Postural control in standing: role of vision and additional support.

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ABSTRACT

The purpose of the study was to investigate the availability of vision and additional support on anticipatory (APAs) and compensatory (CPAs) postural adjustments and their interaction. Eight healthy adults were exposed to external perturbations induced at the shoulder level while standing with and without holding onto a walker in full vision and while blindfolded. Electrical activity of the trunk and leg muscles and center of pressure (COP_{AP}) displacements were recorded and quantified within the time intervals typical of APAs and CPAs. The results showed that with full vision, there was no difference in both APAs and CPAs in standing with and without holding onto a walker. With subjects holding onto a walker, CPAs in standing blindfolded were comparable to CPAs in full vision: this was seen in changes in the electrical activity of most of the muscles at the individual muscle, joint, and the muscle group levels as well as in COP_{AP} displacements. The findings suggest that: (1) in conditions where vision is available, vision overrules simultaneously available proprioceptive information from the support, (2) while in conditions where vision is not available, proprioceptive information from the support or support itself could be substituted for vision. It is possible to suggest that using a non stabilizing support could be a valuable strategy to improve postural control when visual information is not available or compromised.

Keywords: Posture, Anticipatory, Compensatory, Vision, Additional support.

INTRODUCTION

Sudden perturbations applied to the human body act as a destabilizing force, resulting in the displacement of the body's center of mass (COM) beyond or closer to the boundaries of the base of support (Maki and McIlroy 1996), thus compromising balance (Macpherson et al. 1989; Henry et al. 1998). These perturbations can be broadly classified into two major types: (1) Internal and (2) External. Internal perturbations are self-initiated movements such as when reaching up to a shelf or lifting an object while standing. External perturbations are disturbances triggered from outside the body such as getting accidentally hit by another person while walking in a crowd.

Generally, the central nervous system (CNS) uses two main strategies to restore balance if it is distorted by an internal or external perturbation: (1) feedforward control, which is the anticipatory postural adjustments (APA) prior to the expected body perturbations (Belenkiy et al. 1967; Massion 1992) and: (2) feedback control, which is the compensatory postural adjustments (CPA) that are initiated by the sensory feedback signals after the perturbations (Park et al. 2004; Alexandrov et al. 2005). That is, APAs serve to minimize the displacement of the body's COM prior to a perturbation (Bouisset and Zattara 1987; Aruin and Latash 1995b), while CPAs serve as a mechanism to restore the position of COM after a perturbation has already occurred (Macpherson et al. 1989; Maki and McIlroy 1996).

The individual role of APAs and CPAs in control of posture was studied relatively extensively. Consequently, it was demonstrated that the magnitude of APAs depends on direction (Aruin and Latash 1995a; Santos and Aruin 2008) and on magnitude of a perturbation (Aruin and Latash 1996; Bouisset et al. 2000), as well as on body stability (Nouillot et al. 1992; Aruin et al. 1998; Nouillot et al. 2000). Specifically, APAs have been shown to decrease in experiments with both very stable (Nardone and Schieppati 1988) and very unstable posture (Pedotti et al. 1989; Nouillot et al. 1992; Gantchev and Dimitrova 1996; Aruin et al. 1998. Since the need for stabilizing posture is eliminated in stable posture, the requirement for APA is substantially reduced in these conditions (Nardone and Schieppati 1988). On the contrary, APAs could themselves be a potential source of perturbation in case of unstable posture and as such they are suppressed in order not to additionally destabilize posture (Aruin et al. 1998). It was also shown that APAs are affected by the characteristics of a motor action utilized to induce a perturbation (Aruin and Latash 1995b; Aruin et al. 2003; Shiratori and Aruin 2007), body configuration (van der Fits et al. 1998; Aruin 2003), and fear of falling (Adkin et al. 2002).

Previous literature reports that the CPA response depends on the direction and magnitude of the perturbation and on the dimensions of the base of support (Horak and Nashner 1986; Henry et al. 1998; Dimitrova et al. 2004; Jones et al. 2008), predictability of perturbation characteristics (Burleigh and Horak 1996), instructions (McIlroy and Maki 1993), surface contact between the body and the support (Le Bozec et al. 2008), and involvement of a secondary task such as holding an object in the hands (Bateni et al. 2004). Moreover, distinct patterns of muscle activation called the ankle or hip strategy, were described in the leg and trunk muscles in response to external perturbations induced by a sudden movement of the support surface (Horak and Nashner 1986).

