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The influence of footwear on knee joint loading during walking — *in vivo* load measurements with instrumented knee implants

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ABSTRACT

Since footwear is commonly used every day, its influence on knee joint loading and thereby on the development and progression of osteoarthritis may be crucial. So far the influence of footwear has been examined only indirectly. The aim of this study was to directly measure the effect of footwear on tibiofemoral contact loads during walking.

Instrumented knee implants with telemetric data transmission were used to measure the tibiofemoral contact forces and moments in six subjects. The loads during walking with four different shoes (basic running shoes, advanced running shoes, classical dress shoes and shoes with a soft rounded sole in the sagittal plane (MBT)) were compared to those during barefoot walking. Peak values of all six load components were analyzed.

In general, footwear tended to increase knee joint loading slightly, with the dress shoe being the most unfavorable type of footwear. At the early stance phase all load components were increased by all shoe types. The resultant force rose by 2–5%, the internal adduction moment by 7–12% and the forces on the medial compartment by 3–5%. Significant reductions of the resultant force were solely observed for the advanced running shoe (−6%) and the MBT (−9%) shoe at late stance. Also the medial compartment force was slightly yet non-significantly reduced by 2–5% with the two shoes. It is questionable whether such small load changes have an influence on the progression of gonarthrosis. Future research is necessary to examine which factors regarding the shoe design, such as heel height, arch support or flexibility are most decisive for a reduction of knee joint loading.

1. Introduction

Footwear protects and supports the foot. Furthermore, attempts have been made to develop footwear with beneficial effects on joint loading in the lower extremities, on posture or back pain. Especially for sport shoes various damping and cushioning elements have been developed to reduce loading of the lower extremities. Since joint loading is related to the progression of osteoarthritis (OA) (Andriacchi et al., 2004; Englund, 2010; Heijink et al., 2012), footwear has furthermore become one aspect in the therapeutic treatment of OA patients.

Knee OA most often affects the medial compartment, which is assumed to be exposed to higher forces than the lateral one (Jackson et al., 2004). The external adduction moment (EAM) is related to OA severity, progression and pain (Miyazaki et al., 2002; Sharma et al., 1998) and often serves as an indirect measure for the medial compartment force (F_{med}). Several studies have shown that wearing shoes leads to increased instead of decreased EAMs (Radzimski et al., 2012). During walking with conventional walking shoes the EAM was increased by 7.4% (Kemp et al., 2008) to 11.9% (Shakoor and Block, 2006) compared to barefoot walking. Similarly, athletic footwear increased the EAM by 9.7% (Keenan et al., 2011). An increase of even 15% was observed with clogs and a 'stability shoe', no increase was seen with flat walking shoes or flip–flops (Shakoor et al., 2010). Furthermore, high heeled shoes are generally assumed to increase joint loading. The EAM rose by 9–14% with 3.8 cm high heels (Kerrigan et al., 2005) and even by 23% with 6 cm high heels (Kerrigan et al., 1998).

To mimic the barefoot situation a 'mobility shoe' with a flexible sole without heel was developed (Shakoor et al., 2008). In comparison to conventional walking shoes, these shoes resulted in 8% lower peak EAMs. Similar reductions of 11–16% were observed with another flexible, non-heeled shoe (Trombini-Souza et al., 2011). It contained however a 3 mm high internal wedge which might have had an additional effect on the EAM.

Recently the 'MBT' (Massai Barefoot Technology) shoe, a shoe with a soft sole rounded in anterior–posterior direction, was developed with the intention to increase muscle activation and to improve posture and balance. It has been shown that the 'MBT' shoe causes higher muscle activity and changes the movement

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patterns of the ankle (B. Nigg et al., 2006; Romkes et al., 2006). So far, knowledge about the effect of the 'MBT' shoe on knee joint loading is limited. Buchecker et al. (2012) showed that the first EAM peak is decreased by 12% in overweight males, compared to conventional shoes. Furthermore B.M. Nigg et al. (2006) showed that pain of OA patients can be reduced when walking with the 'MBT' shoe. However the same pain reduction was found for the conventional control shoe within a 12-week period.

