Upper extremity load during wheelchair-related tasks in subjects with a spinal cord injury

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Chapter 1

General Introduction

Introduction

Overload injuries of the upper extremity musculoskeletal system are a common problem [168]. For subjects with a spinal cord injury (SCI) the problem is even more serious since they depend on their upper extremities for mobility and are more prone to these injuries. Next to daily wheelchair propulsion, wheelchair-related activities of daily living are generally seen as a risk factor for upper extremity overload. Wheelchair propulsion is considered to be a risk factor because of the repetitiveness of the movement, while the peak loads in tasks such as lifting and transfers are expected to be damaging. These tasks have hardly been studied in terms of mechanical strain. Besides wheelchair propulsion, the focus of this thesis will be on the load and the task characteristics of lifting and reaching and the effect of the SCI on the mechanical strain. This will help to increase the understanding of the underlying mechanisms related to the development of overload injuries of the upper body musculoskeletal system.

In this general introduction, an overview will first be given of the context of this study. The knowledge on the load of wheelchair-related tasks thus far and the influence of lesion level on the strain will then be presented. Furthermore the historical framework of this thesis will be described. Finally the aims and the outline of this thesis will be specified.

Spinal cord injury

Spinal cord injury (SCI) is a disruption of the spinal cord, resulting in complete or incomplete muscle paralysis, loss of sensation and dysfunction of the autonomic system. The severity of the consequences of a SCI is dependent on the level and the extent of the lesion. Lower thoracic and lumbar lesions cause paralysis of the legs, whereas high thoracic lesions also result in paralysis of the trunk muscles (Figure 1.1). Cervical lesions (tetraplegia) result in paralysis of the legs and trunk as well as (partial) paralysis of the arms. The incidence of SCI in the Netherlands is estimated to be around 12 cases per million persons per year [161]. No reliable number is available for the prevalence of SCI in the Netherlands, but this has been estimated at

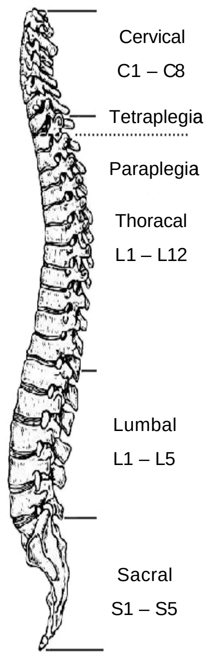


Figure 1.1: Distinction between tetraplegia and paraplegia and their accompanying segment levels.

12.000 persons. In the Netherlands about half of the SCI cases have a traumatic cause (falls, motor vehicle accidents) [161]. According to Post et al. [124], about 80% of the subjects with SCI will remain at least partly wheelchair-dependent.

Due to improved medical treatment, over the last five decades life expectancy of people with a SCI has increased from well below the general population, to more or less equal to the general population today [185]. The rehabilitation of SCI today focuses on the improvement of remaining capacities and functional integration into society. Restoring the level of mobility is expected to positively influence daily activities and functioning which may in turn enhance freedom of movement, social range of action and reintegration and probably therefore also the quality of life of the individual [36, 81, 113].

As a consequence of the increased life expectancy, people with a SCI more often experience secondary complications like increased risk for cardiovascular diseases [42], kidney insufficiency, bladder problems and musculoskeletal complaints [112]. A sedentary lifestyle is generally seen as a metabolic risk factor [44].

Clearly, due to paralysis and wheelchair-confinement, persons with a SCI are prone to a sedentary lifestyle. Increasing the level of physical activity is likely to increase the general health of people with a SCI [37, 66]. A physically active lifestyle not only reduces the risk of cardiovascular diseases but also of obesity and of decubitis by stimulation of blood flow in the legs. For subjects with a SCI it is even suggested that exercise may result in increases in pathways promoting neuronal health and recovery in the SCI population [120]. The downside of increased physical activity in subjects with a SCI is that it is related to upper-body work, and as such may lead to overload on the musculoskeletal system of the upper extremity. Prevalence rates of 50 to 70% for upper extremity complaints, especially at the shoulder and the wrist, in the wheelchair user population have indicated that overload injuries of the musculoskeletal system are a serious long-term problem [7, 19, 110, 119]. Not only daily wheelchair users develop complaints but also wheelchair athletes who probably have a good physical capacity [30, 47]. Complaints could develop from both local overload of muscles but also from imbalance of the muscles around the shoulder. In both cases, these complaints often result in a vicious circle of pain, reduced physical activity, reduced work capacity and increased strain.

The shoulder: mobility vs. stability

Since long it has been acknowledged that the shoulder joint is especially at risk for overuse injuries due to its complex structure and its limited muscle mass [25].

The shoulder mechanism is a complex system which consists of the clavicula, scapula, humerus and the thorax. A movement of the shoulder is therefore actually a movement in four joints: the sternoclavicular, acromioclavicular and glenohumeral joints and the scapulothoracic gliding plane.

The arm is very mobile via the connection to the scapula and the structure of the glenohumeral joint. The glenoid, the saucer of this joint, is relatively small compared to the head of the humerus, therefore the joint requires continuous muscle activity to stabilize it. Stabilization of the glenohumeral joint is the main function of the four muscles of the rotator cuff, while other muscles are prime movers of the arm due to their large moment arm. At any rate, the mobility is at the expense of the stability of the joint [25] and therefore makes it especially vulnerable for the development of related instability complaints [105].

Mechanical load

Mechanical loading on a system can be studied by external and internal variables. Forces and net moments are external variables and as such relate to the occurring internal forces. Although internally, strain can in principle be predicted from external loads [129, 177], the external variables are likely not sufficiently informative about the strain on the muscles and joints when external forces are widely different from the conditions studied in the aforementioned publications, or when the necessity for maintaining joint stability might play an important role.

Compared to the external load, the joint reaction force is a more accurate indicator for strain at the shoulder joint since this parameter reflects the effect of the external forces, the muscle forces around the joint to compensate the external moment and additional muscle force required for joint stability. The joint reaction force therefore reflects both the possibly damaging strain to the joint surface under extreme loading, as well as the strain that may lead to soft tissue damage due to high muscle forces.

In any case, to determine the mechanical load on the upper extremity, external forces need to be measured. One option is to use the custom-made wheelchair ergometer [111],

Figure 1.2: Subject in the instrumented wheelchair during level wheelchair propulsion at 3 km·hr^{-1}.

which has been used in various wheelchair studies [35, 40, 69, 139, 170, 180]. However, to approach reality even closer, an instrumented wheel was designed (Figure 1.2). This wheel can be mounted on a regular wheelchair and therefore propulsion on a treadmill, a track or just in daily life can be measured. But even more importantly, also the wheelchair-related tasks like negotiating a curb or performing a transfer or a lift can now be measured. Therefore, in this thesis the external hand forces, which were applied to the handrim, were measured with this custom-built instrumented wheel.

With the measured external hand forces and hand moments, external moments around the joints can be calculated by inverse dynamics. For accurate calculation, in combination with kinematics, information about the mass and the length of the body segments is necessary. Additionally, to calculate the internal load on the structures of the upper extremity a biomechanical model based on 3D anatomical information is necessary.

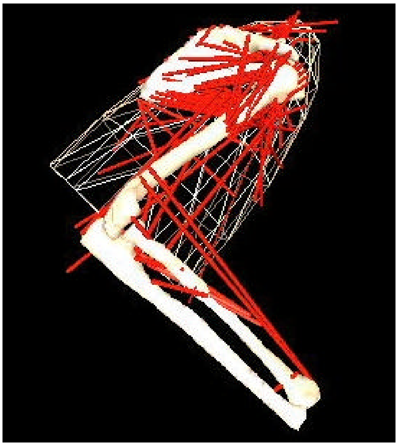


Figure 1.3: Graphical illustration of the Delft Shoulder and Elbow model during wheelchair propulsion. The lines represent muscle elements.

The Delft Shoulder and Elbow Model [163, 165, 179] is a finite element musculoskeletal model consisting of 31 muscles divided into 139 muscle elements (Figure 1.3). The external forces and the rotations of the thorax, clavicle, scapula, humerus and forearm are the input variables for the model. The model can be used to calculate the load on the structures of the shoulder and elbow. In this thesis, next to net moments, the by the model calculated individual muscle forces and joint reaction forces will be used.

What is known about complaints?

In previous studies it has already been shown that the physiological strain on subjects with a SCI is high during wheelchair-related activities [70, 71]. Especially during transfer tasks periods of peak physiological strain of over 60% of the heart rate reserve frequently occurred [76]. Unfortunately, there is limited information about the mechanical load of these tasks and their contribution to upper extremity complaints. Tasks like wheelchair propulsion [166, 177], maneuvering (negotiating a curb or a slope), weight-relief lifting [60] and transfers [183] lead to a high mechanical load. Some information is available about the mechanical strain on the upper extremity in other populations and tasks. For example Anglin et al. [4] measured transfers from sit to stand and Kuijer et al. [88] calculated net shoulder moments for pulling and pushing a refuse container.

Due to the functional changes in the neuromuscular system and consequently the disturbed co-ordination pattern and compensatory muscle activity in the remaining muscle groups after muscle paresis, subjects with a high-level SCI are expected to show notably higher mechanical strain compared to subjects with a low-level SCI, as seen in physiological studies [37, 69, 71]. To study the effect of lesion level, the mechanical strain of subjects with a high-level lesion is compared to subjects with a low-level lesion and able-bodied subjects.

During the rehabilitation process and during daily life, many tasks have to be performed that appear to lead to high mechanical strain on the upper extremities. For damage to occur to the joints and the soft tissues around the joints, not only the load is important but also the physical capacity of the person. Shortly after the spinal cord injury, most subjects do not have well-trained upper extremity musculature. As a consequence, the tasks are highly demanding in absolute terms as well as in relation to the physical capacity of the subjects.

Most often the first indication of an overload injury is pain. Although pain may initially not limit the wheelchair user to perform activities independently, it may have functional costs such as fatigue and discomfort [33]. Eventually, overload injuries are likely to result in reduction in performance of daily life activities, which in turn will reduce physical capacity and increase the risk of subsequent overloading [141]; a vicious circle.

To understand the underlying mechanisms for the origin of shoulder problems a conceptual model is necessary. Many factors play a role in the eventual damage; not only the load and other task characteristics, but especially in subjects with SCI, also the physical capacity. Variables of the load are the intensity, frequency, direction of force and time of exposure. One possible explanation for shoulder damage in the wheelchair-dependent population could be that as a result of peak loading during wheelchair-related tasks, damage occurs (Figure 1.4). Subsequent, continuous submaximal loading during wheelchair propulsion could lead to incomplete recovery and cumulative damage as a result.

Historic framework of this thesis

This study is part of the national multidisciplinary research program ‘Physical strain, work capacity and mechanisms of restoration of mobility in the rehabilitation of individuals with SCI’. The program is performed in collaboration with eight rehabilitation centers and five research groups. The complex adaptations in the organ systems that are fundamental to restoration of mobility in SCI rehabilitation are studied within the context of biomedical and psycho-medical issues of mobility restoration. Issues concerning the level of function as well as activities and participation are addressed.

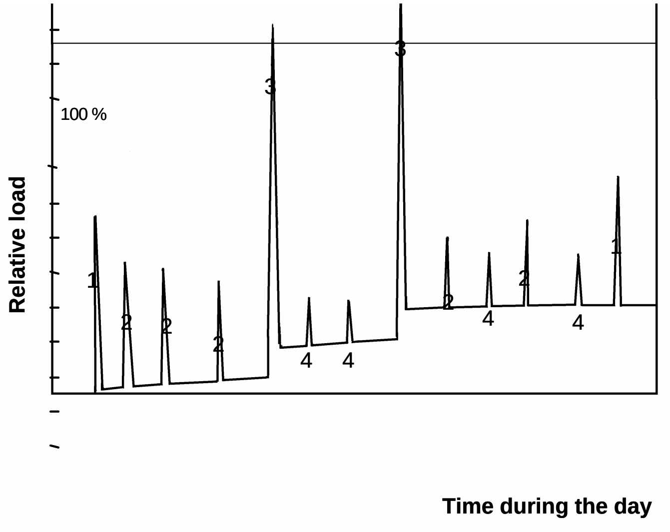


Figure 1.4: Hypothetical load profile of a day of a person with a spinal cord injury. Different wheelchair-related tasks cause a high relative load and even damage. Continuous wheelchair propulsion increases the offset of the load.

1= transfer in/out bed, 2 = lift, 3= transfer in/out car, 4 = reach

In a longitudinal cohort study, the physical capacity [36], pain, performance of wheelchair skills [80] and functional independence [125] are studied among others in the light of the restoration of mobility. In this thesis the slant is the musculoskeletal system, adaptations of the cardiovascular system [39], daily activity [126] and participation [12, 123] are the topics of other projects.

Since 1983, the research program ‘Stress and Strain in the upper extremity’ of the Faculty of Human Movement Sciences in Amsterdam looked into wheelchair propulsion. It was found that the mechanical efficiency of wheelchair propulsion was low (<11%) [172], probably partly due to the stability requirements in the shoulder. To further explore the relation between the mechanical and physiological load on the shoulder, a three-dimensional biomechanical model of the upper extremity was developed in collaboration with Delft University of Technology (The Delft Shoulder and Elbow Model [163, 165]). Since 1987 both parties have been working together intensively in shoulder research [167, 179], while shoulder problems in wheelchair propulsion have been an important point of research [166, 177, 180]. This research program was basis for the foundation of the Dutch Shoulder Group, now part of the International Shoulder Group (www.internationalshouldergroup.org).

Outline of this thesis

The aim of this thesis is to study the underlying mechanisms for the development of overload injuries of the upper body musculoskeletal system in subjects with a SCI.

Pain can be assumed to be an early indicator of overuse, which could show early overuse and damage processes. In chapter 2, the development of upper extremity musculoskeletal pain during and after rehabilitation and its relationship with lesion level, personal characteristics and functionality is studied. This study was performed to find out at what time during the rehabilitation complaints occur and to identify subject characteristics that make persons at a higher risk to develop upper extremity pain after the rehabilitation process.

In chapter 3, the mechanical load on the upper extremity during different wheelchair activities is determined by calculating net moments around the glenohumeral and elbow joints. Net moments are common in biomechanical research and are relatively easy to calculate. This study gives an impression of the load on the shoulder and elbow during different wheelchair-related tasks and the differences between subjects with paraplegia and subjects with tetraplegia.

Another indicator of mechanical load at the shoulder joint is calculated in chapter 4. The reaction force on the joint surface reflects both the compression force and the muscle forces around the joint and therefore might be a more informative predictor of joint and soft tissue damage. In this chapter, the glenohumeral reaction force during three wheelchair-related tasks and for three different subject groups is studied.

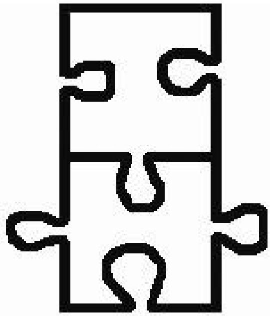
Variation in motor lesion level among individuals with tetraplegia will affect the innervations of the shoulder and arm-hand muscles and therefore has an effect on the muscle load. The effect of lesion level, and specifically muscle paresis, on the mechanical load is studied in a simulation study in chapter 5, in which the shoulder-elbow model is modified to represent different levels of cervical spinal cord injuries.

Impingement is a common diagnosis for shoulder pain. During the performance of a lift an extreme downward pull on the body is present, which is suggested to create a reduction of the sub-acromial space, which might lead to impingement. In chapter 6, the aim is to determine the risk for impingement during the performance of a weight-relief lift.

Finally, in chapter 7, the main findings and conclusions of this thesis are summarized and discussed, and the implications of this thesis for future research and practice are discussed.

Chapter 2

Upper extremity musculoskeletal pain during and after rehabilitation in wheelchair using persons with a spinal cord injury



Based on
S van Drongelen, S de Groot, HEJ Veeger, ELD Angenot, AJ Dallmeijer
MWM Post and LHV van der Woude
Spinal Cord, accepted for publication

Abstract

Study design: Prospective cohort study.

Objectives: To study upper extremity musculoskeletal pain during and after rehabilitation in wheelchair using subjects with a spinal cord injury and its relation with lesion characteristics, muscle strength and functional outcome.

Setting: Eight rehabilitation centers with a SCI unit in the Netherlands.

Methods: Using a questionnaire, number, frequency and seriousness of musculoskeletal pain complaints of the upper extremity were measured. A pain score for the wrist, elbow and shoulder joints was calculated by multiplying the seriousness by the frequency of pain complaints. An overall score was obtained by adding the scores of the three joints of both upper extremities. Muscle strength was determined by manual muscle testing. The motor score of the Functional Independence Measure provided a functional outcome. All outcomes were obtained at four test occasions during and one year after rehabilitation.

Results: Upper extremity pain and shoulder pain decreased over time (30%) during the latter part of inpatient rehabilitation ($P<0.001$). Subjects with tetraplegia showed more musculoskeletal pain than subjects with paraplegia ($P<0.001$). Upper extremity pain and shoulder pain were significantly inversely related to functional outcome ($P<0.001$). Muscle strength was significantly inversely related to shoulder pain ($P<0.001$). Musculoskeletal pain at the beginning of rehabilitation and BMI were strong predictors for pain one year after inpatient rehabilitation ($P<0.001$).

Conclusions: Subjects with tetraplegia are at a higher risk for upper extremity musculoskeletal pain and for shoulder pain than subjects with paraplegia. Higher muscle strength and higher functional outcome are related to fewer upper extremity complaints.

Introduction

Based on epidemiological studies it seems evident that manual wheelchair propulsion and wheelchair-related daily life activities cause a heavy load on the upper extremities, especially for persons with cervical spinal cord injury (SCI) [31, 148]. Other suggested risk factors for the development of shoulder pain are the duration of injury, age (i.e. older people have a higher risk than younger people), higher body mass index (BMI) [17, 90, 119] and wheelchair propulsion style [16]. However, research has been limited in terms of methodology and power. Most of the previous studies were cross-sectional and/or retrospective studies and focused on subjects who had a longer duration of the injury. Some of the studies collected data on pain by questionnaires [38, 141], while other studies performed an additional physical examination [17, 119, 148]. Only few studies looked at complaints which originated during the inpatient rehabilitation [58, 147, 149]. Musculoskeletal complaints during inpatient rehabilitation can be crucial to the progress and the duration of the rehabilitation [58]. Further, there is not much evidence-based information about the relationship between upper extremity complaints and the functionality of subjects with a SCI.

In the current study the course of upper extremity musculoskeletal pain over time is described and analyzed in association with the course of muscle strength and functional outcome during and one year after rehabilitation. Musculoskeletal pain and functionality are expected to be related since persons with a better physical condition have fewer complaints or less intensive pain [47].

Knowledge about the course and prognostic determinants of upper extremity complaints is not only of importance to predict the condition of the persons with SCI at the end of the rehabilitation period, but also for treatment and prevention of pain. Timing of more or less straining training periods could be beneficial to the occurrence and course of pain complaints [33].

The aim of this study was to investigate the course of upper extremity musculoskeletal pain and its relationship with lesion characteristics, muscle strength and functional outcome of subjects with a SCI, with special attention to the shoulder joint. Furthermore, the most important predictors for upper extremity musculoskeletal pain one year after the rehabilitation were investigated.

It is expected that subjects with tetraplegia have more upper extremity musculoskeletal pain and that subjects with higher muscle strength and higher functional outcome develop fewer complaints.

Methods

Subjects

The present study was part of the Dutch research program 'Physical Strain, Work Capacity, and Mechanisms of Restoration of Mobility in the Rehabilitation of Persons with Spinal Cord Injuries'. Persons with an acute SCI were followed during their inpatient rehabilitation and until one year after discharge. Subjects were measured four times: at the start of active rehabilitation, defined as being able to sit in a wheelchair for 3-4 hours (t_1), three months after t_1 (t_2), at the time of discharge from inpatient rehabilitation (t_3) and a year after discharge (t_4). In eight Dutch rehabilitation centers, specialized in the rehabilitation of persons with SCI, a trained local research assistant conducted the measurements according to a standardized protocol.

Subjects were asked to enter the program if they had an acute SCI, were between 18 and 65 years of age and were classified as A, B, C or D on the American Spinal Injury Association (ASIA) Impairment Scale [99]. Exclusion criteria were progressive diseases, psychiatric problems and insufficient knowledge of the Dutch language. Subjects who were able to walk were excluded from this study, but a small group of subjects with a high lesion level who eventually are using a power wheelchair were included.

For some subjects the length of stay in inpatient rehabilitation was no longer than three months. In these cases, their second measurement was performed at the time of discharge. Data of those subjects were considered as measurements at t_3, meaning that of those subjects no data were available at t_2.

After they had been given information about the testing procedures, the subjects gave their written, informed consent. All tests and protocols were approved by the Medical Ethics Committee of Rehabilitation Centre Hoensbroek, The Netherlands.

Procedure

Lesion and personal characteristics

At each test occasion, the lesion characteristics (level and completeness) were assessed by a physician according to the International Standards for Neurological Classification of Spinal Cord Injury [99]. Also the body mass index (BMI) was determined (body mass·height^{-2}).

Pain

At the four test occasions (t_1-t_4) the participants were asked in a separate standardized questionnaire if they experienced pain on the joints or muscles of the upper extremity joints, i.e. the wrist, elbow and shoulder of both arms. Musculoskeletal pain in each joint was scored as 0 when no pain was present or as 1 when pain was present. Subjects were also asked about the seriousness of pain complaints and about the frequency of pain occurrence. The seriousness was measured subjectively on a Likert scale of 1-5 ranging from very mild to very severe. Frequency was scored as 1 if pain occurred once a week or less, as 2 if pain occurred two or three times a week and as 3 when pain occurred more than three times a week.

An overall upper extremity pain score was obtained by multiplying the seriousness by the frequency of complaints when pain was present. The scores of the three joints of both upper extremities were added together to obtain an upper extremity pain score, ranging from 0 to 90. The pain score was also analyzed for the shoulder joints separately (score ranging from 0 to 30).

Another section of the survey asked for other (among others neurogenic) pain complaints with the specific note about other pain complaints than musculoskeletal pain.

Manual muscle strength

To assess the strength of six muscle groups of the upper extremities, standardized manual muscle tests (MMT) were performed for the wrist extensors, elbow flexors and extensors, shoulder internal and external rotators, and shoulder abductors. The MMT for each muscle group was performed in a standardized position [77]. Muscle force was measured subjectively by the research assistant on a scale of 0-5 as follows: 0) no muscle contraction, 1) palpable or visible muscle contraction, 2) active movement through full range of motion (ROM) with gravity eliminated, 3) active movement through full ROM against gravity, 4) active movement through full ROM against resistance, 5) normal muscular strength. The muscle group scores of the right and left upper extremities were added together to obtain an overall MMT score, ranging from 0 – 60.

FIM motor score

The motor score of the Functional Independence Measure (FIM) [57] was used to measure the level of independence in activities of daily living. The Dutch version of the FIM was used for this study [125]. The items were scored on a seven-point scale (1 (completely dependent) – 7 (independent)) and the 13 items were added together to obtain an overall FIM motor score, ranging from 13 – 91.

Statistical analyses

Only those subjects who took part in more than one test occasion were included in the analyses. Furthermore, subjects who were community walkers (persons who walk indoors and outdoors possibly with aid and use a wheelchair only for long distances outdoors) at the last test occasion were removed from the database to prevent a positive selection.

To determine whether pain in the upper extremities increased or decreased during and after rehabilitation, the multi-level modeling program MlwiN [131, 160] was used. In the longitudinal data set of this study, the hierarchy in the data is the repeated measurement 'test occasion (t_1-t_4)' (level 1), which is grouped within the individual participants (level 2), who are grouped in the rehabilitation centers (level 3). Overall pain in the upper extremity (ranging from 0 - 90) and shoulder pain (0 - 30) were the outcome variables of this multilevel regression analysis. Since pain was not normally distributed, this count variable was analyzed with a Poisson model. Time was modeled with three categorical dummy variables, with t_1 as the reference to t_2, t_3 and t_4, e.g. the regression coefficient of t_1 – t_4 indicated the difference in pain score between t_1 and t_4. A priori, time was also modeled with t_3 as the reference to t_1, t_2 and t_4, to examine when significant changes over the time occurred.

To investigate changes in upper extremity pain over time, only the time dummies were included in the basic model (model 1).

To investigate the effect of lesion characteristics, lesion level (paraplegia = 1; tetraplegia = 0) and completeness (incomplete = 0; complete = 1), were added to the basic model and a backward regression technique was used (model 2). Personal characteristics (age, gender, BMI) were added to the final model 2 to investigate their effect on the relation between pain complaints and lesion level or completeness. Variables that changed the regression coefficients of lesion level or completeness by at least ten percent were identified as confounders and were corrected for in the final analysis. Interaction terms between lesion level and time were also added to check for possible effect modifications ($P<0.05$). To investigate the effect of muscle strength and motor FIM on pain, these variables were added separately to model 2. These parameters were not put together in the same model since they could have an effect on each other.

To find out which parameters were the most important predictors for pain, a prediction model was set-up with upper extremity pain or shoulder pain at t_4 as the outcome variable. Lesion characteristics, personal characteristics, muscle strength and functional outcome at t_1 were used in separate analyses to predict the pain at t_4. Pain at t_1 was a variable that was always added first to the model to account for pain complaints at t_4. First the independent variables were included separately, and if they showed a p-value below 0.1 the variables were added to the Poisson model. Second, a backward elimination technique was used until only significant determinants remained ($P<0.05$).

Results

Descriptive

A total of 169 subjects participated in this study. The t_1 measurement was performed by 169 subjects. At t_2 to t_4 the number of participants was 133, 161 and 116 respectively (Table 2.1).

