SHOULDER LOAD DURING WEIGHT RELIEF LIFTING: A SIMULATION STUDY

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INTRODUCTION
High and frequent loading during wheelchair ADL is a generally recognized factor contributing to the development of shoulder complaints. A previous study has shown that the external load on the shoulder in subjects with a spinal cord injury (SCI) is high (40 Nm) during weight relief lifting [1]. Simulation of internal load that includes the effect of (partial) muscle paralysis is likely to lead to higher glenohumeral contact forces and forces in the remaining muscles, when compared to a complete musculoskeletal system. The purpose of this study was to evaluate the effect of lesion level on the estimated glenohumeral contact force and muscle load.

METHODS
Four subjects with tetraplegia (TP) and four able-bodied (AB) male subjects participated. Three-dimensional kinematics of the thorax, humerus, clavicula, scapula, forearm and hand were recorded with a 3-camera opto-electronic system (Optotrak, Canada) during 3 trials of weight relief lifting. External forces were recorded with an instrumented wheelchair (AMTI 6df; Quickie Triumph, The Netherlands). The orientation of the scapula was determined in a calibration measurement with a scapula-locator system. From this measurement and the orientation of the humerus during the tasks, the orientation of the scapula and clavicula were calculated using a regression model [2]. Position and force data were used as input for the Delft Shoulder and Elbow Model which calculates muscle forces and joint glenohumeral contact forces (GHCF). 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 [3], based on the assumption that the maximum relative force of this study was to evaluate the effect of lesion level on the estimated glenohumeral contact force and muscle load.

RESULTS AND DISCUSSION
The peak GHCF (Figure 1) was higher for the TP profiles than for the AB profiles (P=0.037) and for the TP profiles the peak GHRF was significantly higher for the first successful simulation (S1) compared to T1 (P=0.029). For the T1 simulation, higher forces are calculated for the TP profiles in the serratus anterior, pectoralis major, deltoideus and in the rotator cuff compared to the AB profiles. However the muscle forces are not significantly higher for the TP profiles compared to the AB profiles (P=0.36). The calculated forces for the biceps and triceps show much more predicted force in the triceps for the AB profiles compared to the TP profiles.

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CONCLUSIONS
The higher GHCF found in our simulations was mainly due to a different task performance by the TP subjects. The model modifications had a minor effect on the calculated GHCF. 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.

REFERENCES