Journal of Biomechanics 44 (2011) 1354–1360

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

The effect of valgus braces on medial compartment load of the knee joint – *in vivo* load measurements in three subjects

Ines Kutzner [a,*], Steffen Küther [a], Bernd Heinlein [b], Jörn Dymke [a], Alwina Bender [a], Andreas M. Halder [c], Georg Bergmann [a]

[a] Julius Wolff Institute, Charité – Universitätsmedizin Berlin, Augustenburger Platz 1, 13353 Berlin, Germany
[b] ZHAW Zurich University of Applied Science, Biomechanical Engineering, Winterthur, Switzerland
[c] Hellmuth-Ulrici-Kliniken, Klinik für Endoprothetik, Sommerfeld, Germany

ARTICLE INFO

Article history:
Accepted 3 January 2011

Keywords:
Knee joint
Braces
Forces
Valgus
Medial load

ABSTRACT

Knee osteoarthritis occurs predominately at the medial compartment. To unload the affected compartment, valgus braces are used which induce an additional valgus moment in order to shift the load more laterally. Until now the biomechanical effect of braces was mainly evaluated by measuring changes in external knee adduction moments. The aim of this study was to investigate if and to which extent the medial compartment load is reduced *in vivo* when wearing valgus braces.

Six components of joint contact load were measured *in vivo* in three subjects, using instrumented, telemeterized knee implants. From the forces and moments the medio-lateral force distribution was calculated. Two braces, MOS Genu (Bauerfeind AG) and Genu Arthro (Otto Bock) were investigated in neutral, 4° and 8° valgus adjustment during walking, stair ascending and descending.

During walking with the MOS brace in 4°/8° valgus adjustment, medial forces were reduced by 24%/30% on average at terminal stance. During walking with the GA in the 8° valgus position, medial forces were reduced by only 7%. During stair ascending/descending significant reductions of 26%/24% were only observed with the MOS (8°).

The load reducing ability of the two investigated valgus braces was confirmed in three subjects. However, the load reduction depends on the brace stiffness and its valgus adjustment and varies strongly inter-individually. Valgus adjustments of 8° might, especially with the MOS brace, not be tolerated by patients for a long time. Medial load reductions of more than 25% can therefore probably not be expected in clinical practise.

1. Introduction

Osteoarthritis (OA) is the most common joint disease, associated with pain and loss of mobility. When the knee joint is highly loaded, e.g. during the stance phase of gait, most of the axial force is transmitted by the medial compartment (Hsu et al., 1990; Johnson et al., 1980; Morrison, 1970; Shelburne et al., 2005; Zhao et al., 2007). This fact is believed to be responsible for the observation that gonarthrosis predominately starts at the medial compartment (Hernborg and Nilsson, 1977; Jackson et al., 2004).

Besides surgical treatments, several conservative methods, such as lateral shoe wedges, the use of crutches, weight reduction and valgus bracing are common to reduce the axial tibial force and/or to shift it laterally. Reduced loading of the affected compartment is related to pain reduction and improved function, and may thus delay the need for joint replacement. Valgus braces induce an additional external valgus (abduction) moment at the knee joint, which counteracts the external adduction moment (EAM) in order to shift the axial force from the medial knee compartment towards the lateral one. A correlation between the medial contact force and peak EAMs was found analytically and measured *in vivo* in one subject (Shelburne et al., 2008; Zhao et al., 2007).

In previous studies the load reducing effect of braces was predicted by brace and EAM measurements. Reductions of peak EAMs of about 10–15% were reported when walking with valgus braces (Lindenfeld et al., 1997; Self et al., 2000). Studies have furthermore predicted that larger valgus angulations of the brace lead to higher load reduction of the medial compartment. Reductions of peak EAMs of up to 15% and 19% were reported when walking with a brace adjusted in 4° and 8° valgus positions, respectively (Fantini Pagani et al., 2010). Using an analytical model, Pollo et al. (2002) estimated a medial load reduction of 11% when the brace was adjusted in a 4° valgus position, and 17% for the 8° valgus position. Furthermore, several clinical studies

* Corresponding author. Tel.: +49 30 450559678; fax: +49 30 450559980.
E-mail address: ines.kutzner@charite.de (I. Kutzner).

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doi:10.1016/j.jbiomech.2011.01.014

have shown that the use of braces leads to reduced pain and improved function (Kirkley et al., 1999; Lindenfeld et al., 1997; Pollo et al., 2002; Richards et al., 2005).

Up to now no direct measurement of the medial contact force (F_{med}) has been performed to confirm the unloading effect of valgus braces. The aim of this study was therefore to investigate the ability of valgus braces to reduce the medial compartment load, by taking measurements directly in the knee joint. Knee contact loads (3 forces and 3 moments) were measured using instrumented knee implants with telemetric data transmission. Medio-lateral load distribution was determined during the most frequent strenuous activities of daily living: walking, stair ascending and descending.

