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Glenohumeral Contact Forces and Muscle Forces Evaluated in Wheelchair-Related Activities of Daily Living in Able-Bodied Subjects Versus Subjects With Paraplegia and Tetraplegia

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ABSTRACT. van Drongelen S, van der Woude LH, Janssen TW, Angenot EL, Chadwick EK, Veeger DH. Glenohumeral contact forces and muscle forces evaluated in wheelchair-related activities of daily living in able-bodied subjects versus subjects with paraplegia and tetraplegia. *Arch Phys Med Rehabil* 2005;86:1434-40.

Objective: To estimate the differences in glenohumeral contact forces and shoulder muscle forces between able-bodied subjects and subjects with paraplegia and tetraplegia during wheelchair-related activities of daily living (ADLs).

Design: Kinematics and external forces were measured during wheelchair ADLs (level propulsion, weight-relief lifting, reaching) and processed by using an inverse dynamics 3-dimensional biomechanical model.

Setting: Biomechanics laboratory.

Participants: Five able-bodied subjects, 8 subjects with paraplegia, and 4 subjects with tetraplegia (N=17).

Interventions: Not applicable.

Main Outcome Measures: Glenohumeral contact forces and shoulder muscle forces.

Results: Peak contact forces were significantly higher for weight-relief lifting compared with reaching and level propulsion ($P < .001$). High relative muscle force of the rotator cuff was seen, apparently needed to stabilize the joint. For weight-relief lifting, total relative muscle force was significantly higher for the tetraplegia group than for the able-bodied group ($P = .022$).

Conclusions: Glenohumeral contact forces were significantly higher for weight-relief lifting and highest over the 3 tasks for the tetraplegia group. Without taking paralysis into account, more muscle force was estimated for the subjects with tetraplegia during weight-relief lifting.

Key Words: Activities of daily living; Biomechanics; Muscles; Rehabilitation; Shoulder; Spinal cord injuries.

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SHOULDER PAIN OFTEN INTERFERES with activities of daily living (ADLs) essential for the functional independence of people with a spinal cord injury (SCI).¹ One factor that could contribute to the development of shoulder complaints is the relatively heavy and frequent loading of the upper extremity during wheelchair ADLs, such as transfers and weight-relief lifts.²⁻⁴

The load on the shoulder has often been quantified as net joint moments, probably because net moments are fairly simple to determine. However, net joint moments do not necessarily reflect the magnitude and distribution of muscle forces and the stability requirements in the shoulder. Compared with net shoulder moments, glenohumeral contact forces might be a more accurate indicator of mechanical load at the shoulder joint because contact forces reflect the sum of the external forces and the muscle forces around the joint. The compression force on the joint surface may cause damage to the joint surface, whereas the muscle forces can be high to stabilize the joint and therefore may lead to soft-tissue damage. Glenohumeral contact forces have been shown to correlate well with net moments for wheelchair propulsion in subjects with a low-lesion SCI,⁵ but it is unlikely that the same relationship would hold for subjects with high-level or different levels of SCI.

In a previous study,⁴ no differences were found in net joint moments during wheelchair ADLs between subjects with a high-level SCI and subjects with a low-level SCI. The glenohumeral contact forces are, however, expected to be higher for people with high-level SCI than for people with low-level SCI and able-bodied individuals. Glenohumeral contact forces for ADLs in subjects with SCI have not yet been studied. Contact forces have been estimated for wheelchair propulsion by Veeger et al,⁵ who reported peak glenohumeral contact forces between 800 and 1400N. In studies with able-bodied subjects, Anglin et al⁶ reported values up to 1750N for lifting a 10-kg suitcase. Kuijer et al⁷ reported contact forces between 500 and 1500N for pulling and pushing containers 40 to 74kg.

Because of muscle paralysis in subjects with a (high-level) SCI, other muscles must be more active to provide joint stability and to provide for the necessary external force, placing active muscles at increased risk for overuse injuries. Muscle activity during ADLs has been studied by using electromyography by Reyes,³ Perry,² and Newsam⁸ and colleagues, who showed high muscle activation of the latissimus dorsi, triceps caput longum, and the pectoralis major pars sternalis muscles during transfers and weight-relief maneuvers, respectively. For wheelchair propulsion, the rotator cuff muscles⁵ and the pectoralis major muscle⁹ seem to be at risk for muscle damage.

