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Shoulder load during handcycling at different incline and speed conditions

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ABSTRACT

Background: The manual wheelchair user population experiences a high prevalence of upper-limb injuries, which are related to a high load on the shoulder joint during activities of daily living, such as handrim wheelchair propulsion. An alternative mode of propulsion is handcycling, where lower external forces are suggested to be applied to reach the same power output as in handrim wheelchair propulsion. This study aimed to quantify glenohumeral contact forces and muscle forces during handcycling and compare them to previous results of handrim wheelchair propulsion.
Methods: Ten able-bodied men propelled the handbike on a treadmill at two inclines (1% and 4% with a velocity of 1.66 m/s) and two speed conditions (1.39 and 1.94 m/s with fixed power output). Three-dimensional kinematics and kinetics were obtained and used as input for a musculoskeletal model of the arm and shoulder. Output variables were glenohumeral contact forces and forces of important shoulder muscles.
Findings: The highest mean and peak glenohumeral contact forces occurred at 4% incline (420 N, 890 N respectively). The scapular part of the deltoideus, the triceps and the trapezius produced the highest force.
Interpretation: Due to the circular movement and the continuous force application during handcycling, the glenohumeral contact forces, as well as the muscle forces were clearly lower compared to the results in the existing literature on wheelchair propulsion. These findings prove the assumption that handcycling is mechanically less straining than handrim wheelchair propulsion, which may help preventing overuse to the shoulder complex.

1. Introduction

A high prevalence of upper limb injuries exists in the manual wheelchair user population. The point prevalence of shoulder pain in persons with spinal cord injury ranges from 30 to 73% (Ballinger et al., 2000; Pentland and Twomey, 1991), whereas in the general population a point prevalence of 7 to 27% is found (Luime et al., 2004). The most common causes of shoulder pain in individuals with chronic spinal cord injury are musculoskeletal, particularly overuse injuries to the rotator cuff (Dyson-Hudson et al., 2007). A relatively high load on the shoulder joint and a high frequency of this load, as it occurs during handrim wheelchair propulsion (van Drongelen et al., 2005a; van Drongelen et al., 2005b; Veeger et al., 2002), are suggested to be contributors to the development of shoulder injuries (Leclerc et al., 2004; Mercer et al., 2006). Transfers and vertical lifts, where the frequency is low but the shoulder load particularly high (van Drongelen et al., 2006), are other activities that may add to the risk to develop shoulder complaints.

In finding a good balance between lowering the load on the upper extremities and still living an active lifestyle, the handbike might be the preferred choice for outdoor transportation. A handbike can be used for everyday outdoor propulsion over longer distances or for training (Valent et al., 2010). This will strengthen the upper extremity muscles in a more balanced form and could lower the risk of overuse injuries, since the impingement syndrome appears to be related to weakness of the rotator cuff and scapular stabilizers (Miyahara et al., 1998). The handbike is energetically more efficient and less straining for the cardio-respiratory system (Dallmeijer et al., 2004; van der Woude et al., 2006). Whether it is also beneficial for lowering the mechanical load and for the prevention of overuse injuries is likely, but not yet known.

To date, details on the external force applied during handcycling or arm cranking are becoming available (Bafghi et al., 2008; Faupin et al., 2010; Kramer et al., 2009a; Kramer et al., 2009b; Smith et al., 2008). The externally applied hand forces give an indication about the possible shoulder moments and the shoulder load. The external forces of handcycling appear to be lower in comparison to handrim wheelchair propulsion, which suggests a lower shoulder load and thus a lower risk for shoulder injury (Arnet et al., in press; Veeger et al., 2002). The benefit of handcycling, with regard to shoulder injuries,

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has also been shown in studies analyzing muscle activity during handcycling or arm cranking. Comparing handbike to wheelchair propulsion, DeCoster et al. (1999) reported that during the latter the supraspinatus and the pectoralis major are much more active, suggesting that these structures are exposed to excessive strain in combination with abduction and internal rotation of the shoulder, which makes impingement likely to occur (Neer, 1972). For this reason, the handbike has been recommended for prevention of shoulder dysfunction (DeCoster et al., 1999).

