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Shoulder and lower back joint reaction forces in seated double poling

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Abstract

Overuse injuries in the shoulders and lower back are hypothesized to be common in cross-country sit-skiing. Athletes with reduced trunk muscle control mainly sits with their knees higher than hips (KH). To reduce spinal flexion, a position with the knees below the hips (KL) was enabled for these athletes using a frontal trunk support. The aim of the study was to compare the shoulder joint (glenohumeral joint) and L4-L5 joint reactions between the sitting positions KL and KH.

Five able-bodied female athletes performed submaximal and maximal exercise tests in the sitting positions KL and KH on a ski-ergometer. Measured pole forces and 3-dimensional kinematics served as input for inverse-dynamics simulations to compute the muscle forces and joint reactions in the shoulder and L4-L5 joint.

This was the first musculoskeletal simulation study of seated double poling. The results showed that the KH position was favorable for higher performance and decreased values of the shoulder joint reactions for female able-bodied athletes with full trunk control. The KL position was favorable for lower L4-L5 joint reactions and might therefore reduce the risk of lower back injuries. These results indicate that it is hard to optimize both performance and safety in the same sit-ski.

Keywords: musculoskeletal modeling, inverse-dynamics simulations, muscular metabolic power, cross-country sit-skiing

Word count: (Introduction- Discussion): 3529 (4000)

Introduction

For wheelchair users, lower back pain[1,2] and shoulder injuries[3] are common. Several risk factors have been identified for lower back pain including bad posture and instability of the spinal column as well as high peak shear force and high cumulative compression on the lumbar spine.[4] Flexion of the spinal column is an inappropriate posture because it is related to other risk factors. For example, flexion of the spinal column while lifting cause high anterior shear force in the intervertebral discs.[5] Also, sitting causes the pelvis to tilt backward and the spine to flex slightly compared to standing, and as a consequence the lumbar joint reactions forces increase.[6] After long periods of lifting or using a flexed posture of the spine, the stretching of the viscoelastic structures in the spinal region reduces the proprioceptive function of the mechano-receptors.[7] Reduced proprioception impairs the stabilization of the spine and thereby increases the risk of injuries.

Shoulder pain is a major problem for people with spinal cord injuries because the arms care for all mobility during daily activities. A common shoulder injury is subacromion impingement syndrome.[8] One mechanism of this syndrome is reduced subacromial space, which compresses the tissue in that region and causes inflammation and pain.[9] In wheelchair athletes this injury is often present together with muscular imbalance, i.e. greater strength of the shoulder abductors than adductors and higher strength of the inward rotators than outward rotators.[10] It is also discussed that the ergonomics of the sitting posture, due to a backward tilt of the pelvis and flexion of the spine, increase the risk of shoulder injuries.[11] As far as the authors know, there have not been any studies of shoulder reaction forces in relation to spinal curvature published previously.

Joint reactions are difficult to measure in vivo because all alternatives are invasive procedures. For example, lumbar intervertebral disc pressure has been measured by an implanted pressure transducer[12] and shoulder joint reactions have been measured by an instrumented implant.[13] Another option is inverse-dynamics simulations, which compute muscle forces and joint reactions from non-invasive measurements of body kinematics and external forces, e.g. the AnyBody Modelling System (Anybody Technology A/S, Aalborg, Denmark). The AnyBody Modelling System has been used for injury prevention purposes, where joint reactions have been studied in for example lifting,[14, 15] wheelchair propulsion[16] and in volley-ball.[17] Validation studies of simulated compression forces between vertebrae L4 and L5 show good agreement to measured reactions.[15, 18] Comparison between the AnyBody Modelling System and in vivo measurements[13] has shown quite good agreement for shoulder joint reactions.[19]

In able-bodied cross-country skiing, lower back injuries are a common problem and believed to arise because of the large range of motion and monotony of lower back flexion-extension.[20] Cross-country sit-skiing is an endurance Paralympic sport where the athletes sit in a sledge that is mounted on a pair of skis and propel themselves using poles. Even though no study of injury prevalence in cross-country sit-skiing exists as far as the authors know, we speculate that there is a risk for overuse injuries in cross-country sit-skiing in the shoulders (as in wheelchair usage) and in the lower back (as in cross-country skiing).

The cross-country sit-skiing athletes use many different sitting positions.[21] Athletes with full trunk muscle control mainly use a knee-seated position (knees lower than hips). Athletes with severely reduced trunk muscle control often sit with their knees higher than their hips and the lower back resting against a support from the sledge (KH). The KH position can lead to a kyphotic spinal curvature (flexion of the spine) and therefore the KH position is hypothesized to be associated with high joint reactions in both the lumbar spine and the shoulders. For this study, a new sit-ski sledge was designed with the intention to enable athletes with reduced hip and trunk control to sit in a knee-seated position (knees lower than hips) using a frontal trunk support (KL).

This was the first musculoskeletal simulation study of seated double poling. The overall objective was to further the understanding of how the sitting position influences performance and risk for injuries. The specific aim was to compare the joint reaction forces, in the L4-L5 joint (the joint between lumbar vertebra 4 and 5) and in the shoulder joint, between the sitting positions KL and KH during seated double poling.

Methods

Participants: Five female able-bodied national class athletes in cross-country skiing (28.5 ± 4.5 year, 62.6 ± 8.1 kg, 1.67 ± 0.05 m) participated in the study. The study was approved by the Regional Ethical Review Board in Umeå, Sweden (Dnr 2013-412-31M and Dnr 2015-74-32M) and informed consent was obtained from all participants.

