MUSCLE RECRUITMENT PATTERNS IN PERTURBED SITTING

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INTRODUCTION

Trunk instability is a major problem for people with spinal cord injury (SCI). Most spinal cord lesions situated above the first lumbar vertebra will cause full or partial paralysis of the lumbar musculature. As a result, the lumbar muscles cannot produce sufficient forces to stabilize the lumbar spine, which is inherently unstable [1]. Often, individuals with SCI employ one arm around their wheelchairs to prevent falling, making bimanual tasks difficult to perform.

Functional electrical stimulation (FES) uses bursts of short electric pulses to generate muscle contraction [2, 3]. By stimulating a specific set of muscles and properly sequencing the stimulation, FES can generate body functions such as grasping, standing and walking. The use of FES to increase trunk stability and improve sitting function remains a largely unexplored topic in biomedical engineering. It has been shown that continuous, bilateral stimulation of the lumbar erector spinae, using surface stimulation, can increase sitting stability in complete paraplegia [4]. This suggested that FES could be used effectively for maintaining trunk stability during sitting when applied to specific muscles of the trunk, which would greatly benefit individuals with SCI. To develop such a system however, a fundamental knowledge of the biomechanics of the trunk must be acquired.

The direction dependencies of trunk muscle responses have been poorly reported, since most studies applied perturbations from a single direction. Only a few studies applied perturbations from multiple directions [5-8]. The results of these studies provided general descriptions of muscle responses but specific details, such as amount of muscle response, were not reported. The purpose of this study was to determine the direction dependency of trunk muscle responses and the amount of muscle activities to perturbations applied from multiple directions during sitting of able-bodied subjects. The direction dependency of trunk muscle responses could then be formulated and implemented with FES to increase trunk stability of individuals with SCI.

METHODS

Subjects

Twelve healthy, right-handed male adults (age 21-39 years; height 178.0 ± 4.7 cm; body mass 70.3 ± 10.0 kg) participated in this study. They had no medical history of neurological disorders. All subjects gave informed consent to participate in the study. The experimental procedures used in this study were approved by the local ethics committee.

Measurements

Sixteen channels of electromyogram (EMG) were recorded from trunk and neck muscles bilaterally using disposable silver-silver chloride surface electrodes with a diameter of 10 mm and a fixed interdistance of 18mm. After careful abrasion of the skin, the electrodes were placed longitudinally over the muscles. The location of each trunk electrode was selected according to [9, 10]: rectus abdominis (RA) was located 3 cm lateral to the umbilicus, aligned vertically; external obliques (EO) was located 15 cm lateral to the umbilicus, aligned 45º to the vertical; internal obliques (IO) was located on the midpoint between ASIS and symphasis pubis, above the inguinal ligament, aligned 45º to the vertical; thoracic erector spinae (T9) was located 5 cm lateral to the T9 spinous process, aligned vertically; lumbar erector spinae (L3) was located 3 cm lateral to the L3 spinous process, aligned vertically; latissimus dorsi (LD) was lateral to T9 spinous process, over the muscle belly; sternocleidomastoid (SM) was 1/3 the distance from the sternal notch to the mastoid process at the distal end overlying the mucle belly; and splenius capitis (SC) was over the C4-C5 level, aligned vertically. The reference electrode was placed over the clavicle.

External force perturbations were applied to the subjects at chest levels by an experimenter manually pulling a rope in series with a force transducer (MLP-100-CO-C, Transducer Techniques, Temecula, USA: Amplifier; Model 9243, Burster, Germany) to a custom harness. The resulting perturbation force profile resembled an impulse function.

All EMG signals and the force transducer signal were collected using a 64-channel, 12-bit analog-to-digital converter (NI 6071E, National Instrument, Austin, USA) at a sampling frequency of 2000 Hz.

Protocol

Resting EMG values were recorded while the subjects laid down supine on a bench. Then, 3 sets of maximum voluntary isometric contraction (MVC) exercises were performed [11, 9]. The exercises were, for the abdominal muscles, 1) sit-up, 2) lateral bend to the left, 3) lateral bend to the right; for the back muscles, 4) back extension; for the neck muscles, 5) neck flexion, 6) neck extension, 7) neck left flexion, 8) neck right flexion. In each exercise, the subjects were manually braced by a research assistant.

Subjects were instructed to cross their arms in front of their chest, to close their eyes, and to sit relaxed and naturally on a custom seating apparatus. A total of 40 perturbation trials (8 directions, 5 trials each) were applied to the subjects. The perturbation directions were labeled 1 to 8, with direction 1 corresponding to the anterior direction, and incrementing clockwise by 45º. The perturbation trials were given in random order. The subjects wore a headphone and listened to asynchronous whale music and nature sounds found in national parks. In addition, the subjects counted numbers aloud to prevent anticipation. During the perturbation trials, two researchers took up the slack in the ropes of two different directions. One direction was the intended pulling direction, where the force transducer was attached and the other direction served as a decoy to prevent subjects from anticipating the pulling direction. To maintain consistency, all external perturbations were pulled by one researcher. Breaks were given after every 10 trials.

Analysis

The force transducer data was filtered using a 10 Hz, 4th-order, zero-phase lag, low-pass Butterworth filter. The onset of perturbation was defined as the time when the first derivative of the force transducer signal exceeded 12N/s. A time window of 1.0 s before and 3.0 s after the onset of perturbation was selected for subsequent analyses for all signals. EMG signals were rectified, then low pass filtered with a cutoff frequency of 2.5 Hz using the fourth-ordered, zero-phase-lag Butterworth filter [12, 9, 10]. The filtered EMG was averaged among 5 trials for each direction for each muscle. Then the averaged EMG was normalized using the resting and maximum values, i.e., *normalized EMG*=(*EMG-resting value*)/(*MVC value-resting value*) x 100.

