Title: Postural Reactions of the Trunk Muscles to Multidirectional Perturbations in Sitting

Authors: Kei Masani[1,2], Vivian W. Sin[1], Albert H. Vette[1], T. Adam Thrasher[3], Noritaka Kawashima[1,4], Alan Morris[2], Richard Preuss[1,2], and Milos R. Popovic[1,2]

Affiliations:
[1] Rehabilitation Engineering Laboratory,
Institute of Biomaterials and Biomedical Engineering, University of Toronto
164 College Street, Toronto, Ontario, M5S 3G9, Canada
[2] Rehabilitation Engineering Laboratory,
Lyndhurst Centre, Toronto Rehabilitation Institute
520 Sutherland Drive, Toronto, Ontario,M4G 3V9, Canada
[3] Health & Human Performance, University of Houston,
3855 Holman Street, Garrison Rm 104, Houston, TX, 77204-6015, USA
[4] Japan Society for the Promotion of Science,
6 Ichibancho, Chiyoda-ku, Tokyo 102-8471, Japan

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Correspondence to:
Kei Masani PhD
Rehabilitation Engineering Laboratory, Lyndhurst Centre,
Toronto Rehabilitation Institute,
520 Sutherland Drive, Toronto, ON, M4G 3V9, Canada
Phone: +1-416-597-3422 ext 6213, Fax: +1-416-425-9923
E-mail: k.masani@utoronto.ca

Abstract

Background: The dynamic role of the trunk musculature, with respect to stability, has not been fully explored to date. The purpose of this study was, using a transient and multi-directional perturbation, to: 1) quantify the tonic level of activity in the superficial trunk musculature prior to any perturbation; 2) quantify the phasic activity in those same muscles following application of a transient, horizontally directed load; and 3) quantify the direction-dependent behavior of this phasic response.

Methods: Twelve healthy individuals were perturbed during sitting via a chest harness in eight horizontal directions. Surface electromyograms were measured bilaterally from the abdominal (rectus abdominis, internal and external obliques) and back musculature (thoracic and lumbar erector spinae) to determine the tonic muscle activity prior to perturbation, and the phasic response following perturbation. A descriptive model was used to characterize the relationship between the phasic response of the muscles due to perturbation and the pulling direction.

Findings: Tonic activity in the trunk musculature in upright sitting is low, but still above resting levels by at about 1-3 % of the MVC for the abdominal muscles, and 4-6 % for the back muscles. Each trunk muscle also showed a direction-specific, phasic activation in response to perturbation, above these tonic levels of activation. This phasic activation was accurately modeled using a descriptive model for each muscle.

Interpretation: The obtained muscle activation level and the identified descriptive model will be applied in the design of a closed-loop controller for Functional Electrical Stimulation.

Key Words: Human, Sitting, Balance, Motor Control

Introduction

The instability of the trunk is a major problem for people with spinal cord injuries (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 trunk stability. Trunk stability has received a great deal of attention in both the clinical and scientific literature. Generally, this research has had two primary foci: inter-vertebral instability in mechanical back pain (Cholewicki and McGill, 1996); and postural instability in neurological conditions such as stroke (Verheyden et al., 2007) and spinal cord injury (Seelen et al., 1997). The current study fits in this second group, and represents a preliminary step in the development of a controller for a functional electrical stimulation (FES) system to support a sitting posture following a lesion of the spinal cord above the first lumbar level. This work specifically focuses on the role of trunk musculature in maintaining an upright sitting posture following a transient horizontal perturbation.

Tonic muscle co-activation has been shown to provide postural stability to the trunk, often without the need for any additional perturbation-specific phasic response (Stokes et al., 2000). Under most conditions, however, excessive co-activation may be viewed as inappropriate, as it does not leave the trunk free to move or bend, and may lead to rapid muscle fatigue (Hunter et al., 2004). Furthermore, the resulting decrease in trunk compliance may necessitate an increase in the forces required to adequately stabilize the pelvis/hips complex in sitting (which will be addressed in future work). A more desirable strategy, therefore, may be one in which low levels of spine stiffness, sufficient only to prevent immediate spine buckling, are maintained by tonic activation of the musculature, with phasic, feedback-driven muscle activation occurring in response to perturbations (Preuss et al., 2005).

To date, the phasic electromyographic (EMG) responses of various trunk muscles have been investigated in both sitting and standing, using a variety of loading and perturbation techniques (Brown et al., 2003, Cresswell et al., 1994, Mannion et al., 2000, McMulkin et al., 1998, Preuss and Fung, 2007, Radebold et al., 2000, Stokes et al., 2006, Stokes et al., 2000, Thomas et al., 1998, Vera-Garcia et al., 2007, Zedka et al., 1998). One common finding is that the trunk muscles tend to respond differently depending on the direction in which they are loaded. This direction-dependent behavior, however, has yet to be quantified to the extent required to implement a controller capable of mimicking this physiological action. This quantitative analysis is particularly important, given that the anatomy of the trunk musculature suggests that these muscles may act across a wide range of perturbation directions (Stokes and Gardner-Morse, 1999).

In order to develop a controller for an FES system for the superficial trunk muscles, the tonic activity in quiet sitting must be quantified, along with the phasic response for multiple directions of perturbation. The purpose of this study was to: 1) quantify the tonic level of activity in the superficial trunk musculature prior to any perturbation; 2) quantify the phasic activity in those same muscles following application of a transient, horizontally directed load; and 3) quantify and model the direction-dependent behavior of this phasic response.

Methods

Subjects

Twelve healthy male adults (age: 21-39 years; height: 178.0 (SD 4.7) cm; body mass: 70.3 (SD 10.0) kg) were recruited from university students and postdoctoral fellows. Subjects were excluded if they had any prior diagnosis of neurological disease, injury or impairment; any prior diagnosis of spinal scoliosis or other condition affecting spine posture; and/or any prior history of acute or chronic back pain. All subjects gave informed consent prior to participation. The experimental procedures used in this study were approved by the local ethics committee.