2. Methods and materials

2.1. Investigated braces

Two braces with monocentric joints were investigated: MOS (MOS Genu, long version, Bauerfeind AG, Germany) and GA (Genu Arthro, Otto Bock HealthCare GmbH, Germany) (Fig. 1A, and B). Both braces are designed to apply an external valgus moment about the knee through a three-point bending system. Whereas the GA is a unilateral brace, the MOS has a bilateral frame.

The braces were adjusted by skilled orthopaedic technicians. The joints were aligned with Nietert's (1976) compromise axis . The braces were first fitted to the leg in a neutral position to examine whether the brace itself already has an influence on joint loading. After performing the activities with the brace in neutral position, additional valgus angles of 4° (MOS) and 8° (MOS and GA) were adjusted. The valgus adjustment of MOS is set by scaled, eccentric screws while the brace is still attached to the leg. The GA has to be removed from the leg to set the additional valgus angle.

When fixing the brace at the leg, it is bent in varus direction until it reaches its final position (Fig. 1C). This back-bending causes the required valgus moment M_b. The magnitude of M_b depends not only on the adjusted angle, but also on the stiffness of the brace.

2.2. Instrumented implant

An instrumented tibial tray with telemetric data transmission (Fig. 2) was developed to measure the 6 components (3 forces and 3 moments) of the knee contact load *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 custom-made tibial component has a stem-in-stem design. Six semiconductor strain gauges (KSP 1-350-E4, Kyowa, Japan) measure the strains in the inner stem and deliver, via a 6×6 calibration matrix, the 6 load components. The signals are multiplexed by a custom-made telemetry chip and transferred via an antenna to the external receiver (Graichen et al., 2007). The electronics are powered inductively by an external magnetic field.

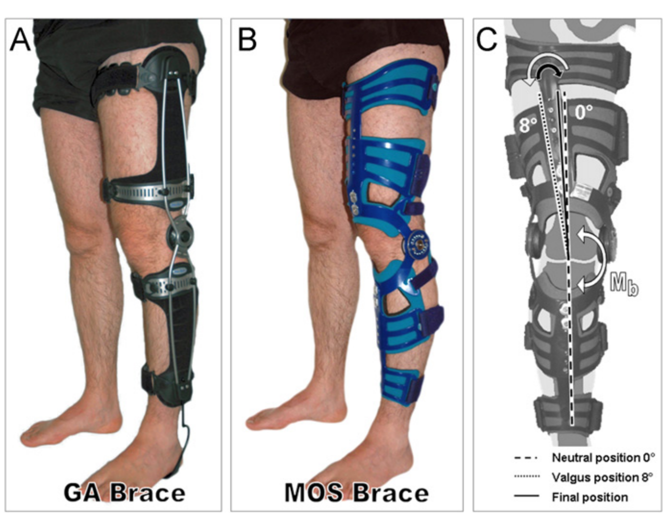

Fig. 1. Investigated braces GA (A) and MOS (B) and schematic illustration of brace adjustment (C). The braces are adapted in neutral position (0°) and then adjusted in 4° or 8° valgus position (grey arrow). When fixing the braces at the leg, they are bent back (black arrow) and thereby apply an external moment M_b.

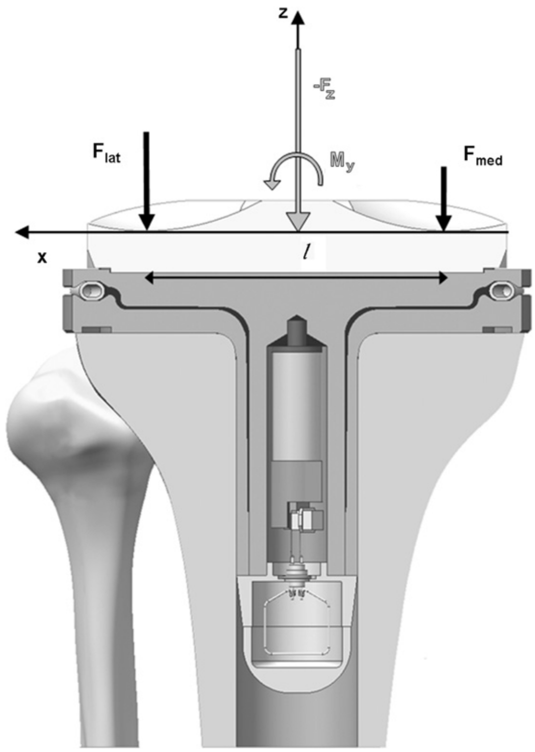

Fig. 2. Coordinate system of the instrumented tibial tray in the frontal plane. The axial force $-F_z$ is the sum of the medial and lateral force (F_{med}+F_{lat}) acting on the tray.

Table 1
Subject data.

Subject	K1L	K3R	K5R
Age (years)	64	71	60
Body mass (kg)	103	96	96
Height (cm)	177	175	175
Time post-op (months)	23	12	6
Mechanical axis angle (°)	3 varus	4 varus	1 varus

2.3. Coordinate system

The right-handed coordinate system of the implant is fixed at the right tibia (Fig. 2). Its origin lies on the extended stem axis at the height of the lowest part of the tibial insert.