Besides external applied hand force, net shoulder moments or muscle activity, the mechanical load on the shoulder can also be quantified as the glenohumeral contact force. The glenohumeral contact force can be calculated by the Delft Shoulder and Elbow Model (DSEM) (van der Helm, 1994). The input of the model comprises the external forces and the orientations of the upper extremity segments including the scapula. The output of the DSEM includes among others muscle lengths, muscle forces, moments around various axes and joint contact forces. The DSEM has been validated by comparing the calculated glenohumeral contact force and muscle forces with values measured using an instrumented shoulder prosthesis (Nikooyan et al., 2010). During dynamic motions, the glenohumeral contact forces predicted by the model showed compatibility with the contact forces measured with an instrumented shoulder prosthesis, although at arm elevations above 60° the measured and predicted values showed different trajectories. So far, glenohumeral contact force has been studied during various activities of daily living such as pushing and pulling (Hoozemans et al., 2004), reaching (van Drongelen et al., 2005a), lifting objects (Magermans et al., 2005), combing hair (Magermans et al., 2005), performing weight relief lifts (van Drongelen et al., 2005a) and handrim wheelchair propulsion (van Drongelen et al., 2005a; Veeger et al., 2002). To date no studies quantifying glenohumeral contact forces during handcycling are available. In order to enable recommendations about the use of a handbike in the context of reduced risk for shoulder overuse injuries, the glenohumeral contact forces should be studied and described.

The aim of this study was to quantify glenohumeral contact forces and muscle forces in able-bodied males, during synchronous handcycling in everyday conditions on a motor driven treadmill. It was hypothesized that due to the continuous force application and the direction of force application, the glenohumeral contact force and muscle forces will be lower compared to previous studies in handrim wheelchair propulsion.

2. Methods

2.1. Subjects

Ten able-bodied men participated in this study (mean (SD): age 30 (4) years, height 1.77 (0.06) m and body mass 75 (9) kg). Inclusion criteria were: no experience in handbike and wheelchair use, no current complaints of the musculoskeletal system of the upper extremities or shoulder surgery. All subjects gave their written informed consent. The study was approved by the local ethics committee.

2.2. Experimental design

After familiarization with the experimental procedure and setup, subjects propelled the handbike on a motor driven treadmill (Mill, Forcelink B.V., Culemborg, The Netherlands) at low, everyday speeds and power levels. To evaluate the shoulder load at different inclines, the subjects propelled with a constant velocity of 1.66 m/s and two inclines of respectively 1% and 4% for 1 min per slope. To evaluate the shoulder load at different speed conditions, all subjects propelled at the same constant external power output (PO) with two belt velocities of respectively 1.39 and 1.94 m/s for 1 min per speed. The power output was achieved by the rolling resistance of the handbike–user combination on the level treadmill and an additional external force acting via a pulley system on the handbike (Veeger et al., 1989). The individual drag force was determined in a separate drag test (van der Woude et al., 1986). The additional force was reduced proportionally with increasing velocity to achieve the same power output at both velocities.

2.3. Instrumented handbike

The handbike used in these experiments (Fig. 1) was an attach-unit system (Tracker Tour, Double Performance, Gouda, The Netherlands) with a synchronous crank setting, attached to a hand rim wheelchair (Pro Competition, Cyclone Mobility & Fitness, Bromborough, UK) (Van Drongelen et al., in press). The gear ratio of the handbike was fixed throughout the experiment at 0.741. Subjects were not instructed on how to propel the handbike.

The instrumented handbike measures the external forces on the left handle bar with a special purpose crank unit (Faculty of Human Movement Sciences, VU University, Amsterdam, The Netherlands). The handle bar contains a 6-axis force transducer (FS6-500, Advanced Mechanical Technology Inc., Watertown, Massachusetts, USA) which records three-dimensional forces at 100 Hz. Two incremental optical encoders (Type 19, Elcis, Collegno, Italy) measure the angles of the handle bar relative to the crank and the crank relative to the handbike (Van Drongelen et al., in press).

The measured forces were transformed from the local coordinate system of the force transducer to the local coordinate system of the crank (tangential, radial, lateral) and to forces in the global coordinate system. The total force (F_{tot}) acting on the crank was calculated as the norm of the three measured force components.

