INTRODUCTION

The majority of manual wheelchair users will experience upper extremity injuries and/or pain [1], which have been linked to the considerable physical demand placed on the upper extremity during wheelchair propulsion. As a result, much effort has been devoted to developing propulsion techniques that reduce upper extremity demand. However, the functional roles of individual muscles and how they work in synergy to satisfy the mechanical energetic demands of wheelchair propulsion are poorly understood. Understanding muscle function is a prerequisite for identifying how different propulsion techniques influence upper extremity demand and potential injury mechanisms.

Previous wheelchair propulsion studies have used EMG analyses to examine muscle function, which classified muscles broadly as either having push or recovery functions [e.g., 2]. Others have used joint power analyses to examine function based on net joint power [3]. However, how individual muscles contribute to the power flow that generates wheelchair propulsion is not well understood. Therefore, the purpose of this study was to use a forward dynamics simulation of the push phase of wheelchair propulsion to identify muscle function based on how individual muscles generate, absorb and/or transfer mechanical power to overcome the external workload.

METHODS

An upper extremity musculoskeletal model and forward dynamics simulation of wheelchair propulsion was used within a dynamic optimization framework that identified the muscle excitation patterns required to produce a simulation of normal wheelchair mechanics. The cost function used in the optimization minimized the difference between the simulation and experimentally collected push phase joint kinematics and handrim forces of a representative wheelchair user.

The upper extremity model was developed based on the work of Holzbaur et al. [4] and consisted of six rotational degrees of freedom representing trunk, shoulder, elbow and forearm articulations. Trunk lean and hand translations were prescribed over the entire stroke based on experimental kinematic data. Twenty-six Hill-type musculotendon actuators governed by intrinsic muscle force-length-velocity relationships were used to drive the model. Muscle excitation-activation dynamics were modeled using a first order differential equation with muscle specific activation and deactivation time constants.

Following the optimization, individual muscle contributions to the mechanical power of each body segment and the external handrim power were determined using a segment power analysis [5]. Individual body segments were combined into a single group representing the upper body.

RESULTS AND DISCUSSION

The optimization framework successfully identified muscle excitation patterns that reproduced the experimental data, with average joint kinematic and handrim differences of 0.6 degrees and 2.7 N, respectively, which were within one standard deviation of the experimental data. The segment power analysis showed that the proximal muscles (e.g. shoulder flexors) generated most of the mechanical power during the push phase, with the anterior deltoid (ADELT) being the primary contributor (Fig. 1A). During the first half of the push phase, ADELT provided similar amounts of power to the body and handrim (Fig. 1A; dotted and
Figure 1: Muscle contributions to body and handrim power during the push phase. Note the change in scale with ADELT. Total is the total musculotendon power, which is the sum of the Body and Handrim power.

dashed lines equal). During the second half, ADELT delivered most of its power directly to the handrim (Fig. 1A; dashed line). The functional role of the proximal muscles as the primary power generators is consistent with a previous study showing the net shoulder flexion moment increases the mechanical energy of the arm and hand/handrim during the push phase [3].

The brachioradialis and brachialis muscles (BRA group), which are elbow flexors that do not cross the shoulder joint, were active over a large portion of the push phase. These muscles provided power to the handrim during the first half of the push phase (Fig. 1B; dashed line positive), but then absorbed power from the handrim in the second half (Fig. 1B; dashed line negative). The biceps brachii (BIC), elbow flexors that also cross the shoulder, acted to transfer power from the arm (Fig. 1C; dotted line negative) to the handrim (Fig. 1C; dashed line positive) as well as generate power directly to the handrim (Fig. 1C; solid line positive).

The most distal upper extremity muscles (e.g., pronator teres and pronator quadratus, PT Group) did not generate or absorb power from the handrim or body for most of the push phase (Fig. 1D; solid line near 0). Instead, these muscles acted primarily to transfer power from the handrim to the body (Fig. 1D; dashed line negative, dotted line positive). Similar energy transfer mechanisms have been shown in other movement tasks such as pedaling and walking [e.g., 6].

CONCLUSIONS

Identifying individual muscle function during wheelchair propulsion is challenging due to the multi-articular and multi-muscle nature of the upper extremity. However, the forward dynamics simulation and segment power analysis framework used in this study is ideally suited to quantify muscle contributions to the mechanical energetics of wheelchair propulsion and identify muscle function.

ADELT was found to be the primary contributor to mechanical power during the push phase. This muscle also acts to elevate the humeral head in the glenohumeral joint, which requires high activity from antagonistic muscles to stabilize the joint. This may explain why overuse injuries (e.g., in the rotator cuff muscles) and pain (e.g., due to glenohumeral joint inflammation) at the shoulder are common among wheelchair users. The elbow flexors (BIC and BRA Group) generated power directly to the handrim as well as transferred power from the arm to the handrim. Although these muscles provide almost as much energy as ADELT early in the push phase, the prevalence of elbow injuries is much lower. This may be due to the lower muscle co-contraction required to stabilize the elbow joint.

REFERENCES


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