The force components $+F_x$, $+F_y$ and $+F_z$ act in lateral, anterior and superior directions, respectively. The moments $+M_x$, $+M_y$, $+M_z$ act in the sagittal, frontal and horizontal plane of the tibia, respectively and turn right around their belonging axes. The negated moment $-M_y$ counteracts the external adduction moment, i.e. it attempts to abduct the tibia.

2.4. Subjects and activities

After approval by the ethics committee and the patients' informed consent, three male patients obtained an instrumented implant after medial compartment osteoarthritis (Table 1).

The knee alignment in the frontal plane was determined from radiographs during two-legged stance. The mechanical axis angle was defined as the angle formed by the mechanical axes of femur (hip to knee centre) and tibia (Specogna et al., 2007). All subjects showed a slight varus alignment.

Three activities of daily living were investigated: walking at a self selected speed on level ground, ascending stairs, and descending stairs (stair height: 200 mm). All subjects performed the activities fluently and free of pain.

2.5. Medio-lateral load distribution

All moment components can be caused by friction in the joint and/or by forces acting eccentrically to the origin of the coordinate system. It is not generally possible to distinguish how these two factors influence the moments. In order to calculate the medio-lateral load distribution, the following reasonable assumptions were made: during the investigated activities, the knee moves nearly in the sagittal plane. Friction in the frontal plane of the tibia is therefore negligible. Since an ultra-congruent tibial inlay was used, the femoral condyles with the constant distance l do not move in medio-lateral direction. The moment M_y is therefore caused solely by the axial force component -F_z, acting with an offset to the centre of the implant in medio-lateral direction. Negative values of M_y indicate a force shift in medial direction; positive values indicate a lateral shift.

The axial force $-F_z$ is transferred via the two condyles from the femur to the tibia and is split into the medial (F_{med}) and lateral (F_{lat}) axial force component, with F_{med} being:

$$F_{med} = \frac{-F_z}{2} - \frac{M_y}{l} \tag{1}$$

It was tested with the instrumented implants during the calibration process that F_{med} can be calculated with an error below 3% if $|F_z|$ is greater than 1000 N. Values of F_{med} were therefore determined only during the stance phase of gait when $|F_z|$ was above this level (Fig. 3).

2.6. Data analysis

All activities were repeated 5 times (stair ascending/descending) to 30 times (level walking) for each subject. The continuous load patterns from repeated trials of the individuals were averaged by a Dynamic Time Warping procedure (Wang and Gasser, 1997) to create typical load-time curves (Fig. 4).

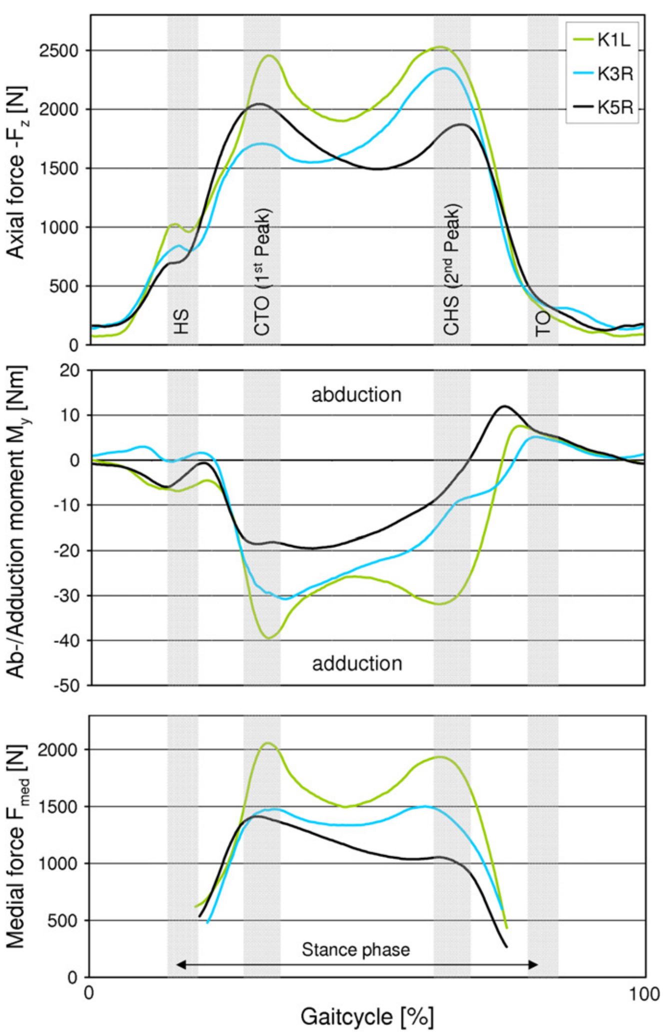

Fig. 3. Load components $-F_z$, M_y and F_{med} during walking without brace. HS: heel strike; CTO: contralateral toe-off; CHS: contralateral heel strike; TO: toe-off. Calculation of F_{med} only if $|F_z| > 1000$ N. Averaged data from 30 trials.

