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Immediate effects of valgus knee bracing on tibiofemoral contact forces and knee muscle forces

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ABSTRACT

Background Valgus knee braces have been reported to reduce the external knee adduction moment during walking. However, mechanistic investigations into the effects of valgus bracing on medial compartment contact forces using electromyogram-driven neuromusculoskeletal models are limited.

Research question What are the immediate effects of valgus bracing on medial tibiofemoral contact forces and muscular loading of the tibiofemoral joint?

Methods Sixteen (9 male) healthy adults (27.7 ± 4.4 years) performed 20 over-ground walking trials at self-selected speed both with and without an Ossür Unloader One® brace. Assessment order (i.e., with or without brace) was randomised and counterbalanced to prevent order effects. While walking, three-dimensional lower-body motion, ground reaction forces, and surface electromyograms from eight lower-limb muscles were acquired. These data were used to calibrate an electromyogram-driven neuromusculoskeletal model of muscle and tibiofemoral contact forces (N), from which muscle and external load contributions (%) to those contact forces were determined.

Results Although walking with the brace resulted in no significant changes in peak tibiofemoral contact forces at the group-level, individual responses were variable and non-uniform. At the group-level, wearing the brace resulted in a 2.35% (95% CI 0.46–4.24; $p = 0.02$) greater relative contribution of muscle to lateral compartment contact loading (54.2 ± 11.1%) compared to not wearing the brace (51.8 ± 12.1%) ($p < 0.05$). Average relative contributions of muscle and external loads to medial compartment loading were comparable between brace and no brace conditions ($p \geq 0.05$).

Significance Wearing a valgus knee brace did not immediately reduce peak tibiofemoral contact forces in healthy adults during normal walking. It appears this population may modulate muscle activation patterns to support brace-generated valgus moments, thereby maintaining normal walking knee moments and tibiofemoral contact forces. Future investigations are warranted to better understand effects of valgus knee brace in people with medial knee osteoarthritis using an electromyogram-driven neuromusculoskeletal model.

1. Introduction

Knee bracing is a potential treatment option for people suffering from knee osteoarthritis, and brace use is associated with improvements in knee pain and function [1]. However, the biomechanical effects of knee bracing are not entirely understood. Valgus knee braces are designed to apply a valgus moment about the knee, potentially altering knee joint loading (i.e., moments and forces both external and internal to the joint) during ambulatory tasks. Specifically, valgus knee braces aim to alter medial tibiofemoral loading [2] where structural degeneration is present and improve clinical symptoms.

Valgus knee braces have been reported to reduce the external knee adduction moment during walking [3]. The external knee adduction moment has been used extensively as a surrogate of medial-to-lateral tibiofemoral contact force distribution and has been associated with structural disease progression [4]. However, tibiofemoral contact forces depend on both external knee loads and muscle forces, which can independently vary due to muscle activation patterns and/or variations in

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gait kinematics [5–7]. Critically, muscles account for ~50% of medial compartment contact forces and 100% of lateral compartment loads during walking [5]. Investigating how valgus knee braces affect muscle and tibiofemoral contact forces may elucidate mechanisms by which valgus braces improve knee function.

Few studies have examined the effects of knee braces on tibiofemoral contact forces, and reported results are conflicting [8,9]. The concurrent effects of bracing on muscle activation patterns, muscle forces, and tibiofemoral contact forces remain unknown. Furthermore, heterogeneous muscle activation patterns in people with knee osteoarthritis [10] may account for, in part, inconsistent effects of valgus knee bracing on tibiofemoral contact forces [1,8,9]. Therefore, to understand the immediate effects of valgus knee bracing on tibiofemoral contact forces and identify potential mechanisms by which braces affect function, it is necessary to first eliminate the heterogeneous adaptations often reported with knee osteoarthritis [10] and evaluate pain-free individuals without lower-limb pathology. Second, it is critical to use a modelling method sensitive to subject- and task-specific muscle activation patterns, as well as external biomechanics (i.e., joint kinematics and kinetics). Electromyogram (EMG)-driven neuromusculoskeletal models use experimental measures of muscle activation (i.e., EMG), together with the individual's joint kinematics and kinetics, to predict muscle and tibiofemoral contact forces, as well as relative contributions of muscle and external loads to those contact forces [5,11–13].