The first aim of this study was to determine the glenohumeral contact forces and the muscle forces during 3 wheelchair-related activities. The second aim was to determine if there are differences in the glenohumeral contact forces and muscle forces among able-bodied subjects, subjects with paraplegia, and subjects with tetraplegia. It was hypothesized that

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Table 1: Subject Characteristics

Group	Age (y)	Height (m)	Body Mass (kg)	Injury Level	Time Since Injury (y)
Able-bodied (n=5)	22±3	1.82±0.11	73±5	NA	NA
Paraplegic (n=8)	39±12*	1.86±0.08	79±9	T3-12	14±10
Tetraplegic (n=4)	28±5	1.88±0.05	70±14	C6-7	7±6

NOTE. Values are mean ± standard deviation.

Abbreviation: NA, not applicable.

*Significantly older compared with able-bodied subjects ($P<.05$).

the glenohumeral contact forces would be higher for specific wheelchair ADLs compared with wheelchair propulsion and that the rotator cuff muscles would be highly active to stabilize the glenohumeral joint. Further, we expected that the contact forces and muscle forces would be higher for subjects with tetraplegia than for able-bodied subjects and subjects with paraplegia.

METHODS

Participants

A convenience sample of 5 able-bodied subjects, 8 subjects with paraplegia, and 4 subjects with tetraplegia participated in this study after giving written informed consent. Subjects were eligible to participate if they had no current shoulder complaints, did not have cardiovascular diseases, and had sufficient cognitive capacity to understand the goal of the study and the testing methods. Two of the subjects with paraplegia and 1 subject with tetraplegia had an incomplete lesion. Subject characteristics are listed in table 1. The protocol of this study was approved by the Medical Ethical Committee of the Vrije Universiteit Medical Center.

Protocol and Tasks

Subjects performed a set of 5 standardized ADL tasks under experimental conditions in an instrumented wheelchair. From the 11 ADLs measured in a pilot experiment, 5 tasks were selected by the researcher and a physician. These 5 tasks were selected for their commonality in daily life and for their suggested variation in their strenuous nature. Both 3-dimensional external forces and moments and 3-dimensional kinematics of the upper extremity were determined during each activity. Before testing, all subjects were allowed to become accustomed to the experimental setup, by propelling freely to feel the properties of the experimental wheelchair and by practicing each task before measurements started. For this study, only 3 tasks were analyzed: wheelchair propulsion, weight-relief lifting, and placing a bottle on a platform.

To ensure a submaximal exercise level for all subjects, wheelchair propulsion was performed on a level treadmill^a at .83m/s. Subjects were instructed before the test and reminded during the test to use only the handrim for propulsion.

For the second task, subjects had to perform a weight-relief lift. Because of the design of the recording system, the lift had to be performed with the hands on the handrims. However, subjects were allowed to place the left (nonmeasured) hand on the tire to create a wider support base. This task was performed 3 times with a 20-second rest between trials.

The third task was a reaching task; subjects had to place a 1.5-kg bottle on a platform 0.5m high with their right hand. The bottle had a ring under the cap so that the subjects with tetraplegia were also able to grasp the bottle with the help of the ring. At the beginning of the task, subjects sat in the wheelchair and held the bottle in their lap; subsequently, they

placed the bottle at the platform in front of them, held it there, and then took it back to the starting position.

Instrumented Wheelchair

All tasks were performed in a standard design Quickie Triumph wheelchair.^b A 6 degrees-of-freedom force transducer^c had been built in the right wheel. The handrim was connected to the transducer by an aluminum shell. Next to the transducer, a portable data acquisition device^d and a custom angular position sensor were built into the wheel (fig 1). The width of the back of the wheelchair was .42m; the height was .40m. The seat was .42m wide and deep. The seat height was .55m, the seat angle to the horizontal was 10°, and the angle of the back to the vertical was 5°. The radius of the wheels and rims were, respectively, .305 and .265m. The diameter of the rim tube was 20mm, the pressure of the rear tires was 4.5 bar, and the camber of the wheels was set at 5°. After the instrumented wheel was balanced, the inertia was calculated and subsequently the inertia of the other wheel was corrected by adding extra weights. The total mass of the wheelchair was 18.6kg.