Measurements

Surface EMG was acquired using two AMT-8 EMG systems (Bortec Biomedical Ltd., Calgary, Canada) with a total amplifier gain of 2000 (including pre-amplification), a frequency response of 10 to 1000 Hz, and a common mode rejection ratio of 115 dB (at 60 Hz). Disposable silver-silver chloride surface electrodes (10mm diameter) were placed in a bipolar configuration (18mm fixed inter-electrode distance) over the bilateral rectus abdominis (RA) (vertical alignment, 3cm lateral to the umbilicus), external oblique (EO) (aligned 45 deg to vertical, 15cm lateral to the umbilicus), internal oblique (IO) (aligned 45 deg to vertical at the midpoint between the anterior superior iliac spine and the symphysis pubis, above the inguinal ligament), thoracic erector spinae (T9) (vertical alignment, 5cm lateral to the T9 spinous process), and lumbar erector spinae (L3) (vertical alignment, 3cm lateral to the L3 spinous process). A reference electrode was placed over the clavicle.

Perturbations were applied via a manual pull, using a rope attached to a body harness surrounding the subject's chest, below the axilla. Tensile force in the rope was recorded by an in-series force transducer (Sensor: MLP-100-CO-C, Transducer Techniques, Temecula, USA: Amplifier; Model 9243, Burster, Germany) with a range of Â±444.8 N, and an output from 0 V to 10 V. The resulting perturbation force profile resembled an impulse function.

All data were synchronously acquired, at a sampling frequency of 2000 Hz, using a 64-channel, 12-bit analog-to-digital converter (NI 6071E, National Instrument, Austin, USA) and custom-designed data acquisition software embedded in LabView (National Instruments, Austin, USA).

Protocol

Resting EMG levels were recorded for 30s with the subjects supine, fully relaxed, eyes closed and without speaking. Maximum voluntary isometric contractions (MVC) were then performed according to (Davidson and Hubley-Kozey, 2005, McGill et al., 2006). MVC for the abdominal muscles were performed in a bent-knee sit-up position, with the trunk flexed to ~45 deg, and the arms placed across the chest. The subjects were asked to produce maximal isometric efforts in trunk flexion, left lateral bend and right lateral bend, with trunk movement restrained by one researcher, and the feet secured by a second.

MVC for the erector spinae were performed with the subjects prone, with their hips at the edge of the test bench and their arms placed across the chest. The subjects were asked to produce a maximal isometric effort in trunk extension against symmetrical manual resistance provided to the upper thoracic region by one researcher, with the legs secured by a second. Each MVC effort was held for approximately 3s, and was repeated 3 times.

Perturbation trials began only after a sufficient period of rest (based on the subjects' feedback), in order to minimize any potential effect of muscle fatigue. The experimental set-up is illustrated in Fig. 1A. Subjects were seated with their eyes closed, the lower legs unsupported and the arms crossed over the chest. Instructions were to "...sit up tall, and imagine a string attached to the top of your head pulling you up..." and to "...try and keep your neck and shoulders relaxed." To minimize the effect of auditory cues, subjects wore headphones, playing asynchronous nature sounds, and were asked to count out loud at various prescribed intervals. The pelvis was surrounded by a wooden restraint, which provided a margin of safety, but did not contact or constrain the pelvis.

Each subject experienced a total of 40 perturbation trials (8 directions, 5 trials each) randomly ordered, with a rest period provided every 10 trials. Perturbations were delivered manually, via ropes attached to a harness placed around the subject's chest, just inferior to the axilla. In order to minimize any anticipatory postural adjustments, ropes were cocked by two researchers in both the intended pulling direction (where the force transducer was attached) and in a second direction randomly selected among the remaining seven directions (Fig. 1A). Tension in the latter was reduced to zero when the perturbation was initiated. All perturbations were performed by the same researcher. The subjects confirmed post-testing that they were not able to anticipate the direction of the pull.

Data Analysis

Perturbation Force

Prior to analysis, force transducer data was low-pass filtered at a cut-off frequency of 10Hz, using a 4th-order, zero-phase lag Butterworth filter. Onset of perturbation was determined from the first derivative of this signal, as the point closest to the peak value which first exceeded 12 N/s. This threshold was selected based on data collected in preliminary studies. A time window of 1.0 s before and 3.0 s after perturbation onset was selected for subsequent analyses. For each subject, the peak force over this period was averaged over the 5 trials for each perturbation direction. A one-way repeated-measures ANOVA, with the peak perturbation force as the dependent variable, and the direction of perturbation as the independent variable, was performed to determine the consistency of the perturbation force across directions.

<u>EMG</u>

Prior to analysis, all acquired EMG signals were rectified, and low-pass filtered at a cutoff frequency of 2.5 Hz (Vera-Garcia et al., 2006) using a 4th-order, zero-phase-lag Butterworth filter. Resting EMG levels and MVC values were taken as the mean amplitude of the signals acquired during these procedures, with the maximum value from the MVC trials, for each muscle, used for normalization purposes.

For each subject, the pre-processed EMG for each muscle was averaged across the 5 trials in each perturbation direction. These averaged values were then normalized according to the following procedure (Dewald et al., 1995, Vasavada et al., 2002):

$$\textit{normalized EMG}=(\textit{EMG-resting value})/(\textit{MVC value-resting value})\times 100$$

The mean EMG amplitude, over the 0.25s window immediately preceding perturbation onset, was initially used to describe the tonic activity of each muscle, for each upcoming perturbation direction. A one-way repeated-measures ANOVA was then run with these values as the dependent variable, and the upcoming perturbation direction as the independent variable, in order to confirm that no direction-specific anticipatory effects were present. As no significant main effect of direction was found (p=0.093 to 0.662), the tonic activity for each muscle was taken as the mean of these values across all perturbations directions. This tonic activity, for each subject and each muscle, was then subtracted from the corresponding EMG signals, for all directions of perturbation. The phasic response for each muscle was then determined as the peak value of the remaining signal over the 0.5s window immediately following the onset of the perturbation.

Prior to further statistical analysis, the perturbation directions for the left-sided muscles were named for the angle of pull in a clockwise direction (with 0 as the anterior direction of pull), and the directions for the right-sided muscles were similarly named, but in a counter-clockwise direction (Fig. 3, inset). This was done to allow for direct comparison of the symmetry of the response for the bilateral musculature.

