

# Influence of an unstable shoe on compensatory postural adjustments: An experimental evaluation

Andreia S. P. Sousa

*Centro de Estudos de Movimento de Actividade Humana, Escola Superior de Tecnologias da Saúde do Instituto Politécnico do Porto, Portugal, [asp@estsp.ipp.pt](mailto:asp@estsp.ipp.pt)*

Rui Macedo

*Centro de Estudos de Movimento de Actividade Humana, Escola Superior de Tecnologias da Saúde do Instituto Politécnico do Porto, Portugal, [rmacedo@estsp.ipp.pt](mailto:rmacedo@estsp.ipp.pt)*

Rubim Santos

*Centro de Estudos de Movimento de Actividade Humana, Escola Superior de Tecnologias da Saúde do Instituto Politécnico do Porto, Portugal, [rss@estsp.ipp.pt](mailto:rss@estsp.ipp.pt)*

João Manuel R. S. Tavares

*Faculdade de Engenharia da Universidade do Porto, Departamento de Engenharia Mecânica / Instituto de Engenharia Mecânica e Gestão Industrial, Portugal, [tavares@fe.up.pt](mailto:tavares@fe.up.pt)*

**ABSTRACT:** This study attempted to evaluate the influence of using an unstable shoe in muscle recruitment strategies and center of pressure (CoP) displacement after the application of an external perturbation. Fourteen healthy female subjects participated in this study. The electromyographic activity of medial gastrocnemius, tibialis anterior, rectus femoris, biceps femoris, rectus abdominis and erector spinae muscles and the kinetic values to calculate the CoP were collected and analyzed after the application of an external perturbation with the subject in standing position, with no shoes and using unstable footwear. The results showed increased in medial gastrocnemius activity during the first compensatory postural adjustments and late compensatory postural adjustments when using an unstable shoe. There were no differences in standard deviation and maximum peak of anteroposterior displacement of CoP between measurements. From the experimental findings, one can conclude that the use of an unstable shoe leads to an increase in gastrocnemius activity with no increase in CoP displacement following an unexpected external perturbation.

**Keywords:** Postural control strategies, Electromyography, Center of pressure, Masai Barefoot Technology.

## 1 INTRODUCTION

The human postural control system manages body position in space in order to promote balance and orientation, based on the central integration of proprioceptive, visual and vestibular information and an internal representation of body orientation in space. The internal model of body position is continually updated based on this multi-sensorial feedback that is used to create motor commands to control body position in space, taking into account environmental constraints [Massion, 1994; Mergner, 1998].

Any perturbation, either external, such as a sudden change in the base of support, or internal, such as a rapid movement of the upper and lower extremities, changes the projection of the center of mass (CM) closer to the limits of the base of support and the alignment between the CM and the center of pressure (CoP), which can result in postural imbalances. To minimize the danger of loss of balance, the central nervous system uses anticipatory postural

adjustments (APA) in the form of feedforward mechanisms prior to the imbalance [Aruin, 1995b; Belenkiy, 1967; Li, 2007; Massion, 1992], and compensatory postural adjustments (CPA) that are initiated by sensory feedback signals [Alexandrov, 2005; Park, 2004].

There are different balance strategies. The most common strategy of movement in response to a forward imbalance is the ankle strategy, which involves shifting the CM by rotating the body about the ankle joints with minimal movement of hip and knee joints. The hip strategy changes the CM position through flexion or extension of the hip. A stepping strategy realigns the base of support under the center of body mass with rapid steps towards the external source of perturbation [Horak, 1987]. The use of each strategy depends on the configuration of the base of support and on the intensity of the perturbation. Postural adjustments occur not only due to sensory feedback in response to unexpected external perturbations but also as a consequence of feedforward in anticipation of expected disruptions. Main-

taining posture on unstable bases of support requires higher levels of the control system and a fundamental change in the mode of using proprioceptive information [Ivanenko, 1997].

Maintaining balance in the standing position has been described as an effective method for the rehabilitation [Wester, 1996] and prevention of musculoskeletal injuries [Bahr, 1997; Caraffa, 1996; Wedderkopp, 1999]. The Masai Barefoot Technology (MBT), an unstable shoe, aims to promote continuous stability training. This study aims to evaluate the influence of using an unstable shoe, MBT Sport Black model, USA, on kinetic and electromyographic parameters during CPA following an external perturbation.

