



Contents lists available at ScienceDirect

Clinical Biomechanics

journal homepage: www.elsevier.com/locate/clinbiomech

Changes in running kinematics and kinetics in response to a rockered shoe intervention

Katherine A. Boyer^{a,b,*}, Thomas P. Andriacchi^{a,b}

^a Mechanical Engineering, Stanford University, Stanford, CA, USA

^b Bone and Joint RR&D, VA Palo Alto Hospital, Palo Alto, CA, USA

ARTICLE INFO

Article history:

Received 5 February 2009

Accepted 10 August 2009

Available online xxx

Keywords:

Locomotion
Joint mechanics
Joint forces
Kinematics
Kinetics
Motion analysis
Ankle

ABSTRACT

Background: A suggested link between ambulatory mechanics and injury development has resulted in significant interest the development of footwear to change locomotion patterns. The purpose of this study was to test the hypothesis that there will be significant changes in the kinematics and kinetics at the ankle and minimal changes at the knee and hip in the mechanics of running in a shoe with a sagittal plane curvature relative to a flat soled shoe.

Methods: During running 3-D lower extremity kinematics and kinetics for 19 healthy volunteers were quantified using an optoelectronics system and a force plate. Data were collected for a flat sole conventional shoe (New Balance 658 (Control)) and a shoe with a rounded sole in the sagittal plane (Masai Barefoot Technologies (MBT)). Data were compared for the two shoe conditions using paired Student *t*-tests ($\alpha = 0.05$).

Findings: The ankle dorsi-flexion angles at heel-strike and mid-stance were greater, while the ankle plantar and dorsi-flexion moments and peak ankle joint power were significantly lower with the MBT relative to the control ($P < 0.05$). Decreases in the first medial GRF peak and the peak anterior GRF peak were also found for running in the MBT shoe.

Interpretation: Despite a major difference in sole geometry, accommodations to the rockered sole were found only at the ankle. These results suggest changes in ankle kinematics and kinetics may be used to minimize the effect of changes in sole rocker on limb dynamics. Thus, changes in shoe rocker may offer potential therapeutic opportunities for running related conditions at the ankle without substantial risk to the knee or hip.

© 2009 Elsevier Ltd. All rights reserved.

1. Introduction

There is increasing interest in methods to modify patterns of human locomotion to address specific needs for load modification, rehabilitation and/or disease prevention (Nigg et al., 1999; Romkes et al., 2006; Li and Hong, 2007; Fisher et al., 2007). Possible explanations for the various changes in locomotor patterns in response to changes in external stimuli occur to maintain biological loading of tissue within a specific range (Hardin et al., 2004); to control the external forces acting on the body by maintaining a preferred movement pattern (Nigg, 2001); and/or to minimize the metabolic cost of a specific task (Nigg, 2001).

Investigations into the effects of shoe interventions designed to modify patterns of locomotion have primarily focused on interventions that could increased shoe stability and thus limit excessive

foot and/or knee motions (Frederick, 1986; Nigg et al., 1999). Foot orthotics introducing material or structural changes in medial-lateral (ML) stability are commonly prescribed to treat a large number of running related injuries, including Achilles tendinopathy, patella-femoral pain and illiotibial band syndrome (Johnston et al., 2003; MacLean, 2001). Some controversy as to the specific effect of orthotics inserts exists in the literature (Nigg, 2001) as results from different studies are often contradictory. Changes primarily in foot eversion, ankle inversion moments and ankle plantar/dorsi-flexion range of motion have been reported in some studies in response to foot orthotics (Nigg et al., 1998; Mundermann et al., 2003). In contradiction another study reported no changes in foot eversion but changes in tibial rotation and knee ab/adduction moments (Williams III et al., 2003). Changes in the shoe heel construction have also been investigated, an increase in the ML heel flare of the shoe sole was shown to influence the initial shoe and ankle pronation however, not systematically in all subjects (Stacoff et al., 2001; Nigg and Morlock, 1987). Despite a limited understanding of why and how the runners adapt their gait to shoe interventions, clinical success rates for footwear interventions are reported to be

* Corresponding author. Address: Mechanical Engineering, Stanford University, 219 Durand, 496 Lomita Mall, Stanford, CA 94305-4038, USA.

E-mail address: kboyer@stanford.edu (K.A. Boyer).

high (McNicol et al., 1981; Gross et al., 1991) and thus merit more research.

Footwear stability is most often considered in the ML plane. However, more recent developments have focused on the effects of less rigid, minimalist or unstable interventions (Nigg et al., 2006; Romkes et al., 2006) assuming that less support for the feet and lower extremity would stimulate and strengthen the muscles that contribute to static and dynamic stability thereby minimizing injury risks. Anterior–posterior (AP) stability of a shoe can be altered by changing the amount of sole rocker or heel construction in the AP direction. To date, these studies have only focused on walking and shown that altering the AP rocker of the shoe produces the greatest adaptations at the ankle joint (Li and Hong, 2007; Nigg et al., 2006; Romkes et al., 2006; Long et al., 2007). These studies reported significant decreases in the ankle dorsi-flexion angles in stance, trends for reduction in the ankle sagittal and frontal plane joint moments and small changes in the knee or hip sagittal plane joint motions and moments. The changes in AP rocker of shoe soles have also been shown to be effective in stimulating muscle activity (Nigg et al., 2006; Romkes et al., 2006). Limited changes higher up the kinematic chain in these studies were attributed to changes in muscle activation patterns, which compensated or overcame the change in the foot–ground interface to maintain the kinematic pattern of the knee, hip, and trunk segment during walking.

While the body of literature on interventions for medial lateral stability is growing there is a paucity of information on the influence of modifying the AP curvature of the shoe on running mechanics. Yet this information might be an important consideration in the evaluation of potential interventions that can assist in addressing overuse injuries, muscle weakness and rehabilitation following injury or with disease. The recently developed Masai Barefoot Technologies (Switzerland) shoe (MBT) with a rounded sole offers the opportunity to study the various strategies of adaptation to a rockered sole design. Therefore the purpose of this study was to investigate the mechanism of adaptation to a rockered sole shoe in running. In particular, to address a subject's ability to accommodate to a rockered shoe design at a single joint or whether the adaptations will take place at the ankle, knee and hip joints. Specifically, this study tested the hypothesis that there will be significant changes in the kinematics and kinetics at the ankle and minimal changes at the knee and hip during running.

2. Methods

19 healthy volunteers were tested, all with no history of serious lower-limb injury or lower-extremity pain for a minimum of 6 months prior to testing: 11 women (age: 28.9 Standard Error (SE) 7.3 years; BMI: 22.7 SE 2.9) and 8 men (age: 32.6 SE 7.5 years; BMI: 23.5 SE 1.8). Informed consent was obtained from all subjects prior to testing per Stanford University IRB guidelines. The experimental sample size was estimated from previous walking kine-

matic and kinetic data in the MBT shoe (Nigg et al., 2006; Romkes et al., 2006) using a power analysis (Lieber, 1990). A minimum of 15 subjects was deemed necessary to provide the statistical power (0.80) to detect a parameter difference of 15%.

The control shoe in this study was the New Balance 658 ($m = 269$ g). The unstable shoe condition tested in this study was the MBT M-walk shoe (Masai Barefoot Technology, Switzerland) ($m = 625$ g). The tested MBT shoe was characterized by a rounded shoe sole design (Fig. 1) in the anterior posterior direction, making the shoe unstable in the AP direction. The sole was composed of two materials, one located in the heel region (orange colour in Fig. 1) and a different material for the anterior sole section. As the subjects had not previously worn the MBT shoe they were asked to wear the shoe for a two week period prior to testing so that they were comfortable moving in the shoe before testing.

