Medial and lateral foot loading and its effect on knee joint loading

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Background: The medial knee contact force may be lowered by modified foot loading to prevent the progression of unilateral gonarthrosis but the real effects of such gait modifications are unknown. This study investigates how walking with a more medial or lateral rollover of the foot influences the in vivo measured knee contact forces.

Methods: Five subjects with telemetered knee implants walked on a treadmill with pronounced lateral or medial foot loading. Acoustic feedback of peak foot pressure was used to facilitate the weight bearing shift. The resultant contact force, Fres, the medial contact force, Fmed, and the force distribution Fmed/Fres across the tibial plateau were computed from the measured joint contact loads.

Findings: During lateral foot loading, the two maxima of Fres during the stance phase, Peak 1 and Peak 2, increased by an average of 20% and 12%, respectively. The force distribution was changed by only −3%/+2%. As a result, Fmed increased by +16%/+17%. Medical foot loading, on the other hand, changed Fres only slightly, but decreased the distribution by −18%/.11%. This led to average reductions of Fmed by −18%/−18%. The reductions were realized by kinematic adaptations, such as increases of ankle eversion, step width and foot progression angle.

Interpretation: Medial foot loading consistently reduced the medial knee compartment, and may be a helpful gait modification for patients with pronounced medial gonarthrosis. The increase of Fmed during lateral foot loading was mostly caused by muscular co-contractions. Long-term training may lead to more efficient gait and reduce co-contractions.

1. Introduction

Knee osteoarthritis (OA) is one of the most frequent orthopedic diseases, leading to pain, high economical costs, and reduced life quality. During walking, 60–80% of the contact force, −Fz, which acts distally in the direction of the tibia (Fig. 1), is transferred by the force, Fmed, which acts at the medial knee compartment (Halder et al., 2012). Therefore, OA mostly affects the medial knee compartment (Ahlbäck, 1968; Felson and Radin, 1994). In order to decelerate OA progression and reduce pain, treatments aim to shift −Fz more to the lateral compartment, thus reducing Fmed.

Due to the difficulties in measuring Fmed directly, several studies have used the external adduction moment (EAM) as an indirect measure. The EAM is mainly determined by the ground reaction force (GRF) and its lever arm (d) relative to the knee joint center, calculated in the frontal plane of the tibia (Fig. 2A). Hence, the EAM decreases if either the magnitude of the GRF is lowered, or its lever arm is decreased. It is mostly assumed that a reduced EAM is equivalent to a lateral shift of −Fz, and thus a smaller force Fmed.

A non-surgical approach to lower the EAM is to modify foot loading during walking. Studies have investigated the effect of passive methods, such as laterally wedged shoes, on the EAM, with partially controversial results (Hinman et al., 2008, 2009, 2012; Kutzner et al., 2011; Segal, 2012). A common belief among orthopedists is that the lateral wedge shifts the center of pressure (COP) to the lateral side of the foot and thus also shifts −Fz laterally (Fig. 2A). Another research group aimed to lower the EAM using shoes with higher stiffness on the lateral side, thus also reducing the EAM (Jenkyn et al., 2011). In contrast to this assumption, a medial shift of the COP was found but the EAM was still lowered: the medio-lateral (m-l) component of the GRF became smaller and the combined effects (shift of the GRF/COP shift to the medial side of the foot and lateral tilt of the GRF vector) were realized and reduced the EAM. Kinematic gait adaptations in the frontal plane were also found with that type of shoe (Boyer et al., 2012; Erhart et al., 2008). A medio-lateral movement of the center of mass (CoM) and the resulting acceleration forces could also cause a tilt of the GRF vector (Jenkyn et al., 2011) and change d (Fig. 2B). The variable stiffness shoe was also tested in an older subject with instrumented knee prosthesis, and average reductions of −22% in the EAM and −19% in Fmed were found throughout the stance phase (Erhart et al., 2010a). In contrast to the controversial results for lateral shoe wedges, the shoe also led to significant reductions in pain (Erhart et al., 2010b).