Table 2.1: Group characteristics

		t_1 (n=169)	t_2 (n=133)	t_3 (n=161)	t_4 (n=116)
Lesion	*% paraplegics*	*59.2*	*52.6*	*60.2*	*66.4*
Completeness	*% complete*	*74.9*	*69.2*	*70.3*	*76.3*
Gender	*% male*	*74.0*	*75.2*	*74.5*	*73.3*
BMI ($kg{\cdot}m^{-2}$)	*Mean ± SD*	*22.8 ± 3.8*	*23.1 ± 3.9*	*23.5 ± 4.0*	*24.6 ± 4.6*

The mean number of days (SD) between t_1 and t_3 was 230 (136) for all subjects, 179 (90) days for those with paraplegia (PP) and 309 (157) days for those with tetraplegia (TP).

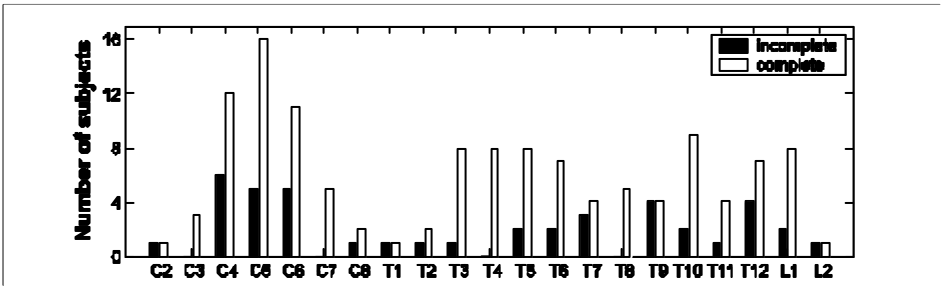


Figure 2.1: Distribution of the highest (one-sided) lesion levels and completeness of the lesion of all subjects at the start of the active rehabilitation (t_1).

The neurological level of the (incomplete) injuries ranged from C2 to S5 (Figure 2.1). Of the 169 subjects 20 subjects eventually used a power wheelchair next to their manual wheelchair, whereas seven subjects used only a power wheelchair.

For the parameters muscle strength and motor FIM, the number of subjects that took part in the measurements was smaller and varied over time and over tests. The FIM was scored for almost all subjects while about 20 subjects did not perform the manual muscle tests. The mean values (SD) of the manual muscle score for the time intervals t_1-t_4 were respectively 49.4 (15.1), 50.8 (14.1), 52.9 (12.6) and 55.1 (10.4). The mean values (SD) for the FIM motor score were 39.0 (19.0), 51.0 (22.5), 61.0 (21.8) and 63.4 (21.2).

Table 2.2: Descriptive pain data

	Hand & wrist		Elbow		Shoulder	
	Left	Right	Left	Right	Left	Right
t_1 : PP = 100 TP = 69						
Complaints	6 / 20	4 / 20	9 / 13	6 / 13	28 / 38	28 / 36
Seriousness	2.88 (1.21)	2.71 (1.30)	2.91 (1.07)	2.89 (1.37)	2.74 (1.11)	2.66 (1.04)
Frequency	2.69 (0.62)	2.79 (0.51)	2.95 (0.21)	3.00 (0.00)	2.61 (0.70)	2.58 (0.73)
t_2 : PP = 70 TP = 63						
Complaints	5 / 16	1 / 15	4 / 6	1 / 5	23 / 32	22 / 37
Seriousness	2.81 (1.12)	3.25 (1.13)	2.70 (1.06)	3.33 (1.03)	2.93 (1.24)	2.76 (1.22)
Frequency	2.70 (0.57)	2.75 (0.58)	2.80 (0.42)	2.50 (0.84)	2.57 (0.75)	2.45 (0.78)
t_3 : PP = 97 TP = 64						
Complaints	8 / 9	6 / 11	4 / 5	3 / 8	21 / 24	21 / 27
Seriousness	3.00 (1.19)	2.41 (0.94)	3.00 (1.41)	2.91 (1.22)	2.47 (1.04)	2.55 (1.12)
Frequency	2.72 (0.57)	2.76 (0.56)	2.91 (0.30)	3.00 (0.00)	2.35 (0.87)	2.43 (0.79)
t_4 : PP = 77 TP = 39						
Complaints	6 / 8	6 / 8	3 / 6	4 / 10	17 / 13	20 / 17
Seriousness	2.53 (1.26)	2.87 (0.99)	2.80 (1.40)	2.67 (1.29)	2.53 (1.16)	2.28 (0.97)
Frequency	2.40 (0.83)	2.50 (0.85)	2.70 (0.67)	2.33 (0.90)	2.31 (0.82)	2.32 (0.81)

The number of subjects with pain complaints, the seriousness (1-5) and frequency (1-3) of pain per week of the upper extremity joints for subjects with paraplegia and tetraplegia at the four test occasions. Complaints are presented by the number of subjects with paraplegia and tetraplegia. Values for the seriousness and frequency are mean and standard deviation.

Musculoskeletal pain was reported most frequently for the left and right shoulder (Table 2.2). In addition, subjects with TP reported more complaints in the hand, wrist and elbow when compared to subjects with PP. Pain in all joints was experienced as mild to moderate and occurred three or more times a week. The seriousness and the frequency of the shoulder pain complaints seemed to decrease over time.

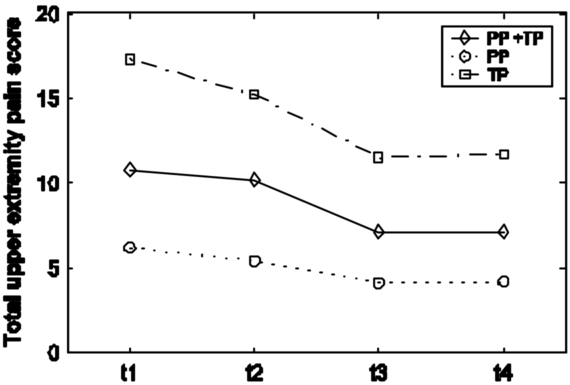
Course of upper extremity musculoskeletal pain over time and the relationship with lesion level, muscle strength and functional outcome

The basic model, to describe the course of upper extremity musculoskeletal pain, showed a significant reduction in upper extremity complaints (P<0.001) between t_2 and t_3 of 30% (Figure 2.2). There were no significant changes between t_1 and t_2 or t_3 and t_4.

For subjects with TP the risk of upper extremity complaints was a factor 2.8 higher than for subjects with PP (P<0.001) (Figure 2.2). The interaction term between time and lesion level showed that subjects with TP showed a stronger decrease in upper extremity musculoskeletal pain than subjects with PP over time intervals t_1-t_3 and t_1-t_4.

None of the personal characteristics were found to be a confounder for the relationship between pain complaints and lesion level over time.

Figure 2.2: Effect of lesion level on upper extremity pain. Estimation of the total upper extremity pain score for all subjects (PP + TP), based on model 1 and separately for subjects with paraplegia (PP) and for subjects with tetraplegia (TP), based on model 2.



When the parameter ‘functional outcome’ was added to the model including the time dummies and lesion level, the FIM score was a significant explanatory variable for upper extremity musculoskeletal pain. A 10 point increase of the FIM motor score between subjects or within subjects over a period of time is associated with an 11% decrease in upper extremity pain (*P*<0.001).

Muscle strength was not found to be an explanatory variable of upper extremity musculoskeletal pain.

Shoulder pain

The basic model, used to describe the course of shoulder pain, showed that there was a significant increase in total shoulder pain between t_1 and t_2 (*P*=0.018) and a significant decrease between t_2 and t_3 (*P*<0.001).

Lesion level had a significant effect on shoulder complaints (*P*<0.001) and the personal characteristics had no confounding effect. The risk of shoulder complaints was a factor 2.2 higher for subjects with TP than for subjects with PP. When lesion level was included in the model the increase in shoulder pain between t_1 and t_2 was not significant anymore.

The separate analyses of muscle strength and functional outcome showed that both the MMT score (*P*<0.001) and FIM motor score (*P*<0.001) were explanatory variables of shoulder pain complaints. A 10 point increase of the FIM motor score resulted to 14% less shoulder pain and a 10 point increase of the MMT score resulted to 12% less shoulder pain (Figure 2.3).

The interaction terms between time and lesion showed again that subjects with TP showed a stronger decrease in shoulder pain than subjects with PP over the time intervals t_1-t_3 and t_1-t_4.

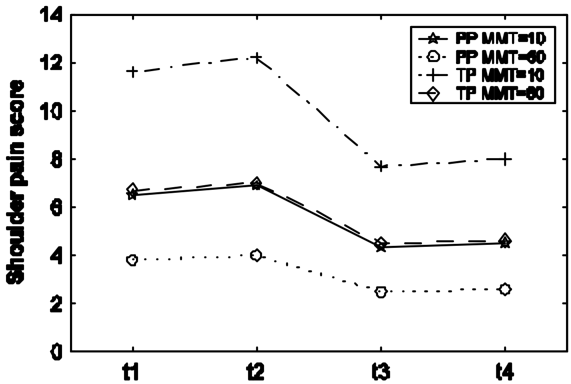


Figure 2.3: Effect of manual muscle test score on shoulder pain. Estimated shoulder pain score is plotted for a manual muscle score of 10 and a score of 60 for the subjects with tetraplegia (TP) and the subjects with paraplegia (PP) at the four test occasions.

Prognostic model

When the significant univariate variables were combined in a model together with upper extremity musculoskeletal pain at t_1, the BMI and the FIM motor score at t_1

were significant (Table 2.3) predictors for upper extremity pain at t_4. A 10 point higher FIM score at t_1 was associated with a 10% increased risk on developing upper extremity musculoskeletal pain at t_4 (*P*<0.001).

Table 2.3: Results of the Poisson analysis for the prediction of total upper extremity pain score at t_4 with variables at t_1.

Variable	*CorrCoef*	*S.E.*	*p-value*	*IDR*	*95% CI*	*95% CI*
Constant	*-0.679*	*0.213*				
Complaints t_1	*0.042*	*0.002*	*< 0.001*	*1.043*	*1.039*	*1.047*
BMI t_1	*0.064*	*0.008*	*< 0.001*	*1.067*	*1.050*	*1.082*
FIM t_1	*0.010*	*0.002*	*< 0.001*	*1.010*	*1.006*	*1.014*

The results are based on the final model after backward elimination.
Abbreviations: CorrCoef = correction coefficient, S.E. = standard error of the mean
IDR = incidence density ratio, CI = confidence interval, BMI = body mass index, FIM = functional independence measurement

For shoulder pain, age, BMI, completeness of the lesion and FIM motor score at t_1 were significant predictors when corrected for shoulder pain at t_1. Except for the completeness of the lesion (78%) the effect on pain was small (<10%). Older persons with a higher BMI, a higher FIM motor score and an incomplete lesion were at a higher risk for developing shoulder pain.

Discussion

Pain is defined as an unpleasant sensory and emotional experience associated with actual or potential tissue damage, or described in terms of such damage [68]. Not only is pain related to the physical state, but also to the psychological state of the persons. Since pain is a subjective score, and not a physical outcome measurement, it is a difficult variable to work with.

The musculoskeletal pain survey we used in this study was not validated or used in previous studies. As in several other studies, the survey was solely used to ask whether subjects experienced pain. Next to musculoskeletal pain, in a separate section subjects were asked for neurogenic pain at or under the lesion (tight band feeling, phantom pain, and dull pain). In this study it was tried to distinguish musculoskeletal pain from neurogenic pain. It is not always possible to distinguish clearly between these two types of pain, and therefore part of the scored musculoskeletal pain could be related to neurogenic pain.

In a large-scale research like the current project it is impossible to perform physical and technical (MRI) exams at all test occasions. However, for a better analysis of the problem these exams might be necessary. Therefore, further research would benefit from surveys combined with physical and technical exams.

It has long been acknowledged that the shoulder joint is especially at risk for overuse injuries due to its complex functional anatomy and its limited muscle mass. The arm is very mobile via the connection to the scapula and its joint structure (a small saucer within a large cup). This mobility goes at the expense of the stability of the joint [25] and makes it especially vulnerable for the development of instability complaints [105].

In the current study, we looked at the development of musculoskeletal pain during and after the rehabilitation process and found that pain complaints develop very quickly, which leads to the conclusion that pain was not simply due to overuse. Musculoskeletal pain could also have developed due to adaptation, in which case pain will probably develop in the long run due to overuse. In this study the shoulder pain score significantly increased during the first three months of active rehabilitation and decreased after that period. Since subjects with SCI in rehabilitation had become dependent on the use of their upper extremities, which are often not well trained in early rehabilitation, training of new skills might have led to stressed muscles and pain. During rehabilitation, when subjects get more experience with arm exercise and had had muscle strength training, pain could diminish. This study showed that the upper extremity musculoskeletal pain score decreased most between t_2 and t_3. However, after discharge of inpatient rehabilitation the musculoskeletal pain complaints did not increase but stabilized. It might have been possible that increased independence, more frequent performance of ADL tasks in a less adjusted environment after rehabilitation and less specific muscle training would have led to an increase in pain after the rehabilitation.

Lesion level was highly correlated to upper extremity and shoulder pain. This finding was not supported by Dalyan et al. [38] and Subbarao et al. [155], but consistent with Sie et al. [148] and Turner et al. [159]. Due to partial muscle paralysis of thoracohumeral muscles and shoulder muscle imbalance, individuals with high-level SCI are at higher risk of developing musculoskeletal pain [31, 127]. It is suggested that muscle paralysis will influence the remaining upper extremity muscles which have to stabilize the joints and have to produce the necessary external force to perform the task. The subjects with TP also need more external stabilization to maintain trunk balance to perform the task. On the condition that most subjects with a C6 or C7 lesion performed ADL independently, the extra muscle force needed for stabilization could have been responsible for the higher incidence of overload injuries for subjects with higher level lesions. It is possible that shoulder instability, which can arise after muscle paralysis, leads to shoulder complaints, but instability might contribute to the (earlier) onset of complaints due to overuse in subjects with high SCI lesions.

Musculoskeletal pain in SCI is experienced during daily life activities and especially weight-bearing tasks such as transfers and weight-relief lifts [119, 155]. For these tasks, balance is very important and the net moments and compression forces around the shoulder are high [175, 174]. It is common that more than 25% of the body weight is transferred through the humerus to the thorax during the performance of these tasks. Reduction of the subacromial space and impingement of the supraspinatus muscle are possible explanations for the development of pain. However, in this study all subjects with an acute SCI were included, including seven subjects with a high cervical lesion who had to use a power wheelchair for mobility. Although a small minority group in this study, overload injuries due to manual wheelchair use will not apply to these subjects.

How damage to the joints and the muscles develops is unfortunately unknown, which makes the evaluation of the load on the upper extremity during various ADL one of the utmost importance. More specifically, there is a need for a combined biomechanical epidemiological study, where the dose-response relationship between everyday mechanical load and pain can be studied.

We found that both muscle strength and functional outcome were inversely related to shoulder pain. Subjects with higher maximal muscle forces may perform the ADL tasks at a lower relative force compared to subjects with lower muscle strength. Therefore these subjects had a lower risk for stress on the muscles. As previously described, higher muscle forces and, as a consequence, higher compression forces in the joint [176], increase the risk on impingement, joint damage and muscle damage. It has been shown [33] that a specific exercise protocol of stretching and strengthening shoulder muscles can lead to a decrease in upper extremity pain. Curtis et al. [33] found an effect of a six months exercise protocol, where the intensity of the experienced pain was decreased. The effect was almost twice as high for subjects with PP than for subjects with TP, reflecting the interaction between functional status and lesion level.

MMT is a fairly crude way to measure muscle strength [63, 114, 146]. However, to include a large number of subjects the MMT data were used instead of the strength data measured with a hand-held dynamometer because the latter were only collected when the MMT score was 4 or 5. In our study not only the strength of the muscles, which provide the score for the lesion level were measured, but the strength of the shoulder internal and external rotators and shoulder abductors as well. Lesion level will have an influence on the MMT score, but will not necessarily dictate the MMT score since there is variation in the paralysis due to variation in the segment innervation and variation in the completeness of the lesion. From the results it might be concluded that higher muscle strength would lead to less pain and that exercise could reduce pain complaints.

The FIM motor score can be seen as an indicator of the number of ADL tasks subjects can perform independently. Neurological level of injury will have an effect on the FIM score but will not determine this score, because coordination, skill level, physical capacity and attitude will have an effect on the FIM score as well. Since the FIM motor score is positively correlated to the muscle score [8, 46], it was no surprise that both parameters were explanatory variables for shoulder pain. We expected that subjects with a higher FIM score developed fewer complaints because they have a better physical capacity. However, it was also possible that these subjects with a higher FIM score perform too many ADL tasks and develop complaints as a consequence of overuse, while on the other hand subjects with a low FIM score perform fewer tasks and are therefore less susceptible to overuse pain complaints.

In this study the possibility of having musculoskeletal pain one year after the rehabilitation was found to be much higher when pain was already present at an earlier time (t_1). Silfverskiold [149] reported that 33% of subjects with TP who had pain after six months still had shoulder pain after 18 months. Therefore early onset of shoulder pain within six months has a high predictive value for pain after 18 months.

Further, BMI, FIM motor score and completeness of the lesion were consistent predictors of musculoskeletal pain in the upper extremity and shoulder. It is not surprising that BMI was found to be a significant variable in the prognostic model, since BMI is obviously related to the amount of physical strain experienced. Heavier persons, due to more body fat or more muscle mass, have a higher mass to transfer and experience a higher drag force during wheelchair propulsion.

It is difficult to relate a higher FIM motor score at the beginning of the rehabilitation to more upper extremity musculoskeletal pain after the rehabilitation, but it is possible that due to their good functionality these persons started too early performing all sorts of tasks while they are generally not aware of the increased risk for upper extremity complaints. They probably have enough strength, but they have no experience and are not trained for specific arm function.

The same might be true for the predictor variable completeness of the lesion for shoulder pain. Persons with a complete lesion may perform fewer tasks and have a lower risk on developing pain, while persons with an incomplete lesion are less needy and use less assistive devices with the risk on overload.

Conclusion

The occurrence of upper extremity musculoskeletal pain during inpatient rehabilitation decreased over time. In the studied SCI population using wheelchairs as their primary means of mobility, the reduction in pain complaints was 30% between t_2 and t_3. There was a significant relationship between upper extremity and shoulder pain and lesion level. Subjects with TP were at a higher risk for pain than subjects with PP.

Muscle strength and functional outcome were identifiers for shoulder pain. Higher scores on these tests resulted in 10-15% fewer shoulder pain complaints.

Early onset of upper extremity pain seemed to be the most important predictor of pain at a later time. Thus, in the beginning of the inpatient rehabilitation one should be very careful to prevent overload (by performing heavy ADL like transfers). The rehabilitation should focus on a balanced training of the upper extremity to make up for the lack of strength of the upper extremity. Further, overweight should be prevented and one should strive for optimal wheelchair qualities.

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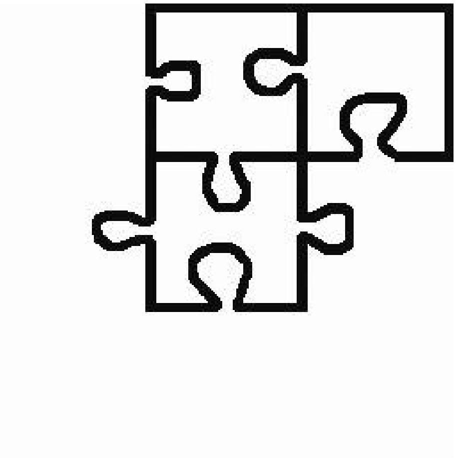
Chapter 3

Mechanical load on the upper extremity during wheelchair activities



Based on
S van Drongelen, LH van der Woude, TW Janssen, EL Angenot, EK Chadwick and DH Veeger
Archives of Physical Medicine and Rehabilitation 86(6): 1214-20, 2005

Abstract

Objective: To determine the net moments on the glenohumeral joint and elbow joint during wheelchair activities.

Design: Kinematics and external forces were measured during wheelchair activities of daily living (level propulsion, riding on a slope, weight-relief lifting, reaching, negotiating a curb) and processed in an inverse dynamics biomechanical model.

Setting: Biomechanics laboratory.

Participants: Five able-bodied subjects, eight subjects with paraplegia, and four subjects with tetraplegia.

Interventions: Not applicable.

Main Outcome Measure: Net moments on the glenohumeral joint and elbow joint.

Results: Peak shoulder and elbow moments were significantly higher for negotiating a curb and weight-relief lifting than for reaching, level propulsion, and riding on a slope. Overall, the elbow extension moments were significantly lower for subjects with tetraplegia than for those with paraplegia.

Conclusions: The net moments during weight-relief lifting and negotiating a curb were high when compared with wheelchair propulsion tasks. Taking the effect of frequency and duration into account, these loads might imply a considerable risk for joint damage in the long term.

Introduction

In handrim wheelchair users, the upper extremities are at serious risk of overuse injuries. Wheelchair use requires continuous use of the upper extremities, not only for mobility, but also for transfers, weight-relief lifts, and reaching activities. Studies [31, 119] have shown that shoulder pain and impingement frequently occur among people with a spinal cord injury (SCI). Pain is experienced during wheelchair-related activities of daily living (ADLs), such as wheelchair propulsion and performing transfers. Because these activities are essential for functional independence, quality of life, and even the life expectancy of people after a SCI [116], evaluating the mechanical load on the shoulder is important to an understanding of the mechanisms that may cause upper-extremity joint degeneration. Factors that have been mentioned as contributors to the development of shoulder complaints are the relatively high load and high frequency of this load on the shoulder during wheelchair propulsion [7]. In addition, and possibly even more important, the load on the shoulder during other wheelchair-related tasks, such as transfers and weight-relief lifts, has been mentioned [60, 121, 134].

In our study, we used net moments around the elbow and the glenohumeral joint to quantify the mechanical load on those joints. Net joint moments are generally used to analyze (working) conditions and to classify these conditions [24]. To show the high loading at the shoulder, studies [83, 137, 177, 180] have presented net joint moments for wheelchair propulsion at various speeds and for varying external power outputs. Some studies have reported high net moments during ADLs and work-related activities of able-bodied subjects [3, 67, 88]; however, little is known about the mechanical load during wheelchair-related ADLs. In studies with able-bodied subjects, Anglin and Wyss [3] reported unilateral net moments on the shoulder of 16Nm for coming from sit to stand and 28Nm for lifting a suitcase; Kuijer et al. [88] calculated net moments between 10 and 30Nm for pulling a refuse container.

Harvey and Crosbie [60] are the only authors thus far who have estimated shoulder and elbow moments (respectively, 45Nm and 30Nm) for subjects with tetraplegia during a weight-relief maneuver. Muscle activity was studied by Reyes [134], Perry [121], and Newsam [109] and colleagues, who showed high muscle activation of the latissimus dorsi, the long head of the triceps, and the sternal part of the pectoralis major during transfers and weight-relief maneuvers, respectively. The study by Harvey and Crosbie [60] reported far higher shoulder moments than were found for ADL wheelchair propulsion [177]. It is likely that wheelchair-related daily activities can result in higher peak mechanical loads on the shoulder (especially) than everyday wheelchair propulsion. However, until now, no systematic analysis of several wheelchair-related ADLs for both able-bodied and subjects with a SCI has been conducted.

The purpose of this study was to compare the mechanical load between subjects with a high-level SCI to subjects with a low-level SCI. Subjects with a high-level SCI show a higher prevalence and intensity of shoulder pain than subjects with a low-level SCI [31]. Not only are key muscles, such as the triceps brachii, lattisimus dorsi, and the sternal part of the pectoralis major, often compromised [107, 156], but subjects with a high-level SCI also have less trunk control. It is to be expected that more compensatory activity is needed in the remaining shoulder muscles to stabilize the glenohumeral joint, which might be revealed by kinetic and kinematic analysis. Combined with different kinematics, this will be reflected in a difference in, and likely higher, net moments for the shoulders.

The aims of this study were to determine (1) the net moments acting on the shoulders and the elbows during various wheelchair-related activities and (2) the differences between net moments on the glenohumeral joint and elbow joint for subjects with a high-level or low-level SCI versus able-bodied subjects.

Methods

Participants

Seventeen subjects participated (Table 3.1): five able-bodied subjects, four with tetraplegia, and eight with paraplegia. Two subjects with paraplegia and one subject with tetraplegia had an incomplete lesion. The inclusion criteria for this study were that subjects be male and have no current shoulder problems. All subjects were informed about the nature of the study before giving written informed consent to participate. The protocol of this study was approved by the Medical Ethical Committee of the Vrije Universiteit Medical Center.

Table 3.1: Subject characteristics.

Characteristics	*Able-bodied (n=5)*	*Paraplegic (n=8)*	*Tetraplegic (n=4)*
Age (y)	*22 ± 3*	*39 ± 12 **	*28 ± 5*
Height (m)	*1.82 ± 0.11*	*1.86 ± 0.08*	*1.88 ± 0.05*
Weight (kg)	*73 ± 5*	*79 ± 9*	*70 ± 14*
Injury level (min-max)	*NA*	*T3-T12*	*C6-C7*
Years after SCI	*NA*	*14 ± 10*	*7 ± 6*

Note. Values are mean ± standard deviation or range
* Significantly different compared with able-bodied (*P* <0.05)
Abbreviation: NA = not applicable

Protocol and Tasks

To determine the net moments during wheelchair-related ADLs, subjects performed different standardized ADLs under experimental conditions in an instrumented wheelchair (Figure 3.1). Both 3-dimensional external forces and moments and 3-dimensional kinematics of the upper extremity were determined in each activity. Before testing, all subjects were allowed to become accustomed to the experimental wheelchair and the experimental setup.