While less attention was paid to the investigation of a relationship between the APAs and CPAs, it is known that predictability of the upcoming perturbation is a major factor in generation of the APAs and their effect on the subsequent CPA magnitude. Thus, it was demonstrated that blindfolded subjects do not show APAs while exposed to unpredictable perturbations such as getting hit by an object (Santos et al. 2010a; Santos et al. 2010b). However, the participants do produce a larger reactive compensatory response (CPA) aimed to restore the balance of the body after the perturbation has already occurred (Santos et al. 2010a). Although it was reported that producing stronger APAs are associated with smaller compensatory EMG activation and COP displacements after the perturbations (Massion 1992; Aruin and Latash 1995b), it is not yet completely known to what extent the association between APAs and CPAs exist (Bouisset and Zattara 1987; Le Bozec et al. 2008). The nature of the relationship between APAs and CPAs could shed additional light on the rehabilitation protocols that constantly use repeated perturbations to the trunk in neurologically impaired individuals for the balance control training (Kisner C 2007).

Few studies have examined the EMG characteristics and kinetics of a standing posture stabilized by an additional external support in response to either internal (Slijper and Latash 2000; Hall et al. 2010) or external perturbations (Hausbeck et al. 2009). However, none of the above mentioned studies investigated how the availability of visual information and additional support affect the relationship between the APAs and CPAs. Therefore, our main goal was to explore the role of additional support and visual information in maintaining balance that was disturbed by an external perturbation. In our study, subjects were exposed to the external perturbations while standing with and without holding onto a walker with full vision and while blindfolded. We hypothesized that with the addition of a walker, (1) the APA and CPA response would be reduced in the vision condition, as visual information adds up to the proprioceptive information along with body stabilization, and (2) in case of no-vision condition, where APA itself is negligible, the CPA response would be very much reduced with a walker than without it, as proprioceptive information along with body stabilization will substitute for the lack of the visual information. Moreover, the effect of a walker would be more prominent in blindfolded conditions than when vision is available.

Furthermore, the role of COP displacement *per se* in describing the relationship between the anticipatory and compensatory components of control of posture is not sufficiently described in the past literature. Therefore, we also hypothesized that the APA and the CPA response of the postural muscles will be mirrored by a parallel change in the anticipatory and the compensatory response of the COP displacement.

METHODS

Participants

Eight healthy participants (5 males and 3 females; mean ± SD: age 25 ± 4 years, body mass 69 ± 11 kg) without any known neurological or musculoskeletal disorders participated in the experiment. All the participants had normal vision and were right-handed and right-legged based on their self report on using preferential hand and leg during daily activities. The experimental procedure was approved by the Institutional Review Board of the University of Illinois at Chicago and the participants provided their informed consent.

Instrumentation

A force platform (AMTI, OR-5, USA) was used to record the ground reaction forces and the moments of forces. An accelerometer (Model 208CO3, PCB Piezotronics Inc., USA) was taped to the participant's left clavicle proximally to record the moment of the pendulum impact (see further text for details). Disposable self-adhesive electrodes (Red Dot 3M) were used to record the surface electrical activity (EMG) of the following muscles: tibialis anterior (TA), rectus femoris (RF), rectus abdominis (RA), soleus (SOL), biceps femoris (BF), and erector spinae (ES). The pairs of electrodes were placed over the right side of the participant's body over the muscles bellies. Prior to the placement of the electrodes, the skin area was cleaned with alcohol swipes. A ground electrode was attached to the anterior aspect of the leg over the tibial bone. The EMG signals were collected, filtered and amplified (10-500 Hz, gain 2000) with a commercially available EMG system (Myopac, RUN Technologies, USA). All the signals were sampled at 1000 Hz frequency with a 16-bit resolution. Customized LabView software (LabView 8.6 National Instruments, Austin, Tx, USA) was used in a desktop computer to collect the data.

Procedure

The participants were instructed to maintain upright stance while standing barefoot on the force platform with their feet shoulder width apart and in parallel. This foot position was marked on top of the platform and reproduced across the trials. The verticality of the body position was controlled using a pointer that was placed laterally to the subject at the shoulder level. The participants were positioned in front of an aluminum pendulum attached to the ceiling. The pendulum consisted of a height adjustable central rod with the distal end designed as two padded pieces positioned shoulder width apart and projected towards the participant (Fig. 1). A load (4% of the body weight of the participant) was attached to the distal end of the central rod, above the padded pieces. A rope fastened to the distal end of the central rod of the pendulum was passed through a pulley system and used to release the pendulum (for more details see (Santos and Aruin 2008)). Before its release, the experimenter secured the pendulum to a trigger at a fixed distance away from the participant (0.6 m). Then the experimenter released the trigger by pulling the rope so that the pendulum produced a uni-directional perturbation to the standing participant. The participants received the impact of the pendulum on their shoulders symmetrically and were required to maintain their balance after the perturbation.