Data was captured as subjects ran in a straight line along an 11 m long runway. The experiment consisted of three running trials first for the rockered test shoe and then for the control shoe at a self selected speed. Kinematic data throughout the trial was collected using an 8-camera system (Qualisys AB, Gothenburg, Sweden) sampling at 120 Hz. Ground reaction force data was collected using a force plate (Bertec Corporation, Columbus, Ohio), built into the floor, sampling at 120 Hz and synchronized to the kinematic measurements. Subjects were allowed as many practice trials as necessary prior to collection in each condition and were instructed to move their starting position along the runway so that they could land reliably on the force platform with the right foot without altering their stride pattern. Lower extremity joint kinematics for the right leg were quantified using the point cluster technique (PCT) (Andriacchi et al., 1998). With the PCT, clusters of nine and seven reflective markers were distributed on the thigh and shank, respectively, to predict the movements of the underlying femur and tibia. A two second static trial in each of the shoe conditions was recorded with the subject standing still in order to create the anatomical reference frames for each limb segment. Additional markers placed on bony landmarks (medial and lateral malleolus, lateral and medial tibial plateau, lateral and medial femoral condyles, greater trochanter, anterior and posterior superior iliac spine, and the iliac crest) were used to establish the tibial, femoral and pelvic anatomic coordinate systems. Cluster coordinate systems were calculated for the marker clusters on the thigh and shank separately by calculating principal axes assuming a unit weight for each marker. The relative position between the marker cluster coordinate systems and the anatomical coordinate systems were also calculated in the static run for the thigh and shank. The running trials were then carried out after removing the femoral condyle markers, and the medial tibial plateau and medial malleolus markers. Marker trajectories were reconstructed using Qualisys Track Manager software (Qualisys AB, Gothenburg, Sweden), using the information about the relative locations of the anatomic coordinate systems to cluster coordinate systems obtained in the static trial, the locations of the femur, tibia and foot and projected angles between segments were calculated. The net external interseg-

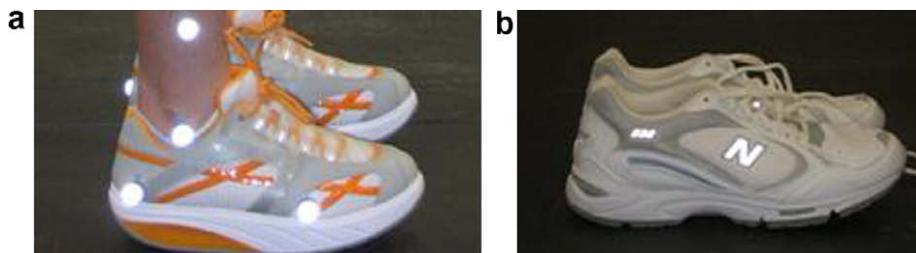


Fig. 1. (a) Unstable shoe condition used in the study. MBT M-walk shoe Masai Barefoot Technology, Switzerland and (b) New Balance 658 control shoe.

mental joint forces and moments were calculated using inverse dynamics methods and sagittal plane joint powers were calculated using the net joint moment and the joint angular velocity.

To compare among subjects of different body sizes, the ground reaction forces (GRF) were scaled to % body weight (BW) and joint moments scaled to % BW \times height. The joint kinematics and kinetics data were time normalized to 100% of stance and for each subject the average of three trials for each subject and shoe was used for analysis. Joint kinematics were quantified at the heel-strike, mid-stance and toe-off time points. For the joint kinetics and powers, the peak values in mid-stance were quantified for both the MBT and the control shoe.

The kinetic, kinematic and power variables for each shoe condition were compared using paired Student's T-tests (SPSS 16.0, SPSS Inc © 1987–2007) at the level of significance of 0.05.

3. Results

The general pattern of sagittal plane joint motion in the MBT shoe was the same as in the control shoe (Fig. 2). However, there were differences in the ankle joint kinematics at heel-strike, mid-stance and at toe-off for the MBT shoe trials. In the MBT shoe there was a greater ankle dorsi-flexion angle ($P = 0.001$) at heel-strike and this was maintained through mid-stance ($P = 0.01$). At toe-off the ankle was less plantar-flexed in the MBT shoe compared with the control shoe ($P = 0.03$). There were no significant differences in the hip and knee joint kinematics during the stance phase of running.

The magnitudes of the first medial GRF peak and the peak anterior (push-off) force were significantly lower in the MBT shoe (Fig. 3). A significantly greater posterior force during the impact phase of stance (first 20%) was also found for the MBT shoe. There were no differences in the magnitude of the impact or active peaks in the vertical GRF, or in the loading rate of the impact peak between the MBT and control shoe (Fig. 3). The self-selected running speed was statistically slower (2.5 ± 0.1 MBT; 2.6 ± 0.1 m/s control; $P < 0.001$) and thus stance time slightly longer in the MBT shoe than for the control shoe (293 ± 7 ms MBT; 285 ± 6 ms control; $P = 0.008$).

There was a significant decrease in the external ankle dorsiflexion, plantarflexion and inversion joint moments ($P < 0.0001$;

$P = 0.002$; $P = 0.003$) (Fig. 4). The decrease in the ankle dorsi-flexion moment was on average 12%. There was a trend for a decrease in the ankle eversion, and internal rotation moments ($P = 0.084$, $P = 0.062$). No changes in the external joint moments were found at the knee or hip however, there was considerable variability in these results.

The peak negative and peak positive ankle joint powers were significantly lower for the MBT shoe ($P < 0.0001$) (Fig. 5). The opposite was found for the knee, where an increase ($P = 0.034$) in peak positive joint powers was found. There were no differences found for the minimum or maximum hip joint powers. The total peak positive joint power in stance (sum of hip, knee and ankle) was lower in the MBT shoe than in the control shoe.

4. Discussion

This is the first study to our knowledge to investigate the effects of a rockered shoe sole such as the MBT on the joint kinematic and kinetics in running. In agreement with our primary hypothesis the accommodation to the MBT shoe occurred primarily at the ankle joint with an increase in the ankle dorsi-plantar flexion angle at heel-strike that was maintained through the stance phase, a reduction in the sagittal plane ankle joint moments and the peak negative and positive ankle joint powers. Additional changes in the frontal plane ankle moments were also found. The kinematic and kinetic accommodations at the knee and hip were not significant and highly variable among tested subjects.

The changes in ankle joint moments suggest that the net contribution of the muscles crossing the ankle joint complex during running with the MBT are smaller than with the control shoe, a direct consequence of the change in landing position specifically the decreased ankle dorsi-flexion angle. This reduction in the net moment can reflect a reduction in the agonistic group or an increase in the antagonistic activity and thus may also indicated a reduced joint load in the case of the former (Andriacchi et al., 1997; Herzog et al., 2003). Associated with the kinematic and kinetic changes at the ankle were reductions in both the peak positive and negative ankle joint powers providing further support for the suggestion that there were changes in forces of the muscles surrounding the ankle joint associated with the movement task.

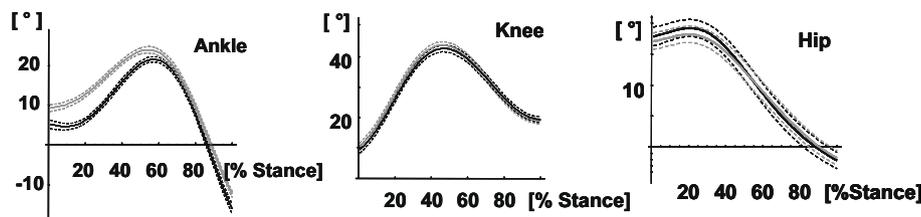


Fig. 2. Mean and SE sagittal plane joint angles for the stance phase of running for the ankle (left), knee (middle) and hip (right). MBT shoe is shown in grey and the control shoe in black.

