Available online at www.sciencedirect.com
SCIENCE DIRECT
ELSEVIER
Clinical Biomechanics 21 (2006) 82–88
CLINICAL BIOMECHANICS
www.elsevier.com/locate/clinbiomech

Effect of an unstable shoe construction on lower extremity gait characteristics

Benno Nigg *, Sabrina Hintzen, Reed Ferber

Human Performance Laboratory, Faculty of Kinesiology, University of Calgary, Calgary, Alberta, Canada T2N 1N4

Received 9 March 2005; accepted 17 August 2005

Abstract

Background. To compare kinematics, kinetics and muscle activity during standing and walking for healthy subjects using an unstable test shoe (Masai Barefoot Technology, MBT) and a stable control shoe.

Methods. Eight subjects volunteered for this study. During quiet standing, center of pressure excursion and muscle activity were determined. During walking, lower extremity kinematics, kinetics, and muscle electromyographic (EMG) signals were determined. Data were collected for the two shoe conditions after a 2 week accommodation period. Statistics included repeated measures ANOVAs ($\alpha = 0.05$) and post hoc tests where appropriate.

Findings. During quiet standing, the center of pressure excursion was significantly and substantially greater in the unstable compared to the control shoe. Electromyographic intensity increased in the unstable test shoe for all tested muscles, but only significantly for the tibialis anterior. During locomotion, kinematics were similar in the two shoe conditions except for the initial plantar–dorsiflexion, which showed a significant more dorsiflexed position during the first half of stance in the unstable test shoe compared to the stable control shoe. The angular impulses did not show any significant differences between the two shoe conditions for all three joints but some trends towards a reduction for the knee and hip joint. There were no significant differences in electromyographic activities between the control and the unstable shoe. However, several muscles showed some trends.

Interpretation. The unstable shoe produced changes and trends in kinematic, kinetic and electromyographic characteristics that seemed to be advantageous for the locomotor system. Further studies should investigate muscle strength, dynamic stability, pain reduction for arthritic knees and injury prevention for high performance athletes when using the unstable shoes.

Keywords: Unstable shoe; MBT; Stability training; Walking; Kinematics; Kinetics; EMG

1. Introduction

Stability of human locomotion is important for children as they learn to walk, for athletes to perform well, and for the elderly to be mobile as long as possible. Several strategies can improve stability during locomotion including shoe constructions that provide support, strengthening lower extremity muscles, altering gait mechanics and/or stability training.

Shoes for walking and other locomotion activities are typically constructed to provide stability for the user. However, by using "stable" shoes, the muscles that would usually contribute to static and dynamic stability may get weaker over time because they are very little used (Jackman and Kandarian, 2004; Spector, 1985). Unstable training devices are often used to compensate for the missing training effects improving ankle and knee strength and proprioception (Waddington et al., 2000; Waddington and Adams, 2004). Standing balance training has also been reported to be effective for rehabilitation (Wester et al., 1996) and prevention of

* Corresponding author.
E-mail address: nigg@ucalgary.ca (B. Nigg).

0268-0033/$ - see front matter © 2005 Elsevier Ltd. All rights reserved.
doi:10.1016/j.clinbiomech.2005.08.013

musculoskeletal injuries (Bahr et al., 1997; Caraffa et al., 1996; Wedderkopp et al., 1999) assuming the muscles have been strengthened.

Currently, stability and muscle training are two aspects that are addressed separately. Stability is typically provided through shoe construction. Muscle training is typically done during independent training sessions. It would be advantageous if the daily locomotion activities and the stability training could be combined. Such a training device combining the daily locomotion activities with stability training has recently been developed by Masai Barefoot Technologies, Switzerland (MBT). The MBT shoe has a rounded sole in the anterior–posterior direction (Fig. 1), thus providing an unstable base. It is speculated and supported by anecdotal evidence that these shoes act as a training device for lower extremity muscles.

The conceptual thinking behind the functioning of an unstable shoe construction relates to the strengthening of muscles that are anatomically closer to the axes of rotation. If stability is provided with large muscles (e.g. triceps surae) with a line of action further from the joint axis, joint loading will be large. If stability is maintained through smaller muscles with a line of action closer to the joint axis, these muscles can react faster and joint loading will be substantially reduced. However, joint pain is assumed to be associated with joint loading. Thus, reduced joint loading should reduce joint pain.

