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Time evolution of the organization of multi-muscle postural responses to sudden changes in the external force applied at the trunk level

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Abstract

We studied muscle activation patterns in response to perturbations of posture (sudden changes in the external force applied to the thorax) during two time intervals corresponding to preprogrammed postural reactions and voluntary corrections of posture. A hypothesis was tested that a set of postural muscles could be used to form stable groups (M-modes) whose composition changes in different time intervals after a perturbation. Perturbations were applied at the sternum level to standing subjects at an unexpected time. Principal component analysis with factor extraction allowed to identify sets of three factors (M-modes) during the two time intervals, 80–180 ms (T_1) and 250–450 ms (T_2) after the perturbation. The composition of M-modes was similar within each time interval across subjects and perturbations but differed significantly between T_1 and T_2 . In particular, M-modes during T_1 were characterized by more co-contraction patterns. The results suggest that the neural controller is able to rearrange M-mode composition in real time based on a safety-efficacy trade-off. The results also support the idea that M-modes represent synergies in the muscle space, while they may be used as elemental variables to form synergies at a higher hierarchical level to produce desired mechanical effects.

Keywords

synergy; posture; muscle mode; postural perturbation; electromyogram

In several recent studies, the multi-muscle control of vertical posture has been viewed as based on a hypothetical hierarchical scheme [12,15]. At the upper level, the task is shared among a relatively small number of elemental variables (muscle modes or M-modes), while at the lower level M-modes are projected on numerous individual muscle activations [2–3,5–9,15]. M-modes (sometimes addressed as “synergies”) have been studied in a variety of postural tasks, from voluntary sway [3,9] to responses to external perturbations [16]. In most of these studies, M-modes have been viewed as relatively stable eigenvectors in the muscle activation space based on similarities of their directions across subjects and tasks [3,8]. However, a few studies have suggested that M-mode composition may change, in particular in challenging postural tasks and with practice [1,10]. Although changes in the relative **magnitude** of the M-modes at different times following a perturbation have been described [16], time changes in the

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composition of M-modes in such tasks have not been documented. The main goal of this study is, therefore, to investigate the time evolution of the composition of M-modes following a perturbation. In addition, this study considers effects of prior knowledge of the perturbation on the M-mode composition, as this question remains unclear [13].

Seven subjects, 5 males and 2 females, took part in the experiments. Their mean age was 25.6 years (SE 1.1), mean height was 1.71 m (SE 0.02) and mean weight was 68.6 kg (SE 2.7). All subjects were healthy, without any known peripheral or neurological disorder. All the subjects gave informed consent according to the procedures approved by the Office of Research Protection of the Pennsylvania State University.

A force platform (AMTI, OR-6) was used to measure the moments of force about the central frontal and sagittal axes of the platform and the three components of the ground reaction force. Disposable self-adhesive electrodes (3M Corporation) were used to record the surface muscle activity (EMG) of the following muscles on the right side of the subject's body: soleus (SOL), gastrocnemius medialis (GM), tibialis anterior (TA), biceps femoris (BF), rectus femoris (RF), vastus medialis (VM), lumbar erector spinae (ES), and rectus abdominis (RA). The distance between the two electrodes of each pair was 3 cm. The signals from the electrodes were pre-amplified ($\times 3000$) and band pass filtered (60–500 Hz). All the signals were sampled at 960 Hz with a 12-bit resolution.

The subjects were equipped with an adjustable, custom made upper body harness. Masses could be attached to this harness, at the subject's lower sternum level, via a pulley and string system (Fig. 1). As a result, horizontal forces could be applied to the subjects in the forward and/or backward direction (F_{front} and F_{back}). The loads were attached via two electromagnets, and thus could be released unexpectedly for the subject. Two low-drift force sensors (PCB Piezotronics, 208C02) were used to record F_{front} and F_{back} . The resultant of F_{front} and F_{back} was computed and is referred to as F_{perturb} .

The experiment began with three control trials with quiet standing: without a load and while holding in the extended arms a 5 kg load that acted downward in one trial and upward in another one. These trials were later used for the EMG normalization, as in previous studies [8–9]. The control trials were followed by 96 trials with perturbations. For each of these trials, the subjects were given visual feedback on X_G , the location of the COM projection onto the anterior-posterior axis; X_G was estimated assuming static posture (see Fig. 1) and expressed as a deviation from its value observed during the quiet standing control trial. The subjects were always instructed to keep X_G as close as possible to zero, to avoid stepping, and to keep their arms crossed on their chest. Subjects stood barefoot, with the feet parallel to each other, spaced by 20 cm, symmetrical relative to the X axis of the force platform. The midline of the body was aligned with the line of action of the loads.