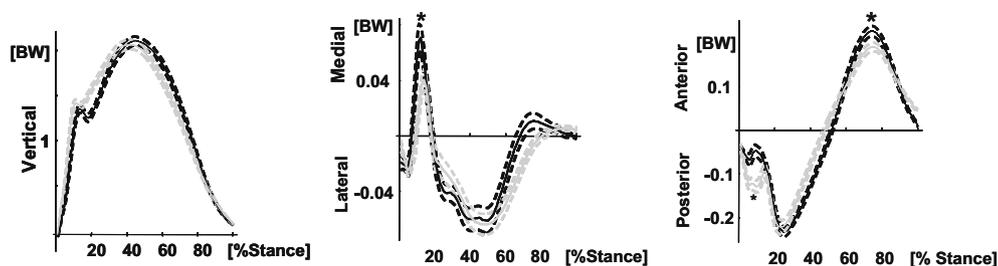


Fig. 3. Mean and SE ground reaction force data for the MBT (grey) and control shoes (black) running.

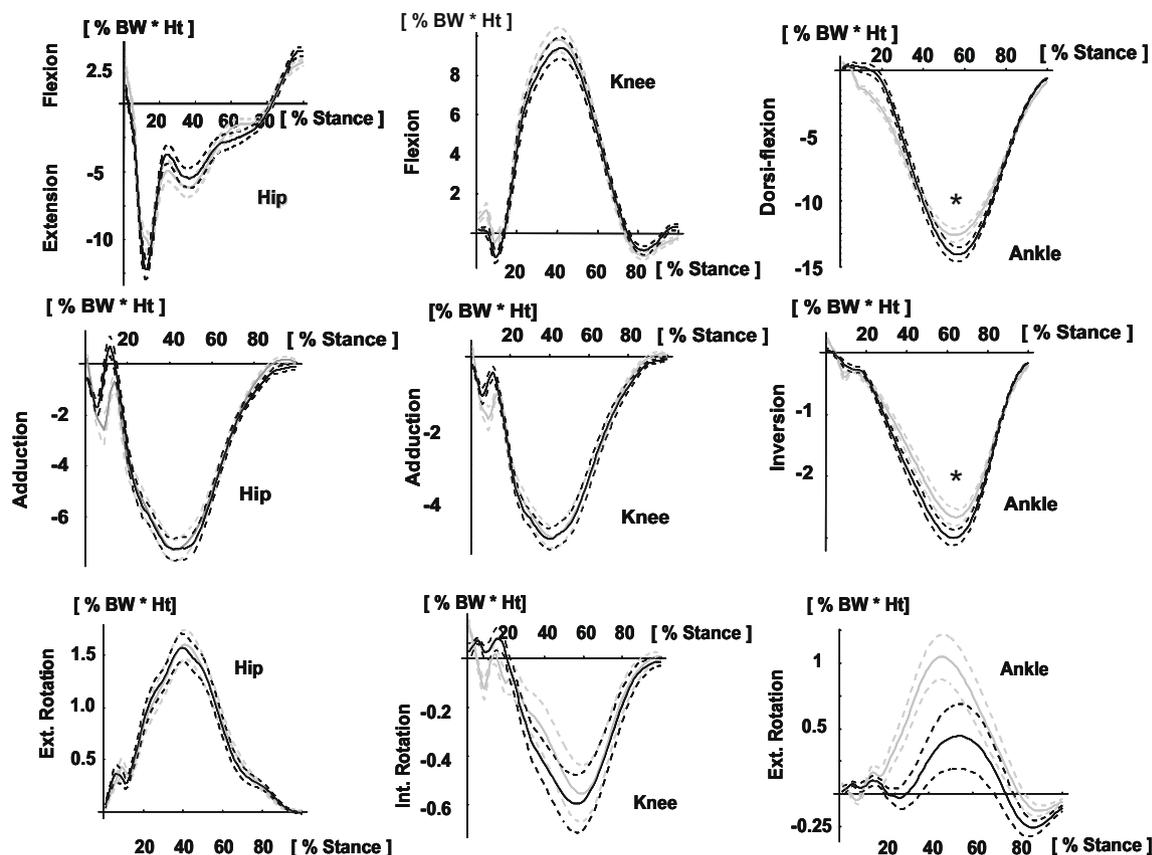


Fig. 4. Stance phase mean and SE external joint moments for the hip (left), knee (middle) and ankle (right) joints for the MBT (grey) and Control (black) shoes. *Indicates a significant difference in the peak joint moment ($P < 0.05$).

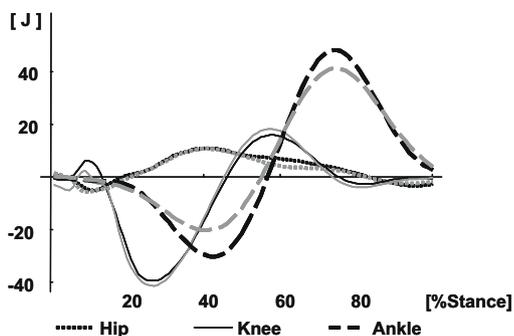


Fig. 5. Mean joint power during stance for the hip, knee and ankle joints for running in the MBT (Grey) and control (Black) shoes.

Yearly incidence of running injuries in recreational runners has been estimated at 37–56% (van Mechelen, 1992). Overuse injuries that appear to be due to constant repetition of the same movement with insufficient rest account for 50–70% of all injuries. Numerous intrinsic and extrinsic factors have been associated with specific injuries. In the case of Achilles tendinopathy (AT), one of the more common injuries, biomechanical factors that have been linked to the injuries are greater plantar and dorsi-flexion moments, earlier and increase peak pronation, greater foot eversion, ankle dorsiflexion and knee flexion in the stance phase of running for AT injured versus healthy controls (Donoghue et al., 2008). Medial posted orthotics inserts are one of the non invasive treatments of AT injuries (Werd, 2007). The reported functional effects of orthoses in AT injured runners are reduced ankle dorsi-flexion range of motion,

decreased maximum dorsiflexion position and increased ankle eversion (Donoghue et al., 2008). A reduced ankle range of motion in the early portion of the stance phase in the MBT shoe, similar to that found in the orthoses study (Donoghue et al., 2008), and reduce sagittal plane joint moments suggest it could be a beneficial intervention in the return to the sport phase of AT recovery. These potentially beneficial changes at the ankle occurred without systematically affecting the sagittal plane hip and knee joint motions or moments.

The subjects in this study were all healthy young active individuals tested while running at a slow pace. Although the results indicate there may be positive implications for those with low leg injuries the adaptations to the MBT shoe may not be the same for injured persons or for running at faster speeds. Frontal plane foot-ankle motions have previously been identified as important factors in lower leg injuries but were not quantified in this study. These variables should be measured to further understand the changes in ankle motions and the potential for this intervention to be used in rehabilitation overuse injuries (i.e. Achilles tendonopathy) of foot and ankle structures. While adaptations at the hip and knee were not found, it is possible that pelvic or upper body adaptations may have occurred in response to the rockered sole. Future investigations that include upper body analysis would be of interest. A more complete understanding of the adaptation to the rockered shoe sole may be obtain by also including measurements of muscle activations in future work. Quantification of muscle activations are of interest as changes in level of agonist/antagonistic muscles can result in a increase in joint loads despite decreases in the net external joint moments. The results for the knee and hip were varied among the subject for some people these changes when running in the MBT shoe may have clinical significance.

The finding that the ankle can accommodate for substantial changes in the shoe rocker is in agreement with other studies, however in general the kinematic and kinetic accommodations were smaller in running than for walking (Nigg et al., 2006; Romkes et al., 2006). This is not surprising as simple dynamic models of walking and running (Adamczyk et al., 2006; Seyfarth et al., 2001) indicate that AP curvature (foot size and shape) effects are greater in walking than in running. This is (again) not surprising as running is considered a bouncing gait and “easily” modeled with simple point-mass-spring models (Blickhan, 1989). While walking can be modeled as a simple inverted pendulum motion (Kuo, 2001), more complex models that include a foot segment and curvature more closely model the experimentally determined metabolic cost of walking (Adamczyk et al., 2006; Seyfarth et al., 2001).