Another study investigated the effect of medial foot loading on EAM, controlled by vibration feedback (Dowling et al., 2010). The first peak of
the EAM during the stance phase was reduced by $-14.2\%$. Wheeler et al. (Wheeler et al., 2011) attempted to reduce the EAM during gait under feedback control and let the subjects choose the kinematic adaptations. Medial foot loading was one of the most frequently used adaptations. The EAM was then lowered by $-3\%$ to $-50\%$.

However, whether EAM is a reliable surrogate measure for $F_{med}$ has not yet been fully clarified. Studies with instrumented knee implants have revealed limitations for using EAM as predictor for $F_{med}$ (Kutzner et al., 2013; Trepczynski et al., 2014). One of the reasons seems to be that antagonistic muscle activities, which raise the joint contact force, are not reflected in gait analysis data.

Despite numerous investigations, there is still no consensus whether a lateral or medial shift of the COP would unload the medial compartment and the underlying mechanisms have not fully been revealed.

The aim of this study was therefore to determine how altered foot loading influences the medial knee joint loading in reality. The effect of medial and lateral foot loading on the knee joint loads was measured in vivo by using instrumented knee implants with telemetric data transmission.

2. Methods

2.1. Subjects

Five patients (anthropometrics see Table 1) were investigated. The clinical study was approved by the Charité Ethics Committee, under the registry number EA2/057/09, and is registered at the ‘German Clinical Trials Register’ (DRKS00000563). All patients gave written informed consent to participate in this study.

2.2. Instrumented implants

The knee joint forces and moments were measured in vivo by instrumented knee implants with inductive power supply and telemetric data transmission (Fig. 2). The technical details have been described previously (Heinlein et al., 2007). An inner cavity in the stem of the clinically proven tibial tray (INNEX, Zimmer GmbH, Winterthur, Switzerland) was equipped with 6 strain gauges for measuring the 3 force and 3 moment components, with a measurement error below $2\%$. The coordinate system is located on the extended stem axis at the lowest point of the inlay of the implant. The force component, $F_x$, acts laterally, $F_y$ acts anteriorly, and $-F_z$ acts distally along the stem axis.

The medio-lateral force distribution between the two tibial compartments is specified by the ‘Medial Ratio’ (MR), which represents $F_{med}$ as a percentage of $-F_z$ (Heinlein et al., 2009).

$$MR = \frac{F_{med} \times 100}{-F_z}$$

All forces were computed in percentage of the patient’s body weight [%BW]. The data from the left implants were mirrored to the right side. The resultant force, $F_{res}$, was determined from its 3 force components as follows:

$$F_{res} = \sqrt{F_x^2 + F_y^2 + F_z^2}.$$  \hspace{1cm} (1)

The medial knee contact force, $F_{med}$ (Fig. 1, right), can be computed from $-F_z$, the moment $M_y$, which acts around the a-p axis, and the distance, $l$, between the two femoral condyles.

$$F_{med} = \frac{-F_z - M_y}{l}.$$  \hspace{1cm} (2)

The computation was described in detail previously (Halder et al., 2012) and tests showed an error of less than $3\%$ if the axial force is $>1000$ N. Therefore, $F_{med}$ is computed only for the stance phase when $-F_z > 1000$ N is given.

The medio-lateral force distribution between the two tibial compartments is specified by the ‘Medial Ratio’ (MR), which represents $F_{med}$ as a percentage of $-F_z$ (Heinlein et al., 2009).

Table 1: Patient data.

<table>
<thead>
<tr>
<th>Patient</th>
<th>K1L</th>
<th>K3R</th>
<th>K5R</th>
<th>K7L</th>
<th>K8L</th>
<th>Average</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age [years]</td>
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<td>75</td>
<td>63</td>
<td>78</td>
<td>74</td>
<td>72</td>
</tr>
<tr>
<td>Body mass [kg]</td>
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<td>97</td>
<td>90</td>
<td>68</td>
<td>79</td>
<td>86</td>
</tr>
<tr>
<td>pOP time [months]</td>
<td>70</td>
<td>58</td>
<td>51</td>
<td>51</td>
<td>45</td>
<td>55</td>
</tr>
<tr>
<td>TFA</td>
<td>varus</td>
<td>varus</td>
<td>varus</td>
<td>varus</td>
<td>varus</td>
<td>varus</td>
</tr>
</tbody>
</table>

Age at measurement, pOP = postoperative time. TFA = tibio-femoral angle.