The primary aim of this study was to test the hypothesis that wearing a valgus knee brace would immediately reduce peak medial tibiofemoral contact forces during walking compared to walking without the brace. Secondary aims were to investigate immediate effects of wearing a valgus knee brace on lateral tibiofemoral contact forces, total tibiofemoral contact forces, relative contributions of knee muscles to tibiofemoral contact forces, knee muscle co-activation patterns, and lower-limb joint kinematics and kinetics, and their coordination.

2. Methods

2.1. Participants

Sixteen healthy adults (mean ± standard deviation for age = 27.7 ± 4.4 years and body mass index = 23.4 ± 3.0 $kg{\cdot}m^{-2}$; n = 9 (56%) males; 3.3 ± 2.5° static varus alignment) were recruited from the university community. Participants were recreationally active, but had no current musculoskeletal pain/impairment or history of lower-limb injury that would affect their walking. Participants provided their written informed consent prior to testing. Ethics approval was obtained from the institutional Human Research Ethics Committee.

2.2. Procedures

Participants completed a minimum of 20 over-ground walking trials, both with and without wearing the brace (Unloader One®, Össur, Reykjavik, Iceland), along a 15-metre walkway at their self-selected speed. The brace was designed to provide an assistive valgus knee moment. Brace and no brace conditions were randomized across participants and counterbalanced to prevent order effects. Prior to testing, each participant's right leg was fitted with the Unloader One® according to manufacturer guidelines. Participants then walked wearing the brace for 10-minutes prior to testing to ensure familiarization.

Surface electromyography was performed using a Telemyo DTS wireless telemetry system (Noraxon, AZ, USA) on eight superficial muscles of the right limb: gluteus medius (GM), vastus medialis (VM), vastus lateralis (VL), rectus femoris (RF), semimembranosus (SM), biceps femoris (BF), tibialis anterior (TA), and medial gastrocnemius (MG). The skin-surface above the muscle belly was cleaned with alcohol, shaved, and lightly abraded to reduce impedance prior to application of bipolar surface Ag/AgCl electrodes (Duo-Trode, Myotronics, USA) (diameter: 12.5 mm; inter-electrode distance: 19 mm) oriented parallel to muscle fibres. Raw EMG signals, acquired at 1500 Hz, were pre-amplified and low-pass filtered, and time-synchronized with the camera motion analysis system (Vicon, MX, Oxford, UK) via a hardware trigger.

Participants performed six maximum voluntary contraction (MVC) tasks to elicit maximum activation of instrumented muscles: (i) single-leg heel raise; (ii) standing hamstring curl; (iii) seated knee extension; (iv) side-lying hip abduction; (v) ankle dorsiflexion from neutral ankle position; and (vi) single-leg hop. Participants performed three maximal efforts (3-second duration) for each contraction (i.e., i–v) against manual resistance with 30 s rest. Participants received standardized verbal encouragement to contract maximally. Participants performed three repetitions of a single-leg hop following demonstration and standardized instructions, and after demonstrating task proficiency.

Retroreflective markers and marker clusters were applied bilaterally to the pelvis, and both lower-limbs [14]. Supplementary Fig. S1 illustrates electrode and reflective marker placement with respect to the brace. Three-dimensional marker trajectories were collected at 100 Hz using a 12-camera motion analysis system (Vicon, MX, Oxford, UK) and Vicon Nexus (version 2.5). Two ground-embedded force plates (Advanced Mechanical Technology, MASS, USA) acquired ground reaction data at 1500 Hz. Twenty unique gait cycles with clean right foot force plate strikes were obtained. Participants rested for 10-minutes before completing walking trials under the second condition (i.e., brace or no brace).