Physical tests: During the week before the physical tests, the participants performed two familiarization sessions, with seat adjustments and 45 min exercise time in each. The sessions involved training in both sitting positions including the initial stages of the sub-maximal protocol and five 15 s maximal effort intervals. The physical tests included two sets of exercise tests, one in each sitting position, KL and KH, in randomized order separated by at least 48 hours. Each set included a sub-maximal incremental test and a maximal time-trial of 3 min (MAX). The submaximal test comprised 4-7 exercise intensities of 3 min depending on participants’ fitness. The exercise intensity nr 4 (37W, approximate blood lactate concentration 4 mmol/L and respiratory exchange ratio 1), and MAX were chosen for further analysis. These physical tests was a part of a larger study; for a more extensive description of the tests, standardization, measurements and equipment see, Lund Ohlsson and Laaksonen.[22]

Measurements and equipment: Two different sit-skis with sitting positions, KL and KH respectively (Ableway AB, Östersund, Sweden), were mounted on a ski-ergometer (ThoraxTrainer, ThoraxTrainer A/S, Kokkedal, Denmark). The sitting height was for KL 0.38

m and for KH 0.33 m (mimicking similar height to the skiers'center of mass, KL: 0.63 ± 0.01 m, KH: 0.62 ± 0.01 m).

Power output for each stroke was computed by the software of the ergometer (ThoraxTrainer ver 1.01, ThoraxTrainer A/S, Kokkedal, Denmark). Blood lactate concentration was determined from ear lobe capillary blood samples with a Biosen C-line (EKF diagnostic GmbH, Magdeburg, Germany). Pole forces were measured at 250 Hz using a uniaxial strain gauge load cell mounted between handle and pole (Biovision, Wehrheim, Germany).

Three-dimensional kinematic data was recorded at 200 Hz with eleven Oqus3+ cameras (Qualisys AB, Gothenburg, Sweden). A full-body marker set was employed (modified Plug-in-gait, www.vicon.com) with 40 markers (Fig. 1), diameter 12 mm, including three on each pole, for details see Lund Ohlsson et al.[23] Joint angles were defined as: hip and knee according to Holmberg and Lund[24], spine angle as angle between pelvis and trunk in the sagittal plane (flexion negative), and shoulder angles according to International Society of Biomechanics.[25] The shoulders angles used rotation order x, z, y (y – axis directed towards the glenohumeral joint from the mid of medial and lateral epicondyles, z – axis perpendicular to the plane formed by the glenohumeral joint, lateral and medial epicondyles pointing backwards).

Analysis: Performance was defined as

$$Performance = \frac{\text{Mean power output in MAX}}{\text{body mass}}. \qquad (1)$$

Kinematics and kinetics were analyzed and simulations were performed over 4 cycles after 120 s in exercise intensity 37W and after 60 s in MAX. The start of the poling cycle was defined as when the pole tip was in its foremost position. A complete poling cycle comprised a poling phase (pole tip moving backward) and a return phase (pole tip moving forward).

Calibration of load cells in the poles were made for 0, 5, 10, 15 and 20 kg to compute the transformation function from voltage to force. The loads were applied axially on the handles with no poles attached to the handles; therefore calibration was made without bending the poles.

Simulations: Participant specific inverse-dynamics musculoskeletal simulation models were built in the AnyBody Modelling System v. 6.0 for both sitting positions. The actual dimensions of the skiers, i.e. body mass and height, were used to scale the models. The segment masses and inertia properties were scaled from body mass according to Winter.[26] Data from measurements of kinematics and pole forces of the 37W and MAX intensity over 4 poling cycles was used to drive the simulation models. The simulations derived the muscle forces (f_i), joint reactions and the muscle specific metabolic power (mMP_i). The simulation models were modifications of the full-body MoCap model, available in the AnyBody Managed Model Repository v.1.6.3 (www.anybodytech.com), with added poles and sit-ski. This full-body model comprised 41 rigid segments and around 700 muscle actuators. To limit total simulation model complexity the muscle actuators used a constant force model, i.e. maximum attainable force was constant over both length and velocity, and included no tendon unit. The body model and the sit-ski were connected to each other by both hard constraints (no motion) and soft constraints (motion). The hard constraints were defined in the ankles (both in KL and KH), knees and seat (only in KL). The soft constraints were created by a dynamic contact model comprising a contact point on the body model and a contact zone on the sit-ski (cylindrical boxes in Figure 1, KL: frontal trunk support, KH: knee support, seat, and backrest). Whenever the contact point was in the contact zone a set of virtual muscles were included to represent the reaction forces.[27]

The glenohumeral joint (shoulder joint) was modelled as a spherical joint (three rotational degrees of freedom) between the humerus and scapula (wiki.anyscript.org). The only ligament included in the shoulder model was the conoid ligament. The direction of the shoulder reaction forces were modelled from the center of the head of humerus into the glenoid cavity (wiki.anyscript.org). The lumbar spine comprises 5 vertebrae (with spherical joints in-between), 188 muscle fascicles and intra-abdominal pressure.[28]

A digital low-pass filter with a cut-off frequency of 10 Hz was used for filtering the measured kinematics data. The data was then matched to the body model in the AnyBody Modeling System by an optimization procedure;[29] through this procedure together with the estimated segments masses the body models' segment lengths were scaled to match the participants' body height and kinematics. Thereafter, muscle strengths were scaled using segment masses and segment lengths through the ScalingLengthMassFat function in the software. The joint angles were also a result of this procedure.