Fig. 1. Instrumented tibial tray of the knee endoprosthesis.

Fig. 2. Mechanisms for altering the EAM. (A) The external adduction moment (EAM) is mainly determined by the magnitude of the ground reaction force vector (GRF, black arrow) and its lever arm, $d$, to the knee joint center. A basic model suggests that a lateral shift of the COP (green arrow) shortens $d$. (B) A more complex model considers that a dynamic medio-lateral shift of the center of mass (CoM) may cause additional acceleration forces, which influence the inclination of the GRF (dotted arrows). This would have an additional influence on $d$ and the EAM.
An MR value of 70%, for example, indicates that 70% of the vertical force, $-F_z$, is transferred by the medial knee compartment. A decrease of MR therefore implies a force shift from the medial towards the lateral compartment.

2.4. Motion analysis

A 10-camera infrared motion analysis system (VICON Metrics, Oxford, UK) was used to quantify the kinematics at a rate of 120 Hz. A total of 55 reflective markers were attached to the subject’s skin. From the marker coordinates, the functional joint centers and axes were determined (Taylor et al., 2010; Heller et al., 2011). Six gait parameters (see below) were computed throughout the entire gait cycles to determine how the modified foot loading was realized by the subjects. Measurements were taken on a treadmill; therefore, the ground reaction force GRF, its lever arm to the knee joint and the EAM could not be measured.

2.5. Activity and acoustic feedback

In order to facilitate a more medial or lateral foot loading, a resistive pressure sensor (Interlink Electronics, Type FSR 402, 18 mm diameter) was embedded in an insole (Fig. 3). The sensor had been tested using a 2 mm silicone sheet under the sensor and a 1 mm sheet between the sensor and the upper metal loading plate. Within a load range of 0–350 N, output voltage and load deviated by less than 15% from a linear behavior.

![Shoe sensor. A resistive pressure sensor was embedded in an insole under the fifth metatarsal head.](image)

Our main intention was to affect the second peak of $F_{med}$ during the late stance phase because its magnitude is higher than the first peak in almost all our patients (Halder et al., 2012; Bergmann et al., 2014). This second peak typically occurs when the metatarsal is loaded at the instant of contralateral heel strike (Perry and Burnfield, 2010; Vaughan et al., 1999; Chiu et al., 2013). The location of the sensor under the metatarsal head was tested with a healthy, elderly subject. The subjects first stood barefoot on an insole and the location under the 5th metatarsal head was marked. Then, the pressure sensor was embedded into the insole. The pressure signal was displayed on a monitor. At a selectable pressure level, a beep tone was triggered.

The subjects first walked ‘normal’ without instructions on the treadmill at 2.5 km/h, and the average pressure level was set as baseline. For achieving more lateral foot loading, the threshold was then increased by 20%, and the subjects were verbally instructed to roll the foot on the lateral side and thereby trigger a short beep during every step. We did not give instructions how exactly to achieve the beep tone and let the subjects choose their own strategy. Prerequisite was that the subjects would maintain a gait pattern that was as physiological and fluent as possible. More medial foot loading was accomplished by decreasing the threshold by 20%, and instructing the subjects to roll their foot on the inner side, thereby avoiding the beep tone. The 20% changes had been tested prior to the study to verify that it was accomplishable. Subject K3R was not able to realize medial foot loading and could not avoid the beep. The data is still included in the analysis. Prior to taking the measurements, the subject practised the changed gait patterns for at least 5 min until they were able to achieve the altered foot loading with every step.

2.6. Analysis of loads

30 gait cycles per subject and foot loading condition were included in the analysis except for K3R, for whom the kinematic data during normal foot loading allowed the analysis of only 20 cycles. The time courses of $F_{med}$ and MR were calculated (Eqs. (2), (3)) during stance phase when $-F_z$ was higher than 1000 N. To illustrate their behavior with time, the patterns from all trials of the single subjects were averaged, using a time warping procedure which retains the peak values (Bender and Bergmann, 2011).