These considerations suggest that it may be beneficial to understand the locomotion characteristics in unstable shoes. However, a comprehensive comparison of the gait characteristics in stable and unstable shoes has, to the knowledge of the authors, not been done yet.

Thus, the purpose of this investigation was to compare lower extremity kinematics, kinetics, and electromyographic muscle activity during quiet standing and during locomotion for healthy subjects using an unstable test shoe and a conventional stable control shoe.

Specifically, the purposes were

(a) to identify the excursion of the center of force during quiet standing to provide evidence for the differences in stability,

Fig. 1. Illustration of the unstable MBT test shoe used in the study.

(b) to quantify the differences in EMG activity during standing and walking to provide evidence for changes in muscle activation,
(c) to quantify the differences in resultant ankle, knee and hip joint moments during walking to provide evidence for changes in joint loading, and
(d) to describe changes in kinematics and kinetics,

for an unstable test shoe (MBT shoe) and a stable control shoe. Measurements were performed after using the unstable test shoe intensively for 2 weeks.

The following hypotheses were tested:

Excursion of the center of mass
- H1 During quiet standing, the unstable shoe exhibits greater anterior–posterior and medio-lateral excursion of the center of force than the stable control shoe.

Kinematics
- H2 During walking, there are differences in ankle joint kinematics but no differences in knee and hip joint kinematics.

Resultant ankle, knee and hip joint moments
- H3 During walking, there are differences in the resultant ankle joint impulses between the unstable test and the stable control shoe.
- H4 During walking, there are no differences in resultant knee and hip joint impulses between the unstable test and the stable control shoe.

EMG activity
- H5 During quiet standing, the EMG intensity is higher for the unstable shoe condition compared to the stable control shoe.
- H6 During walking, the EMG intensity is lower for the unstable shoe condition compared to the stable control shoe.

2. Methods

2.1. Subject population

Eight subjects volunteered for this study, five males and three females with the mean age: 28.0 yr (SD 3.6 yr), mean mass: 70.1 kg (SD 7.5 kg) and mean height: 169.5 cm (SD 6.4 cm). All subjects were free of lower-extremity pain and injury for a minimum of 6 months prior to testing and had never used the unstable shoe before. All subjects gave informed written consent corresponding to the guidelines of the University of Calgary Ethics Committee.

According to an a priori power analysis ($\beta = 0.20$; $P = 0.05$), based on pilot data for CoP excursion, kinematic, resultant joint moments and EMG intensities, eight subjects were sufficient (Lieber, 1990). The variable that determined the minimal number was the EMG intensity. A coefficient of binomial distribution showed that for the eight subjects used in the current investigation, six subjects would have to demonstrate a minimum 10% change from the control condition to gain statistical significance at a 95% confidence level. Therefore, the sample size was sufficient for the purposes of this study.

2.2. Shoe condition

The control shoe tested in this study was the Adidas SuperNova running shoe (mass: 358 g). The control shoe was not considered a motion control shoe and was used as a representation of a commercially available shoe commonly used by the general population. The unstable shoe condition tested in the study was the MBT shoe from Masai Barefoot Technology, Switzerland (mass: 650 g). The tested MBT shoe (Fig. 1) was characterized by a rounded shoe sole design in the anterior–posterior direction, making the shoe unstable.

2.3. Testing procedure

Retroreflective markers were placed on the segments of the rearfoot, shank, thigh and pelvis of the right lower extremity. Markers were placed on both sides of the body over the greater trochanters, the medial and lateral femoral condyles and the medial and lateral malleoli. The three rearfoot markers were placed directly on the posterior and posterior-lateral aspect of the shoe heel counter. The position of the markers were determined for a standing neutral trial to define the anatomical coordinate system for each segment for both shoe conditions, first for the stable control shoe and second for the unstable test shoe.

2.3.1. Standing quiet

The standing test (Emery, 2004) consisted of three trials of 10 s each while standing on both feet on a force plate to determine the anterior–posterior and medial–lateral excursion of the center of pressure and the corresponding EMG activity.

2.3.2. Walking

The walking test consisted of 10 walking trials first for the unstable test shoe and second for the control shoe at a walking speed of 5.0 ± 0.5 km/h. For the walking trials, lower leg kinematics and kinetics and EMG data from selected muscles were collected.

