Electomyographic activities of shoulder muscles during Handwheelchair.Q vs pushrim wheelchair propulsion

Paride Cavallone [a,*], Taian Vieira [b], Giuseppe Quaglia [a], Marco Gazzoni [b]

[a] Department of Mechanical and Aerospace Engineering, Politecnico di Torino, Turin, Italy

[b] Department of Electronics and Telecommunications, Politecnico di Torino, Turin, Italy

[*] Corresponding author: paride.cavallone@polito.it

Abstract

Different mechanisms of force transmission have been developed for the movement of wheelchairs, from the standard pushrim propulsion to the handbike. Contributing to this repertoire, we recently developed a system of propulsion based on a pulley-cable mechanism, the Handwheelchair.Q. In contrast to other propulsion systems, the Handwheelchair.Q requires users to extend the shoulders and flex the elbows to move the wheelchair forward, mimicking the rowing gesture. Whether however our proposed, propulsion system imposes a similar degree of shoulder muscles excitation with respect to the conventional, pushrim system is yet to be addressed. In this study we therefore assess whether the Handwheelchair.Q demands a similar degree and timing of muscle excitation with respect to the pushrim wheelchair, for a given travelled distance. We address this issue by sampling the angular speed of the two wheels and the surface EMGs from ten, shoulder muscles, while seven subjects use the two propulsion systems at constantly low and high speeds, one at a time. As expected, results revealed opposite muscle groups were excited when comparing the two mechanisms for wheelchair propulsion. ANOVA statistics indicated the amplitude of EMGs was greater for shoulder flexors and elbow extensors during the drive phase of pushrim propulsion, with the opposite being observed for the Handwheelchair.Q. Interestingly, from the angular speed we observed a significantly greater average displacement was achieved with the Handwheelchair.Q. Our results support therefore the notion that, with respect to pushrim propulsion, subjects were able to move faster without overloading the shoulder muscle with the Handwheelchair.Q.

1. Introduction

Spinal cord injuries (SCI) affect a large fraction of the world population, severely impairing the activities of daily living. Notwithstanding the marginal (0.2%) decrease in SCI occurrences from 1990 to 2016, the incidence of SCI was still markedly high in 2016, amounting to roughly 0.93 (0.78 – 1.16) million of new cases worldwide [1]. The consequences of SCI span a broad spectrum, from the emergence of disorders typically associated with physical inactivity [2], [3] to mental health problems [4]. Of our particular interest is the loading of upper limb muscles during wheelchair propulsion. Locomotion of SCI subjects is ensured through the use of wheeled mobility equipment, with the most popular examples being manual wheelchairs, power wheelchairs, and scooters [5]. Manual wheelchairs are the most employed means of mobility in the SCI population: among the 3.6 million users in North America, nearly 90% use a manual wheelchair [6]. Despite their popularity, musculoskeletal complications arising from the use of these conventional wheelchairs have been documented.

The upper limb is the most commonly affected site in manual wheelchair users. Indeed, different studies have reported that over 70% of manual wheelchair users suffer from shoulder pain [7]–[9]. Moreover, repetitive stress injury has been often observed in SCI subjects [10], [11], likely because of the overt necessity of using the upper limbs for locomotion. In addition to hindering the mobility of SCI subjects, these injuries impact on whichever activity demands the use of the shoulder joint. With the goal of preventing or minimising the consequences of these musculoskeletal injuries, wheelchairs based on different systems of propulsion have been designed in the last years [12], [13]. For instance, recent studies have reported a diminished excitation of the shoulder muscles when subjects were asked to reproduce the movement necessary to move a wheelchair using reverse propulsion [14] and lever system [15] approaches than when relying on the standard, pushrim propulsion system [16]–[18]. While these results are encouraging, the efficiency of the movement with these innovative systems remains an issue, at least in paraplegia. In addition to alleviating the muscle demand, these systems should ensure the same extent of mobility as that experienced with the conventional wheelchair. This was indeed one of the reasons motivating the design of a wheelchair propelled through a pulley-cable system [19]–[21], which effect on the neuromuscular system has yet to be tested.