A two-way repeated-measures ANOVA was then performed on the tonic trunk muscle activity, with activation level as the dependent variable, and the muscle and side (left vs. right) as the independent variables.

Three-way repeated-measures ANOVA was performed on the phasic response of the trunk musculature, with activation level as the dependent variable, and the direction of perturbation, the muscle and the side of the muscle (left vs. right) as the independent variables.

Curve-fitting

The relationship between the phasic response and the perturbation direction for each muscle, for each subject, was modeled using the Gaussian function:

$$y = a + b\exp^{(-((x-c)/d)^2))}, \tag{1}$$

where y is the phasic response, x is the angle of the pulling direction, and a, b, c, d are the coefficients of the function. The interpolation between data points provided by this modeling approach was used to provide a more accurate, quantitative estimate of the direction-specificity of the observed phasic response. The specific coefficients of interest were 'c', which represents the direction in which a maximum phasic response would be expected, and 'd', which represents the estimated width of the Gaussian function at half-maximum amplitude (i.e. the range of perturbation directions for which muscle activation will be at least half of the value at '*c*'). This function was chosen based on a preliminary visual examination of the data, and on previously reported direction-specific behavior in the trunk musculature (e.g. Preuss and Fung, 2007). The curve-fitting was executed using a Levenberg-Marquardt algorithm to search for the coefficient values that minimize χ-square (Kaleidagraph ver. 3.6.4, Synergy Software, USA). For the left and right-sided muscles, the directions of perturbation were named as described above (Fig. 3, inset).

Two-way repeated measures ANOVA were performed for the parameters '*c*' and '*d*', using only the data for which a statistically significant fit was obtained using the curve-fitting approach described above. For these analyses, the independent variables were muscle and side of the body (left and right).

All of the statistical analyses described above were performed in Statsview (version 5, SAS Institute Inc., USA). Post hoc analyses, where applicable, were carried out using a Bonferroni-Dunn test. Analyses were performed using an α-level of 0.05 for all tests.

Results

Missing Data

Surface EMG data from certain muscles (LL3 for 3 subjects, LT9 for one subject) showed inexplicably high levels of tonic activation (>100 %MVC). For the subjects in question, these muscles, along with their right-sided counterparts, were excluded from subsequent analysis.

Perturbation Force

The peak perturbation force was significantly different between directions of perturbation ($p<0.0001$), with group means ranging from 131 N to 148 N (Fig. 1B). Post-hoc analysis found that forces were often significantly higher in the diagonal directions than in the non-diagonal (in-plane) directions (Fig. 1B). The intra-subject variation among directions was 7.8±2.2 % (coefficient of variation) on average, while the intra-direction variation among subjects was 21.5±1.9 %.

Tonic Activity

Tonic EMG levels were significantly different between trunk muscles (main effect of muscle: $p<0.0001$), but generally symmetrical (no main effect of side: $p=0.825$; and no interaction effect between muscle and side: $p=0.929$) (Fig. 1C). The differences between trunk muscles were roughly explicable by differences between the abdominal muscles (1 % to 3 %MVC) and those of the back (4 % to 6 %MVC). Post-hoc analyses found bilateral tonic levels for L3 to be higher than for all of the abdominal muscles, while those for T9 were higher than for EO alone.

Phasic Response

Fig. 2 illustrates the group ensemble average traces (all subjects) of the phasic response to perturbation, along with the pulling forces (bottom row) for each direction.

The phasic response for the trunk muscles was highly direction dependent (main effect of direction: $p<0.0001$) and symmetrical (no main effect of side: $p=0.563$). Significant differences were also found in the phasic activation levels of the trunk muscles tested (main effect of muscle: $p<0.0001$). The phasic response for all muscles and directions is illustrated in Fig. 3.

Curve-fitting

The results of the curve fitting were statistically significant for most subjects and muscles, with certain notable exceptions: LT9 for one subject, LL3 for three subjects, and RIO for one subject ($r<0.632, df=8$). Additionally, there were obvious outliers in two cases: *b* of REO for one subject and *b* of RL3 for one subject. These muscles were eliminated from the descriptive analysis and from the ANOVA tests. Taking into account the missing data addressed above, the total number of subjects for the subsequent results were 6 for the left-sided muscles and 7 for the right-sided muscles. For the remaining cases, the correlation coefficient for the curve-fitting was strongly significant (p<0.0001 for all). These results are presented in Table 1. The results of the curve-fitting for a representative subject are shown in Fig. 4.

<u>Direction of the Maximum Phasic Response (coefficient '*c*')</u>

For these interpolated results, the direction of the maximum phasic response (c) was significantly different between muscles (main effect of muscle: p<0.0001), but was generally symmetrical (no main effect of side: p=0.358, and no significant interaction between muscle and side: p=0.393) (Fig. 5A). For the abdominal muscles, the phasic response was strongest for posterior perturbations oriented slightly contralateral to the midline, with a progressive shift in orientation to the contralateral side for the RA, EO and IO, respectively. For the back muscles, the phasic response was strongest for anterior perturbations oriented approximately 30 deg contralateral to the midline.

<u>Range of Phasic Response (coefficient '*d*')</u>

The range of the phasic response (d) was also significantly different between muscles (p<0.0001), but generally symmetrical (no main effect of side: p=0.051; and no significant interaction between muscle and side: p=0.405) (Fig. 5B). EO and IO had the widest distribution of directions for their half-maximum phasic response, while RA, T9 and L3 responded in a more focused manner.

Discussion

The purpose of this study was to quantify the response of the superficial trunk musculature to transient, multi-directional loading via a chest harness. We demonstrated that tonic levels of trunk muscle activation, above resting levels, were observed in the quiet sitting posture, suggesting that a degree of muscle activation is required to support spine and postural stability in the unperturbed, upright posture. We also demonstrated that a phasic, direction-specific increase in trunk muscle activation, above tonic levels, was observed following perturbation, suggesting that tonic levels of muscle activation provide only a limited margin of stability, while postural stability following perturbation requires a specific, dynamic response from the trunk musculature. Further, especially for the purpose of designing a controller for an FES system for trunk muscles, we identified the descriptive model for the direction-dependent phasic activity for each muscle.