The procedure for the perturbed trials was as follows. Two loads of equal mass, one in the front and one in the back, were attached via the electromagnets at the chest level. The resultant force applied to the subject was zero, and the subject stood in a relaxed state with X_G close to zero. The data collection started and, after a delay varying randomly between 3 and 9 s, one of the two loads was unexpectedly released leading to a non-zero resultant F_{perturb} . Note that this method results in a predictable simple time pattern of F_{perturb} , while the commonly used platform translations result in a more complex transient perturbation [13]. The subjects had to bring X_G to zero as quickly as possible and keep it at zero until the end of the trial (15 s). Note that this posture did not correspond to the resting posture any more.

Six different perturbation amplitudes were used in each direction corresponding to the load masses of 1.3 kg, 1.75 kg, 2.21 kg, 2.66 kg, 3.02 kg and 3.47 kg. For each load mass, 16 trials were recorded in 3 blocks. The first two blocks consisted of 4 trials each, all four trials with

perturbations in the same direction, forward or backwards. The subjects were always aware of the direction of upcoming perturbations. Before each of the two blocks, the subjects had 2 practice trials to get used to the perturbation. The order of the two blocks was balanced across loads. The third block consisted of 8 trials (4 trials for each perturbation direction) in a random order. Within this block, the subjects were not informed in advance of the direction of perturbations. The order of the perturbation amplitude was randomized across subjects. If the subject made a protective step, this trial was discarded and repeated at the end of the block. The largest perturbation in the backward direction appeared to be most challenging: Five subjects stepped at least once. However, with up to 3 additional attempts, they were all able to perform correctly the required number of trials.

The data were processed off-line using Matlab 7.3 software package. All signals were filtered with a 50 Hz low-pass second-order zero-lag Butterworth filter. The EMG signals were rectified and normalized by the averaged EMG signals in one of the control trial [8–9]. The EMGs of the dorsal muscles (SOL, GM, BF, ES) were normalized using the data from the control trial where the load was acting downward. The EMG of the frontal muscles (TA, RF, VM, RA) were normalized using the data from the control trial where the load was acting upward. For each trial, the onset of the perturbation (t_0) was defined, using the change in force induced by the load release, at the time when the force derivative exceeded 5% of its peak value in that trial. Only the data 0.5 s before and 4 s after t_0 were kept for further analyses.

An example of the normalized EMG time profiles, averaged over four trials, in a typical subject in response to the predictable backward perturbation produced by releasing the 2.21 kg load is shown in Fig. 2. The lower right panel shows F_{perturb} . As expected, the muscles opposing the perturbation (the frontal muscles, TA, RF, VM, RA) showed a higher level of activation as compared to the dorsal muscles (SOL, GM, BF, ES).

Two phases of the response to the perturbation were analyzed, from 80 to 180 ms after t_0 (T_1), and from 250 to 450 ms after t_0 (T_2). These time intervals were defined based on the literature [4], and the visual inspection of the EMG signals (Fig. 2), to reflect a relatively short-latency, pre-programmed postural response and a later voluntary postural adjustment. For each of the two intervals, the normalized EMG data were integrated over 4 bins of equal duration (25 ms and 50 ms for T_1 and T_2 , respectively).

Further, muscle activation levels were analyzed to define muscle groups with activation levels co-varying across perturbation directions and magnitudes, namely muscle modes or M-modes [3,8–9,14]. For each subject, each block (known and unknown direction of perturbation), and each interval of analysis (T_1 and T_2), the 600×8 IEMG correlation matrix (4 time windows * 4 trials * 2 directions * 6 loads; 8 muscles) was subjected to principal component analysis (PCA). Consistently with the cited previous studies, only the three first PCs were considered and subjected to factor extraction using Varimax rotation. On average the three factors (M-modes) accounted for 78.4% of the total variance (40.1%, 22.9% and 15.3% respectively). ANOVAs with factors *Period* (two levels, T_1 and T_2) and *Block* (two levels, known and unknown direction of perturbation) performed on the amounts of variance explained by individual M-modes and on the total amount of variance explained showed no significant effects.

Table 1 shows the M-mode composition (factor loadings) averaged across subjects. Note that the M-mode composition was very similar across blocks but not across the two time intervals. For T_2 , the muscles formed two groups, as reported in earlier studies [3,8–9], M_1 and M_2 uniting the dorsal and frontal muscles respectively. For T_1 , however, the M-mode composition was more complex, with frontal and dorsal muscles being significantly loaded within the same M-mode.

