MUSCULOSKELETAL MODELLING OF HANDCYCLING MOTION ON AN EROGOMETER: INFLUENCE OF CRANK POSITION ON TRAINING PURPOSES.

Felix A.J. Wolbert[1], Markus Peters[2], Josef Viellehner[2], Johannes Funken[2] and Wolfgang Potthast[2]

Institute of Health Science and Technology, Aalborg University, Aalborg, Denmark[1]
Institute of Biomechanics and Orthopedics, German Sport University Cologne, Cologne, Germany[2]

The objective of this study was to examine the influence of crank position during hand cycling on muscle forces and Glenohumeral joint reaction forces. Twelve male subjects performed 16 trials with variation in crank parameters. 3D motion capturing (Vicon) and custom made handlebars (Kistler force sensors) were used to determine kinematic and kinetic data. Musculoskeletal modelling (AnyBody) was used for calculations of joint contact and muscle forces. Significant difference was found between settings and muscle forces (p=0.006) and Glenohumeral joint reaction forces (p=0.000). For the m. Brachialis, the m. Latissimus Dorsi and the m. Deltoideus Scapularis, an optimal setting was found. The antero- posterior and distraction force are minimized with horizontal handlebars, a small crank width and long crank arms.

KEY WORDS: Musculoskeletal modelling, hand cycling, crank position, muscle forces, joint reaction forces.

INTRODUCTION: To train a certain muscle (group), for building muscle mass or rehabilitation, it is necessary to know how to do the corresponding exercise in such a way that it fits the purpose optimally. Janssen et al. (2001) showed that hand cycling is well suited for aerobic training. Understanding the effect of the crank position, and therefore the body position during hand cycling can provide useful information, which can be applied in training settings and therefore used to enhance performance.

In contrast to cycling, there has been a small amount of research in hand cycling in terms of the effect of different variables. Previous research for cycling showed us that there are two parameters for the crank position that may have an influence on the joint reaction forces and the muscle recruitment. This is comparable for hand cycling due to similarity in the propulsion system. For training purposes, it is important to know what the most efficient setup is for training certain muscles, while injuries are prevented and pain is limited as much as possible. 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). The relationship between shoulder forces and the crank position may therefore indicate the need for a certain position, when it comes to preventing injury and minimizing shoulder pain during training. Only Arnet et al. (2012) and Ahlers and Jakobsen (2016) performed a study about shoulder load during hand cycling. Arnet et al. tested the effect of different incline and speed conditions on a hand bike on a treadmill, while Ahlers and Jakobsen looked at Glenohumeral joint forces during a single condition.

When it comes to previous research investigating muscle recruitment during handcycling, a few more studies have been conducted (Felsner et al., 2016; Ahlers and Jakobsen, 2016; Litzenberger et al., 2015). However, not many studies looked at the effect of the crank position. Only Litzenberger et al. (2015) looked at the effect of different crank lengths and heights on muscular recruitment.

Not a single previous study looked at the effect of the crank position on the Glenohumeral joint forces. Also, to our knowledge no studies have been conducted about the effect of the crank position on muscle recruitment when it comes to the crank width, crank length and handlebar position separately, let alone in combination with each other and the crank height.

Without comprehensive understanding about the effect of the crank position on muscle recruitment and joint reaction forces, the settings for optimal training circumstances stays mainly intuitive. Furthermore, knowledge on that topic may reduce the risk of injuries and minimize pain of existing injuries, during the exercise. Related information however, is so far missing in scientific literature. Therefore, the purposes of this study were therefore to analyze the effect of different crank positions on the Glenohumeral joint forces (1) and muscle recruitment (2) with the use of a musculoskeletal model.

METHODS: Twelve male subjects without neurological or physical impairments and no hand cycling experience (height 180.2 ± 3.0 cm; mass 76.5 ± 6.8 kg) volunteered to participate in this study. The research was done in a laboratory on a hand cycling ergometer (SRM, Jülich, Germany) and differed from hand cycling with respect to steering and standing versus sitting position.