Inverse-dynamics computations were performed in the AnyBody Modeling System to obtain joint reactions and muscular forces. The inverse problem was formulated as a static optimization problem where the objective function described how the redundant muscle recruitment problem should be solved subject to the multi-body system.[30] The muscle recruitment objective function was a 5th order polynomial of the relative muscle forces (force divided by strength for each muscle). The 5th order polynomial criteria was chosen as a tradeoff between 2nd order and infinite order polynomial based on higher validity of the L4-L5 compression forces using the software AnyBody Modeling System,[15] and high cooperation between muscles during high-intensity endurance exercise.[31] The L4-L5 joint compression force (orthogonal to L4) and anterior shear force (tangential to L4 pointing anterior) were normalized to the joint reactions for each participant in a standing position. The participant specific standing joint reactions were obtained using the individualized body models from the sitting position KL. The individualized body models were simulated in a theoretical standing position (derived from the StandingModel in the AnyBody Managed Model Repository v.1.6.3) with the same joint angles (i.e. the same body posture for all five individual simulation models).

The total shoulder joint reaction force and its inferior-superior component (in the scapula reference frame) were computed. The total shoulder joint reaction was normalized to body weight.

Based on Beltmann et al.[32] for each muscle *i* in the simulation model, muscular metabolic power ($\mathrm{mMP_i}$) was defined as

$$\mathrm{mMP_i} = \begin{cases} f_i \cdot v_i/1.25 & \text{if } v_i > 0 \\ -f_i \cdot v_i/0.25 & \text{if } v_i < 0 \end{cases}, \qquad (2)$$

where f_i is the muscle force and v_i is the contraction speed of muscle *i*. Positive contraction speed was defined as lengthening of muscle fiber. By Eq. 2, eccentric muscle work costs less mMP than concentric and no cost was associated with contraction where muscle length was constant.[32]

Muscular metabolic power for two muscle groups, arms (muscles with insertion on the arm) and spine (muscles in the trunk and neck without insertion on the legs or arms), was computed as

$$\mathrm{mMP_{group}} = \frac{\sum_{i,group} \int_0^{cycle\ time} \mathrm{mMP_i}\, dt}{\text{cycle time}}. \qquad (3)$$

The proportion of the power from each muscle group mMP to total mMP were defined as Rel $\mathrm{mMP_{group}}$.

Statistics: Statistical significance was set at $\alpha \leq .05$. Data was compared pair-wise between the two sitting positions with two-sided paired Student t-tests when normality was observed (Shapiro-Wilk), otherwise with Wilcoxon's signed rank test.

Results

Performance was higher in position KH (KL: 0.77 ± 0.08 W·kg^{-1}, KH: 1.00 ± 0.14 W·kg^{-1}, p = 0.008). No difference was observed in cycle length or cycle time (p > .05). Joint kinematics showed that KL had more extended hips, less maximal spine flexion and smaller range of motion in flexion (Figure 2; Supplemental Movie 1). The KH position was associated with a smaller pole angle (a more horizontal pole) at the end of the poling phase (Table 1) and a wider elbow in the start of the poling phase for MAX (maximal angle of shoulder angle 2, p = .043, and range of motion, p = .041, Figure 2). Pole forces showed higher mean values for KH compared to KL in 37W but no significant difference in MAX. The peak values of the pole forces were not different between sitting positions (Figure 2).

The peak L4-L5 joint reactions for compression and anterior shear force (Table 1) were larger for KH in MAX but showed no significant difference between sitting positions in 37W. For normalization purposes the upright standing joint reactions were computed (compression 354 ± 45 N, anterior shear 32 ± 11 N). The normalized L4-L5 joint reactions (Figure 3) were larger in KH than KL, especially during MAX when also the power output was higher for KH. For exercise intensity 37W, the anterior shear force mean was larger for KH compared to KL while the compression force mean was not significantly different (p = .063).

The accumulation of compressive load was higher during the poling phase than the return phase (Figure 4). The accumulated anterior shear load was more evenly distributed during the poling phase (Figure 4).

In the shoulders (Figure 5), KL resulted in larger peak and mean reactions than KH for exercise intensity 37W (i.e. equal power output) while no difference between the sitting positions was observed during MAX (higher power output for KH). For the direction of the shoulder reaction vectors see Supplemental Movie 1. The inferior-superior component of the shoulder reaction was not different between the sitting positions (Table 1). The reaction force profile (Figure 6) for exercise intensity 37W showed that peak shoulder joint reactions occurred during the poling phase. The peak shoulder joint reactions normalized to body weight were not different between sitting positions (Table 1).

mMP_{arms} was larger for KL than KH in 37W but showed no difference in MAX. Instead, mMP_{spine} showed no difference in 37W but was higher for KH compared to KL in MAX. In relative terms, Rel mMP_{arms}, was larger in KL and Rel mMP_{spine} larger in KH (Table 1). Relative muscle forces for selected muscles in both sitting positions are shown in Figure 7.

