Role of hand contact in continually challenged postural equilibrium

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Abstract

Balance is defined as the ability to maintain postural stability (equilibrium) during quiet standing, perturbed standing and voluntary movement. Additional hand contact helps in retaining balance by extending the base of support. Little is known, however, about the actual contribution of the hand support in a continually challenged stability during standing. With this study we aimed to investigate the activity of hand contact when balance was challenged by a force perturbation applied to the waist, to determine to what extent balance corrective responses are generalized across continuous perturbation.

We measured the muscle activity of right shoulder, trunk and right lower leg following application of a mild, random and continuous forward and backward pulls at the waist. Results show that in situations with additional hand support, the contribution of lower leg muscles was minimized and suggests that the postural equilibrium was dominantly obtained by the use of hand muscles. Interestingly, the activation of the trunk flexors remained unchanged where the activation of trunk extensors rose significantly in cases where the subjects used the additional hand support.

1 Introduction

During biological evolution, man has developed many motor abilities that enable us to stand and move on only two feet. Maintaining balance during standing and moving is one of these highly important, basic human motor abilities which has been under investigation by many researchers [1]. Among other, there was an extensive focus on compensatory steps and rapid responses of hands which do not necessarily have a direct influence on preserving stability [2]. Three major sensory systems govern human balance [1] – vision, vestibular and somatosensory system. For static stability of the posture, the vertical projection of the body centre of mass (COM) should be within the base of support (BOS) [3] [1]. The BOS, or supporting area, is defined as the possible stable range of the ground reaction force vector, known also as the centre of pressure (COP).

In order to investigate the sensory systems, much of the research has attempted to perturb the balance in various ways and under various conditions [1]. The most common experimental approach is to perturb the support surface, which displaces the BOS under the body's COM, but there are also other approaches which more directly displace the COM, such as pulls or tugs at the waist [4].

For evaluation of postural stability, electrical activity (electromyography or EMG) of muscles, kinematic and kinetic variables are most commonly used. Force plates and motion tracking systems provide data regarding the displacement of the COM and/or the COP. Insight into EMG variables provides important information regarding postural responses such as onsets and magnitude of muscle activation.

While standing over a moving support surface, human motion can be modelled with a combination of ankle and/or hip motions [5]. Horak and Nashner suggested that the central nervous system (CNS) employs ankle strategy if the perturbation can be compensated solely by the ankle torque and in case of bigger perturbations, hip strategy (or combination of both) is required. A principle of abundance by Gelfand and Latash [6] has also been suggested, which states that all elements (degrees of freedom or DOFs) of a structural unit always participate in all of the tasks, assuring both stability and flexibility of the performance. Therefore no strategy is ever eliminated or frozen. Similarly, a more recent study by Scholz et al. [7] suggests that CNS makes use of multi-joint redundancy and performs postural control by a single strategy that considers all DOFs.

In presence of larger perturbations, ankle and hip strategy may not be able to stabilise the body. Therefore stepping strategy can be employed [5][2]. An alternative to the stepping strategy is using arms either as a counterbalancing tool [8] or to make a contact with the environment [2]. Studies indicated that even a light contact with environment can improve postural control by providing additional sensory feedback [9]. Moreover, a firm contact with the environment provides much better stabilising potential [10]. Holding a handle or handrail is therefore a common and suitable option for maintaining balance and preventing falls [11]. Some recent studies [12][13] investigated the effects of handle locations on the postural control and concluded [12] that the most suitable location of the handle is approximately at the shoulder height. Even though these studies include the additional hand contact in balancing situations, they are in most cases describing only reactive compensatory movements. The descriptions of a contribution or influence of the supportive contact in continuous perturbed situations remain to be conducted. Therefore, the aim of our study was to investigate the compensatory aid of the supportive hand contact, when balance is continually challenged by a perturbation applied to the waist.

2 Methods

2.1 Subjects

Experiments were performed on twelve adult male subjects (average age = 22.2 years, *SD* = 2.2 years, average height 179 cm, *SD* = 6.2 cm and average weight = 76.7 kg, *SD* = 8.4 kg). All subjects were right handed and free of any injuries or health issues with the potential to influence balance control. Subjects were informed about the course of the study and consent, approved by the National medical ethics committee (No. 112/06/13), was obtained prior the start of their participation.

2.2 Experimental setup

A motorized waist-pull system [14] was used to continuously destabilize balance. A special waist belt equipped with force sensors and connected to the motorized waist-pull system, motion markers and surface EMG (sEMG) electrodes were mounted on the subject.

Testing sessions began with instrumentation setup and calibration. The first step involved preparation (i.e. shaving and cleaning) and application of electrodes on the skin overlying twelve selected muscles on the subjects right hand side (Biceps brachii - BB, Triceps brachii (long head) - TB, Deltoideus (anterior) - DA, Deltoideus (posterior) - DP, Latisimus dorsi - LD, Trapezius (middle) - TM, Trapezius (higher) - TH, Pectoralis major - PM, Multifidus - MF, Obliques externus - OE, Tibialis anterior – TA and Gastrocnemius lateralis - GA).