2.3. Data analysis

The MOtoNMS (version 2.2) toolbox [15] was used to process marker, force plate, and EMG data in Matlab (R2016b, The Mathworks) for subsequent use in OpenSim (version 3.3) [16]. Hip joint centres were calculated using Harrington regression equations [17]. Knee and ankle joint centres were defined using midpoints of medial and lateral femoral condyles and malleoli, respectively. A vertical ground reaction force threshold of 20 N was used to detect heel strike and toe-off gait events. Markers and ground reaction data were low-pass filtered using zero-lag 2^{nd} order Butterworth filters with 6 Hz cut-offs, and then transformed from laboratory to OpenSim coordinate systems. The EMG data were high-pass filtered (30 Hz cut-off), full-wave rectified, and then low-pass filtered (6 Hz cut-off) using zero-lag 2^{nd} order Butterworth filters producing linear envelopes. The linear envelopes were subsequently amplitude-normalized to the maximum linear envelope EMG value recorded across all MVC tasks [18].

A generic full-body anatomic model [19], available in OpenSim, was used. The model had three rotational degrees of freedom (DOF) at the hip, one DOF at the knee with abduction/adduction and internal/external rotations as well as anterior/posterior and superior-inferior translations prescribed as functions of knee flexion, and one DOF at the ankle. Lower-limb body lengths were linearly scaled to match individual anthropometry using distances between joint centres calculated from a trial of upright quiet stance. Once scaled, inverse kinematics analysis in OpenSim was used to determine model hip, knee, and ankle kinematics by minimizing error between experimental marker and model marker positions. A surrogate measure of the standing static frontal plane alignment was determined from a single frame of inverse kinematic analysis, where the knee model was permitted to move into varus-valgus, and static alignment was extracted. Subsequently, inverse dynamics analysis was used to calculate joint moments. The brace action was modelled in OpenSim as an additional external load applied as a pure abduction moment about the knee. Brace abduction moments varied as a function of knee flexion-extension angle, as specified by the manufacturer. All joint moments were normalized to body mass ($Nm{\cdot}kg^{-1}$).

The muscle analysis tool in OpenSim was used to calculate instantaneous muscle-tendon unit (MTU) kinematics (i.e., lengths and

moment arms). The MTU kinematics, joint moments, and processed EMG were then used to calibrate an EMG-driven neuromusculoskeletal (NMS) model toolbox (i.e., Calibrated EMG-informed Neuromusculoskeletal Modelling Toolbox (CEINMS)) [20]. The CEINMS operates by first calibrating neuromuscular parameters via optimization, and subsequently estimating muscle and joint contact forces. For each participant, neuromuscular parameters of muscles spanning the knee were optimised using morphometric scaling [21] followed by functional calibration in CEINMS using four walking trials (two brace and two no brace). The objective of function calibration was to minimize (i) squared error between predicted NMS model-generated and inverse dynamics-calculated knee flexion-extension moments, and (ii) peak tibiofemoral contact forces [22]. The eight experimental EMG signals were mapped to thirteen MTU in CEINMS, similar to Lloyd and Besier [11]. Once calibrated, CEINMS was used to estimate muscle forces for the remaining walking trials. Tibiofemoral contact forces were calculated using the planar mechanism described previously [5] and were normalized to body weight ($N{\cdot}kg^{-1}$).