For each gait cycle, the 2 maxima (‘Peak 1’ and ‘Peak 2’ in the following) of $F_{res}$ were identified, and the 3 ‘load values’ $F_{res}$, $F_{med}$ and MR were determined at the times of these force peaks. If $F_{res}$ did not exhibit two distinct peaks, the load values were determined at the times when the peaks usually occur during normal foot loading. Thus, we obtained 30 values of $F_{res}$, MR and $F_{med}$ per subject and foot loading condition for each of the 2 peaks.

These numerical values were averaged intra-individually for each foot loading condition. The reported data from single subjects refer to these intra-individual mean values. The load changes, $\Delta F_{res}$, $\Delta MR$ and $\Delta F_{med}$, between normal walking and walking with medial or lateral foot loading, were computed as percentage of the values during normal walking. The mean values from all steps of all subjects were then averaged again.

2.7. Definition of gait parameters

The following 6 kinematic parameters were determined from the gait data:

- **SW** step width = the distance between the medio-lateral ankle midpoints
- **TLA** trunk lean angle, the angle defined by a vector determined by the midpoint of the shoulder and pelvis markers in the frontal plane and a vertical line, positive if the trunk is leaned towards the ipsilateral side
- **AIA** ankle inversion angle, positive if the medial foot side is raised
FPA the foot progression angle, positive if the toe points outwards
LAA leg adduction angle, the angle between the longitudinal axis of leg and a vertical line
m-l COM the maximal lateral amplitude of the m-l center of mass movement. COM is defined as the geometric midpoint of the pelvis markers

The parameters were determined at the instants of Peak 1 and Peak 2 (except for m-l COM, for which the maximum value during stance phase was determined), and averaged first intra- and then inter-individually. Then, their changes ∆ between normal and medial or lateral foot loading were determined. The correlations between these parameter changes and the 3 load value changes were calculated from the individual mean values. Values of R² above 0.5 were regarded to prove a good correlation between a certain parameter on one of the load values.

3. Results

Examples of measured contact loads during walking with normal and shifted foot loading are accessible from the public database, www.OrthoLoad.com.

3.1. Sensor pressure

The pressure sensor output during normal, lateral and medial foot loading is given in Table 2 for Peak 1 and Peak 2. Average values during normal walking were 0.87|1.33 V at Peak 1|Peak 2. All subjects were able to increase/decrease the pressure during lateral/medial foot loading except for K3R, who could not lower the pressure during medial loading. None of the subjects really changed the pressure by the demanded 20% compared to normal walking. The average signal changes from all subjects were much larger with +130%|+67% for lateral and −40%|−41% for medial loading at Peak 1|Peak 2.

3.2. Changes of joint contact loads%

The individual changes (∆) of Fres, MR and Fmed at Peak 1 and Peak 2, compared to normal walking, are shown in Fig. 4. All changes individually varied very much.

3.2.1. Lateral foot loading

Averaged across the 5 subjects, lateral foot loading increased Fres by 20%|12% for Peak 1|Peak 2. An increase was observed in 4 of 5 subjects. Only subject K3R exhibited a subtle decrease of Fres by −3%|−2%. MR was altered slightly by −3%|+2% on average. Fmed was increased by 16%|17% on average. Subject K3R was the one with the smallest increase of Fmed by 1%|2%.

3.2.2. Medial foot loading

Average changes of Fres by +5%|−3% were calculated. MR was consistently decreased in all subjects by −18%|−11% on average. Fmed was also decreased in all subjects, with average changes of −18%|−18%.

3.3. Load-time patterns

Fig. 5 shows exemplary load-time patterns of Fres, MR and Fmed of subject K7L. This subject was chosen because her data best demonstrates the combined effect of Fres and MR on Fmed. Lateral foot loading raised Fres substantially throughout the whole stance phase, as indicated by an arrow. MR was initially slightly higher than during normal walking, but smaller after Peak 1. Here, Fres was therefore shifted to the lateral compartment at Peak 1 and Peak 2. However, the strong increase of Fres overbalanced the small decrease of MR, resulting in a pronounced increase of Fmed during lateral foot loading throughout the whole stance.

