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# Effect of boot shaft stiffness on stability joint energy and muscular co-contraction during walking on uneven surface

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ABSTRACT

Increased boot shaft stiffness may have a noticeable impact on the range of motion of the ankle joint. Therefore, the ability of the ankle joint to generate power for propulsion might be impaired. This might result in compensatory changes at the knee and hip joint. Besides, adaptability of the subtalar joint to uneven surface might be reduced, which could in turn affect stability. The aim of the study was therefore to investigate the influence of boot shaft stiffness on biomechanical gait parameters.

Fifteen healthy young adults walked over coarse gravel wearing two different hiking boots that differed by 50% in passive shaft stiffness. Leg kinematics, kinetics and electromyography were measured. Gait velocity and indicators for stability were not different when walking with the hard and soft boot shaft over the gravel surface. However, the hard boot shaft decreased the ankle range of motion as well as the eccentric energy absorbed at the ankle joint. As a consequence, compensatory changes at the knee joint were observed. Co-contraction was increased, and greater eccentric energy was absorbed. Therefore, the efficiency of gait with hard boots might be decreased and joint loading at the knee might be increased, which might cause early fatigue of knee muscles during walking or hiking. The results of this study suggest that stiffness and blocking of joint motion at the ankle should not be equated with safety. A trade-off between lateral stiffness and free natural motion of the ankle joint complex might be preferable.



## 1. Introduction

Efficient, safe and comfortable walking is an indispensable part of human life. Therefore, improving gait performance is an important issue in medicine and recreational sport activities. Good walking performance relies on efficient transformation of mechanical power output produced by the musculoskeletal system through footwear. Hereby, the design of the shoe plays an important role. For example, the viscoelastic characteristics of the midsole or the weight of the shoe can influence the energy expenditure (Stefanyshyn and Nigg, 2000). The focus of interest in this study is the design of the shoe shaft. High shoe-shafts are worn in several types of orthopaedic, sport, hiking or military boots. In these boots the shaft may have a noticeable impact on the ankle range of movement (ROM) and therefore to the ability of the ankle joint to generate power for propulsion (Cikajlo and Matjacic, 2007). In addition, adaptability of the subtalar joint to uneven surfaces may be limited. The main reason for wearing a high boot shaft is to provide support and stability at the ankle joint (Stefanyshyn and Nigg, 2000). In particular ankle sprains of the lateral ligaments occur frequently while walking, hiking (Blake and Fergussoon, 1993) and during other sport activities (O'Loughlin et al., 2009). For this reason the biomechanical function of the boot shaft is to restrict excessive inversion. To protect the ankle joint from injuries caused by excessive inversion a considerable boot shaft stiffness might be required (Stacoff and Stüssi, 1993; Müller and Hintermann, 1996; Avramakis et al., 2000; Verhagen et al., 2001; Fong et al., 2007). Since the protective effect of the boot shaft in lateral direction requires a circular embracing of the ankle and shank, the desired increased boot shaft stiffness in lateral direction might be involved with a reduction of the functional ROM of the ankle joint. It has been shown that military boots with increased shaft stiffness, decreases the peak dorsiflexion of up to 4$^{\circ}$ and reduced the peak power produced at the ankle joint of 33%, compared to a soft boot shaft (Cikajlo and Matjacic, 2007). Sufficient ankle power is important for forward motion during gait and therefore necessary to obtain appropriate walking velocities (Requaio et al., 2005). Besides the reduced power generation at the ankle joint, the boot-shaft stiffness may have a considerable impact on the ability of the ankle joint to adapt the foot via the subtalar joint to uneven surfaces. This might impair the stability during gait (Hijmans et al., 2007). However, it has been reported in the review by Bot and van Mechelen (1999), that the majority of studies on ankle bracing showed no effect on movement performance such as jump height, running speed and agility. Consequently, the lack of