The mean amplitude of EMG 0.25 s before the onset of the perturbation was considered the tonic activity for each muscle. The tonic activity was subtracted from the corresponding muscle activity to determine the phasic response of each muscle. The peak value within 0.5 s after the onset of the perturbation was quantified. Then, the direction dependency of the peak values of the phasic response for each muscle was formulated. The relationship between the peak value of the phasic response and the perturbation angle was represented by a fitted normal distribution function. The fitted function was

$$y = c_0 + c_1 \exp^{(-((x-c_2)/c_3)^2))} \tag{1}$$

where y is the phasic response, x is the perturbation angle, and $c_{0,1,2,3}$ are coefficients.

RESULTS

Only the EMGs from the right side were analyzed because symmetry in EMG response was found between the left and right sides of the body. EMGs from LD were not analyzed. Fig. 1 shows the group ensemble average of the phasic responses and of the load, to illustrate the tendency of the direction dependency of the phasic response. Clear direction dependencies could be seen for RA, EO, IO, T9 and L3. The response amplitudes of SM and SC were about the same in all directions, which suggests that those muscles do not behave with a direction dependency. This observation was verified by one-way repeated measures analysis of variance (ANOVA). The muscle responses among directions were significantly different for most of the muscles except the SM and SC. Thus, the formulation was performed except for these two muscles.

Fig. 2 shows the curve fit results for each muscle, and Table 1 shows the coefficients of the Gaussian formula. All curve fits were statistically significant, as shown by the coefficients of determinations (R in Table 1). The direction at the peak of the Gaussian distribution and the width of the Gaussian distribution were represented as the values of c_3 and c_4 in Table 1, respectively.

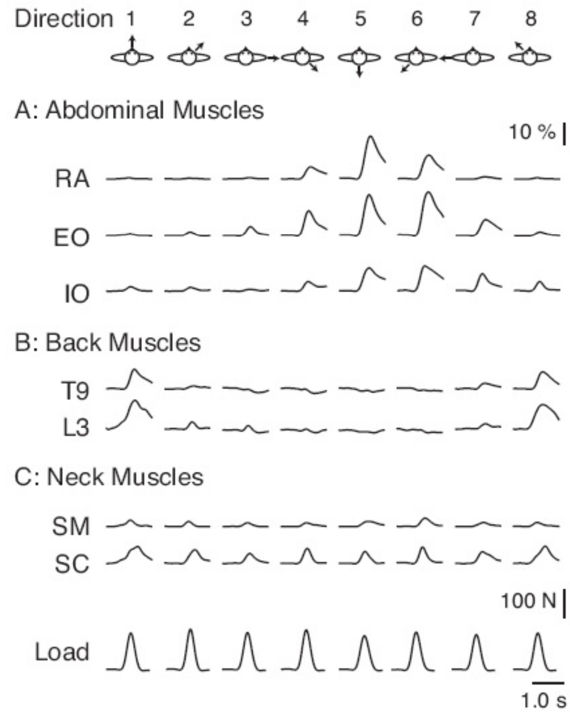

Figure 1: Group ensemble average of the phasic response for each muscle and direction

Table 1: Coefficients for the direction dependency formulas (average value)

Muscles	c_0	c_1	c_2	C_3	*R*
RA	0.498	19.6	188	45.9	0.999
EO	0.815	20.1	197	75.8	0.991
IO	2.01	10.9	217	71.6	0.990
T9	0.669	9.09	339	52.8	0.994
L3	1.10	15.3	337	49.4	0.989

DISCUSSION

In this study, perturbations were applied from 8 different directions to the torso of able-bodied subjects during sitting to investigate the direction dependency of the phasic responses in trunk and neck muscles. Clear direction dependencies were observed in RA, EO, IO, T9 and L3. The neck muscles responded equally to all directions. The direction dependencies for the trunk muscles were successfully formulated using a Gaussian distribution.

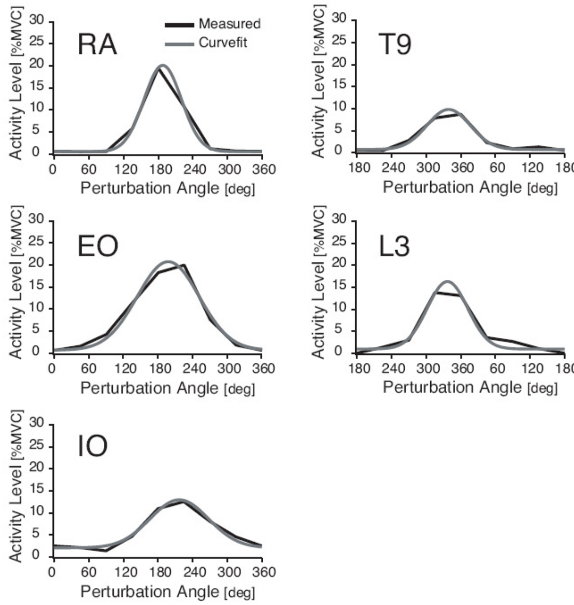

Figure 2: Results of the curve fitting for abdominal and back muscles

The present study was the first to formulate the muscle responses to approximately equal loads from different perturbation directions. Using the formula, the direction where each muscle was maximally activated could be identified. The formula was also sensitive enough to quantify the range of the activation for each muscle.

CONCLUSION

In conclusion, clear direction dependencies were found in RA, EO, IO, T9 and L3 to a sudden, transient horizontal load at the chest, while the neck muscles responded equally to all directions. The direction dependency of each muscle was successfully formulated using a Gaussian distribution. These formulas could be used in implementing a FES system for trunk muscles to stabilize sitting posture for people with spinal cord injury.

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