The subjects were tested after a 2 week accommodation period to the unstable test shoe. When receiving the MBT shoes, subjects were instructed on the proper walking methods with the unstable test shoe by an instructor trained in the correct use of the shoe. Subjects completed approximately 5–10 min of walking in the laboratory until the instructor felt that they walked properly and the subjects felt comfortable walking in the unstable test shoe. Subjects were asked to wear the unstable test shoe as much as possible for 2 weeks. On average, the subjects wore the shoes for 9.5 h (SD 2.1 h) each day over the 2 week period. The test protocol consisted of the standing tests first followed by the walking tests. The unstable test shoe was tested first followed by the control shoe. Prior to using the control shoe during the test session, subjects were allowed approximately 5–10 min of walking to re-familiarize themselves with the normal control shoe.

2.4. Kinematic and kinetic data

Kinematic data were collected using an eight high-speed video camera system (Motion Analysis Corporation, Santa Rosa, CA, USA) at a sampling rate of 240 Hz. The pre-determined criterion for tolerable error in space calibration was set at 0.06%, corresponding to a 0.6 mm maximum error for a 1 m^3 volume. Three-dimensional marker traces were reconstructed using Expert Vision Three-Dimensional Analysis software (Motion Analysis Corporation, Santa Rosa CA, USA). Kinetic data were collected simultaneously with the kinematic data using a force platform (Kistler: Winterthur, Switzerland) that was placed in the center of the walkway level with the ground at a sampling rate of 2400 Hz. Kinematic and kinetic data were filtered using a zero-lag quadratic low-pass Butterworth filter with a cut-off frequency of 12 and 50 Hz, respectively.

The results from the ground reaction force measurements were used to determine the excursion of the center of pressure (CoP) as illustrated in Fig. 2.

From the kinematic data linear and angular positions, velocities and accelerations were determined and exported for further analysis. Kintrak software (Human Performance Laboratory, Calgary, Canada) was used to calculate kinematic and kinetic variables. Resultant joint moments and powers were determined using inverse dynamics techniques. Results were normalized to ground contact. Peak joint angles, peak ground reaction force values and angular impulses were determined for the first and second half of the stance phase for each trial and condition. The resultant joint moments were used to calculate the angular impulses, I,

$$I\ (\text{joint/axis}) = \int M \cdot \mathrm{d}t$$

for the total ground contact and/or for the first and second half of ground contact, providing a variable that describes an integrated effect of the resultant joint moments. The angular impulses were determined for the three rotational axes for the ankle, knee and hip joint.

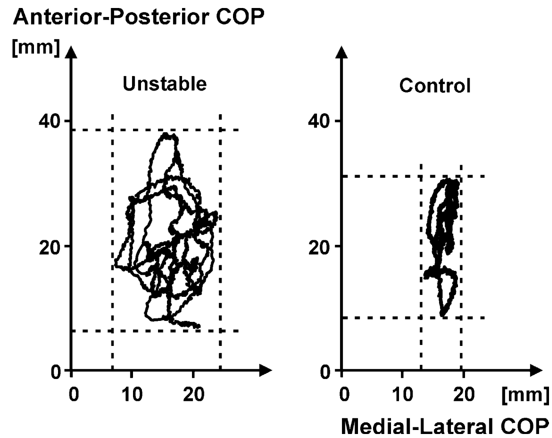

Fig. 2. Illustration of the differences in center of pressure movement during quiet standing in an unstable test shoe and a stable control shoe (one representative trial for one subject during quiet standing for 10 s). The horizontal lines illustrate the measurement of the CoP excursion.

2.5. EMG data

Myoelectric signals were recorded using round bipolar surface electrodes (Ag/AgCl: Biovision, Wehrheim, Germany). To achieve an optimal EMG signal and low impedance ($<5\ k\Omega$) hair was removed and skin cleaned with isopropyl wipes prior to the electrodes being placed on the skin. Each electrode was 10mm in diameter and had an inter-electrode spacing of 22 mm and was placed midway between the motor end plate and the distal myotendinous junction. A ground electrode was placed on the lateral condyle of the knee. Five EMG electrodes were placed on the skin overlying the muscle belly of the tibialis anterior, medial gastrocnemius, biceps femoris, vastus medialis, and gluteus medius of the limb of interest. All raw EMG analog signals were on-line pre-amplified (×7000) and then converted into digital signals sampled at 2400 Hz simultaneously with the kinematic and kinetic data. Timing of touch-down for each trial was determined from kinetic data.