Even though the EAM is commonly taken as a surrogate measure for medial compartment load, profound evidence about the direct correlation is still lacking. Only a few studies with controversial results have reported on the correlation between EAM and F_{med} in a single subject (Erhart et al., 2010a; Walter et al., 2010; Zhao et al., 2007). In these studies EAMs were altered by a variable-stiffness shoe or gait modifications like walking speed, foot path and the use of walking poles. Whereas no correlation between the peak EAMs and F_{med} was found by Zhao et al. (2007), others reported a correlation solely for the first peak EAM at the early stance phase (Erhart et al., 2010a) or solely for the second peak EAM at the late stance phase (Walter et al., 2010). Due to these inconsistencies it is unclear to what extent the commonly used indirect measures are able to predict changes in joint loading.

The aim of this study was to measure the influence of footwear on tibiofemoral contact forces and moments directly in vivo using instrumented knee implants.

2. Methods

2.1. Instrumented implant

An instrumented tibial tray with telemetric data transmission (Fig. 1) was used to measure all 6 components of knee contact forces and moments with a mean measurement error $<2\%$ in vivo (Heinlein et al., 2007). It is based on the INNEX FIXUC total knee system (Zimmer GmbH, Winterthur, Switzerland) with a standard femoral component and a standard ultra-congruent tibial insert. The tibial component is modified to enable a slight deformation of an inner stem which is measured by 6 semi-conductor strain gages (KSP 1-350-E4, Kyowa, Japan). The signals are sensed and transmitted by an inductively powered telemetry circuit (Graichen et al., 2007). The right-handed coordinate system of the implant is fixed at the right tibia. Its origin lies on the extended stem axis at the height of the lowest part of the tibial insert. Forces and moment measured in left knees were transformed to the right side. The force components $+F_x$, $+F_y$ and $+F_z$ act in lateral, anterior and superior directions onto the tibial component. The resultant force F_{res} consists of all 3 force components. The moments $+M_x$, $+M_y$ and $+M_z$ act in the sagittal, frontal and horizontal plane of the tibia and turn right around their belonging axes. The contact force F_{med} acting on the medial plateau of the tibia was calculated using the axial force F_z, the moment M_y in the frontal plane and the distance l between the femoral condyles (Kutzner et al., 2011).

$$F_{med} = (-F_z/2) - (M_y/l)$$

Forces are stated as percentage of bodyweight (%BW), moments as %BW times meter (%BWm).

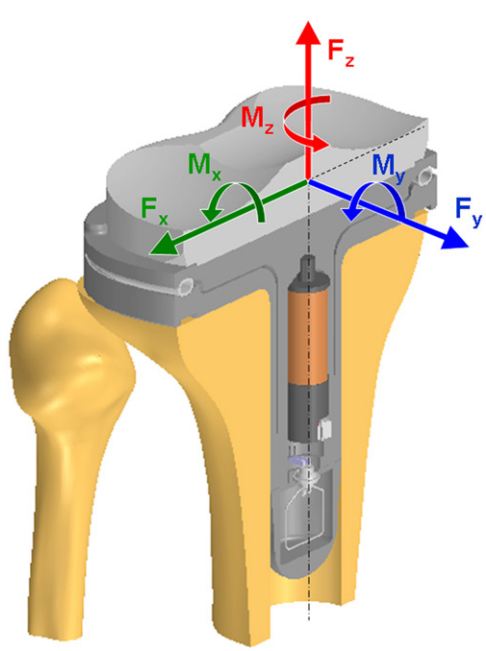

Fig. 1. Coordinate system of the instrumented tibial tray.

2.2. Footwear and subjects

Four different shoes were investigated and compared to barefoot walking (Fig. 2):

'Basic' shoe: basic, low cost running shoe (Adidas Duramo).
'High Level' shoe: advanced, high cost running shoe (Adidas Cushion 6).
'MBT' shoe: shoe with a soft rounded sole in anterior–posterior direction (MBT Tariki).
'Dress' shoe: classical man's (Rieker) or woman's shoe with a 2.5 cm high heel.

After obtaining approval of the ethics committee and the subjects' informed consent, 6 subjects with instrumented knee implants participated in this study (Table 1). Three weeks prior to the measurements, the subjects received the shoes (except for the 'MBT') and were asked to wear them as often as possible. With the 'MBT' shoe the patients exercised for about 15 min before measurements were taken. Data were captured during treadmill walking at a constant velocity of 4 km/h. The different shoes were worn in a random order.

