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Does wearing shoes affect your biomechanical efficiency?

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ABSTRACT

Studies involving minimalist shoes have dramatically increased this past 10 years. While a deeper knowledge of the related modifications has ensued regarding the kinematics, electromyographic, and dynamic patterns, little is known regarding the modifications at the muscle forces and muscle fiber levels. The aim of the present study was to assess at a muscular level the modifications brought up when running barefoot, using 0 mm midsole height running shoe, or using classical midsole height running shoes. An EMG-Driven model that combines the kinematics, dynamics, and electromyographic data was used to estimate the Triceps Surae (TS) muscle forces and fiber behavior during running using different footwear conditions. Despite differences at the joint level between barefoot and shod running when looking at ankle joint range of motion, or foot-ground angle at touchdown, the results showed no effect of footwear neither on the maximal muscle forces nor on the relative amount of force produced by each muscle within the TS muscle group when wearing different footwear. On the contrary, different behaviors of muscle fibers were shown with lower amplitudes of fiber lengths for the Gastrocnemii biarticular muscles when running barefoot. This particular results reveal that wearing a shoe, even with a very thin sole, could deeply modify the intricate muscle–tendon mechanics of running.
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1. Introduction

A Pubmed search using "barefoot running" or "minimalist running" returns over 160 matches. More than 130 of these studies have been published since 2000, denoting a striking growing interest in research on this topic for these last years.

Effects of barefoot and minimalist shoe on running pattern have been widely studied and authors consistently observed shorter stride length, shorter contact time and higher stride frequency in barefoot running compared to classical shod running (De Wit et al., 2000; Divert et al., 2005b). Many kinematic differences were highlighted between barefoot and shod running in habitually shod runners. Some authors have observed at touch-down a flatter foot and a more plantar-flexed ankle in barefoot running (Chambon et al., 2014; De Wit et al., 2000; Hamill et al., 2011). Moreover, it has been shown that foot eversion and internal tibial rotation were increased during barefoot running (Barnes et al., 2010; Fukano et al., 2009). These kinematic variations seem to be accompanied by muscular adaptations mainly located around the ankle joint. Indeed, greater tibialis anterior preactivation (von Tscharner et al., 2003) and lower preactivation of the three calf muscles (Divert et al., 2005b) during shod running compared to barefoot running have been observed.

These kinematic and muscular adaptations may influence body loading characteristics, but all previous observations did not converge. Some studies have shown a decrease of the first Ground Reaction Force (GRF) peak in barefoot condition (Divert et al., 2005b; Hamill et al., 2011), while others observed no difference (De Wit et al., 2000; Paquette et al., 2013). Concerning loading rate, the reported results diverge largely according to the studies that showed an increase (Chambon et al., 2014; De Wit et al., 2000; Paquette et al., 2013) or a decrease (Hamill et al., 2011) during barefoot versus shod running. The different levels of subject experience in barefoot running may explain the diverging results. Indeed, habitually barefoot runners tend to strike the ground on midfoot/forefoot, sometimes showing an absence of first peak on vertical GRF (VGRF) (Lieberman et al., 2010), with this result not being always confirmed (Hatala et al., 2013). Comparing barefoot and shod running using inverse dynamics, Kerrigan et al. (2009) have observed an increase of joint torques at ankle, knee and hip level for shod running. Divert et al. (2005a) have shown an increase of leg stiffness during barefoot running, while Hamill et al. (2011) have shown an increase of ankle stiffness.

While all the above studies focused on experimental data, none is sufficient to explain as a whole the running pattern. Muscle force represents a key variable that encompass all the above-mentioned factors into a single easily interpretable data, which can be a highly valuable addition to the experimental data. Indeed, the muscle force used to set a segment in motion or produce a force against an external

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support depends on muscle activation (usually measured using electromyography), fiber length and velocity (through the force-length-velocity relationship), and moment arm (Buchanan et al., 2004; Lloyd and Besier, 2003). Each of the classically studied parameter (i.e., muscle activation, joint kinematics, and ground reaction force) could only exhibit little to no effect of minimalist footwear. However, a variable such as muscle force combining some of these parameters in an intricate way could present stronger and clearer evolution patterns in between shod and barefoot running. Moreover, as previously shown during hopping and running, fiber lengths trajectories estimated from an EMG-Driven model are pretty close to those measured using ultrasonography (Gerus et al., 2012). The aim of the present study was to investigate the effect of barefoot, minimal, and classical thickness footwear at the level of muscle forces and muscle fiber behavior.