Data were stored on a memory Flash card with a sampling rate of 100Hz, which is high enough to accurately collect both kinematic and kinetic data.¹⁰ The instrumented wheel enabled us to measure the (propulsive) forces applied on the handrim, as well as the moments on the handrim. The hand moments applied by the hand on the rim were calculated from the difference between the moment measured around the wheel axis and the moment produced by the applied force on the handrim.¹¹ The point of force application of the hand was assumed to be at the third metacarpal. The sensitivity of the instrumented wheel for the forces was (F_x [forward]=3.0N, F_y

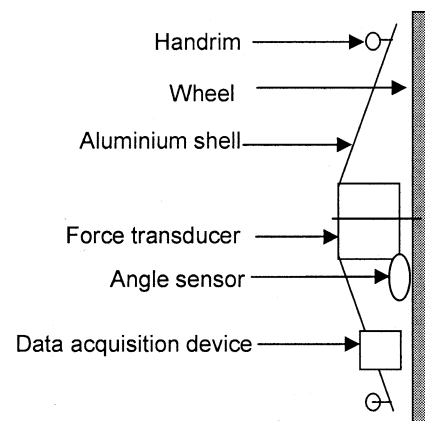


Fig 1. Schematic drawing of the instrumented wheel, placed at the right side of the wheelchair; cross-sectional front view.

[downward]=2.8N, F_z [medial]=4.1N) and for the moments ($M_x=0.3\text{Nm}$, $M_y=0.7\text{Nm}$, $M_z=0.4\text{Nm}$).

The AMTI force transducer was synchronized with the Optotrak^e computer by a telemetric system.^f Forces and moments were low-pass filtered by using a 10-Hz second-order recursive Butterworth filter. All moments and forces from the wheelchair were transformed from the rotating (local) coordinate system of the force transducer to forces and moments in the global coordinate system and subsequently corrected for the camber of the wheelchair and for the offset (the weight of the rim and the shell connected to the transducer).

Kinematics

Kinematics were recorded during each experimental trial by using 3 Optotrak cameras^e operating at 100Hz. Seventeen active markers were placed on the right side of the subject's body (thorax, upper arm, forearm, hand) as well as on the wheelchair.^{12,13} Recordings were performed with additional technical markers on the elbow (epicondylus medialis humeri) and the forearm (processus styloideus ulnae). Before the actual measurements, a calibration measurement was performed in which the orientation of the technical markers was defined relative to the bony landmarks. Also, the orientation of the scapula was determined in a calibration measurement with a scapula locator system⁴ while the subject was sitting in the wheelchair with the arms in the anatomic position. From the scapula calibration measurement and the orientation of the humerus during the tasks, the orientation of the scapula and clavícula were calculated by using a linear regression model of Pascoal.¹⁴ From the position of the landmarks, the local coordinate systems of the trunk, humerus, and forearm were reconstructed according to the guidelines of the International Shoulder Group.¹⁵ This guideline proposes a definition of a joint coordinate system for the shoulder, elbow, wrist, and hand.

Biomechanical Model

Kinematics of the right arm and shoulder and the exerted forces at the hand were used as input for the Delft shoulder and elbow model.¹⁶ The model is an inverse-dynamic model consisting of 31 muscles, divided into 139 muscle elements. For muscles with large attachment sites or complex architectures, more than 1 muscle element is necessary to represent the mechanical effect of the muscle. Joint moments are calculated by inverse dynamics, whereas the joint contact forces are the sum of all forces acting on the bones, thus both external forces and muscle forces.