2.3. Data evaluation

After short adaptation periods of about 4 min for each shoe, 20–30 gait cycles per subject and shoe were recorded. For each gait cycle the following distinctive peak forces and moments were identified (Fig. 3):

- Peak *resultant forces* (F_{res}) immediately before heel strike (HS peak), at contralateral toe off (1st peak) and at contralateral heel strike (2nd peak).
- Peak *medial* forces (F_{med}) at early (1st peak) and late stance (2nd peak).
- Peak *shear forces* in lateral ($+F_x$), medial ($-F_x$), anterior ($+F_y$) and posterior ($-F_y$) direction during one gait cycle. The instants at which peak shear forces occur within one gait cycle differ individually.
- Peak *flexion moments* ($+M_x$) at early (1st peak) and late stance (2nd peak).
- Peak *adduction moments* ($-M_y$) at early (1st peak) and late stance (2nd peak).
- Peak external *rotational moments* ($+M_z$) at early stance (1st peak) and internal rotational moments ($-M_z$) at late stance (2nd peak).

Mean peak values from repeated trials were calculated per subject. The load differences between barefoot and shod walking were analyzed statistically using a paired t-test (SPSS Inc.). Due to multiple comparisons ($k=4$), Bonferroni adjustment was applied. The significance level α of 0.05 was reduced to 0.0125 (α/k). In the following, all numbers refer to median values of peak loads if not stated otherwise.

3. Results

3.1. Resultant force

During barefoot walking, peak resultant forces F_{res} of 102%BW (HS peak), 231%BW (1st peak) and 263%BW (2nd peak) were measured. Both the HS peak and the 1st peak of F_{res} increased when walking with shoes (Table 2, Fig. 4). However, only the increase at HS with the 'Dress' shoe was found to be significant. The 2nd force peak increased only with the 'Dress' shoe. Significant force reductions were observed with the 'High Level' shoe and the 'MBT'.

3.2. Medial force

During barefoot walking, peak medial forces F_{med} of 190%BW (1st peak), and 182%BW (2nd peak) were measured. Whereas the

Fig. 2. Investigated shoes. From left to right: 'Basic' running shoe, 'High Level' running shoe, man's or woman's 'Dress' shoe, 'MBT' shoe.

Table 1
Subject data.

Subject	K1L	K2L	K3R	K5R	K7L	K8L	Average
Gender	Male	Male	Male	Male	Female	Male	–
Age (years)	64	72	71	61	76	71	69
Body mass (kg)	106	90	96	97	69	80	90
Height (cm)	177	171	175	175	166	174	173
Time post-op. (months)	18	16	7	17	31	12	17

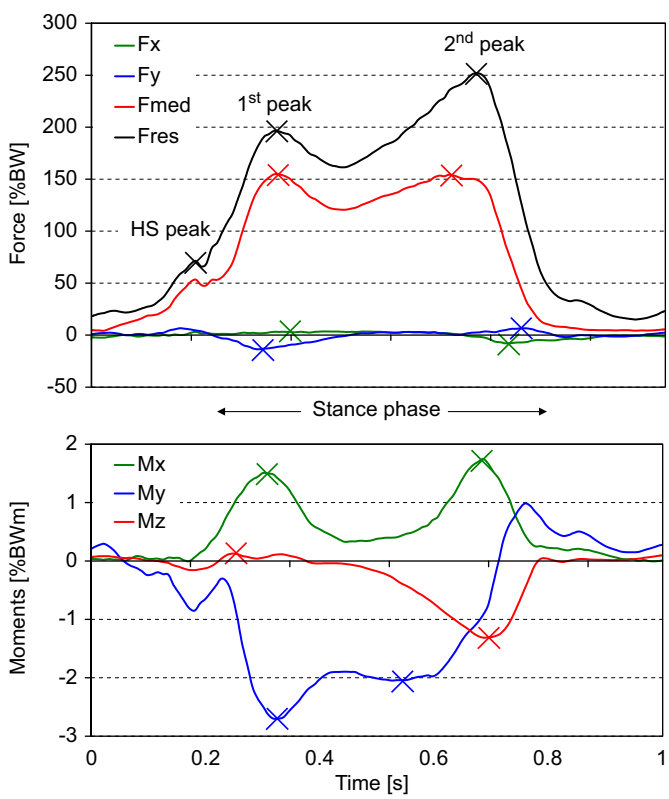

Fig. 3. Identified peak forces and moments within one gait cycle. Exemplary trial from one subject during barefoot walking.

1st peak increased slightly during shod walking, the 2nd peak increased only with the 'Dress' shoe but decreased with the 'High Level' and 'MBT' shoe (Fig. 5, Table 2). All these changes were, however, not significant.