For statistical analysis of the M-mode composition, we computed the cosines between the individual M-mode vectors in the muscle activation space and a reference vector defined as the central vector (analogous to the mean vector, see [8]) across all the subjects for a given combination of block and time interval. Different cosine values reflect different orientations of individual vectors relative to the reference vector. These cosine values were log-transformed into z-scores. Two-way ANOVAs on the z-scores with factors *Period* (T_1 and T_2) and *Block* (predictable and unpredictable direction of perturbation) showed no effects of *Block* and no interaction. Hence, the data were collapsed over the factor *Block* and a one-way ANOVA with the factor *Period* was run. It showed a significant effect for the two first M-modes ($F_{(1,26)} = 11.77, p < 0.01$ and $F_{(1,26)} = 4.34, p < 0.05$ for M_1 and M_2 , respectively) and an almost significant effect for M_3 ($F_{(1,26)} = 4.13, p = 0.052$).

We also estimated the occurrence of co-contraction patterns in the M-modes. Co-contraction patterns were defined as those involving high loading factors (> 0.5 ; cf. [8,14]) with the same sign within the same M-mode for two muscles with opposing actions at one of the major joints. Such patterns were observed in *individual* data for the VM/RF & BF in both blocks, and also in ES & RA and TA & SOL for the block with perturbations of unpredictable direction. The corresponding loadings (averaged across subjects) are highlighted in Table 1. The total number of co-contractions observed in individual results summed over the subjects and over both blocks of trials was twice as high for the T_1 time interval as for the T_2 time interval (18 and 9 respectively). The occurrence of co-contraction patterns was the same whether the direction of perturbation was known or not. The influence of the factor *Period* on the number of co-contraction patterns was checked using Wilcoxon's signed-rank test. It confirmed that the occurrence of co-contraction patterns was significantly higher ($p < 0.05$) for T_1 than for T_2 .

The main result of the study is the demonstration of changes in the M-mode **composition** at different times following a perturbation. The early postural responses that come at a latency of about 80 ms have been addressed as long-latency reflexes, automatic postural corrections, and pre-programmed reactions (reviewed in [11]). The purpose of these responses is to provide an approximate compensation for the perturbation based on an estimate of its expected magnitude. The co-contractions M-modes (more common in T_1) may be viewed as the means of increasing the apparent stiffness of the postural joints and stabilizing the body to an upcoming perturbation of any magnitude and direction. Such M-modes have been reported for challenging postural conditions, for example while standing on a board with a decreased support area [1,10]. The more commonly seen "push-back" and "push-forward" M-modes [3,8–9] may be viewed as more efficient in moving the center of mass but also less safe in poorly predictable external conditions. In such conditions, using an inappropriately scaled M-mode may itself turn into a source of postural perturbation.

The observed different M-modes during the two time intervals suggest that the neural controller is able to rearrange M-mode composition in real time based on a safety-efficacy trade-off [14]. The early response based on an estimate of the perturbation characteristics uses a set of M-modes with more co-contraction patterns, while the later response based on actual perturbation characteristics primarily uses more efficient "push-back" and "push-forward" M-modes. Note that both sets of M-modes project on the same postural muscles suggesting that M-modes indeed represent synergies in the muscle space, while they may be used as elemental variables to form synergies at a higher hierarchical level to produce desired mechanical effects [14].

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Figure 1.

Schematic representation of the experimental set-up, and estimation of the center of gravity location along the antero-posterior axis X_G . Abbreviations: m_{back} and m_{front} - the masses of the back and front loads respectively, F_{back} and F_{front} - the forces applied on the harness by the back and front loads respectively, Z_F - coordinate of the point of application of these forces along the Z axis, M_Y - the torque measured by the force platform along the Y axis, m - the mass of the subject and g - gravity constant.

Figure 2.

Typical time profiles of the rectified, filtered and normalized EMG signals averaged across 4 trials performed by a representative subject in a series with load = 2.21 kg, backward perturbation, known direction of perturbation. TA – tibialis anterior, RF – rectus femoris, VM – vastus medialis, RA – rectus abdominis, SOL – soleus, GM – gastrocnemius medialis, BF – biceps femoris, ES – erector spinae. EMG signals of the dorsal muscles (SOL, GM, BF, and ES) are inverted for better visualization. The lower right panel shows the perturbation force applied to the subject at time t=0.

Results of the PCA Analysis

Loadings coefficients averaged across subjects ($\pm$ SEM) are shown for each muscle, the two blocks (known or unknown direction of perturbation) and the two periods of analysis (T₁ and T₂). The highest loadings are shown in bold. TA – tibialis anterior, RF – rectus femoris, VM – vastus medialis, RA – rectus abdominis, SOL – soleus, GM – gastrocnemius medialis, BF – biceps femoris, ES – erector spinae.