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Modulation of shoulder muscle and joint function using a powered upper-limb exoskeleton

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ABSTRACT

Robotic-assistive exoskeletons can enable frequent repetitive movements without the presence of a full-time therapist; however, human-machine interaction and the capacity of powered exoskeletons to attenuate shoulder muscle and joint loading is poorly understood. This study aimed to quantify shoulder muscle and joint force during assisted activities of daily living using a powered robotic upper limb exoskeleton (ArmeoPower, Hocoma). Six healthy male subjects performed abduction, flexion, horizontal flexion, reaching and nose touching activities. These tasks were repeated under two conditions: (i) the exoskeleton compensating only for its own weight, and (ii) the exoskeleton providing full upper limb gravity compensation (i.e., weightlessness). Muscle EMG, joint kinematics and joint torques were simultaneously recorded, and shoulder muscle and joint forces calculated using personalized musculoskeletal models of each subject's upper limb. The exoskeleton reduced peak joint torques, muscle forces and joint loading by up to 74.8% (0.113 Nm/kg), 88.8% (5.8%BW) and 68.4% (75.6%BW), respectively, with the degree of load attenuation strongly task dependent. The peak compressive, anterior and superior glenohumeral joint force during assisted nose touching was 36.4% (24.6%BW), 72.4% (13.1%BW) and 85.0% (17.2%BW) lower than that during unassisted nose touching, respectively. The present study showed that upper limb weight compensation using an assistive exoskeleton may increase glenohumeral joint stability, since deltoid muscle force, which is the primary contributor to superior glenohumeral joint shear, is attenuated; however, prominent exoskeleton interaction moments are required to position and control the upper limb in space, even under full gravity compensation conditions. The modeling framework and results may be useful in planning targeted upper limb robotic rehabilitation tasks.

1. Introduction

Intensive task-directed therapy and repetitive motion of an affected limb immediately following traumatic brain injury or a stroke event is performed to exploit the brain's neuroplasticity, and has been shown to contribute to reduced motor impairment and improved functional performance (Teasell and Kalra, 2004). While rehabilitation programs can be time-consuming and labor-intensive for the patient (Lo and Xie, 2012), powered exoskeletons enable more frequent repetitive movements without the presence of a full-time therapist, thereby reducing the time and cost burden of the rehabilitation (Huang and Krakauer, 2009).

Exoskeleton-based rehabilitation programs that have been shown to improve shoulder and elbow joint motion in cases of neuromotor impairment have employed progressive-resistive therapy, upper limb weight compensation and task trajectory guidance (Crocher et al., 2012; Fasoli et al., 2003; Frisoli et al., 2012); however, some studies have shown no difference in upper limb motor improvement with powered exoskeleton-based therapy compared to manual therapy, suggesting that further development of robot-control strategies may be required to create more targeted and purposeful limb assistance (Crocher et al., 2012; Patton et al., 2008). Gravity compensation during assisted motion is clinically significant in upper-limb rehabilitation, as it decreases the magnitude of joint torques required to generate movement (Platz et al., 2001). Coscia et al. (2014) showed that using a sling to unload the upper limb during reaching resulted in a decrease in muscle EMG of up to 50%, with preservation of muscle synergy patterns (Coscia et al., 2014); however, the way in which shoulder muscle and joint loading are modulated during movements undertaken using powered assistive devices is not well understood.

Glenohumeral joint function during upper limb motion is achieved by the activation of the surrounding musculature and

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resultant compression of the humeral head into the glenoid fossa (Labriola et al., 2005). The muscle and joint-contact loading patterns to achieve this are critical in upper limb function, since each muscle's force multiplied by its moment arm represents the muscle's net contribution to glenohumeral joint torque, while the resultant glenohumeral joint force magnitude and orientation is primarily governed by the vector sum of individual forces produced by the spanning muscles, and is a key determinate of joint stability. Unfortunately, the difficulty in directly measuring muscle and joint loading non-invasively in humans poses a major challenge in understanding neuromuscular control of upper limb movement and shoulder joint behavior. Inverse kinematics and inverse dynamics have been used to provide information on limb motion and net joint moments at the human shoulder joint (Anton et al., 2001); however, these approaches alone cannot provide quantitative muscle loading information.

Computational modeling is one of the only means available to non-invasively quantify muscle and joint loading during upper limb movement. The shoulder complex is a highly mechanically redundant system with many muscles spanning each joint, and a net joint torque can be produced by an infinite combination of muscle forces. Musculoskeletal models of the upper limb have been developed to solve this redundancy problem and investigate shoulder implant behavior (Masjedi and Johnson, 2010), wheelchair propulsion (Dubowsky et al., 2008), and plan reconstructive surgery (Magermans et al., 2004), but to our knowledge, they have not been used to investigate modulation of upper limb muscle and joint forces through human-machine interaction with assistive rehabilitation exoskeletons. This study aimed to develop a subject-specific musculoskeletal modeling framework to quantify the influence of a powered robotic assistive exoskeleton on shoulder muscle and joint function during movements with and without gravity compensation. We hypothesize that upper limb weightlessness will significantly attenuate the loading of the prime mover muscles and therefore the magnitude and direction of the glenohumeral joint force as well as glenohumeral joint stability.