< Fig. 1 is about here>

To examine the effect of a walker and vision on APAs and CPAs four experimental conditions were performed: (1) perturbations were induced in conditions with no walker (arms hanging loosely by the sides of the participant's body, N) and full vision (V): This condition will be called NWV, (2) the subjects were holding onto a walker with full vision available (WV), (3) perturbations were induced in conditions with no walker, i.e., with arms hanging loosely by the sides of the participant's body and no vision (NWNV), (4) holding onto a walker with no vision available (WNV). The participants were instructed to position themselves inside the walker such that the handles of the walker were close to the hip joints. The height-adjustable walker had no wheels and was positioned on the surface outside of the force plate. The subjects were also instructed not to apply force to the handles of the walker, but rather, use it merely as a point of contact. The force applied to the walker was measured in a pilot experiment involving three subjects holding onto a walker in vision and no vision conditions. The force applied to the walker was 0.32 N ± 0.1 (mean ± SD), which could not be considered as stabilizing force (Jeka and Lackner 1994; Jeka 1997). This base level of force applied to the walker increased as a result of the perturbation to 1.29 N ± 0.88 in the vision conditions and 2.47 N ± 0.66 in the no vision conditions. This small peak of force seen only after the perturbation onset could be explained by a reflective activation of arm muscles in response to the pendulum impact delivered to the shoulders. No advance warning of the impending perturbation was provided in any of the conditions. In the vision conditions, the participants were asked to look straight ahead at the load attached to the pendulum and as a result subjects were fully aware of the moment of the pendulum release. In no-vision conditions (NWNV and WNV), the participants wore the eye cover and earphones playing music to prevent them from obtaining visual or auditory information about the moment of the pendulum release. Since the mass of the pendulum and the distance from which it was released were the same, similar perturbations were induced in all the conditions. Prior to data collection, all the participants received two practice trials in each experimental condition to familiarize with the perturbation and to make sure the magnitude of the perturbation was large enough to evoke compensatory feet-in-place reactions. Five trials were performed in each experimental condition and the order of the condition was randomized across each participant. The rest intervals between trials within a condition were 5s. There was a 3-min rest interval between the conditions to avoid fatigue. In addition, rest period was given to the participant whenever they asked for.

Data processing

All signals were processed offline using customized Matlab 7.6 software (MathWorks, Natick, MA). EMG signals were rectified and filtered with a 50 Hz low-pass, 2nd order, zero-lag Butterworth filter, while the reaction forces and the moments were filtered with a 20 Hz low-pass, 2nd order, zero-lag Butterworth filter. The accelerometer signal was corrected for offset and ‘time-zero’ (T_0=0) was calculated by a computer algorithm as a point in time at which the signal exceeded 5% of the maximum acceleration. This value was confirmed by visual inspection by an experienced researcher. Data in the range from -1000 ms (before T_0) to +1000 ms (after T_0) were selected for further analysis. Aligned trials within each condition were averaged for each subject.

The muscle latency (beginning of activation/inhibition) was detected in a time window from -250 ms to +250 ms in relation to T_0 by a combination of computer algorithm and visual inspection of the averaged trials. To identify the baseline, mean and standard deviation (SD) of the EMG data were calculated from -500 ms to -400 ms before T_0. The latency for a specific muscle was defined as the instant lasting for at least 50 ms when its EMG amplitude was greater (activation) or smaller (inhibition) than the mean ± 2 SD of the baseline. The latencies of the muscles were categorized and averaged by the following levels: (1) distal muscles (TA and SOL), (2) intermediate muscles (RF and BF), and (3) proximal muscles (RA and ES).

The averaged for 5 trials EMG signals were integrated (Int_{EMGi}) with 150 ms time windows. The two epochs were: (1) from -100 ms to +50 ms (anticipatory reactions, APA); (2) from +50 ms to +200 ms (compensatory reactions, CPA). The APA interval of integration was selected because APAs had been previously documented starting no earlier than 100-150 ms prior to a predictable perturbation or action initiation (Slijper and Latash 2000; Santos et al. 2010a). Each of the epochs was further corrected by the averaged 150 ms baseline activity time window of EMG integral from -1000 ms to -850 ms in relation to T_0.

$$Int_{EMG_i} = \int_{150}^{0} EMG - \left(\int_{-1000}^{-850} EMG \right) \qquad (1)$$

Int_{EMGi} is the integral of EMG activity of muscles inside each 150 ms which is corrected by the baseline activity. Then the Int_{EMGi} data were normalized to peak activity across all the conditions. This was done for each muscle for each subject.

$$IEMG_{NORM} = \frac{Int_{EMG_i}}{IEMG_{\max}} \qquad (2)$$

Note that positive values indicate an activation of the muscle, while negative values indicate a decrease in the background activity (inhibition) and due to the normalization, all the $IEMG_{NORM}$ values are within the range from +1 to -1 for each 150 ms epochs.