Tonic Activity

Tonic levels of trunk muscle activation, prior to perturbation, were found to be low: generally between 1 % and 6 % MVC. These values are in line with those predicted by different trunk models for the maintenance of a stable, upright posture (Cholewicki and McGill, 1996, Cholewicki et al., 1997, Gardner-Morse and Stokes, 1998, Granata and Wilson, 2001). Furthermore, these levels of tonic activation appear practical as a means to ensure postural stability in quiet sitting, as they can likely be maintained for prolonged periods of time without inducing muscular fatigue (Hunter et al., 2004, Kahn et al., 1997).

The inter-muscular differences in tonic activation levels observed in the current study (Fig. 1C) are likely explained by two factors. First, the centre of mass of the trunk falls somewhat anterior to the vertebral bodies, particularly at thoracic levels (Pearsall et al., 1996). Add to this the fact that the arms were crossed over the subjects' chest, rather than by the sides, and the higher levels of tonic activation in the erector spinae, compared with the abdominal musculature, may have simply been the result of the equilibrium requirements of the posture. Second, the instructions to the subjects to "sit up tall", as opposed to the somewhat more relaxed (and common) slouched or slumped posture, may have led to elevated levels of tonic activity in the back muscles (O'Sullivan et al., 2006, O'Sullivan et al., 2002).

Phasic Response

As expected, a phasic increase in trunk muscle activation, above tonic levels, was observed following perturbation (Fig. 2), with this phasic response being highly direction dependent for all muscles tested (Fig. 3). The results of the curve-fitting of this phasic response also found the direction of maximum response (coefficient '*c*') to be significantly different between the different trunk muscles (Fig. 5A), suggesting a unique mechanical action for each trunk muscle studied. This assertion is further strengthened by the finding that the range of the phasic response (coefficient '*d*') was also significantly different between muscles (Fig. 5B). This is, to our knowledge, the first quantitative data providing both a direction of maximum response for the trunk musculature, outside of the constraints of the actual directions tested, as well as a quantitative description of the relative distribution of the response amplitude relative to direction.

Although several previous studies have investigated the phasic response of the trunk musculature following perturbation (Brown et al., 2003, Cresswell et al., 1994, Mannion et al., 2000, McMulkin et al., 1998, Preuss and Fung, 2007, Radebold et al., 2000, Stokes et al., 2006, Stokes et al., 2000, Thomas et al., 1998, Vera-Garcia et al., 2007, Zedka et al., 1998)., only a few studies tested perturbations in multiple directions. McMulkin et al. (1998) reported trunk muscle responses to perturbations in multiple directions during standing, and showed different muscle response amplitudes among perturbation directions. The direction dependency of the phasic response was not quantified, however, as the main purpose of their study was to compare two different perturbation methods (applied via a harness or to subjects' hands). Thomas et al. (1998) did conclude that the EMG amplitude of the muscle response of erector spinae was dependent on the perturbation direction. This was based, however, on loading from only two directions: forward and 45 deg right of forward. Stokes et al. (2006) investigated responses of trunk muscles to perturbations in multiple directions, in standing, under preloaded conditions, and also found that the muscle response depended on the pulling direction. Preuss and Fung (2007) recently investigated the phasic response of the trunk musculature, using multi-directional translations of the support surface (8 directions, similar to the current study), in both sitting and standing. These authors concluded that the trunk musculature responds to perturbation not only in a direction-specific manner, but in a manner specific to the actual perturbation experienced by the trunk depending on the given motor task. As such, these results should be viewed as highly complementary to the results of the current study, given the difference between the perturbation paradigms. The application of the Gaussian fit model to the results of the current study also builds upon these previous results, providing more sensitive information with respect to the unique direction-specificity of the phasic activation of each trunk muscle.

This unique direction-specificity may be largely explained by the musculoskeletal geometry of the trunk (Stokes and Gardner-Morse, 1999). As RA, T9 and L3 span the trunk vertically, in the sagittal plane, those muscles are likely to have the greatest mechanical advantage against perturbations in the sagittal plane. Conversely, since EO and IO have fibers which cross the trunk diagonally and transversely, the range covered in the horizontal plane is wider than for RA, T9 and L3, suggesting a wider range of action for the abdominal obliques. This is supported by the values of coefficient '*d*' of the Gaussian fit for these muscles (Fig. 5B). In addition to these mechanical differences, it has also been suggested that the central nervous system may tune the activation level based on the musculoskeletal geometry (Vasavada et al., 2002). This would suggest that these phasic responses are not simply a result of a mechanical interaction between an external perturbing force and a mechanical system, but represent (at least in part) a specific postural strategy adopted by the CNS, based on factors such as prior experience.

Relevance of Findings for the Development of an FES System

The findings of the current study provide several pieces of information that are essential to the development of an FES system to support a stable sitting posture following a spinal cord lesion. First, these results confirm previous observations that the tonic activation of the trunk musculature, above resting levels, is present during upright sitting. Second, a phasic response from the superficial trunk musculature is required to maintain this upright sitting posture following application of a transient, horizontally directed load. Third, this phasic response differs between muscles, but may be quantitatively modeled using a Gausian function with respect to the distribution and relative amplitude of the response between perturbation directions. Finally, both the tonic and phasic activity of the trunk musculature, under the conditions tested, are statistically symmetrical.

The tonic activation observed in quiet sitting is likely present to maintain what has been termed 'sufficient stability' (McGill and Cholewicki, 2001): a level of stiffness sufficient to provide stability under the current loading conditions, while minimizing joint loading and the potential for muscle fatigue. This strategy may also provide a degree of pre-loading to the tendons to maximize the efficiency (i.e. minimize the electromechanical delay) of any additional phasic muscle activation. This suggests the need for an open-loop component in the design of a controller for an FES system for the trunk.