Produced external power output per propulsion cycle was calculated under the assumption that equal forces were applied with the left and the right hand. Power output was calculated from the product of the tangentially applied force component and the linear velocity of the crank: $PO = 2 \cdot F_{tan} \cdot v_{crank}$.

2.4. Kinematics

Kinematics of the upper limb were recorded synchronously to the kinetic data collection with a 6-camera movement analysis system (Oqus, Qualisys AB, Gothenburg, Sweden) operating at 100 Hz. Five unique clusters of reflective markers (4 markers each) were placed onto the upper extremities on the left-hand side (thorax, acromion,

Fig. 1. Subject sitting in the instrumented handbike with cluster markers attached to the thorax, acromion, upper arm, lower arm and hand.

upper arm, forearm and hand) (Fig. 1). Prior to the actual tests, reference measurements of bony landmarks were performed while the subject was sitting with the arms in the anatomical position. The reference measurements were used to relate the clusters to the local anatomical coordinate systems of the body segments, following the descriptions published in Wu et al. (2005) and in line with previous studies on joint contact forces in wheelchair propulsion (van Drongelen et al., 2005a; Veeger et al., 2002).

2.5. Inverse dynamic model

The Delft Shoulder and Elbow model was used to calculate the mechanical load (van der Helm, 1994; Veeger et al., 1997). Since the model represents a right shoulder and arm, all input data were mirrored to the right side. Kinematic input was the position of the incisura jugularis and the orientations of the thorax, scapula, humerus, forearm and hand. Further, the 3-dimensional external forces and moments applied by the hand on the crank served as kinetic input. The kinematic input data were optimized to the model structure as descibed by de Groot (1998). Output variables of the model used in this study were the glenohumeral contact force and muscle forces. A minimum stress cost function was used to calculate the muscle forces during handcycling (Praagman et al., 2006), and the total force produced by each muscle was obtained by summing up the forces of the muscle elements. To enable comparison of muscle forces, the 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/cm^2 of the physiologic cross-sectional area, obtained by Veeger et al. (1991a).

2.6. Data analysis

From the last 30 s of each exercise bout, five regular consecutive propulsion cycles were selected for data analysis. The propulsion cycle was defined as one rotation of the crank, starting at the position where the crank was parallel to the propulsion surface and the handle bar aimed towards the person sitting in the handbike. Since only a limited amount of data can be processed by the model, input data were resampled to 50 Hz and filtered with a second order Butterworth filter with a cutoff frequency of 10 Hz.

The mean and peak glenohumeral contact force and muscle forces per cycle were calculated and subsequently averaged over the five cycles. Of the 31 muscles, which were output of the model, only those muscles were analyzed which are relevant for the shoulder load (scapulothoracic muscles, scapulohumeral muscles and upper arm muscles).

2.7. Statistical analysis

The mean and peak values of glenohumeral contact forces, as well as the mean and peak values of the relative and absolute muscle forces were analyzed. To compare the results between the different incline conditions (1 vs. 4%) and speed conditions (1.39 vs. 1.94 m/s), non-parametric Friedman-tests were used since the data were not normally distributed. Level of significance was set at $Pi<0.05$.

3. Results

All subjects were able to perform all conditions. Mean achieved power and total external force applied to the crank are listed in Table 1. The achieved power output during the speed conditions did not differ between the two conditions ($P=0.682$).

Table 1
Mean measured power and external force applied to the crank.

Condition	1%	4%	1.39 m/s	1.94 m/s
Measured PO [W]	15.0	56.4	31.9	30.3
F_{tot} [N]	12.8	33.8	24.2	18.2

3.1. Glenohumeral contact forces

Mean glenohumeral contact forces over the propulsion cycle ranged between 270 and 420 N (Fig. 2). They were significantly higher during handcycling at 4% compared to 1% ($P<0.001$), but they were not significantly different for the two speed conditions (1.39 vs. 1.94 m/s, $P=0.241$). The peak glenohumeral contact forces ranged between 520 and 890 N (Fig. 2) and were significantly different for the two incline conditions (1 vs. 4%, $P=0.001$), but not for the two speed conditions at equal power output (1.39 vs. 1.94 m/s, $P=0.905$).