In equation 1, C and R values were calculated at different joint and muscle group levels (Feldman 1986; Slijper and Latash 2004). C represents co-activation (sum) and R represents reciprocal activation (difference) of (i) agonist-antagonist muscles at a joint level or (ii) dorsal-ventral muscles at a muscle group level. We expect to see an increase in C values in the case of increased co-contraction and an increase in R value in the case of increased reciprocal activation (Slijper and Latash 2004; Li and Aruin 2007).

i. Joint level (agonist-antagonist)

$$\begin{aligned} C_{TA+SOL} &= \int EMG_{TA} + \int EMG_{SOL} & R_{TA-SOL} &= \int EMG_{TA} - \int EMG_{SOL} \\ C_{RF+BF} &= \int EMG_{RF} + \int EMG_{BF} & R_{RF-BF} &= \int EMG_{RF} - \int EMG_{BF} \\ C_{RA+ES} &= \int EMG_{RA} + \int EMG_{ES} & R_{RA-ESL} &= \int EMG_{RA} - \int EMG_{ES} \end{aligned} \qquad (3)$$

ii. Muscle group level (dorsal-ventral)

$$C_{D+V} = \sum \int EMG_{TA,RF,RA} + \int EMG_{SOL,BF,ES} \qquad (4)$$

$$R_{D-V} = \sum \int EMG_{TA,RF,RA} - \int EMG_{SOL,BF,ES}$$

Time-varying COP_{AP} was calculated using the following approximation (Winter et al. 1996).

$$COP_{AP} = \frac{Mx - (Fy * dz)}{F_Z} \tag{5}$$

Where Mx is the moment in sagittal plane, Fz and Fy are the vertical and the anterior-posterior components of the ground reaction force, and dz is the distance from the origin of the platform to the surface (0.038 m). As the perturbations were induced symmetrically, only COP displacements in the anterior-posterior direction (Y-axis according to our experimental set-up) will be reported. The COP_{AP} data were shifted 50 ms forward to account for the electro-mechanical delay (Cavanagh and Komi 1979; Corcos et al. 1992). We calculated the magnitude of COP_{AP} displacement at T_0 (A_ΔCOP), which is anticipatory in nature and the peak displacement of the COP (C_ΔCOP) that is compensatory in nature. Please note that the greater the value of peak displacement during the compensatory postural control, the larger the body perturbation.

Statistical analysis

Two way repeated measures analysis of multi-variance (MANOVA) with factors (1) condition (4 levels: NWV, WV, NWNV, WNV) and (2) epochs (2 levels: APA, CPA) were used to compare the $IEMG_{NORM}$ of all six muscles. Significant MANOVAs were followed up by uni-variate ANOVAs. Two way repeated measures ANOVA with factors (1) condition (4 levels: NWV, WV, NWNV, WNV) and (2) epochs (2 levels: APA, CPA) were used to compare the C and R values at joint level and at muscle group level and the COP_{AP} displacement. For the

COP_{AP} displacement, the factors were A-ΔCOP (displacement at T_0) and C-ΔCOP (peak displacement). For the latencies of the EMG activity, 2-way repeated measures ANOVA with factors of condition (4 levels: NWV, WV, NWNV, WNV) and group (3 levels: distal muscles, intermediate muscles, proximal muscles) was used. Post-hoc analysis with Bonferroni correction was used for further comparisons within the four conditions. In all the repeated measures ANOVA, whenever the Mauchly's test of sphericity was not met, Greenhouse-Geisser correction was made. The statistical significance was set at $p<0.05$ in all the tests. Statistical analysis was performed in SPSS 17 for Windows XP (SPSS Inc., Chicago, USA). Data are presented in the text and figures as means and standard errors.