2. Methods

Eight male subjects (23.8 ± 3.7 years, 176 ± 4 cm, 69.8 ± 5.6 kg) agreed to participate in the experiment and gave informed written consent. The experiment was approved by the local ethical committee. All of them were habitually shod runners without having already experienced running using minimalist shoes.

2.1. Experimental conditions

Three types of footwear were tested during the experiment; no shoe: Bare condition (barefoot running condition, 0 mm heel to toe drop), minimalist shoe: 00 mm condition (no EVA midsole, 3 mm rubber outsole, 0 mm heel to toe drop), classical thickness shoe: 16 mm condition (16 mm EVA midsole, 3 mm rubber outsole, 0 mm heel to toe drop). The uppers and drops of these prototype shoes were identical for the minimalist and regular running shoes in order to avoid any unwanted effect of these factors (Morio et al., 2009). For each subject, 3 min of treadmill running at 3.33 m s^{-1} was allowed in order for them to adapt to the footwear condition. Then, the subjects were asked to run indoor at 3.33 m s^{-1} along a track. Five trials for each footwear condition were recorded. The speed was controlled using photocells and all the subjects' speeds were between 3.27 and 3.39 m s^{-1}. Three 5 s isometric Maximal Voluntary Contraction (MVC) trials with 1-min rest in between were performed for ankle plantarflexion and dorsiflexion before the running trials. For these MVC tasks, the subjects sat on a chair with their foot set into a shoe firmly fixed on the ground. Verbal encouragements were given during each MVC to ensure maximal activation of the muscles.

2.2. Data acquisition and processing

Three-dimensional kinematics data were recorded using 40 markers located over the subject lower limbs and torso and sampled at 125 Hz using an 8 cameras Vicon system. Ground reaction force data of the right foot were recorded using a Kistler (Kistler 9281 CA, dimensions: 600×400 mm^2) force plate sampled at 2000 Hz. The force plate was located at 10 m from the beginning of the running track. Electromyographic data of the Tibialis Anterior (TA), Lateral Gastrocnemius (LG), Medial Gastrocnemius (MG), and Soleus (SL) muscles were recorded at 2000 Hz using a Trigno Delsys wireless EMG system. All data were acquired synchronously using the Vicon Nexus software.

A generic model of the human body including the head, trunk, and lower limbs segments was created in Opensim (Delp et al., 2007). This model comprises 2 degrees of freedom to represent the ankle joint (one for flexion–extension and one—located at the subtalar joint—for the pronation–supination), one degree of freedom at the knee joint to represent the flexion–extension (2 additional translational degrees of freedom are dependent on the angular one), and 3 degrees of freedom at the hip joint. This model also comprised the Tibialis Anterior, Gastrocnemius Lateralis, Gastrocnemius Medialis, and Soleus muscles considered as the main contributors of the ankle joint flexion/extension movement. This generic model was scaled to match the subject's anthropometric measurements based on experimentally measured marker positions from static poses. In order to reduce the measurements errors due to marker movement on the subject's skin, an inverse kinematics algorithm was used to solve for the minimum difference between experimental and virtual markers (Delp et al., 2007). For this inverse kinematics step based on the difficulty to assess the sole mechanical characteristics during the complete stance phase, the ground contact model was kept identical for all the experimental conditions. The outputs of this inverse kinematics step consisted in ankle, knee, and hip joint angles through time. By combining kinematics, anthropometric, and force plate data, this model was further used to estimate the muscle–tendon lengths and moment arms for each muscle as well as the ankle net joint torque.

EMG data of the four muscles investigated in this study were band pass filtered (Butterworth zero time lag, 4th order, 10–450 Hz), full wave rectified, and low pass filtered (Butterworth zero time lag, 4th order, 5 Hz). The EMG data from the MVC trials were processed in the same way. Then, the overall maximum of each muscle MVC was used to normalize the remaining data set. The force plate was used to detect heel contact and toe off instants and a time window from 150 ms before ground contact to toe off was used to analyze the data of processed EMG, muscle–tendon lengths, moment arms and net joint torque.