Different studies [22]–[24] have shown the low efficiency of the pushrim system, from different points of view. A common characteristic of the pushrim, lever and handbike systems of propulsion is that in the drive phase these systems impose a fixed trajectory for the hand. In addition, the pushrim and the lever system require a pushing movement. The pulley-cable system does not impose a fixed trajectory for the hand and requires a pulling movement. So, the pulley-cable system represents an alternative, not only to the pushrim system, but also to the lever system and the Handbike. The idea of the innovative, pulley-cable system of propulsion is to propel the wheelchair relying on a gesture that reproduce the trunk and arm movements taking place during the last third of the rowing drive phase [25]. The user bilaterally pulls a pair of cables that are wrapped around a pulley transmitting the end-point force to the rear wheels. Handhweelchair.Q is a prototype of wheelchair with a rigid, not foldable, standard frame. The prototype has been equipped with two telescopic rods, on the left and right side. At the end of the telescopic rods, two return pulleys are mounted in a location to enable the user to perform the rowing gesture.

In this study, we evaluate the neuromuscular demands and the degree of mobility provided by our innovative wheelchair. We specifically ask whether the pulley-cable propulsion system would lead to a more efficient movement, requiring less muscular effort for a given movement speed, when compared with the standard, pushrim approach. Movement efficiency was assessed in terms of speed and travelled distance and excitation of the shoulder muscles for both wheelchair propulsion systems. Given the pulley-cable propulsion demands the shoulder extension rather than flexion during the propulsion phase, we expect to observe a smaller or comparable degree of muscle excitation. Moreover, as the pulling propulsion is ensured by a pulley system, which radius is remarkably smaller than that of the wheel, individual movement cycles are expected to result in greater speeds and distances for the pulley-cable rather than the pushrim propulsion system without overloading the shoulder muscles.

2. Methods

2.1. Participants

Seven able-bodied adults (six male) were recruited to participate in the study (range values; age: 24-40 yrs; height: 168-183 cm; body mass: 55-75 kg). None of the subjects reported any musculoskeletal disorders that could affect their upper limb movements at the occasion of experiments, which commenced after subjects provided written informed consent. Experimental procedures conformed with the *Declaration of Helsinki* and were approved by the Local Ethics Committee. Able-bodied subjects, inexpert on the use of wheelchairs, were chosen because, otherwise, we would be unable to discriminate the effect of experience and wheelchair propulsion system on the degree of muscle excitation.

2.2. Experimental protocol

Subjects were first asked to practice on a wheelchair specifically designed to accommodate both pulley-cable and pushrim, propulsion systems: the Handwheelchair.Q. The Handwheelchair.Q was positioned over a commercial roller test bench. This test bench (Manufacturer: Invictus Active, City: Walsall (UK), WS2 8TL, Model: Invictus Active Trainer) is a device to practice indoor activity for wheelchair users. The roller test bench has low inertia and high friction torque (Figure 1; [19]). The familiarisation session lasted about 5 min and comprised practicing with both propulsion systems at variable speeds.

After that, participants were asked to propel the wheelchair using the pushrim (Figure 2A) and the pulley-cable (Figure 2B) systems, at low and high speeds. Each of these four trials lasted roughly 2 min, ensuring at least 30 movement cycles per trial. Low and high speeds were self-selected, according to the instruction given to subjects: to perform at a comfortable speed, as if they had to move around normally (low speed), and to perform twice as fast as for the comfortable speed condition (high speed). Each trial was applied twice, providing a total of eight recordings. Trials were applied in random order, with rest periods of 2 min in-between.