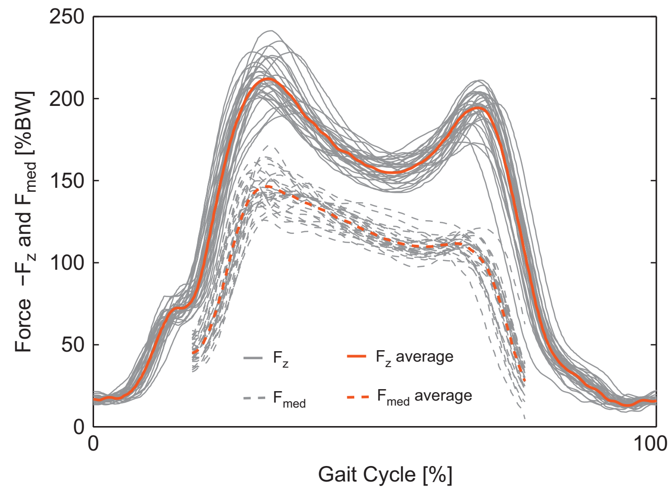

Fig. 4. Intra-individual variability of axial force $-F_z$ and medial force F_{med} during walking. Axial forces (solid lines) and medial forces (dashed lines) from 30 repetitive trials (thin lines) and their average (thick lines). Exemplary trials from subject K5R.

During all 3 activities, double-peak patterns of the axial and medial forces were observed during the gait cycles (Fig. 5). Peak values ("1st Peak" and "2nd Peak") were evaluated separately for each trial and then averaged by their arithmetic mean. They are stated in percent of bodyweight (%BW).

To determine walking speed and stride length, reflective markers were placed on the subjects' heel and captured with a camera-based gait analysis system (Vicon, Oxford, UK).

Differences between the conditions with and without brace were analysed within each subject using a Wilcoxon test ($\alpha = 0.05$).

2.7. Stiffness test

A Zwick 1455 material testing machine (Zwick, Germany) was used to determine the stiffness of the braces in the frontal plane. The braces were mounted on rigid test blocks having the size of shank and thigh, and were supported at their outer ends. An increasing test load was applied in medial direction at the joint centre with a velocity of 30 mm/min up to a maximum load of 100 N. The spring constant was calculated at this maximum force.

3. Results

3.1. Walking (Fig. 6A, and B)

Peak forces occurred at contralateral toe off (1st peak) and before contralateral heel strike (2nd peak) (Figs. 3, and 5A). A small peak was furthermore seen immediately before heel strike.

Without brace, 83%/91%/70% (subject K1L/K3R/K5R) of the total axial force F_z were transferred through the medial compartment during the 1st peak, and 76%/65%/59% during the 2nd peak.

With the MOS brace in neutral position, F_{med} was already reduced by 10%/9% on average (1st/2nd peak). In the 4° valgus adjustment F_{med} decreased by 18%/24% (1st/2nd peak). In the 8° valgus adjustment F_{med} was reduced by 23%/30% (1st/2nd peak). Except for the values of the 2nd peak with the brace in neutral position, changes were significant in all subjects.

With the GA brace medial force reductions were much smaller than with the MOS brace. Significant force reductions of 7% on average (2nd peak) were only observed for the 8° valgus adjustment.

3.2. Stair ascending (Fig. 7A, and B)

Peak forces occurred at CTO (1st peak) and after contralateral stair contact (2nd peak) (Fig. 5B).