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propulsion power and the reduced adaptability of the ankle joint due to the boot-shaft stiffness may be compensated for by changes at the knee and hip joints. As a result performance, e.g. walking speed and stability, can be maintained. Possible compensatory changes observed in the literature are increased joint moments at the knee and hip as well as increased stiffness at the knee. Higher peak hip flexion moments were observed when subjects were instructed to push less with their feet as they walked (Lewis et al., 2008). Increased peak knee flexion and peak extension moments were observed when walking with short-leg walking boots compared to normal shoe conditions (Zhang et al., 2006). Increased knee stiffness caused by an enhanced activity of the hamstring muscles (m. biceps femoris and m. semitendinosus) was observed when the ankle was immobilized (Gruber et al., 2006).

The aim of this study was to investigate the influence of boot shaft stiffness on gait performance on uneven surface. First, it was investigated whether a stiff boot shaft reduces ankle ROM, power production and muscular co-contraction at the ankle joint. Second, it was studied if a stiff boot shaft increases hip and knee joint power production and muscular co-contraction at the knee joint.

## 2. Methodology

### 2.1. Experimental setup

15 healthy male subjects 29 years (SD=5), 177 cm (SD=5) and 77 kg (SD=8), volunteered to participate in this study. The subjects familiarized themselves with the experimental procedure and potential risks before providing written consent, as approved by the local ethic committee. Motion data was collected using a 6-camera Vicon MX-460 system (Vicon, Oxford, UK) operating at 240 Hz. The Vicon Plug-in-Gait marker set was used to generate kinematic and kinetic data. In addition medial ankle and knee markers were used during the standing trial to improve the reliability of ankle, knee and hip joint angles as well as joint moments (Stief et al., 2009). Ground reaction forces were recorded using a Kistler 9281 force plate $60 \times 90\,cm^2$ (Kistler, Winterthur, Switzerland) sampled at 960 Hz. Surface EMG signals for tibialis anterior, gastrocnemius medialis, peroneus longus, vastus lateralis and semitendinosus were acquired at 1000 Hz using Noraxon TeleMyo 2400 system (Noraxon Inc., Scottsdale Arizona, USA). The walking surface was covered with coarse gravel of varying diameter up to 4 cm, filled up to a height of 8 cm, along a length of 6 m. It has been previously shown that the dispersion through the ballast to the force plate does not have a significant effect on force measurements, either in shear or normal directions, if the load was applied in the approximate centre of the plate (Wade and Redfern, 2007). Therefore, walking trials were found to be valid when the measured point of force application was less than 10 cm away from the border of the force plate, other trials were repeated. After a familiarisation period with the surface and boot condition a total of 10 valid walking trials for each boot condition were captured. The order of boots was randomized. Preferred walking frequency on gravel surface was determined for each subject during the familiarisation period and was controlled by a metronome throughout the test. This acoustic signal of walking frequency has been shown to improve repeatability of walking speed compared to free locomotion (Heller and Haake, 2006).

### 2.2. Material

Two hiking boots "Tibet" and "Tahoe GTX" (Lowa Sportschuhe GmbH, Jetzendorf, Germany) with similar sole and substructures were tested. The boot with the softer shaft (Tahoe GTX) was additionally weakened in the manufacturing process such that a part of the lateral reinforcement of the shaft was not inserted. Passive shaft stiffness of both boots was measured with a prosthesis inserted into the boot (Fig. 1). Force was applied by a rope fixed to the shaft of the prosthesis. Force in the rope was measured using a load cell (AST GmbH, Wolnzach, Germany) having an accuracy of 0.3 N. The movement of the top of the prosthesis shaft was determined using the motion analysis system previously described. Position of the boot with respect to the direction of pull was varied clockwise from $0^\circ$ to $360^\circ$ in steps of $15^\circ$, which resulted in 24 measured directions. Each direction was measured twice, and the procedure was repeated clockwise to control for possible effects due to expandable lacing with increasing number of loading cycles.