2. Materials and methods

2.1. Subject recruitment

Six healthy male participants (age: 25–38 years old, body mass: 56–85 kg, height: 170–175 cm) were recruited for upper-limb motion experiments. This study was approved by the University of Melbourne Human Ethics Advisory Group, and all participants gave written informed consent prior to testing.

2.2. Upper limb motion experiments

Subjects were seated and asked to generate peak voluntary isokinetic and isometric contractions during coronal plane abduction and adduction; sagittal plane flexion and extension; and internal and external rotation at both 30° and 90° of humeral elevation in the coronal plane using a dynamometer (Biodex Pro 4, Shirley, NY), following a previously reported protocol (Wu et al., 2016). Activities of daily living were then performed, including abduction and adduction; flexion and extension; horizontal flexion and extension; reaching; and nose touching. Each task was executed over a duration of 2 s (Table 1). During the tasks, trajectories of three-dimensional marker trajectories attached to the upper limb were measured using an 8-camera, video-based motion capture system (Vicon, OxfordMetricsLtd., Oxford), and used to calculate upper limb kinematics. Markers were attached to bony landmarks recommended by the ISB (Wu et al., 2005), with additional markers attached to the acromion process, olecranon process, middle of the dorsal side of wrist, and the inferior aspect of the 11th rib.

A customized 3D-printed marker cluster was attached to the scapular spine and used to track scapular motion (van Andel et al., 2009). A two-stage calibration method was subsequently used to minimize scapula skin motion artefact (Brochard et al., 2011). Briefly, an adjustable scapula locator was used to palpate and digitize the locations of the trigonum spinae, angulus acromialis and angulus inferior at their initial and final positions for each upper limb motion task, while the positions of the 3D-printed marker cluster were simultaneously recorded. For subsequent trials, the relative locations of the scapular bony landmarks, and thus the scapula position and orientation, were calculated from the position of the 3D-printed marker cluster alone. Marker trajectories were sampled at 200 Hz and filtered using a fourth-order, low-pass Butterworth filter with a 10 Hz cut-off frequency. The electromyographic (EMG) data of eleven major muscle subregions, including trapezius (upper and lower), deltoid (anterior, middle, and posterior), infraspinatus, pectoralis major (upper and lower) and middle latissimus dorsi were recorded simultaneously using pre-amplified EMG surface electrodes (Cometa, Bareggio, Italy) at a sample rate of 2000 Hz (Dickerson et al., 2008). The EMG data were high-pass filtered using a second order 25 Hz Butterworth filter, rectified and low-pass filtered using a second order 2 Hz Butterworth filter (Nikooyan et al., 2012). All EMG data were normalized to peak values obtained from a series of maximum voluntary isometric contractions (Halaki and Ginn, 2012).

Subjects were then fitted to a robotic upper-limb exoskeleton (ArmeoPower, Hocoma, Switzerland) while seated. The 6-degree-of-freedom exoskeleton included 3-degrees-of-freedom of shoulder joint motion (i.e. elevation-depression, plane of elevation and axial rotation), 2-degrees-of-freedom of elbow joint motion (i.e. flexion-extension and pronation-supination) and 1-degree-of-freedom of wrist joint motion (i.e. flexion-extension). Exoskeleton joint centers were manually aligned to corresponding anatomical joints on each subject by adjusting the height and lengths of the robotic limb segments. Comfortable and unobstructed movement of the upper limb through its range of motion served to confirm appropriate exoskeleton joint center alignments. Each activity of

Table 1
Descriptions of upper-limb tasks for abduction, flexion, horizontal flexion, reaching and nose touching. The tasks of adduction, extension and horizontal extension were kinematically equivalent to the tasks of abduction, flexion and horizontal flexion, respectively, performed in reverse.

	Start position	End position
Abduction	Humerus abducted to 45° in the coronal plane; elbow fully extended; palm facing down	Humerus abducted to 100° in the coronal plane; elbow fully extended; palm facing down
Flexion	Humerus flexed to 45° in the sagittal plane; elbow fully extended; palm facing inward	Humerus flexed to 100° in the sagittal plane; elbow fully extended; palm facing inward
Horizontal flexion	Humerus abducted to 90° in the coronal plane; elbow fully extended; palm facing down	Humerus flexed to 90° in the sagittal plane; elbow fully extended; palm facing down
Reaching	Humerus abducted to 45° in the coronal plane; elbow fully extended; palm facing down	Humerus flexed to 90° in the sagittal plane; elbow fully extended; palm facing inward
Nose touching	Humerus abducted to 45° in the coronal plane; elbow fully extended; palm facing down	Thumb touching the nose