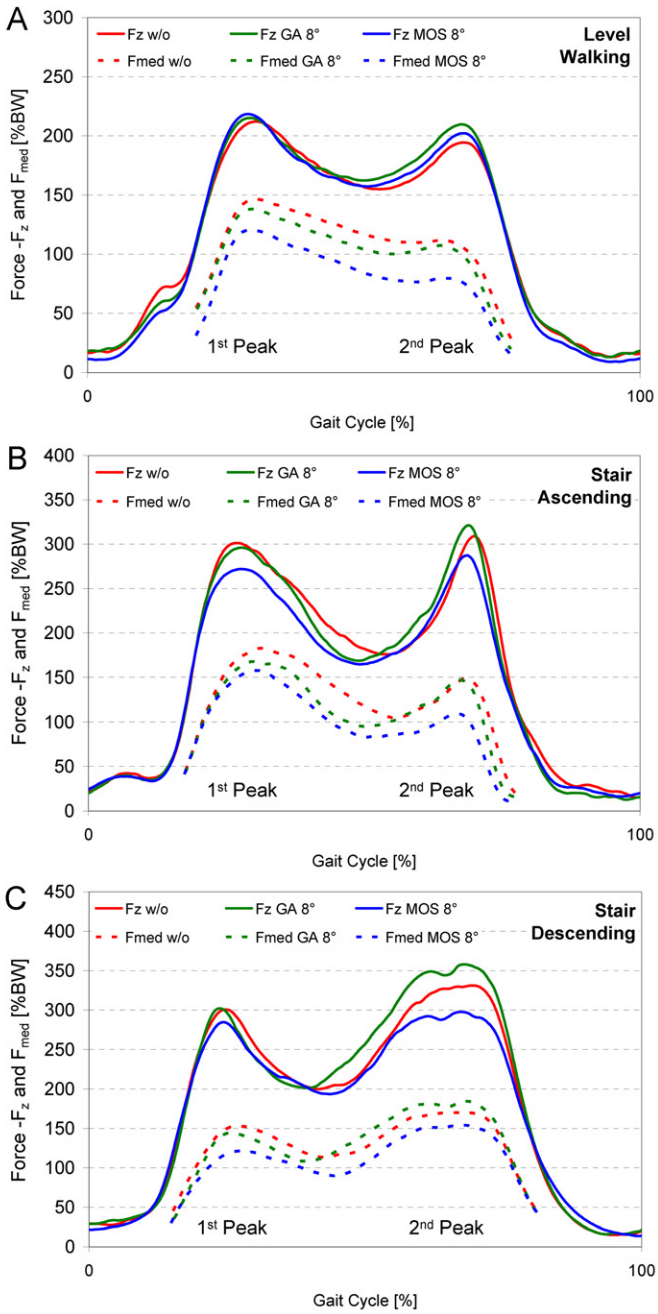

Fig. 5. Axial force $-F_z$ and medial force F_{med} during investigated activities. Solid lines: F_z. Dashed lines: F_{med}. (A) level walking. (B) stair ascending, (C) stair descending. Forces during walking without and with GA and MOS brace in 8° valgus adjustment. Average force-time patterns from subject K5R.

Without brace, 71%/74%/63% (subject K1L/K3R/K5R) of the total axial force F_z were transferred through the medial compartment during the 1st peak and 65%/68%/50% at the 2nd peak.

With the MOS brace, significant reductions of F_{med} of 26% on average (2nd peak) were observed in the 8° valgus position.

With the GA brace, F_{med} was less reduced than with the MOS brace. With the 8° valgus adjustment, F_{med} was reduced by 6%/9% (1st/2nd peak) on average. However, force reductions were not significant.

3.3. *Stair descending (Fig. 7C, and D)*

Peak forces occurred at CTO (1st peak) and immediately before contralateral stair contact (2nd peak) (Fig. 5C).

Without brace 72%/71%/56% (subject K1L/K3R/K5R) of the total axial force F_z were transferred through the medial compartment during the 1st peak and 56%/62%/52% at the 2nd peak.

With the MOS brace significant force reductions of F_{med} of 17% on average (2nd peak) were observed in the 8° valgus adjustment.

With the GA brace in the 8° valgus adjustment only small reductions of F_{med} of 7%/6% (1st/2nd peak) occurred. However, force reductions were not significant and differed only slightly from those of the neutral adjustment.

3.4. *Changes of axial force F_z (Fig. 6C, and D)*

Besides F_{med} also F_z was often reduced when walking with braces (Table 2). Changes of F_z varied strongly between the subjects. Especially in K3R the axial force was reduced by 20% and more at CHS while walking with the MOS (Fig. 6D).

Significant reduction of F_z in all 3 subjects was only observed with the MOS during stair descending (2nd peak) (mean reduction = 16%). During walking, stair ascending and descending with the GA brace, mean reductions of F_z did not exceed 7% and were not significant.

3.5. *Walking speed and stride length*

Walking speed without brace was slowest in subject K3R (1.10 m/s) followed by K5R (1.17 m/s) and K1L (1.26 m/s) (Table 3). No significant differences in walking speed were observed while walking with braces except for K3R. In this subject walking speed was slightly increased by 5% while walking with the GA brace in neutral position.

Stride length was smallest in subject K3R (1.23 m) followed by K5R (1.41 m) and K1L (1.43 m). No significant differences in stride length were observed while walking with braces.

3.6. *Stiffness test*

A spring constant of 9.8 N/mm was determined for the MOS brace, but only 4.0 N/mm for the GA brace. About 2.5 times higher external valgus moments must therefore be expected for the MOS than for the GA brace when using the same adjustment.

4. Discussion

The use of valgus braces is a common conservative method with the intention to reduce the medial compartment load and to achieve improved function and pain reduction. This study is a case report; the limited number of 3 subjects does not allow general conclusions. The subjects underwent total knee replacement, were free of pain when performing the activities and had only slight varus alignments. Due to these facts, their kinematics may differ from those of typical OA patients. However, the authors are convinced that the axial force as well as the medio-lateral force distribution is not affected much by the knee replacement itself. The principle biomechanical mechanism of valgus braces can therefore be evaluated and transferred into clinical practise.

The results of this study confirm previous findings that the total axial force is predominantly transferred through the medial compartment during the stance phase of gait (Hsu et al., 1990; Johnson et al., 1980; Morrison, 1970; Shelburne et al., 2005; Zhao et al., 2007). 59–91% of the peak axial forces were transferred medially during walking and 50–72% when going up or down stairs.

A reduction of the medial compartment load by the investigated valgus braces could be confirmed in all three subjects.

