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Muscle co-contraction around the knee when walking with unstable shoes

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ABSTRACT

Walking with unstable shoes has been discussed to decrease joint loading. Typical estimates of joint loading using an inverse dynamic approach only account for net joint moments, not considering the potential role of muscular co-contraction. Therefore, the purpose of this study was to compare muscular co-contraction levels when walking with two different unstable shoe constructions (rocker-bottom and toning shoes) compared to walking with regular shoes. For each shoe condition, 12 healthy subjects walked with both, a regular shoe and with an unstable shoe at self-selected walking speed at a 10-m walkway. Surface EMG data of selected muscles were recorded and time normalized for calculating co-contraction indices (CCI) for opposing muscle groups.

Results showed an increase of co-contraction primarily for vastii and gastrocnemius muscles for the first and second half of stance when walking with an unstable shoe construction. Therefore, when using an inverse dynamic approach to analyze joint loading differences between regular shoes and unstable shoes, one should be cautious in interpreting the data, as these methods base their estimates of joint moments upon the net joint torque.

1. Introduction

In recent years, the influence of footwear on gait biomechanics has been of high interest for research. Typically, shoes and footwear are designed to support the foot in joint stability throughout locomotion and recreational activities, as well as to protect from environmental conditions (Reinschmidt and Nigg, 2000). One special concept of shoe design is the unstable shoe construction. This concept aims to increase instability during standing and locomotion, for example by using a rocker-bottom (e.g. Masai Barefoot Technology, MBT) or balance-pods centered in the forefoot and heel region (e.g. Reebok Easy Tone, ET). The majority of companies, who promote these type of footwear, claim, that these shoes can positively affect health, in terms of increased muscle activation during locomotion and/or a reduction of joint loads of the lower extremities.

Research has addressed these claims intensively in the last decade (e.g. Nigg et al., 2006; Romkes et al., 2006; Landry et al., 2012; Buchecker et al., 2013; Horsak and Baca, 2013). The majority of research conducted, has used motion analysis and inverse dynamic approaches to analyze effects of unstable shoe constructions on kinematic and kinetic parameters during gait, typically of the lower extremities. Some of the researchers have additionally used electromyography (EMG) to quantify changes in muscle activation patterns (Romkes et al., 2006; Buchecker et al., 2010; Horsak and Baca, 2013). The majority of studies analyzing gait biomechanics during level walking found significant alterations in sagittal and frontal plane kinematics and kinetics for the knee and ankle joints (e.g. Romkes et al., 2006; Landry et al., 2012). Briefly, when walking with rocker-bottom shoes, people tend to walk with reduced hip flexion–extension and ankle adduction–abduction range of motion (Landry et al., 2012), and an increased dorsiflexion angle at initial contact, followed by a continuous plantarflexion movement (Romkes et al., 2006). Researchers also found some notable changes for gait kinetics. Ankle moments tend to be greater for walking with the unstable shoes and at the hip and knee, both increases and decreases in moments in frontal and sagittal plane were observed (e.g. Buchecker et al., 2010; Landry et al., 2012). Linked to the kinematic changes found at the ankle-joint-complex, researchers also identified some significant alterations in muscle activity patterns of the gastrocnemius, vastii and tibialis anterior muscles. Buchecker et al. (2010) reported an increase of vastus

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lateralis and gastrocnemius medialis activity during late stance. This was also observed by Romkes et al. (2006), who reported a significant increase of muscle activity for gastrocnemius, vastus medialis, vastus lateralis and rectus femoris. It is noteworthy that in muscle-driven forward dynamics simulations of normal walking, vastii and gastrocnemius muscles have been shown to be the primary contributors to the tibia-femoral joint forces (intersegmental and compressive forces), with the peak force being even greater than the muscle forces themselves (Sasaki and Neptune, 2010).

A major limitation of the inverse dynamic approach to determine joint kinetics is, that only the net effect of muscle activity at a joint can be estimated (Winter, 2005). If co-contraction, the simultaneous activation of various muscles acting at a joint, is taking place, the analysis only yields the net effect of these muscles, hence, only net joint moments. The external knee adduction, for example, is often used as an indicator of internal joint loading. Research has shown, that only low to moderate correlation exists between the external adduction moment and internal compartment loads (e.g. Winby et al., 2013). This, often accepted, inaccuracy in estimating joint forces is even increased, if co-contraction is taking place. One way to overcome this problem is the use of some form of electromyography-driven modeling approach to estimate joint contact force and all muscle forces apparent (e.g. Sasaki and Neptune, 2010; Gardinier et al., 2013). However, this approach is not only time consuming and difficult to apply, it sometimes lacks validity as these calculations are typically handled as a standard minimization problem.