## 2 METHODOLOGY

### 2.1 *Subjects*

Fourteen healthy female individuals were tested (age =  $34.6 \pm 7.7$  years, body weight =  $65.3 \pm 9.6$  kg, height =  $1.59 \pm 0.06$  m and Q angle =  $15.14 \pm 0.79$  degrees; mean  $\pm$  S.D.), being excluded subjects presenting one or more of the following conditions: 1) history of recent musculoskeletal injury in the lower limbs [Lord, 1994], 2) history or signs of neurological dysfunction that could affect motor performance, balance and sensory afferents [Lord, 1994; Ramstrand, 2010], 3) history of surgery of the lower limbs, 4) presence of pain in the legs and lower trunk in the 12 months preceding the study [Ramstrand, 2010; Tinetti, 1988], 5) cognitive changes [Lord, 1994], 6) individuals under the influence of medication, 7) balance disorders and visual deficits, 8) individuals with experience of using unstable footwear prior to the study [Ramstrand, 2010], 9) individuals with abdominal skinfold thickness exceeding 0.2 cm.

All trials were performed using the dominant limb, which was identified by asking subjects to kick a ball [Keating, 1996]. In all individuals, the right lower extremity was the dominant member.

The study was conducted according to the involved Institutions' ethical norms and conformed to the Declaration of Helsinki, dated 1964, being informed consent obtained from all participants.

### 2.2 *Instrumentation*

A Biopac Systems, Inc. – MP 100 Workstation™ (Biopac Systems, Inc. 42 Aero Camino Goleta, CA 93117) was used to collect all electromyographic (EMG) data, which were sampled at 2000 Hz with a bandpass filter between 10 and 500 Hz, amplified (common mode rejection ratio (CMRR)  $>110$  dB, gain = 1000) and analogical-to-digital converted (12 bit). Data were collected on tibialis anterior (TA),

medial gastrocnemius (MG), rectus femoris (RF), biceps femoris (BF), rectus abdominis (RA) and erector spinae (ES) muscles using steel surface electrodes (TSD150, from BIOPAC Systems, Inc. (USA)), with bipolar configuration, a 11.4 mm contact area and an inter-electrode distance of 20 mm, and a ground electrode. This equipment presents good reliability and validity [Soderberg, 2000].

CoP values were obtained from a force plate, model FP4060-10 from Bertec Corporation (USA), connected to an amplifier with default gains and a 1000 Hz sampling rate. The amplifier was connected to a Biopac 16-bit analogical-digital converter from BIOPAC Systems, Inc. (USA). The intraclass correlation coefficient (ICC) reliability of the instrument is 0.88 [Hanke, 1992].

The magnitude of the destabilizing force induced to subjects was monitored using an isometric dynamometer (Globus Italia, Italy), ICC = 0.97-0.98 [Bohanon, 1986].

We used a caliper to measure the abdominal skinfold thickness (Harpenden Skinfold Caliper HSB-BI model, Victoria Road Burgess Hill, UK).

### 2.3 *Assessment*

Each individual was exposed to a postural stress, whose protocol was adapted from [Wolfson, 1986], being applied a forward destabilizing force at the lower trunk level with a magnitude of 4.5% of body weight.

All individuals were asked to remain upright, comfortably standing, with the base of support aligned across the width of the shoulders, upper limbs along the body, and not to take any step or elevate the heels, keeping the balance [Fiedler, 2005]. Additionally, they had to focus a target that was two meters away and at the eye level [Fiedler, 2005].

No advance warning of the impending perturbation was provided; instead, the subjects wore earphones and listened to music delivered via a mini audio player. A forward destabilizing force was applied, maintained for at least three seconds and subsequently eliminated instantaneously. Each subject performed three repetitions of the procedure.

We evaluated the electromyographic activity (EMG<sub>a</sub>) of TA, MG, RF, BF, RA and ES muscles at predetermined intervals. The integral of the EMG<sub>a</sub> during the procedure was analyzed in two epochs, each of 150 ms duration in relation to the time of application of the destabilizing force, herein designated by “time zero” (T0). The time windows for the two epochs were the following: 1) from -100 ms to +50 ms (compensatory postural adjustments 1 (CPA1)) and 2) from +200 ms to 350 ms (late compensatory postural adjustments (CPA2)). The window of CPA was chosen based on the literature data

regarding the time of corrective responses observed in the trunk and leg muscles following external perturbations, see, for example, [Henry, 1998], and following the protocol described in [Santos, 2009]. This interval was divided in two epochs to differentiate reflex responses (CPA1) from voluntary reactions (CPA2) [Latash, 2008].

The EMG<sub>a</sub> integral,  $\int_{-500}^{T_0} EMG dt$ , for each epoch was subsequently corrected by the EMG integral of the baseline activity from -500 ms and -450 ms in relation to T<sub>0</sub>:

$$(1)$$

The  $\int_{-500}^{T_0} EMG dt$  within each 150 ms epoch  $t_{wi}$ ,  $i=1, 2$ , and  $T_0$  is the 50 ms of the EMG baseline activity defined as the integral of EMG signal from -500 ms to -450 ms in relation to T<sub>0</sub> [Aruin, 1995a; Santos, 2009].