The distribution of the glenohumeral contact force throughout a propulsion cycle in this group of inexperienced able-bodied subjects showed that the highest contact force occurred in the phase where the crank is pulled towards the subject and lifted up, which is between 270 and 0°. The highest amount of total force, however, was applied to the crank while pulling the crank down and towards the subject (210°–300°). A typical example is shown in Fig. 3.

3.2. Muscle forces

The highest mean value of absolute muscular force over the cycle was seen in the scapular part of the deltoideus, which reached mean forces of 175 N and peak forces of 344 N (Fig. 4). The muscle with the highest peak values of absolute force was the triceps, which reached mean forces of 150 N and peak forces of 525 N. Further, the trapezius was also one of the most activated shoulder muscles with mean forces of 125 N and peak forces of 290 N.

Analyzing the relative muscle force, the deltoideus (scapular part) was the muscle showing the highest values. At the 4% incline, the scapular part of the deltoideus produced 10% of its maximal force over the propulsion cycle and reached peak values of 20% (Fig. 5). Other muscles, which reached peak values of more than 20% of their modeled maximal force capacity, were infraspinatus, supraspinatus and the clavicular part of the deltoideus. Muscles that were least active were teres minor and teres major with relative mean forces of less than 3%.

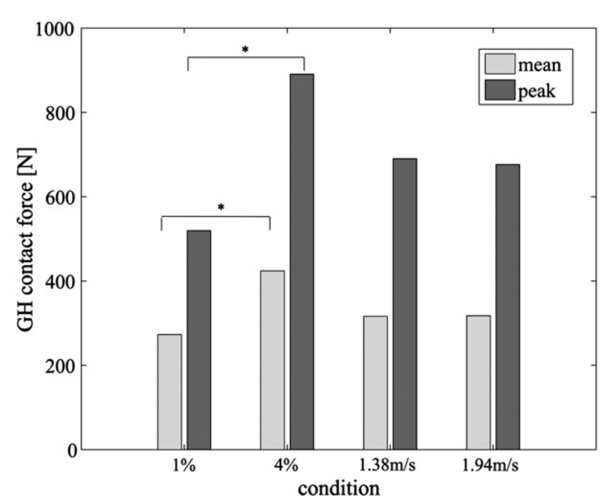

Fig. 2. Mean and peak glenohumeral contact forces (n = 10). * Significantly different for the incline conditions (1 vs. 4% with 1.66 m/s) but not for the speed conditions (1.39 m/s (32 W) vs. 1.94 m/s (30 W)), $P<0.05$.

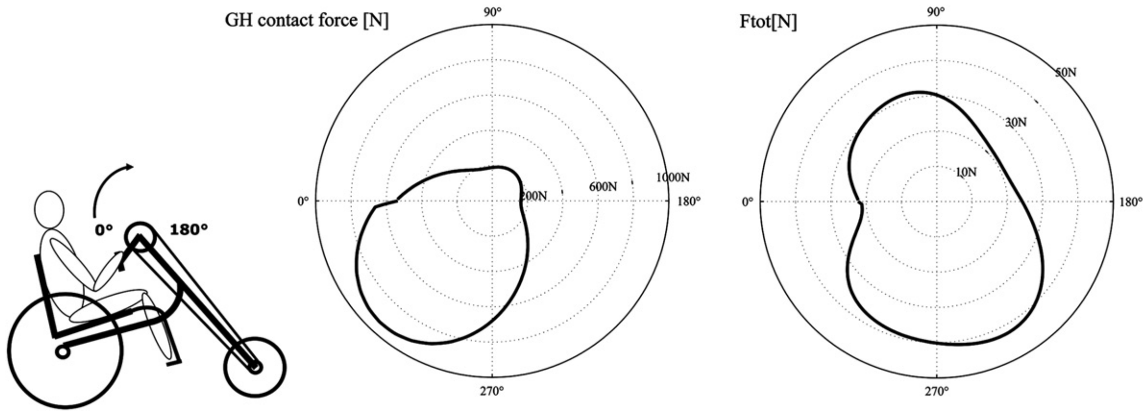

Fig. 3. Typical example (n = 1) of the glenohumeral contact force and the total force at 4% (56.4 W) at 1.66 m/s, visualized from the right hand side of the handbike. 0° is the position where the crank is aiming towards the subject sitting in the handbike.