EMG signals were resolved into time–frequency space using wavelet analysis technique (Spector, 1985) and then averaged across the 10 trials for each condition. The 11 wavelet intensities with center frequencies between 7 (wavelet 0) and 542 Hz (wavelet 11) were determined. The total EMG intensity was defined as the sum of the EMG wavelet intensities for wavelet domains 1 through 8. The intensity is a measure of the power of the signal. In this study, only the total EMG intensity will be discussed.

For the quiet standing condition, EMG data were normalized to the average total intensity of the control shoe condition. The total intensities were averaged over the middle 8 s of quiet standing period.

For the walking condition, the data for the total EMG intensity were normalized to the intensity of the control shoe. The intensities were determined for the time interval from 200 ms before heel strike to 500 ms after heel strike based on previous studies from our laboratory using similar techniques (von Tscharner, 2001; Waddington and Adams, 2004; Wakeling et al., 2001).

2.6. Statistical analysis

All statistical tests were performed using SPSS version 12.0.0. The effects of the unstable test shoe on kinematics, kinetics, and EMG were determined using repeated measures ANOVAs ($\alpha = 0.05$).

3. Results

3.1. Standing test

3.1.1. Center of pressure

Center of pressure excursions were significantly greater for the unstable test shoe compared to the control shoe in anterior–posterior and in medio-lateral direction (Fig. 2). The mean excursion in medio-lateral direction was 11.91 mm (SD 5.39 mm) for the MBT and 5.82 mm (SD 2.39 mm) for the control shoe. The mean excursion in anterior–posterior direction was 27.25 mm (SD 9.13 mm) for the MBT and 17.88 mm (SD 5.82 mm) for the control shoe.

3.1.2. EMG

All five measured muscles showed in the average an increase in EMG intensity. Compared to the control shoe, the specific mean increases were 70% (SD 85%) for the tibialis anterior (significant difference), 38% (SD 41%) for the gastrocnemius, 37% (SD 46%) for the vastus medialis, 11% (SD 69%) for the biceps femoris and 38% (SD 78%) for the gluteus medius. The average increase was 39%, also not significant.

3.2. Walking test

3.2.1. Kinematics

The ankle joint was significantly more dorsiflexed during the first half of stance in the unstable test shoe compared to the stable control shoe (Fig. 3). No other significant differences in lower extremity joint kinematics were observed between the unstable test and the stable control shoe condition.

3.2.2. Kinetics

The angular impulses did not show any significant differences between the two shoe conditions for all three joints (Fig. 4). The angular impulse for in-eversion showed a trend for the MBT being more towards inversion. However, the variability was large.

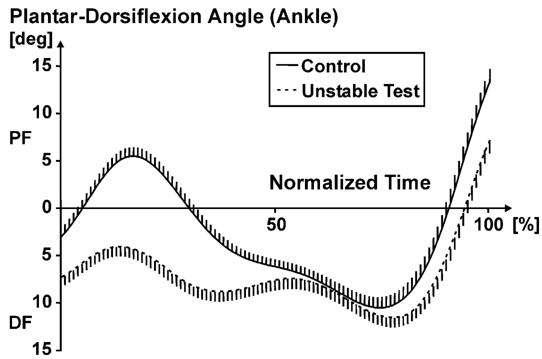

Fig. 3. Ankle plantar–dorsiflexion angles during the stance phase of walking for the control shoe and the unstable test shoe. In all graphs, the standard error bars are depicted into one direction to allow for optimal reading of the graph.