Throughout the sessions, electromyography signals were collected from these sites at a sampling rate of 1000Hz. Three maximal voluntary contraction (MVC) trials were recorded from the selected muscles. The MVCs trials were 5s in duration and were obtained via voluntary isometric contractions of selected muscles.

The motorized waist-pull system (Fig. 1) was set to produce a perturbation waveform of both forward and backward direction and with different forces.

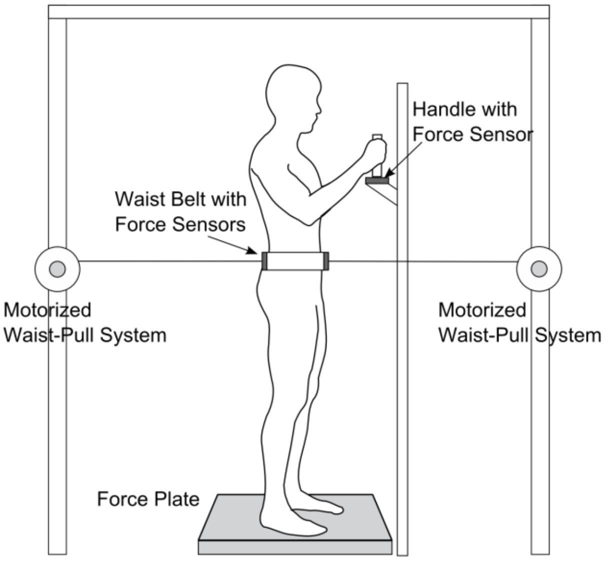

Fig. 1. Experimental setup of the study. The subject is standing on a force plate, wearing a waist belt connected to the motorized waist-pull system which generated translational force perturbations

The perturbation waveform (Fig. 2) was generated using a mild random white noise signal in frequency range of 0.25-1.00 Hz with peak force of 11% of a subject's body weight which was measured using a force plate. Frequency and maximum force were determined by preliminary experimental trials and waveform was constructed in such a manner to eliminate possible muscle reactions associated with reflexes.

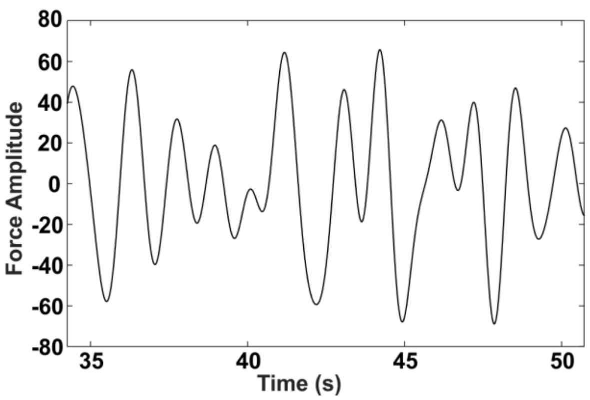

Fig. 2. A 15s sample of a perturbation waveform.

A handle (diameter = 3.2 cm, length = 12 cm) was mounted beside the force plate to the right of the subject (distance-to-body-midline = 0.25 x body-height, height = subjects shoulder height). Handle was mounted on a 3-axis force sensor (45E15A, JR3, Woodland, USA) to record the forces generated by the hand, during perturbations.

2.3 Experimental protocol

The experiment was divided into two distinct sessions – 'handle' and 'no handle'. Each session consisted of fifteen, five-minute continuous perturbations and after every fifth repetition there was a ten-minute break. Subjects were instructed to stand straight with the barefoot feet placed at hip width on the force plate and were required to look straight ahead.

In the first, 'handle' session, the subjects were instructed to hold the handle to assist them with maintaining their balance. In the second session ('no handle'), the subjects were not allowed to hold the handle and had to fold their arms across their chest. They were instructed to try to maintain balance without using their hands and not to make any corrective steps.

2.4 Statistical analysis

A paired-samples t-test was conducted to compare integrated EMG (iEMG) activity of two trunk and two lower leg muscles in 'handle' and 'no handle' conditions.

3 Results

Fig. 3 shows a representative comparison of a recorded sEMG activity of TA muscle. The comparison is between sessions where the subject was using a handle and where the subject was not allowed to use the handle as an additional supportive contact. After visual inspection it is apparent that there was a big difference in the activity of the muscle. The same observation was also noted for the other lower leg muscle, GA. This could already suggest that when using the hand as a complementary stabilizer, the most of the stabilization activity transfers from lower leg muscles to other muscles of the body.