The participants had to perform a total of 16 trials. In 14 trials, the crank width (3 variations), crank length (3 variations) and handlebar position (2 variations) differed from each other. The crank height was the only variating factor for the remaining 2 trials. The order of the trials was chosen in such a way that the time between the trials, needed for adjusting the parameters, was minimized. Due to limitations of the ergometer, the crank height couldn't be adjusted, so the height of the standing platform was adjusted to realize a difference in relative crank height. In the neutral position the subject had to stand on a box of 150 mm, which resulted in a position with the upper arm in approximately 90 degrees ante flexion when the crank was in horizontal position. The other settings of the neutral position were a crank length of 200 mm, a crank width of 50 mm, a power output of 60 Watt and with handlebars in the horizontal position (perpendicular to the crank arm). The longest crank length, of 250 mm, was not used with a crank width of 150 and 250 mm due to physiological limitations. The participants had to maintain a crank rate of approximately 50 rpm during the trials. The power output was automatically kept constant by the SRM analysis software (SRM, Jülich, Germany).

All the data of the subjects during a total of 17 trials per person were recorded for a period of at least 8 cycles. Kinematic data were captured using a 3D motion capturing system (100 Hz, Vicon™, Oxford, UK). 52 markers were placed at bony landmarks, to drive the full body musculoskeletal model. A cycle started with the right handlebar in the lowest position, and ended when it reached that same position. Custom made handlebars, including Kistler 3-Component Force sensors (model 9251A), were used to capture hand force data at 1000 Hz.

A computational model was prepared in a multibody simulation software, the AnyBody Modelling System (version 6.0.5 (AnyBody Technology, Aalborg, Denmark). The model is made by adjusting the basic MoCapModel from the AnyBody Managed Model Repository 1.6.3. The data of each separate cycle was used to build a musculoskeletal model with AnyBody for the corresponding cycle.

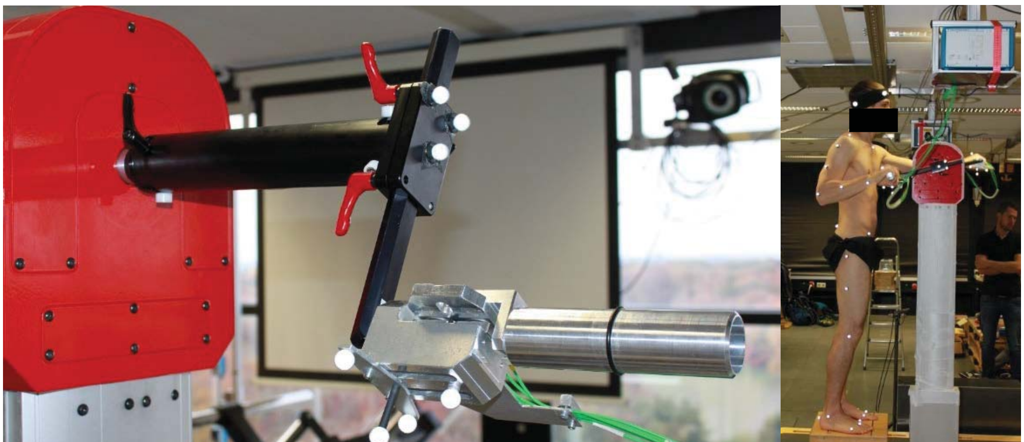

Figure 1: Custom made handlebar including a 3D force sensor (Kistler) in horizontal setup with a 150 mm crank arm and a 250 mm crank width (left). A subject performing a trial of the testing protocol (right).

MATLAB R2015b (MathWorks, Inc.) was used for post processing and descriptive statistics. This resulted in an average value for each muscle and joint reaction force for each trial. The muscles in the model comprised of serval subdivisions, which constitute the different directions of muscle movement. Therefore, all subdivisions for each muscle, were enveloped in order to represent the muscle force. Furthermore, joint reaction forces for the Glenohumeral joint were exported as well as kinematic trajectories for the wrist, the shoulder and the elbow.

RESULTS AND DISCUSSION: The data that is presented corresponds to that of the right arm. The cycle started when the right arm was in the lowest position, so with vertical crank arms.
Mauchly's test of sphericity indicated that the assumption of sphericity had been violated for the muscle forces (x^2 = 203,537, *p* = .00) and for the joint reaction forces (x^2 = 203,537, *p* = .00), so the Greenhouse-Geisser correction was applied. There was a significant difference between crank settings and muscle forces (*F* = 5.636, *p* = 0.006) and the crank settings and Glenohumeral joint reaction forces and moments (*F* = 7.138, *p* = 0.009). Figure 1 presents the mean Glenohumeral joint reaction forces during one cycle.