2.3. Kinematics and electrophysiological data acquisition

To study upper arm kinemtics, 17, reflective markers, were positioned bilaterally in the upper limbs according to a modified version of the full-body, plug-in gait marker protocol (Nexus Plug-in Gait marker set, Oxford Metrics, Oxford, UK), to model only the trunk, arm, and forearm segments. After sampling (100 Hz) and labelling markers, their coordinates were used to computecardan angles for the shoulder and elbow joints using the Nexus software (Nexus 2.11 software, Vicon Motion System, Oxford, UK).

Pairs of circular surface electrodes (24 mm diameter with roughly 30 mm center-to-center distance, Spes Medica, Battipaglia, Italy) were used to collect surface electromyograms (EMGs) from ten muscles in the dominant side: anterior deltoid (AD), middle deltoid (MD), posterior deltoid (PD), middle trapezius (MT), upper trapezius (UT), infraspinatus (IS), supraspinatus (SS), pectoralis major, biceps brachii (BB) long head and triceps brachii (TB) lateral head. The selected muscles are either prime movers or stabilisers of the shoulder [26], [27] and represent an extended set in relation to that considered in previous studies comparing different wheelchair propulsions [14], [15]. After carefully shaving and cleaning the skin with abrasive paste, surface electrodes were positioned on the skin surface over the muscles of interest (Figure 1). Pairs were centred at locations recommended by SENIAM [28], except for infraspinatus and supraspinatus, for which the positioning procedure was as described by [29]. Bipolar EMGs were recorded with a wireless system (200 V/V gain; 10–500 Hz bandwidth amplifier; DuePro system, OTBioelettronica and LISiN, Politecnico di Torino, Turin, Italy). EMGs were digitized at 1,000 Hz with a 16 bits A/D converter, synchronously with the sampling of markers' coordinates (Nexus 2.11 software, Vicon Motion System, Oxford, UK).

In addition to subject's biomechanical and electrophysiological data, the speed of the wheels of the Handwheelchair.Q was assessed. The angular velocity of the wheelchair wheels was determined with two Hall sensors (SS490 MRL, Honeywell, Charlotte, USA) each mounted on the wheelchair frame beside the left and right rear wheels (Figure 1). With this sensor, which output was sampled with a 16-bit A/D converter at 1000 Hz (NI USB-6341, National instrument, Austin, Texas, USA), we could detect the passages of sixteen equidistant magnets positioned on each rear wheel. A TTL trigger pulse was used for the offline synchronisation of the Hall sensor readings and biomechanical and EMG data.

2.4. Data analysis

Individual movement cycles were identified from the acceleration of the wheelchair wheels. First, angular velocities were obtained for the right (ω_r) and left (ω_l) wheels according to:

$$\omega_r[i] = \left(\frac{1}{\Delta t_r[i]}\right)\frac{\pi}{8} \tag{1}$$

$$\omega_l[j] = \left(\frac{1}{\Delta t_l[j]}\right)\frac{\pi}{8} \tag{2}$$

where Δt_r and Δt_l respectively denote the time interval taken for the right and left wheels to move by $\pi/8$ radians each. Both angular speeds were linearly interpolated (1000 Hz), averaged, and then multiplied by the radius (r_{rw}) of the wheels to provide the equivalent wheelchair speed:

$$Xd[k] = \frac{(\omega_r[k] + \omega_l[k])}{2} r_{rw} \quad (3)$$

where k is an integer ranging from 1 to the total number of samples interpolated over the trial duration.

Drive and recovery phases were then identified based on instants when acceleration values crossed the 0 reference value (Figure 3). The acceleration is the time derivative of the equivalent speed, Xd. Crossings from negative to positive values denote the transition from recovery to drive whereas crossings from positive to negative values indicate transition from drive to recovery. A single cycle was defined by consecutive instants corresponding to the transition from drive to recovery. Once cycles were identified, the first five cycles were discarded to remove any transient effects associated with the commencement of the trial. From the wheel velocity we further computed the projected travelled distance and the average, linear speed of the wheelchair.