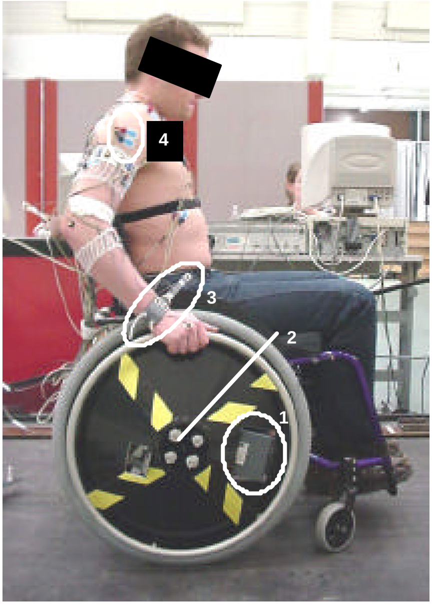


Figure 3.1: Able-bodied subject in instrumented wheelchair during level wheelchair propulsion. Legend: 1 = data logger, 2 = instrumented wheel, 3 = technical marker, 4 = EMG electrodes (EMG not used in current study).

Subjects performed three tasks: wheelchair propulsion, a weight-relief lift and a reaching task. The subjects with a SCI performed two additional tasks. Wheelchair propulsion was performed at $0.83 m \cdot s^{-1}$, to ensure a submaximal exercise level for all subjects. When the level treadmill (Enraf Nonius) was at speed and the subject was propelling comfortably, data were collected for a period of 30s.

Because of the design of the recording system, the weight-relief 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 larger support base. This task was performed three times with 20s rests between trials.

The third task was placing different bottles on a platform, 0.5m off the ground. The bottles varied in mass (0.1, 0.75, 1.5kg). At the start of each trial, subjects sat in the wheelchair and held the bottle at their lap; subsequently, they placed the bottle on the platform in front of them and took it back to the starting position. For this task, the exerted hand force was the force needed to compensate for the gravitational force on the bottle.

Subjects with a SCI also performed the following two additional tasks: subjects had to propel on a slope of 3% at a speed of $0.56 m \cdot s^{-1}$. When the treadmill reached the preset slope, 30s of propulsion were recorded.

The fifth task was negotiating a curb of 0.1m. Before negotiating the curb, the subjects were allowed to practice with the experimenter behind the wheelchair. If the subject was not comfortable performing the task, the task was cancelled. If the subject was comfortable, three successful trials were recorded.

Instrumented Wheelchair

All tasks were performed in a Quickie Triumph™ (Sunrise Medical Benelux) wheelchair (see Figure 3.1). A six degrees of freedom force transducer (M6-1000, Advanced Mechanical Technology Inc.) was built into the right wheel. The handrim was connected to the transducer by an aluminum shell. Next to the transducer, a portable data acquisition device (Porti™, Twente Medical Systems) and an angular position sensor were built into the wheel.

The wheelchair had a standard design with the backrest of the chair 0.42m wide and 0.40m high. The seat was 0.42m wide and deep. Seat height was .55m, 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, 0.305 and 0.265m. The diameter of the rim tube was 0.02m, 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; subsequently, the inertia of the other wheel was corrected by adding extra weights. The total weight of the instrumented wheelchair was 18.6kg.

Data were stored on a memory flash card. The instrumented wheel enabled us to measure the (propulsive) forces applied on the handrim as well as the torques on the handrim. The hand torque applied by the hand on the rim was calculated from the difference between the torque that was measured around the wheel axis and the torque produced by the applied force on the handrim [180]. It was assumed that the force was applied at the third metacarpal as the point of hand contact. The accuracy of the instrumented wheel was measured in newtons (F_x = forward = 3.0N, F_y = downward = 2.8N, F_z = medial = 4.1N) and for the moments in newton meters (M_x = 0.3Nm, M_y = 0.7Nm, M_z = 0.4Nm).

The force transducer was synchronized with the Optotrak™ computer (Northern Digital) by a telemetric system (Faculty of Human Movement Sciences). Forces and torques were low-pass filtered by using a 10Hz second-order recursive Butterworth filter. All torques and forces from the wheelchair were transformed from the rotating (local) coordinate system of the force transducer to forces and torques in the global coordinate system and subsequently corrected for the camber of the wheelchair and for the offset; i.e. the weight of the rim and the shell connected to the transducer.

Kinematics

Kinematics were recorded with a 3-camera optoelectronic system (Optotrak™, Northern Digital). 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 [164, 178]. The 3-dimensional positions of markers were recorded at 100Hz during each experimental trial. Recordings were performed with technical markers on the epicondylus medialis humeri and the processus styloideus ulnae. Before the actual measurements, a calibration measurement was performed in which the orientation of the technical markers was defined relative to bony landmarks. Also, the orientation of the scapula was determined by a calibration measurement with a scapula-locator system [75], while the subject sat in the wheelchair with the arm in the anatomic position. From the scapula calibration measurement and the orientation of the humerus during the tasks, the orientation of the scapula and clavicula were calculated by using a regression model of Pascoal [118]. 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 [164].

Biomechanical Model

The 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 [163, 165]. The input kinematics were derived from the position of the incisura jugularis and the orientation of the thorax, humerus, forearm and wrist. Orientation of the scapula and clavicula was obtained from regression equations [118]. Further, the 3-dimensional external forces and the torques applied by the hand on the rim were used as input. Output variables of the model used in this study were net joint moments around the glenohumeral joint and around the elbow joint.

Data Analysis

The moments around the glenohumeral joint were expressed as moment components (flexion and extension, endo- and exorotation, abduction and adduction) relative to the thorax. The moment components were used to calculate the resultant net moments on the glenohumeral joint. The net elbow moment was calculated as the moment around the flexion-extension axis of the elbow joint only (extension = positive, flexion = negative).

From the 30s recorded during the wheelchair propulsion tasks, five consecutive pushes were selected for data analysis. For every push, the peak net shoulder and elbow moment were determined. The push phase was defined as the phase in which the external force was above the level of noise in the recovery phase. For the other ADL tasks the peak values for each trial were determined. However, to compare the reaching task with the other tasks, only the peak shoulder and elbow moments of the trial with the 1.5kg weight were used. This specific trial was chosen to create a broad range of variation of external loading and thus of net moments. For the weight-relief lift, the moments around the elbow and the shoulder were corrected for body weight because the applied forces highly depend on the body mass.

Statistical Analysis

To detect significant differences among the subject characteristics of the three subject groups, independent *t*-tests were applied.

For each task, the mean of the peak moments over the trials was calculated. To compare the peak moments among the tasks a general linear model for repeated measures was used (within-subject factor: task; between-subject factor: groups). Depending on the tasks that were compared, different numbers of subjects were used. The level of significance was set at *P* less than 0.05 for all statistical tests.

Results

Participants

All subjects were able to perform the requested tasks except for negotiating the curb. The latter task could be performed by only five of the eight subjects with paraplegia and by none of the subjects with tetraplegia. The data of one of the tetraplegia subjects worked out to be erroneous because of missing values in the Optotrak data for both propulsion tasks and had to be discarded.

Except for age between the able-bodied and the paraplegic group, no differences were found for subject characteristics.

Wheelchair Propulsion

The peak net moments for the shoulder and elbow for low-intensity wheelchair propulsion were between 4.1 and 11.3Nm and between –0.5 and 7.9Nm, respectively (Table 3.2). The highest components around the shoulder were the adduction and the anteflexion components. Figure 3.2 gives a typical example of the net shoulder moment during the whole push.

No significant differences were found between the able-bodied subjects and the subjects with a high or a low SCI for both the shoulder and elbow peak moments.

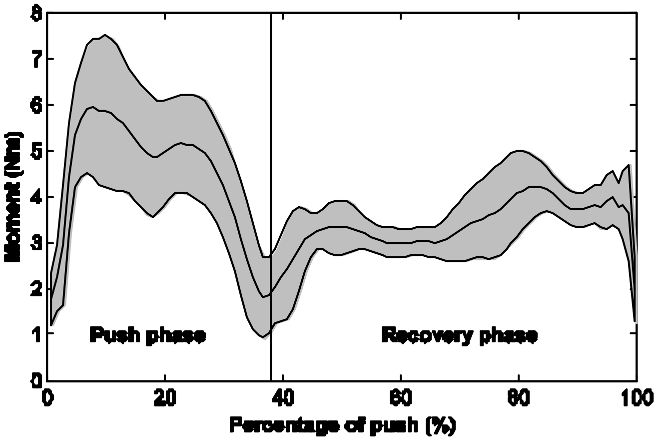


Figure 3.2: Typical example of the net shoulder moment for a subject with paraplegia, during wheelchair propulsion. Mean over five pushes and standard deviation, time normalized to a full cycle (100%).

Weight-Relief Lift

For the weight-relief lift, the peak moments on the shoulder and the elbow for the three trials of lifting were calculated. Figure 3.3 gives a typical example of the moments around the glenohumeral joint.

The two large moment components at the shoulder were retroflexion and adduction. After correction for body mass, the mean peak shoulder moment was 0.56Nm·kg^{-1} and for the elbow 0.47Nm·kg^{-1} for all subjects.

The absolute peak net moments for the shoulder and elbow were, respectively, between 24 and 70Nm and between 8 and 51Nm (see Table 3.2). The absolute values were used to compare

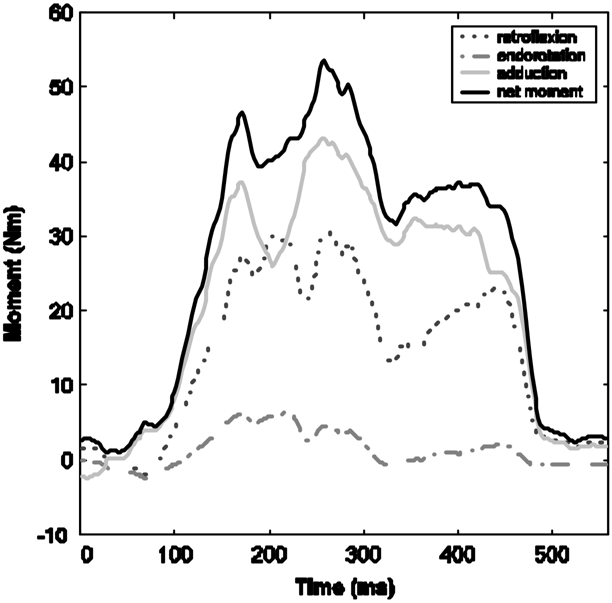


Figure 3.3: Typical example of the net shoulder moment and moment components for an able-bodied subject during a weight-relief lift.

the weight-relief lift with the other ADL tasks. For the shoulder moments, no significant differences were found between the subject groups.

However, a significant difference was found for the absolute peak elbow moment between the subject groups (P=0.008), leaving only a trend for the elbow moments when corrected for body mass (P=0.062). For comparison with the other tasks, the average over the three trials of the peak moments was calculated.

Reaching

The peak shoulder and elbow moments for the three trials of placing a bottle on the shelf were calculated (Figure 3.4, Table 3.2). The net moment on the glenohumeral joint increased from 5.8 to 12.7Nm with the increasing weight, the net moment on the elbow increased from –1.1 to –5.2Nm. A flexion moment in the elbow was needed to hold up the weight, whereas a mainly anteflexion moment in the shoulder was needed to hold up the arm in front of the body. For a higher weight, significantly higher shoulder and elbow moments were found (P<0.001). No significant differences were found between the three groups.

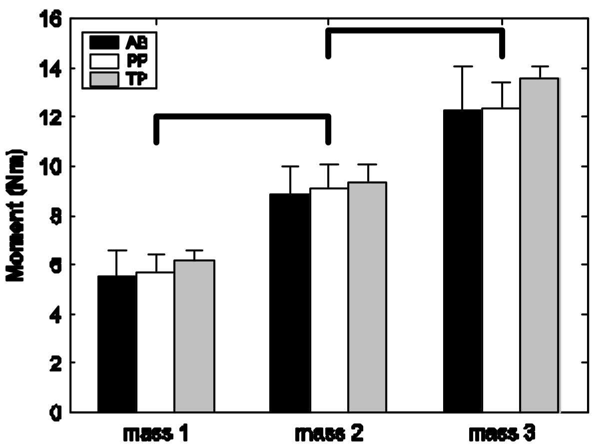


Figure 3.4: Net shoulder moments (mean and SD) during the reaching tasks for able-bodied subjects (AB) and subjects with paraplegia (PP) and tetraplegia (TP) for three different mass conditions. Reaching height was 0.5m.

? Significantly different (P<0.001).

Propelling on a Slope

Only the subjects with a SCI performed this task. The peak net moments for the shoulder were between 9.7 and 20.6Nm and for the elbow between 3.3 and 9.7Nm (see Table 3.2). No significant difference was found for the net moments between the subjects with paraplegia and with tetraplegia.

Negotiating a Curb

Five of 12 subjects with a SCI (PP only) were able to perform this task in the experimental wheelchair. The peak shoulder moments were between 36 and 97Nm for the different subjects (see Table 3.2). The elbow moments were between 32 and 75Nm.

Table 3.2: Peak net shoulder and elbow moments for the five ADL tasks for able-bodied subjects and subjects with paraplegia and tetraplegia.

Tasks	*Moments or power*	*Able-Bodied (n=5)*	*Paraplegic (n=8)*	*Tetraplegic (n=4)*	P
wheelchair propulsion	*GH peak (Nm)*	*6.7 ± 2.8*	*7.2 ± 2.4*	*9.0 ± 1.8*	
	EL peak (Nm)	*3.6 ± 2.0*	*3.0 ± 2.3*	*2.0 ± 2.3*	
	Power output (W)	*5.1 ± 0.8*	*4.4 ± 0.5*	*4.3 ± 1.6*	
reaching	*GH peak (Nm)*	*12.3 ± 1.8*	*12.3 ± 1.0*	*13.6 ± 0.5*	
	EL peak (Nm)	*-5.5 ± 0.4*	*-5.6 ± 0.2*	*-4.6 ± 1.6*	
riding a slope	*GH peak (Nm)*	*NA*	*14.6 ± 3.8*	*18.0 ± 1.8*	
	EL peak (Nm)	*NA*	*5.7 ± 2.1*	*7.6 ± 2.5*	
	Power output (W)	*NA*	*10.8 ± 2.1*	*11.8 ± 0.9*	
weight-relief lift	*GH peak (Nm)*	*36.1 ± 8.7*	*43.9 ± 9.4*	*44.4 ± 17.9*	
	EL peak (Nm)	*35.3 ± 10.6*	*42.2 ± 6.2*	*21.5 ± 11.9*	*0.008*
	GH peak ($Nm\cdot kg^{-1}$)	*0.5 ± 0.1*	*0.6 ± 0.1*	*0.6 ± 0.1*	
	EL peak ($Nm\cdot kg^{-1}$)	*0.5 ± 0.1*	*0.5 ± 0.1*	*0.3 ± 0.2*	
negotiating a curb	*GH peak (Nm)*	*NA*	*75.1 ± 23.5*	*NA*	
	EL peak (Nm)	*NA*	*60.2 ± 16.5*	*NA*	

Note. Values are mean ± standard deviation, NA = not applicable
Abbreviations: EL = elbow joint; GH = glenohumeral joint

Discussion

This study was conducted to gain insight into the external loading on the shoulder and the elbow during various wheelchair-related ADLs among subjects with a SCI and nonimpaired subjects. Although the mechanical load of wheelchair propulsion has been studied extensively, few studies [60, 109, 121, 134] have looked at wheelchair-related ADLs.

Wheelchair Propulsion Tasks

Subjects propelled the wheelchair on a level surface at a speed of $0.83 m\cdot s^{-1}$ as well as at a speed of $0.56 m\cdot s^{-1}$ on a slope of 3%. Therefore, external power output was limited; that is 4.6 ± 0.9W for level propulsion and 11.0 ± 1.9W for riding on a slope. However, a setup with a low speed without extra resistance was chosen, so that all subjects were able to ride at a submaximal level. The net moments we found seem to deviate from other studies [89, 137, 138, 177, 180] on wheelchair propulsion. However, considering the differences in power output among these studies and our study, the net moment values we found did not differ from those in the literature.

Weight-Relief Lift

Apparently, subjects with a high lesion level performed this task in a somewhat different way because they were not able to use full triceps activity to extend their arms. They seemed to first lock the elbow joint, after which they lifted their body weight from the shoulder with the clavicular part of the pectoralis muscle and the deltoid muscle. Therefore, the trend for a difference in elbow moments between the subjects with a high and low SCI could be explained by different kinematics. However, different activation levels of the triceps can cause these differences as well.

In addition, the constraint that subjects were required to use the handrim (and the combination of handrim and tire on the left side) 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 and the magnitude of the net moments. For small subjects, the handrims were further away from the body center than for larger subjects, which may increase external loading further. The elbow extension during weight-relief lifting (able-bodied, 18°±4°; paraplegia, 29°±6°; tetraplegia, 20°±11°) is a risk factor, and in combination with the high elbow moment it could compromise the integrity of the elbow joint.

Recently, the guidelines for pressure relief by weight-relief lifting have been revised in the Netherlands [22], as a result of a study by Coggrave and Rose [26], who found that traditional lifting was not efficient. In the light of the results of our study and other research [3, 26] this policy change makes sense.

Reaching

The results of the reach tasks showed that the net moments are dependent on the mass of the object. However, the actual moments were probably underestimated because of the low segment mass in the model. The moment on the shoulder joint during reaching with the empty bottle was comparable to wheelchair propulsion at a low speed. Because reaching with a weight of 1.5kg is a much more straining task, the shoulder moments were almost identical to the moments for propelling on the slope.

In this study, all subjects were able to perform all the tasks, and no differences were found between groups. This implied that no essential differences were found in kinematics and in external forces. For subjects with a high lesion level, different strategies may be necessary to stabilize the joint as a result of partial muscle paralysis. This compensation activity may lead to a high muscle stress and/or a high joint reaction force.

Negotiating a Curb

Only five subjects with a low-level SCI (of our 12 subjects with a SCI) could negotiate the curb. These subjects were very fit and were well able to handle the rather heavy (18.6kg) experimental wheelchair. Clearly, this task is accompanied by very high net moments in both the shoulder and the elbow. Subjects have to lift their body weight against gravity while rolling up the curb.

Methods

In this study, the mechanical load was expressed as net joint moments, which is a generally accepted measure to define mechanical load [3, 89, 137]. Net joint moments are the resulting moments around a joint to compensate for the external moments and to perform a certain task. Therefore, net joint moments are sensitive to the kinematics of the task. If, for the same external load, the kinematics differ, a difference in the net joint moments will be found. Yet no difference will be found if the kinematics do not differ or if the kinematic differences are relatively small compared with the force requirements, as in the weight-relief lift.

In subjects with a high-level SCI, key muscles are often compromised. It was expected that this would become visible in strategy or technique and in external force parameters. However, the subjects with a high-level SCI did not perform the tasks in a completely different way; therefore, we did not find differences between the net joint moments among the groups.

The model used in our study was not individualized but based on the morphology of an older cadaver [163, 182]. Therefore, we may have under- or overestimated the net moments because the moment component caused by the mass of the limb is constant for different subjects. The choice to use a single model is arbitrary but highlights the effects of kinematics and external forces. Also, our results will be in line with future comparisons between subject groups for individual muscle forces.

The load on the shoulder and elbow is considerable during ADL tasks. Therefore, apart from the ergonomics of the task layout, therapists should be aware of patients' physical capacity before starting to practice these heavy ADL tasks, to prevent early damage to the joints. We believe that overall muscular work capacity plays an important role in the height of the mechanical load. A recent study by Fullerton et al. [47] is indicative in this respect. Their results showed that highly-trained wheelchair athletes experience significantly less shoulder pain than nonathletic wheelchair users. It is expected that for subjects with a high-level SCI, the mechanical load (compression forces, muscle load) can be notably higher - for example, for weight-relief - because of the often complete absence of active support from the legs and the compensatory muscle activity in the remaining muscle groups after muscle paralysis. Therefore, an active training or exercise could be beneficial, probably even in early rehabilitation, to increase work capacity.

The risk for musculoskeletal injuries is not only affected by the peak forces occurring during a task, as presented in this study, but also by the frequency, the duration, the direction of the force, and the point of force application of a given task [162]. Even though the load on the shoulders and the elbows is relatively low during normal wheelchair tasks, wheelchair propulsion is a repetitive task and could lead to overuse injuries as a consequence of the combination of load and repetition. ADL tasks like weight-relief lifting are relatively low frequent but extremely straining. There is also the absence of sufficient recovery time: subjects must perform lifts during the day and propel themselves. Tasks like making a transfer are performed around 15 times a day [119], and, as shown by Janssen et al. [71], a physical strain of 60% of the heart rate reserve occurs frequently during transfers. The high loads during ADL tasks might be a risk factor for overuse of the upper-extremity joint, which would be in line with epidemiologic data [119]. When these high loads lead to trauma in the upper extremity, it is likely that no recovery occurs because of the regular (and almost inevitable) repetitive submaximal loading of the upper extremity during wheelchair propulsion. It is therefore likely that neither wheelchair propulsion nor weight-relief lifts by themselves are responsible for the high prevalence of overuse injuries but that the combination of both forms of loading comprise a high-risk factor.

Conclusions

Negotiating a curb and performing a weight-relief lift were accompanied by a significantly higher net moment in the shoulder and elbow than were found for wheelchair propulsion and reaching. Propelling on a slight slope caused a higher shoulder moment than did normal wheelchair propulsion.

No significant differences were found in the estimated loads on shoulders among the three groups. For the subjects with paraplegia, the elbow moments during the weight-relief lift were significantly higher than for the subjects with tetraplegia.

Acknowledgment
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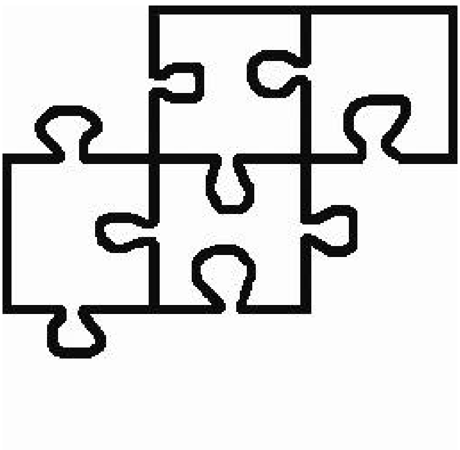
We greatly acknowledge the technical assistance of Jos van den Berg and the experimental assistance of Brechje Tijssen, Manon Faijdherbe, and Marijke Schep.

Chapter 4

Glenohumeral contact forces and muscle forces evaluated in wheelchair-related activities of daily living in able-bodied subjects versus subjects with paraplegia and tetraplegia



Based on
S van Drongelen, LH van der Woude, TW Janssen, EL Angenot, EK Chadwick and DH Veeger
Archives of Physical Medicine and Rehabilitation 86(7): 1434-40, 2005

Abstract

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, eight subjects with paraplegia, and four 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<0.001). 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<0.022).

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

Introduction

Shoulder pain often interferes with activities of daily living (ADLs) essential for the functional independence of people with a spinal cord injury (SCI) [116]. 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 [121, 134, 175].

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 [177], but it is unlikely that the same relationship would hold for subjects with a high-level SCI.

In a previous study [175], 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 subjects than for people with low-level SCI and able-bodied individuals. Glenohumeral contact forces for ADLs in subjects with a SCI have not yet been studied. Contact forces have been estimated for wheelchair propulsion by Veeger et al. [177], who reported peak glenohumeral contact forces between 800 and 1400N. In studies with able-bodied subjects, Anglin et al. [4] reported values up to 1750N for lifting a 10kg suitcase. Kuijer et al. [88] 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 [134], Perry [121], and Newsam [109] 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 muscles [177] and the pectoralis major muscle [104] seem to be at risk for muscle damage.

The first aim of this study was to determine glenohumeral contact forces and muscle forces during three wheelchair-related activities. The second aim was to determine whether 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 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 five able-bodied subjects, eight subjects with paraplegia, and four 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 one subject with tetraplegia had an incomplete lesion. Subject characteristics are listed in table 4.1. The protocol of this study was approved by the Medical Ethical Committee of the Vrije Universiteit Medical Center.

Table 4.1: Subject characteristics

	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-T12*	*14 ± 10*
Tetraplegic (n=4)	*28 ± 5*	*1.88 ± 0.05*	*70 ± 14*	*C6-C7*	*7 ± 6*

NOTE. Values are mean ± standard deviation
Abbreviation: NA = not applicable
* Significantly older compared with able-bodied subjects (P<0.05)

Protocol and tasks

Subjects performed a set of five standardized ADL tasks under experimental conditions in an instrumented wheelchair. From 11 ADLs measured in a pilot experiment, five tasks were selected by the researcher and a physician. These five 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 three tasks were analyzed: wheelchair propulsion, weight-relief lifting, and placing a bottle on a platform ('reaching').

To ensure a submaximal exercise level for all subjects, wheelchair propulsion was performed on a level treadmill (Enraf Nonius) at $0.83 m \cdot s^{-1}$. 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 three times with a 20s rest between trials.

The third task was a reaching task; subjects had to place a 1.5kg bottle with their right hand on a platform 0.5m high. 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 (Sunrise Medical Benelux). A six degrees of freedom force transducer (M6-1000, Advanced Mechanical Technology Inc.) 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 (Porti™ Twente Medical Systems) and a custom angular position sensor were built into the wheel (Figure 4.1). The width of the back of the wheelchair was 0.42m; the height was 0.40m. The seat was 0.42m wide and deep. The seat height was 0.55m, 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, 0.305 and 0.265m. The diameter of the rim tube was 0.02m, 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.

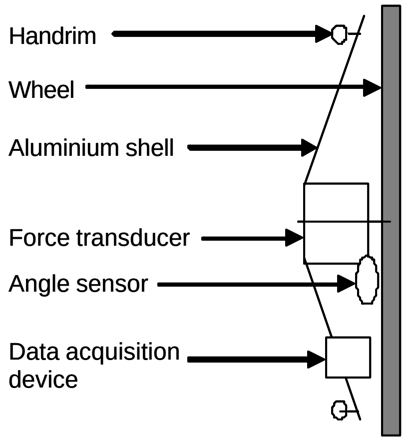


Figure 4.1: Schematic drawing of the instrumented wheel, placed at the right side of the wheelchair; cross-sectional front view.