### 2.3. Data processing and statistical analysis

Kinematics and EMG data were filtered with a 4th order zero lag Butterworth digital low pass filter with a cut-off frequency of 10 Hz. EMG data of each muscle were then normalized on the respective maximum signal of each subject during all trials. The co-contraction index (Rudolph et al., 2000) of muscle antagonistic pairs at the knee and ankle joints was calculated using Eq. 1.

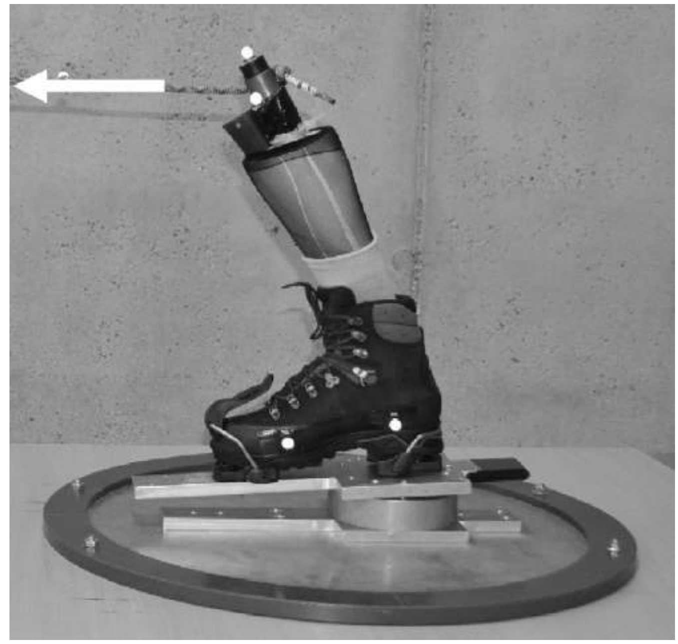

**Fig. 1.** Measurement of boot shaft stiffness. A load, indicated with the white arrow, was applied on top of the prosthesis fit into the boot. The load, elongation of the top of the prosthesis as well as the moment arm of the load with respect to the ankle joint centre was measured.

$$CI = \frac{EMG_{NL}}{EMG_{NH}}(EMG_{NL} + EMG_{NH}) \tag{1}$$

$EMG_{NL}$ is the normalized EMG-data of the less active muscle in the pairing. $EMG_{NH}$ is the normalized EMG-data of the higher active muscle in the pairing. High co-contraction values indicate a high level activation of both muscles, whereas low co-contraction values indicate either a low activation of both muscles or a high level activation of one muscle in conjunction with a low activation of the other muscle in the pairing. For the knee joint vastus lateralis and semitendinosus were paired. Ankle co-contraction was separated into medio-lateral (peroneus longus and tibialis anterior) and anterior–posterior (gastrocnemius medialis and tibialis anterior) direction. The EMG data of the stance phase was further subdivided into three phases (weight acceptance, single limb support and take off); timing of the three phases were determined by contact of the contra-lateral foot, which thereby results in two phases of double limb support separated by a phase where only the stance leg had contact with the ground. The co-contraction index was integrated with respect to phase duration and normalized on the same.

Joint energies were evaluated during stance phase of gait. Joint energies were functionally separated into eccentric and concentric behavior during stance. To indicate stability of gait the variability of step time and step width (Richardson et al., 2008) as well as the average step length (Maki, 1997) were determined. All data processing and analysis was performed using MatLab 7.3 (The MathWorks Inc., Natick, MA, USA). Paired *t*-tests of dependent samples were performed across the two different boot conditions to determine if the boot shaft stiffness had a significant effect on joint kinematics, joint kinetics, kinematic stability parameters and co-contraction indices. Normality of differences was tested with Shapiro–Wilk tests. Distribution of data was checked for skewness and kurtosis values before applying the parametric test. Significance level was set at $\alpha = 0.05$. Cohen's *d* was calculated according to Kotrlik and Williams (2003) to analyze the effect size and interpreted using Cohen's descriptors.