RESULTS

EMG profiles

Fig.2 shows the EMG traces (averaged across five trials) from the ventral (TA, RF, RA) and dorsal (SOL, BF, ESL) muscles of a participant during all the four experimental conditions. Note the presence of visible APAs in the NWV and WV conditions, while APAs in the NWNV and WNV conditions are negligible. Also note a much larger compensatory reaction in the NWNV and WNV conditions

< Fig. 2 is about here>

Latency of EMG activity

Fig.3 shows the muscle latencies calculated for the three muscle levels (distal, intermediate and proximal) across all the participants for the four experimental conditions. In conditions with full vision, all the muscles became active well before the moment of perturbation. When vision was not available, the muscles became active after the perturbation. A 2-way repeated measures ANOVA (condition * level) applied on the muscle latencies revealed a significant main effect of condition $F(1.6, 11.4)=77.5$, $p<0.001$, with no main effect of level and interaction. In the subsequent post-hoc analysis, NWV and WV conditions show muscle latencies at about -150 ms before the pendulum impact, suggesting the existence of the anticipatory activity (namely, APA), while the NWNV and WNV conditions show an increased level of EMG activity only after the pendulum impact (no APAs).

< Fig. 3 is about here>

EMG integrals

Integrals of EMG activity ($IEMG_{NORM}$) for all the muscles (TA, RF, RA, SOL, BF, ES) in each of the two epochs are shown in Fig. 4. Note that the integrals of EMG for the NWNV and WNV conditions are negligible during the APA epoch with huge integrals of EMG seen during the CPA epoch. At the same time, integrals of EMG in the NWV and WV conditions are of a moderate magnitude during the APA epoch with reduced magnitudes of EMG integrals during the CPA epoch. Repeated measures 2-way MANOVA applied to all six muscles revealed a significant main effect of condition [Wilks' Lambda=.07, $F_{(18,45.7)}=3.9$, $p<0.001$, $\eta^2=.58$], epochs [Wilks' Lambda=.00, $F_{(6,2)}=2018.8$, $p<0.001$, $\eta^2=1$] and interaction [Wilks' Lambda=.06, $F_{(18,45.7)}=4.2$, $p<0.001$, $\eta^2=.60$]. The results of the univariate analyses for all muscles are presented in Table 1. The univariate analysis results show that except SOL, all other muscles show significant interaction, suggesting that in blindfolded conditions, the EMG activity in CPA epoch is large and the presence of a walker reduces the CPA response only when there is no vision (in the WNV condition).

< Fig. 4 is about here>

<Table 1 is about here>

EMG analysis at a joint level

The way trunk and leg muscles were activated was analyzed using C and R values (see Methods). The sums of integrals of EMG for muscle pairs that suggests the existence of the co-activation of muscles at each joint level is illustrated in Fig.5A. Table 2A presents the results of repeated measures ANOVA. The significant interaction for all the C values suggests the following: (1) C values in the NWV and the WV conditions were larger during the APA epoch and were smaller during the subsequent CPA epoch, while C values in the NWNV and WNV conditions were negligible during APA epoch and they were larger during the CPA epoch, and (2) the presence of a walker decreased the C values during the CPA epoch in the WNV but not in the WV condition. In addition, there was a significant main effect of epochs (CPA epoch was significantly larger than the APA epoch (all $ps<0.05$)) and conditions. Specifically, the post-hoc tests showed the following for the muscle pairs: (1) $C_{(TA+SOL)}$: NWNV and WNV conditions showed larger co-contraction than the NWV and WV conditions, (2) $C_{(RF+BF)}$: no difference was seen between all the conditions, (3) $C_{(RA+ES)}$: NWNV had significantly higher co-contraction values when compared to NWV and WV ($ps<0.05$), while the WNV was comparable to the NWV and WV ($p=0.6$ and 0.1, respectively).

< Fig. 5 is about here>

<Table 2 is about here>

The difference of integrals of EMG for muscle pairs that represents reciprocal activation of agonist-antagonist muscles at each joint level is shown in Fig. 5B. Table 2B presents the results of repeated measures ANOVA. There was no significant interaction seen at all the joint levels, with $R_{(TA\text{-}SOL)}$ and $R_{(RA\text{-}ES)}$ showing a significant main effect of condition and epoch respectively. The post-hoc analysis of the $R_{(TA\text{-}SOL)}$ did not show any differences between the conditions, while $R_{(RA\text{-}ES)}$ had a greater EMG activity in the CPA epoch than the APA epoch.