Discussion

This was the first musculoskeletal simulation study of seated double poling investigating joint reaction forces and muscular forces in the body. The purpose of this study was to compare lower back and shoulder joint reaction forces between the sitting positions KL and KH. The results of this study confirmed the hypothesis that the KH position causes higher normalized L4-L5 joint reaction forces, both compression and anterior shear, along with a larger spine flexion. The results also rejects the hypothesis that larger spinal flexion would be associated with higher shoulder joint reaction forces because the opposite is shown. The posture with less spinal flexion (KL) enabled generation of more arm muscle power and induced a higher shoulder joint resultant reaction but there was no difference in the inferior-superior component. This study also showed that larger spinal flexion increased the lumbar compression force, which is consistent with the result from Wilke et al.[12] who measured higher pressure in the L4-L5 disc during spinal flexion compared to normal spinal curvature. Simulation studies of lifting tasks have also shown that the L4-L5 compression force increases with more spinal flexion.[15, 18]

The range of the resultant of the reaction force in the shoulder joint was 266-312 % of body weight, and therefore higher than the 150 % of body weight measured during daily activities.[13] This seems realistic because double poling is a very strenuous activity. Even though our study had higher relative exercise intensity (KL 79 %, KH 61 %) and maximal intensity, the peak shoulder reactions (1200-2500 N, Figure 5) were similar to a quasi-static 2-dimensional simulation of wheelchair rolling (1900 N, 40% of maximal intensity).[33] A 3-dimensional simulation of wheelchair rolling in comfortable self-selected speed showed much lower peak force of 300 N.[16] The hypothesis in this study was that a larger spinal flexion would have caused scapula to rotate forward and downward in the sagittal plane, thus depressing the acromial process onto the glenohumeral joint and reducing the subacromial space, inducing a higher inferior-superior shoulder joint reaction component. The results showed that higher arm muscle power and lower spinal flexion were related to higher values of shoulder joint resultant reactions, but not the inferior-superior component. It is possible that the shoulder model was too simplistic to capture the impact of spinal flexion on the shoulder because no mechanical contact model between acromion and humerus was included.

It is difficult to predict whether the order of magnitude for the reaction forces in this study could increase injury risk. For comparison, in the current study the lumbar spine peak compression force was 1855 N in sitting position KH, which is well below the recommended limit for males of 3400 N by Waters et al.[34] (no limit for females found). High cumulative lumbar shear forces have been shown to be related to higher injury risk in industry workers.[4] Full flexion of the spine can reduce the ability of the major lumbar extensor muscles to resist shear force and should therefore be avoided.[5] According to these arguments, the KH position with its higher shear force throughout the poling cycle, higher peak compression force, larger spinal flexion and higher pole forces might therefore be associated with a higher risk of lower back injury, e.g. disc herniation.

Shoulder pain and injuries can arise from many different causes, for example lack of joint stability, low flexibility, bad spinal posture and overloading.[8, 9, 33] The KL position had the highest shoulder joint resultant reactions but the KH position had higher spinal flexion.

Therefore, according to the results of this study no conclusion can be made with regard to shoulder injury risk for each sitting position. Muscular balance around the shoulder is important to reduce shoulder pain[10] and the current study showed higher relative muscle forces in the posterior shoulder muscles m. deltoideus posterior part and m. trapezius, compared to the anterior muscles m. deltoideus anterior part and m. pectoralis major (Figure 7). However, seated double poling might be positive as complementary training for wheel chair users because of larger utilization of posterior shoulder muscles.[34]

The current study showed a connection between muscular metabolic power in the region around the joint and the joint reaction force. In the spine, both higher mMP_{spine} and higher joint reactions (compression and anterior shear) in the L4-L5 joint were present, i.e. higher for KH than KL, especially in MAX. In the arms, both higher mMP_{arms} and higher shoulder joint reaction in 37W (the equal power output) were present, i.e. higher for KL than KH. Similar results are also present in the relative muscle forces (Fig 7), higher force in m. erector spinae for KH during both 37W and MAX and higher force in m. deltoideus posterior part, m. infraspinatus, m. latissimus dorsi, m. triceps brachii for KL.

Both the L4-L5 joint and the glenohumeral joint were modelled as spherical joints with no sliding. Therefore, no muscle activation was needed for translational stabilization of the joint. If sliding had been included in the model, the relative muscle forces might have been different.

To conclude, for female able-bodied athletes with full trunk control, the KH position was favorable for performance and shoulder joint resultant reactions. However, the KL position was favorable due to lower L4-L5 joint reactions and might thus be associated with a decreased risk of lower back injuries. This implies that both the parameters performance and safety cannot both be optimized in the same sit-ski. This study also showed that the joint reactions were related to both the muscular metabolic power and the relative muscle forces in the region around the joint. Therefore, a para-athlete with highly reduced trunk muscle control might respond differently than the athletes tested in this study.

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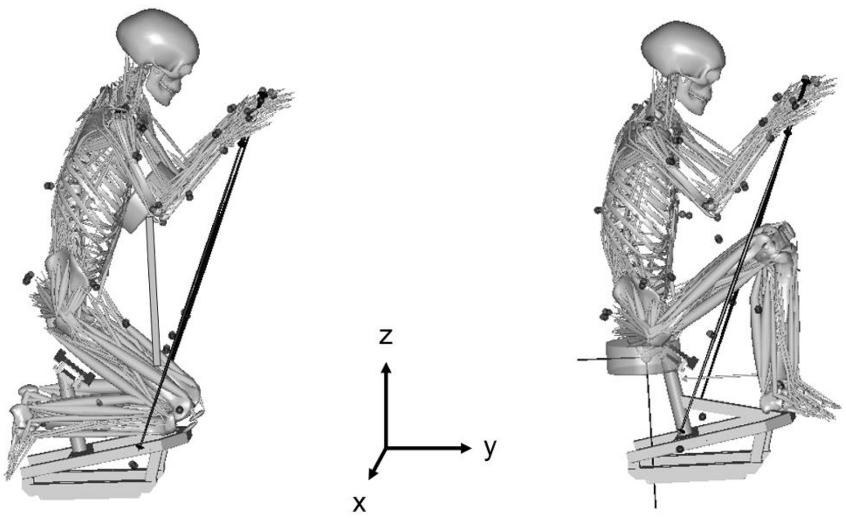

Figure 1 - The simulation models at the start of the poling phase. To the left (a) the position with knees lower than hips (KL) and to the right (b) the position with knees higher than hips (KH). The cylindrical boxes are conditional contact points to the sit-ski, for KL (frontal trunk support) and for KH (back support, the seat and the support in the fold of the knees).