The range of motion of the shoulder joint and the timing and degree of muscle excitation were computed separately for each cycle and subject. Range of motion values were defined as the maximal flexion-extension excursion of the shoulder and elbow joints within cycles. The timing of muscle excitation was estimated as detailed in [30]. Briefly, after full-wave rectification, EMGs were low pass-filtered (2nd order Butterworth filter; 30 Hz cut-off) into their envelopes [31]. We then computed the baseline level as the mean plus three standard deviations of the EMG envelope during the first second of acquisition (rest). Within each movement cycle, for each time sample, we computed how many out of the 200 (200 ms) preceding and the 200 succeeding samples did not and did respectively exceed the baseline level. The time sample providing the greatest count was deemed as the onset of muscle excitation [30]. Onset of silencing was defined using the same procedure, considering though the sum of preceding samples exceeding and succeeding samples not exceeding the baseline level. Onsets were identified separately for each muscle and cycle. Occasionally, multiple maxima and minima were observed in the counts, indicating multiple bursts of excitation were present. Whenever it occurred, we retained the burst providing the greatest, time-integrated envelope value (cf. horizontal, black traces in Figure 3). In addition to the timing of muscle excitation, we computed the duty cycle of muscle excitation, considering the highest burstwithin cycles, and the average amplitude of EMG envelopes, computed over samples for which the muscle was indeed excited. Subjects for which no onset values could be identified, because the muscle was either not excited or did not rest within cycles, a duty cycle of 0% or 100% was respectively assigned.

2.5. Statistical analysis

Parametric, inferential statistics was used to assess the effect of test conditions on kinematics data, after ensuring their distribution was Gaussian (Shapiro-Wilk test, $p>0.15$) and the homogeneity of variances (Levene's test, $p>0.2$). Two-way ANOVA was applied to assess whether the propulsion system, taken as repeated measures, and the movement speed affected the duration of cycles and their phases, the joint range of motion and the chair speed and displacement. The assumption of equal variances for the different groups (propulsions and speeds) was not verified for muscle excitation data. We therefore applied the Wilcoxon test to assess the separate effect of propulsion and movement speed on the onset of muscle excitation and silencing, the duty cycle, and the RMS amplitude, after applying Bonferroni correction to compensate for the inflation of type I error.

3. Results

Qualitative differences in wheelchair mobility, joint kinematics, and muscle excitation can be appreciated for the representative data shown in Figure 3. The same time scale (2 s) was used to represent data for both propulsion systems, highlighting the shorter time taken to complete a cycle with the pushrim than pulley-cable propulsion. During drive, elbow and shoulder moved in opposite directions, both for the pushrim and pulley-cable propulsion systems, with the latter being associated with a roughly 20° greater elbow range of motion (cf. traces in Figure 3 middle panel). Excitation of posterior and anterior deltoid took place at different, relative instants of the movement cycle, with the posterior and anterior being respectively elicited at recovery and drive for the pushrim propulsion and vice-versa for the pulley-cable propulsion (Figure 3). As reported below, group results confirm the observations just made for a single subject.

3.1. Wheelchair and joint kinematics

Even though we were unable to control for the movement cadence in real time, subjects performed the tasks at a consistent cadence. For the two propulsion systems and the two movement speeds, the coefficient of variation of the duration of drive and recovery phases did not exceed 8%. Similarly, the relative duration of drive and recovery phases did not change with both the propulsion system (ANOVA $F<0.4$, $p>0.59$) and the movement speed ($F<4.2$, $p>0.07$; Figure 4*A*). The duration of cycles was however significantly greater for the pulley-cable system and the low speed (ANOVA main effect, $F>7.4$, $p<0.02$).

Regardless of the propulsion system, the fast speed resulted in greater and faster, linear displacements of the wheelchair (F>31.1, p<0.01; Figure 4*B*). The pulley-cable propulsion led to wheelchair displacements, in the drive phase, roughly three times greater in relation to the pushrim propulsion (F=46.0, p<0.01), even though no effect on the wheelchair speed was observed (F=2.8, p=0.12).