3.3. Shear forces

During barefoot walking, peak forces of 6%BW in lateral ($+F_x$) and 11%BW in medial direction ($-F_x$) were measured. However, among the subjects, the force magnitudes and patterns varied extremely. Individual peak values were up to 5 times higher than the median values of all subjects. Even though not significant, peak

Table 2
Changes of peak forces during shod walking compared to barefoot walking. Median values in percent (%).

Forces	Basic	High Level	Dress	MBT
F_{res} (HS peak)	+14	+15	+17*	+13
F_{res} (1st peak)	+5	+2	+2	+3
F_{res} (2nd peak)	−3	−6*	+4	−9*
F_{med} (1st peak)	+3	+3	+4	+5
F_{med} (2nd peak)	0	−2	+8	−5
$+F_x$ (lateral)	+17	+15	+11	+21
$-F_x$ (medial)	+11	+2	+13	+1
$+F_y$ (anterior)	+5	+15	+19	+14
$-F_y$ (posterior)	+15*	+12*	+19*	+6

* Significant changes ($p < 0.0125$).

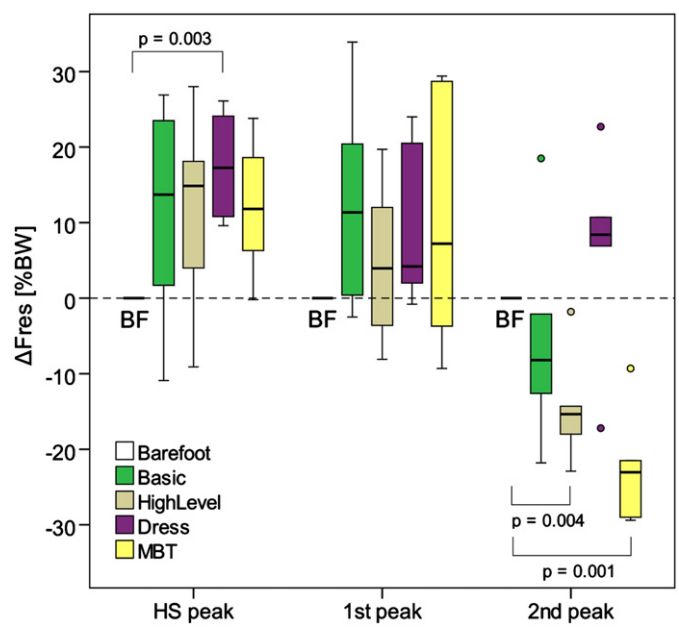

Fig. 4. Differences of peak resultant forces (ΔF_{res}) between shod and barefoot (BF) walking. p-values only stated if $p < 0.0125$.

forces increased in both medial and lateral direction during shod walking (Table 2).

Peak forces in anterior ($+F_y$) and posterior ($-F_y$) direction during barefoot walking were 10%BW and 23%BW, respectively. Both increased during shod walking. A significant increase of the posterior force was observed with the 'Basic' ($p=0.012$), 'High Level' ($p=0.003$) and the 'Dress' shoe ($p=0.001$).

3.4. Moments

During barefoot walking peak flexion moments ($+M_x$) of 1.7%BWm (1st peak) and 1.6%BWm (2nd peak) were observed. Whereas M_x increased with all shoes at early stance, it decreased at late stance with all shoes except for the 'Dress' shoe (Table 3). Nevertheless, none of these changes were found to be significant.