5. Conclusions

Early research suggested that shoes might be a powerful tool for manipulating human movement and thus for preventing injuries (Frederick, 1986; Clarke et al., 1983; Nigg et al., 1981). Despite a large body of literature in the area it remains unclear how and why the human musculoskeletal system adapts to external influences on the foot/ground interface (Nigg and Wakeling, 2001; Nigg, 2001; Hardin et al., 2004; Frederick, 1986). It has been suggested that experimental evidence of these effects is limited in part because of the limited conditions under which the adaptations and interventions are studied (Hardin et al., 2004; McNair and Marshall, 1994). The rockered sole of the MBT shoe used in this investigation provided an opportunity to investigate a condition far outside the normal range of shoe designs and yet the kinematic differences for the knee and hip joint were not significant. These results suggest it is possible to accommodate substantial changes in the curvature of the sole by changes in the motion and forces sustained at the ankle and can occur with minimal change to the knee or hip motions or moments. Changes in shoe sole geometry may offer potential therapeutic opportunities for conditions at the ankle without substantial risk to the knee or hip.

Conflict of interest

Financial support by Masai Barefoot Technologies. The sponsoring company was not involved in the study design, data collection/analysis, or preparation of the manuscript.

Acknowledgments

Authors wish to thank Ms. Katerina Blazek for help with data collection and processing and Dr. Benno Nigg for many helpful discussions and manuscript review.

References

Adamczyk, P.G., Collins, S.H., Kuo, A.D., 2006. The advantages of a rolling foot in human walking. *J. Exp. Biol.* 209, 3953–3963.
 Andriacchi, T.P., Johnson, L.M., Hurwitz, D.E., Natarajan R., 1997. Musculoskeletal dynamics, locomotion, and clinical applications. In: Mow V.C. (Ed.), *Basic Orthopedic Biomechanics*. pp. 37–68.

Andriacchi, T.P., Alexander, E.J., Toney, M.K., Dyrby, C., Sum, J., 1998. A point cluster method for in vivo motion analysis: applied to a study of knee kinematics. *J. Biomech. Eng.* 120, 743–749.
 Blickhan, R., 1989. The spring-mass model for running and hopping. *J. Biomech.* 22, 1217–1227.
 Clarke, T.E., Frederick, E.C., Hamill, C.L., 1983. The effects of shoe design parameters on rearfoot control in running. *Med. Sci. Sports Exerc.* 15, 376–381.
 Donoghue, O.A., Harrison, A.J., Laxton, P., Jones, R.K., 2008. Lower limb kinematics of subjects with chronic achilles tendon injury during running. *Res. Sports Med.* 16, 23–38.
 Fisher, D.S., Dyrby, C.O., Mundermann, A., Morag, E., Andriacchi, T.P., 2007. In healthy subjects without knee osteoarthritis, the peak knee adduction moment influences the acute effect of shoe interventions designed to reduce medial compartment knee load. *J. Orthop. Res.*
 Frederick, E.C., 1986. Kinematically mediated effects of sport shoe design: a review. *J. Sports Sci.* 4, 169–184.
 Gross, M.L., Davlin, L.B., Evanski, P.M., 1991. Effectiveness of orthotic shoe inserts in the long-distance runner. *Am. J. Sports Med.* 19, 409–412.
 Hardin, E.C., van den Bogert, A.J., Hamill, J., 2004. Kinematic adaptations during running: effects of footwear, surface, and duration. *Med. Sci. Sports Exerc.* 36, 838–844.
 Herzog, W., Longino, D., Clark, A., 2003. The role of muscles in joint adaptation and degeneration. *Langenbecks Arch. Surg.* 388, 305–315.
 Johnston, C.A., Taunton, J.E., Lloyd-Smith, D.R., McKenzie, D.C., 2003. Preventing running injuries. Practical approach for family doctors. *Can. Fam. Physician* 49, 1101–1109.
 Kuo, A.D., 2001. A simple model of bipedal walking predicts the preferred speed-step length relationship. *J. Biomech. Eng.* 123, 264–269.
 Li, J.X., Hong, Y., 2007. Kinematic and electromyographic analysis of the trunk and lower limbs during walking in negative-heeled shoes. *J. Am. Podiat. Med. Assoc.* 97, 447–456.
 Lieber, R.L., 1990. Statistical significance and statistical power in hypothesis testing. *J. Orthop. Res.* 8, 304–309.
 Long, J.T., Klein, J.P., Sirota, N.M., Wertsch, J.J., Janisse, D., Harris, G.F., 2007. Biomechanics of the double rocker sole shoe: gait kinematics and kinetics. *J. Biomech.* 40, 2882–2890.
 MacLean, C.L., 2001. Custom foot orthoses for running. *Clin. Podiatr. Med. Surg.* 18, 217–224.
 McNair, P.J., Marshall, R.N., 1994. Kinematic and kinetic parameters associated with running in different shoes. *Brit. J. Sports Med.* 28, 256–260.
 McNicol, K., Taunton, J.E., Clement, D.B., 1981. Iliotibial tract friction syndrome in athletes. *Can. J. Appl. Sport Sci.* 6, 76–80.
 Mundermann, A., Nigg, B.M., Humble, R.N., Stefanyshyn, D.J., 2003. Foot orthotics affect lower extremity kinematics and kinetics during running. *Clin. Biomech.* 18, 254–262.
 Nigg, B.M., 2001. The role of impact forces and foot pronation: a new paradigm. *Clin. J. Sport Med.* 11, 2–9.
 Nigg, B.M., Morlock, M., 1987. The influence of lateral heel flare of running shoes on pronation and impact forces. *Med. Sci. Sports Exerc.* 19, 294–302.
 Nigg, B.M., Wakeling, J.M., 2001. Impact forces and muscle tuning: a new paradigm. *Exerc. Sport Sci. Rev.* 29, 37–41.
 Nigg, B.M., Denoth, J., Neukomm, P.A., 1981. Quantifying the load on the human body: problems and some possible solutions. In: Morecki, A., Fidelus, K., Kedzior, K., Wit, A. (Eds.), *Biomechanics*. University Park Press, Baltimore, pp. 88–99.
 Nigg, B.M., Khan, A., Fisher, V., Stefanyshyn, D., 1998. Effect of shoe insert construction on foot and leg movement. *Med. Sci. Sports Exerc.* 30, 550–555.
 Nigg, B.M., Nurse, M.A., Stefanyshyn, D.J., 1999. Shoe inserts and orthotics for sport and physical activities. *Med. Sci. Sports Exerc.* 31, S421–S428.
 Nigg, B.M., Hintzen, S., Ferber, R., 2006. Effect of an unstable shoe construction on lower extremity gait characteristics. *Clin. Biomech.* 21, 82–88.
 Romkes, J., Rudmann, C., Brunner, R., 2006. Changes in gait and EMG when walking with the Masai Barefoot Technique. *Clin. Biomech.* 21, 75–81.
 Seyfarth, A., Gunther, M., Blickhan, R., 2001. Stable operation of an elastic three-segment leg. *Biol. Cybern.* 84, 365–382.
 Stacoff, A., Reinschmidt, C., Nigg, B.M., van Den Bogert, A.J., Lundberg, A., Denoth, J., Stussi, E., 2001. Effects of shoe sole construction on skeletal motion during running. *Med. Sci. Sports Exerc.* 33, 311–319.
 van Mechelen, W., 1992. Running injuries. A review of the epidemiological literature. *Sports Med.* 14, 320–335.
 Ward, M.B., 2007. Achilles tendon sports injuries: A review of classification and treatment. *J. Am. Podiat. Med. Assoc.* 97, 37–48.
 Williams III, D.S., McClay I, D., Baitch, S.P., 2003. Effect of inverted orthoses on lower-extremity mechanics in runners. *Med. Sci. Sports Exerc.* 35, 2060–2068.