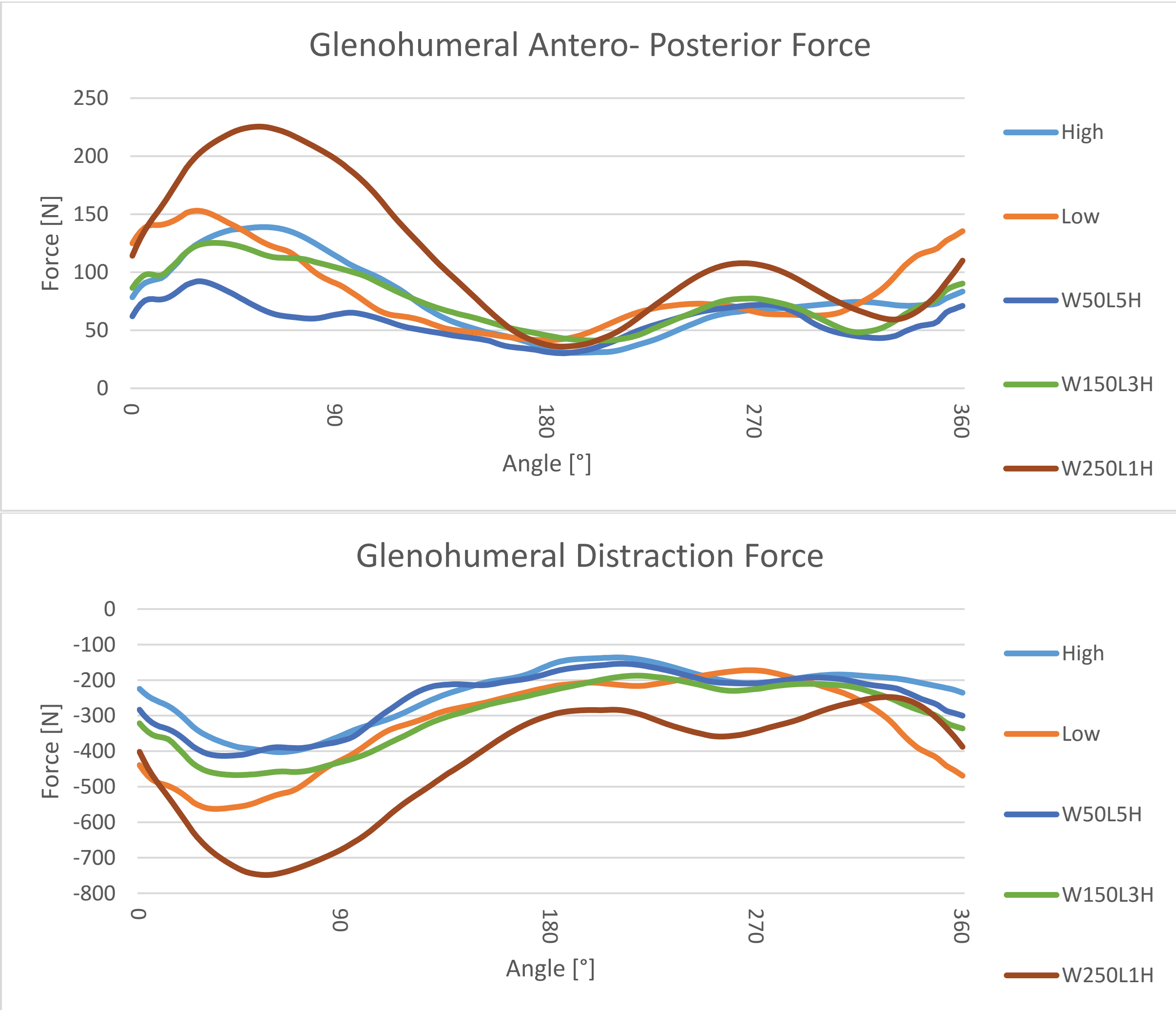

Figure 1: Mean Glenohumeral joint reaction forces (N) during one cycle. The Antero force and distraction force are viewed as positive in the figures above. A cycle started with the right handlebar in the lowest position, and ended when it reached that same position. Crank setting description: 'H' represents horizontal position, which is the position with the handlebars perpendicular to the crank arm. 'W' shows the width of the crank and 'L' the length of the crank arm. 'High' is with the subject on a box of 300 mm, and 'Low' is with the subject on the platform. All the other trials are done on a box of 150 mm, which is the neutral height.