<table>
<thead>
<tr>
<th>Subject</th>
<th>Sensor pressure [V]</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Peak 1</td>
</tr>
<tr>
<td></td>
<td>Lateral</td>
</tr>
<tr>
<td></td>
<td>Mean</td>
</tr>
<tr>
<td>K1L</td>
<td>2.56</td>
</tr>
<tr>
<td>K3R</td>
<td>1.37</td>
</tr>
<tr>
<td>K5R</td>
<td>1.82</td>
</tr>
<tr>
<td>K7L</td>
<td>2.10</td>
</tr>
<tr>
<td>K8L</td>
<td>2.14</td>
</tr>
<tr>
<td>Average</td>
<td>2.00</td>
</tr>
</tbody>
</table>

Pressure signal in Volt at Peak 1 and Peak 2 during lateral, normal and medial foot loading conditions. Means and standard deviations (SD) for each subject and averaged across all subjects (bold).
phases. Medial foot loading had no substantial effect on \( F_{\text{res}} \). MR became much smaller throughout the whole stance phase. The permanent decrease of MR plus the nearly unchanged pattern of \( F_{\text{res}} \) resulted in a pronounced decrease of \( F_{\text{med}} \).

### 3.4. Changes of kinematic parameters

The subjects realized the altered foot loading by adaptations of the kinematic parameters. Some of the 6 investigated parameters changed in the same direction in at least 4 of the subjects (Tables 3A, 3B), for example \( \Delta TLA \) and \( \Delta AIA \) at both peaks. However, both parameters increased for lateral as well as for medial foot support. This means that the same kinematic changes of ankle inversion and trunk lean were performed, both for increasing and decreasing the pressure at the sensor site. In a similar way, \( \Delta SW \) increased for lateral as well as for medial foot loading (except K3R) but the increase in SW was larger for medial foot loading.

Some of the parameters differed extremely between subjects. During medial foot loading, for example, \( \Delta SW \) increased by 9 mm in K3R, but by 11 mm in K8L, compared to normal walking.

In only a few cases, good correlations (\( R^2 > 0.5 \)) were found between changes of one of the 6 kinematic parameters and changes of a load value:

For lateral foot loading, these included the following:

- A higher \( \Delta TLA \) correlated with higher \( \Delta F_{\text{med}} \) (\( R^2 = 0.52 \)) at Peak 1.
- A reduced \( \Delta FPA \) correlated with higher \( \Delta F_{\text{res}} \) (\( R^2 = 0.68 \)) at Peak 1.
- A reduced \( \Delta FPA \) correlated with smaller \( \Delta MR \) (\( R^2 = 0.82 \)) at Peak 1.
- A higher \( \Delta SW \) correlated with higher \( \Delta F_{\text{med}} \) (\( R^2 = 0.63 \)) at Peak 2.
- A reduced \( \Delta SW \) correlated with higher \( \Delta F_{\text{med}} \) (\( R^2 = 0.67 \)) at Peak 2.

For medial foot loading, they included the following:

- A higher \( \Delta LAA \) correlates with higher \( \Delta F_{\text{res}} \) (\( R^2 = 0.68 \)) at Peak 1.

### 4. Discussion

The focus of this study was to gain a better understanding of how medial or lateral foot loading during gait influences the magnitude of the knee contact force and its distribution between the two tibial compartments.

The medial force \( F_{\text{med}} \) is a result of both, the total force magnitude \( F_{\text{res}} \), and the force distribution MR across the knee. Generally, an increase of \( F_{\text{res}} \) would cause an increase of \( F_{\text{med}} \); whereas a decrease of the MR, indicating a force shift towards the lateral compartment, would cause a decrease of \( F_{\text{med}} \). The resulting effect, whether \( F_{\text{med}} \) is decreased or increased, depends on the amount by which both parameters are changed.

Lateral foot loading caused small changes in the MR that were overcompensated by a pronounced increase in \( F_{\text{res}} \). The combined effect was an increase of \( F_{\text{med}} \). The raised resultant force is most likely due to

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### Table 3A

Changes of kinematic parameters at Peak 1, caused by lateral or medial foot loading.