The phasic behavior of the trunk muscles observed in the current study also support the need for a closed-loop component for the controller of an FES system, using kinematic feedback to drive an appropriate phasic response from the trunk musculature (Vette et al., 2007). Based on these findings, it appears that this control will have to be separate for each muscle, in order to properly mimic the observed direction-specific. Further work will be required, however, in order to expand upon the potential inter-dependency of the response from each of these muscles.

Study Limitations

The use of manual perturbation represents a limitation of the current study protocol; particularly in light of the significant difference in perturbation force found across perturbation directions (Fig. 1B). The within-subject variability in these forces, however, was small (about 7 %), suggesting that the peak force was relatively consistent for each subject. Further investigation using motorized equipment with various levels of perturbation force will be required to investigate the extent of this limitation.

The use of surface EMG also presents a limitation for the current study. In addition to the limitations inherent to surface EMG (De Luca, 1997), the muscles of the trunk tend to be large in relation to the collection area of the electrodes, raising the possibility that activation is not uniform throughout the muscle. The abdominal obliques, in particular, have variable fiber orientation across their area, suggesting potential variability in the mechanical actions of different fiber groups within the same muscle (Dickstein et al., 2004). Certain groups have suggested multiple electrode sites for the RA and EO (Davidson and

Hubley-Kozey, 2005), as a means to better capture the true activation patterns of these muscles. Unfortunately this cannot be accomplished with surface EMG for muscles such as the IO, and poses particular problems with respect to normalization of the EMG signals (i.e. the collection of true MVC for each "fiber group" within the muscle). Additionally, other muscles which might contribute to the postural stability of the trunk (such as Lumbar Multifidus and Quadratus Lumborum) were not monitored. These issues will be more thoroughly addressed in future studies.

Finally, the pelvis and lower limbs were relatively unconstrained in the current study. Further work is required to determine the impact of different degrees of pelvic and lower limb constraint on the postural response in the trunk.

Conclusion

Tonic activity in the trunk musculature in upright sitting is low, but still above resting levels by at about 1-3 % of the MVC for the abdominal muscles, and 4-6 % for the back muscles. This suggests the need for open-loop control in the design of an FES system for sitting postural control. Each trunk muscle also showed a unique, direction-specific, phasic increase in activation in response to perturbation, above these tonic levels of activation, likely based (at least in part) on its musculoskeletal geometry. This phasic activation, however, can be accurately modeled using a descriptive model, which may be applied in the design of closed-loop controller for FES.

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Figure Legends

Fig. 1 A: Experimental setup. The perturbation force was applied from 8 directions (top). Two researchers cocked two ropes in two different directions (bottom). One direction was the intended pulling direction, where the force transducer was attached. The purpose of cocking the other direction was to prevent the subject from anticipating the pulling direction. To maintain consistency, all external perturbations were performed by a single researcher. B: The peak value of the perturbation force. C: The tonic activity for all muscles, across all directions of perturbation. The black bars represent the left muscles, and the grey bars represent the right muscles. Data shown in B and C are the group means ± a standard deviation, for all directions. Results of the post-hoc multiple comparison are shown in B and C using horizontal bars.

Fig. 2 The group ensemble averaged traces of the phasic responses for all muscles, in 8 pulling directions. Each trace is the ensemble average of all subjects. The bottom traces represent the group average pulling force in each direction.

Fig. 3 The phasic response for all muscles and directions. The black bars represent the left muscles, and the grey bars represent the right muscles. Data are the group mean ± a standard deviation, for each direction. The pulling directions are illustrated in the figure legend (bottom right), and are mirror-imaged for the left and right-sided muscles to allow for comparison.

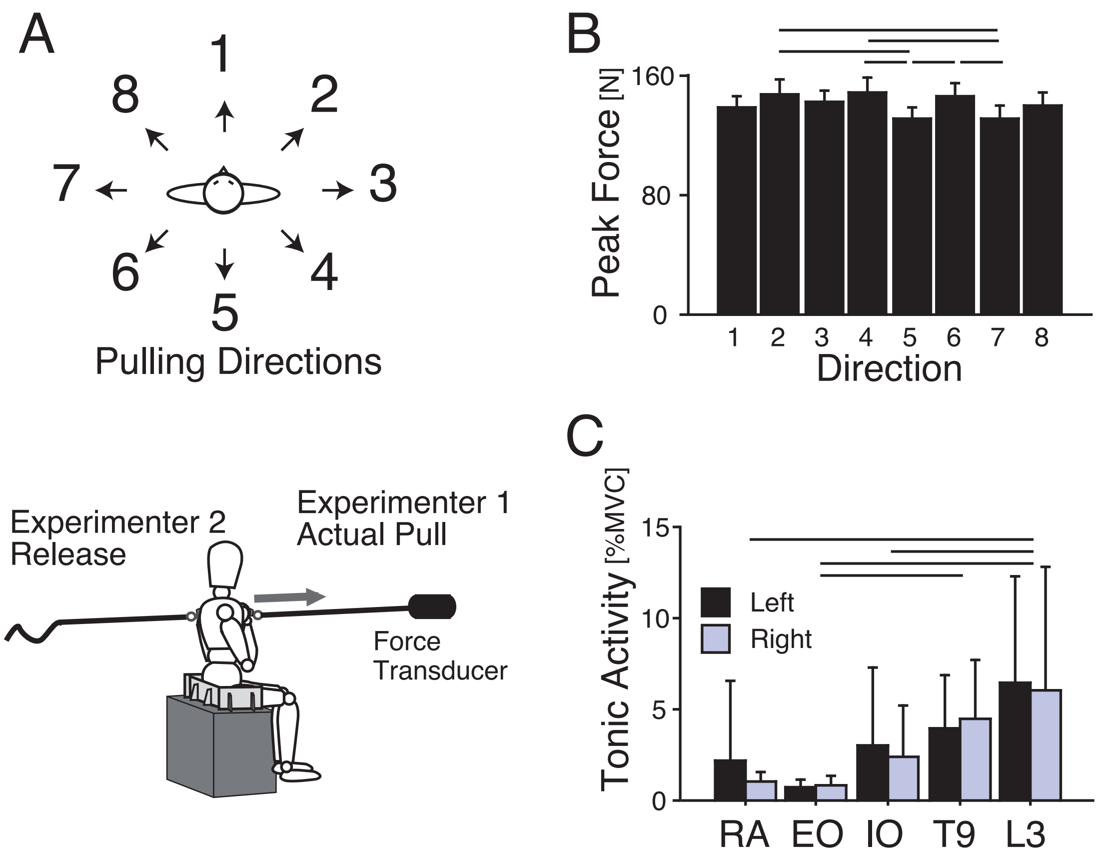
Fig. 4 Results of the curve-fitting for a single representative subject. The closed circles and the open circles represent the phasic responses of left and right-sided muscles, respectively. The thick lines and thin lines are the fitted Gaussian functions for the left and right-sided muscles, respectively. In each plot, the coefficients c and d as well as the correlation coefficient *r* of the curve-fitting are shown. The pulling directions are illustrated in the figure legend (bottom right), and are mirror-imaged for the left and right-sided muscles to allow for comparison. The horizontal axes of the T9 and L3 plots have been shifted to better illustrate the fitted Gaussian functions.

