results were reported in the previous investigation on geared manual wheelchairs [9]. Use of the geared wheels had a greater effect on the peak activity of the shoulder flexor muscles during propulsion on level floor and on the peak activity of the infraspinatus during ramp propulsion. Reducing peak shoulder muscle activity is the primary benefit of using geared manual wheelchair wheels. A reduction in the peak muscular demands could lead to a reduction in peak forces applied to glenohumeral joint and consequently a reduction in the risk of shoulder secondary injury and pain, thereby improving independence of wheelchair users [33]. Using the geared wheels decreased the integrated shoulder flexors muscle activity substantially in comparison to standard wheels.

Combined metrics, such as normalized integrated muscle activity, incorporate both muscle activity and stroke cycle characteristics. Normalized integrated muscle activity provides a more comprehensive characterization of the impact that the geared wheel system has on upper extremity biomechanics. This metric, among others, may provide a better understanding of how geared wheels use affects wheelchair propulsion. The normalized integrated muscle activity of the infraspinatus increased significantly during the geared wheels condition. This could be due to the significant increase in the maximum external rotation of the glenohumeral joint during geared wheel use. The increased activity of infraspinatus muscle could better stabilize the glenohumeral joint by externally rotating the humerus and exerting a depression force on the humeral head (increasing joint space width); therefore, decreasing the risk of shoulder impingement. The normalized integrated muscle activity for the shoulder flexor muscles were not significantly different between the geared and standard wheel conditions. However, they were significantly higher for the geared condition in the investigation by Howarth et al. [9]. The difference in gear ratio (2:1 vs. 1.6:1) and slower velocity with the geared wheel with larger gear reduction (2:1) could be the main reasons for this difference. This might be an evidence that the smaller gear reduction (1.6:1) could be a more optimized gear ratio for manual wheelchair users, particularly for longer distances.

## 5. Conclusion

The spatiotemporal parameters, glenohumeral joint kinematics, and shoulder muscle activity during standard and geared manual wheelchair propulsion on level floor and ramp were quantified in this study. Overall, using geared wheels resulted in substantial reduction of shoulder muscle activity and improvement in glenohumeral joint kinematics during manual wheelchair propulsion. These results suggest that using the geared wheels may have the potential to decrease the risk of secondary musculoskeletal injuries and joint pain in manual wheelchair users. We also found a significant increase in the stroke frequency and significant decrease in the speed and stroke distance which translate to a more repetitive task during geared wheels use. Transition to a geared manual wheelchair might be an alternative to standard manual wheelchair mobility in the context of preservation of upper limb function as well as the need to remain physically active. However, further biomechanical investigations including hand-rim biomechanics, upper extremity joint dynamics, and muscle activity with manual wheelchair users with spinal cord injury are warranted.

**Ethical Approval**

This study was approved by the University of Wisconsin-Milwaukee Institutional Review Board. All subjects provided

written consent to the collection and use of data reported in the manuscript.

## Disclosure Statement

Elizabeth Hsiao-Wecksler is a co-founder of IntelliWheels, Inc, the manufacturer of the geared manual wheelchair wheels that was used in this study. IntelliWheels, Inc. was sold in July 2018 and Dr. Hsiao-Wecksler is no longer involved in the company.

## Supplementary material

Supplementary material associated with this article can be found, in the online version, at doi:10.1016/j.medengphy.2019.06.018.

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