Regarding the influence of the test condition, all mean and peak muscle forces increased significantly with higher incline ($P<0.05$). While in propelling at increased speed the triceps ($P<0.001$), deltoideus (clavicular part, $P=0.033$) and the pectoralis major ($P=0.032$) produced significantly lower force over the whole cycle while the mean forces of teres minor increased significantly ($P=0.017$). Concerning the peak forces only the value of the triceps ($P<0.001$) decreased significantly with higher velocity (Figs. 4 and 5).

4. Discussion

4.1. Glenohumeral contact forces

The glenohumeral contact force increased while propelling at the higher incline due to the higher power output and consequently the higher external force applied to the crank (Table 1). While propelling with a higher speed, and therefore a higher cadence at equal power output, the external force decreased. The glenohumeral contact force, however, did not change (Fig. 2) similar to the force produced by most of the shoulder muscles (Fig. 4). At the faster speed conditions the shoulder muscles produced therefore muscle force which did not contribute to the external force. In previous studies on handrim wheelchair propulsion it has been shown that with higher angular velocity of the rim, and therefore faster extension of the shoulder and elbow joint, the force application became less effective (Veeger et al., 1991b).

The results of the glenohumeral contact forces of this study, compared to mean values reported during activities of daily living, showed that glenohumeral contact forces measured during weight-relief lifts (van Drongelen et al., 2005a), pushing a cart (Hoozemans et al., 2004), lifting a 10 kg suitcase (Anglin et al., 1997) and walking with a cane (Anglin et al., 1997) were higher (800 N, 470 N, 1750 N and 1241 N, for each activity respectively) than values measured in this study (Fig. 2).

To evaluate if handcycling is a more optimal alternative propulsion mode for persons depending on a wheelchair the glenohumeral contact forces of both devices should be compared. To date only one study on wheelchair propulsion (Veeger et al., 2002) is comparable to the present study, since they measured glenohumeral contact forces at comparable conditions and reported on both the push and the recovery phase. During wheelchair propulsion at 10 W and 20 W mean contact forces were reported between 500 and 850 N during the push phase and between 300 and 400 N during the recovery phase (Table 2). When comparing the glenohumeral contact force of both devices during the whole propulsion cycle, wheelchair propulsion (10 W) resulted in 40% higher value than handcycling (15 W, Table 2).

This difference between devices cannot be attributed to the difference in external force since the mean external applied force during handcycling was higher than during wheelchair propulsion at

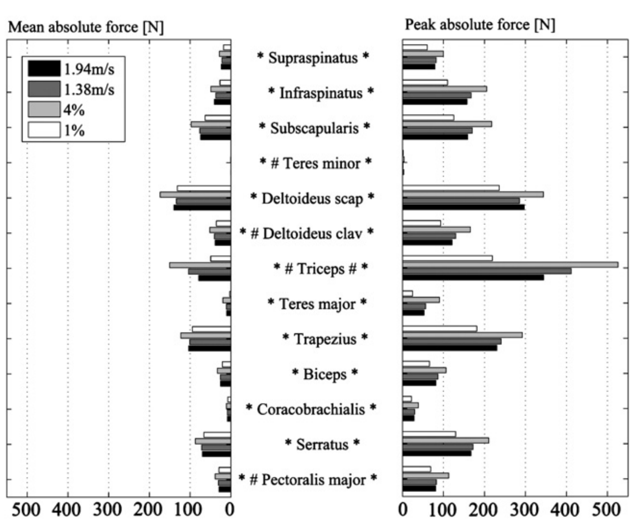

Fig. 4. Mean and peak absolute muscles forces (n = 10). * Significantly different for the incline conditions (1% (15.0 W) vs. 4% (56.4 W) with 1.66 m/s), $P<0.05$. # Significantly different for the speed conditions (1.39 m/s (32 W) vs. 1.94 m/s (30 W)), $P<0.05$.

