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

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ABSTRACT

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 telemeterized 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, F_{res}, the medial contact force, F_{med}, and the force distribution F_{med}/F_{res} across the tibial plateau were computed from the measured joint contact loads.

Findings: During lateral foot loading, the two maxima of F_{res} 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, F_{med} increased by +16%/+17%. Medial foot loading, on the other hand, changed F_{res} only slightly, but decreased the distribution by −18%/−11%. This led to average reductions of F_{med} 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 F_{med} during lateral foot loading was most likely 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, $-F_z$, which acts distally in the direction of the tibia (Fig. 1), is transferred by the force, F_{med}, 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 $-F_z$ more to the lateral compartment, thus reducing F_{med}.

Due to the difficulties in measuring F_{med} 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 $-F_z$ and thus a smaller force F_{med}.

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 $-F_z$ laterally (Fig. 2A). Another research group aimed to lower the EAM using shoes with higher stiffness on the lateral side of the sole. Similar as for the wedges, it was initially hypothesised that this shoe would shift the COP laterally, thus shortening the lever arm and decreasing 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) shortened d and thus 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 F_{med} 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

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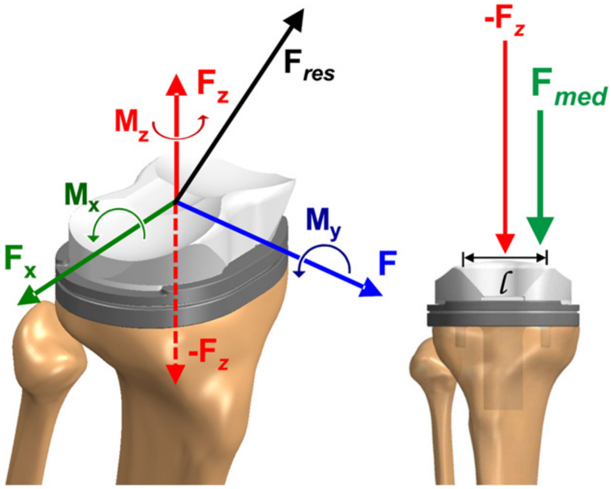

Fig. 1. Instrumented tibial tray of the knee endoprosthesis.

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.

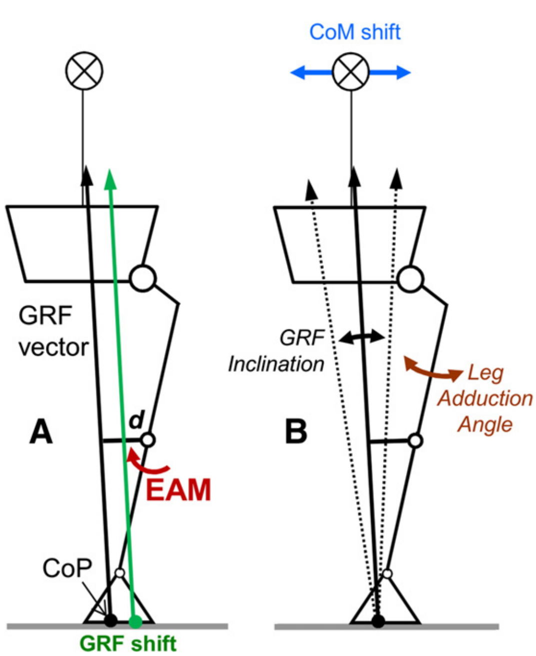

Fig. 2. *Mechanisms for altering the EAM.* (A) The external adduction moment (EAM) is mainly determined by the magnitude of the ground reaction force vector (GRF, black arrow) and its lever arm, *d*, to the knee joint center. A basic model suggests that a lateral shift of the COP (green arrow) shortens 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.

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.

2.3. Knee joint loading

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:

$$F_{res} = \sqrt{F_x^2 + F_y^2 + F_z^2}. \tag{1}$$

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

$$F_{med} = \frac{-F_z}{2} - \frac{M_y}{l} \tag{2}$$

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).

$$MR = \frac{F_{med} * 100}{-F_z} \tag{3}$$

Table 1
Patient data.

Patient		K1L	K3R	K5R	K7L	K8L	Average
Age	[years]	68	75	63	78	74	72
Body mass	[kg]	98	97	90	68	79	86
pOP time	[months]	70	58	51	51	45	55
TFA		3.0 varus	3.5 varus	1.0 varus	6.5 varus	4.0 varus	4.0 varus

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