The effect of material characteristics of shoe soles on muscle activation and energy aspects during running

B.M. Nigg*, D. Stefanyshyn, G. Cole, P. Stergiou, J. Miller

Human Performance Laboratory, Faculty of Kinesiology, University of Calgary, Calgary, Canada

Accepted 11 November 2002

Abstract

The purposes of this study were (a) to determine group and individual differences in oxygen consumption during heel–toe running and (b) to quantify the differences in EMG activity for selected muscle groups of the lower extremities when running in shoes with different mechanical heel characteristics. Twenty male runners performed heel–toe running using two shoe conditions, one with a mainly elastic and a visco-elastic heel. Oxygen consumption was quantified during steady state runs of 6 min duration, running slightly above the aerobic threshold providing four pairs of oxygen consumption results for comparison. Muscle activity was quantified using bipolar surface EMG measurements from the tibialis anterior, medial gastrocnemius, vastus medialis and the hamstrings muscle groups. EMG data were sampled for 5 s every minute for the 6 min providing 30 trials. EMG data were compared for the different conditions using an ANOVA ($\alpha = 0.05$). The findings of this study showed that changes in the heel material characteristics of running shoes were associated with (a) subject specific changes in oxygen consumption and (b) subject and muscle specific changes in the intensities of muscle activation before heel strike in the lower extremities. It is suggested that further study of these phenomena will help understand many aspects of human locomotion, including work, performance, fatigue and possible injuries.

© 2003 Elsevier Science Ltd. All rights reserved.

Keywords: Muscle activation; Oxygen consumption; Running; Visco-elastic heel material

1. Introduction

Heel–toe runners experience impact forces between 1.0 and 2.5 times body weight. One would expect impact forces to produce substantial vibrations of the soft tissue packages of the lower extremities (e.g. quadriceps, hamstrings, triceps surae). However, impact related soft tissue vibrations are small or not apparent. Thus, the soft tissue packages must have mechanical characteristics that correspond to heavily or critically damped mechanical systems. Since vibrations are small for any shoe–surface combination the mechanical characteristics of the soft tissues must be adjusted by muscle pre-activation (Nigg, 1997; Wakeling and Nigg, 2001) and one should expect muscle and subject specific reactions for different shoe conditions.

Thus, one may speculate that muscles are programmed to avoid vibrations (Nigg, 1997). A specific muscle group should change its activity when the frequency of the input signal is close to its natural frequency and the changes in activity should depend on the mass of the soft tissue package and its mechanical characteristics. Furthermore, it is speculated that any change in muscle activity affects work and performance of the human locomotor system (Nigg, 2000). Thus, effects on muscle activity should be observable by studying EMG signals and effects on work by studying oxygen consumption. However, experimental evidence for such phenomena is not available.

The purposes of this study were:

- (a) to determine group and individual differences in oxygen consumption during heel–toe running and
- (b) to quantify the differences in EMG activity for selected muscle groups of the lower extremities.

*Corresponding author. University of Calgary, Faculty of Kinesiology, 2500 University Drive NW, Calgary, Alta, Canada T2N 1N4. Tel.: +403-220-3436; fax: +403-284-3553.

E-mail address: nigg@ucalgary.ca (B.M. Nigg).

When running in shoes with different mechanical heel characteristics.

Specifically the following hypotheses were tested:

- H1 The group differences in oxygen consumption between the elastic and the viscous shoe condition are small and not significant.
- H2 There are groups of subjects with significantly less, equal and more oxygen consumption when running in shoes with elastic and viscous heels.
- H3 Changes in shoe conditions change EMG activity of selected lower extremity muscles. The differences are muscle and subject specific.

2. Methods

Twenty proficient male *runners* participated in this study and gave written informed consent. They were free of any serious injuries at the time of study.

Two *shoe conditions* were used in this study. The two shoes were identical (same uppers, outsoles, insoles, etc.) but differed in the midsole materials of the heel. One heel material was of medium hardness (shore $C = 45$) and mainly elastic. The other heel material was softer (shore $C = 26$) and more viscous. To quantify the differences between the materials the heels of the two shoes were tested in a testing machine (MTS Systems Corporation, Eden Prairie, Minnesota, USA). A heel-shaped indenter attached to the actuator of the MTS machine was used to compress the heel of each shoe at a speed of 300 mm/s. The tests were performed to approximately 14 mm deformation to simulate actual deformation during locomotion. Three trials were performed on each shoe. The results of these tests are illustrated in Fig. 1.

The two shoes differed in hardness (as measured with a durometer) and in visco-elasticity (as indicated by the

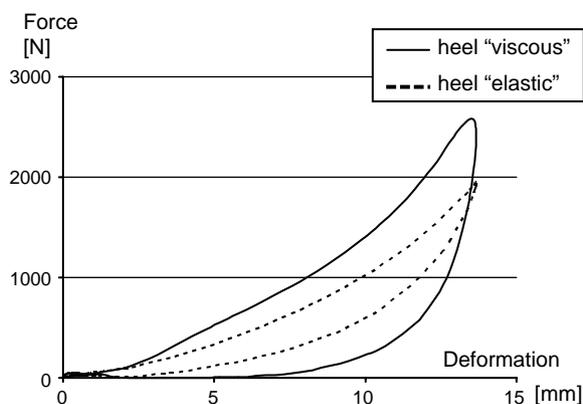


Fig. 1. Force–deformation diagrams for the two heels of the tested shoes as determined through MTS testing at 300 mm/s. Each curve is the average of three test results under the same conditions.

remaining deformation after the task). One shoe (solid line) was harder and more visco-elastic. The other shoe (dotted line) was softer and less visco-elastic. Shoe A will be named *viscous*, shoe B will be named *elastic* in the text. The mass of the shoes differed by less than 6 g with the elastic shoe (size 9) having a mass of 293 g and the visco-elastic shoe (size 9) having a mass of 288 g. All runners used at the time of the experiment size 9 or 10 (US) running shoes.

Oxygen consumption was quantified while running on a portable treadmill (Quinton: Q65, Seattle, Washington) in three testing sessions (Williams, 1985; Morgan and Craib, 1992; Martin et al., 1993) conducted at a similar time of day for each subject to eliminate the potential variation in VO_2 due to circadian rhythm. First, a *calibration session* was used to determine aerobic threshold, anaerobic threshold and $\text{VO}_{2\text{max}}$. Two *testing sessions* were used to quantify the oxygen consumption for the two shoe conditions. The two *testing sessions* started with a 10–15 min warm-up period. The actual test consisted of steady-state runs of 6 min duration, running slightly above the aerobic threshold. The four VO_2 values measured every 30 s for the final 2 min were used to determine the mean oxygen consumption for a given shoe and test. A 3-min rest period was allowed between each run. One testing session had a testing sequence of v–e followed by e–v. The other testing session had the testing sequence e–v followed by v–e (where e and v corresponded to the “elastic” and the “viscous” shoe condition). The sequence used in the first test session was randomly assigned. This procedure provided 4 pairs of oxygen consumption results for comparison.

Individual oxygen consumption results were defined as different for the two shoe conditions when the VO_2 values for all four paired comparisons (v–e, e–v, e–v, v–e) were consistently higher or lower for one shoe condition compared to the other shoe condition. Based on the oxygen consumption results, three groups were defined:

Viscous group: $\text{VO}_2(\text{elastic}) > \text{VO}_2(\text{viscous})$ for all four comparisons.