For each participant, kinematic, kinetic, EMG, and tibiofemoral contact force data over each gait cycle were spline interpolated to 101 time points, and ensemble-averaged to generate mean waveforms for brace and no brace conditions. Spatiotemporal parameters, lower-limb joint kinematics and knee joint moments were extracted. As the brace-generated knee torque was non-linearly influenced by knee flexion angle, this may change the temporal relationships between knee contact forces, moments, and kinematics. Consequently, polar plots were developed to present the coordination between medial tibiofemoral contact forces and knee adduction moments, internal knee extension moments, and knee flexion angle for brace and no brace conditions. Directed muscle co-activation ratios and total muscle activation were calculated for two agonist/antagonist pairs around the knee: (i) medial (VM, SM, MG) and lateral (VL, BF, LG) muscle groups; and (ii) extensor (SM, BF) and flexor (VM, VL, RF) muscle groups [23].

2.4. Statistical analysis

We aimed to detect a moderate-to-large effect of bracing (0.6) on medial tibiofemoral contact force. Assuming 80% power and alpha level of 0.05, and a correlation between measurements on the same individual of 0.875 [24], a sample of at least 13 participants was required. In the absence of previous estimates of effects of valgus knee bracing on medial tibiofemoral contact force using EMG-driven NMS models, we based effect size on meta-analysis reporting effect of valgus bracing on external knee adduction moment (standardized mean difference: -0.61) [1]. Statistical analyses were performed using Statistical Package for the Social Sciences (SPSS), version 24 (IBM, New York, USA). Shapiro-Wilk tests were used to examine data normality and conditions were compared using paired t-tests and Wilcoxon signed rank tests, as required ($p < 0.05$).

3. Results

Walking speed, stride length, and stride width did not differ between conditions (Table 1). Participants walked with significantly higher peak hip flexion, lower peak hip extension, and lower knee flexion when wearing the brace (Table 1). Participants walked with a significantly higher peak external knee flexion moment and lower peak external knee adduction moment when wearing the brace (Table 1). There were no other differences in kinematics or kinetics between brace and no-brace conditions (Table 1).

Peak medial and lateral tibiofemoral contact forces did not differ between brace and no-brace conditions (Table 1; Fig. 1). The first peak of total tibiofemoral contact force was 5% higher in the brace compared to no brace condition, and the second peak of the total tibiofemoral contact force was 5% lower in the brace compared to no brace condition (Table 1). Notably, there was substantial inter-participant variation in the magnitude of first peak medial tibiofemoral contact force between brace and no brace conditions (-6.1%–17.2% relative to the mean timing). Similarly, we found large variation in the change in magnitude of the peak external knee adduction moment (-19.6% to -1.2%) and external knee flexion moment (-14.8%–125%) (Fig. 2) between brace and no brace conditions. Polar plots (Fig. 3) depict the time-continuous coordination between medial tibiofemoral contact force and external knee adduction moment, external knee flexion moment, and knee flexion angle for both brace and no brace conditions.

Relative contributions of muscle and external loads to medial compartment loading did not differ between conditions (Fig. 4). However, mean relative contribution of muscle to lateral compartment loading was 2.35% (95% CI 0.46–4.24; $p = 0.02$) greater when wearing the brace compared to not wearing the brace. There were no significant differences in total muscle activation between brace and no brace conditions. When wearing the brace, participants exhibited significantly lower peak medial-to-lateral muscle co-activation ($p = 0.03$), and significantly higher peak extensor-to-flexor muscle co-activation ($p = 0.02$) during mid-late stance (Fig. 5).

4. Discussion

Knee loading during dynamic tasks is causally affected by activation of muscles spanning the joint [25]. However, there is a paucity of investigations into the concurrent effects of bracing on muscle activation patterns [26], muscle forces, and resulting tibiofemoral contact forces [8,9]. We overcame the limitations of external biomechanical measures by using an EMG-driven NMS model to estimate tibiofemoral contact forces and calculate the contributions of muscle and external loads to those contact forces. Given muscle activation patterns are altered in people with knee osteoarthritis [10], the aim of this mechanistic study was to evaluate immediate effects of valgus bracing on tibiofemoral contact forces and muscle activation patterns in pain-free individuals.

