Postural EMG responses in man after perturbations in multiple directions

Michael Voigt[1,2], Zlatko Matjacic[1],Dejan Popovic[1], Thomas Sinkjær[1]
[1]Center for Sensory –Motor Interaction, Aalborg University, Frederik Bajers Vej 7 D3, 9220 Aalborg Ø. [2]Institute of Exercise and Sport Sciences, University of Copenhagen
Denmark
E-mail: mv@smi.auc.dk

Abstract – *The aim of the present study was to analyze the pattern of muscle synergies that underlies our recently proposed control strategy of unexpectedly perturbed stance that reduces the multivariable control problem of stance into two decoupled univarible problems. Eight healthy subjects participated in the present study. During quiet standing they were unexpectedly perturbed with a special custom made apparatus in 8 different directions with three different perturbation strengths. Additionally, the orientation of the feet was varied between the preferred position, widely outward rotated and parallel. Surface EMG was recorded from 11 leg and trunk muscles on the right side. The magnitude of the muscle responses showed a characteristic directional pattern for each muscle. The patterns were rather consistent between the subjects, they scaled with perturbation strength and they were not clearly influenced by the changes in the orientation of the feet. These observations suggests that the responses to perturbation generates a series of muscle synergies specific to the perturbation direction, and that the generation of these synergies mainly are determined by the perturbation direction i.e. sensory information elicited by the perturbation independent of the biomechanical configuration of the legs. However, afferent information from the legs may fine tune the relative muscle activation within the different synergies.*

1. Introduction

It has been proposed that the central nervous system use synergies as fundamental building blocks to synthesize postural responses after perturbed stance [3]. Use of synergies allows the central nervous system to utilize reduced number of inputs to control redundant multiple degrees of freedom; thus effectively reduces the number of degrees of freedom of the complex biomechanical system. This hypothesis evolved from studies utilizing perturbations in the anterio-posterior (AP) direction alone. Only few studies have analyzed the patterns of muscle synergies underlying the control strategy behind responses to multidirectional postural perturbations. From one of the earlier studies it became obvious that the relations between the muscles in the postural responses must be thought as functions of several variables rather than fixed entities and the composition of postural responses most likely is a complex process that includes perturbation direction as a continuous variable [7]. In a recent study [2] it was suggested that a flexible continuum of muscle synergies, rather than fixed muscle synergies (and simple reflex patterns), that are modifiable in a task-dependent manner is used for equilibrium control in stance. Additionally, evidence was found for a centrally mediated timing of muscle latencies combined with a peripheral influence on the magnitude of the muscle responses [2].

Recently we have proposed [6] that the postural responses in a standing human, following a perturbation delivered simultaneously in two planes, result from a control strategy that transforms a multivariable control problem into two decoupled univariable problems. These are i) the control of balance in the AP plane and ii) control of balance in the ML plane. This control strategy was suggested on the basis on the observation of net joint torque responses, which are the mechanical or functional consequence of the underlying muscle activation patterns also termed the muscle synergies [7]. However, the net joint torque responses do not reveal how the central nervous system (CNS) selects or generates functional muscle synergies.

The aim of the present study was to analyze the pattern of muscle synergies underlying the functional postural responses in the light of our proposed postural control strategy.

2. Methods

Eight male subjects with no known neurologic or orthopedic disorders (mean age 31, SD 6.7 years; height 182, SD 8.8 cm; weight 75.8, SD 11.4 kg) participated in this study

Perturbations were generated at the level of pelvis of a standing subject by means of a two degrees of freedom (DOF) servo-controlled mechanical apparatus named “Multi-purpose Rehabilitation Frame” (MRF). Subjects stood in such a way to have legs in parallel while being braced at the level of pelvis by the adjustable bracing system as shown in Fig. 1a. A detailed description of the MRF device is given in [5].