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 [28]. 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 [180]. 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 (Fx [forward] = 3.0N, Fy [downward] = 2.8N, Fz [medial] = 4.1N) and for the moments (Mx = 0.3Nm, My = 0.7Nm, Mz = 0.4Nm).

The AMTI force transducer was synchronized with the Optotrak™ (Northern Digital) computer by a telemetric system (Faculty of Human Movement Sciences). Forces and moments were low-pass filtered by using a 10Hz 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 (i.e. the weight of the rim and the shell connected to the transducer).

Kinematics

Kinematics were recorded during each experimental trial by using three Optotrak™ cameras (Northern Digital) 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 [164, 178]. 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 system [175] 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 clavicula were calculated by using a linear regression model of Pascoal [118]. 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 [187]. 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 [165]. The model is an inverse-dynamic musculoskeletal model of the upper extremity consisting of 31 muscles, divided into 139 muscle elements. For muscles with large attachment sites or complex architectures, more than one 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.

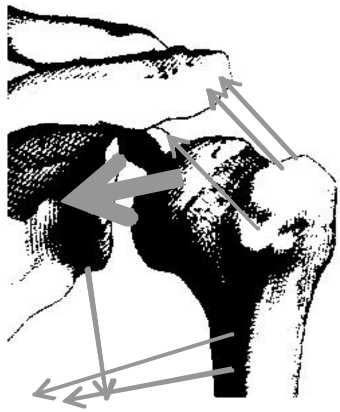


Figure 4.2: Picture of the shoulder joint that explains the principle of the glenohumeral contact force. In the figure the glenohumeral contact force (from center of rotation of the humeral head to the glenoid) is presented by a thick arrow. The lines of work for six hypothetical muscle elements are presented by thin arrows.

The input kinematics were the position of the incisura jugularis and the orientations of thorax, humerus, forearm, and wrist. Orientations of the scapula and clavicula 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 (Figure 4.2). Muscle forces were calculated based on a minimum stress cost function. 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 $100 N \cdot cm^{-2}$ of the physiologic cross-sectional area and obtained from Veeger et al. [179, 182].

In our study, the lesion level was not simulated in the model by reducing muscle force in paralyzed muscles; therefore, all 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 30s. From these 30s of raw data, five 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 five 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 three 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 0.05 for all statistical tests.

Results

Participants

Four subjects with tetraplegia, five able-bodied subjects, and eight subjects with paraplegia participated in this study. No differences were found for subject characteristics, except that the able-bodied group was younger than the paraplegia group (Table 4.1).

Of the subjects with tetraplegia, one subject had no triceps muscle tension at all (manual muscle test [MMT] score, 0), two subjects were unable to act against gravity (MMT score, 2), and one 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.

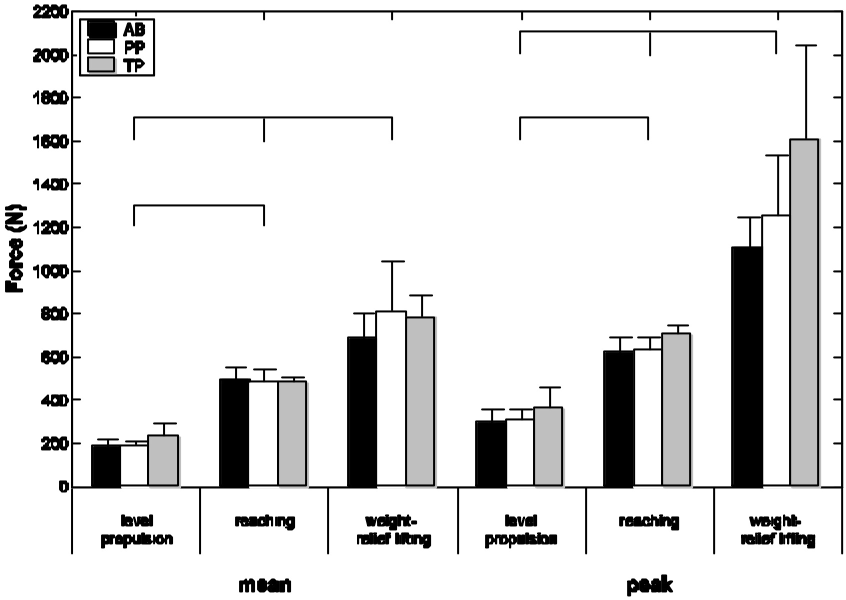


*Figure 4.3 Mean and peak glenohumeral contact forces for the three ADL tasks for able-bodied subjects (AB), subjects with paraplegia (PP) and those with tetraplegia (TP). **?** = Significantly different among the tasks.*

Glenohumeral contact forces

For both peak and mean values, performing a lift was accompanied by a significantly (P<0.001) higher glenohumeral contact force when compared with both wheelchair propulsion and reaching (Figure 4.3). Reaching caused a significantly (P<0.001) 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=0.01) - 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 performed 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 (Figure 4.4). When expressed as a percentage of their maximum force, the supraspinatus was the muscle with the highest load (12%). The relative forces of the other muscles were between 5% and 10%. No overall significant differences were found among groups.

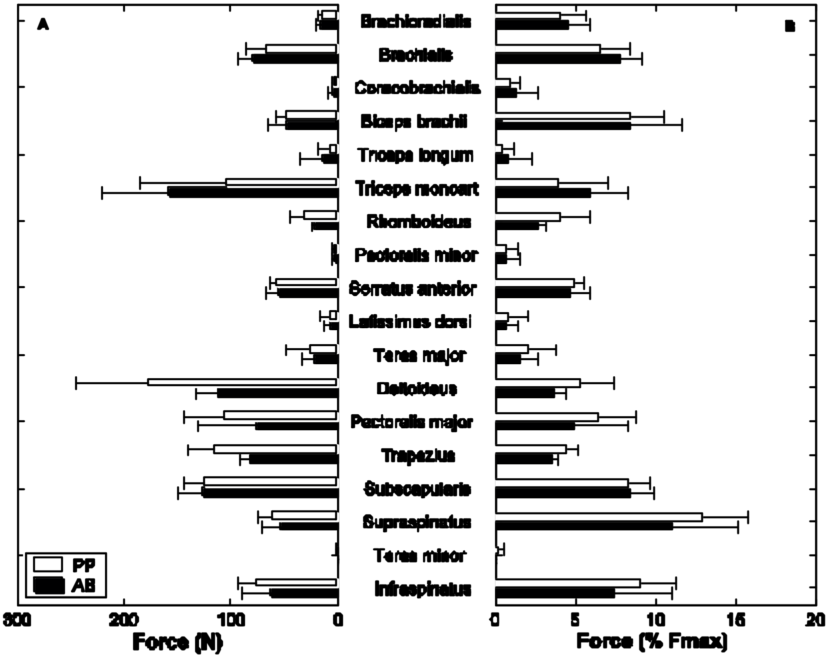


Figure 4.4: (A) Peak absolute muscle forces and (B) peak relative muscle forces during the push phase for both the able-bodied and paraplegia groups. Abbreviations: max = maximum, monoart= monoarticular.

Reaching

The muscles that produced the largest peak force during the reaching task were the deltoideus, the brachialis, and the trapezius muscles (Figure 4.5). 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 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 (Figure 4.6). When expressed as a percentage of their maximum force, for the three subject groups, the latissimus dorsi, the biceps brachii, and the monoarticular part of the triceps

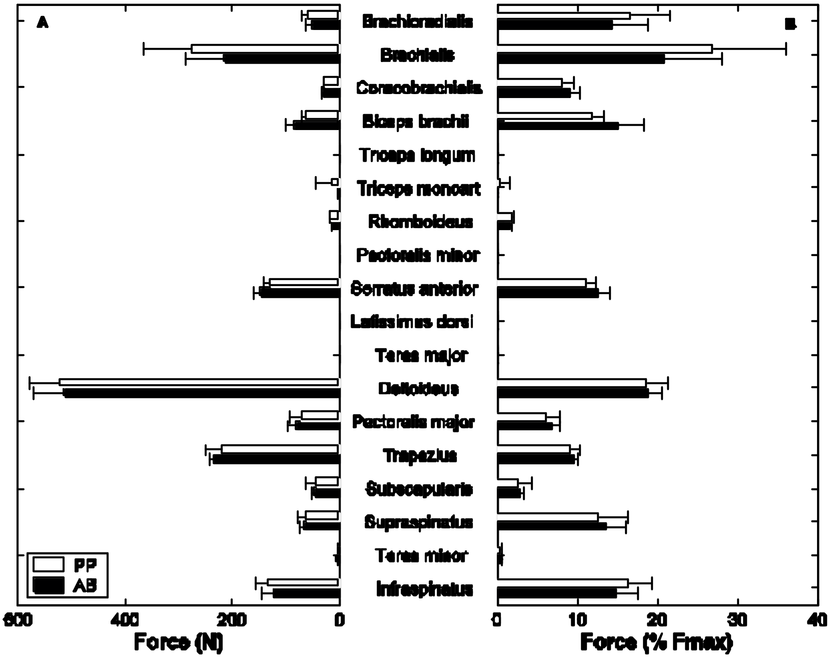


Figure 4.5: (A) Peak absolute muscle forces and (B) peak relative muscle forces for the reaching task with a 1.5kg mass at 0.5m for both the able-bodied and paraplegia groups.

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 4.6, 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=0.022).

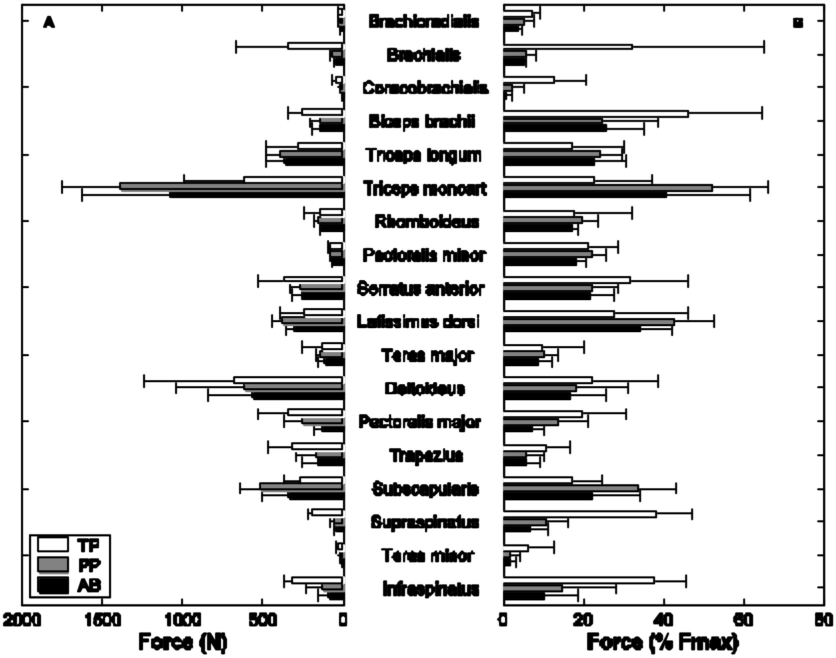


Figure 4.6: (A) Peak absolute muscle forces and (B) peak relative muscle forces for weight-relief lifting for the able-bodied, paraplegia, and tetraplegia groups.

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 ±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 ± 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 al. [177] 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 occurring. 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 Kuijer et al. [88] in which contact forces between 500 and 1500N have been reported for pulling and pushing containers of 40 to 74kg.

The mechanical load had not been calculated yet for weight-relief lifting in SCI. However, in able-bodied subjects, Anglin et al. [4] 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 groups [175]; 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 Figure 4.4). These findings are in accordance with Veeger, [177] 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 al. [104] found relatively high and prolonged electromyographic activity in the supraspinatus for wheelchair propulsion at 5km·hr^{-1}. The other prime movers for wheelchair propulsion showed muscle forces in accordance with the previously mentioned studies. Our study reported 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 is required (i.e. other wheeling conditions, higher velocity, or a suboptimal wheelchair design), the load on, among others, the supraspinatus, will be higher, and the risk for shoulder complaints will therefore 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 3 km·hr^{-1}, approximately 45 pushes·min^{-1} 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 injury [16] and has led to research into mechanisms to reduce the peak force [79, 135].

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 electromyographic activity of these muscles recorded during forward flexion [87, 101].

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 Newsam [109] and Reyes [134] 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 [100].

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 significant 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 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 be 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.

Even 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.

Acknowledgments

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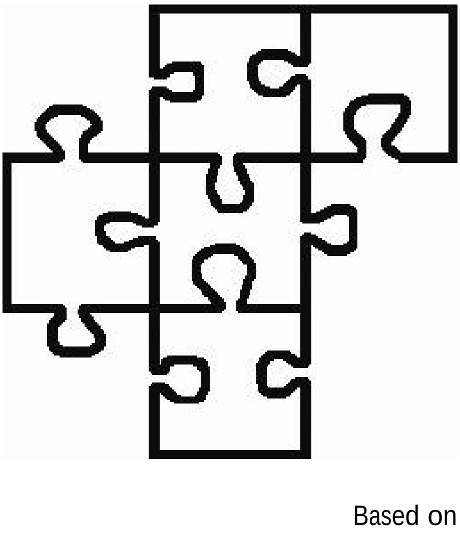
Chapter 5

Glenohumeral joint loading in tetraplegia during weight-relief lifting: a simulation study



Based on
S van Drongelen, LHV van der Woude, TWJ Janssen, ELD Angenot
EKJ Chadwick and HEJ Veeger
Clinical Biomechanics, accepted for publication

Abstract

Background: The incidence of shoulder complaints in wheelchair users is high and the etiology is poorly understood. The goal of this study was to examine the effect of lesion level and isolated triceps muscle paresis on the internal load on the shoulder by simulation.

Methods: Kinematic and kinetic profiles from four able-bodied subjects and four subjects with tetraplegia were used as input for an inverse dynamics biomechanical model. The model was modified to simulate lesion level and triceps muscle paresis.

Findings: The simulations resulted in a significantly higher (+56%) glenohumeral contact force (*P*=0.037) for tetraplegic profiles than for able-bodied profiles. The model modifications to simulate lesion level only had a minor effect (+7%) on the calculated glenohumeral contact force. More simulations were successful at lower triceps force levels for tetraplegic profiles compared to able-bodied profiles (*P*=0.012). The muscle forces at the simulated T1 lesion were not significantly higher in tetraplegic profiles compared to able-bodied profiles.

Interpretation: The glenohumeral contact force for the tetraplegic profiles is mainly higher due to different task performance. Model modifications only have a minor effect on the calculated glenohumeral contact force. For able-bodied profiles the triceps force seems to be an important factor. The high internal load at the shoulder recommends new techniques of weight-relief lifting and proper training of the arm-shoulder muscles in rehabilitation.

Introduction

Shoulder pain is common in individuals with a spinal cord injury (SCI) and is associated with the level and severity of injury. Pain appears to be more common in individuals with tetraplegia (TP) and individuals with complete injuries [31, 114, 148]. Due to muscle paresis and shoulder muscle imbalance, individuals with TP are at higher risk of developing shoulder pain, especially during weight-bearing activities [31, 109, 127, 134]. Weight bearing activities, with sustained high forces, are not the sole factor for developing shoulder complaints [7], repetitive applied forces during wheelchair propulsion are also a risk factor.

A previous study showed that for weight-relief lifting and negotiating a curb the net load on the shoulder was high [175]. However, the variation in motor lesion level among individuals with TP will affect the innervation of the shoulder and arm-hand muscles and will therefore have an effect on the muscle balance. In subjects with TP the triceps muscle is a key muscle for the positioning of the arm in space. When the triceps is paralyzed, extension of the elbow is only possible when gravity or inertia of the arm and hand is used or when the shoulder is anteflexed in a closed chain. To perform a weight-relief lift without triceps force an elbow hyperextension is needed to lock the elbow joint in order to lift the body weight with the remaining shoulder and trunk muscles. It is suggested that this variation in muscle paresis will influence the remaining upper extremity muscles which have to stabilize the joints and have to produce the necessary external force to perform a task. With the paresis of upper extremity muscles, a different movement pattern and deviant muscle activity have been recorded during the performance of a weight-relief lift [109]. Elbow extension was achieved prior to trunk elevation and greater activity of the anterior deltoid and of the infraspinatus was recorded for subjects with TP when compared to subjects with a low paraplegia.

Muscle paresis due to SCI can lead to higher loads on the glenohumeral and the elbow joints, because other muscles have to stabilize the joint with possibly unfavorable torque components which have to be compensated by additional muscle force. Above that, the non-paretic muscles are likely to be at a higher risk for soft tissue damage due to the higher muscle forces in the remaining functional muscle mass. Until now, there are no data available on the effect of lesion level on the joint and muscle loads during wheelchair related tasks.

To examine the effects of lesion levels and triceps muscle paresis on the load on the shoulder, a biomechanical model of the upper extremity [165] will be used to estimate this load. The purpose of this study was to answer the following two questions by model modifications: 1) what is the effect of simulated lesion level on the estimated glenohumeral contact force and muscle load, given exerted external hand forces and kinematics during a weight-relief lift of able-bodied subjects and subjects with tetraplegia? 2) What is the effect of simulated triceps paresis on the load on the shoulder?

It was hypothesized that in a simulated high cervical lesion the load on the glenohumeral joint would be higher compared to a simulated low cervical lesion, and at least partially associated with an increase in force of the rotator cuff muscles. Further it was expected that when the simulations with triceps paresis of the subjects with tetraplegia were successful, the glenohumeral joint would be higher due to a different performance.

Methods

Design

Kinematic and kinetic data from four able-bodied (AB) subjects and four subjects with TP were collected during the performance of weight-relief lifting. These data were used as inputs for the Delft Shoulder and Elbow Model. By modifying this model, the effects of lesion level and reduced triceps force on the load on the shoulder were analyzed.

Data collection

Kinematics and kinetics were collected during three standardized trials of weight-relief lifting. Four subjects with TP (mean (SD): 28 (5) yrs, 70 (14) kg, 1.88 (0.05) m, 7 (6) yrs since injury, C6-C7, three complete and one incomplete lesion) and four AB subjects (mean (SD): 22 (4) yrs, 71 (2) kg, 1.78 (0.05) m) were informed of the nature of the study before they gave their written, informed consent to participate. The protocol of this study was approved by the Medical Ethical Committee of the Vrije Universiteit Medical Center.

Subjects performed three trials of weight-relief lifting under standardized laboratory conditions in an experimental wheelchair. Kinematics of the trunk and the right arm were recorded with a three-unit motion analysis system (Optotrak™, Northern Digital) operating at 100Hz. Prior to the actual measurements, a calibration measurement was performed to define the orientation of the technical markers relative to bony landmarks. The subjects with TP were not capable of performing the lift quasi-statically, therefore the orientation of the scapula was determined with a scapula-locator system [75], while the subject was sitting in the wheelchair with the arms in the anatomical position. From the orientation of the scapula measured during the calibration and the orientation of the humerus during the tasks, the orientations of the scapula and clavicula during the task were calculated using linear regression. Assuming a normal kinematic motion of the scapula for subjects with a SCI, we applied the regression model of Pascoal [118] applying a 0N load (Table 5.1). External forces were recorded with an instrumented wheelchair [175]. The point of external force application on the hand was assumed to be at the base of the third metacarpal.

From both subject groups, twelve profiles were created (4 subjects x 3 trials) consisting of the kinematic and kinetic data, these profiles were used as input for the model.

Table 5.1: Coefficients of the regression model of Pascoal to calculate angles of clavicula and scapula

	Constant	*Hy*	*Hz*	*X_0*	*Load*
Cy	*1.079*	*+ 0.130*	*- 0.244*	*+ 0.803*	*-0.628*
Cz	*-3.960*	*- 0.0284*	*+ 0.118*	*+ 1.094*	*0.197*
Sy	*14.759*	*+ 0.106*	*- 0.0736*	*+ 0.677*	*-0.183*
Sz	*-11.008*	*- 0.0811*	*+ 0.413*	*+ 0.730*	*0.811*
Sx	*0.433*		*+ 0.040*	*+ 0.973*	

The output parameters Cy (clavicula protraction), Cz (clavicula elevation), Sy (scapula protracion), Sz (scapula latero-rotation) and Sx (scapula tilt) are dependent on the orientation of the humerus (Hy = plane of elevation, Hz = angle of elevation) and the initial angles of the clavicula and scapula (X0) i.e. $Cy = 1.079 \cdot Constant + 0.130 \cdot Hy - 0.244 \cdot Hz + 0.803 \cdot X_0 - 0.628 \cdot load$.

Delft shoulder and elbow model

The Delft Shoulder and Elbow Model [165] is a finite element musculoskeletal model, consisting of 31 muscles, divided into 139 muscle elements. The model has to satisfy two constraints: the external moments must be balanced by muscle forces, termed the ‘moment constraint’ and the glenohumeral joint contact force vector must be directed into the glenoid cavity, which is termed the ‘stability constraint’. Inputs to the model were the profiles, consisting of kinematics (position of the Incisura Jugularis, the rotations of the thorax, scapula, clavicle, humerus, forearm and wrist) and kinetics (the exerted forces). Output variables of the model are the glenohumeral contact forces and the muscle forces (Figure 5.1).

Muscle forces were calculated based on a minimum stress cost function (minimize the sum of the squared muscle stresses) and the total force produced by each muscle was obtained by summing the forces of the muscle elements [23, 177]. Muscle forces were expressed in Newton and not relative to the maximum muscle force. The maximum muscle forces were based on the physiological cross-sectional area [179, 182] multiplied by a force of $100 N \cdot cm^{-2}$.

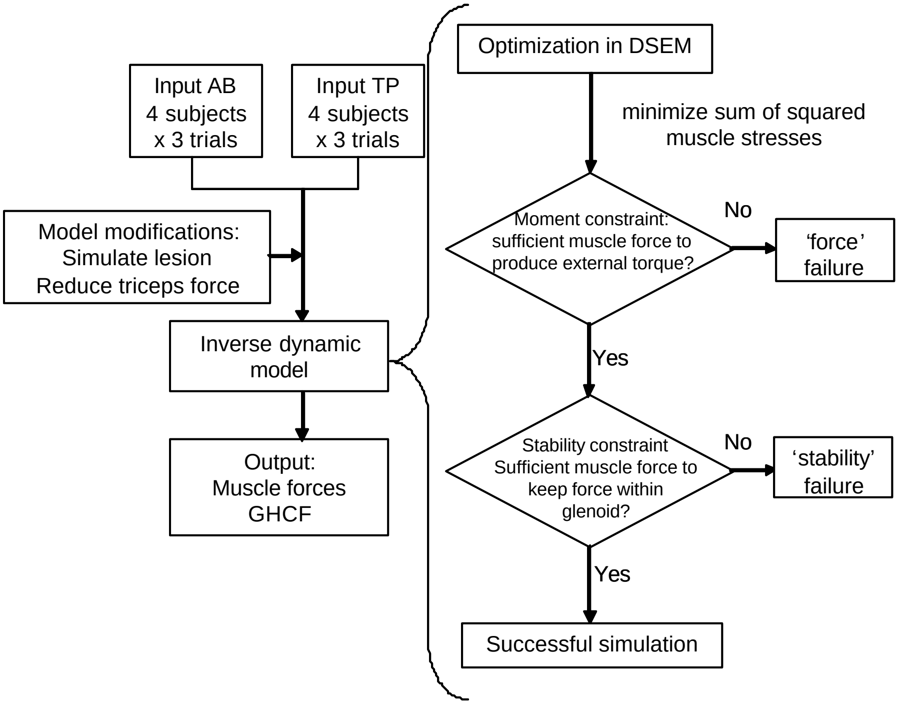


Figure 5.1: Simulation procedure (left) and optimization in the DSEM (right). On the left hand side the experimental procedure is presented. Kinematics and kinetics of able-bodied and tetraplegic subjects served as input for the inverse dynamic model. The model could be modified to i.e. simulate lesion level. On the right hand side the optimization (by minimizing the muscle stress and 2 constraints) of the model is given.

Model modifications

Lesion level

To modify the model to simulate complete lesion levels, we made a classification of muscle force at each lesion level, based on muscle segment innervations as described in Gray [54]. The assumption was made that the maximum relative force of each muscle was expressed relative to the number of innervating segments above the lesion (Table 5.2). For example, the pectoralis minor is innervated from C6 - C8 and at a C7 lesion we assumed that only two thirds of the muscle was innervated, therefore this muscle had only 66% of its maximal relative force. By this method the model was modified to simulate lesions from C5 to T1, whereby a T1 lesion was equal to the complete, fully functional, shoulder-elbow model. C4 lesions were not simulated because individuals with a C4 lesion are not capable of performing a weight-relief lift.

Since the model is a shoulder and elbow model, hand function was not modified and only 25 arm and shoulder muscles are presented (Table 5.2). All 24 profiles, both from AB and TP subjects, were used as input to the modified models. In this way, the profiles of the AB subjects were also used as input to a model with a SCI.

Table 5.2: Percentage of maximum relative force (%) for each lesion level.