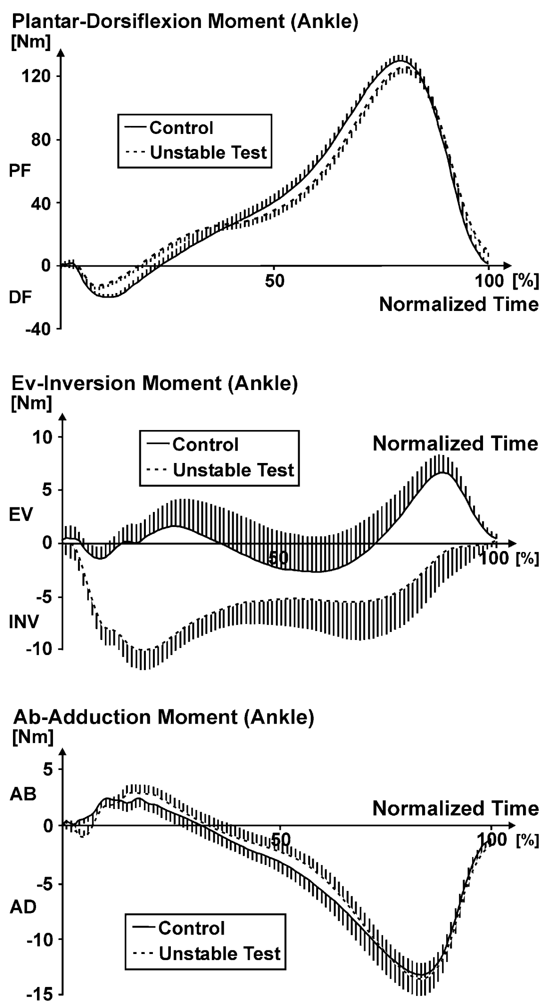

Fig. 4. Ankle plantar–dorsiflexion moments (top), ankle ab-adduction moments (center) and ankle in-eversion moments (bottom) during the stance phase of gait for the control shoe and the unstable test shoe.

3.2.3. EMG

There were no significant differences in EMG activities between the control and the MBT shoe. However, several muscles showed some trends. In comparison to the control shoe condition the tibialis anterior muscle showed an average reduction of 26% (SD 24%), the biceps femoris of 55% (SD 60%), the gastrocnemius an average increase of 52% (SD 82%), the vastus medialis of 4% (SD 13%) and the gluteus medius of 16% (SD 25%).

4. Discussion

As expected, the MBT shoes showed a much larger excursion of the center of force in the anterior–posterior and in medio-lateral direction. The test shoes produced a substantial and significant increase in the movement of the center of force, which is a sign for increased instability. Thus, the test shoes were, as hypothesized in H1 more unstable than the control shoes. In analogy to the wobble boards, which are often used for strength and proprioceptive training during rehabilitation, one may speculate that MBT shoes may serve as an effective training device for muscle strength, stability and proprioception (Woollacott and Shumway-Cook, 2002), a speculation which is tested in another ongoing study in our center.

Based on its construction, the MBT shoe forced the user to land more towards the midfoot and, consequently, the plantar–dorsiflexion movement was different between the two tested shoes during the first half of the stance phase as hypothesized (H2). However, the changes in kinematics were around the neutral axis and, consequently, no increased joint loading (Nigg, 2001; Nigg and Wakeling, 2001) for the MBT shoe should result from this kinematic change.

The resultant joint moments and joint impulses are an indication for the joint loading. All nine joint impulses did not show any significant differences between the two shoe conditions. Thus, hypothesis H3 could not be supported. For the plantar–dorsiflexion rotational loading, the effects of the changes were small and the differences were at low moment values. For the rotational ab-adduction loading during take-off, the MBT shoe showed a reduction. The rotational eversion loading was higher for the control shoe, the rotational inversion loading was higher for the MBT for the first half of ground contact. The rotational impulses for the knee joint showed a trend towards slight decreases (between 16% and 83%) for the MBT shoe for all three rotational knee axes, also not significant (Fig. 5). The rotational impulses for the hip joint axes for the second half of ground contact showed for the MBT shoe a not significant trend towards a reduction for the internal/external rotation axis and only small changes for the ab/adduction and the flexion/extension rotation axis (Fig. 5). Thus, hypothesis H4 was supported. Consequently, based on the resultant joint

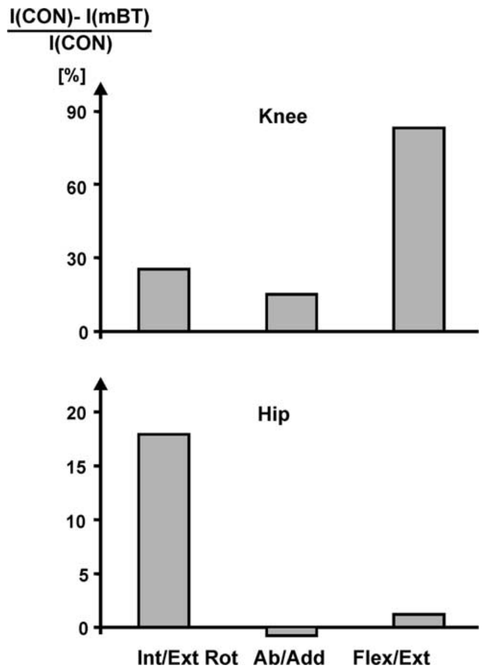

Fig. 5. Relative changes of resultant joint impulses for the three knee rotation axes for the total ground contact (top) and the three hip rotation axes for the second half of ground contact (bottom).