Elastic group: $\text{VO}_2(\text{elastic}) < \text{VO}_2(\text{viscous})$ for all four comparisons.

Neutral group: $\text{VO}_2(\text{elastic}) \approx \text{VO}_2(\text{viscous})$ for not consistent comparisons.

A paired *T*-test was performed to compare between the two shoe conditions ($\alpha = 0.05$) for the whole group.

Muscle activity was quantified for the right leg from EMG (1200 Hz, Biovision, Germany) measured in a special testing session on the same treadmill. Oxygen consumption measurements and EMG measurements were not made simultaneously because simultaneous measurements could influence each other negatively. Bipolar surface EMG measurements were taken from for the tibialis anterior, medial gastrocnemius, vastus

medialis and the hamstrings group. Vertical force data from the back right support leg of the treadmill (Kistler AG, Winterthur, Switzerland, 1200 Hz) were used to determine the time of the right foot contact. EMG data were sampled for 5 s every minute for the 6 min. A total of 5 steps per data collection were analyzed providing 30 trials (steps) for analysis. EMG data were filtered (von Tscherner, 1999) with a Bandpass Filter (20–400 Hz cut-off). A root mean square (RMS) calculation was performed on the filtered EMG data from 50 ms prior to contact to the time of first contact with the treadmill (muscle pre-activation). The top and bottom 10% of the RMS data were removed in order to eliminate possible outliers. Therefore, results from 24 foot-contacts were used in the analysis for each shoe condition and subject. This data were compared for the different conditions using an ANOVA ($\alpha = 0.05$).

The results for subject 13 were excluded from further analysis because the oxygen consumption difference between the 2 days was more than 10%. The results for subject 15 were excluded because the subject did not complete the oxygen consumption measurements. For the remaining 18 subjects, EMG data were excluded for

the vastus medialis muscle for four subjects (no. 4, 7, 10 and 14) because there was too much noise in the signal that could not be filtered out.

3. Results

The results are grouped with respect to the following aspects, (a) description of a typical EMG data set for treadmill running, (b) group results and (c) individual results for oxygen consumption and EMG data.

3.1. EMG-force data set

A data set for one ground contact for the four muscles involved is illustrated in Fig. 2 for subject #20.

3.2. Group differences for the two shoe conditions

For *oxygen consumption*, the mean group differences between the elastic and the viscous shoe condition were small and not significant for both days of testing [Day 1: $\text{VO}_2(\text{elastic}) = 38.9 \pm 4.3 \text{ ml/kg/min}$,

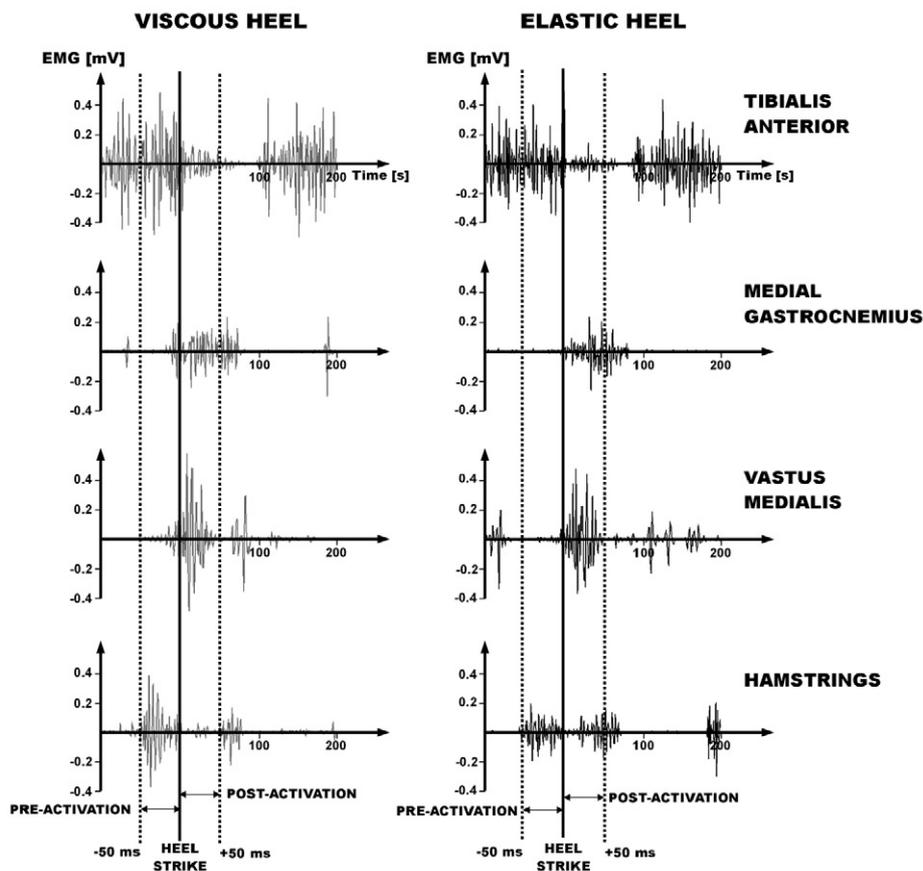


Fig. 2. Illustration of the EMG signals for the four muscles tibialis anterior, gastrocnemius medialis, vastus medialis and hamstrings for one subject (subject 20) while running with the shoe with the viscous (left) and the elastic (right) heel. The solid vertical line indicates first ground contact, the dotted lines indicate 50 ms before and 50 ms after first ground contact.

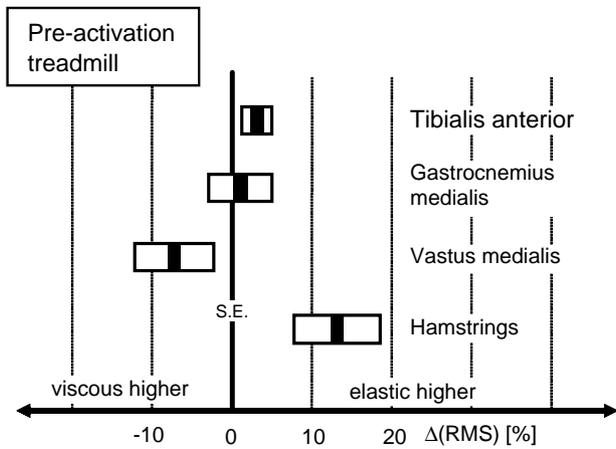


Fig. 3. Percentage changes of the mean RMS values with SE for the four muscles for the 18 subjects. A positive sign indicates that the RMS was higher for the elastic than for the viscous shoe condition.

$VO_2(\text{viscous}) = 38.9 \pm 4.5$ ml/kg/min; Day 2: $VO_2(\text{elastic}) = 39.4 \pm 3.4$ ml/kg/min, $VO_2(\text{viscous}) = 39.4 \pm 3.7$ ml/kg/min]. Thus, hypothesis H1 was supported by the results of this study.

The RMS of the *EMG measurements* for the four muscles measured showed no significant group differences between the two shoe conditions (Fig. 3). The relative differences in the pre-activation between the averages for the elastic and viscous shoe conditions were 3.2% for the tibialis anterior, 0.9% for the gastrocnemius, -7.2% for the vastus medialis and 13.2% for the hamstrings.

4. Individual results

The individual differences in *oxygen consumption* are illustrated in Fig. 4. Five subjects (2, 3, 8, 18 and 19) used in all four comparisons less oxygen for the elastic shoe-heel situation. Five subjects (4, 6, 7, 16 and 20) used in all four comparisons less oxygen for the viscous shoe heel situation. Eight subjects (1, 5, 9, 10, 11, 12, 14 and 17) had not consistent results for the four comparisons. Thus, hypothesis H2 was supported by the results of this study.