Therefore, when reporting joint loading differences between walking with normal and with unstable shoes, it is important to understand, how these shoe constructions contribute to the amount of co-contraction. In respect to the published evidence reporting biomechanical changes when walking with unstable shoes, up to date, only one study could be identified (Buchecker et al., 2010) which reported co-contraction levels.

However, the data of Buchecker et al. (2010) are limited to a specific population, as they only recruited overweight men for their study. In contrast to this specific study population, the main population which intends to use such shoe concepts will also include a more healthy and young population. Beside the rocker-bottom shoe design, which was analyzed in most of the present studies, several other unstable shoe concepts have emerged in the last years, e.g. the balance pod shoe design. Up to date no scientific literature reported analyses of co-contraction when walking with balance-pod shoe designs, even though these shoes are widely used and popular.

Due to the lack of knowledge in the potential role of unstable shoe constructions in increasing the amount of co-contraction around the knee joint, the purpose of this study was to compare co-contraction levels when walking with two different unstable shoe constructions compared to walking with regular shoes in a young healthy population. We hypothesized that walking with unstable shoes will increase the level of co-contraction during stance.

Information regarding the influence of unstable shoe constructions on co-contraction levels may help to better interpret already published data on biomechanical changes during walking with unstable shoes. It may also serve as a guide for future research and its underlying methodological approaches to quantify kinetic changes when wearing these shoes.

2. Methods

In this study data of two previously conducted studies focusing on gait biomechanics during level walking with regular and with two different types of unstable shoe constructions (Fig. 1), a

Fig. 1. The rocker-bottom shoe (MBT, Mahuta) (A) and the balance-pod shoe (Easy Tone, Reenew) (B) used in this study as unstable shoe constructions.

rocker-bottom and a balance-pod shoe design, were used retrospectively (Horsak and Baca, 2012, 2013). Both data sets were captured within the same methodological framework, as briefly described as follows, and were used separately for analyses in the present study.

2.1. Participants

For the rocker-bottom shoe data set, 12 healthy participants (7 male and 5 female) volunteered (age: 25 ± 6 years, height: 174 ± 7 cm, mass: 68 ± 10 kg). The balance-pod shoe data comprised 12 participants (5 male and 7 female; age: 25 ± 4 years, height: 172 ± 11 cm, mass: 67 ± 11 kg). In total for each data set, 14 participants were recruited, but because of technical issues, two participants of each data set had to be removed from analyses. All participants were eligible if they were free from any lower and upper extremity orthopedic or neurological problems and pain. Participants were excluded, if they had used the unstable shoe construction prior to the study. All participants were informed of the aims of this study and signed an informed consent form, which was approved by the Local Ethics Committee.

2.2. Data acquisition and processing

Participants were asked to walk with both, a regular shoe (control situation) and with the unstable shoe at self-selected walking speed at a 10-m walkway. Prior to data acquisition, participants were allowed to walk along the walk-way as often as necessary until they felt comfortable during walking with their regular shoes. This was typically reached after five minutes of walking at self-selected walking speed. After this accommodation phase, the data were captured using the different shoes. For each shoe, participants again had approximately five minutes of accommodation time to feel comfortable during walking at the walkway. Surface electromyography (EMG) data were recorded using a telemetry system (DELSYS, Myomonitor IV, Boston, MA, USA) and the software EMGworks version 3.6, which was installed on a stationary computer. Single differential Ag bar-electrodes with a contact dimension of 10.0 × 1.0 mm, an inter-electrode distance of 10.0 mm and a resulting detection area of 10 mm^2 (DELSYS, DE-2.1) were attached parallel to the muscle fiber direction over the mid-muscle belly (Hermens et al., 2000). The EMG amplifier system gain was 1000 with an common mode rejection ratio of −92 dB, an input impedance of >10^{15} Ω and an overall channel noise ⩽1.2 μV (RMS, R.T.I)^2. The analog signals were converted into digital signals using a sampling rate of 1000 Hz and a 16-bit analog–digital converter, using a ±5 V signal input range, which was integrated into to the EMG telemetry handheld device (DELSYS, Myomonitor IV, Boston, MA, USA).

Prior to electrode placement, the skin was shaved, slightly abraded and cleaned using skin preparation gel. The ground electrode was placed at the wrist. In total muscle activities of the