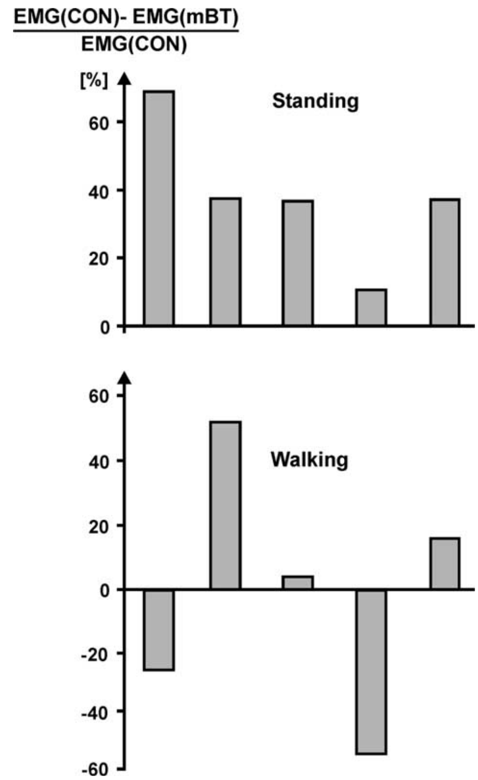

Fig. 6. Relative changes of EMG intensity for the five muscle groups tested for 8 s of standing (top) and walking (bottom) from 200 ms before to 500 ms after heel strike.

moments and impulses, there is a trend that the loading due to the resultant joint moments is decreased at the knee and hip joints.

However, joint loading depends not only on the resultant joint moments but also on the muscle contraction. Specifically, co-contraction of muscles can add substantially to joint loading. Therefore, EMG intensities may provide an indication of additional joint loading or unloading. The EMG measurements during standing showed an average increase in all EMG activities (Fig. 6), also only significant for the tibialis anterior muscle, providing weak support for hypothesis H5. Since these EMG intensities are small compared to locomotion one should conclude that they are not a concern with respect to joint loading. However, it is suggested that this increased activity may be advantageous for the training of these muscles.

The EMG measurements during walking showed an increase in the gastrocnemius activity and decreases for all other muscles, all not significant (Fig. 6). Thus, when estimating the additional joint loading from a muscle point of view, one cannot conclude for the ankle and knee joint. However, the additional muscle joint loading for the hip joint should be slightly reduced.

The fact that changes in muscle activity and resultant joint moments were for many variables not significant, despite the fact that the average changes were quite substantial is related to the different strategies used by the subjects when changing from the control shoe to the MBT (von Tscharner et al., 2003). Some subject applied an ankle joint control strategy, some a combination of knee and/or hip strategy. The different demands of the specific strategies are expressed in different joint moments and/or different EMG activities. It is interesting that on an individual basis, the changes in joint moments and EMG intensities were very consistent between the two shoe conditions.

Another aspect of muscle effects may consider the intermittent aspect of the activity changes. A visual control of the EMG signals showed more variation of the EMG signals for the MBT than for the control shoe. This may be associated with a possible loosening of muscle tenseness. However, quantifiable and conclusive evidence for this observation is missing.

Previous studies have suggested that balance training devices, such as a wobble-board or an unstable surface, can significantly improve ankle and knee muscle strength, proprioception, general rehabilitation (Waddington et al., 2000; Waddington and Adams, 2004; Wester et al., 1996), and can help prevent lower extremity musculoskeletal injuries (Bahr et al., 1997; Caraffa et al., 1996; Wedderkopp et al., 1999). However, only one study has investigated changes in muscle EMG following a balance training protocol (Cosio-Lima et al., 2003). It was reported that a 5 week balance training program resulted in improved balance and increased EMG activity of the pectoral girdle and abdominal muscles compared to a conventional exercise protocol (Cosio-Lima et al., 2003). The results of the present study are in agreement with the cited previous studies and suggest that the use of a standing balance training device, such as the MBT shoe, can alter muscle EMG activity over time.