Fig. 5 The estimated angle of the maximum phasic response *c* (A) and the estimated width of the phasic response *d* (B) for all muscles. The black bars represent the left muscles, and the grey bars represent the right muscles. Data are the group mean ± a standard deviation, for all directions. Results of the post-hoc multiple comparison are shown using horizontal bars. The pulling directions are illustrated in the figure legend (bottom), and are mirror-imaged for the left and right-sided muscles to allow for comparison.

Table 1: Results of the curve-fitting. a, b, c, and d are the coefficients in eq. 1. χ^2 and r are the χ-square value and the correlation coefficient for the curve-fitting, respectively. Coefficients are reported as group mean (SD).

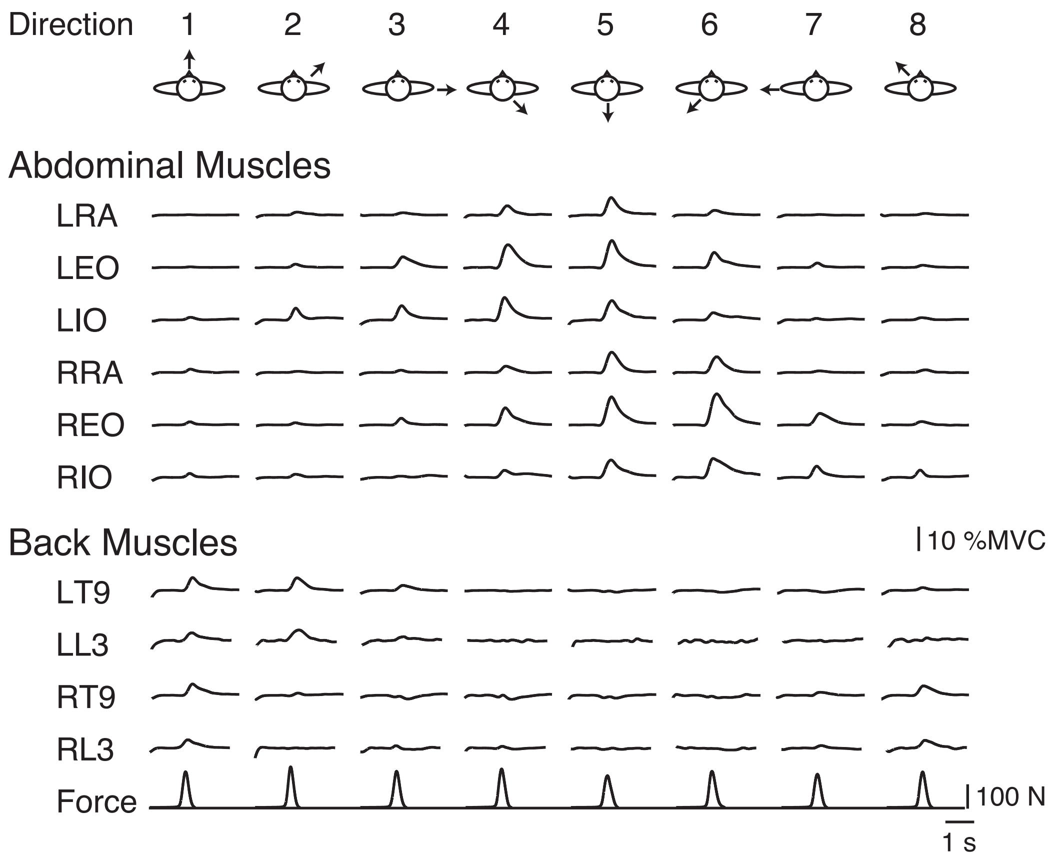
		a	b	c	d	χ^2	R^2
RA	Left	0.108(0.123)	11.4(4.6)	166(6)	49.2(7.2)	0.846(0.785)	0.990(0.012)
	Right	0.295(0.342)	23.9(12.5)	172(7)	44.4(7.3)	4.69(5.93)	0.988(0.011)
EO	Left	0.225(0.490)	14.3(12.7)	150(9)	68.7(17.1)	13.7(29.0)	0.977(0.018)
	Right	0.549(0.987)	21.2(10.6)	166(10)	74.6(15.0)	22.6(26.6)	0.965(0.031)
IO	Left	1.0165(0.999)	12.9(8.7)	142(24)	88.9(35.2)	32.6(60.7)	0.845(0.169)
	Right	1.942(1.381)	16.0(4.9)	140(25)	68.3(14.8)	11.3(11.0)	0.953(0.049)
T9	Left	0.355(0.824)	10.0(6.0)	29(14)	54.9(17.7)	3.36(2.74)	0.944(0.052)
	Right	0.585(1.230)	17.5(15.2)	24(10)	45.6(17.2)	15.7(33.4)	0.924(0.082)
L3	Left	0.301(0.659)	11.1(6.1)	25(10)	42.9(10.1)	4.10(6.80)	0.950(0.068)
	Right	0.241(1.281)	19.7(20.1)	29(11)	42.2(6.6)	11.7(24.6)	0.969(0.049)