The standard deviation (SD) and maximum peak-to-peak amplitude (CoP<sub>max</sub>) of displacement of the CoP for each interval of 150 ms was calculated and corrected for baseline values between -500 ms and -450 ms. The time durations for each interval for the CoP were similar to those used to calculate the  $\int_{-500}^{T_0} EMG dt$ . However, they were shifted 50 ms forward for each epoch to account for the electromechanical delay [Cavanagh, 1979; Howatson, 2008]. This resulted in the following intervals: (1) +100 a 250 ms (CPA1); (2) +250 a 400 ms (CPA2).

## 2.4 Statistics

The data were processed using the Statistic Package Social Science (SPSS Inc., an IBM Company Headquarters, 233 USA) version 13.0. The sample characterization was performed using descriptive statistics.

The Wilcoxon test was used to examine the influence of using an unstable shoe on the degree of muscle activity recruited after the application of an external perturbation, and the T-test for paired samples was used to analyze the same influence on the parameters for the CoP. For inferential analysis, a statistical significance of 0.05 was adopted.

## 3 RESULTS

Tables 1 and 2 show mean values, SD, maximum and minimum EMG<sub>a</sub> integral of TA, MG, RF, BF, RA and ES muscles, obtained in the standing position after the application of an external perturbation, with and without MBT shoes. Both for CPA1 and CPA2, the Wilcoxon test has shown statistically significant differences between measurements obtained

with and without the shoes in MG muscle activity, Table 3.

Table 1. Mean values, SD, maximum (Max) and minimum (Min) EMG<sub>a</sub> integral of TA, MG, RF, BF, RA and ES muscles, measured with and without MBT shoes during CPA1.

Muscles	Series	Mean	SD	Max	Min
TA	Barefoot	0.00005	0.000039	0.00015	0.000001
	MBT	0.00008	0.00009	0.00038	0.000019
MG	Barefoot	0.00011	0.000063	0.00015	0.000001
	MBT	0.00049	0.000080	0.00300	0.000087
RF	Barefoot	0.00001	0.000008	0.00003	0.000006
	MBT	0.000003	0.000004	0.00017	0.000007
BF	Barefoot	0.00002	0.000037	0.00015	0.000007
	MBT	0.00003	0.000040	0.00017	0.000007
RA	Barefoot	0.00001	0.000007	0.00003	0.000005
	MBT	0.00002	0.000023	0.00008	0.000005
ES	Barefoot	0.00001	0.000005	0.00002	0.000006
	MBT	0.00003	0.000044	0.00018	0.000006

Table 2. Mean values, SD, maximum (Max) and minimum (Min) EMG<sub>a</sub> integral of TA, MG, RF, BF, RA and ES muscles, measured with and without MBT shoes during CPA2.

Muscles	Series	Mean	SD	Max	Min
TA	Barefoot	0.00004	0.000047	0.00019	0.000009
	MBT	0.00006	0.000077	0.00032	0.000014
MG	Barefoot	0.00010	0.000070	0.00004	0.000007
	MBT	0.00040	0.000680	0.00257	0.000085
RF	Barefoot	0.00001	0.000008	0.00004	0.000007
	MBT	0.00003	0.000038	0.00015	0.000007
BF	Barefoot	0.00002	0.000025	0.00010	0.000006
	MBT	0.00003	0.000038	0.00014	0.000008
RA	Barefoot	0.00001	0.000001	0.00001	0.000005
	MBT	0.00001	0.000020	0.00008	0.000005
ES	Barefoot	0.00001	0.000006	0.00003	0.000007
	MBT	0.00003	0.000039	0.00015	0.000007

Table 3. P values obtained using the Wilcoxon test to compare muscle activity between measurements taken with and without MBT shoes, during time windows (Tw) CPA1 and CPA2.

Muscles	TW	P value	TW	P value
TA	Barefoot	0.221	CPA1	0.245
	MBT			
MG	Barefoot	0.001	CPA1	0.008
	MBT			
RF	Barefoot	0.551	CPA2	0.572
	MBT			
BF	Barefoot	0.331	CPA2	0.064
	MBT			
RA	Barefoot	0.660	CPA2	0.346
	MBT			
ES	Barefoot	0.245	CPA2	0.173
	MBT			

When analyzing the displacement of the CoP (SD and CoP<sub>max</sub>) following the application of an external perturbation, it can be seen that, both in terms of SD and CoP<sub>max</sub>, there was an increase in the CoP displacement in the series performed with MBT shoes, Figures 1 and 2. However, the results ob-

tained after applying the T-test for paired samples showed no evidence of statistically significant differences in these two variables, with and without MBT shoes (PCA1,  $p=0.315$  (SD),  $p=0.331$  (CoP-max); PCA2,  $p=0.712$  (SD),  $p=0.650$  (CoPmax)).