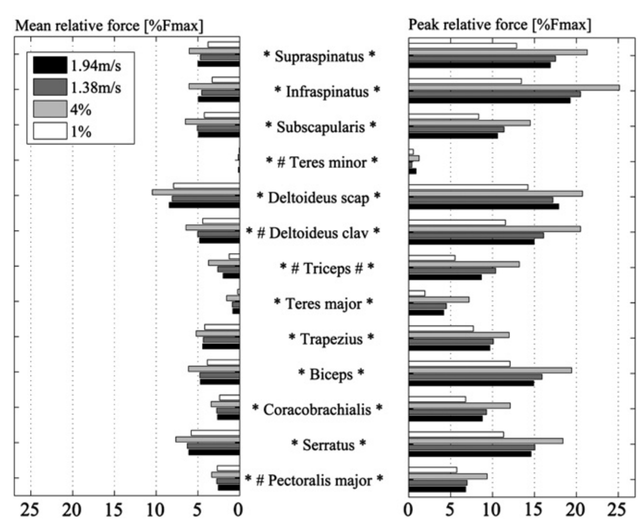

Fig. 5. Mean and peak relative muscles forces (n = 10). * Significantly different for the incline conditions (1% (15.0 W) vs. 4% (56.4 W) with 1.66 m/s), $P<0.05$. # Significantly different for the speed conditions (1.39 m/s (32 W) vs. 1.94 m/s (30 W)), $P<0.05$.

Table 2
Comparison of results the from handbike (HB) propulsion with results from previous studies conducted in wheelchair (WC) propulsion. The results of wheelchair propulsion are given for the push phase and for the propulsion cycle, which includes push and recovery phase.

Study	Device	PO [W]	F_{tot} [N]		GH contact force [N]		
			Push phase	Cycle	Push phase	Cycle	
			Mean	Mean	Mean	Mean	Peak
Veeger 2002	WC	10.0	19.6	8.0	640	439	1000
		20.0	30.0	14.4	850	616	1370
Present study	HB	15.0		12.8		273	519

10 W. Other than the amount of the external force, also its direction could have been a reason for the lower glenohumeral contact forces in handcycling. The force applied during the guided circular movement of handcycling in front of the body might have led to lower moments acting on the shoulder joint. In turn, this would have resulted in a lower glenohumeral contact force.

The continuous force application of handcycling was a further factor why the glenohumeral load in handcycling was lower. In wheelchair propulsion an idle period exists, where the hands have to be brought back to the starting point on the pushrim. Obviously, this movement also generated glenohumeral contact forces even though no force was applied to the pushrim. During this recovery phase, Veeger et al. calculated mean glenohumeral contact forces which were 70% of the mean values calculated during the push phase (Veeger et al., 2002). The intermittent force application of wheelchair propulsion further provoked the high peak values of glenohumeral contact force. In wheelchair propulsion, the external force had to be applied to the pushrim during only half the time of the propulsion cycle, whereas in handcycling the external force could be applied more evenly to the crank during the whole cycle. This resulted in a less prominent peak. Veeger et al. calculated peak values of 1000 to 1370 N during wheelchair propulsion (Veeger et al., 2002) while the forces found for handcycling were less than half of these values (Table 2).

Analyzing the distribution of the forces over the propulsion cycle showed that the highest external force was on average applied pulling the crank down (210–270°, Fig. 3). In this section F_{tot} was to some extent produced by the weight of the arm. The highest amount of glenohumeral contact force, however, was produced during pulling and lifting up the crank (270–0°, Fig. 3). This was the time period where external force had to be produced against gravity and therefore muscles were more active. Additionally, the distance from the shoulder to the handgrip was low which resulted in a smaller moment arm with which the propulsion force could be produced. Therefore, higher muscular force was needed to produce that propulsion force.