Figure 1



A
1
2
3
4
5
6
7
8
Pulling Directions
Experimenter 2
Release
Experimenter 1
Actual Pull
Force
Transducer
B
Peak Force [N]
160
80
0
1 2 3 4 5 6 7 8
Direction
C
Tonic Activity [%MVC]
15
10
5
0
Left
Right
RA EO IO T9 L3

Figure 2



Direction
1
2
3
4
5
6
7
8
Abdominal Muscles
LRA
LEO
LIO
RRA
REO
RIO
Back Muscles
10 %MVC
LT9
LL3
RT9
RL3
Force
100 N
1 s

Figure 3

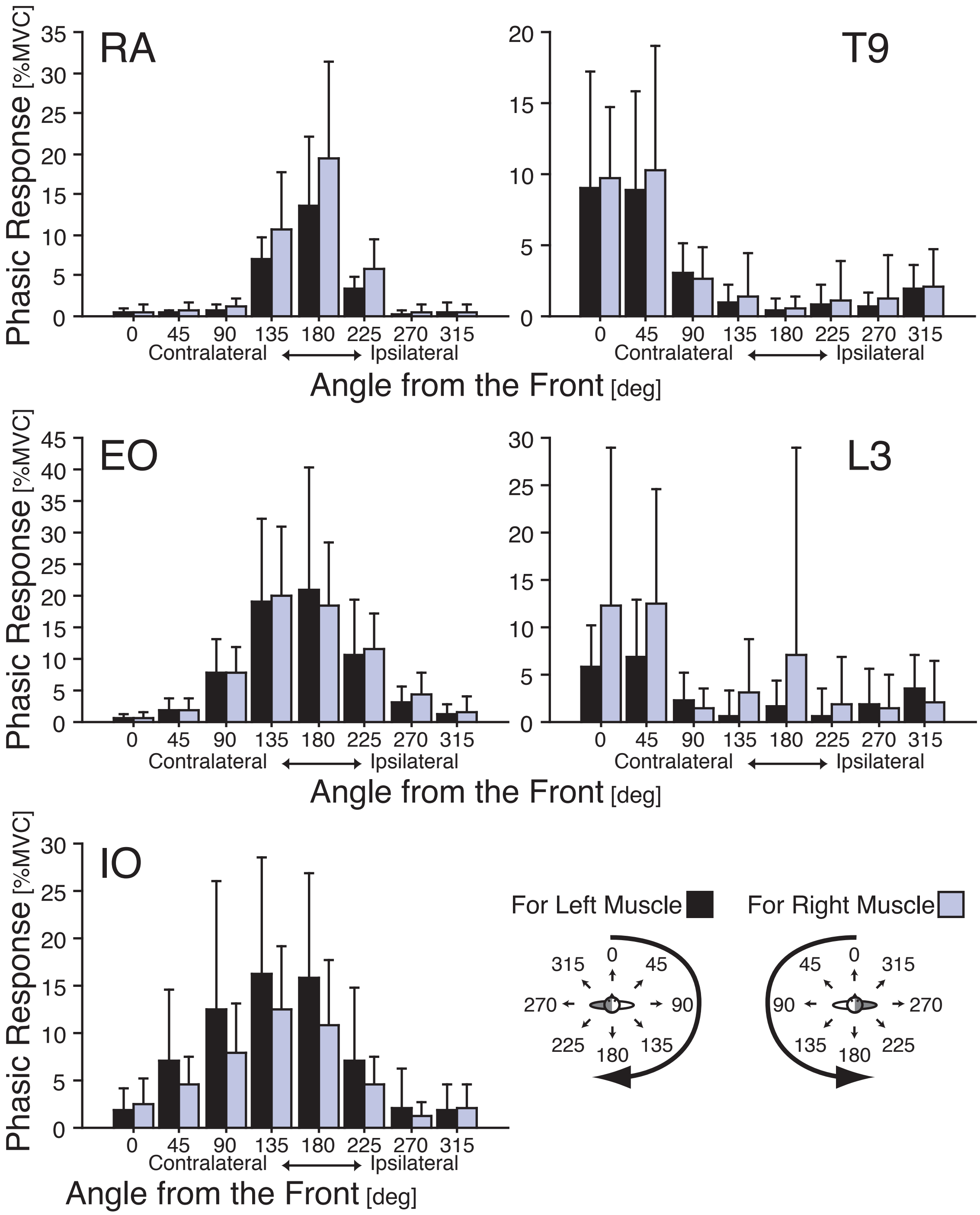

RA
T9
EO
L3
IO
Phasic Response [%MVC]
0
5
10
15
20
25
30
35
45
40
0 45 90 135 180 225 270 315
Contralateral
Ipsilateral
Angle from the Front [deg]
For Left Muscle
For Right Muscle
315
0
45
270
90
225
180
135
45
0
315
90
270
135
180
225