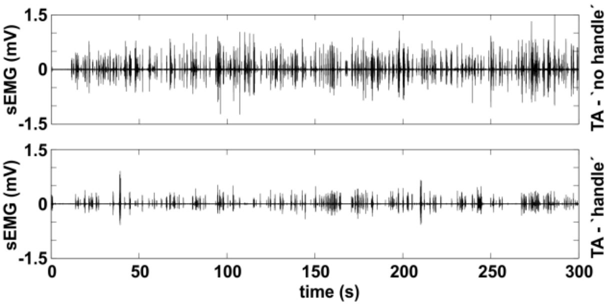

Fig. 3. Raw sEMG signals of TA muscle in sessions with and

In further analysis of sEMG, we included only the last five repetitions in both sessions. In first ten repetitions in both sessions, the subjects were getting familiar with the perturbations in order to exclude possible sudden, unexpected reactive movements. The sEMG signals were rectified, band filtered, normalized and then integrated (iEMG), to calculate the area under the curve of the rectified sEMG signal.

The analysis of average iEMG scores of both lower leg muscles (Fig. 4) showed that there was a significant difference between ‘handle’ and ‘no handle’ conditions.

On average the iEMG of GA in ‘no handle’ condition (M = 34.9, SD = 12.5), showed significantly greater values than in ‘handle’ condition (M = 24.7, SD = 10.9), $t(11)$ = 3.9 p = .002. Even greater difference in iEMG values between the ‘no handle’ (M = 18.0, SD = 7.8), and ‘handle’ condition (M = 3.9, SD = 1.6), $t(11)$ = 6.8 p < .001 was shown for the TA muscle.

These results suggest that a supportive hand contact has a major effect on lower leg muscles. Specifically, our results suggest that, when standing with additional supportive contact, lower leg muscles can be more relaxed and that the body possibly utilizes other available stabilizers (i.e. hand and/or trunk muscles).

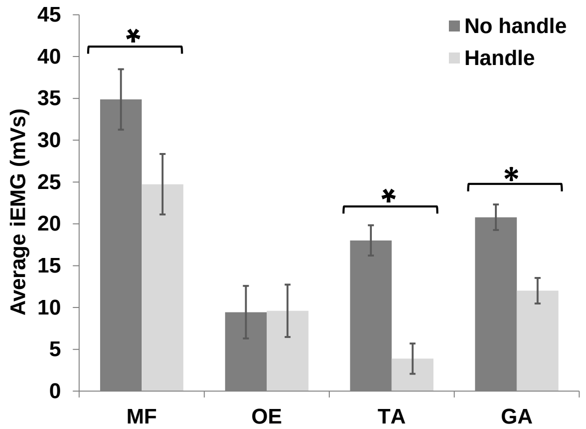

Fig. 4. Average iEMG of last five repetitions. The error bars represent SEM. Significant differences are indicated * (p < .05).

However, in the trunk muscles, the results were less distinctive. The significant difference in iEMG scores between ‘handle’ (M = 24.7, SD = 10.9) and ‘no handle’ (M = 34.9, SD = 12.5) condition was shown only for the MF muscle, $t(11)$ = 4.9, p < .001.

The results of the activity of OE did not show any significant difference. The values in ‘handle’ (M = 9.6, SD = 5.3) and ‘no handle’ (M = 9.4, SD = 6.3) conditions were almost the same and therefore proven to be non-distinctive $t(11)$ = -.1, p =.916. This indicates a constant activation of these muscles, regardless of whether the subject was using the handle or not.

4 Discussion

Reactive balance control during standing and locomotion is already well studied and documented [2] [10] [13] [15], but the findings about corrective movements of the continually perturbed balance are limited. The main aim of this study was to find out the actual role of the supportive hand contact and its influence to other postural muscles.

Results revealed the undisputed evidence that the muscles of the lower leg and back are very much affected by the additional support. Even though the perturbation was mild, all of the subjects transferred the balance corrections to the hand, thus lowering the activity of lower leg muscles (Fig. 4). This result is something to be expected. The torques in ankles that are needed for whole body stabilization are larger compared to the ones in the elbow and shoulder. The location of the ankle joint is more distant from the COM, therefore the lever arm is longer and greater torques are generated. The production of such torques is strenuous and energy less efficient. On the other hand in case of additional hand support, where the handle is at the appropriate location (i.e. shoulder height) [13], the shoulder and elbow are closer to the COM, thus less effort for balancing is needed.

The fact that the activation of the muscles flexors of the trunk (OE) did not change could be due to the close location of the perturbation origin. Since the perturbation was applied at the waist, the activation of OE was constant in order to provide trunk stability. In cases where there was no additional hand support, the OE was in coactivation with MF. However, in cases where the subjects were able to use additional hand support the OE coactivated with the muscles of the shoulder and hand. In order to prove these assumptions we will need to do some more detailed analyses.

Even though our results show the significant differences in muscle activation between the ‘handle’ and ‘no handle’ conditions, a few more questions have arisen. Furthermore, a frequency analysis of the muscle activation in relation to the perturbation signal frequency might provide us with additional insight to postural stability. In our study we additionally observed the forces applied to the handle by the contacting hand, kinematic parameters and other kinetic data, therefore there is enough data to further investigate the stabilizing role of additional hand support.

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