The RMS of the EMG showed muscle and subject specific results. Thus, hypothesis H3 was supported. The RMS of the EMG pre-activation showed some systematic changes for the vastus medialis muscle (Fig. 5). All five subjects in the “elastic” group, the group that used more oxygen in the visco-elastic shoe, showed a higher vastus medialis pre-activation for the viscous shoe condition. The average difference in the RMS for this muscle was 20.3% with a range between 3.5% and 65.7%. All three subjects in the “viscous” group for which EMG results were available (6, 16 and 20) showed a higher vastus medialis pre-activation for the elastic

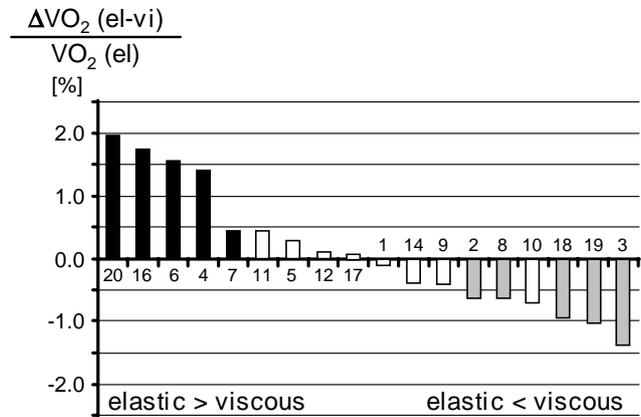


Fig. 4. Individual normalized differences in oxygen consumption between the elastic and the viscous shoe condition. Positive values indicate that $VO_2(\text{elastic}) > VO_2(\text{viscous})$. Negative values indicate that $VO_2(\text{elastic}) < VO_2(\text{viscous})$. The differences were determined based on four oxygen measurements during four different trials per shoe condition (a total of 16 measurements per shoe and 32 measurements per difference). The shaded bars indicate the five subjects that had the same positive difference between the elastic and the viscous shoe condition for both days.

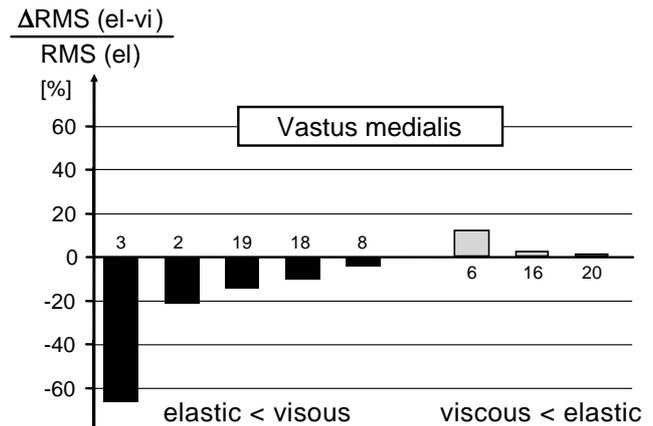


Fig. 5. Relative difference of the EMG-RMS [(elastic–viscous): (elastic)] for the vastus medialis muscle for the elastic (black) and the viscous group (gray). The viscous group has only three results since the EMG data for the other two members of this group were rejected.

shoe condition. The average difference in the RMS for this muscle was 4.7% with a range between 0.9% and 11.7%. The other muscles did not show a systematic difference.

5. Discussion

The purposes of the study were to determine group and individual differences in oxygen consumption during heel-toe running in shoes with elastic or visco-elastic heel materials and to investigate whether

observed differences would be associated with changes in muscle activity.

5.1. Oxygen consumption

The results of this study confirmed that oxygen consumption did, in the average, not differ between the two shoe conditions. However, individual subjects showed systematic and consistent differences of up to 2% (Fig. 4) in oxygen consumption for the two tested shoe heel materials. The finding that runners use less oxygen when running with visco-elastic heel material is counterintuitive and is generally not expected. However, the experimental results for the group that used less oxygen with the visco-elastic heels (the “viscous” group) is in line with theoretical predictions made earlier that viscous shoes sole material may, under certain conditions, require less work for a given locomotion task (Nigg and Anton, 1995).

Thus, the often-used statement that elastic shoe midsole materials are more advantageous for performance than visco-elastic materials is not correct. Some subjects performed better when using visco-elastic and others when using elastic heel midsole materials. Taking into account that the shoes used in this study were arbitrarily selected (the only goal was to have one shoe with an elastic and one with a visco-elastic heel) and that the selected visco-elastic materials were determined by availability, one could speculate that differences in oxygen consumption may even be higher if the shoe sole material was optimally tuned to each subject. However, the criteria for *optimal tuning* are not known yet.

5.2. Muscle activity

Changes in muscle activity can occur with respect to timing, intensity and frequency. There were no changes in timing registered for the two shoe conditions. The changes in the intensity of the muscle activity (RMS) before heel strike when changing footwear were in some cases quite substantial (Fig. 5). The muscle “package” with the smallest mass had the lowest range of change of RMS (TA: 35%), the larger muscle “packages” had a range of changes in muscle activity between 85% and 91% (GM: 91%; VM: 90%; HA: 85%) indicating that larger muscle activity changes were needed for the soft tissue packages with the bigger masses.

The intensity of the EMG pre-activation showed some systematic changes for the vastus medialis muscle (Fig. 6). All five subjects in the “elastic” group, the group that used more oxygen in the visco-elastic shoe, showed a higher vastus medialis pre-activation for the visco-elastic shoe condition. The average difference in the RMS for this muscle was 20.3% with a range between 3.5% and 65.7%. All three

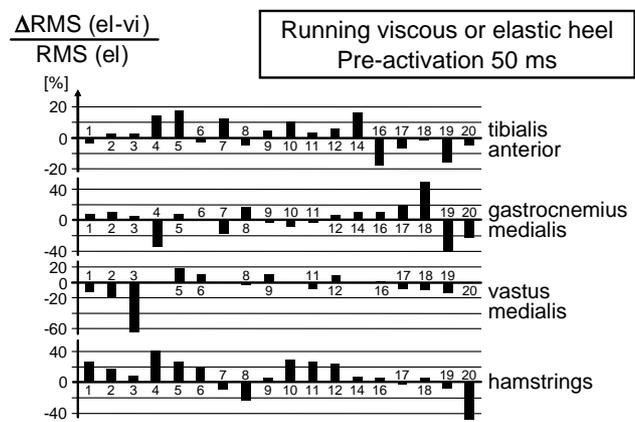


Fig. 6. Relative EMG RMS between the elastic and the viscous shoe condition for the vastus medialis muscle for two groups of subjects, subjects that used less oxygen in the elastic shoe condition (left) and subjects that used less oxygen in the viscous shoe condition (right).

subjects in the “viscous” group (6, 16 and 20) showed a higher vastus medialis pre-activation for the elastic shoe condition. The average difference in the RMS for this muscle was 4.7% with a range between 0.9% and 11.7%. The other muscles did not show a systematic difference.

Changes in the mean wavelet frequencies between shoe conditions were not significant for the group comparisons. However, in analogy to the individual RMS data, there were substantial subject and muscle specific changes in the mean frequencies for the different shoe conditions.

The EMG signals before heel strike had a significant and substantially higher mean wavelet frequency than the signals after heel strike confirming previous experimental results (Wakeling et al., 2001). The frequency characteristics of the EMG signals are influenced by the conduction velocity and the shape of the action potential (Lindström et al., 1970; Stulen and de Luca, 1981; Solomonow et al., 1990). Differences in conduction velocities may be attributed to fast and slow twitch fibers. However, experimental evidence for this functional relationship is contradictory (Karlson, 1999; Wakeling et al., 2002). The experimental results of this study suggest that EMG activities before and after heel strike belong to two different events. It is speculated that the EMG activities before heel strike are preprogrammed based on the expected impact shock and are related to a “muscle tuning” activity (Nigg and Wakeling, 2001) while the EMG activities after heel strike belong to a reflex arc event.