In summary, the MBT shoe produced changes in kinematic, kinetic and EMG characteristics that seem to be advantageous for the locomotor system. Based on these initial results it seems warranted that further studies investigate aspects such as improvement of muscle strength, improvement of stability during locomotion, pain reduction for arthritic knees and injury prevention for high performance athletes.

Acknowledgements

The authors thank Masai Barefoot Technology (Switzerland), for their financial support and Tim Gormley, for his technical assistance and help with data processing.

References

Bahr, R., Lian, O., Bahr, I.A., 1997. A twofold reduction in the incidence of acute ankle sprains in volleyball after the introduction of an injury prevention program: a prospective cohort study. Scandinavian Journal of Medicine and Science in Sports 7, 172–177.

Caraffa, A., Cerulli, G., Projetti, M., Aisa, G., Rizzo, A., 1996. Prevention of anterior cruciate ligament injuries in soccer. A prospective controlled study of proprioceptive training. Knee Surgery, Sports Traumatology, Arthroscopy 4, 19–21.

Cosio-Lima, L.M., Reynolds, K.L., Winter, C., Paolone, V., Jones, M.T., 2003. Effects of physioball and conventional floor exercises on early phase adaptations in back and abdominal core stability and balance in women. Journal of Strength and Conditioning Research 17, 721–725.

Emery, C.A., 2004. Is there a clinical standing balance measurement appropriate for use in sports medicine? A review of the literature. Journal of Science and Medicine in Sport 6, 492–503.

Jackman, R.W., Kandarian, S.C., 2004. The molecular basis of skeletal muscle atrophy. American Journal of Physiology. Cell Physiology 287 (4), C834–C843.

Lieber, R.L., 1990. Statistical significance and statistical power in hypothesis testing. Journal of Orthopaedic Research 8, 304–309.

Nigg, B.M., 2001. The role of impact forces and foot pronation: a new paradigm. Clinical Journal of Sports Medicine 11, 2–9.

Nigg, B.M., Wakeling, J.M., 2001. Impact forces and muscle tuning—a new paradigm. Exercise and Sport Sciences Review 29, 37–41.

Spector, S.A., 1985. Trophic effects on the contractile and histochemical properties of rat soleus muscle. Journal of Neuroscience 5, 2189–2196.

von Tscharner, V., 2001. Intensity analysis in time–frequency space of surface myoelectric signals by wavelets of specified resolution. Journal Electromyography and Kinesiology 10, 433–445.

von Tscharner, V., Goepfert, B., Nigg, B.M., 2003. Changes in EMG signals for the muscle tibialis anterior while running barefoot or with shoes resolved by non-linearly scaled wavelets. Journal of Biomechanics 36, 1169–1176.

Waddington, G., Adams, R.D., 2004. The effect of a 5-week wobble-board exercise intervention on ability to discriminate different degrees of ankle inversion, barefoot and wearing shoes: a study in healthy elderly. Journal of the American Geriatrics Society 52, 573–576.

Waddington, G., Seward, H., Wrigley, T., Lacey, N., Adams, R., 2000. Comparing wobble board and jump–landing training effects on knee and ankle movement discrimination. Journal of Science and Medicine in Sport 3, 449–459.

Wakeling, J.M., Pascual, S.A., Nigg, B.M., von Tscharner, V., 2001. Surface EMG shows distinct populations of muscle activity when measured during sustained sub-maximal exercise. European Journal of Applied Physiology 86, 40–47.

Wedderkopp, N., Kaltoft, M., Lundgaard, B., Rosendahl, M., Froberg, K., 1999. Prevention of injuries in young female players in European team handball. A prospective intervention study. Scandinavian Journal of Medicine and Science in Sports 9, 41–47.

Wester, J.U., Jespersen, S.M., Nielsen, K.D., Neumann, L., 1996. Wobble board training after partial sprains of the lateral ligaments of the ankle: a prospective randomized study. Journal of Orthopaedic Sports and Physical Therapy 23, 332–336.

Woollacott, M., Shumway-Cook, A., 2002. Attention and the control of posture and gait: a review of an emerging area of research. Gait and Posture 16, 1–14.