	C5	*C6*	*C7*	*C8*	*T1*
m. trapezius scap part	*100*	*100*	*100*	*100*	*100*
m. trapezius clav part	*100*	*100*	*100*	*100*	*100*
m. levator scapulae	*100*	*100*	*100*	*100*	*100*
m. pectoralis minor	*0*	*33*	*66*	*100*	*100*
m. rhomboideus	*100*	*100*	*100*	*100*	*100*
m. serratus anterior	*33*	*66*	*100*	*100*	*100*
m. deltoideus scap part	*50*	*100*	*100*	*100*	*100*
m. deltoideus clav part	*50*	*100*	*100*	*100*	*100*
m. coracobrachialis	*33*	*66*	*100*	*100*	*100*
m. infraspinatus	*50*	*100*	*100*	*100*	*100*
m. teres minor	*50*	*100*	*100*	*100*	*100*
m. teres major	*0*	*50*	*100*	*100*	*100*
m. supraspinatus	*66*	*100*	*100*	*100*	*100*
m. subscapularis	*50*	*100*	*100*	*100*	*100*
m. biceps caput longum	*50*	*100*	*100*	*100*	*100*
m. biceps caput breve	*50*	*100*	*100*	*100*	*100*
m. triceps caput longum	*0*	*33*	*66*	*100*	*100*
m. latissimus dorsi	*0*	*33*	*66*	*100*	*100*
m. pectoralis major thor part	*0*	*0*	*33*	*66*	*100*
m. pectoralis major clav part	*50*	*100*	*100*	*100*	*100*
m. biceps caput longum	*50*	*100*	*100*	*100*	*100*
m. triceps caput mediale	*0*	*33*	*66*	*100*	*100*
m. brachialis	*33*	*66*	*100*	*100*	*100*
m. triceps caput laterale	*0*	*33*	*66*	*100*	*100*
m. anconeus	*0*	*0*	*50*	*100*	*100*

Based on the muscle segment innervations as described in Gray [54], a classification of muscle force was made. It was assumed that the relative maximum force of each muscle was relative to the number of innervating segments above the lesion. In bold are the innervating segments of the muscles based on Gray [54].

Triceps force

To study the effect of loss of triceps force in detail, paresis of the triceps muscle was simulated by scaling the maximum relative force up from 0% (no force) to 100% (maximum force) in steps of 10%. Because the anconeus is also an elbow extensor and is on the same neurological level as the triceps, the force of the anconeus was modified in conjunction with the triceps. During these simulations, all the other muscles in the model were assumed to be fully functional.

Data analysis

On completion of a simulation, the model returns the exit status of the simulation. This can be either successful or non-successful completion of the task. A non-successful simulation indicates that the model was unable to complete the given task, and can be further divided into the lack of ability to generate a sufficiently large moment balance ('force failure') or the lack of ability to direct the glenohumeral contact force into the glenoid ('stability failure') (Figure 5.1).

For the successful simulations, the peak glenohumeral contact forces and the projection of the glenohumeral contact forces on the glenoid surface were determined.

The model was not able to satisfy the constraints for all profiles at the same simulated lesion level, therefore different profiles were successful at a different lesion level or at a different level of triceps force. As a consequence, we analyzed the outcomes of the simulations with the complete model (defined as T1 and triceps100) and the outcomes of the highest successful lesion level (defined as S1) or the lowest successful triceps force level (defined as triceps1). Further, the muscle forces, at the time of the peak glenohumeral contact force, were determined for all successful profiles.

Statistics

Success of the simulations

To detect significant differences in the number of successful simulations between the able-bodied profiles and the tetraplegic profiles, a non parametric Mann-Whitney test was used.

Lesion level

To detect significant differences between the glenohumeral contact forces of the able-bodied and tetraplegic profiles a three factor General Linear Model for repeated measures was used (2 within-subject factors: 1) lesion level (n=2: S1 and T1) and 2) trial (n=3); one between subject factor: able-bodied profile versus tetraplegic profile). In this way we were able to detect differences between the subject groups, between the highest successful lesion level (S1) and the T1 lesion level and among the different trials. The level of significance was set at P=0.05.

Triceps force

A three factor General Linear Model for repeated measures was used (2 within-subject factors: 1) triceps force (n=2: triceps1 and triceps100) and 2) trial (n=3); one between subject factor: able-bodied profile versus tetraplegic profile) to find significant glenohumeral contact force differences between the subject groups, between the lowest successful triceps force (triceps1) and the simulation with 100% triceps force (triceps100) and among the trials. The level of significance was set at P=0.05.

Muscle forces

To test whether the tetraplegic profiles differed from the able-bodied profiles in terms of estimated muscle forces at the T1 level a two factor General Linear Model for repeated measures was used (within-subject factor: trial (n=3); one between subject factor: able-bodied profile versus tetraplegic profile). For the summed forces of the rotator cuff muscles as well for the forces of the separate muscles of the rotator cuff, a two factor General Linear Model for repeated measures (1 within-subject factor: lesion level (n=4); one between subject factor: able-bodied profile versus tetraplegic profile) was performed to find significant differences among the lesion levels.

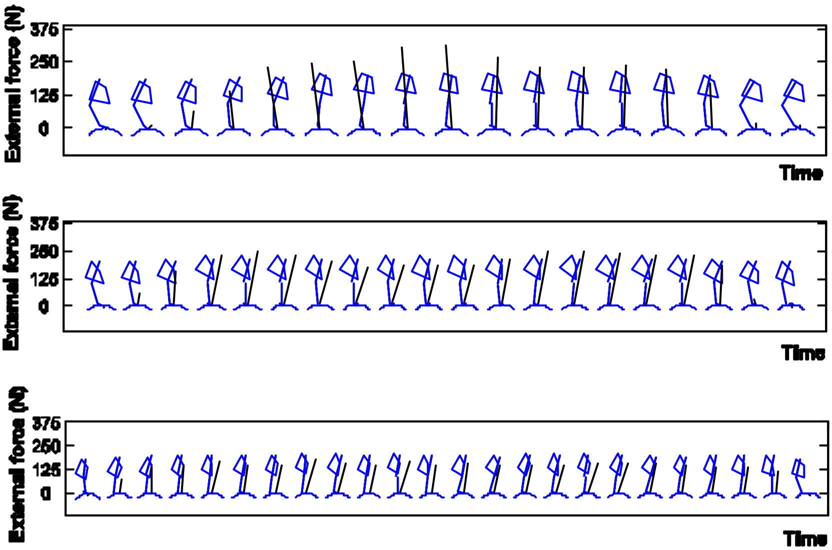


Figure 5.2: Typical example of the kinematics and kinetics of the performance of a weight-relief lift by an able-bodied profile (top panel), a tetraplegic profile without elbow locking (middle panel) and a tetraplegic profile with elbow locking (bottom panel). The stick figures represent the trunk and the arm with the hand on top of the handrim. The solid black lines indicate the reaction force to the applied hand force.

Results

Success of simulations

Kinematic profiles for the AB and TP subjects differed, which manifested itself in differences in net moments as shown previously [175]. Stick figures of the sagittal view of three profiles show the differences in the task performance between the profiles. In the AB profile (Figure 5.2, top panel) the contact force is directed through the glenohumeral joint while in the TP profiles (Figure 5.2, middle and bottom panel) the contact force is directed anterior of the joint. Variation in the performance of the lift in the tetraplegic group is illustrated by the presence and absence of elbow locking (Figure 5.2, middle and bottom panel). The TP profile in the middle panel was not able to come to a stabile position at once, but comes to the end position in two phases.

Lesion level

At the C5 lesion level, the model was unable to complete the task for all profiles (Figure 5.3). At the C6 lesion level seven simulations were successful for the tetraplegic profiles compared to only four simulations for the able-bodied profiles. However, at the C6 level and over all the lesion levels (C6-T1) no differences were found in the number of successful simulations (P>0.696).

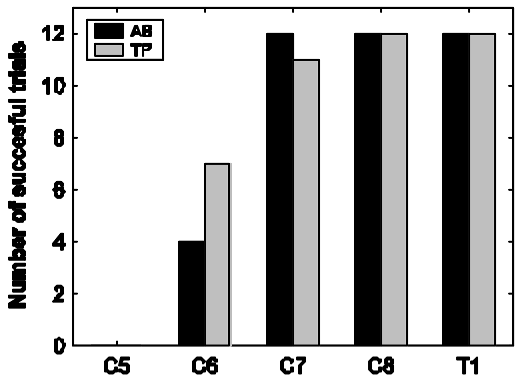


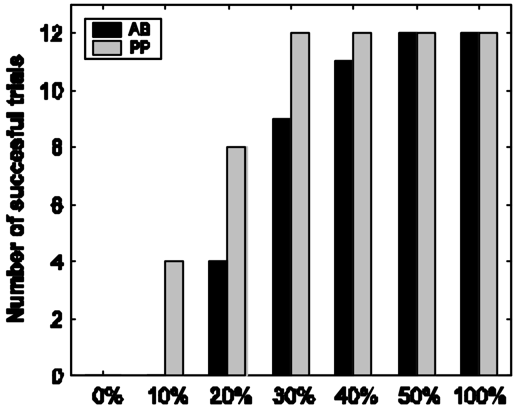
Figure 5.3: Number of successful trials out of 12 profiles per group (4 AB and 4 TP subjects x 3 trials each) for simulations with different lesion levels.

Triceps force

For both tetraplegic and able-bodied profiles without triceps and anconeus force, the model was unable to complete the task because the model was not able to satisfy the moment constraint (Figure 5.4). Overall, no differences were found in the number of successful simulations (P>0.05). However, at the lower force levels

(10 - 40%), significantly more simulations were successful for tetraplegic profiles compared to able-bodied profiles (*P*=0.012).

Figure 5.4: Number of successful trials out of 12 profiles per group (4 AB and 4 TP subjects x 3 trials each), for the simulations with reduced triceps force. At the lower force levels (10 - 40%), significantly more simulations were successful for tetraplegic profiles compared to able-bodied profiles (*P*=0.012).

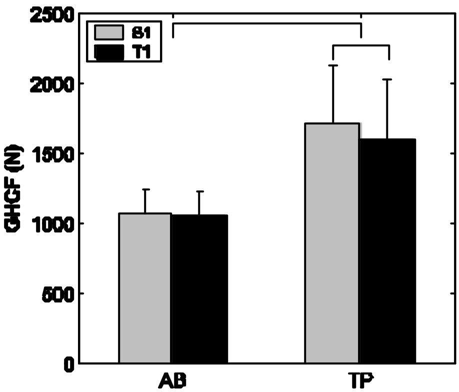


Glenohumeral contact force

Lesion level

The peak glenohumeral contact force was significantly higher for the tetraplegic profiles compared to the able-bodied profiles (*P*=0.037), the mean values were respectively 1656 ± 437N vs. 1062 ± 177N for the tetraplegic and able-bodied profiles (Figure 5.5).

Figure 5.5: Mean values and standard deviations of the glenohumeral contact force (GHCF) for the able-bodied (n=12) and tetraplegic profiles (n=12) for both the first successful simulation (S1) and the complete model simulation (T1). **?** = Significantly different (*P*<0.037).



There was also a significant difference between the highest successful lesion level (1390 ± 464N) and the T1 lesion level (1327 ± 436N) simulation (*P*=0.006). Further there was a significant interaction effect (*P*=0.013) for the profiles and the simulations, indicating that for the tetraplegic profiles the glenohumeral contact force for the highest successful lesion level was higher compared to the glenohumeral contact force for the T1 lesion level. No differences were found among the trials.

The direction of the glenohumeral contact force was (borderline) not significantly different between the highest successful lesion level and the T1 lesion level (*P*>0.061) or between the tetraplegic and able-bodied profiles (*P*>0.269). The direction of the glenohumeral contact force was on the whole in the anterior superior quadrant of the glenoid (Figure 5.6).

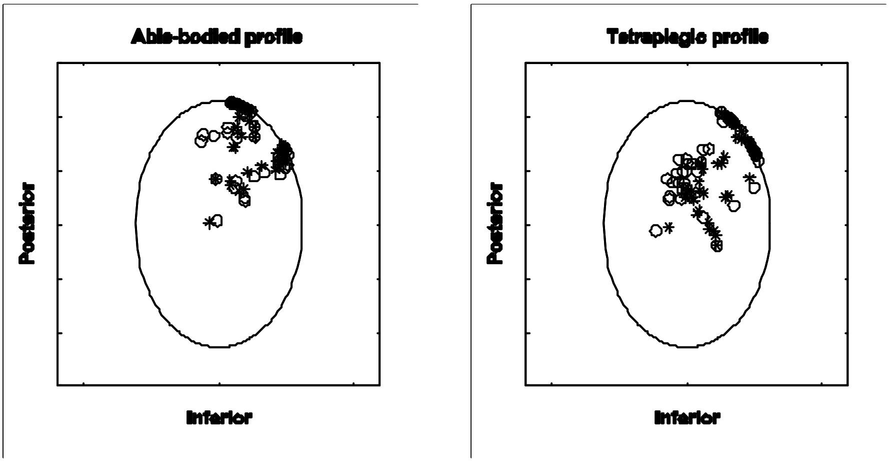


Figure 5.6: Typical examples of the position of the glenohumeral contact force on the glenoid surface for an able-bodied profile and a tetraplegic profile. The dots represent the C6 lesion level and the stars represent the T1 lesion level.

Triceps force

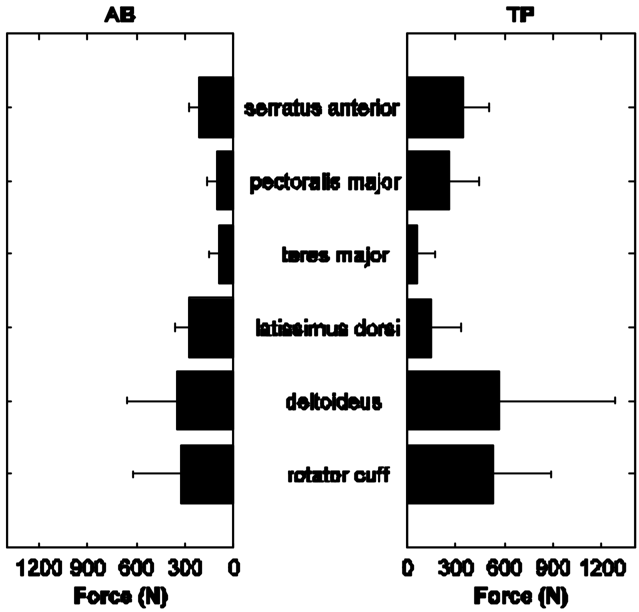
For the peak glenohumeral contact forces there were no significant differences between the able-bodied and the tetraplegic profiles (*P*=0.067), the mean values were 1070 ± 184N vs. 1613 ± 463N, respectively for the able-bodied and tetraplegic profiles. The difference in the glenohumeral contact forces between the lowest successful triceps force level and the 100% triceps force simulation was not significant. Also no significant differences were found for the trials (*P*>0.3).

Muscle forces

In the absence of muscle paresis, differences in the glenohumeral contact forces were already found between the able-bodied and tetraplegic profiles, therefore only the predicted forces for the T1 simulations are presented (Figure 5.7).

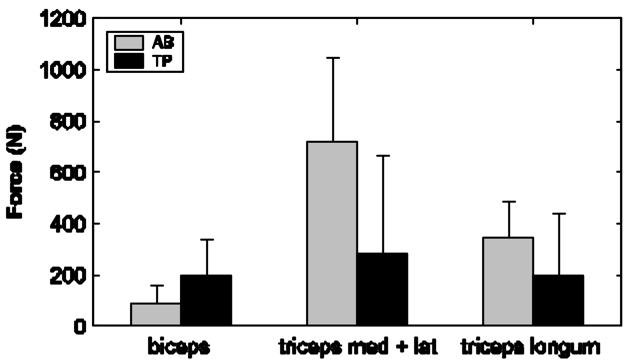
Higher forces were calculated for the tetraplegic profiles in the serratus anterior, the pectoralis major, the deltoideus and in the rotator cuff compared to the able-bodied profiles. However, these muscle forces were not significantly higher in the tetraplegic profiles compared to the able-bodied profiles (*P*>0.144).

Figure 5.7: Mean values and standard deviations of absolute muscle forces of six important muscles to perform the lift for the able-bodied and tetraplegic profiles during the T1 simulations.



Also the calculated forces for the biceps and triceps (Figure 5.8) seemed to show much more predicted force in the triceps for the able-bodied profiles compared to the tetraplegic profiles. However, these differences were not significant (*P*>0.160).

Figure 5.8: Mean values and standard deviations of the absolute biceps and triceps muscle forces for the able-bodied and tetraplegic profiles at the T1 lesion level.



For the model modifications C6 - T1, the effect on the rotator cuff can be seen in figure 5.9. For the tetraplegic profiles more force was calculated in the rotator cuff compared to the able-bodied profiles (*P*=0.023), but no differences were found for the different lesion levels. For the separate muscles of the rotator cuff a significantly higher force was predicted in the supraspinatus for the tetraplegic profiles (*P*=0.016) and a significantly higher force was predicted for the C6 level compared to the C8 and T1 lesion levels (*P*=0.040).

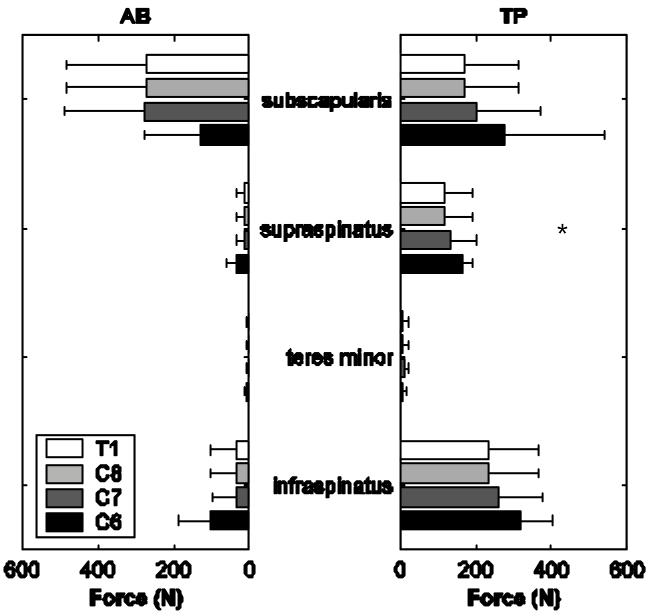


Figure 5.9: Mean values and standard deviations of the absolute muscle forces of the muscles of the rotator cuff. Forces for both the able-bodied and tetraplegic profiles for the different model modifications (lesion level C6 to T1) are presented.
* For the supraspinatus a significantly higher force was predicted for the tetraplegic profiles than for the able-bodied profiles (P=0.016).

Discussion

Lesion level

Results showed that it is possible to have a successful simulation of the performance of a weight-relief lift with a complete C6 spinal cord injury, but not with a C5 lesion. In real life this appears indeed to be the case: only rarely an individual with a C5 lesion is able to perform a weight-relief lift independently [153]. All persons with a C6 lesion are usually capable of independently performing a weight-relief lift, although there is considerable variation in the overall independence of persons with a C6 lesion. The use of the shoulder muscles is a determining factor in a these subjects' independence [46].

At the C6 lesion level, fewer able-bodied profiles were successful compared to the tetraplegic profiles, which suggests an useful adaptation in the kinematics of the subjects with TP. The able-bodied profiles were only successful at a lower lesion level, in accordance with a less modified model. To simulate a C6 lesion the triceps force in the model was changed to 33% of the maximum force. This percentage was similar to the percentage at which most simulations were successful in the triceps study (30% triceps force), However, in the simulated C6 lesion more muscles were scaled down.

Since the glenohumeral contact forces were not much affected by the triceps force, the combination with other adaptations must have been responsible for successful lifts. The model lesion modifications had an effect on the glenohumeral contact force, but this effect was small; for the tetraplegic profiles the glenohumeral contact force was only 7.3% lower in the T1 simulation compared to the S1 simulation. We expected the effect to be larger, but it is well likely that the adaptations in the TP kinematics and hand forces already compensated for the loss of muscle force and the loss of muscle function by the use of different muscles. It even appears that because the effect of lesion modification is so small that the TP subjects performed the weight-relief lift in the most economic manner given their lesion level.

The direction of the glenohumeral contact forces did not differ significantly between both profiles, and simulated lesion levels, for all conditions, this force was mostly directed through the forward-upper quadrant of the glenoid. This upward directed glenohumeral contact force could cause a reduction of the subacromial space with subacromial disorders as a result.

Triceps force

In a previous study [175] we found that the elbow moment was lower (39%) for TP subjects compared to AB subjects. The different movement pattern, featuring locking of the elbow to relieve the triceps, was visible in the number of successful simulations for the tetraplegic profiles and in the amount of triceps force needed. The able-bodied profiles needed much more triceps force at the T1 lesion level to generate the elbow extension moment (Figure 5.8) and had less successful simulations at lower force levels. For the able-bodied profiles the triceps weakness appeared to be the most important factor for failing the simulations.

But what caused the difference of 550N in the glenohumeral contact force between the able-bodied and tetraplegic profiles? This force could not be explained by difference in body mass, or difference in size and direction of the applied external force (expressed as % BW). Probably the difference not only came from compensation in the shoulder but from support of the trunk and lower extremities as well. The subjects with TP needed more external stabilization to maintain balance and to perform the task. Detailed information about the movement and positioning of the thorax during the task could give more insight into this compensation mechanism. Further, with the model we were able to simulate SCI in terms of shoulder and elbow function. However, persons with a high-level SCI also suffered loss of hand function. The loss of hand function could have had an effect on the kinematics of the hand and wrist during the task, changing the results of this study at the level of shoulder and elbow. An extended shoulder-elbow-hand model will be needed to estimate these effects.

Muscle forces

Only few studies reported on the muscle activity of weight-relief lifting in individuals with SCI [60, 109, 134]. Comparison of these studies to the current study is complicated by the difficulties of translating electromyography (EMG) to force. Reyes [134] and Newsam [109] both measured high activity (>50% MVC) in the latissimus dorsi in subjects with low-level paraplegia. The latissimus dorsi can elevate the trunk in combination with a fixed and stable arm by depressing the shoulder girdle. Harvey and Crosbie [60] found that some subjects with TP showed high muscle activity in the latissimus (63% MVC), but that not all subjects had full function of this muscle and did not depend on it. Apart from high latissimus activity, Newsam [109] measured high to moderate activity in the anterior deltoid, the rotator cuff and the serratus anterior in subjects with a C6 lesion. These results were comparable to the results in the current study where these muscles, except for the latissimus, showed more force for the TP subjects. For the performance of a weight-relief lift, it seems that the function of the latissimus and the triceps are taken over by the pectoralis major, the deltoid and the serratus anterior in subjects with TP.

A critical issue in this study is whether the predicted forces of the model are correct. The maximum triceps force for the tetraplegic profiles was almost 485N (summed force of medial, lateral and longum). This value was much lower than the maximum force in the able-bodied profiles (1055N) but was still 11% of the maximum triceps force in the (adjusted) model. This percentage is close to the percentage measured by Needham-Shropshire et al. [107]. They reported that subjects with a manual muscle test score of 3/5 for the elbow extension only had 9% of the maximum voluntary force production of healthy controls. In the study of Newsam for subjects with a functional elbow extension the EMG intensity ranged from 32% to 50% maximum voluntary contraction (MVC). However, especially for TP subjects, the EMG intensity in terms of % MVC is of little value without information about the moment that can be generated at 100% MVC. Information about muscle force instead of muscle activity is an important reason to use a biomechanical model.

Methods

With the Delft Shoulder and Elbow Model it was possible to obtain insights into the changes in shoulder load when simulating different lesion levels. However, the model is based on a single set of anthropometry and therefore the effects of inter-individual differences in force or anthropometry are not accounted for. We only measured four subjects with TP, each of them with a different lesion level, completeness of the lesion and triceps function. It is obvious that this will limit the application of these results to the whole TP population. Also variations in motor level innervation would likely have had an effect on the force capacity of shoulder and arm muscles relative to lesion level. Although some information is available about the segmental innervations of muscles the literature shows no consensus on this issue [77]. This implies that the amount of force we assigned to each lesion level is arbitrary, although due to neurological variability all classifications are in fact assumptions.

The regression equation of Pascoal [118] used to calculate the orientation of the scapula and clavicula was based on the motions of the scapula and clavicula of healthy subjects and may not be the best one for subjects with a SCI. Another regression equation, formulated by Koontz et al. [84], was based on the scapular movement of subjects with a low thoracic SCI. Six Dijkstra [150] showed that the scapulo-humeral rhythm was not different for able-bodied subjects and those with a low-level spinal cord injury, but assumed that this would not be the case for subjects with higher lesion levels. Although a sensitivity analysis indicated that in this study the effect of scapula orientation was quite small, it might be advisable to use directly recorded scapula orientation values whenever possible.

Conclusions

The glenohumeral contact force was 56% higher for tetraplegic profiles when compared to able-bodied profiles. The glenohumeral contact force was mainly higher due to different task performance, it even appears that the subjects with tetraplegia perform the weight-relief lift in a most economic, but different manner, given their lesion level.

The simulation of specific lesion levels had a minor effect (0.9 - 7.3%) on the calculated glenohumeral contact force, the effect was higher for the simulated SCI lesions compared to the simulations with a modified triceps force. Only for able-bodied profiles the triceps force seemed to be an important factor, for the tetraplegic profiles only 30% of the maximum relative triceps force was needed for a successful weight-relief lift.

Using the complete model (T1), it could not be shown that the predicted muscle forces in the rotator cuff, the deltoideus, the pectoralis major and the serratus anterior were higher for the tetraplegic profiles than for the able-bodied profiles. Within the group of the rotator cuff muscles, significantly more force was predicted in the supraspinatus for the tetraplegic profiles.

Due to the higher load on the shoulder joint and shoulder muscles in subjects with tetraplegia, these subjects run a higher risk of muscle overload and damage to the shoulder joint.

Acknowledgement

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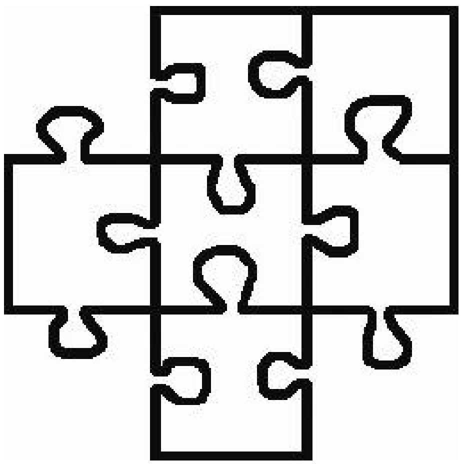
Chapter 6

Risk for impingement during weight-relief lifting



Abstract

Background and Purpose: Shoulder pain is a common phenomenon among wheelchair users, and might be related to impingement due to specific wheelchair tasks such as a weight-relief lift. The aim of this study was to analyze whether the weight-relief lift might be causative of shoulder impingement.