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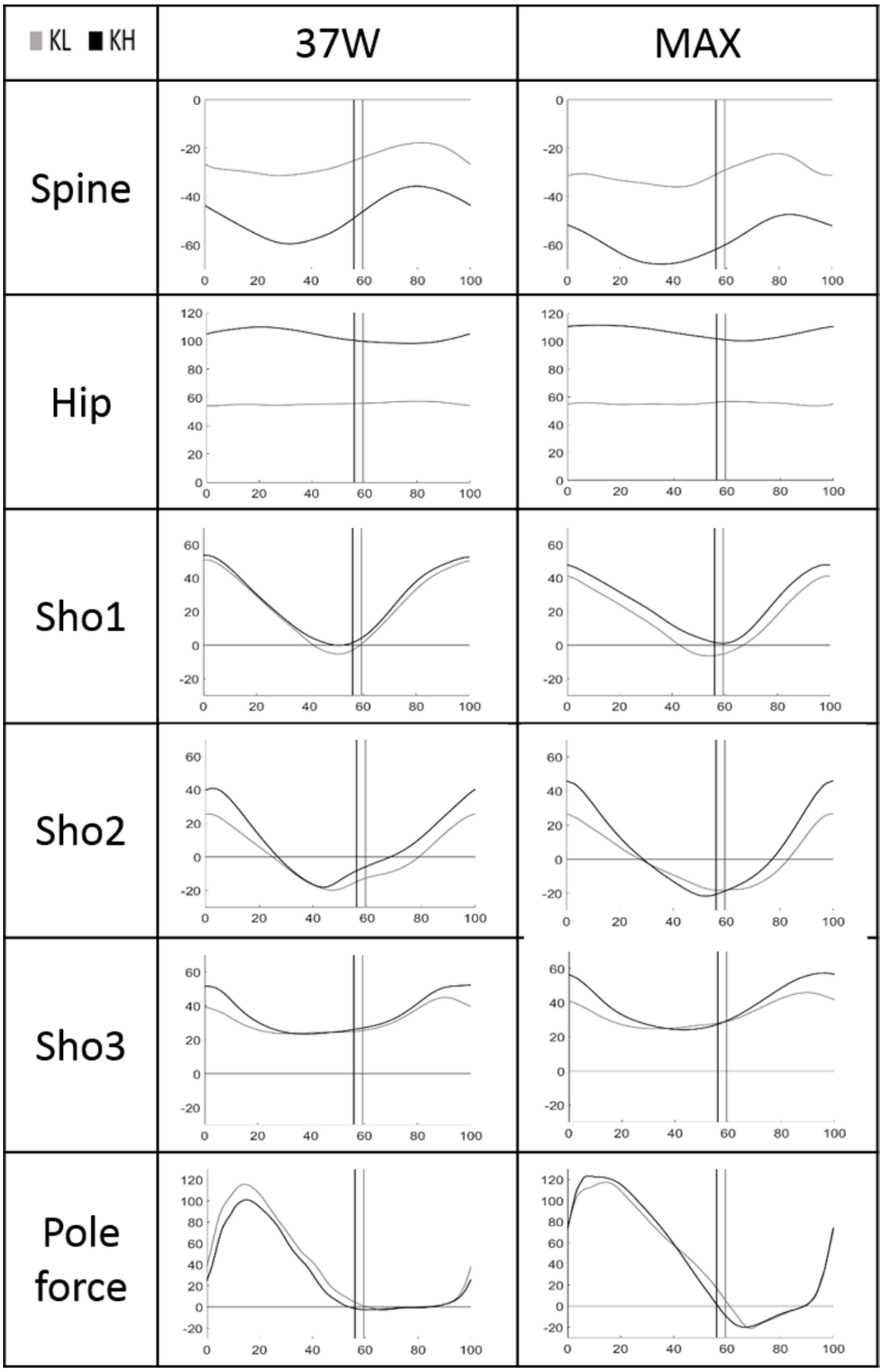

Figure 2 - Selected joint angles [°] and the pole force [N] over the poling cycle (0-100 %) for exercise intensity 37W and maximal intensity (MAX). Mean profile of the five participants for sitting position KL (grey) and KH (black). Definitions of joint angles, in anatomical position, are: spine flexion = 0° (angle between pelvis and trunk in the sagital plane with flexion negative), hip = 0° (flexion positive), shoulder angle 1 (Sho1) = 0° (flexion positive, extension negative), shoulder angle 2 (Sho2) = 0° and shoulder angle 3 (Sho3) = 0°.