The Glenohumeral distraction force has been acknowledge as a risk factor for developing shoulder injuries (Stuelcken et al., 2010, Werner et al., 2006). Therefore, minimization of the Glenohumeral distraction force due to certain crank settings potentially minimize the injury risk. The distraction forces are lower with the use of horizontal handlebars compared to vertical handlebars. However, there was no significant difference. This can be explained due to the large standard deviations with the vertical settings. Nevertheless, there was a significant difference in crank arm lengths between settings with horizontal handlebars. The short crank arm lengths resulted in higher distraction forces then the other lengths.

The antero- posterior force is provided by the rotator cuff and is critical for stability and concentric rotation of the humeral head on the glenoid (Reuther et al., 2014). An example of shoulder instability is dislocation. In over 95% of shoulder dislocations, the humerus is displaced anteriorly. Meaning the minimization of the antero- posterior force should be taken seriously. The settings with vertical handlebars results in higher, but not significant different forces then the other settings. It does appear that a short crank arm results in significantly higher antero- posterior forces then other settings. In this case, a long crank arm, horizontal handlebars and small crank width (W50L5H) seems to be the most optimal setting.

Three muscles had a significant higher muscle force, expressed as a percentage of the maximum force for that particular muscle, in certain position compared to others. For the m. Brachialis, this seems to be the case with a long crank arm and horizontal handlebars. The m. Deltoideus Scapularis and m. Latissimus Dorsi have higher muscle forces with a wide crank, short crank arm and horizontal handlebars.

CONCLUSION: The most important findings of this research are: 1) There seems to be a lot of variation in muscle forces and Glenohumeral joint reaction forces between subjects with the use of vertical handlebars, 2) The m. Brachialis has a higher muscle force with long crank arms and horizontal handlebars, and the m. Deltoideus Scapularis as well as the m. Latissimus Dorsi have higher muscle forces with a wide crank, short crank arm and horizontal handlebars and 3) a setting with horizontal handlebars, a small crank width and long crank arms seems to be optimal for minimizing the risk on shoulder injuries.

REFERENCES:

Arnet U., Van Drongelen S., Van Der Woude L.H.V., Veeger D.H.E.J. (2012). Shoulder load during handcycling at different incline and speed conditions. *Clinical biomechanics*, 27, 1-6

Ahlers F.H., Jakobsen L. (2016). Biomechanical analysis of hand cycling propulsion movement: A musculoskeletal modelling approach. *Master thesis – Sports technology, Aalborg University*

Felsner E., Litzenberger S., Mally F., Sabo A. (2016). Musculoskeletal modelling of elite hand cycling motion: evaluation of muscular on- and offset. *Procedia Engineering*, 147, 168-174

Janssen T.W., Dallmeijer A.J., Van Der Woude L.H.V. (2001). Physical capacity and race performance of handcycle users. *J Rehabil Res Dev*, 38(1): 33-40

Litzenberger S., Mally F., Sabo A. (2015). Influence of different seating and crank positions on muscular activity in elite handcycling – a case study. *Procedia Engineering*, 112, 355-360

Reuther E.K., Thomas S.J., Tucker J.J., Sarver J.J., Gray C.F., Rooney S.I., Glaser D.L., Soslowsky L.J. (2014). Disruption of the anterior-posterior rotator cuff force balance alters joint function and leads to joint damage in a rat model. *J Orthop Res*, 32(5), 638-644

Stuelcken C.M., Ferdinands R.E.D., Ginn K.A., Sinclair J.P. (2010). The shoulder distraction force in cricket fast bowling. *Journal of Applied Biomechanics*, 26, 373

Van Drongelen S., Van Der Woude L.H.V., Janssen T.W., Angenot E.L., Chadwick E.K., Veeger D.H. (2005a). Glenohumeral contact forces and muscle forces evaluated in wheelchair- related activities of daily living in able-bodied subjects versus subjects with paraplegia and tetraplegia. *Arch Phys Med Rehabil*, 86, 1434-1440

Van Drongelen S., Van Der Woude L.H.V., Janssen T.W., Angenot E.L., Chadwick E.K., Veeger D.H. (2005b). Mechanical load on the upper extermity during wheelchair activities. *Arch Phys Med Rehabil*, 86, 1214-1220

Veeger H.E., Rozendaal L.A., Van Der Helm F.C. (2002). Load on the shoulder in low intensity wheelchair propulsion. *Clin Biomech*, 17, 211-218

Werner S.L., Jones D.G., Guido J.A. jr., Brunet M.E. (2006). Kinematics and kinetics of elite windmill softball pitching. *The American Journal of Sports Medicine*. 34(4), 597