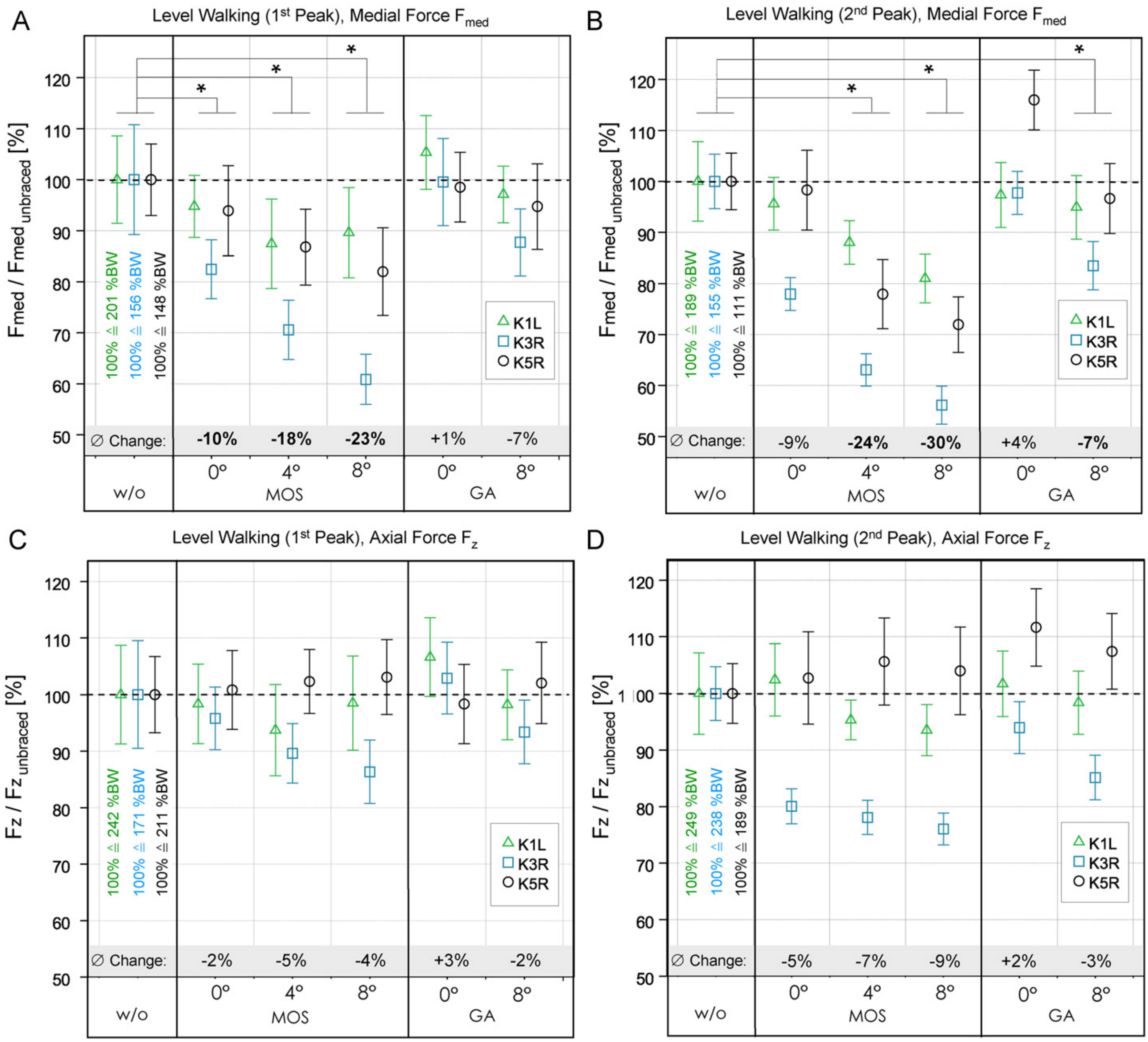

Fig. 6. Medial (A, and B) and total axial forces (C, and D) during level walking with braces. Forces are given in percent of forces without braces (mean ± 1 SD) at 1st peak (A, and C) and 2nd peak (B, and D). Vertical numbers indicate forces in percent bodyweight (BW) during walking without brace (w/o). MOS and GA: brace type. 0° (neutral), 4°, 8°: valgus adjustment of the braces. Average force changes from all 3 subjects are given at the bottom of the diagrams. Bold numbers indicate significant force changes in all 3 subjects.

However, large differences were observed between the two investigated brace types as well as between the subjects themselves.

The reduction of F_{med} can result from two different effects: a reduction of F_z and/or from a shift of F_z to the lateral compartment. Reduced axial forces can be caused by a lower walking speed, by smaller stride lengths and by other changed gait patterns such as greater trunk sway.

With braces, walking speed did increase or decrease by 5% at maximum. Former own *in vivo* load measurements have shown that an increase in walking speed of 5% lead to increased axial forces (1st peak) of about 3%. The second axial force peak is even less influenced. Slight changes in walking speed can therefore not be the reason for substantial changes in the axial force. Further possible changes in kinematics, which might have an influence on knee joint loading, have not been quantified.