## 3. Results

### 3.1. Boot shaft stiffness

Fig. 2 shows a contour plot of shaft elongation in the transversal plane. Compared to the hard boot shaft, the soft boot shaft allowed about twice as much flexion in lateral direction for the same load applied to the prosthesis inserted into the boot. Surface contours showed a reniform shape, which is more pronounced in the soft shaft condition, allowing more flexion in

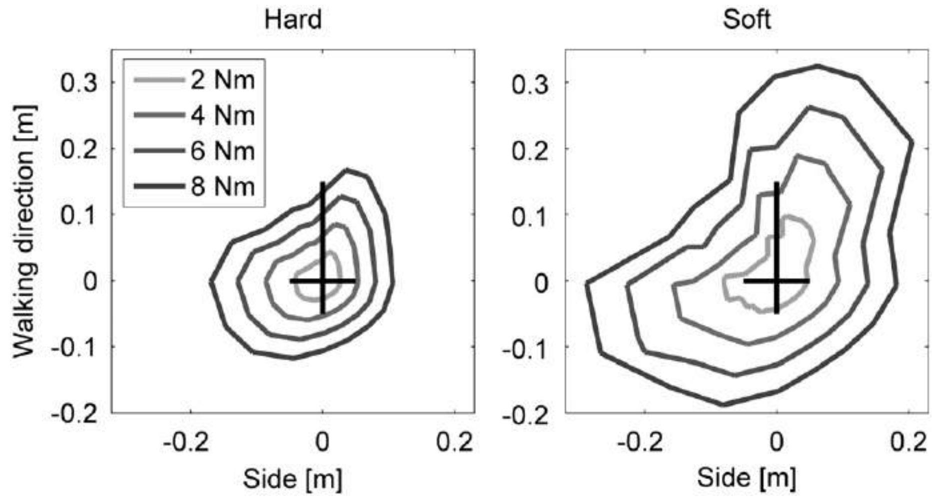


**Fig. 2.** Boot shaft stiffness, illustrated by elongation of the top of the prosthesis for different moments applied to the ankle joint of the prosthesis. The centre of the cross indicates the ankle joint centre within the boot.

**Table 1**
Spatio-temporal gait parameters.

| Gait parameter | Hard shaft | Soft shaft | *p*-value |
|---|---|---|---|
| Velocity (m/s) | 1.108 (0.082) | 1.102 (0.091) | 0.77 |
| Weight acceptance (ms) | 150 (22) | 160 (25) | **0.02** |
| Single limb support (ms) | 532 (41) | 527 (37) | 0.22 |
| Propulsion (ms) | 152 (22) | 158 (24) | 0.11 |
| Step length (cm) | 64.07 (3.70) | 64.41 (4.03) | 0.40 |
| Step time variability (ms) | 20.97 (6.88) | 23.62 (8.99) | 0.18 |
| Step width variability (cm) | 3.95 (1.20) | 3.68 (1.39) | 0.44 |

anterior-lateral and posterior-medial directions. Clockwise repetition of measurements did not demonstrate considerable changes, so that a possible effect of expandable lacing was not observed.

### 3.2. Gait performance

Gait velocity was 1.108 m/s (SD=0.082) and 1.102 m/s (SD=0.091) for the hard and soft boot, respectively. The walking speed was not significantly different between the different boots ($p=0.77$). Duration of gait phases is shown in Table 1. Significant shorter duration of weight acceptance was found for the hard boot shaft ($p=0.02$, $d=0.72$), the other gait phases were not significantly different.

Stability indicators step length, variability of step width and variability of step time were not significantly different; their values are shown in Table 1.