Subjects: Five subjects with paraplegia and six able-bodied subjects.

Methods: During weight-relief lifting, kinematic, kinetic and electromyographic data were collected. In addition, subjects underwent static strength tests of the abductors, adductors, internal and external rotators. Reduction of the of subacromial space and muscle activity were quantified as possible contributing factors to impingement.

Results: No significant differences were found for the muscle force ratios between the able-bodied subjects and the subjects with paraplegia. Triceps caput longum, pectoralis pars thoracalis and teres major showed significantly higher activity for the subjects with paraplegia compared to the able-bodied subjects. No reduction of the subacromial space was found, and no relations were found between muscle activity and reduction of the subacromial space.

Discussion and Conclusion: A causative relation between weight-relief lifting and shoulder impingement could not be established. The high inter-individual variability in the measured subacromial space and the fact that subjects without shoulder complaints were included could explain the fact that we did not find significant differences between the able-bodied subjects and the subjects with paraplegia. However, attention must be paid to the strength of the thoracohumeral muscles and muscle imbalance to prevent shoulder impingement in people with a spinal cord injury.

Introduction

Epidemiological studies have shown a high prevalence of shoulder complaints and shoulder pain in persons with a spinal cord injury (SCI). This in itself is a strong indicator of the effect of repetitive high loading. Sie et al. [148] reported that 55% of the patients with tetraplegia reported upper extremity (UE) pain, most commonly at the shoulder (46%). In subjects with paraplegia 64% of the patients reported UE pain, where carpal tunnel syndrome and shoulder pain were most commonly seen. Although shoulder pain may not initially limit the wheelchair user's ability to perform daily activities independently, it may have functional costs such as rapid fatigue, loss of endurance capacity or decreased speed of performance. Eventually wheelchair users with shoulder pain are forced to eliminate functional activities that are associated with pain [33]. In a review, Lee and McMahon [91] suggested that shoulder problems in persons with SCI begin with muscle imbalance, leading to glenohumeral instability, impingement syndrome, rotator cuff tears and subsequent degenerative joint disease.

Bigliani and Levine [10] reported that subacromial impingement syndrome has become an increasingly common diagnosis for patients who have shoulder pain. Nichols et al. [110] reported that in most cases (73%) the cause of shoulder pain in people with SCI involved soft tissue injury. As a result of the narrowing of the subacromial space, subacromial impingement syndrome affects the structures of the subacromial space, which are the tendons of the rotator cuff and the subacromial bursa [102]. Subacromial impingement syndrome can be classified as either intrinsic or extrinsic impingement [102]. Intrinsic impingement has been described as a result of the degenerative process that occurs over time with overload, tension overload or trauma to the tendons. Extrinsic impingement occurs as a result of mechanical compression by external structures. Etiological factors for intrinsic impingement are amongst others weakness of the scapular musculature and imbalance of the rotator cuff muscles.

It has been assumed that due to the unbalanced muscle action during daily wheelchair propulsion, a muscular imbalance of the shoulder joint might possibly develop [19, 103]. In wheelchair propulsion the primary muscles involved around the shoulder are the internal rotators and the adductors of the humerus [104] and one could expect to find higher forces for these muscle groups and lower abduction/adduction and higher exorotation/endorotation force ratios. An analogous development has been shown in swimmers [6] for whom it is also supposed to lead to shoulder complaints.

There have been previous studies about shoulder impingement related to wheelchair use, but these studies concentrated on the incidence of this pathology through MRI or radiography [17, 43, 52]. Only little is mentioned about the development of impingement in relation to wheelchair related tasks.

Reyes et al. [134] reported that making a weight-relief lift comprised of two biomechanical tasks in low-level paraplegia: trunk elevation and elbow extension. The thoracohumeral muscles elevate the trunk on a stabilized humerus, thereby transferring the load on the humerus directly to the trunk. Insufficient contribution of these muscles when making the weight-relief lift may create a reduction of the subacromial space.

The purpose of this study was to study: 1) the existence of a different force balance in subjects with a SCI versus able-bodied (AB) controls. 2) the contribution of the thoracohumeral muscles during a weight-relief lift and 3) the effect of weight-relief lifting on the in-vivo estimated magnitude of the subacromial space.

We hypothesize that the task properties of making a weight-relief lift aggravate the occurrence of etiological factors to the development of impingement. During weight-relief lifting there is an extreme downward pull on the body, which may create a reduction of the subacromial space and will be compensated by the thoracohumeral muscles.

From the considerations above four questions were formulated.

1) Are the force ratios of abduction/adduction and external/internal rotation decreased for subjects with a SCI in comparison to the AB subjects?
2) Is muscle weakness of the primary muscles, with regard to the weight-relief lifting, present in subjects with a SCI or AB subjects?
3) Does making a weight-relief lift lead to a reduction of the subacromial space? If so, is this reduction greater for subjects with a SCI than for AB subjects?
4) Is there a relationship between muscle activity and a reduction of the subacromial space?

Methods

Design

Eleven male subjects were measured during three standardized trials of weight-relief lifting. Kinematic, kinetic and electromyographic (EMG) data were registered during the task. The force of the abductors, adductors, internal rotators and external rotators were measured to analyze muscle strength balance.

Subjects

Five subjects with paraplegia (PP) and six able-bodied subjects (AB) gave written informed consent after an explanation about the nature of the study. The inclusion criteria for this study were that subjects were male and that they had no current shoulder complaints. Two PP had an incomplete lesion. A list of subject characteristics is given in table 6.1. The protocol of this study was approved by the Medical Ethical Committee of the Vrije Universiteit Medical Centre.

Table 6.1: Subject characteristics

	Age (y)	*Length (m)*	*Body Mass (kg)*	*Years after SCI (y)*	*Level of lesion*
PP (n=5)	*36.8 ± 1.9**	*1.87 ± 0.09*	*78.4 ± 8.8*	*18.4 ± 8.5*	*T3-T9*
AB (n=6)	*22.2 ± 2.8*	*1.82 ± 0.10*	*75.3 ± 6.6*	*NA*	*NA*

NOTE. Values are mean ± standard deviation or range (level of lesion)
Abbreviation: NA = not applicable.
* Significantly older compared with able-bodied subjects (P<0.05).

Instrumented wheelchair

All experiments were performed in a standard design Quickie Triumph™ (Sunrise Medical Benelux) wheelchair of which the right wheel was instrumented with a force-transducer (M6-1000, Advanced Mechanical Technology Inc.) [175]. This transducer measures six degrees of freedom: both forces and moments of the hand exerted on the handrim were measured in three directions (x, y, z). During the experiment the data were recorded on a Porti™ data logger (Twente Medical Systems) and stored on a Flash memory card. Forces and moments were sampled at 100Hz and were low-pass filtered using a 10Hz second-order recursive Butterworth filter.

Protocol

The strength of the adductors, abductors, internal rotators and external rotators of the glenohumeral joint were measured with a one degree of freedom force-transducer (AE Sensors). While the experiment leader generated a force in opposite direction, the subjects had to perform an adduction, abduction, internal rotation and external rotation. For the adduction, 90° abduction was used as starting position and for abduction the starting position was at 0° abduction. In both measurements the force-transducer was held on the forearm as close as possible to the elbow. For the internal and external rotation the starting position was 0° abduction and 90° of elbow flexion, while the force-transducer was connected to the wrist. Each muscle group was tested three times.

In addition, subjects underwent static strength tests to record the EMG during maximum voluntary contraction (MVC) of 15 muscles of the upper extremity (Table 6.2). Muscle activity was recorded with bipolar Ag/AgCl (Medicotest A/S) surface electrodes at a sample frequency of 1000Hz.

The weight-relief lift had to be performed with the hands on the handrims to measure the forces and moments of the right hand with the transducer. This task was performed three times with 20s of rest between the trials. During a trial the subjects had to lift themselves, hold for a few seconds in lifted position and then lower themselves to the seated position (see Figure 6.1).

Table 6.2: Recorded muscles and electrode placement

Muscles	*Abbreviation*	*Electrode location*
m. biceps brachii caput longum	*BL*	*1/3 from fossa cubiti on the line between acromion and fossa cubiti*
m. triceps brachii caput longum	*TL*	*In the midst between the crista acromion and the olecranon, two finger widths medial to this line*
m. triceps brachii caput laterale	*TLa*	*In the midst between the crista acromion and the olecranon, two finger widths lateral to this line*
m. brachioradialis	*BR*	*In the midst of the muscle belly*
m. pectoralis major pars clavicularis	*PC*	*In the midst between sternoclavicular joint and processus coracoideus, 2 cm below the clavicula*
m. pectoralis major pars thoracalis	*PT*	*In the midst of the muscle belly*
m. trapezius pars descendens	*TD*	*1/2 on the line between acromion and C7*
m. trapezius pars ascendens	*TA*	*2/3 on the line between trigonum spinae and T8*
m. deltoideus pars clavicularis	*DC*	*One finger width distal and anterior to the acromion*
m. deltoideus pars acromialis	*DA*	*On the line from acromion to epicondylus lateralis*
m. deltoideus pars scapularis	*DS*	*Two finger widths posterior to the acromion*
m. latissimus dorsi	*LD*	*1/2 on the posterior line of the armpit*
m. teres major	*TM*	*In the midst between trigonum spinae and angulus inferior, exo- and endorotation to differentiate from the m. infraspinatus*
m. serratus anterior	*SA*	*Anterior to the lateral scapular border, distal to the latissimus dorsi*
m. infraspinatus	*IS*	*In the midst between trigonum spinae and angulus inferior, 2 cm of the medial border of the scapula*

Kinematics

Kinematics of the upper body during the weight-relief lift were measured with Optotrak™ (Northern Digital) with a sample frequency of 100Hz. In total, we used 17 active markers on the right side of the subjects' body (thorax, upper arm, forearm, hand) [187], of which four markers were on a cuff that could be reshaped to fit around the upper arm (see Figure 6.1). With the markers on this cuff, the centroid of the humerus was determined. Further, two technical markers for the epicondylus medialis (EM) and three technical markers for the processus styloideus ulnaris (SU) were used (see Figure 6.1). Prior to the actual measurements, a calibration measurement was performed in which the orientation of the technical markers was defined relative to bony landmarks. The orientations of the processus styloideus radialis and the processus coracoideus were determined with a pointer. During these measurements, subjects were sitting in the wheelchair with the arms in the anatomical position. Optotrak, EMG and the force measurements were synchronized when collecting data.

Data processing

Of the force data measured during the static muscle strength tests, the peak force for the abductors, adductors, internal rotators and external rotators of the glenohumeral joint were determined for each test. The average value of the peak forces was used to calculate the muscle strength ratios (Q), representing muscle strength balance:

1) Qab/ad = (max. force abduction / max. force adduction)
2) Qab/exo = (max. force abduction / max. force exorotation)
3) Qab/endo = (max. force abduction / max. force endorotation)
4) Qexo/endo= (max. force exorotation / max. force endorotation)

To calculate the median intensity of the EMG, first the raw EMG data were corrected for movement artefacts by subtraction of the low pass (Fc = 5Hz) component of the signal. The data were rectified and filtered with a recursive low-pass filter of 4Hz to create a linear envelope and finally the data were expressed as a percentage of the MVC.

The lift was divided into three phases using a combination of kinetic and kinematic data. The raise phase started when force was applied to the handrim and the acromion moved upward. The lift phase started when the force was maximal and the velocity of the acromion was around zero. The recovery phase began when the acromion lowered and the applied force decreased again.

The recovery phase was not analyzed in detail because the recovery phase was a

Figure 6.1: Subject performing a weight-relief lift. The circle marks the cuff around the upper arm.

controlled lowering of the body by use of gravity and EMG activity was generally low [134].

Before processing the kinematic and kinetic data, a representative period of the lift phase was selected based on two criteria (Figure 6.2):

1) the vertical force had to be higher than 25% bodyweight (BW). The criterion of 25% BW was chosen because subjects will not lift their full bodyweight during a weight-relief lift because the legs are still resting on the edge of the seat and on the foot rest.
2) the velocity of AC had to be between -0.01 and $0.01 m \cdot s^{-1}$.

When the data met both criteria the subjects were assumed to be in the end position of the lift.

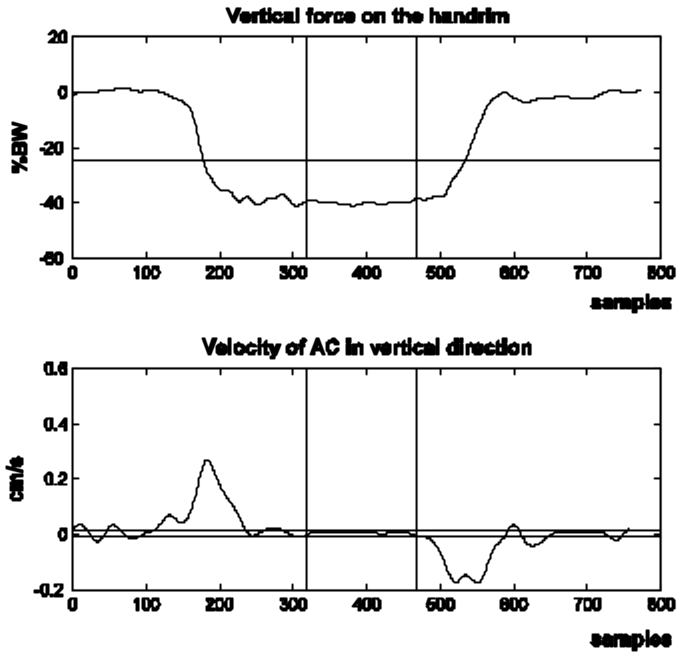


Figure 6.2: Typical example of the selection in time where the kinematic and kinetic parameters met the selection criteria. Upper plot: the vertical force had to be higher than 25% BW. Lower plot: the velocity of AC had to be around zero, between -0.01 and $0.01 cm \cdot s^{-1}$.

For the kinematic data missing values were interpolated, followed by the calculation of the landmarks EM, SU and the centroid of the humerus using the available technical markers. The reduction of the magnitude of the subacromial space was calculated as the difference between the distance EL_AC in the end position of the lift compared to the distance EL_AC in the anatomical reference position (? EL_AC). All kinematic parameters were expressed in millimeters.

The forces measured on the handrim of the instrumented wheelchair were expressed as a percentage of BW. Next to the magnitude of this force, the direction of the total force, expressed as the angle with the vertical, was determined.

Statistical analysis

To detect significant differences between the subject characteristics of the two subject groups, independent *t*-tests were applied.

Peak forces and force ratios, magnitude and direction of the exerted hand forces and kinematics were tested with a two-way factorial ANOVA (between-subject factor: 2 groups, within-subjects factor: 3 trials). For the distance AC_EL, the hand forces and direction of hand forces, this was followed by an intra-class correlation estimation over the three trials.

For the EMG, a three-way factorial ANOVA (one between-subject factor: 2 groups, 2 within-subjects factor: 2 phases (raise and lift phase) and 3 trials) was performed to find significant differences between the groups and between the phases.

A univariate regression analysis was performed on the averaged data over the three trials to determine if the mean EMG (triceps caput longum, triceps caput laterale, latissimus dorsi and pectoralis major pars thoracalis) could predict the alteration of the distance EL_AC. The total force (% BW) and test group (PP = 1, AB = 2) were added to the regression model as confounders. The level of significance was set at $P<0.05$ for all statistical tests.

Results

Subjects

The two groups did not differ significantly in size and weight, but subjects with PP were significantly older ($P<0.003$) compared to the able-bodied subjects.

Forces and force ratios

The force ratios calculated with the peak forces of the abductors, adductors, internal rotators and external rotators are given in table 6.3. For the peak forces a significantly higher abduction force was found for the able-bodied subjects than for the subjects with PP ($P=0.019$). No significant differences were found for the force ratios between the able-bodied subjects and the subjects with PP.

Table 6.3: Mean values for the peak forces and force ratios for the subjects with paraplegia (PP) and the able-bodied (AB) subjects.

		PP (n=5)	*AB (n=6)*
Peak force (N)	*Abduction*	*199.7 ± 31.0*	*250.1 ± 35.4**
	Adduction	*281.8 ± 72.1*	*344.5 ± 61.3*
	Exorotation	*122.8 ± 31.1*	*139.4 ± 22.9*
	Endorotation	*214.6 ± 45.4*	*224.1 ± 30.5*
Force ratio	*Qab/ad*	*0.74 ± 0.16*	*0.75 ± 0.18*
	Qab/exo	*1.70 ± 0.38*	*1.83 ± 0.32*
	Qab/endo	*0.97 ± 0.26*	*1.14 ± 0.25*
	Qexo/endo	*0.60 ± 0.19*	*0.63 ± 0.12*

NOTE: values are mean over trials and subjects ± standard deviation.
* Significantly higher compared with subjects with paraplegia (P<0.05)

EMG

With respect to the EMG-parameters significant differences were found between the groups and between the phases (Table 6.4). The activations of triceps caput longum and the triceps caput laterale were significantly higher during the raise phase than during the lift phase while the trapezius pars ascendens had higher activity in this phase. The triceps caput longum, the pectoralis pars thoracalis and the teres major showed higher activity for the subjects with PP compared to the able-bodied subjects (Figure 6.3). During the raise phase moderate activity (>25%) was found in the latissimus dorsi, the triceps caput longum and caput laterale, the teres major and the pectoralis major pars thoracalis. During the lift phase only the latissimus dorsi, the triceps caput laterale and the pectoralis major pars thoracalis showed moderate activity.

Table 6.4: Peak EMG for 15 muscles during the raise and lift phases of the weight-relief lift

	Raise		*Lift*	
Muscle	*PP*	*AB*	*PP*	*AB*
BL	*4.0 ± 1.9*	*3.3 ± 2.5*	*4.8 ± 2.8*	*8.5 ± 8.3*
BR	*4.1 ± 1.9*	*2.9 ± 1.9*	*4.7 ± 2.8*	*3.6 ± 2.3*
DA	*8.2 ± 5.0*	*4.7 ± 1.7*	*8.6 ± 5.6*	*6.5 ± 2.3*
DC	*8.6 ± 7.3*	*3.9 ± 1.6*	*9.8 ± 9.2*	*5.6 ± 2.6*
DS	*4.6 ± 2.9*	*2.7 ± 1.5*	*3.7 ± 1.3*	*2.6 ± 1.8*
IS	*16.9 ± 13.3*	*9.6 ± 4.7*	*16.4 ± 11.4*	*14.6 ± 6.9*
LD	*36.4 ± 27.0*	*25.3 ± 10.3*	*33.0 ± 15.8*	*31.2 ± 12.4*
PC	*6.3 ± 3.3*	*4.0 ± 2.0*	*6.5 ± 5.8*	*4.2 ± 2.5*
PT	31.0 ± 16.5	16.9 ± 4.6	30.0 ± 11.6	21.7 ± 12.7
SA	*12.2 ± 8.9*	*5.7 ± 3.7*	*12.3 ± 9.9*	*7.5 ± 2.7*
TA	***16.8 ± 9.7***	***13.2 ± 3.2***	***18.7 ± 10.9***	***20.7 ± 5.0***
TD	*2.6 ± 1.4*	*3.5 ± 1.9*	*2.5 ± 0.9*	*3.1 ± 0.8*
TL	**39.0 ± 6.1**	**26.6 ± 13.2**	**22.4 ± 8.7**	**9.9 ± 6.6**
TLA	***41.0 ± 15.4***	***20.3 ± 8.4***	***27.2 ± 15.5***	***14.1 ± 9.9***
TM	25.0 ± 17.6	4.9 ± 2.8	21.4 ± 15.3	5.6 ± 4.4

Bold: significant differences between the phases, Italic: significant differences between the groups
NOTE: values are mean values (in %MVC) over trials and subjects ± standard deviation.
Muscle abbreviations are listed in table 6.2.

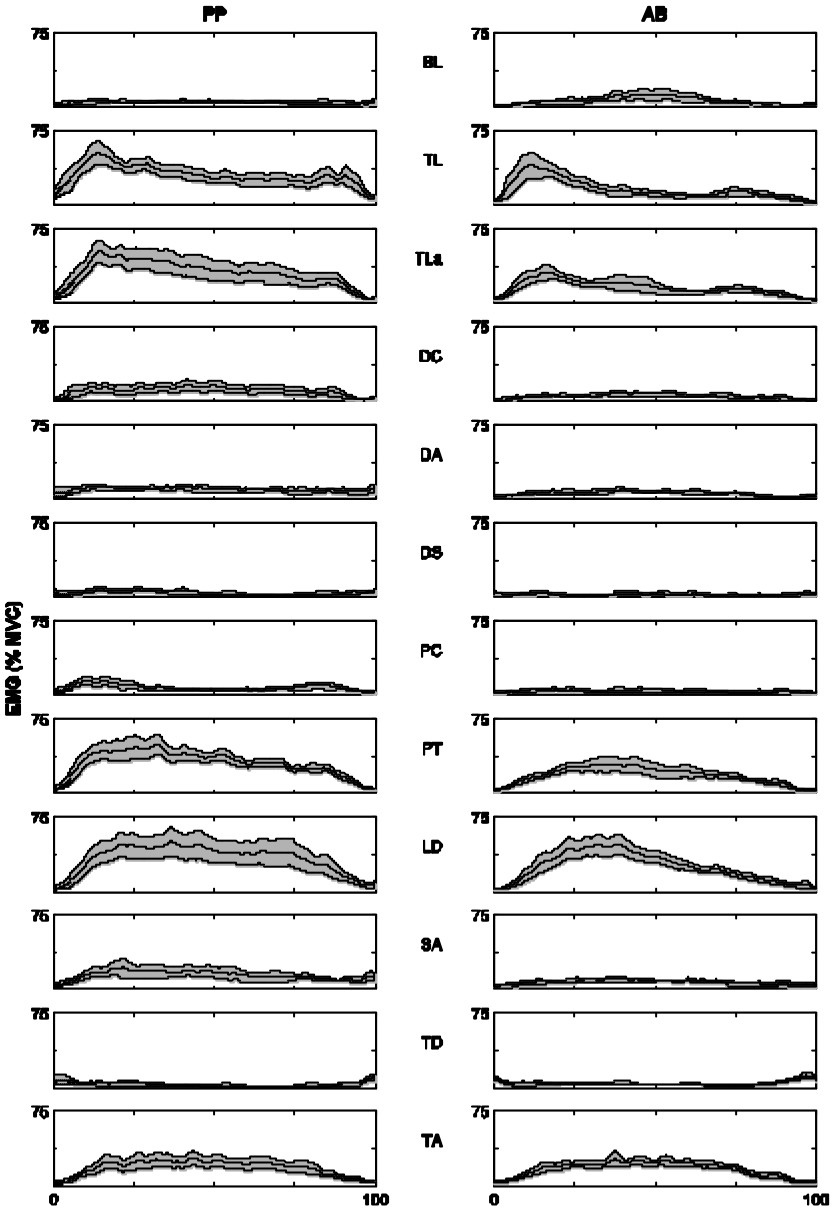


Figure 6.3: Mean EMG (% MVC) and the standard error of the mean of 12 muscles (see Table 6.2) for subjects with paraplegia (PP) and able-bodied subjects (AB). Time is normalized to 100% of the weight-relief lift.

Kinematics and Kinetics

The alterations in the distance EL_AC showed a high consistency (ICC > 0.89). No significant differences were found in the changes of the distance EL_AC between trials or between groups.

The values of ? EL_AC were 1.5 ± 1.3mm and -0.5 ± 2.3mm respectively for the subjects with PP and able-bodied subjects. Although no significant differences were found between the groups, there was a high variability within the groups (Figure 6.4). For seven subjects (5 PP, 2 AB) the distance EL_AC increased compared to the distance measured in the anatomical position (1.6 ± 1.3mm) and for four subjects (4 AB) EL_AC decreased (-1.7 ± 1.7mm).

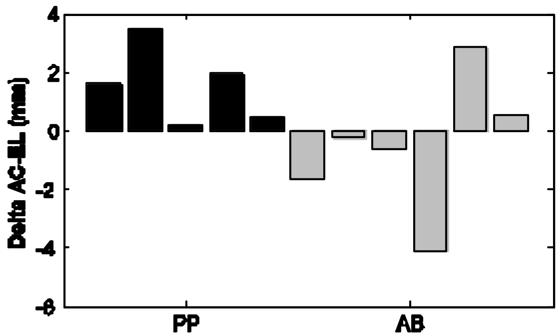


Figure 6.4: ?EL_AC: difference (in mm) between the vertical distance measured during the weight-relief lift and the values determined in the anatomical position. Mean values over the trials are presented for each subject.

Also for the kinetic parameters the three trials of the weight-relief lift were performed consistently. The intraclass correlation showed an alpha of 0.89 for the total force and an alpha of 0.97 for the direction of this force.

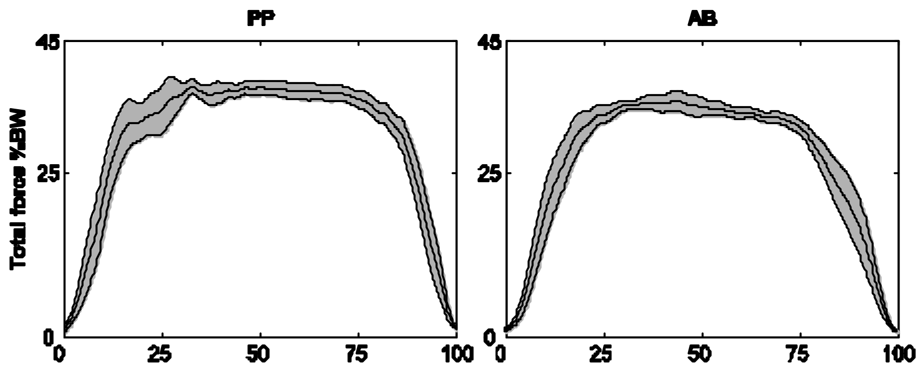


Figure 6.5: Mean total force (% BW) and the standard error of the mean for subjects with paraplegia (PP) and able-bodied subjects (AB). Time is normalized to 100%.