Perturbing torque pulses in duration of 200 ms were delivered either in one of four principal directions (Forward and Backward in the AP plane; Right and Left in the ML plane) or in one of four combinations of the principal directions (Forward & Right, Forward & Left, Backward & Right, Backward & Left) as shown in Fig. 1b. The subjects were instructed to stand

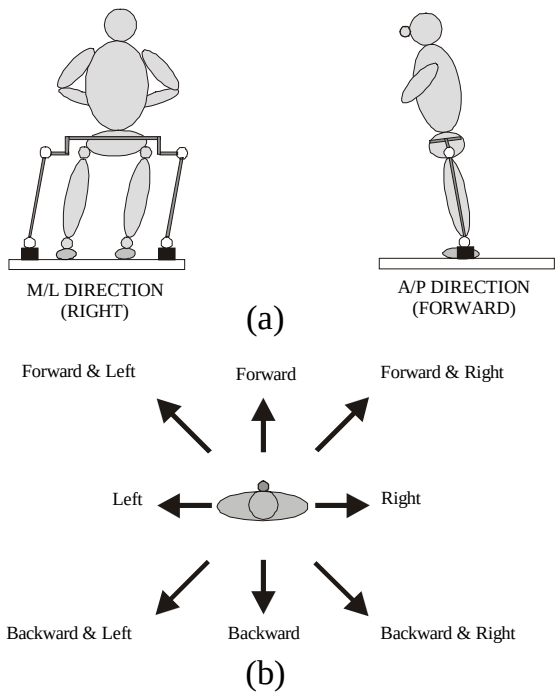

Figure 1. a) the experimental setup. b) the eight directions of perturbation from each perturbation.

relaxed prior to perturbation and return the same initial posture throughout the experiment after the recovery. The perturbations were always delivered at times unknown to the subjects.

Forty unexpected perturbations were delivered in random order between the different directions, giving 5 perturbations in each direction. This was repeated for five randomized trials under the following experimental conditions: 1-3) for three different perturbation strength's (MIN, MED and MAX) with the feet placed with the preferred orientation of the feet and 4-5) perturbation strength MED and the feet placed either widely outward rotated between 30 – 40 degrees (\ /) or with the feet parallel (I I). In the preferred position the feet were typically outward rotated about 15 degrees. Prior to initiation of each experiment an introductory an series of perturbations was undertaken where the amplitude of perturbation was varied. The amplitude where the subjects could comfortably recover balance after the perturbation applied in Backward direction and keeping feet flat on the blocks throughout the postural response was selected as a MAX, (MED=MAX/2, MIN=MAX/4).

EMG responses were recorded from 11 muscles on the right side of the subjects with surface electrodes (Medicotest 720-01-K): tibialis anterior (TA), soleus (SO), medial gastrocnemius (MG), peroneus longus (PL), vastus lateralis (VL), rectus femoris (RF), semimembranosus (SEM), gluteus medius (GME), adductor magnus (ADM, erector spinae (ES) and rectus abdominis (RAB). The skin was carefully shaved, abrased with fine sandpaper and cleaned with alcohol before the electrode pairs were mounted. Each electrode pair was placed at the most prominent part of the muscles, with the line between the electrode centers approximately parallel with the muscle fibers and an interelectrode distance of 3 cm The EMG signals were amplified with custom made amplifiers and sampled at 1 KHz with a PC-based data acquisition system and stored for later analysis. During each perturbation the recording was initiated 500 ms prior to the onset of the perturbation and had duration of 3.5 s.

The EMG signals were high pass filtered at 20 Hz, rectified and low pass filtered at 20 Hz (4th order zero lag Butterworth filters) to obtain linear envelopes and the five recordings in each direction were averaged. The onset of the EMG responses were determined as the time after the initiation of the perturbation where a deflection of the EMG envelope deviated more than one standard deviation of the background EMG level. The activity in the EMG bursts was calculated as the mean amplitude over a 200 ms period starting from the EMG onset.