Reductions of F_z were greater with the MOS brace than with the GA brace. The greater reduction of F_{med} while walking with the MOS brace is therefore partly caused by a greater reduction of F_z and not only by a force shift in lateral direction or changes of M_y (Eq. (1)). In subject K3R (MOS brace, 8° valgus) for example, F_{med} was reduced by 39%. In that case, 20% of the reduction of F_{med} was caused by a reduction of F_z and 80% by the change of M_y.

The evaluated stiffness of the two braces is an approximation since the deformation of thigh and shank was not simulated. Nevertheless, the greater load-reducing effect of the MOS corresponds to its stiffness, which is 2.5 times higher than that of the GA. As a consequence it is possible to apply higher valgus moments with the MOS brace than with the GA using the same adjustment. From a mechanical point of view, very elastic braces with a large adjustment angle are less sensitive against soft tissue movement and adjustment differences of the brace. We speculate

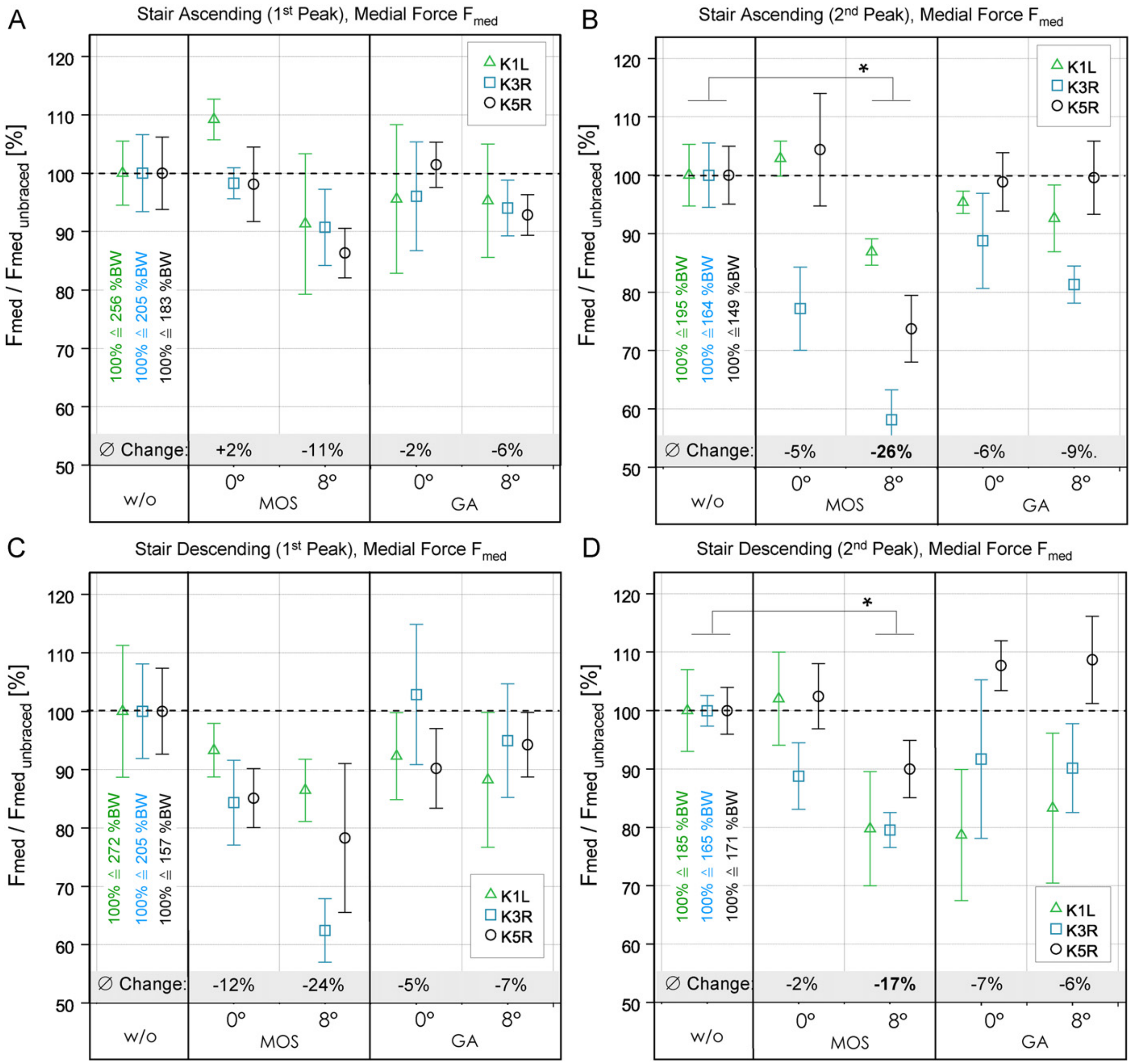

Fig. 7. Medial forces during stair ascending (A, and B) and stair descending (C, and D) with braces. Forces are given in percent of forces without braces (mean ± 1 SD) at 1st peak (A, and C) and 2nd peak (B, and D). For further explanations see Fig. 6.

that the observed strong inter-individual differences of the medial load reduction are partly caused by these factors.