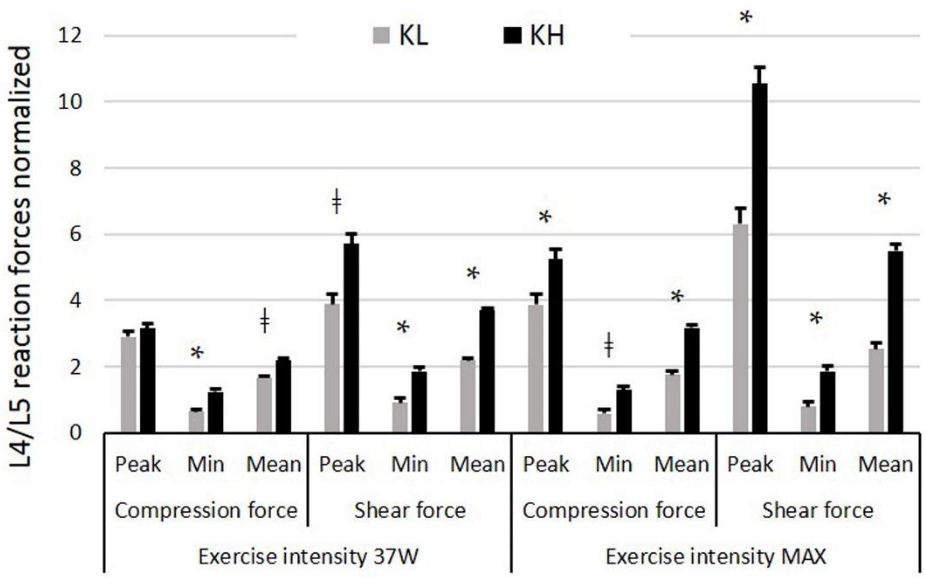

Figure 3 - Normalized L4-L5 joint reaction forces [N/N] (normalized to the joint reactions in standing for each participant), compression and anterior shear forces, in the joint for the two sitting positions knee high (KH) in black and knee low (KL) in grey. Peak - maximal force, Min - minimal force, and Mean - mean force over the poling cycle are presented for exercise intensity 37W and maximal intensity (MAX). Significant difference marked (*) and tendency of difference (.05 ≤ p < .10) marked (ǂ). Error bars shows standard deviation.

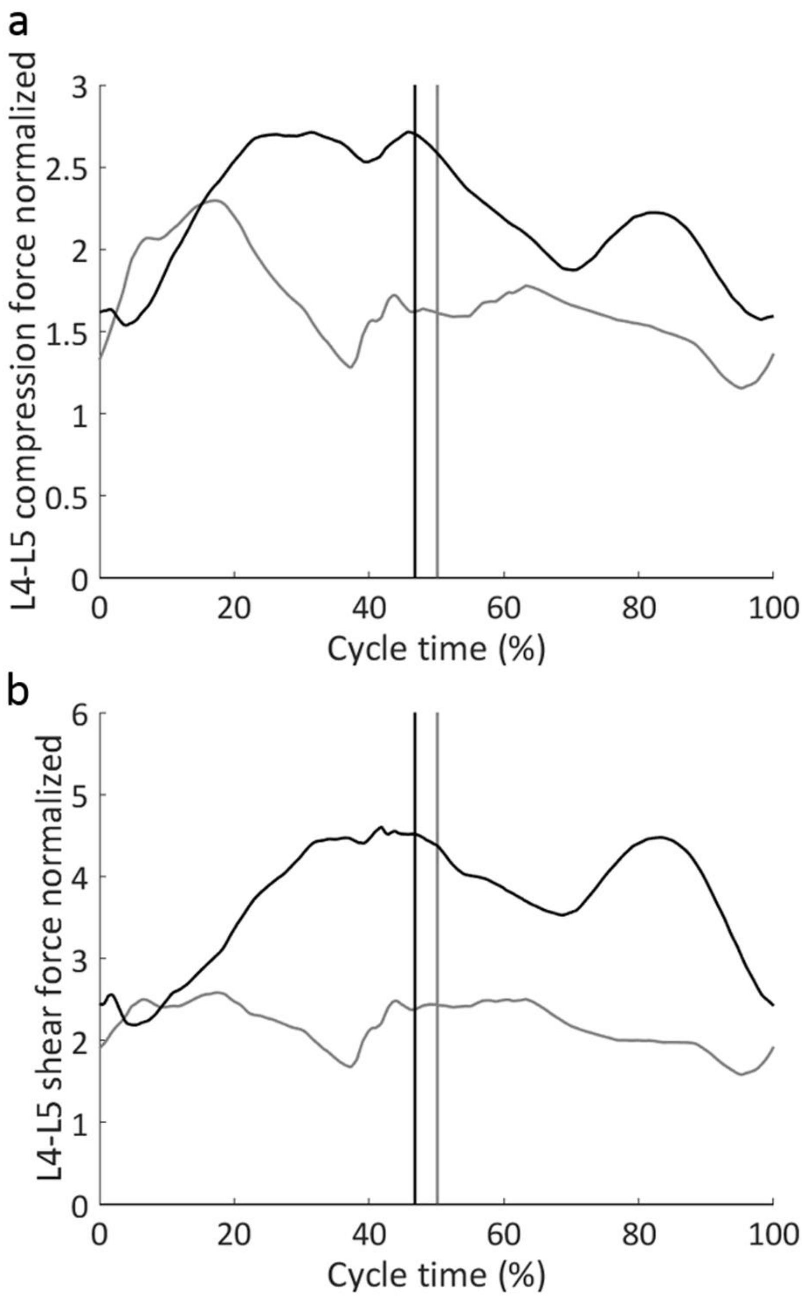

Figure 4 - Force profiles during exercise intensity 37W for the L4-L5 compression force (a) and anterior shear force (b) normalized to standing [N/N]. Sitting position knee high (KH) in black, and knee low (KL) in grey.