EMG analysis at a muscle group level

Similar patterns were seen when comparing the $IEMG_{NORM}$ calculated for the ventral muscles groups (TA, RF, RA) against the $IEMG_{NORM}$ obtained for the dorsal muscle groups (SOL, BF, ESL) (Fig.6). For the summed activity of the dorsal and the ventral muscles (C value), there was a significant interaction along with main effects of condition and epoch (Table 2A). Specifically, the interaction highlighted that with the addition of vision, the C values increased in the APA epoch and decreased in the CPA epoch, while a contrast was seen between the NWNV and WNV conditions. In addition, the C values for the CPA epoch decreased in the WNV conditions and not in the WV conditions when compared to their respective no walker conditions. The post-hoc analysis showed vision conditions were significantly different from the no-vision conditions ($ps<0.05$) and the CPA epoch was greater than the APA epoch ($p<0.001$). The difference between the summed activity of the dorsal muscles and the ventral muscles (R value; Table 2B) showed only a main effect of condition and epoch and not of interaction. The post-hoc analysis did not show any differences between the conditions, while APA epoch was smaller than the CPA epoch ($p<0.01$).

< Fig. 6 is about here>

COP_{AP} displacement

Fig. 7 shows the COP_{AP} traces averaged across five trials in all four of the experimental conditions. There is a clear lack of anticipatory COP_{AP} displacement during the experiments with no vision, which results in a large COP_{AP} displacement after the perturbation. In the vision conditions, clear anticipatory COP_{AP} displacements are seen, resulting in lesser compensatory COP_{AP} displacements. Moreover, the availability of the walker in the no vision condition allow for minimization of the COP_{AP} displacement after the perturbation when compared to the no walker condition. In addition, within vision conditions, when the walker is available, the anticipatory COP_{AP} displacement is smaller.

< Fig. 7 is about here>

The averaged across subjects COP_{AP} displacements calculated for the four experimental conditions are shown in Fig. 8. Small anticipatory changes in the COP_{AP} displacement (A-ΔCOP) are observed in the NWV and WV (vision) conditions. Negligible anticipatory changes in the COP_{AP} displacement could be seen during the no-vision conditions (NWNV and WNV) and there was no difference in the magnitudes of the anticipatory COP_{AP} displacements between the two conditions. The compensatory changes in COP_{AP} displacement, measured as a peak displacement of the COP_{AP} (C-ΔCOP), were the highest in the NWNV condition. Repeated measures ANOVA showed a significant interaction ($F_{3,21}=65.6$, $p<0.001$) along with a main effect of condition ($F_{3,21}=11.8$, $p<0.001$) and epoch ($F_{1,7}=124.3$, $p<0.001$). These results are in line with the results on changes in the EMG integrals presented above. Of particular importance would be the results of the post-hoc analysis of changes in the COP_{AP} displacements between the four experimental conditions. In blindfolded condition, addition of a walker reduced the compensatory COP_{AP} displacement significantly ($p<0.01$), while the COP_{AP} displacement did not significantly differ between the WV and WNV conditions ($p=1.0$).

< Fig. 8 is about here>

DISCUSSION

The goal of the study was to investigate the role of additional support and vision while maintaining balance in response to an external perturbation. We were interested to know the difference between the vision and no-vision conditions when walker is used, and how much the presence of APA contributed to the attenuation of CPA response. Specifically, we wanted to know how the availability of vision and a walker affected the generation of anticipatory and compensatory EMG activity in the leg and trunk muscles at the individual muscle level, joint level, and muscle group level. We also were interested to know if the changes in the COP_{AP} displacements reflected the condition-related changes in the EMG activity. To answer these questions, four experimental conditions were implemented: standing with no walker and full vision, holding onto a walker with full vision, standing with no walker and no vision, and holding onto a walker with no vision. The results of the study showed that with the addition of a walker, (1) APA response was reduced slightly but this reduction was not statistically significant and there was no change in CPA response in vision condition, and (2) in no-vision condition, where APA itself is negligible, the CPA response was reduced and was comparable to that of the vision conditions. This pattern was seen in the EMG integrals of the most of the muscles while calculated either at the muscle level or at the joint and the muscle group levels. In addition, the changes in the displacement of COP_{AP} across the four experimental conditions resembled the pattern of changes in the EMG integrals.