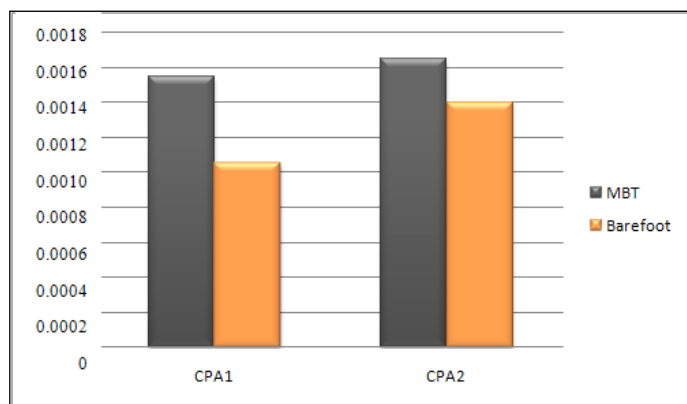


Figure 1. SD of CoP displacement after the application of an external perturbation with unstable shoes (MBT) and without shoes (barefoot).

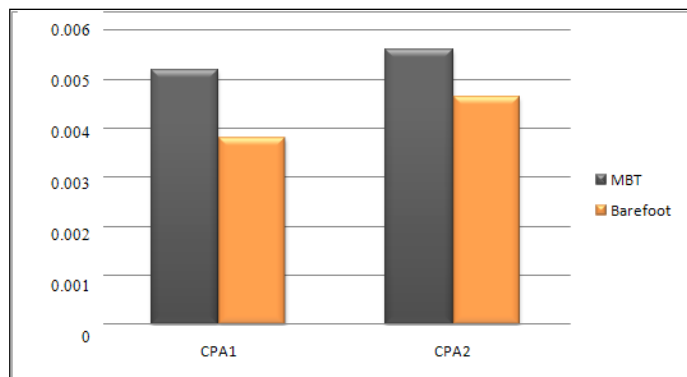


Figure 2. CoPmax displacement after the application of an external perturbation with unstable shoes (MBT) and without shoes (barefoot).

## 4 DISCUSSION

The results of this study show that when using MBT shoes, the muscle activity during CPA (CPA1, CPA2) is superior. According to [Santos, 2009], there is a relationship between APA and CPA in the control of posture and the possibility of optimal use of CPA in postural control. These findings are supported by several previous observations. Firstly, the EMG of the trunk and leg muscles during CPA may be associated with a failure in APA, as was observed in children [Hadders-Algra, 2005; van der Fits, 1998] and in individuals with neurological damage [Bazalgette, 1987]. Moreover, as already mentioned, APA are mitigated in situations of postural instability [Arruin, 1998]. Thus, in this case, compensatory muscle activity becomes necessary to maintain body equilibrium.

In terms of CPA, this study has shown an increase in activity only in the MG muscle when using

unstable shoes. According to [Ivanenko, 1997], when standing on an unstable support base, the CM deviation is accompanied by changes in ankle and plantar pressure distribution, which is compensated by triceps surae muscle activation. When standing on a movable platform, the postural pattern regulation is slightly different: usually humans do not move the CM, shifting instead the point of contact of the rocking platform with the ground under the CM, which leads to an increased need for MG activation.

While analyzing the anteroposterior Cop displacement, no differences were found in SD and CoPmax during CPA, with and without the use of unstable footwear. The values of the CoP anteroposterior displacement did not correlate with changes in MG.

According to [Shumway-Cook, 2003], the time needed to stabilize the CoP is a variable to take into account during postural adjustments. Thus, although there were no changes in terms of SD and CoPmax displacement, there might have been differences in the time needed to stabilize the CoP. Therefore, it is suggested as future work to monitor this variable in order to ascertain whether it may relate to changes in MG activity during CPA.

## 5 CONCLUSIONS

The use of unstable footwear leads to an increase in muscle activity recruited by the medial gastrocnemius muscle during compensatory postural adjustments. Additionally, no differences were observed in the tibialis anterior, rectus femoris, biceps femoris, rectus abdominis and erector spinae muscles after the application of an external perturbation.

Finally, the use of unstable footwear did not imply an increase in anteroposterior center of pressure displacement, both in terms of standard deviation and peak-to-peak amplitude in compensatory postural adjustments after the application of an external perturbation.

## 6 ACKNOWLEDGEMENT

The first author would like to thank the support and contribution of the PhD grant (with reference SFRH/BD/50050/2009) from IPP – Instituto Politécnico do Porto, ESTSP – Escola Superior de Tecnologia da Saúde and FCT – Fundação para a Ciência e a Tecnologia, in Portugal.

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