<table>
<thead>
<tr>
<th>Peak 1 loading</th>
<th>( \Delta SW ) [mm]</th>
<th>( \Delta TLA ) ['']</th>
<th>( \Delta AIA ) ['']</th>
<th>( \Delta FPA ) ['']</th>
<th>( \Delta LAA ) ['']</th>
<th>( \Delta m-l \text{COM} ) [mm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>K1L</td>
<td>22</td>
<td>61</td>
<td>2</td>
<td>2</td>
<td>14</td>
<td>3</td>
</tr>
<tr>
<td>K3R</td>
<td>-13</td>
<td>9</td>
<td>1</td>
<td>2</td>
<td>0</td>
<td>1</td>
</tr>
<tr>
<td>K5R</td>
<td>36</td>
<td>58</td>
<td>1</td>
<td>1</td>
<td>12</td>
<td>2</td>
</tr>
<tr>
<td>K7L</td>
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<td>67</td>
<td>1</td>
<td>5</td>
<td>5</td>
<td>2</td>
</tr>
<tr>
<td>K8L</td>
<td>18</td>
<td>13</td>
<td>2</td>
<td>-2</td>
<td>5</td>
<td>3</td>
</tr>
<tr>
<td>Average</td>
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<td>62</td>
<td>1.4</td>
<td>1.7</td>
<td>7</td>
<td>2</td>
</tr>
</tbody>
</table>

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### Table 3B

Changes of kinematic parameters at Peak 2, caused by lateral or medial foot loading.

Explanations see Table 3A.

<table>
<thead>
<tr>
<th>Peak 2 loading</th>
<th>( \Delta SW ) [mm]</th>
<th>( \Delta TLA ) ['']</th>
<th>( \Delta AIA ) ['']</th>
<th>( \Delta FPA ) ['']</th>
<th>( \Delta LAA ) ['']</th>
</tr>
</thead>
<tbody>
<tr>
<td>K1L</td>
<td>31</td>
<td>68</td>
<td>0</td>
<td>2</td>
<td>14</td>
</tr>
<tr>
<td>K3R</td>
<td>-4</td>
<td>9</td>
<td>1</td>
<td>1</td>
<td>0</td>
</tr>
<tr>
<td>K5R</td>
<td>24</td>
<td>68</td>
<td>1</td>
<td>1</td>
<td>7</td>
</tr>
<tr>
<td>K7L</td>
<td>23</td>
<td>80</td>
<td>1</td>
<td>6</td>
<td>4</td>
</tr>
<tr>
<td>K8L</td>
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<td>115</td>
<td>4</td>
<td>0</td>
<td>1</td>
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<tr>
<td>Average</td>
<td>20</td>
<td>68</td>
<td>1.4</td>
<td>1.6</td>
<td>5</td>
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</table>
muscle co-contractions, an assumption that is supported by similar data obtained by another instrumented knee endoprosthesis (Erhart et al., 2010a). Other studies that investigated the effect of lateral shoe wedges (Hinman et al., 2012; Kakihana et al., 2005) found a lateral shift of the COP and a reduction of the EAM lever arm. Similarly, we found a light lateral force shift (i.e. a reduction of the MR) in 3 subjects. It should be taken into account though that altered foot loading was an active, *muscularly induced* gait modification—as opposed to shoes or wedges which induce a COP shift *passively*. This may be an explanation why the active alteration of the foot loading led to such high muscle co-contractions: Our subjects realized lateral foot loading, among others, by an increased ankle inversion and, in some subjects, in-toeing. The ankle inversion minimizes the support base, so that muscular co-contractions are most likely required by the unstable condition. However, as muscle activation was not quantified by EMG, this assumption could not be proven. This phenomenon of co-contractions is important, maybe even decisive, with regard to the clinical results of interventions: Some patients may be able to utilize an intervention without additional co-contractions whereas others may not be. This might explain controversial clinical results.

