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 each muscle was relative to the number of innervating segments above the lesion. 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. All 24 input profiles (3 trials x 8 subjects) were used. This implied that the profiles of the AB subjects were used as input to a model with a SCI and vice-versa.

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.

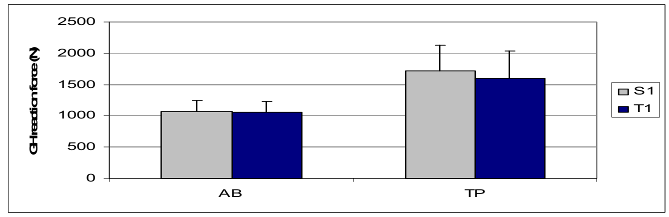

Figure 1: GHCF (mean + SD) for AB and TP profiles for the first successful (S1) and complete model simulation (T1).

Results show that it is possible to have a successful simulation of the performance of a weight relief lift with a complete C6 SCI but not with a C5 lesion, which appears indeed to be the case in real life. At the C6 lesion level, fewer successful AB profiles suggest an adaptation in the kinematics of the subjects with TP. The lesion modifications do have a small effect on the GHCF: T1 is 7.3 % higher than S1 for TP profiles. We expected the effect to be larger, but it is well likely that the adaptations in the TP kinematics already compensate for the loss of muscle force and the loss of muscle function by the use of different muscles.
A critical issue in this study is the question of whether the forces and particularly the forces of the triceps are correct. The predicted maximum triceps force for the TP profiles still 11 % of the maximum triceps force in the model. This percentage is close to the percentage measured by Needham-Shropshire [4]. 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. 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.

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

1. Van Drongelen S, et al. *Arch Phys Med Rehabil*, accepted.
2. Pascoal AG. *Ombro e Elevação do Braço*, Universidade tecnica de Lisboa, Lisboa.
3. Gray H. *Gray's Anatomy*, Churchill Livingstone, New York NY.
4. Needham-Shropshire BM, et al. *J Spinal Cord Med* **20,** 324-30, 1997.