If one accepts the speculation that changes in EMG wavelet frequencies are related to changes in muscle fiber recruitment, one may suggest that “muscle tuning” uses more fast twitch muscle fibers and the “muscle tuning” be related to fatigue. Thus, it should be studied,

whether a reduction of muscle activity before heel strike is associated with fatigue and/or performance.

5.3. Signal frequencies

During running and many other human locomotion activities various frequencies are rather close. The mean frequency of the impact force signal during heel strike is between 5 and 20 Hz. When running on a treadmill this treadmill has a natural frequency about 10–20 Hz. The soft tissue packages (e.g. triceps surae, hamstrings, etc.) have a natural frequency around 5–60 Hz. Thus, the mean frequencies of impact force signals during heel–toe running are close to the natural frequencies of the soft tissue packages of the lower extremities. Consequently, it seems important to understand strategies that can be applied to avoid resonance phenomena of the soft tissue packages. One strategy consists of using the filtering abilities of shoe soles, sport surfaces and shoe inserts; frequency considerations must be performed to understand the possibilities and the limitations. For some subject in the tested population the viscous for others the elastic heel was the “better” solution with respect to oxygen consumption. Another strategy consists in changing the mechanical characteristics (F and c) of the muscles in the soft tissue packages, moving their natural frequency away from the input frequency of the shock wave. It may be that such changes in input frequencies were responsible for earlier published differences in oxygen consumption with different shoes or surfaces (Bates et al., 1978; Caitlin and Dressendorfer, 1979; Clarke et al., 1983a,b; Frederick et al., 1983; Fukuda et al., 1983; Stacoff and Kaelin, 1983; Nigg and Bahlens, 1988; Hamill et al., 1988; Milani et al., 1995).

5.4. Conclusion

The findings of this study showed that changes in the heel material characteristics of running shoes are associated with (a) subject specific changes in oxygen consumption, (b) subject and muscle specific changes in the intensities of muscle activation before heel strike in the lower extremities and (c) subject and muscle specific changes in the mean wavelet frequencies of the EMG signals. The functional understanding of these changes is in its infancy. However, it is suggested that the further study of these phenomena will help understand many aspects of human locomotion, including work, performance, fatigue and possible injuries.

Acknowledgements

This study has been financially supported by Adidas, NSERC and the da Vinci Foundation.

References

- Bates, B.T., Osternig, L.R., Mason, B., James, S.L., 1978. Lower extremity function during the support phase of running. In: Asmussen, E.A., Jorgensen, K. (Eds.), *Biomechanics VI-B*. University Park, Baltimore, pp. 30–39.
- Caitlin, M.J., Dressendorfer, R.H., 1979. Effect of shoe weight on the energy cost of running. *Medicine and Science in Sports* 11, 80.
- Clarke, T.E., Frederick, E.C., Cooper, L.B., 1983a. In: Nigg, B.M., Kerr, B.A. (Eds.), *Biomechanical Measurement of Running Shoe Cushioning Properties. Biomechanical Aspects of Sport Shoes and Playing Surfaces*. University of Calgary, Calgary, AB, pp. 24–33.
- Clarke, T.E., Frederick, E.C., Hamill, C.L., 1983b. The effects of shoe design parameters on rearfoot control in running. *Medicine and Science in Sports and Exercise* 15, 376–381.
- Frederick, E.C., Clarke, T.E., Larsen, J.L., Cooper, L.B., 1983. The effects of shoe cushioning on the oxygen demands of running. In: Nigg, B.M., Kerr, B.A. (Eds.), *Biomechanical Measurement of Running Shoe Cushioning Properties. Biomechanical Aspects of Sport Shoes and Playing Surfaces*. University of Calgary, Calgary, AB, pp. 107–114.
- Fukuda, H., Ohmichi, H., Miyashita, M., 1983. Effects of shoe weight on oxygen uptake during submaximal running. In: Nigg, B.M., Kerr, B.A. (Eds.), *Biomechanical Measurement of Running Shoe Cushioning Properties. Biomechanical Aspects of Sport Shoes and Playing Surfaces*. University of Calgary, Calgary, AB, pp. 115–122.
- Hamill, J., Freedson, P.S., Boda, W., Reichsman, F., 1988. Effects of shoe type on cardiorespiratory responses and rearfoot motion during treadmill running. *Medicine and Science in Sports and Exercise* 20, 515–521.
- Karlson, S., 1999. Enhancement of spectral analysis of myoelectric signals during static contractions using wavelet analysis methods. *IEEE Trans Biomed Eng* 46 (6), 670–684.
- Lindström, L., Magnusson, R., Petersén, I., 1970. Muscular fatigue and action potential conduction velocity changes studies with frequency analysis of EMG signals. *Electromyography* 4, 341–353.
- Martin, P.E., Heise, G.D., Morgan, D.W., 1993. Interrelationships between mechanical power, energy transfers, and walking and running economy. *Medicine and Science in Sports and Exercise* 25, 508–515.
- Milani, T.L., Schnabel, G., Hennig, E.M., 1995. Rearfoot motion and pressure distribution patterns during running in shoes with varus and valgus wedges. *Journal of Applied Biomechanics* 11, 177–187.
- Morgan, D.W., Craib, M., 1992. Physiological aspects of running economy. *Medicine and Science in Sports and Exercise* 24, 456–461.
- Nigg, B.M., 1997. Impact forces in running. *Current Opinion in Orthopaedics* 8, 43–47.
- Nigg, B.M., 2000. Force acting on and in the human body. In: Nigg, B.M., MacIntosh, B.R., Mester, J. (Eds.), *Biomechanics and Biology of Movement. Human Kinetics*, Champaign, IL, USA, pp. 253–267.
- Nigg, B.M., Anton, M., 1995. Energy aspects for elastic and viscous shoe soles and playing surfaces. *Medicine and Science in Sports and Exercise* 27, 92–97.
- Nigg, B.M., Bahlens, A.H., 1988. Influence of heel flare and midsole construction on pronation, supination, and impact forces for heel-toe running. *International Journal of Sports Biomechanics* 4, 205–219.
- Nigg, B.M., Wakeling, J.M., 2001. Impact forces and muscle tuning – a new paradigm. *Exercise and Sport Sciences Review* 29 (1), 37–41.
- Stulen, F., de Luca, C., 1981. Frequency parameters of the myoelectric signal as measure of conduction velocity. *IEEE Trans Biomed Eng* 28, 515–523.
- Stacoff, A., Kaelin, X., 1983. Pronation and sports shoe design. In: Nigg, B.M., Kerr, B.A. (Eds.), *Biomechanical Aspects of Sport*

- Shoes and Playing Surfaces. University of Calgary, Calgary, AB, pp. 143–151.
- Solomonow, M., Baten, C., Smith, J., Baratta, R., Hermens, H., D'Ambrosia, R., Shoji, H., 1990. Electromyogram power spectra frequencies associated with motor unit recruitment strategies. *Journal Applied Physiology* 68 (3), 1177–1185.
- Wakeling, J.M., Nigg, B.M., 2001. Modification of soft tissue vibrations in the leg by muscular activity. *Journal Applied Physiology* 90, 412–420.
- Wakeling, J.M., Rozitis, A.I., Nigg, B.M., 2002. Muscle activity damps the soft tissue resonance in response to pulsed and continuous vibrations. *Journal Applied Physiology* 93, 1093–1103.
- Wakeling, J., von Tscharnner, V., Nigg, B.M., Stergiou, P., 2001. Muscle activity in the leg is tuned in response to ground reaction forces. *Journal Applied Physiology* 91, 1307–1317.
- Williams, K., 1985. The relationship between mechanical and physiological energy estimates. *Medicine and Science in Sports and Exercise* 17, 317–325.