During *medial foot loading*, the MR was reduced in all subjects, whereas increases in $F_{\text{med}}$ if observed at all, were always smaller. Both effects resulted in average reductions of $F_{\text{med}}$ by ~18% at both peaks. The observed effects are in accordance with previously published studies. The reductions of $F_{\text{med}}$ were caused by a force shift away from the medial and towards the lateral compartment. How can the lateral force shift during medial foot loading be explained biomechanically? A simple biomechanical model suggests that medial foot loading would shift the COP and the GRF vector medially, and the increased lever arm, d, would raise EAM and thus also $F_{\text{med}}$ (Fig. 2A). However, the inclination of the GRF, which is influenced by its m-l component, also affects d (Fig. 2B). Such changes in the m-l component of the GRF have been reported previously (Jenkyn et al., 2011) and support the notion that a change in the COP relative to the foot *alone* does not predict effects on the knee joint. We had expected that the amplitude of m-l shift of the COM, causing m-l acceleration forces, would be an *indirect* measure for such changes in the m-l component of the GRF. However, such a correlation was not found, which may be explained if the maximum m-l acceleration forces occur at other times than the maxima of $F_{\text{med}}$.

From the individual changes of the evaluated kinematic gait parameters, it is obvious that the subjects used different strategies to achieve the altered foot loading. The observed correlations between changes in certain gait parameters and some load values do not prove that the changed parameters are the reason for the load changes. Nevertheless, we found some consistent kinematic adaptations during the stance phase. For example, medial foot loading was realized by an increased SW, which accompanied a decreased LAA. Furthermore, we observed out-toeing, altered AIA and changes in the TLA, although not in all subjects. These kinematic adaptations are similar to the ones found with the variable stiffness shoe (Boyer et al., 2012). This underlines that a whole-body movement can change not only the *global* direction of the GRF but also its *relative* distance to the knee joint center.

The interpretation of the measured loads and gait changes is limited because the measurements were performed on a treadmill. GRF could not be measured, the COP and the EAM were not computed and the correlations between these parameters were not evaluated. Secondly, for medial foot loading, we attempted to increase the pressure under the medial aspect of the foot by unloading the lateral aspect. A pronounced decrease of the pressure on the lateral side was indeed shown during medial foot loading trials. Whether this results automatically in an increased pressure on the medial side is somewhat speculative. Theoretically, a lower pressure on the sensor could be the result of a reduction in vertical GRF magnitude, i.e. would not indicate a higher medial foot loading. However, the peak GRF cannot fall below 100%BW and the literature does not report a significant decrease of the vertical GRF (Hinman et al., 2012; Jenkyn et al., 2011; Erhart et al., 2010a). Some example videos of all three foot loading conditions will be uploaded on our online database [www.OrthoLoad.com](http://www.OrthoLoad.com) to provide a visual impression of how the gait modifications were realized.

The real pressure changes at the sensor site, achieved by a more lateral or medial loading of the foot, were much larger than the demanded changes of 20%. For avoiding or enforcing the sensor signal, all subjects exaggerated the amount of necessary load shift and the individually different amount of pressure de- or increase may partly explain the varying effect on the investigated load values.

As no EMG data were collected, the assumed co-contractions during lateral foot loading could not be proven. The subjects had only trained for this type of walking for a few minutes before the measurements were taken. A longer practice period for this gait modification could possibly make the gait pattern more efficient, thus reducing co-contractions. The increase of $F_{\text{med}}$ may therefore be less pronounced after walking for a long time with lateral foot loading. Another limitation of this study was the small number of investigated subjects; therefore, statistically proven conclusions were difficult to draw. Nevertheless, this study provides new insights into the mechanisms that occur when walking with musically induced altered foot loading.

## 5. Conclusions

Medial foot loading consistently led to a reduction in medial knee joint loading of 18% on average. This reduction was induced by a force shift from the medial to the lateral knee compartment. The resultant force did not change much, which indicates that medial foot loading would raise the proportional force acting in the lateral compartment of the knee. Therefore, such a gait pattern only seems helpful for patients with pronounced medial gonnarthrosis. Lateral foot loading, on the other hand, caused a pronounced increase of the medial contact force, most likely due to muscular co-contractions. It can therefore not be suggested for patients with medial gonnarthrosis.

### Conflict of interest

The authors have no conflict of interest.

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