The force on the handrim is the force the subjects applied on one side of the wheelchair (Figure 6.5). The total force expressed as a percentage of body weight and the direction of this force did not differ significantly between the three groups (Table 6.5).

Table 6.5: Force during the weight-relief lift

		PP (n=5)	*AB (n=6)*
Total Force	*(%bodyweight)*	*37.7 ± 3.2*	*34.8 ± 2.5*
Direction of the force	*(degrees)*	*6.6 ± 4.1*	*8.7 ± 5.2*

NOTE: values are mean values over trials and subjects ± standard deviation

The regression analyses showed that there were no significant relations between ? EL_AC and the activity of the triceps caput longum, triceps caput laterale, pectoralis major pars thoracalis or latissimus dorsi when added as univariate predictors. Also no relations were found when the exerted hand force (expressed as % BW) or group (PP or AB) was added as a confounder.

Discussion

The aim of this study was to determine the existence of a different force balance in subjects with a SCI versus able-bodied subjects and to the quantify muscle activity and subacromial space changes when making a weight-relief lift.

The force ratios found in this study were comparable to the values found by Kotajarvi et al. [85] and Burnham et al. [19], despite the different methods used. Burnham et al. [19] found a significantly higher abduction/adduction ratio for athletes with paraplegia in comparison to a group of able-bodied athletes due to relative weakness of the adductors. However, when the paraplegic subject group was divided in a group with impingement syndrome and a group without, they did not find significant differences between the group without impingement syndrome and the able-bodied men. The group with impingement showed muscle imbalance, emphasizing muscle imbalance as factor for intrinsic impingement. The results also seem to agree with the results found in this study. In this study, only subjects without shoulder problems were selected and there were also no significant differences found between the able-bodied group and the paraplegics.

The results of the EMG recordings resembled the results of Reyes et al. [134] and Newsam et al. [109]. However, both studies found high activity (>50% MVC) in the latissimus and the triceps while we only found moderate activity in these muscles (>25% MVC). We did find differences in the EMG activity between the subject groups in the triceps, the pectoralis and the teres major. The higher triceps activity in the PP group corresponds to the higher internal extension moment needed in the elbow [175]. The higher activity in the thoracohumeral muscles might be needed to protect the rotator cuff from impingement.

Not surprisingly, differences in EMG activity between the phases were found. Like Reyes et al. [134], higher triceps activity was found in the raise phase than during the lift phase. The higher activity in the trapezius during the lift phase could have been responsible for holding the scapula on the thorax in this position. Also the serratus anterior superior and the rhomboideus are responsible for holding the scapula on the thorax. As a result one should expect a significant difference for the serratus anterior as well. However, we did not find a difference and the activity of the serratus was low. Also in a previous modeling study [174] low forces for these muscles were calculated. Despite the fact that we only measured a small part of the serratus anterior, it seems that a lot of additional force is not necessary.

In this study we could not find a reduction of the subacromial space during the performance of a weight-relief lift. During a weight-relief lift there is an extreme downward pull on the body, which might create a reduction of the subacromial space. The inter-individual variability found within the groups was high while the extremes for elongation or depression were still surprisingly small. Also the main effect was small; only a mean increase of 0.1mm (± 2.1mm) compared to the anatomical distance. This inter-individual variability can be compared to the variability found in the study of Graichen et al. [52] who compared individuals with shoulder complaints with healthy subjects. Because of the high inter-individual variability they suggested that it might be better to compare the affected shoulder with the healthy shoulder of the same subject. However, in studies with persons with a SCI this does not seem to be very useful since often both shoulders might be affected and certainly both shoulders have been exposed, because subjects rely on both shoulders for wheeling and different ADL. Despite not finding a significant decrease in the subacromial space, compression of the subacromial tissues (extrinsic impingement) due to the performance of the weight-relief lift is possible because for some subject a reduction was found.

The subacromial space is normally between 6 and 14mm [50] while the changes in the subacromial space during movements are in the order of millimeters [51]. Distances in this order of magnitude should be measurable with Optotrak because of its accuracy (RMS accuracy: x and y, 0.1mm; z, 0.15mm). Although one should take displacement of the markers due to skin movement, which can be considerable, into account [93, 145], it should be kept in mind that here in both conditions, the arm was approximately extended, which will minimize the effect of joint rotation on the marker displacements relative to their anatomical landmarks. In addition, markers were placed on the epicondylus lateralis and the acromion, two bony landmarks where skin movement due to soft tissue deformation is likely to be small.

An explanation for not measuring a decrease in the subacromial space might be found in the performance of the task and the activity of muscles. Graichen et al. [53] and Hinterwimmer et al. [65] found that activity of the adductor muscles reduced the decrease of the subacromial space in contrast with activity of the abductor muscles. During the performance of a weight-relief lift one expects high activity of the latissimus dorsi, pectoralis major and teres major. The activity of these thoracohumeral muscles compensate for the downward pull on the body and might cancel the decrease in the subacromial space [109, 134].

When these muscles counteract on the decrease in subacromial space, one would expect to find a relation between the reduction of the subacromial space and the activity of these muscles. However, no such relation was found, not even when compensated for the amount of mass lifted or for spinal cord injury.

Conclusions

The men with paraplegia did not show muscle weakness of the thoracohumeral muscles relative to the able-bodied men, but showed higher EMG activity for the triceps caput longum, the pectoralis pars thoracalis and the teres major. With the used method we did not find a reduction of the subacromial space during the performance of a weight-relief lift. However, the inter-individual variability was high. No relations were found between muscle activity and reduction of the subacromial space.

In light of the influence of the task properties of making a weight-relief lift on shoulder impingement, both strength and balance of the thoracohumeral muscles deserve increased attention during the rehabilitation.

Acknowledgement

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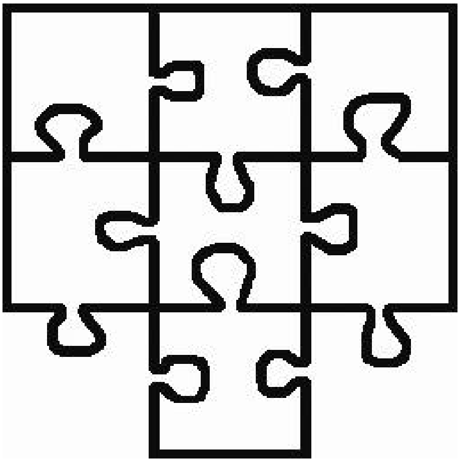
Chapter 7

Epilogue



Mechanical load on the upper extremity in subjects with a spinal cord injury

The aim of this thesis was to acquire an understanding of the underlying mechanisms of the development of overload injuries in the upper body musculoskeletal system in subjects with a spinal cord injury (SCI). Besides the progression of pain complaints during rehabilitation, different loading variables of wheelchair propulsion and specific wheelchair-related activities of daily living (ADL) were studied, namely net moments, muscle forces and joint reaction forces. Both wheelchair propulsion and wheelchair-related tasks are generally seen as risk factors for the development of overuse complaints [7, 14].

Our assumption for the occurrence of damage to shoulder structures was that peak loads during ADL cause damage which can not heal because of the continuous loading of the shoulder during wheelchair propulsion. In this study it was found that the peak loads during ADL are high and probably high enough to cause damage to structures of the shoulder (Chapters 3 and 4). However, another model for shoulder damage might be that the structures of shoulders are weakened through the continuous load of wheelchair propulsion. A peak load during a wheelchair-related ADL could then easily cause damage to the weakened structure (Figure 7.1). Peak loads are thus only part of the total problem and submaximal loads could be just as harmful.

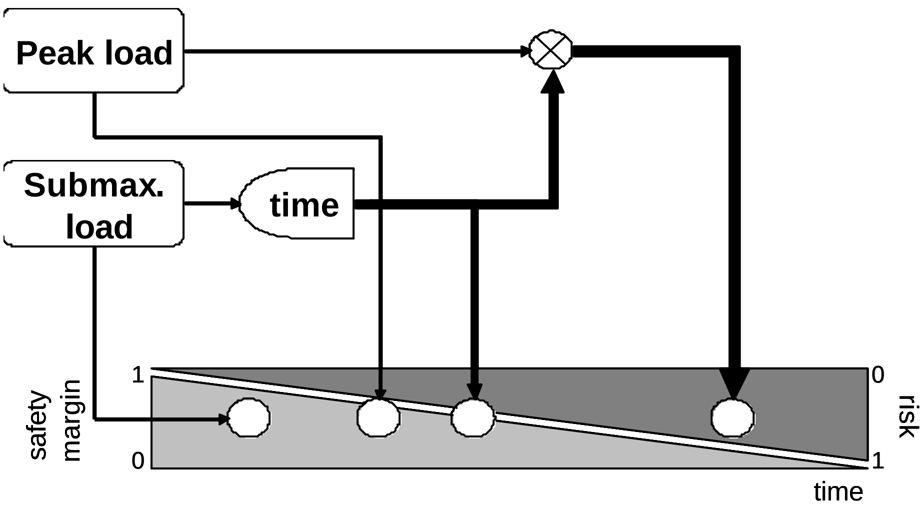


Figure 7.1: Schematic drawing of the relationship between submaximal load and peak load and the effect on the occurrence of damage over time.

This thesis shows that wheelchair propulsion in itself is not that stressful in terms of peak load when compared to wheelchair-related ADL (Figure 7.2). The load of the straightforward, steady state level wheelchair propulsion in the current study will be lower compared to every day wheelchair propulsion where the propelling conditions are less optimal (starting, stopping, turning, obstacles, floor surface etc) [171]. Further, it became clear that reaching is almost twice as heavy as normal level wheelchair propulsion and that weight-relief lifting can be considered to be a heavy task since glenohumeral reaction forces exceeded 1500N. Whether this also implies that level wheelchair propulsion should not be seen as a serious risk factor for upper extremity complaints will be discussed later.

In comparison to wheelchair users with full control over their upper extremities, subjects with tetraplegia have to compensate for their muscle paralysis in the arms and thorax with alternative muscles with possibly unfavorable torque components, which have to be compensated for by additional muscle force. Higher muscle forces and higher glenohumeral reaction forces were expected and also found for subjects with tetraplegia compared to able-bodied subjects (+40%) and subjects with paraplegia. The influence of simulated lesion level was not as high as expected; for a C6 lesion level this resulted in only 7% higher glenohumeral reaction forces compared to the T1 lesion level (Chapter 5). The simulation study showed that the different technique of the weight-relief lift by subjects with tetraplegia was mainly responsible for the higher load.

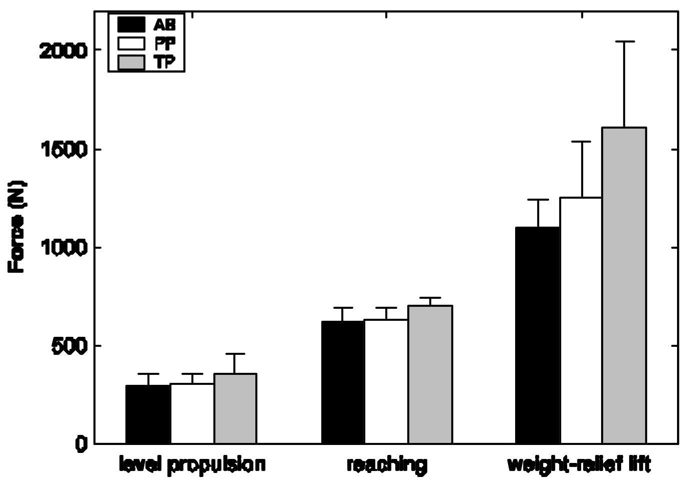


Figure 7.2: Peak glenohumeral reaction forces for level wheelchair propulsion, reaching and weight-relief lifting for able-bodied subjects (AB), subjects with paraplegia (PP) and tetraplegia (TP).

Risk factors for overload

Shoulder problems are a common problem in the general population. The prevalence of these problems is around 10% for persons under the age of 50 and up to 25% for elderly persons. It seems that factors that contribute are physical load and psychosocial work environment [168]. For certain athletes (incl. swimmers and pitchers) the prevalence of shoulder pain is much higher [6, 94]. The repetitions of the movement together with a muscular imbalance around the shoulder are seen as the main factors for shoulder injuries for these athletes.
Risk factors for damage due to wheelchair propulsion or wheelchair-related tasks can be grouped in three domains; individual factors, environmental factors and work requirements (Figure 7.3) [61]. Individual factors include the physical capacity, the posture and the skill level of the persons. Environmental factors determine the propelling conditions i.e. floor surface and the state of the wheelchair. Work requirements refer to the quality and the quantity of the wheelchair tasks. Next to the magnitude and the frequency of the applied load, other work requirements which are relevant to the development of musculoskeletal disorders at the upper extremity are the direction of force [15, 78], time of exposure and rest periods between activities [29, 130].

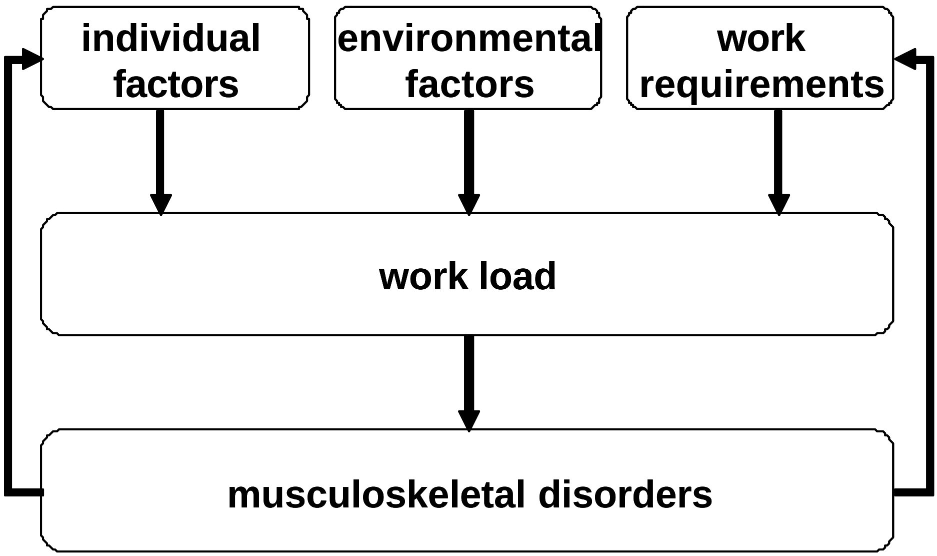


Figure 7.3: Conceptual model describing the relationship between individual factors, environmental factors, work requirements and the work load, possibly resulting in musculoskeletal disorders.

This thesis focused particularly on the peak loads that occur during wheelchair propulsion and wheelchair-related tasks. Peak loads can introduce risk moments for damage and are therefore a risk factor [2, 45, 157]. Of course, peak loads are only one aspect of the many factors that play a role in the occurrence of damage. According to ergonomic studies [2, 45], the frequency of the movement is an important factor for mechanical shoulder disorders. If that is the case, then the repetition of the movement makes wheelchair propulsion a stressful task in terms of mechanical load. Level wheelchair propulsion at 3 $km \cdot hr^{-1}$ was found to be a relatively low intensity task (±400N) [174], but this force is applied approximately 2700 times a day (60 minutes, 45 $strokes \cdot min^{-1}$). On the other hand, wheelchair tasks like lifting and transfers are not performed frequently during the day (approximately 15 transfers) [119], but the peak force is high (±1500N) compared to propulsion [174]. For example the loading effect of wheelchair propulsion can lead to a situation where the load of lifting or of a transfer can become critical, whereas damage occurring due to peak force may lead to chronic effects due to the low-level, high-frequency wheelchair propulsion (see Figure 7.1). At this point it is impossible to say what is more damaging, low-level force, with high-frequency or high-level force and low frequency, but it is not unlikely that both loading effects accumulate.

The activities performed by subjects in a wheelchair can be detected with an accelerometry-based activity monitor [20, 126]. Up till now 24-hour measurements are more common in back pain studies [186] and in physiological studies [70]. In combination with the recording of the activities performed during 24 hours, information about the load of these activities will give a loading profile of subjects with a SCI. These loading profiles will improve the understanding of the influence of frequency and duration in addition to peak loads on damage mechanisms.

Another task which is frequently performed by persons using a wheelchair is reaching. In the area of ergonomic research, overhead reaching is generally recognized as a risk factor for shoulder injuries [48, 62, 122, 151]. Overhead reaching or working is stressful for the arm-shoulder muscles since additional force is necessary to hold the scapula against the thorax, to position the glenoid upwards and to hold up the upper arm. For subjects with a high-level lesion reaching or working overhead is even more straining because of difficulties maintaining trunk stability. It is obvious that when persons in a wheelchair interact with an environment created for standing individuals, the majority of reaching will be overhead. Furthermore, for subjects with a SCI who have no active spinal extension, sit with posterior pelvic tilt, spinal flexion and rounded shoulders, the posture is not at all favorable for reaching tasks. According to Hastings [61], their reaching is impaired and they are even at a greater risk to joint damage. Appropriate seating and trunk support could provide a stable base for the upper extremities, improving the ability to reach [32, 82].

Biomechanical modeling: state of the art and the inevitable limitations

In this thesis a biomechanical model was used to predict the peak load on the upper extremity during wheelchair propulsion and wheelchair-related tasks. The Delft Shoulder and Elbow Model [163, 165, 179] is a unique finite element musculoskeletal model which represents all bones and joints of the upper extremity as well as 31 muscles divided into 139 muscle elements crossing the joints.

This model was not only developed to increase the functional insight in the mechanical behavior of the upper extremity (i.e. the role of the structures of the shoulder), but also to assist in diagnosis and treatment of disorders. To improve the functional outcome of a glenohumeral endoprosthesis the effect of design parameters (radius and thickness of the humeral component) and operation technique parameters (positioning of the humeral and glenoid component) on the muscular effort were investigated [41] as well as the effectiveness of tendon transfers for massive rotator cuff tears [97, 98]. Further, a goal of the model was to enable the estimation of the load on the morphological structures for analysis and prevention of injuries.

With the external forces and the 3D rotations of the thorax, clavicle, scapula, humerus and forearm as input variables, the model can be used to calculate the net moments, individual muscle forces and joint reaction forces in the structures of the shoulder and elbow.

Limitations

An often used argument against application of a model is that it is not an individual model. Indeed, the Delft Shoulder and Elbow model is based on the internal parameters of one cadaver [179, 182] and is not used as a personalized model. The model was applied to explore general tendencies in the behavior of the shoulder since individual analyses are extremely complex and not yet sufficiently applicable, whereas a general impression of effects would already produce relevant important information. The 3D kinematics of several trials of a number of subjects were used as input in our study and showed a clear difference between the subject groups. Other models may be better adjustable to subjects' characteristics, but these models have a limited output [24] or are at least partially based on the same morphological data (Anybody© (AnyBody Technology A/S, Denmark), SIMM© (MusculoGraphics Inc, USA). These models, with net moments as output, are probably easier to use in the rehabilitation centers because they are less complex and less labor-intensive. Under normal situations, the net moments correspond well with the joint reaction forces [129, 177]. In this thesis this relationship was not studied but for both the net shoulder moments and the glenohumeral reaction forces (Chapters 3 and 4), significant differences in the trends among the different tasks were found. For the weight-relief lift, significantly higher peak reaction forces (±25%) were found for the subjects with tetraplegia (1600N) compared to subjects with paraplegia (1250N) and able-bodied subjects (1100N), but no higher net moments. Due to the additional muscular effort needed for stabilization, the actual internal loads are higher than could be derived from an analysis based on net moments only, therefore the net moments will underestimate the load on the shoulder for subjects with tetraplegia.

In the study where the mechanical load for wheelchair-related tasks was quantified (Chapter 4) the glenohumeral reaction forces were expected to be higher for subjects with a high lesion level, since these subjects have to compensate for the muscle paralysis due to their lesion. Indeed this was the case. The additional muscular effect of, among others, the rotator cuff was needed for stabilization of the shoulder and resulted in higher compression forces at the glenohumeral joint.

In the simulation study (Chapter 5) the main interest was to quantify the amount of extra muscle force subjects with a high lesion level exert as compensation for the muscle paralysis. A major challenge was to configure the model to represent an incomplete musculoskeletal system: a subject with a high lesion level, since the original model represented a complete musculoskeletal system. The model was adapted based on segment innervations [54] and on the assumption that the maximum muscle force was related to the number of segments above the lesion. This is only an assumption since there is not only a great deal of variation in the segment innervations but also in the lesions themselves. For further research, more information on the maximal muscle force that subjects with different lesions can apply is necessary. Since the classification chosen here was based on an arbitrary assumption (maximum muscle force related to number of segments above the lesion), the effect of alternative force distribution was explored. One third of the data was simulated with a model in which the maximum muscle force was divided according to a sigmoid curve over the innervating segments. These simulations did not lead to significantly different results; the glenohumeral reaction force was only 4% higher compared to the simulations with the first force distribution. As might be expected in this alternative setup, more simulations at higher lesion levels were successful because more force was available at higher lesion levels.

As all inverse-dynamic models, our model used a cost function to solve the load sharing problem. Cost functions are used to distribute the net joint moments to the force generating structure (muscles and ligaments) in the most economical way. Different cost functions have been proposed [158], but most cost functions are mechanical cost functions, minimizing the muscle stress. Also the cost function used in our model minimized the summed squared muscle stresses.

It has often been assumed that energy cost is minimized by the central nervous system to control a task and not mechanically related measures [1, 59]. Although a relationship between the mechanical parameters and the physiological costs is assumed to exist, the relationships are not unambiguous. Recently Praagman et al. [128] looked at the relationship of two cost functions with muscle energy consumption and found that the energy based cost function led to slightly better modeling results. Better results for predicting muscle force could have been achieved with the new energy related cost function, but this cost function has not been implemented in the model yet.

Damaging forces?

The glenohumeral reaction forces of wheelchair-related ADL are much higher compared to the forces of level wheelchair propulsion, though these forces are not as high as the contact forces measured in the hip joint [9]. For normal walking, forces above 2000N (238% bodyweight) were measured in the hip with an instrumented implant [9]. When comparing these forces, one must bear in mind that the contact forces in the hip joint are highly dependent on the weight of the upper body. Therefore the hip contact force is automatically higher than the glenohumeral contact force. Further, the hip joint is much better shaped than the glenohumeral joint where the head is large compared to the shallow saucer.

The joint reaction force reflects both the compression force to the joint surface and the muscle forces stabilizing the joint. The joint compression force could be damaging to the joint surface [56], but these types of injuries are rare. The high muscle forces are more likely to lead to soft tissue damage [5, 184].

Practical implications

This thesis shows that steady state level wheelchair propulsion is not that stressful in terms of peak load when compared to wheelchair-related ADL. Further it was clear that the glenohumeral reaction force was almost twice as high for reaching as for level wheelchair propulsion. Although the load is often not that large, reaching or working above the head should likely be limited. Important for all these tasks is the posture of the person in the wheelchair; the wheelchair-user interface. For subjects in a wheelchair an adjusted wheelchair is the most important factor for an efficient propulsion [13, 86, 142, 169], while for other tasks a correct posture (due to trunk stabilization, contoured backrest and cushioning) improves stability and therefore reduces the strain on the musculoskeletal system and increases the range of motion and functional outcome [32, 82, 152]. Developments in assistive technology are important in order to allow the use of the newest and lightest materials in the designs of the wheelchair as well as to optimize the wheelchair-user interface.

Wheelchair-related ADL

What has always been suspected has now been confirmed; the load of wheelchair-related tasks is high which could be an important risk factor for damage to the joint and the soft tissues around the shoulder. A simple solution to avoid this high loading is not to perform these tasks anymore, which is for other reasons not realistic or advisable. Alternatively, tasks can be performed differently as for example performing the weight-relief lift in the way able-bodied subjects perform this lift, as was illustrated in the Chapters 4 and 5. It looks that next to muscle paralysis, stability requirements in the thorax and the shoulder prevent subjects with a SCI from doing so. The last few years, alternative methods for the weight-relief lift have been introduced [26]; forward or lateral flexion of the trunk releases the bottom from the seat to allow better perfusion to the buttocks. These new lift methods are even more effective than the traditional lift, where high loads on the upper extremity are needed to lift from the seat.

Transfers are likely even more straining when compared to lifting. During a transfer not only the body weight has to be shifted over quite a large distance, but there is also a rotation of the trunk. The arm is further away from the body, creating a larger moment arm in the shoulder and this position likely requires a considerable effort for stabilization. Especially for subjects experiencing pain and in subjects with a high-level injury, the use of a transfer-assist device has to be encouraged. Transfer-assist devices reduce the amount of force necessary for the lateral movement [55]. In any way, different techniques or changes to the wheelchair-user-interface are necessary for persons (with a high lesion) to remain functioning independently.

Wheelchair propulsion

Not only the peak contact forces for wheelchair propulsion are relatively low, also the mechanical effectiveness of the force application is low [173, 181]. The involved muscle mass in wheelchair propulsion is probably smaller than the involved muscle mass in hand cycling. Therefore, for longer distances it would be better to use the hand cycle, of which the efficiency of the movement is higher and the coupling of the hand on the handrim is no longer a problem. During hand cycling, both the flexors and the extensors employ force and are being trained, which is also better to prevent muscle imbalance. Muscle imbalance is a risk factor for impingement which can arise from high intensity wheelchair propulsion [19]. For weight-relief lifting, especially the large thoracohumeral muscles are important to cancel the decrease in the subacromial space as a result of the downward force. During rehabilitation and thereafter, care must be taken to insure a balanced training of the arm-shoulder musculature to prevent imbalance and to strengthen the thoracohumeral muscles [33, 64].