When considering joint kinematics, the pulley-cable propulsion demanded a greater range of elbow (F=12.0, p<0.01) though not shoulder (ANOVA F=0.3, p=0.62) flexion-extension (Figure 4*C*). No effect was observed for movement speed (ANOVA F<0.5, p>0.47).

3.2. Onset of muscle excitation and silencing

For both movement speeds, the onset of muscle excitation and silencing depended on the propulsion system. Except for the anterior deltoid, which was excited and silenced at roughly mid drive and early recovery respectively, muscles were excited early in recovery and silenced at mid drive for the pushrim propulsion (Figure 5). The opposite was observed for the pulley-cable propulsion, resulting in significantly different onset values for all muscles represented (Figure 5; Wilcoxon paired test; Onset excitation: p<0.035; Onset silencing: p<0.045; henceforth all p values are reported after Bonferroni correction for multiple comparisons). Results for pectoralis major and upper trapezius were not presented because onsets could not be computed for all subjects, as EMG amplitude did not exceed the baseline for the former (0% duty cycle) and was always greater than the baseline for the latter (100% duty cycle).

3.3. Muscle excitation and duty cycle

Movement speed did not affect the duty cycle for all muscles tested (Wilcoxon paired test; p>0.22). With respect to the propulsion system, pulley-cable propulsion resulted in a greater duty cycle for posterior deltoid and supraspinatus and a smaller duty cycle for pectoralis major (Figure 6; p<0.04). Regarding the RMS amplitude of EMGs, movement speed affected eight out of the ten muscles tested, with faster movements leading to greater amplitude values (Wilcoxon paired test; p<0.04). The effect of the propulsion system was variable across muscles, with the pulley-cable propulsion leading to smaller RMS values for middle trapezius, supraspinatus, and middle and posterior deltoid (Wilcoxon paired test; p<0.05) and greater amplitude values for anterior deltoid, upper trapezius, biceps brachii and infraspinatus (Wilcoxon paired test; p<0.04).

4. Discussions

In this study we test the hypothesis that subjects are able to move more efficiently on a wheelchair which propulsion system is based on pulley-cable system (Figure 2) than the conventional, pushrim approach. We assessed efficiency in terms of the timing and degree of muscle excitation and the angular kinematics of the wheelchair. Our results support the notion that, with respect to the pushrim propulsion, subjects were able to move faster without overloading the shoulder muscle with the pulley-cable propulsion. This preliminary test has shown that the total, projected displacement for each cycle obtained with the pulley-cable system is greater than that obtained with the pushrim system (Fig. 4B), notwithstanding the similar, active muscle loading imposed by both systems as revealed by the EMGs collected (Fig. 6).

The excitation and silencing phases (Figure 5) of all eight muscles analysed were in opposition when comparing the pushrim and the pulley-cable propulsion systems. The IS, SS, TB, BB, MT, PD and MD muscles were excited during the transition from the recovery to the drive phase for the pushrim system while these muscles were excited during the transition from the drive to the recovery phase for the pulley-cable system. The AD muscle was excited during the transition from the drive to the recovery phase for the pushrim system while it was excited during the transition from the recovery to the drive phase for the pulley-cable system. This result is presumably due to the opposite demands for muscle loading between propulsion systems: while pushrim propulsion demands shoulder flexion and elbow extension during the drive phase the pulley cable propulsion demands shoulder extension and elbow flexion, as shown in Figure 2. The results of muscle excitation reported here corroborate those documented in a previous study [14], focused on the assessment of excitation of six muscles during forward and reverse-wheeling propulsion. As for the pulley-cable propulsion, the drive phase in reverse-wheeling propulsion is characterized by shoulder extension and elbow flexion and thus by out-of-phase excitation of trapezius and deltoid muscles when compared to the forward-wheeling propulsion [14].