2.3. Estimation of muscle forces

The EMG-driven model used to estimate the muscle forces has already been extensively presented (Buchanan et al., 2004; Lloyd and Besier, 2003) and only a brief description is given here. For each muscle included in the study, a Hill-type muscle model was used to estimate the muscle forces and corresponding joint moment. This Hill-type muscle model was driven by EMG data following an EMG to activation step. The parameters characterizing each muscle force production capacity were adapted from Arnold et al. (2010), which consisted in resting pennation angle, maximum isometric force, tendon slack length and optimal fiber length. In order to tune each of the model parameter, a simulated annealing constrained optimization scheme was used. The optimization criterion consisted in minimizing the difference between the experimentally computed ankle net joint torque and the net joint torque simulated by the EMG-driven model. Following this optimization step, the model can be considered as subject-specific and is thus able to estimate muscle forces individually for each subject.

2.4. Dependent variables

The goodness of fit between the experimentally computed ankle net joint torque and the net joint torque simulated by the EMG-driven model was assessed through the coefficient of determination (r^2) and Root Mean Square (RMS) values.

Over the stance phase, the influence of footwear was tested on the maximal value of the flexion/extension ankle net joint torque as well as on maximal normalized EMG, maximal and mean values of force for each muscle of the plantar flexor group. Moreover, the relative contribution of each muscle to the force of the Triceps Surae (TS) muscle group (sum of LG, MG, and SL muscles) was computed at the time instant when the TS force was maximal. Based on the outputs of the EMG-driven model, the mean values of the normalized fiber length as well as the amplitude values of the fiber lengths were studied during the stance phase for each muscle of the TS muscle group.

2.5. Statistics

Standard statistical methods were used in the calculation of means and standard deviation of the parameters studied for each participant and each condition. One-way repeated measures Analysis of Variance (®Statistica, Statsoft) were used to test the influence of the *footwear* factor on each dependant variables. All significant effects ($p < 0.05$) were followed by Tukey posthoc tests.

3. Results

Given the similar running speeds, the three experimental conditions were equivalent regarding the maximal absolute value of the ankle flexion–extension net joint torque (-185.6 ± 27.9 N m for 00 mm, -172.8 ± 47.0 N m for 16 mm, and -195.1 ± 11.7 N m for Bare).

Significant differences ($F_{12,2} = 12.57$, $p < 0.05$) on kinematics showed that foot angle at touch-down was lower for the Bare condition ($2.06 \pm 9.12^{\circ}$) relative to the 00 mm ($12.24 \pm 14.9^{\circ}$) and 16 mm ($12.69 \pm 13.43^{\circ}$) conditions. The ankle joint amplitude during the stance phase was significantly ($F_{12,2} = 5.65$, $p < 0.05$) greater for the Bare condition ($29.47 \pm 9.52^{\circ}$) than for the 0 mm and 16 mm conditions ($23.66 \pm 12.11^{\circ}$ and $25.79 \pm 12.77^{\circ}$ respectively).

The maximal activations were not different in between the *footwear* conditions. Values for the LG muscle were $46.01 \pm 21.02\%$, $50.57 \pm 12.42\%$, and $54.85 \pm 17.89\%$ MVC for the 00 mm, 16 mm, and Bare conditions respectively. Values for the MG muscle were $69.26 \pm 27.35\%$, $60.11 \pm 20.48\%$, and $71.18 \pm 25.22\%$ MVC for the 00 mm, 16 mm, and Bare conditions respectively, and $60.03 \pm 19.69\%$, $63.55 \pm 26.11\%$ and $67.18 \pm 23.95\%$ MVC for the Soleus muscle for the 00 mm, 16 mm, and Bare conditions respectively. The Tibialis Anterior muscle maximal activation was not affected by *footwear* with maximal values of $41.14 \pm 10.28\%$, $47.54 \pm 24.02\%$, and $35.53 \pm 7.41\%$ MVC for the 00 mm, 16 mm, and Bare conditions respectively.

Regarding the outputs of the EMG-driven model, the optimization process resulted in good agreements between the experimentally computed ankle net joint torque and the net joint torque

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