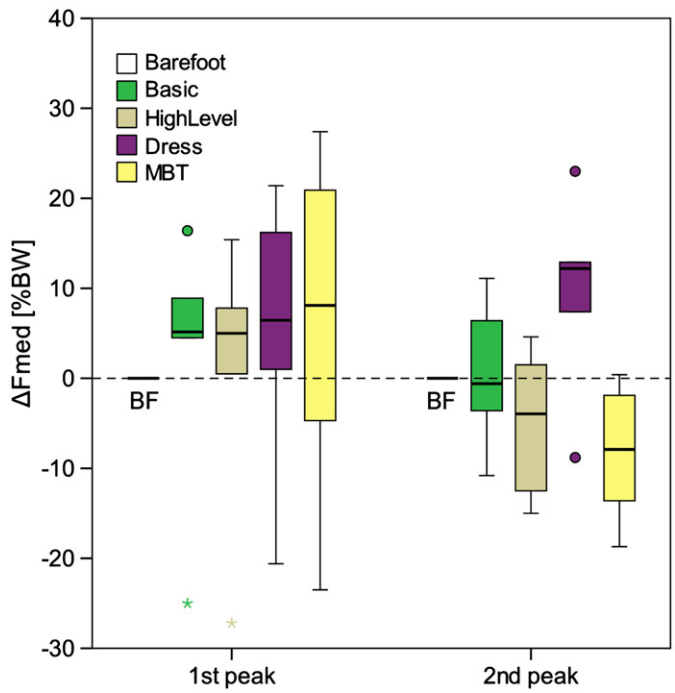

Fig. 5. Differences of peak medial forces (ΔF_{med}) between shod and barefoot (BF) walking.

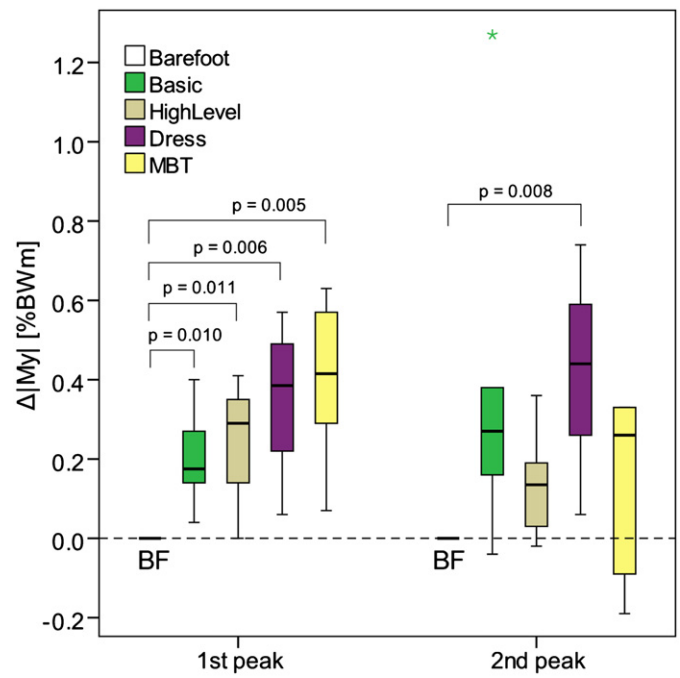

Fig. 6. Differences of peak adduction moments (ΔM_y) during shod and barefoot (BF) walking. p-values only stated if $p < 0.0125$.

Table 3
Changes of peak moments during shod walking compared to barefoot walking. Median values in percent (%).

Moments	Basic	High Level	Dress	MBT
$+M_x$ (1st peak)	+4	+4	+6	+2
$+M_x$ (2nd peak)	−6	−6	+1	−9
$-M_y$ (1st peak)	+7*	+9*	+11*	+12*
$-M_y$ (2nd peak)	+9	+8	+19*	+8
$+M_z$ (external rot)	+31	+28	+28	+8
$-M_z$ (internal rot)	−6	−18	−1	−19

* Significant changes ($p < 0.0125$).

Peak adduction moments ($-M_y$) of 3.4%BWm (1st peak) and 2.9%BWm (2nd peak) were measured during barefoot walking. The 1st peak increased significantly with all shoes (Fig. 6, Table 3). The highest increase was observed with the 'MBT' shoe, followed by the 'Dress', 'High Level' and 'Basic' shoe. The 2nd peak of the adduction moment was only significantly influenced by the 'Dress' shoe. However, all shoes also showed a trend towards increased 2nd peak adduction moments.

External rotational moments ($+M_z$) of 0.4%BWm and internal moments ($-M_z$) of 1.0%BWm were determined during barefoot walking. Whereas external moments increased, internal moments decreased during shod walking (Table 3). Since the peak values of M_z were small in comparison to those of the other moment components, small absolute changes of the moment $+M_z$ resulted in high relative changes of up to 31%. However, all changes of M_z were not significant.

4. Discussion

Footwear generally tended to increase most force and moment components of the tibiofemoral loading during walking. Highest loads were generally observed with the 'Dress' shoe, which had the highest—though moderate—heel height of the investigated footwear. With this shoe, shear forces, resultant forces and adduction moments increased significantly, but also all other load components showed a trend towards higher values. OA patients are commonly advised not to wear high heeled shoes (Kerrigan et al., 2005). The findings of this study tend to support this advice. Since increased instability is assumed to cause higher joint loads (Bergmann et al., 2004) greater heel heights and a narrower heel bases might cause even higher joint loads. However, in this study the investigated footwear did not solely differ in heel height. Various design factors of footwear such as heel width, arch support, sole shape, flexibility or cushioning might influence joint loading. In order to examine the effect of heel height on joint loading further investigations are necessary which address the effect of this single parameter.