The effect of the propulsion system on the degree and timing of muscle excitation was observed to be variable (Figure 6). More specifically, the pushrim and the pulley-cable propulsion demanded longer and more active loading of different muscles: some muscles (AD, UT, PM, BB, and IS) were more strongly excited during pushrim whereas others (MD, PD, MT, and SS) were more strongly excited during pulley-cable propulsion. This difference, expectedly due to the different kinematics imposed by both propulsion systems, suggests the amount of active loading of the upper limb muscles per movement cycle may be comparable. The balanced excitation of different muscles between conditions does not however indicate the two propulsion systems are associated with a similar efficiency. Indeed, the linear displacement of the wheelchair with the pulley-cable system of propulsion is approximately three times greater with respect to the pushrim propulsion per cycle (Figure 4*B*). This fact is likely due to the longer propulsion time required by the pulley-cable propulsion system (Figure 4A). The difference in propulsion time between systems is attributable to the different transmission ratio between the two systems. In fact, the transmission ratio for the pushrim system is defined by the ratio between the radius of the rear wheel r_{rw}, and the radius of the pushrim r_h, being thus approximately equal to one. Conversely, the transmission ratio of the pulley-cable system is defined by the ratio between the radius of the rear wheel and the radius of the pulley r_p: this ratio is nearly 2.2 greater than that for the pushrim system (Figure 2). In addition, the range of motion of the shoulder is similar for the two system of propulsion (Figure 4*C*) while the range of motion of the elbow is roughly 50% higher for the pulley-cable system. Collectively, these results suggest greater distance may be travelled when moving with the pulley-cable propulsion system, for a presumably similar demand for active muscle loading.

The effect of the speed was observed to influence the degree of excitation of the ten muscles tested. In fact, the amplitude of bipolar EMG collected from the ten muscles assessed increased with the increase in movement speed, while the duty cycle was generally variable across muscles and propulsion systems (Figure 6). Similar results have been reported in [15] regarding the pushrim and the lever, propulsion systems. More specifically, these authors reported an increase in the EMG amplitude with increases in movement speed in all muscles analysed, while only 3 out of the 4 muscles assessed showed a longer period of activity [15].

An additional important difference between the two systems is the trajectory of the hand. In the drive phase, the pushrim system imposes a fixed trajectory for the hand whereas no hand trajectory is imposed for pulling the handles with the pulley-cable system. It seems therefore plausible to state that with the pulley-cable propulsion system users can rely on different kinematic strategies to propel the wheelchair, likely adopting that associated with an optimal ergonomic propulsion. In addition, when following a fixed trajectory, the user force on the handrim, or lever, has three components as shown in [32], but only one is useful for the transmission of motion, namely the tangential force, while with the pulley-cable system the end-point force wholly contributes to the generation of the angular momentum driving the wheelchair.

The main limitation of this work is the generalization of the results obtained to disabled people. Seven able-bodied subjects, inexpert on the use of wheelchairs, were chosen to examinate the excitation of ten shoulder muscles with two systems of propulsion: the pushrim and the pulley-cable systems. Able-bodied subjects were chosen because we would be unable to discriminate the effect of experience from that of the wheelchair propulsion system (pushrim and pulley-cable) on the degree of muscle excitation and performance. We acknowledge, therefore, that different results may emerge for a sample of SCI subjects, whose experience with the pushrim propulsion is far greater than with the incipient, pulley-cable system propulsion. While future studies are required to address this issue, our current results suggest that subjects with similar experience on both propulsion systems may be able to move more efficiently with the pulley-cable rather than the pushrim, propulsion system.

Conflicts of interest

The authors have no conflicts of interest to declare.

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Ethical approval

The experimental protocol conformed with the Declaration of Helsinki and was approved by the Regional Ethics Committee (Commissione di Vigilanza, Servizio Sanitario Nazionale—Regione Piemonte— ASL 1—Torino, Italy).

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