### 3.3. Compensation effects

Ankle ROM in plantar-dorsiflexion was 18.1° (SD=3.4) for the soft and 16.7° (SD=3.3) for the hard shaft boot. This reduction in ankle ROM in the sagittal plane was significant ($p=0.04$, $d=0.59$). In lateral direction ankle ROM was 9.4° (SD=3.3) for the soft and 8.7° (SD=3.3) for the hard boot shaft. The reduction of ROM for the hard boot shaft in the frontal plane was not significant ($p=0.16$). Hip and knee joint showed increased ROM for the hard boot, the differences were less than 0.4° and not significant.

Calculated joint powers of ankle knee and hip joint during stance phase are shown in Fig. 3. Integration of positive and negative joint powers resulted in concentric and eccentric joint energies, respectively (Fig. 4). The stiffer boot shaft increased significantly the eccentric energy of the knee joint ($p=0.03$, $d=0.63$) and decreases the eccentric energy at the ankle joint ($p=0.02$, $d=0.72$). Concentric energies as well as eccentric energies at the hip joint did not show significant differences between soft and hard boot shaft conditions.

The integrated contraction index (iCI) during three phases of gait is shown in Fig. 5. Significantly, increased co-contraction in the hard boot shaft condition was observed at the knee joint between muscles vastus lateralis and semitendinosus during single leg stance ($p=0.04$, $d=0.61$). All other phases or muscle antagonistic pairs at the ankle joint did not show significant changes in co-contraction.

## 4. Discussion

Gait velocity and indicators for stability were not different while walking with the hard and soft boot shaft over the gravel surface. However, the hard boot shaft decreased the ankle ROM as well as the eccentric energy absorbed at the ankle joint. At the same time, compensatory changes at the knee joint were observed. Co-contraction was increased, and greater eccentric energy was absorbed. The hip joint did not show significant differences in joint energy between the hard and soft boot shaft. Cohen's *d* for the significant findings ranged from 0.59 to 0.72 representing medium effect sizes.

Mechanical testing of boot shaft stiffness resulted in a reniform surface contour, allowing more flexion in anterior-lateral and posterior-medial directions. This shaft behavior is in agreement with the natural movement of the shank during normal gait, in which the weight-bearing knee joint centre moves from posterior-medial to anterior-lateral direction with respect to the ankle joint centre throughout the stance phase (Jenkyn et al., 2008). Both, the hard and the soft boot shaft showed a reniform surface contour. However, the stiffness of the hard boot shaft was about twice as much as for the soft boot shaft, in lateral as well as in anterior–posterior direction. Since there was a considerable difference in shaft stiffness between boots, a significant reduction of ankle ROM in anterior–posterior direction of 1.4° was observed during walking. The reduction of ankle ROM in lateral direction was about 0.7°, but the reduction was not significant. Since subject variability (SD) was between 35% and 38% of the total ROM in lateral direction, considerable differences in individual