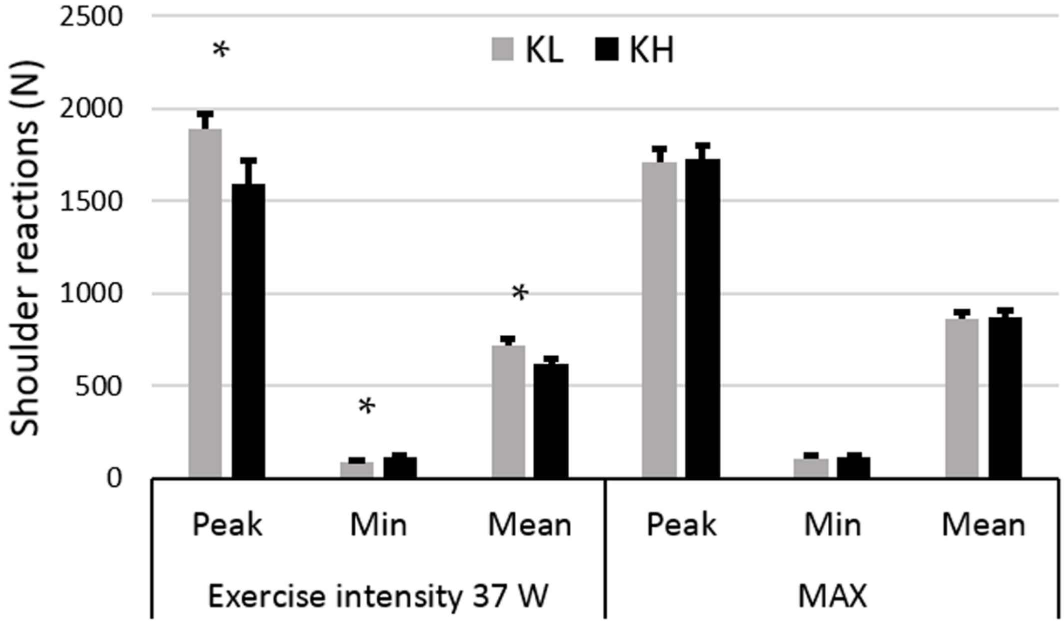

Figure 5 - Shoulder joint resultant reaction force for the two sitting positions knee high (KH) and knee low (KL). Peak - maximal force, Min - minimal force, and Mean - mean force over the poling cycle are presented for exercise intensity 37W and maximal intensity (MAX). Significant difference marked (*). Error bars shows standard deviation.

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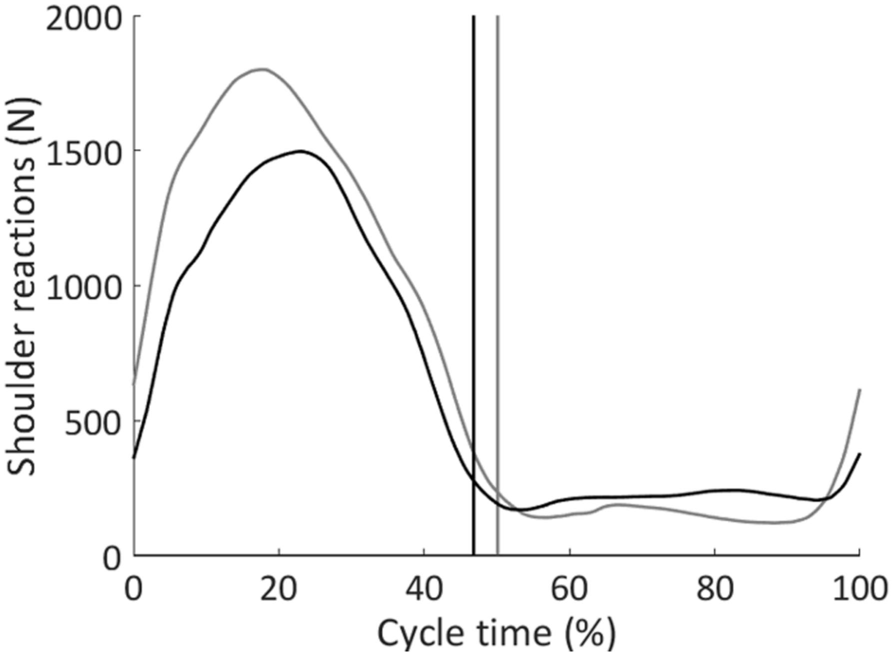

Figure 6 - Force profiles during exercise intensity 37W for the shoulder joint reaction. Sitting position knee high (KH) in black, and knee low (KL) in grey.