4.2. Muscle forces

The scapular part of the deltoideus, the triceps and the trapezius were the muscles that produced the highest force throughout the propulsion cycle (Fig. 4). Considering the produced force in relation to its maximal force gave information about muscular load. The scapular part of the deltoideus produced the highest relative force of 8 to 10% throughout the cycle and peak values up to 21% (Fig. 5). In general, muscles which were mainly active during the pull phase of the propulsion cycle (180°–360°, Fig. 3), such as the deltoideus (scapular part), biceps, trapezius and serratus anterior produced higher forces than muscles mainly active during the push phase. These findings could be explained by the fact that most of the power was produced during the pull phase. The results of electromyogram (EMG) studies conducted during handcycling or arm cranking support these findings. Faupin et al. (2010) reported high peak activities, measured as percentage of maximal voluntary contraction, of the trapezius (descending part, 70% MVC), the biceps (55% MVC) and the deltoideus (clavicular part, 45% MVC) during handcycling. Also during arm cranking Bressel and Heise (2004) reported the highest value of muscle activation in the biceps. The identification of the most active muscles of the model outcomes in this study corresponded with the findings of the EMG studies mentioned earlier. The magnitude of the muscle activation measured with EMG, however, cannot be compared with the values calculated with the inverse dynamic model since the EMG–force relationship is unknown.

Comparison of the muscle forces from the present study with results of wheelchair propulsion at conditions where similar external force was applied (Veeger et al., 2002), showed that mean and peak values measured during handcycling were lower. At a mean applied hand force of 33 N, peak forces were considerably lower in handcycling than in wheelchair propulsion (applied force of 30 N) because force was applied continuously to the crank and more muscles were used to produce the force (flexors and extensors). Concentrating on the shoulder joint, the highest mean values of muscle force produced over the push phase of wheelchair propulsion were measured in the supraspinatus (31%), infraspinatus (21%) and subscapularis (17%), which are all muscles of the rotator cuff. In the present study the highest mean values were found in the scapular part of the deltoideus (10%). These results showed that the rotator cuff, which is prone to overuse symptoms, produced a high relative force during wheelchair propulsion. In handcycling, however, these muscles were not particularly stressed; they all produced a mean relative force of 6%. Due to the fact that the rotator cuff was less stressed during handcycling, these findings could indicate that handcycling might indeed be a good alternative mode of propulsion in order to lower the injuries on the rotator cuff.

Over all it seems that the shoulder muscles were also more equally stressed during handcycling compared to handrim wheelchair propulsion. This was caused by the circular pattern of handbike propulsion where propulsion force can be produced by the extensors and flexors. Thus, more muscles were used at lower intensity level and the risk of overuse of single muscles was evidently reduced.

4.3. Study limitations

The handbike used in this study was a typical handbike used for activities of daily living. The position of the subject in the handbike was not individualized since the handbike was a rather fixed construction. The position of the crank with respect to the shoulder was not the same for all subjects and therefore the moment arm of the forces acting on the shoulder joint differed. This could individually have lead to higher glenohumeral contact forces. Future studies focusing on the setup of the handbike with respect to the subject might indicate to what extent the handbike setup has an effect on shoulder load.

Nikooyan et al. showed that during dynamic motions the glenohumeral contact forces are well predicted by the DSEM model for arm elevations below 60° (Nikooyan et al., 2010). The fixed handbike construction resulted in a slightly different range of motion between subjects. However, only in three subjects the arm elevation reached values above 60°. In these subjects the maximal elevation was 65° which does not limit the validity of the overall results.

The subjects in this study were able-bodied and non-experienced in handcycling. This group was chosen because it was expected that they would respond relatively homogeneous to handbike exercise since they were equally inexperienced and had no restriction due to disabilities. For transferring the results to the population of persons with a spinal cord injury, one has to keep in mind that due to the potential loss of muscle function the relative muscle force might as well be higher as reported in this study. Mainly for persons with a high lesion the muscle force needed for handcycling is distributed over fewer muscles, which could increase the actual stress on these remaining muscles and could result in higher forces on the shoulder joint.

5. Conclusions

This initial comparison showed that due to the circular movement of the hands and the continuous force application, mean and peak glenohumeral contact forces, as well as muscle forces are clearly lower during handcycling compared to wheelchair propulsion. These findings support the assumption that handcycling is mechanically less straining compared to handrim wheelchair propulsion which may reduce the risk of overuse to the shoulder. Therefore, based on this study, the handbike can be recommended as means of transportation for longer distances outdoor or as a training device to build up shoulder muscle strength and protect these muscles from overuse. To strengthen this statement, further studies on the direct comparison of wheelchair and handbike propulsion under different propulsion conditions should be performed.

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