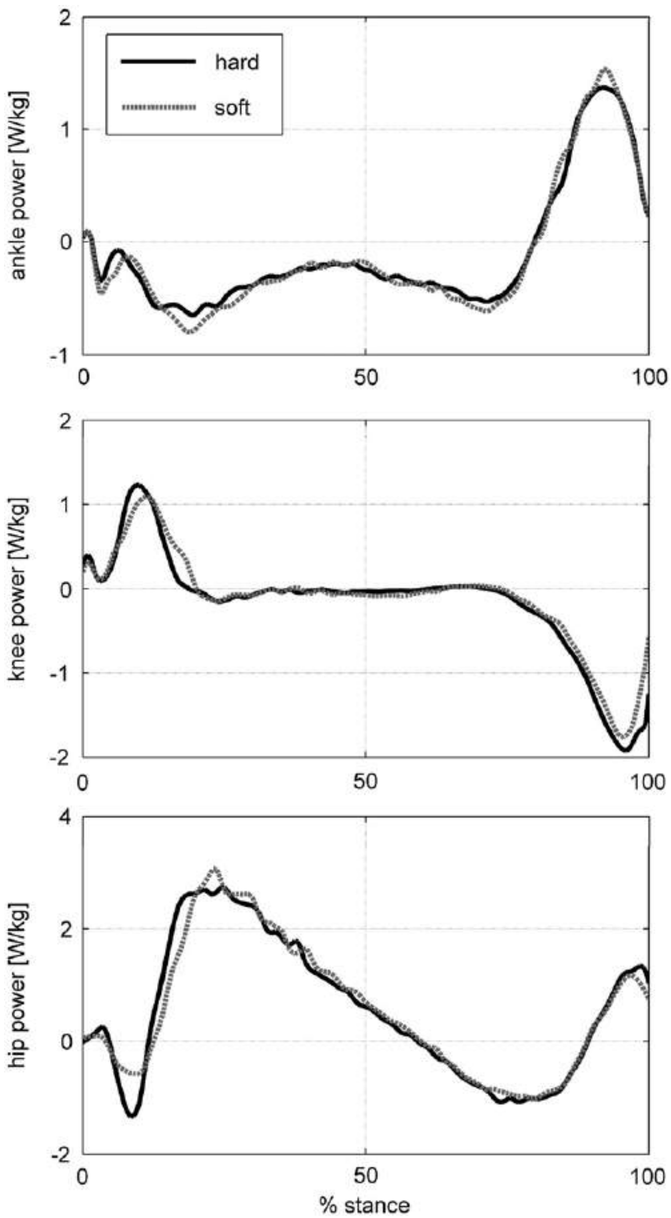


**Fig. 3.** Average joint powers ($n=15$) across the three axes of rotation (sagittal, frontal and transverse) during the stance phase of gait. Negative values correspond to eccentric, positive values to concentric joint power.

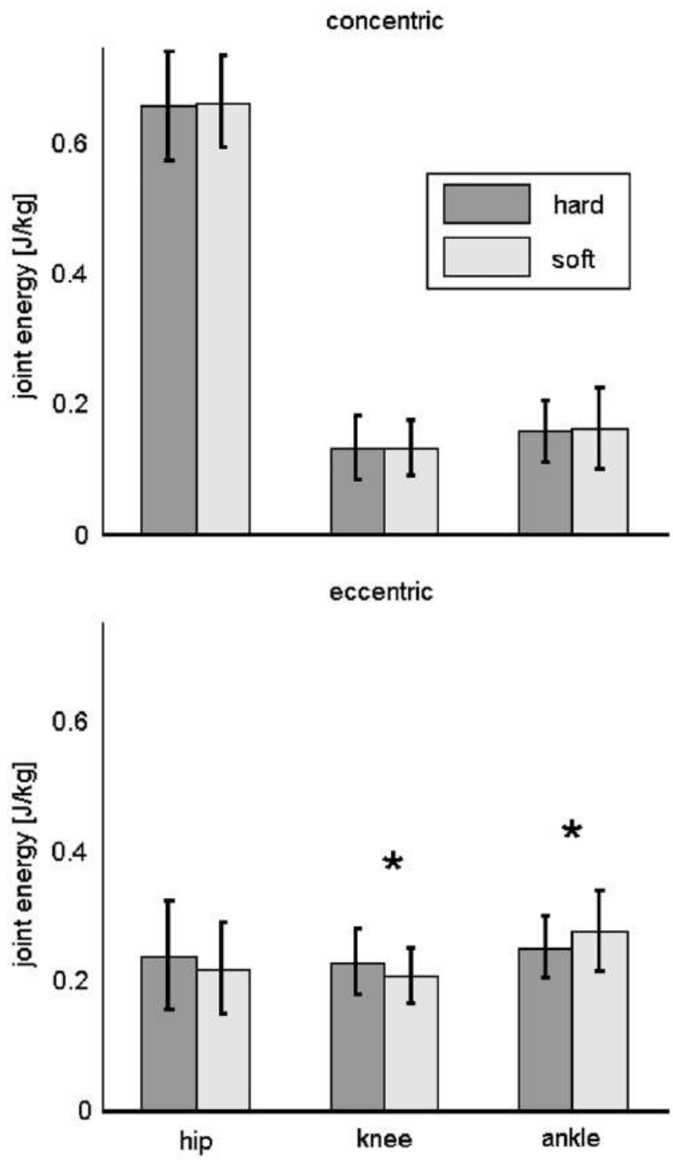


**Fig. 4.** Concentric and eccentric joint energies at the hip knee and ankle joint during the stance phase of gait.

walking technique might be one reason for the missing statistical significance.