The input kinematics were the position of the incisura jugularis and the orientations of thorax, humerus, forearm, and wrist. Orientations of the scapula and clavícula were obtained from the regression equations. Further, the 3-dimensional external forces and the moments applied by the hand on the rim were used as input. For the reaching task, the exerted hand force was the force needed to compensate the gravitational force on the bottle. Output variables of the model used in this study were the glenohumeral contact force and muscle forces (fig 2). Muscle forces were calculated on the basis of a minimum stress cost function, and the total force produced by each muscle was obtained by summing the forces of the muscle elements. To enable comparison of muscle forces, muscle forces were expressed as absolute values as well as a percentage of their maximum. The maximum muscle forces were based on a force of 100N/cm^2 of the physiologic cross-sectional area and obtained from Veeger et al.^{17,18}

In our study, the lesion level was not simulated in the model by reducing muscle force in paralyzed muscles; therefore, all

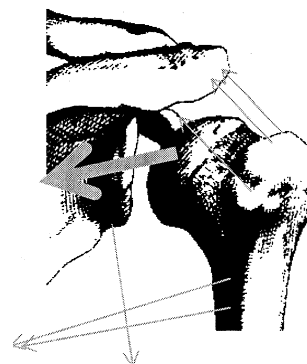


Fig 2. Picture of the shoulder joint that explains the principle of the glenohumeral contact force. In the figure the glenohumeral contact force (from center of rotation of the humeral head to the glenoid) is presented by a thick arrow. The lines of work for 5 hypothetical muscle elements are presented by thin arrows.

the muscles in the model could be used to balance the external moments. Because more force in the remaining muscles would be needed to balance the external moment, the predicted muscle forces would likely be underestimated.

Data Analysis

When the treadmill was at speed and the subject was propelling comfortably, data were collected for 30 seconds. From these 30 seconds of raw data, 5 regular consecutive pushes were selected for data analysis. The push phase was defined as the phase in which the external force was above the level of noise in the recovery phase. The mean and the peak glenohumeral contact forces for the 5 pushes were determined and subsequently averaged over the pushes. The mean and peak glenohumeral contact force was determined for the reaching task. For weight-relief lifting, the mean and peak glenohumeral contact force were calculated for the 3 trials and averaged over these trials.

Of the 31 muscles and muscle parts in the model, 19 muscles (scapulothoracic muscles, scapulohumeral muscles, upper-arm muscles), which are relevant for the load on the shoulder, were selected for analysis. Peak muscle forces were calculated for each task and trial, both absolute and relative to the muscle maximum. Data were subsequently averaged over the trials.

Statistical Analysis

To detect significant differences among the groups, independent *t* tests were applied to the subject characteristics. To compare the peak and mean glenohumeral contact forces among the tasks, a general linear model for repeated measures was used (within-subject factor, task; between-subject factor, group).

To compare the peak absolute and the peak relative muscle forces between the groups, a general linear model for repeated measures was used (within-subject factor, muscle; between-subject factor, group). The level of significance was set at *P* equal to or less than .05 for all statistical tests.

RESULTS

Participants

Four subjects with tetraplegia, 5 able-bodied subjects, and 8 subjects with paraplegia participated in this study. No differences were found for subject characteristics, except that the

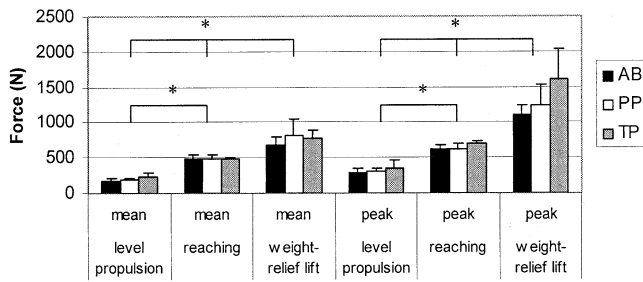


Fig 3. Mean and peak glenohumeral contact forces for the 3 ADL tasks for able-bodied subjects (AB), subjects with paraplegia (PP), and those with tetraplegia (TP). *Significantly different among the tasks.

able-bodied group was younger than the paraplegia group (table 1).

Of the subjects with tetraplegia, 1 subject had no triceps muscle tension at all (manual muscle test [MMT] score, 0), 2 subjects were unable to act against gravity (MMT score, 2), and 1 subject was unable to act against resistance (MMT score, 3). All subjects were able to perform the requested tasks. The kinematic data of one of the subjects with tetraplegia were inaccurate because values were missing for the Optotrak data for the propulsion task. Analysis of the missing value was performed to fill in the glenohumeral contact force, but individual muscle forces were not used for the analysis.