Due to different braces, valgus settings, and diverse measured parameters, a comparison of the results to those of former studies is limited. The same brace (GA) was used in a study of Fantini Pagani et al. (2010). A reduction of the external peak adduction moment of about 19% in the 8° valgus setting was reported, compared to walking without brace. This moment reduction is about 2–3 times higher than the reduction of medial forces now measured directly in the knee joint, which might indicate that no one-to-one conclusions can be drawn from external moment reductions on internal compartment force reductions. The authors also reported that greater reductions occurred at late stance (2nd peak) and that the loading was already reduced with the brace in neutral position, which is in agreement with our findings.

The reduction of medial forces while walking with an unloader brace was calculated by Pollo et al. (2002) using an analytical model. The reported reduction of 11% (4° valgus) and 17% (8° valgus) is within the range of our data. Also force reductions while walking with the brace in neutral position were reported. However, no direct comparison is possible due to the presence of different brace models.

The reported load-reducing effects of braces coincide with clinical observations of pain reduction when walking with braces (Lindenfeld et al., 1997; Pollo et al., 2002; Richards et al., 2005). As the present study shows, the effect depends on the type of brace and its valgus setting. Since the patient's acceptance of the brace and the wearing comfort is of major importance, the amount of applicable external valgus moment is limited. Whereas no major discomfort was reported by the subjects when wearing the GA brace, discomfort was reported when walking with the MOS brace in 8° valgus. Since the chosen valgus settings of 8° with the MOS brace would probably not have been tolerated for a long duration by the subjects, medial load reductions of more than 25% cannot be expected permanently.

Table 2
Changes of medial and axial forces as compared to the unbraced condition.

Activity	Brace	Change (%)			
		F_{med}		F_z	
		1st peak	2nd peak	1st peak	2nd peak
Level walking	MOS 0°	−**10**	−9	−2	−5
	MOS 4°	−**18**	−**24**	−5	−7
	MOS 8°	−**23**	−**30**	−4	−9
	GA 0°	+1	+4	+3	+2
	GA 8°	−7	−**7**	−2	−3
Stair ascending	MOS 0°	+2	−5	+1	−4
	MOS 8°	−11	−**26**	−6	−13
	GA 0°	−2	−6	0	−3
	GA 8°	−6	−9	0	−5
Stair descending	MOS 0°	−12	−2	−6	−6
	MOS 8°	−24	−**17**	−9	−**16**
	GA 0°	−5	−7	−5	−7
	GA 8°	−7	−6	−4	−3

Mean values from all 3 subjects at 1st and 2nd peak. Bold numbers indicate significant changes in all subjects.

Table 3
Average speed and stride length ± 1 SD during walking with and without braces. Bold numbers indicate significant differences to the condition without brace.

Brace	Subject		
	K1L	**K3R**	**K5R**
Walking speed (m/s)			
Without	1.26 ± 0.02	1.10 ± 0.05	1.17 ± 0.02
MOS 0°	1.25 ± 0.04	1.06 ± 0.05	1.15 ± 0.04
MOS 4°	1.25 ± 0.06	1.10 ± 0.04	1.18 ± 0.05
MOS 8°	1.31 ± 0.06	1.14 ± 0.03	1.15 ± 0.03
GA 0°	1.31 ± 0.02	**1.15 ± 0.06**	1.16 ± 0.02
GA 8°	1.26 ± 0.02	1.12 ± 0.03	1.19 ± 0.04
Stride length (m)			
Without	1.43 ± 0.05	1.23 ± 0.03	1.41 ± 0.02
MOS 0°	1.39 ± 0.05	1.24 ± 0.05	1.40 ± 0.02
MOS 4°	1.37 ± 0.04	1.22 ± 0.03	1.40 ± 0.02
MOS 8°	1.44 ± 0.05	1.28 ± 0.03	1.40 ± 0.04
GA 0°	1.49 ± 0.04	1.27 ± 0.03	1.41 ± 0.02
GA 8°	1.46 ± 0.04	1.26 ± 0.02	1.41 ± 0.04

Due to the variability between the subjects, the authors suggest that valgus braces should only be used if a patient reports pain relief. The unloading effect of braces must furthermore be compared to other more comfortable conservative methods such as forearm crutches, laterally wedged shoes or weight reduction.

Conflict of interest statement

The authors declare no conflict of interests. Except from funding, the sponsors were not involved in study design, collection, analysis and interpretation of data, or anything related to this manuscript.

Acknowledgements

The authors thank all subjects for their great contribution and Dr. med. Alexander Beier for his clinical support. This project was supported by the Deutsche Arthrose-Hilfe, by Zimmer GmbH, Winterthur, Switzerland and by the Deutsche Forschungsgemeinschaft (SFB760/ Be 804/18-1).

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