Walking on uneven surface resulted in a reduced walking speed of 30% and step length of 21% compared to the study of Cikajlo and Matjacic (2007) conducted on similar population and similar conditions (boots and walkway length) on even surface. Since Cikajlo and Matjacic did not measure variability of temporo-spatial gait parameters, the change of variability in step width and step time cannot be discussed. Other studies on different populations under different environmental conditions reported greater or comparable variability of step width and step time (Menz et al., 2003; Richardson et al., 2008), so that a considerable increase in variability due to the gravel surface was not observed. For preferred walking speeds of 1.2 m/s on level ground, the hip, knee and ankle joint produces 45%, 20% and 35% of the summed ankle, knee and hip positive mechanical work, respectively, during each stride (Sawicki et al., 2009). In this study, the hip generates 62% of the positive mechanical work. This increased contribution of the hip joint might be explained with the weak coupling of the boot to the gravel surface, which decreases ankle positive energy. This decreased energy at the ankle joint might have to be compensated by increased hip moments (Lewis et al., 2008) changing gait from a push-off into a pull-off mode. Similar increased hip energies of 63% have been described for patients with pathological reduced plantar-flexor power (Sawicki et al., 2009), which supports the previous explanation.

In this study, the eccentric energy of the knee joint increased and the eccentric energy in the ankle joint decreased when wearing the harder boot. Cikajlo and Matjacic (2007) have shown for the shaft of military boots, that the hard shaft decreased concentric peak power at the ankle joint about 33%, but did not alter knee joint power curves. Their results suggest that the major influence of boot stiffness on kinematics and kinetics appears to be limited to the ankle joint. In this study, the ankle joint peak power is only reduced about 12% and boot shaft stiffness has a significant effect on the knee joint too. The lower reduction of ankle peak joint power with the hard boot shaft might be