Glenohumeral Contact Forces

For both peak and mean values, performing a lift was accompanied by a significantly ($P < .001$) higher glenohumeral contact force when compared with both wheelchair propulsion and reaching (fig 3). Reaching caused a significantly ($P < .001$) higher peak and mean glenohumeral contact force compared with level wheelchair propulsion. Peak glenohumeral contact forces for weight-relief lifting were 100% higher than reaching and 300% higher than level wheelchair propulsion. For reaching, the contact forces were twice as high as for wheelchair propulsion.

The tetraplegia group had a significantly higher peak glenohumeral contact force during the tasks than the able-bodied and paraplegia groups ($P = .01$)—a difference mostly caused by the 25% higher contact forces during the weight-relief lift. However, no significant interaction effects were found.

Muscle Forces

For the plain model, without adjustments to simulate muscle paralysis, the results of wheelchair propulsion and the reaching task have been presented for the able-bodied and paraplegia groups. For the weight-relief lift, the muscle forces for the tetraplegia group have also been presented. The results reflect the manner in which these subjects perform this task, highlighting the paralysis of certain muscles.

In line with joint compression forces, the range of the relative forces of the analyzed muscles was higher for weight-relief lifting compared with reaching and wheelchair propulsion. For reaching, the range of the relative forces was higher compared with wheelchair propulsion.

Level Wheelchair Propulsion

The muscles estimated to produce the largest peak forces during the push phase were the monoarticular part of the triceps brachii and the deltoideus muscles (fig 4). When expressed as

a percentage of their maximum force, the supraspinatus was the muscle with the highest load ($\pm 12\%$). The relative forces of the other muscles were between 5% and 10%. No overall significant differences were found among the groups.

Reaching

The muscles that produced the largest peak force during the reaching task were the deltoideus, the brachialis, and the trapezius muscles (fig 5). When expressed as a percentage of their maximum force, the brachialis and the deltoideus muscles were the muscles with the most load during this task. The peak relative force of these muscles exceeded 15% of their maximum on average. The range of relative force of the other active muscles was between 5% and 15%. No differences were found among the groups.

Weight-Relief Lifting

The muscle that produced the largest peak force during the lift was the monoarticular part of the triceps brachii muscle, with peak forces over 1000N for the able-bodied subjects and subjects with paraplegia (fig 6). When expressed as a percentage of their maximum force, for the 3 subject groups, the latissimus dorsi, the biceps brachii, and the monoarticular part of the triceps brachii muscles showed relative muscle activity between 20% and 40% of their maximum force.

For weight-relief lifting, the muscle forces for the tetraplegia group have also been provided. However, one must bear in mind that no modifications were made to the model; all muscles in the model could be used to compensate for the external moment and therefore the forces in the nonparalyzed muscles would be underestimated.

Nonetheless, and as can be seen in figure 6, subjects with a high lesion level showed much more brachialis muscle activity than subjects with paraplegia or the able-bodied subjects and much less activity in the monoarticular part of the triceps brachii muscle. Other muscles with higher relative muscle forces for the tetraplegia group were the supraspinatus, the infraspinatus, and the coracobrachialis muscles. Overall, the relative muscle force was significantly higher for the tetraplegia group than the able-bodied group ($P = .022$).

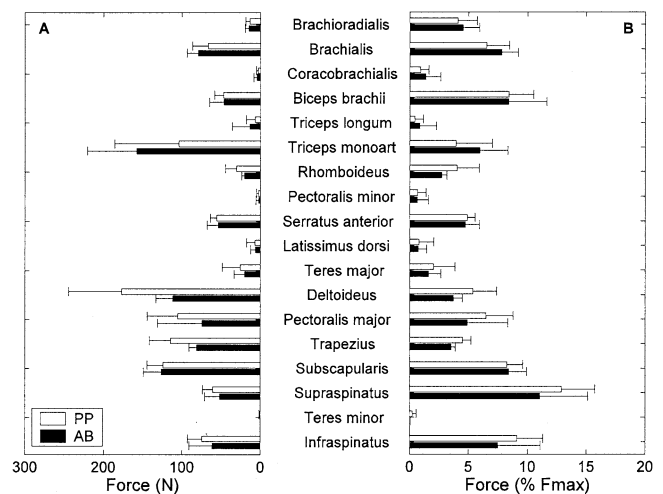


Fig 4. (A) Peak absolute muscle forces and (B) peak relative muscle force during the push phase for both the able-bodied and paraplegia groups. Abbreviations: max, maximum; monoart, monoarticular.