EMG and COP analysis

The patterns of muscle activation seen in the current study are very much in line with that previously described pattern of the individual muscles activity only with respect to co-activation, and not to the reciprocal activation. Specifically, with the addition of a walker, the co-contraction of muscles (that is associated with the increased joint stiffness (Blaszczyk et al. 1997; Bleuse et al. 2006)) in the compensatory phase was reduced significantly only in no-vision condition and not in vision condition. Apart from the joint level, we also analyzed R and C values at the muscle group level, thus looking at the effect of activation of the dorsal and ventral muscle groups on the stabilization of the COP_{AP} (Slijper and Latash 2000; Li and Aruin 2007). Once again our analysis showed that a co-activation of all muscles at the muscle group level was reduced with a walker only in the no-vision and not in the vision condition during the CPA phase, while the changes in the reciprocal activation did not show any significant differences between experimental conditions. Taken together, these observations suggest that the CNS controls the co-activation rather than the reciprocal activation of muscles to compensate for the lack of stability induced by the external perturbations. The reason for choosing the less efficient but safer co-activation strategy to increase the body stability (Tesh et al. 1987; Cholewicki et al. 1997) could be the nature of the perturbation itself which is challenging and could increase the danger of falling. Indeed, co-activation strategy (that allows increasing the muscle stiffness and as such stabilizing the joints) is commonly used by the elderly (Blaszczyk et al. 1997; Bleuse et al. 2006) and individuals with neurological disorders (Aruin and Almeida 1997; Garland et al. 1997; Massion et al. 1999) as it is believed that the CNS deliberately utilizes the co-activation strategy to overcome the limitation associated with age or disease.

The changes in the COP_{AP} magnitudes were robust and mirrored a pattern of the changes in the EMGs. The anticipatory changes in the position of the center of pressure (A-ΔCOP) were reduced in the vision condition with a walker adding kinetic evidence to the findings of the past EMG studies showing that lesser APAs are seen when an additional support is provided (Nardone and Schieppati 1988). It is important to emphasize that APAs could be defined not only as early changes in the EMGs, but also as early changes in certain biomechanical variables described within kinematics and kinetics data (Bouisset and Do 2008). Our study produced similar findings with respect to kinetic as well as to EMG data reinforcing previously described attenuation of APAs in conditions with additional support. The compensatory changes in the position of the center of pressure (C-ΔCOP), as expected, were greatest in the no-vision condition without the walker. However, the most prominent finding here was that when the blindfolded subjects were provided with the walker, the C-ΔCOP magnitude was comparable to that of the conditions with the full vision.

Role of vision in anticipatory control of posture

When APAs were utilized in experiments with full vision, the compensatory EMGs during the CPAs and compensatory COP_{AP} displacements were substantially smaller when compared to the no-vision conditions. At the same time, APAs in the blindfolded conditions were negligible due to the unpredictability of the forthcoming perturbations. Thus APAs (1) are produced in vision conditions, (2) reduce the magnitude of the CPA response, and (3) are negligible in blindfolded conditions. The results are in line with our previous studies describing how body balance is controlled in the presence or absence of information about the forthcoming perturbation (Santos et al. 2010a; Santos et al. 2010b). However, our past and present studies describe the relationship between APAs and CPAs only in the full vision and no-vision conditions. The APA-CPA relationship, however could be affected by partial changes in the vision conditions that are commonly seen in individuals with diabetes mellitus or glaucoma (Dhital et al. 2010). As such, studies are needed to explore the relationship between APAs and CPAs in conditions with less than 20/20 vision or when vision is blurred.

Role of walker in conditions with full vision

The results showed that when vision was available in conditions with a walker or without it, there was no statistically significant difference in the indexes of electrical activity of most of the muscles at the individual muscle, joint, and the muscle group levels as well as in COP_{AP} displacements in each APA and CPA epochs. This could be due to a special role of vision in balance maintenance. Indeed, it is known that humans perceive their body location in space based on visual, vestibular and proprioceptive information (Holmes 1911; Graziano and Gross

1998). However, when vision is available, a limb position is mainly estimated by the visual information alone (Smeets et al. 2006). This suggests that there might be a "visual dominance over the proprioceptive information" (van Beers et al. 1996; Botvinick and Cohen 1998; van Beers et al. 1998), such that the additional proprioceptive information did not play a big part in reducing the APA or CPA response when vision was available. Therefore, it looks like that in conditions where vision is available, it could overrule simultaneously available proprioceptive information and/or additional support provided by the walker.