Exercise or rest during rehabilitation?

An important implication for the rehabilitation is the fact that 40% of the patients developed shoulder pain during the first three months of rehabilitation. Although it was found that upper extremity pain decreased 30% over time during the latter part of inpatient rehabilitation, pain at the beginning of rehabilitation was a strong predictor for pain one year after inpatient rehabilitation (Chapter 2). Pain was related to the muscle force and the functional outcome of the subjects. Especially subjects with tetraplegia have a reduced muscle force and a limited functionality and were at a higher risk for the development of upper extremity pain complaints later on.

At the beginning of the rehabilitation after a spinal cord injury, most subjects do not have well-trained upper extremity musculature which makes them over-sensitive to overuse injuries. The advice to these patients might be to limit their physical activities. However, bed rest will facilitate decubitis, atrophy of the muscles and will reduce the work capacity [115], which will increase the risk on upper extremity pain. It might be that early intervention in the form of training of physical capacity in these patients will be effective for the reduction of the risk to develop shoulder pain. Besides, exercise is known to aid in the prevention of decubitis [115], obesity [27, 49], cardiovascular diseases [92, 133] and diabetes

[117]. Therefore, for subjects with a SCI, exercise can reduce the risk on coronary heart diseases [34] and it is even suggested that exercise may increase the neuronal health and recovery in the SCI population [120]. It is important to increase the physical capacity of patients during the initial rehabilitation, before starting to perform straining wheelchair-related tasks. However, especially for subjects with upper extremity muscle paralysis, there is a fine line between exercise and overuse. Consequently, accurate information on the risk and impact of upper extremity pain and the need to maintain fitness during the rehabilitation process will be of utmost importance.

Future research issues

To unravel the underlying mechanisms of the development of overuse injuries, it will be necessary to look more into the causal aspects of the overuse-damage relation and the influence of recovery on this relation. Also, a longitudinal study to discover the processes which occur due to continuous submaximal loading should be carried out.

Future research could focus on the supraspinatus muscle, since the supraspinatus muscle is one of the shoulder structures most often damaged. Reported disorders of this muscle vary from massive tears to virtually complete atrophy. The degeneration of the tendon, tendinitis, is one of the principal causes of shoulder pain. Impingement against the acromion and the surrounding structures has been addressed as a principal cause of this tendinitis and eventually rotator cuff tears [108]. Further, repetitive activities related to sport, occupation or lifestyle have also been associated with shoulder pain and tendinitis [7, 11, 43].

The effect of the repetitive load of wheelchair propulsion on the tendon could be studied in-vitro. Like in finite element model studies, an in-vitro fatigue analysis based on model loading profiles (from existing wheelchair propulsion simulations) can be carried out to study the effect of the submaximal intermittent load on the stiffness of the tendon [143, 144]. The loading profile for wheelchair propulsion could also be applied to skinned muscle fibers to study the mechanical properties or morphological and histological adaptations to this load [73]. The tendon should be investigated on histopathological changes as well. Increased proportion of type III collagen may be the result of minor injury and points to tendon wound healing [136]. Preferably these studies should be carried out on normal cadaver tendons and on cadavers with serious tendinitis.

The tendon stiffness may even be studied in-vivo as well in both healthy and injured subjects [106] by ultrasound or magnetic resonance imaging. Ultrasound could also be used to measure the subacromial space. During tasks like transfers and lifts, but maybe as well during wheelchair propulsion, there might be a proximal migration of the humerus, reducing the supraspinatus compartment which results in impingement. Impingement of the supraspinatus tendon against the acromion is also found to be a cause of supraspinatus tear [108]. By means of ultrasound or fluoroscopy the size of the compartment could be defined, showing the effect of propulsion style or pain on the compression of the subacromial space.

Swelling due to hyper-oxygenation can decrease the volume of the supraspinatus compartment which can again result in impingement. On the other hand micro-damage can occur as a result of a reduced blood perfusion [72, 74, 96] to the critical zone of the supraspinatus tendon. Tendinitis can therefore be secondary to physiological effects of intensive use.

Another way to detect damage by overuse might be the detection of cytokines and growth factors in the supraspinatus. Repetitive compression and loading can lead to mechanical injury of cellular membranes and intracellular structures [154]. As a result of this tissue injury, cytokines are released [95]. The release of these inflammatory cytokines stimulates a systemic immune reaction. Cytokines and growth factors are indicators for damage which could occur as a result of overuse of the shoulder. For several musculoskeletal conditions the early expression of cytokines has been detected [18, 21, 132, 140]. Detection of cytokines after a submaximal intermittent load like wheelchair propulsion in the supraspinatus is likely an indicator for damage.

Studying these mechanical and physiological parameters to detect damage by submaximal loading in an early stage seems to be the key for the future. Differences on the parameters between wheelchair users and able-bodied subjects and between subjects with and without pain could illustrate the relation between damage and overuse.

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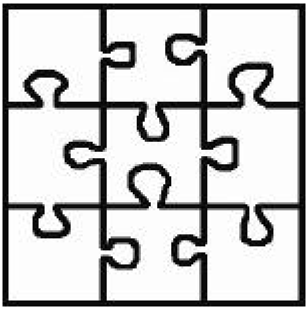
Summary



Upper extremity load during wheelchair-related tasks in subjects with a spinal cord injury

Physical activity is seen as a powerful tool to increase the general health of people with a spinal cord injury (SCI) [66]. The downside of increased physical activity in subjects with a SCI is that it is more or less limited to upper-body work while the upper extremities are particularly sensitive to overload injury. Prevalence rates of 50 to 70% for upper extremity complaints have indicated that overload injuries of the musculoskeletal system are indeed a serious long-term problem [7, 19] in subjects with a SCI. Especially, subjects with a high-level injury appear to be at risk due to a reduced muscle mass in the upper extremities [31, 148].

The aim of this thesis is to increase the understanding of the underlying mechanisms related to the development of overload injuries of the upper body musculoskeletal system in subjects with a SCI.

In chapter 2, the prevalence of upper extremity musculoskeletal complaints and especially shoulder complaints was investigated. One hundred and sixty nine subjects with a SCI were measured and interviewed at four test occasions during and after their rehabilitation. To explain the number of pain complaints, these were related to lesion - and personal characteristics as well as to muscle force (MMT) and functional outcome (FIM motor score). This study showed that pain complaints already developed during the first months of rehabilitation, and that these were decreased by 30% at the time subjects were discharged. After the rehabilitation period no further decrease was found, but rather a slight increase. Subjects with tetraplegia had a higher risk factor of 2.8 for upper extremity complaints when compared to subjects with paraplegia. On the other hand, subjects with a 10 point higher FIM motor score had an 11% lower risk on upper extremity complaints and 12% on shoulder complaints. The same effect was found for the muscle score: subjects with a 10 point higher muscle score had a 14% lower risk on shoulder complaints.

Another interest of this study was to investigate whether personal characteristics led to a higher risk on developing upper extremity pain one year after the rehabilitation. It was found that pain at the beginning of the rehabilitation and a higher body mass index were strong predictors for pain one year after the inpatient rehabilitation.

Wheelchair propulsion and wheelchair-related tasks are both mentioned as risk factors for the development of upper extremity complaints in subjects with a SCI. Both activities have to be frequently performed for mobility and independence from the beginning of the rehabilitation onwards. For wheelchair propulsion the mechanical load has already been studied, but information about the load on the shoulder and the muscles during wheelchair-related activities of daily living was limited. In chapter 3 a description was given of the mechanical load of wheelchair propulsion, reaching, propelling up a slope, negotiating a curb and performing a weight-relief lift, quantified as net moments around the shoulder and elbow. The main interests were the magnitude of the moments and the differences between subjects with paraplegia, subjects with tetraplegia and able-bodied subjects. It was found that the net shoulder moments for the weight-relief lift and negotiating a curb were significantly higher when compared to the other tasks. Further, reaching and riding on a slope caused higher moments compared to level wheelchair propulsion. For the shoulder moments no significant differences were found between the three subject groups. From this study it could be concluded that the wheelchair-related tasks cause a high mechanical strain on the shoulder and elbow, but somewhat surprisingly, there were no differences between able-bodied subjects and subjects with paraplegia or tetraplegia.

It was expected to find differences between the subject groups because subjects with tetraplegia have muscle paralysis of various muscles of the upper extremity and of the thorax. They have to compensate for this paralysis with alternative muscles which may have unfavorable torque components that have to be compensated again. Since additional muscle force cannot be detected from net moments, the tasks were also studied in a more detailed approach with a parameter for mechanical load that does incorporate these muscle forces as well as the strain on the glenohumeral joint. This joint reaction force not only reflects forces needed to overcome the external force but also the summed muscle forces around the joint. The joint reaction force could be calculated with the Delft Shoulder and Elbow model, which is an inverse biomechanical model of the upper extremity. This model also calculated the forces in the individual muscles.

In chapter 4 it was investigated whether the joint reaction forces and the muscle forces were higher for subjects with tetraplegia compared to subjects with paraplegia and able-bodied subjects for three different tasks. In this study it was put forward that there were differences among these subject groups. The different task performances were probably brought about by the muscle paralysis and the decreased stability in subjects with a high-level spinal cord injury. Especially the weight-relief lift seemed to be a task where stabilization of the shoulder and thorax is compulsory. The different task performances led to higher muscle forces and a higher joint reaction force. During wheelchair propulsion and reaching the peak muscle forces did not exceed 20% of the relative maximal force. For the weight-relief lift the latissimus dorsi, the biceps brachii and the monoarticular part of the triceps showed relative muscle forces up to 40% of their maximum force. Muscles which apply high forces are at risk for overuse injuries.

In chapter 4, muscle paralysis was not included in the model, but the study already showed a difference in the magnitude of the joint reaction force due to differences in task performance. Would account be taken of muscle paralysis, the differences were expected to be even higher, since unfavorable muscles have to compensate for muscle paralysis. In chapter 5, the model, which was used in chapter 3 and 4, was modified to represent subjects with a high-level SCI. The maximum relative force of the muscles was adjusted for the different lesion levels (C5 –T1), based on the assumption that the maximum force was relative to the number of innervating segments of the muscle above the lesion. The segment innervations for the muscles were based on Gray [54]. Results showed that the difference in task performance led to a higher joint reaction force for the subjects with tetraplegia. Surprisingly the effect of the different lesion levels was small: values were only 7% higher for the C6 lesion compared to the T1 lesion, which in fact was the intact model. It appears that the subjects with tetraplegia already performed the weight-relief lift in an economic, but different manner, given their lesion level. Besides lesion level, also paralysis of the triceps muscle was studied in this chapter. As expected, it was found that the triceps is an important muscle for the performance of the lift for the able-bodied subjects. They needed around 30% of the maximum force to successfully perform the lift, while lifting following the technique used by the subjects with tetraplegia only 10% of the triceps force was needed.

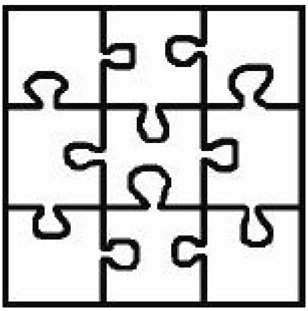
These studies showed that weight-relief lifting is a highly straining task for the shoulder. In chapter 6 it was investigated whether this task led to a reduction of the subacromial space. During this task there is a large downward gravity force on the trunk which has to be compensated by activity of the thoracohumeral muscles. When these muscles can not generate sufficient force, the subacromial space might decrease, which could lead to impingement of the supraspinatus tendon and surrounding tissue against the acromion. In previous studies different force ratios of the shoulder muscles were found for subjects with impingement when compared to subjects without impingement. Therefore, in this study not only the muscle activity was measured, but isometric force of the muscle groups as well. Unfortunately neither a reduction in the subacromial space, nor differences in the force ratios could be detected between the able-bodied subjects and the subjects with paraplegia. Explanations for these results might be that subjects with impingement or shoulder pain were not included in this study or that there is indeed a high inter-individual variability in the subacromial space. During this task the thoracohumeral muscles as well as the triceps caput longum showed levels of muscle activity between 25 and 50% of their voluntary maximum. The activity of these muscles was higher for subjects with paraplegia compared to the able-bodied subjects.

In chapter 7, the epilogue, the main findings of the studies described in this thesis were summarized and discussed. The results of the studies contributed to a better understanding of the risk on overuse injuries of the upper extremity in subjects with a SCI. The peak load during wheelchair-related tasks like weight-relief lifting was found to be high and possibly damaging for the soft tissues. Since lesion level did not influence the load a great deal in the subjects with tetraplegia it is suspected that they perform the weight-relief lift and possibly also other tasks in the most economic way, considering their lesion.

Of course peak load is only one risk factor for overuse injuries. Further research must show what the influence is of continuous submaximal exercise on the occurrence of damage. Detection of mechanical, physiological and biological markers in an early phase of overuse can contribute to the prevention and treatment of these injuries.

Strengthening the muscles, in a highly balanced manner, and improving the overall physical capacity for subjects with a SCI is assumed to be necessary to reduce the relative load. Although exercise is proven to be good for general health and can reduce the risk on secondary impairments, especially for subjects with tetraplegia there appears to be a fine line between improving health and overload as a result of exercise. Further, the need for other, less straining techniques of the lift and transfer is stressed. In addition, improved assistive technology and built environment from early rehabilitation onward is important to reduce the external load.

Samenvatting



Belasting op de bovenste extremiteit tijdens rolstoel-gerelateerde taken bij mensen met een dwarslaesie

Lichaamsbeweging wordt over het algemeen beschouwd als een goede manier om de gezondheid van mensen met een dwarslaesie te verbeteren [66]. Meer bewegen bij mensen met een dwarslaesie levert echter een probleem op: het betreft extra activiteit van de bovenste extremiteiten en juist de bovenste extremiteiten zijn erg gevoelig voor overbelastingsblessures. De hoge prevalentie (50 tot 70%) van overbelastingsblessures aan het bewegingsapparaat van de bovenste extremiteiten maakt duidelijk dat dit een serieus lange termijn probleem is voor mensen met een dwarslaesie [7, 19]. Vanwege een verminderde spiermassa van de bovenste extremiteiten lijken vooral mensen met een cervicale dwarslaesie risico te lopen op overbelastingsblessures [31, 148].

Het doel van dit proefschrift is om meer inzicht te krijgen in de ontstaansmechanismen van aandoeningen aan het bewegingsapparaat van de bovenste extremiteiten bij mensen met een dwarslaesie.

In hoofdstuk 2, is gekeken naar de prevalentie van klachten aan het bewegingsapparaat van de bovenste extremiteiten en in het bijzonder van schouderklachten. Honderdnegenenzestig personen met een dwarslaesie werden op vier tijdstippen tijdens en na hun revalidatie gemeten en geïnterviewd. Om de pijnklachten te verklaren werden deze niet alleen gerelateerd aan persoonlijke- en laesiekenmerken maar ook aan spierkracht (MMT score) en functie (FIM motor score). Deze studie heeft aangetoond dat pijnklachten al voorkwamen tijdens de eerste maanden van de revalidatie. Op het moment dat de personen ontslagen werden was het aantal klachten met 30% gedaald, maar na de revalidatie nam het aantal klachten weer licht toe. In vergelijking tot mensen met een paraplegie hadden mensen met een tetraplegie een factor 2.8 hogere kans op klachten aan de bovenste extremiteiten. Personen die 10 punten hoger scoorden op de FIM motor score hadden 11% minder klachten aan de bovenste extremiteiten en 12% minder schouderklachten. Dezelfde trend werd gevonden voor mensen die een hogere MMT score hadden: een 10 punten hogere MMT score voorspelde 14% minder schouderklachten.

In deze studie werd ook gekeken of bepaalde persoonlijke kenmerken of scores leidden tot een verhoogde kans op pijnklachten één jaar na de revalidatie. Er is gebleken dat pijnklachten aan het begin van de revalidatie en een hogere body mass index voorspellende waarden zijn voor pijnklachten één jaar na de revalidatie.

Rolstoelrijden en rolstoelgerelateerde taken worden vaak genoemd als risicofactoren voor het ontstaan van klachten aan de bovenste extremiteit bij mensen met een dwarslaesie. Beide activiteiten worden vanaf het begin van de revalidatie regelmatig uitgevoerd zowel voor het voortbewegen als om een zelfstandig bestaan te kunnen leiden. De mechanische belasting van het rolstoelrijden is al bestudeerd, maar er is weinig informatie over de belasting van de alledaagse rolstoel taken op het schoudergewricht en de schouderspieren. In hoofdstuk 3 werd de mechanische belasting van rolstoelrijden, helling rijden, een reiktaak, een stoepje nemen en van lften beschreven en uitgedrukt in netto momenten. Het doel van dit hoofdstuk was om te kijken naar de hoogte van de netto momenten van de verschillende taken en om te kijken of er verschillen waren tussen mensen met een paraplegie, mensen met een tetraplegie en gezonde proefpersonen. De netto schouder momenten van het liften en van het nemen van het stoepje waren significant hoger dan de momenten van de andere taken. Ook waren de momenten van het reiken en het helling rijden significant hoger dan normaal rolstoelrijden. Er werden geen verschillen gevonden tussen de drie groepen proefpersonen. Er kan worden geconcludeerd dat de mechanische belasting die wordt uitgeoefend op de schouder en de elleboog bij rolstoeltaken erg hoog is en dat er verassend genoeg geen verschillen werden gevonden tussen mensen met een paraplegie, een tetraplegie en gezonde proefpersonen.

Verschillen tussen de groepen proefpersonen werden wel verwacht omdat personen met een tetraplegie uitval hebben van romp- en arm spieren. Andere spieren, die vaak een ongunstig, te compenseren neveneffect hebben, zullen deze spieruitval moeten opvangen. De extra spierkracht die nodig is voor stabiliteit is niet terug te zien in de netto momenten. Daarom zijn de rolstoeltaken ook op een meer gedetailleerde manier bestudeerd met een parameter voor mechanische belasting (gewrichtsreactiekracht) die zowel de extra spierkracht als de belasting op het schoudergewricht bevat. Deze gewrichtsreactiekracht geeft niet alleen de krachten weer die nodig zijn om het externe moment te compenseren maar ook de som van de spierkrachten rond het gewricht. De gewrichtsreactiekracht kan berekend worden met een driedimensionaal model van de bovenste extremiteit: het Delft Schouder en Elleboog Model. Dit model berekent naast de gewrichtsreactiekrachten ook de krachten van de individuele spieren van de bovenste extremiteit.

In hoofdstuk 4 werd onderzocht of de gewrichtsreactiekrachten en de spierkracht bij drie rolstoeltaken hoger waren voor mensen met een tetraplegie in vergelijking tot mensen met een paraplegie of gezonde personen. Uit deze studie kwam, in tegenstelling tot het voorgaande hoofdstuk, wel naar voren dat er verschillen waren tussen de groepen proefpersonen. Spieruitval en verminderde stabiliteit bij de mensen met een tetraplegie zorgden waarschijnlijk voor een verschil in uitvoer van de taak. Vooral voor het liften bleek stabiliteit van de schouder en romp een vereiste. De verschillen in uitvoering van de taken leidden tot hogere gewrichtsreactiekrachten en hogere spierkracht. De spierkracht bij het reiken en het rijden kwamen niet boven de 20% van de relatieve maximale kracht. Bij het liften leverden de latissimus dorsi, de biceps en het mono-articulaire deel van de triceps meer dan 40% van de relatieve maximale kracht. Spieren die een hogere kracht moeten leveren hebben meer kans op overbelastingsschade.

In hoofdstuk 4 werd in het gebruikte model geen rekening gehouden met spieruitval door verlamming, maar toch werden er verschillen geconstateerd in de gewrichtsreactiekracht door een verschil in uitvoering. Verwacht werd dat de verschillen tussen de groepen groter zouden worden als er in het model rekening werd gehouden met spieruitval, omdat minder gunstige spieren in dat geval de uitgevallen spieren moeten compenseren. In hoofdstuk 5 werd het model, dat in de hoofdstukken 3 en 4 was gebruikt, dan ook aangepast om personen met een hoge laesie te vertegenwoordigen. Gebaseerd op de aanname dat de maximale kracht evenredig was aan het aantal innerverende segmenten boven de laesie, werd de maximale relatieve spierkracht aangepast voor de laesieniveaus C5 tot T1. De innerverende segmenten per spier werden overgenomen uit Gray [54]. De resultaten lieten zien dat het verschil in taakuitvoer hogere gewrichtsreactiekrachten tot gevolg had voor de personen met een tetraplegie. Verrassend genoeg was het effect van de verschillende laesieniveaus erg klein: de gewrichtsreactiekracht was slechts 7% hoger voor de C6 laesie in vergelijking tot de T1 laesie. De T1 laesie stond gelijk aan het intacte model. Het lijkt erop dat de personen met een tetraplegie de lift op een andere, maar gezien hun laesie op de meest economische manier uitvoeren. Naast het effect van laesieniveau werd in dit hoofdstuk ook specifiek het effect van uitval van de triceps bestudeerd. Zoals verwacht was de triceps voor de gezonde proefpersonen een belangrijke spier voor de uitvoer van de lift. De gezonde proefpersonen hadden ongeveer 30% van de maximale triceps kracht nodig om de lift succesvol uit te voeren, terwijl de personen met tetraplegie met hun manier van uitvoeren slechts 10% van de maximale kracht gebruiken.

De vorige studies lieten zien dat het liften een zwaar belastende taak is voor de schouder. In hoofdstuk 6 werd onderzocht of er tijdens het uitvoeren van de lift een verkleining van de subacromiale ruimte plaatsvond. De zwaartekracht die tijdens deze taak aan de romp trekt moet gecompenseerd worden door de thoracohumerale spieren. De subacromiale ruimte zal verkleinen als deze spieren niet genoeg kracht kunnen leveren, en dit kan leiden tot beklemming van de pees van de supraspinatus, en de omliggende structuren, tegen het acromion. In het verleden werden verschillen gevonden in de krachtratio's van de schouderspieren tussen mensen met en zonder impingement, daarom werd in dit hoofdstuk naast spieractiviteit ook isometrische kracht van de schouderspieren gemeten. Helaas werd er geen verandering van de subacromiale ruimte gemeten en ook geen verschil gevonden in de krachtratio's tussen de gezonde personen en de personen met een paraplegie. Een verklaring voor de gevonden resultaten zou kunnen zijn dat er geen mensen met impingement deelnamen aan dit onderzoek. Een andere verklaring is dat er een grote interindividuele variabiliteit bestaat in de grootte van de subacromiale ruimte. De thoracohumerale spieren en de lange kop van de triceps vertoonden tijdens deze taak spieractiviteit tussen 25 en 50% van de vrijwillig maximale contractie. Verder was de activiteit van de bovengenoemde spieren hoger voor de personen met paraplegie dan voor de gezonde personen.

In hoofdstuk 7, de epiloog, werden de belangrijkste resultaten van de in dit proefschrift beschreven hoofdstukken samengevat en bediscussieerd. De resultaten van het gedane onderzoek dragen bij aan het inzicht in de ontstaansmechanismen van aandoeningen aan het bewegingsapparaat van de bovenste extremiteiten bij mensen met een dwarslaesie. De piekbelastingen van de rolstoeltaken zoals het liften waren hoog en waarschijnlijk zo hoog dat ze schade aan de zachte structuren kunnen veroorzaken. Omdat laesieniveau voor personen met een tetraplegie weinig invloed had op de belasting, wordt er aangenomen dat de personen met een tetraplegie de lift, en mogelijk ook andere taken, in de voor hen meest economische manier uitvoeren.

Natuurlijk is de piekbelasting die optreedt slechts één van de risicofactoren voor het ontstaan van overbelastingsklachten. Verder onderzoek zal moeten aantonen wat de invloed is van continue submaximale belasting op het ontstaan van schade. Het vinden van mechanische, fysiologische en biologische markers in een vroeg stadium van overbelasting kan verder bijdragen aan het voorkomen en behandelen van overbelastingsletsels.

Het verbeteren van de fysieke capaciteit van personen met een dwarslaesie door het op een gebalanceerde manier sterker maken van de arm- en schouderspieren is nodig om de relatieve belasting te verlagen. Ondanks het feit dat lichaamsbeweging goed is voor het verbeteren van de gezondheid en het verlagen van de kans op indirecte aandoeningen is er vooral voor de mensen met een tetraplegie een dunne lijn tussen het verbeteren van de gezondheid en het overbelasten als gevolg van extra activiteit. Verder moet benadrukt worden dat er naast andere, minder belastende technieken voor het uitvoeren van de lift en transfers, ook verbeterde hulpmiddelen en een beter aangepaste omgeving nodig zijn na de revalidatie om de belasting te verminderen.

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Curriculum Vitae

Curriculum Vitae

Stefan van Drongelen was born in Terneuzen, the Netherlands on the 10th of December 1975 as the second child of Cees van Drongelen en Leuneke Scheele. He finished high school at the Zeldenrust College in Terneuzen, in 1994.
In 2000 he obtained his MSc in Human Movement Sciences at the Vrije Univerisiteit, Amsterdam. Next to his major in Functional Anatomy he achieved the teacher training at the tertiary level. His research project 'Predicting mechanical load of the glenohumeral joint in dynamic situations' turned out to be the forerunner for his doctorate. After graduation Stefan taught kinesiology at the Teacher Training for Physical Education at The Hague University for a year. From April 2001 until August 2005 he worked as a PhD student at the Faculty of Human Movement Sciences, Vrije Univerisiteit Amsterdam. Under the supervision of Dr. DirkJan Veeger and Dr. Luc van der Woude he investigated the load on the upper extremity during various wheelchair tasks. In this project he worked together with Ed Chadwick (Case Western Reserve University) and Edmond Angenot of the Rehabilitation Center Amsterdam.
Currently, Stefan is heading for new opportunities in the United States of America where he will do postdoctoral research at the Human Engineering Research Laboratories of the University of Pittsburgh.

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