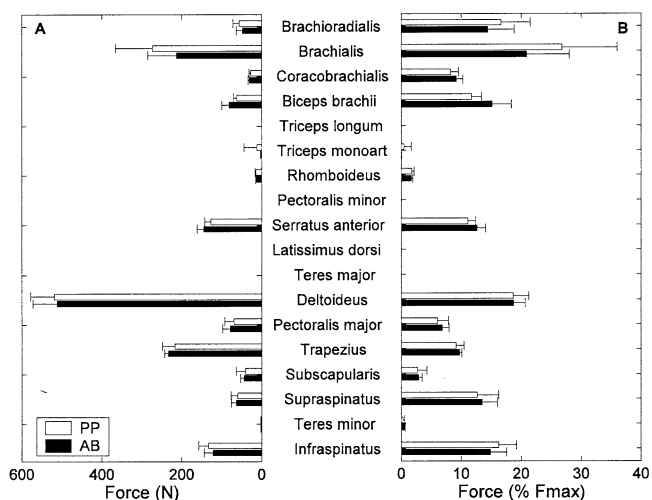


Fig 5. (A) Peak absolute muscle forces and (B) peak relative muscle force for the reaching task with a 1.5-kg mass at 0.5m for both the able-bodied and paraplegia groups.

DISCUSSION

The glenohumeral contact forces were much higher for the weight-relief lifting task than the level wheelchair propulsion and reaching. For subjects with tetraplegia, the contact forces during the weight-relief lift were $\pm 25\%$ higher than for the other groups. Overall, a significant difference among the groups over the tasks was found.

The muscle forces during reaching and level wheelchair propulsion showed no differences between the able-bodied and paraplegia subjects, and both the absolute and relative peak forces were fairly low ($<15\%$). During the weight-relief lift, the peak relative muscle forces were higher (20%–40%) and the tetraplegia group showed much more activity in the rotator cuff and the biceps brachii muscle and less activity in the triceps brachii muscle.

Glenohumeral Contact Forces

The glenohumeral contact forces were low during wheelchair propulsion; however, the external load was low ($4.6 \pm 0.4W$). When the external load is increased by external resistance, increased velocity, or a slope, the contact forces will increase as well. Moreover, as Veeger et al⁵ noted, peak contact forces could be between 800 and 1400N for propelling at 10 and 20W.

Peak values for the reaching task were between 495 and 735N. This task is difficult to compare with other studies because the task was relatively light, with a weight of just 1.5kg and no movement above shoulder level occurred. However, this task is interesting, especially for those with a high lesion, because these subjects need stabilization at the thorax while reaching forward. A much heavier task has been studied by Kuijjer et al,⁷ in which contact forces between 500 and 1500N have been reported for pulling and pushing containers of 40 to 74kg.

The mechanical load has not been calculated yet for weight-relief lifting in SCI. However, in able-bodied subjects, Anglin et al⁶ reported values up to 2075N for coming from sit-to-stand and from stand-to-sit. However, these able-bodied subjects used their legs to lift part of the weight; the glenohumeral contact forces would therefore be higher if only the arms are used.

In a previous study, net joint moments were calculated for the same tasks and the same groups⁴; however, no differences in net shoulder moments were found among subject groups. Because net moments express the mechanical load, but do not reflect the direction of the forces, and the direction of the exerted forces as well as the muscle forces are taken into account in the joint contact forces, the latter can be seen as a better variable to express the mechanical load on a joint.