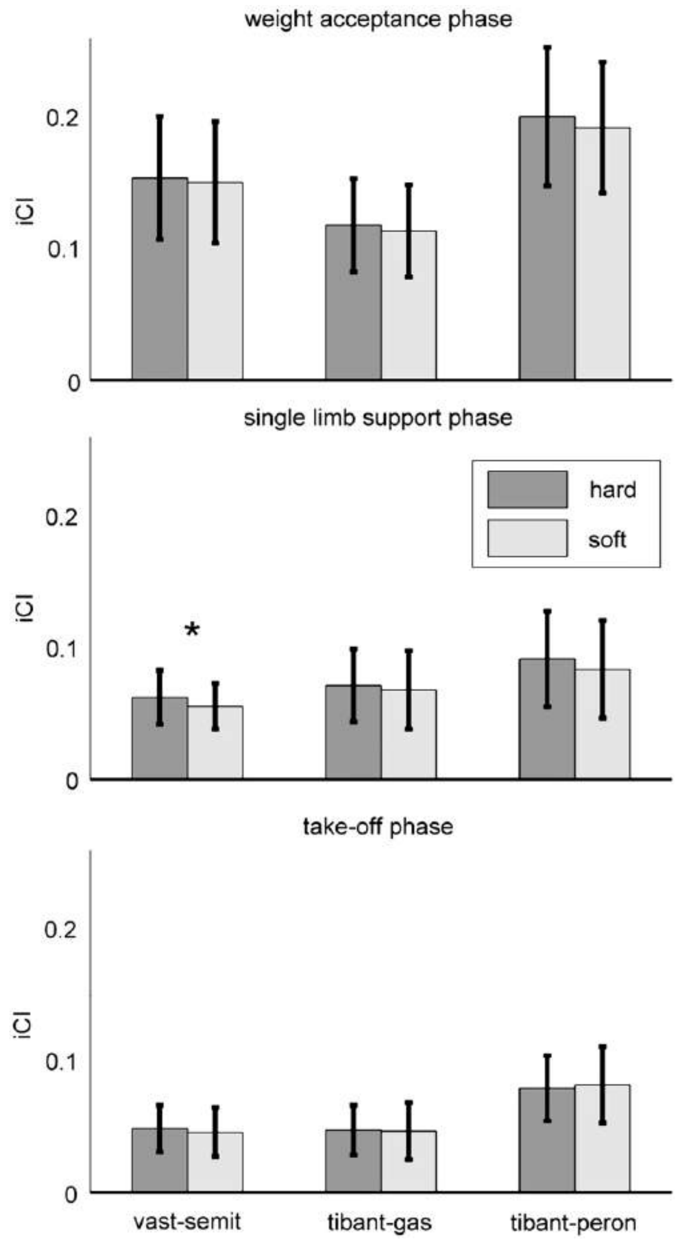


**Fig. 5.** Muscle co-contraction index for 3 antagonistic muscle pairs, integrated over three subsequent phases of gait during the stance phase of gait.

explained with the lower contribution of the ankle joint energy while walking on gravel surface, mentioned in the previous paragraph. If the additional effect on the knee joint observed in this study is an effect of the uneven surface, amplifying possible compensation mechanisms, or due to the extended measures performed in this study (EMG and joint energy) cannot be finally answered. Further studies comparing gravel to level ground surface walking would be required. In this study, co-contraction and eccentric energy at the knee joint increased significantly with the hard boot shaft. Since co-contraction increases joint stiffness, the increased co-contraction might be explained by the higher eccentric joint energy observed at the knee joint. At the ankle joint eccentric energy increased for the soft boot shaft. Since co-contraction at the ankle joint was not significantly different between boots shaft conditions, another reason might be, that increased ROM at the ankle joint enabled greater plantar flexion during weight acceptance. Consequently, the duration of the weight acceptance phase was significantly increased with soft boots (Table 1). Because the ankle joint acts eccentrically during the phase of weight acceptance (Fig. 3), the increased phase duration under eccentric conditions might have increased the joint energy in the soft shaft situation.

Nigg and Segesser (1992) advised that maximizing the energy return is not the adequate approach if one seeks to improve the performance of shoes. If one focuses on energy balance, the shoe should be judged by the amount of energy loss and absorptions. The concept of minimizing the energy loss attempts to reduce the energy that is spent unnecessarily and thus not able to enhance performance. Work loss by stabilizing the knee joint might be the main problem when walking with hard boot shafts on gravel surface. When the foot adapts to the surface via the subtalar ankle joint, which is enabled by a soft boot shaft, the knee needs to be stabilized to a lesser extent. A decrease in muscular effort minimizes redundant energy and therefore improves performance (efficiency of gait). In addition, the stronger co-contraction is likely to increase the compression forces inside the knee joint, so that a higher load was acting on the knee during walking with the hard boot shaft.

In this study, it was shown that the stiff boot shaft increased co-contraction and eccentric energy absorption at the knee joint. As a consequence the efficiency of gait might be decreased. This might cause early fatigue of knee muscles during walking or hiking. Therefore, stiffness and blocking of joint motion at the ankle should not be equated with safety. A trade-off between lateral stiffness and free natural motion of the ankle joint complex might be preferable. High-collar shoes with soft shaft stiffness might already increase tactile sensory feedback around the ankle to improve balance during challenging postural tasks (You et al., 2004).

## Conflict of interest statement

Both authors do not have any financial and personal relationships with other people or organizations that inappropriately influence the work performed.

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