3. Results

In the following the preliminary analyses of the results are presented. The EMG responses were generally burst like increases in EMG activity. However, in muscles with clear postural background activity (especially SO and ES) decreases in EMG activity due to unloading of the muscles were often observed depending on the perturbation direction. The onset times of the EMG responses ranged between 69 and 249 ms depending on muscle and perturbation direction. The pattern was not clear, however there was a tendency to that the most distal muscles (TA, SO, PL) and the most proximal muscles (ES, RAB) responded earlier than the other muscles.

The pattern of directionality in the magnitude of the EMG responses was rather uniform between the subjects. Figure 2. displays a representative example of the patterns of EMG activities from the muscles controlling the ankle, knee and hip joints. In the figure the magnitude of the EMG responses are normalized to the maximum amplitude during the MAX perturbations. Following features should be recognized: a) the amplitude of the responses scales with perturbation strength and b) the shape of the response profile does not change markedly between the situations where the feet are oriented differently while the perturbation strength is kept constant (MED).

4. Discussion

The main observations from the preliminary analysis are 1) that the shape of the EMG response profiles are rather robust both between subjects and within subjects and 2) the profiles scale with perturbation strength and they only change slightly when foot orientation was changed. These observations seems to suggest that the main determinant of the

response profile must be the perturbation direction and not local biomechanical parameters.