Muscle Forces

For wheelchair propulsion under the current conditions, the muscle forces expressed relative to their maximum were low, only the forces of the rotator cuff were relatively higher (10%; see fig 4). These findings are in accordance with Veeger,⁵ if one takes into account the difference in intensity with that study (10 and 20W vs 4.6W in our study). In addition, Mulroy et al⁹ found relatively high and prolonged electromyographic activity in the supraspinatus for wheelchair propulsion at 5km/h. The other prime movers for wheelchair propulsion showed muscle forces in accordance with the previously mentioned studies. Our study reports a low relative force for the long head of the triceps and moderate relative forces for the deltoideus and the pectoralis major. However, the distribution of the force over the muscle parts must be taken into account.

When a higher power output (ie, other wheeling conditions, higher velocity, or a suboptimal wheelchair design) is required, the load on the supraspinatus, among others, will be higher, and the risk for shoulder complaints will increase. However, the risk of complaints is not only affected by the peak forces during propulsion but also by the repetition of the task. One should bear in mind that, at a speed of 3km/h, approximately 45 pushes/min are made. Also, at higher propelling speeds, the push time shortens and the force rise time decreases, which has been mentioned as a serious risk factor for injury¹⁹ and has led to research into mechanisms to reduce the peak force.^{20,21}

During the reaching task, those muscles necessary to elevate the arm forward (deltoideus) and to stabilize the arm (infraspinatus, supraspinatus, serratus anterior) were active. Further, the muscles necessary to hold the bottle upright, such as the brachioradialis and biceps brachii muscles, were active. These muscle forces predicted by the model are in accordance with

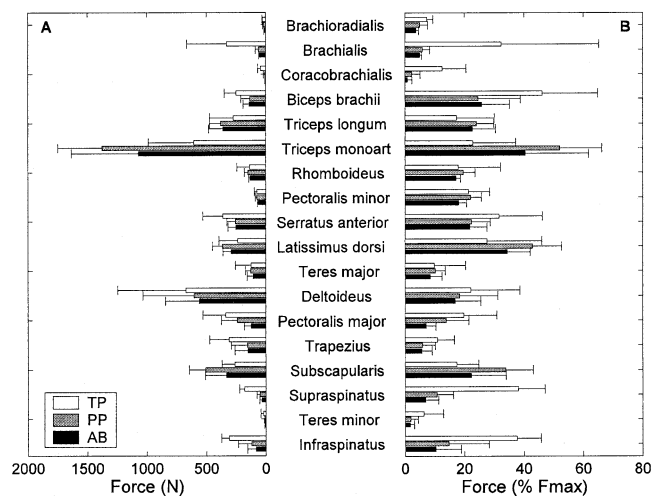


Fig 6. (A) Peak absolute muscle forces and (B) peak relative muscle force for weight-relief lifting for the able-bodied, paraplegia, and tetraplegia groups.

electromyographic activity of these muscles recorded during forward flexion.^{22,23}

For weight-relief lifting, the muscle forces for the tetraplegia group were also provided, because, for this task, the relative muscle forces explain the manner in which these subjects execute the task, making the paralysis of certain muscles visible. In the able-bodied and paraplegia groups, high forces in the triceps were predicted in order to extend the elbow. Force in the pectoralis major and the latissimus dorsi muscles was needed to elevate the trunk. Subjects with tetraplegia performed the lift in a different way because they were unable to use full triceps activity to extend their arms. They did not actively extend the arm but locked the elbow by gravity first (less activity in the triceps), after which they lifted their body weight using the shoulder, which explains the higher muscle activity of the pectoralis major and the deltoideus. With tetraplegia, more force in the rotator cuff was predicted to satisfy the stability constraint of the model, keeping the joint contact force vector in the glenoid cavity.

The predicted muscle forces are in accordance with electromyographic activity of the shoulder muscles reported by Newsam⁸ and Reyes³ and colleagues. These studies reported high muscle activation of the latissimus dorsi, triceps brachii caput longum, and the pectoralis major pars sternalis muscles for subjects with paraplegia. Subjects with a high-level SCI showed higher activity of the deltoideus pars clavicularis and the infraspinatus muscles.