Figure 4

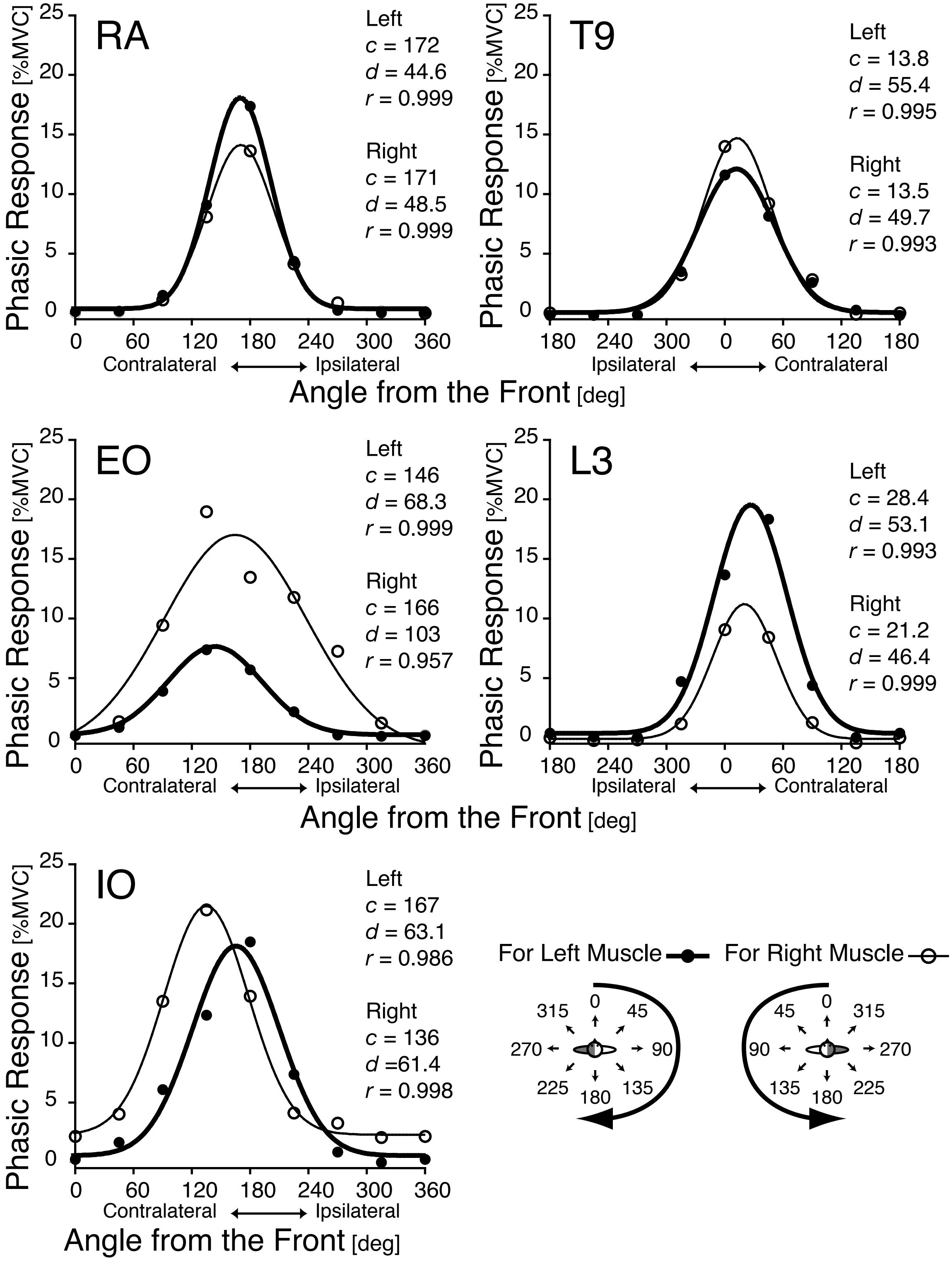
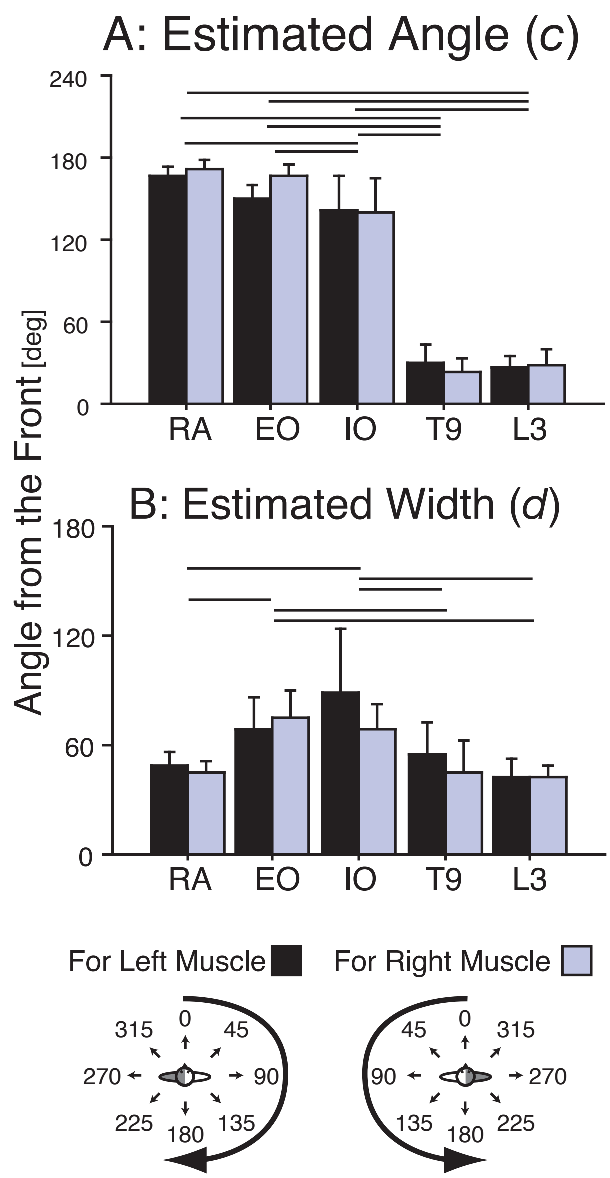

RA
Left
c = 172
d = 44.6
r = 0.999
Right
c = 171
d = 48.5
r = 0.999
T9
Left
c = 13.8
d = 55.4
r = 0.995
Right
c = 13.5
d = 49.7
r = 0.993
EO
Left
c = 146
d = 68.3
r = 0.999
Right
c = 166
d = 103
r = 0.957
L3
Left
c = 28.4
d = 53.1
r = 0.993
Right
c = 21.2
d = 46.4
r = 0.999
IO
Left
c = 167
d = 63.1
r = 0.986
Right
c = 136
d =61.4
r = 0.998
Phasic Response [%MVC]
Angle from the Front [deg]
Contralateral
Ipsilateral
For Left Muscle
For Right Muscle
0
45
90
135
180
225
270
315

Figure 5



A: Estimated Angle (c)
240
180
120
60
0
RA
EO
IO
T9
L3
Angle from the Front [deg]
B: Estimated Width (d)
180
120
60
0
RA
EO
IO
T9
L3
For Left Muscle
For Right Muscle
315
0
45
270
90
225
180
135
45
0
315
90
270
135
180
225