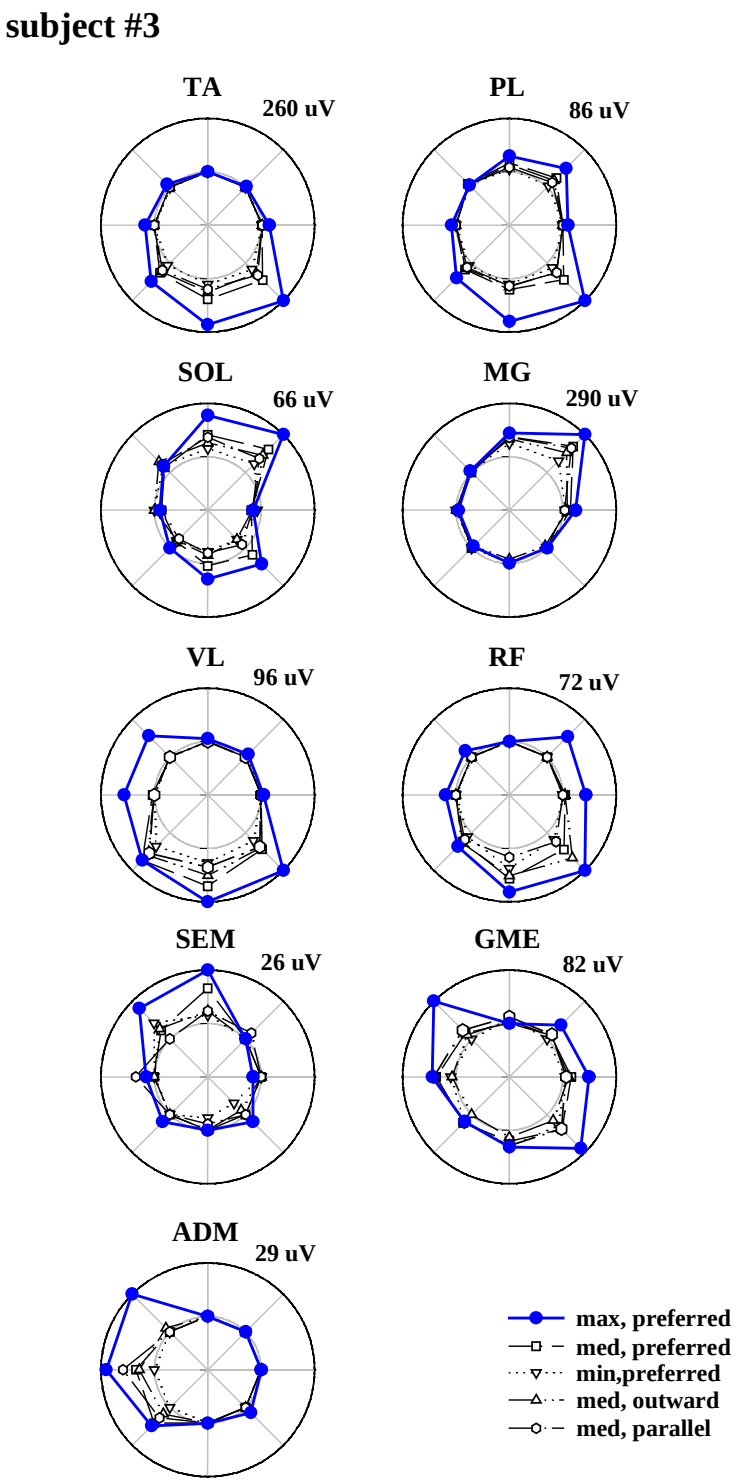

Figure 2. Amplitudes of EMG responses from 9 muscles of the right leg following perturbation of quiet stance from one subject (see text for explanation of abbreviations of muscle names). The perturbations were delivered in 8 different directions according to the radial lines in the polar plots (the person standing in the center face forward (upward)). The perturbations were delivered with three perturbation strengths (MAX, MED, MIN) and with the feet either in the preferred position, widely outward rotated or parallel. MAX is indicated by the fat solid lines. The magnitude of the EMG responses were calculated as the mean amplitude of the EMG envelope over a period of 200 ms starting from the onset of the EMG response. The EMG amplitudes were normalized to the maximum amplitude during MAX perturbations. For each muscle this value is presented upwards left in relation to each polar plot. The center of each plot corresponds to –1, the inner circle corresponds to zero while the outer circle corresponds to +1.

From figure 2 it is clearly seen that the response profiles vary markedly between muscles and this indicates that each perturbation direction has its own functional synergy. When foot orientation is changed the local biomechanical situation for the muscles is changed in relation to the perturbation direction and therefore at the same perturbation strength (MED) the relative weighting of the muscle activation in a given synergy must be fine tuned to keep the torque responses in the principal planes invariant across foot orientations. Information from joint and muscles (i.e. about the local musculo-skeletal configuration) might be important for this fine tuning. However, most likely we cannot 'see' these fine adjustments in this experiment due to a rather low resolution of the EMG-method, or that this fine tuning can only can be observed as 'added noise' because we could not strictly control the change in the biomechanical configuration with changes in foot orientation due to inter-individual anatomical differences.

Our observations indicate that the main determinant for the generation of a specific synergy in response to a given perturbation is sensory information independent of sensory information from the legs. This is in line with the conclusions of Carpenter et al. [1], in which the idea of an ankle triggered signal selection of movement strategy is questioned while it is suggested that the balance corrections may receive strong receptor dependent (otolith or vertical canal) and directionally sensitive amplitude modulating input from vestibulospinal signals.

We find that our observations can be interpreted according to a control system model based on an internal representation of postural orientation (see [4] for a review). This model used sensory information and the constraints of the body dynamics to create internal models of sensory and body dynamics. An external disturbance to posture changes sensory afferent signals and also drives the system to change the internal model of body dynamics as well as to implement a control strategy to reorient the body. In our case the perturbation should then trigger a demand for a desired correction and the afferent input from the legs is used in the comparison of expected and actual afferent input which gives a ‘sensory conflict’ which then is minimized by changing the internal model to drive the response.

Reverting to our proposed postural control strategy with reference to the above mentioned conceptual model framework it seems that the main purpose of the afferent input from the legs might be to signal the degree of fine tuning of the muscle activations in a given synergy to accomplish the invariant torque responses in the principal planes which are characteristic for the proposed control strategy.

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