Role of walker in blindfolded conditions

The addition of a walker in no-vision condition was associated with the reduction of the CPAs despite the fact that APAs were negligible. One possible explanation for such a reduction of the CPA magnitude in the blindfolded conditions with a walker would be that a lack of vision was substituted by the walker. Nevertheless, a number of other factors could have contributed to the observed reduced CPAs in the no-vision condition with the added walker. First, body stabilization could have played an important role. We believe that this was not the case since the magnitude of force applied to the walker was in the range of 0.32 N ± 0.1 (mean ± SD). Such a force applied to the walker could not be considered a stabilizing force based on previous literature (Jeka and Lackner 1994; Jeka 1997). In addition, holding onto the walker did not reduce the CPA response in vision condition; this provides evidence against the possibility of body stabilization contributing for the reduced CPA response in the no-vision condition. Second, the decrease in the CPA response in the condition with no vision and the walker could be due to the effect of the increased APAs. However this is an unlikely possibility because APA responses in no-vision conditions were either negligible or there was no APA response at all. Therefore, it looks as if the only other factor that is present in no-vision condition while using a walker is the additional proprioceptive information obtained from the hands that are in contact with the walker. In fact, studies have shown that improvement in body stability produced by the addition of proprioceptive information could be comparable to that of the improvement produced by vision or sensory inputs from the feet (Fitzpatrick et al. 1994; Slijper and Latash 2000; Rogers et al. 2001). Moreover, studies showed that applying a light finger touch to a stationary object while standing could reduce body sway in healthy and blind individuals (Jeka et al. 1996). Thus, the observed reduction of the CPA magnitude in the blindfolded conditions with a walker could indeed be due to the effect of auxiliary proprioceptive information substituting for a lack of visual information.

Conclusion

The results demonstrated the importance of strong anticipatory adjustments in minimization of compensatory correction while maintaining posture in full-vision conditions with and without using a walker. Moreover, using a walker in blindfolded conditions was associated with smaller compensatory EMGs and displacements of the center of pressure while compared to the same visual conditions without holding a walker. This suggests that using a non stabilizing support such as a walker could be a valuable strategy to improve postural control when visual information is not available or compromised.

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Figure Captions

Fig.1. Schematic representation of the experimental setup. The subjects were exposed to the external perturbations while standing with eyes open and closed and with holding onto a walker and without it. *l* is the length and *m* is a 4% of subject's body weight additional mass.

Fig.2. EMG traces for one of the subjects (average of five trials) obtained from the ventral (TA, RF, RA) and dorsal (SOL, BF, ES) muscles during the four experimental conditions. The vertical line (T_0) represents the onset of the perturbation and the point of alignment. Note the existence of the ES inhibition during the APA epoch in the NWV and WV conditions which is difficult to see due to a need to keep the same scale across all the conditions. Time scales are in milliseconds and EMG scales are in arbitrary units. Muscle abbreviations: TA – tibialis anterior, SOL – soleus, RF – rectus femoris, BF – biceps femoris, RA – rectus abdominus, ES – erector spinae.

Fig.3. Averaged across all subjects latency of EMG activity shown for the four experimental conditions separately for three muscle levels (distal: TA & SOL, intermediate: RF & BF and proximal: RA & ES). Note that for vision conditions, the muscles are active prior to the perturbation (T_0), while the activation of muscles occurred after the perturbation onset in the no-vision conditions.

Fig. 4. Mean and standard error bars of $IEMG_{NORM}$ for APAs (-100 to +50 ms in relation to T_0) and CPAs (+50 to +200 ms in relation to T_0) are shown for all the muscles in the four

experimental conditions. EMG scales are in arbitrary units. * denotes significant changes in post-hoc results.

Fig 5A) Mean and standard error bars of the c values that reflect the co-activation of muscle pairs at the joint level. The c values are in arbitrary units. 5B) Mean and standard error bars of the r values that reflect the reciprocal activation of muscle pairs at the joint level. The r values are in arbitrary units. * denotes significant changes in post-hoc results.

Fig. 6. Mean and standard error bars of the co-activation (C) and the reciprocal activation (R) of muscle pairs at the muscle group levels. Both the C and R values are in arbitrary units. * denotes significant changes in post-hoc results.

Fig. 7. COP_{AP} traces for one subject (average of five trials) during the four experimental conditions. The vertical line (T_0) represents the onset of the perturbation and the point of alignment. Time scale is in milliseconds, COP_{AP}s are in m, and the negative values correspond to displacements of the COP_{AP} backward. Note the existence of an anticipatory shift in COP_{AP} displacement and a lesser compensatory displacement in the vision conditions when compared to the no vision conditions. Additionally, in the no vision condition, the presence of a walker reduced the compensatory COP_{AP} displacements.

Fig. 8. Mean and standard errors of the displacement of the center of pressure (A-ΔCOP) at the time of perturbation (T_0) and the peak of center of pressure displacement (C-ΔCOP) are shown for the four experimental conditions. COP_{AP} magnitudes are in m and the negative values correspond to displacements of the COP_{AP} backwards. * denotes significant changes in post-hoc results.

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