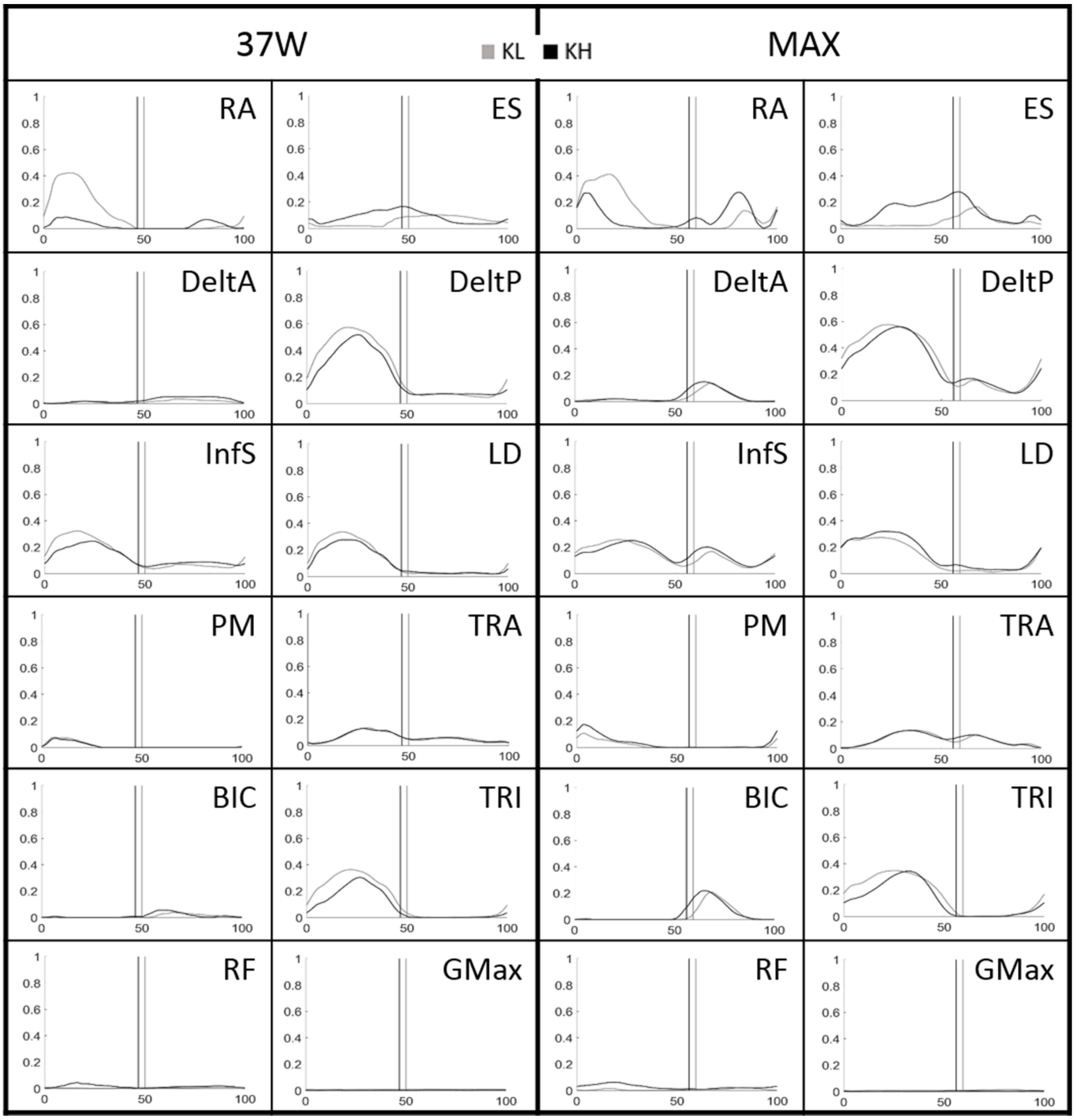

Figure 7 - Relative muscle force for selected muscles during exercise intensity 37W and maximal intensity (MAX) in sitting positions knee low (KL) and knee high (KH). Profiles are the mean of 5 participants and over 4 poling cycles. Muscles are: RA- m. rectus abdominis, ES- mean of muscle parts m. erector spinae passing vertebra L1, DeltA- m. deltoideus anterior, DeltP- m. deltoideus posterior, InfS- m. infraspinatus, LD- m. latissimus dorsi, PM- pectoralis major, TRA- m. trapezius with origin on scapula, BIC- m. biceps brachii, TRI- m. triceps brachii, RF- m. rectus femoris and GMax- m. gluteus maximus.

Table 1. Results of the pole angle at the end of the poling phase, the peak L4-L5 compression force and shear reaction force, the peak shoulder reaction force normalized to body weight, and the mean shoulder reaction force component in the inferior-superior direction from the head of humerus. Muscular metabolic power (mMP) for arms and spine in both absolute (mMP_{arms}, mMP_{spine}) and relative numbers (Rel mMP_{arms}, Rel mMP_{spine}).

	37W			MAX		
Variable	KL	KH	*p*	KL	KH	*p*
Pole angle (°)	18 ± 0.5	13 ± 0.6	< .001	20 ± 0.7	14 ± 0.5	< .003
L4-L5 peak compression (N)	1041 ± 92	1120 ± 53	.587	1359 ± 106	1855 ± 109	.0267
L4-L5 peak shear (N)	122 ± 11	176 ± 10	.051	188 ± 16	321 ± 14	.011
Shoulder peak reaction normalized to body weight (%)	310 ± 65	264 ± 64	.060	282 ± 45	285 ± 54	.824
Shoulder mean inferior-superior component (N)	137 ± 14	130 ± 8	.626	222 ± 13	190 ± 12	.256
mMP_{arms} (W)	483 ± 43	405 ± 54	.017	728 ± 150	830 ± 80	.105
mMP_{spine} (W)	22 ± 19	36 ± 8	.139	37 ± 17	92 ± 30	.012
Rel mMP_{arms} (%)	94 ± 4	87 ± 2	.013	92 ± 4	83 ± 2	.002
Rel mMP_{spine} (%)	4 ± 3	8 ± 2	.063	5 ± 2	9 ± 2	.015

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Supplemental Movie 1 - Musculoskeletal simulations of one of the participants for exercise intensity 37W in sitting positions, knee low (KL) to the left and knee high (KH) to the right. In green is the rigid structure of the sledges. The conditional contact areas are shown in grey. Shoulder joint reaction forces are represented by blue lines with origin from the joint center. When muscle activation increases, the muscle volume increases and the color goes from pink to darker red. The simulations are presented in real-time.