An increase of the medial contact force is especially detrimental for patients with medial compartment OA. Previous studies have shown that the external adduction moment is generally increased when walking with shoes compared to barefoot walking (Radzimski et al., 2012). It is assumed that the non-natural arch support in shoes is a reason for increased EAMs. Franz et al. (2008) showed that arch support cushions lead to an increase of the EAM of 6% during walking. Shakoor et al. (2010) found that the EAM increases by about 15% when walking with clogs or 'stability shoes', compared to barefoot walking, but not when walking with flip–flops, which have no arch support at all. These conclusions are only partly supported by our results. F_{med} increased slightly with all shoes by 3–5% at early stance. At late stance, however, only the 'Dress' shoe led to an increase of 8%. Whereas no change was seen with the 'Basic' shoe, the other two shoes even decreased F_{med} slightly by 2–5%. Similar results were found for the resultant force F_{res} but only the slight reductions of 6% ('High Level') and 9% ('MBT') at late stance were significant. For all investigated subjects, the 2nd peak of F_{res} was higher than the first one. Load reductions at late stance might therefore be more relevant than at early stance. Whether small force reductions of only a few percent are sufficient to reduce the pain of OA patients or even slow the progression of OA is questionable though.

The 'MBT' shoe has a round-shaped sole and differs the most from conventional footwear. It has been shown that the 'instable' 'MBT' shoe causes increased muscle activities (Buchecker et al., 2012; Romkes et al., 2006). Muscle co-contractions, needed to

stabilize the joint, increase the joint contact force (Bergmann et al., 2004). Nevertheless and unexpectedly the 'MBT' shoe with its build-in 'instability' did not lead to a significant increase of the tibiofemoral contact forces during the stance phase. It has been shown that kinematics, especially the ankle movement, is altered when walking with the 'MBT' shoe (Romkes et al., 2006). Since the resultant force is decreased during the late stance phase of gait, it must be assumed that changed gait patterns, rather than damping or cushioning elements of the shoe, lead to reduced joint loading.

This study is limited due to the small number of subjects investigated. Furthermore, the loading of a specific and relatively constraint implant was measured. The TKR design may influence tibiofemoral joint kinematics and therefore the joint contact forces. However, when assuming similar kinematics, the magnitude of the axial force is likely not affected by the chosen TKR system. In contrast, the shear forces acting at the implant are influenced by the design-dependent constraints of the tibial insert. In particular, the antero–posterior shear forces are probably smaller in the natural knee or cruciate retaining TKR designs, since they are partly transferred by the cruciate ligaments. However, we do assume that the load changes due to footwear are transferable. Moreover, the effects of shoes on knee joint loading were only investigated for walking; no statement about the effect of footwear during jogging can be made by this study.

In the current study only the short-term or immediate effect of footwear was evaluated. It is not clear whether an adaptation to the footwear may alter joint loading. No changes over time were found in a study by Hinman et al. (2009) where the effects of a laterally wedged shoe on the EAM were investigated over a period of 4 weeks. The subjects experienced a similar EAM reduction after 4 weeks as that observed at the baseline measurement. In contrast, a study of Erhart et al. (2010b) showed that the EAM reduction improves with long-term use of a variable-stiffness shoe. EAM reductions were higher after 6 months compared to the baseline measurements. Additional follow-up studies are needed to investigate the long-term effect of footwear.

In this study, the influence of footwear on knee joint loading during walking was moderate. Yet, shoes and especially the 'Dress' shoe clearly tended to increase the loads. Only the 'High Level' and the 'MBT' shoes decreased the resultant and the medial compartment force at late stance slightly. There are various design factors of footwear such as heel height and width, arch support, sole shape, flexibility or cushioning which might influence joint loading. Which factor is most decisive for a potential reduction of knee joint loading still has to be analyzed.

Conflict of interest statement

This study was funded by Deutsche Forschungsgemeinschaft (Be 804/18-1) and Zimmer GmbH. The funders had no role in study design, data collection and analysis, decision to publish, or preparation of the manuscript.

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