It is difficult to ascertain which activity is the most taxing in terms of the development of overuse injuries. Wheelchair propulsion, on the one hand, is a highly repetitive task and might therefore lead to more strain than a weight-relief lift if one takes into account the combined effect of peak force and frequency. On the other hand, in weight-relief lifting, the duration of the activity itself is longer compared with wheelchair propulsion in which the pushes are short and the risk of overuse might be much higher. Prevention of overuse injuries should therefore focus on reducing the load of wheelchair use on the upper extremity by improving the material, the environmental conditions, and the technique. The latter applies to both propulsion and performance during ADLs such as lifting. Also, prevention of overuse injuries will benefit from training of the musculoskeletal system, focusing on overall force as well as force balance between muscle groups.²⁴

It is advisable to include strength training of the rotator cuff in the rehabilitation program of patients with SCI, especially for those who have tetraplegia. In addition, special attention to the status of the triceps is warranted because this muscle plays an essential role in tasks related to weight relief.

Methodologic Issues

Although the tasks had to be performed in an experimental instrumented wheelchair that differed from subjects' own wheelchairs, subjects had no problem handling the wheelchair after they had become accustomed to it. In addition, the task constraint—that subjects were required to use the handrims to lift themselves—may have influenced the position and orientation of trunk and arms and thus could have had an influence on the direction of the exerted forces. The different performance with the left and the right arms may have led to a mild asymmetry but was deemed necessary to prevent as much as possible local discomfort of task performance.

A possible training effect may have affected the results because subjects had to repeat the weight-relief task in a different manner than they were used to. An effect related to fatigue of the subjects may have occurred as well; however,

previous statistical analysis showed that there were no differences between the trials.

The model we used was not modified to mimic subjects with a high SCI because we expected the difference in task performance to manifest itself. All muscles were included, and no muscle force reduction related to paralysis was implemented. The muscles in the model can be simulated as (partially) paralyzed simply by reducing the maximum relative force of muscles. The model uses a minimum stress cost function to calculate muscle forces; however, when the model attributes stress to, in reality, paralyzed muscles, the actual stress in the other muscles must be higher than their attributed value. Especially for the tetraplegia subjects, forces in the remaining muscles can be expected to be higher and therefore increase the risk of soft-tissue damage. More activity from the remaining muscles may be needed to stabilize the shoulder joint as well as to perform the task itself, thus resulting in less efficient and higher forces on the joint. Because the glenohumeral contact force is the sum of the external force and the muscle forces, higher muscle force will cause a higher glenohumeral contact force. Therefore, the differences found in this study between the subjects with a high lesion and the subjects with a low lesion or able-bodied subjects will be larger when these modifications in the model take place.

However, considering the relative force, one must take into account that the physiologic cross-sectional areas (CSAs) of the model's muscles were measured in older specimens and the tasks were performed by younger subjects. Most subjects with a lesion have well-trained shoulder muscles and therefore the relative muscle stress for these muscles will be lower. To approximate the reality even more, it may not only be necessary to simulate paralysis but also to increase the maximum muscles stress or to enlarge the physiologic CSAs of some muscles in the model as well.

CONCLUSIONS

A significantly higher glenohumeral contact force was found for the subjects with tetraplegia compared with the able-bodied subjects and subjects with paraplegia, the difference mostly attributable to the higher values for the weight-relief lift in the tetraplegia group. For this ADL task, the load on the glenohumeral joint was twice as high as the load for reaching.

Without taking the paralysis into account, more muscle force was estimated for the subjects with tetraplegia during weight-relief lifting. Modifications to the model would likely increase the forces produced by the remaining muscles.

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Suppliers

- a. Model 3446; Enraf-Nonius BV, Röntgenweg 1, PO Box 810, 2600 AV Delft, The Netherlands.
- b. Sunrise Medical Benlux, Pascalbaan 3, 3439 MP Nieuwegein, The Netherlands.
- c. Type MC3A-6-1000; Advanced Mechanical Technology Inc, 176 Waltham St, Watertown, MA 02472.
- d. Porti; Twente Medical Systems, H. ter Kuilestraat 181, 7547 SK Enschede, The Netherlands.
- e. Northern Digital Inc, 103 Randall Dr, Waterloo, ON N2V 1C5, Canada.
- f. Faculty of Human Movement Sciences, Vrije Universiteit, Van der Boechorststr 9, 1081 BT Amsterdam, The Netherlands.