While wearing the valgus knee brace, participants walked with lower peak hip extension and knee flexion, but higher hip flexion, which together resulted in a higher peak knee flexion moment. Despite these observations and contrary to our hypothesis, medial tibiofemoral contact forces were not reduced when walking with the brace compared to without when we considered the cohort as a whole. Some individuals generated larger, similar, or smaller medial tibiofemoral contact forces while walking with the brace compared to no brace, irrespective of changes to their external knee flexion and adduction moments (Fig. 2). This variation is particularly evident in participants 7 and 8, who had similar changes in external knee flexion and adduction moments, yet experienced opposing changes in medial contact forces. This observation highlights that subject-specific activation patterns are critical when examining mechanistic responses to bracing, and that pain-free individuals without knee pathology appear to have the capacity to recruit lower-limb muscles to support or counteract brace-generated valgus knee torque. Furthermore, visual inspection of participant-specific responses to walking with the brace (Fig. 2) failed to identify a consistent role of frontal plane alignment in medial tibiofemoral joint contact force magnitude. However, study participants did not have substantial varus alignment, thus varus deformity cannot be dismissed as a potential contributor to the mechanism by which valgus bracing might affect tibiofemoral joint contact forces.

At the group level, the first and second peaks of the total tibiofemoral contact force were approximately 5% higher and lower, respectively, when walking wearing the brace compared to without. The elevated first peak of total tibiofemoral contact force may relate to greater activation of muscles at near-peak knee extension during stance, where their moment arms are largest, and thus favourable for torque generation [19], and their lines of action create tibiofemoral compression. Similarly, it is possible a lower second peak in total tibiofemoral contact force may have resulted from reduced muscle contraction. Although these differences approached statistical significance, we

Table 1
Spatiotemporal and kinematic variables for brace and no brace conditions (n = 16).

	Brace	No brace	Mean difference (95%CI)	p-value
Spatiotemporal				
Stride length (m)	1.52 (0.10)	1.52 (0.11)	−0.01 (-0.02, 0.01)	0.55
Step width (m)	0.07 (0.03)	0.07 (0.03)	0.00 (0.00, 0.01)	0.23
Walking speed ($m{\cdot}s^{-1}$)	1.43 (0.14)	1.45 (0.16)	−0.02 (-0.05, 0.01)	0.11
Hip joint kinematics (deg)				
Max flexion	33.0 (4.5)	31.7 (4.6)	**1.3 (0.1, 2.6)**	**0.04**
Max extension	12.1 (6.3)	13.9 (6.7)	**1.8 (0.7, 2.8)**	**< 0.01**
Sagittal plane excursion	45.2 (3.7)	45.6 (3.9)	−0.5 (-1.4, 0.4)	0.26†
Flexion at heel strike	32.2 (4.6)	30.6 (4.3)	**1.6 (0.5, 2.6)**	**0.01**
Knee joint kinematics (deg)				
Max flexion	65.5 (3.6)	67.6 (2.8)	**−2.1 (-3.0, -1.2)**	**< 0.01**
Sagittal plane excursion	70.8 (5.9)	71.2 (3.7)	−0.4 (-2.8, 1.9)	0.70
Flexion at heel strike	14.7 (4.4)	12.4 (3.7)	**2.4 (0.5, 4.3)**	**0.02**
Ankle joint kinematics (deg)				
Max dorsiflexion	15.2 (4.1)	14.5 (3.9)	0.6 (-0.8, 2.1)	0.37
Max plantarflexion	21.7 (5.7)	21.5 (4.6)	−0.2 (-2.0, 1.7)	0.85
Sagittal plane excursion	36.9 (5.3)	36.1 (5.1)	0.8 (-0.7, 2.3)	0.27
Plantarflexion at heel strike	0.0 (3.9)	0.1 (3.5)	0.1 (-1.3, 1.4)	0.96†
Knee moment ($Nm{\cdot}kg^{-1}$)				
Peak knee extension moment	0.40 (0.13)	0.45 (0.17)	0.05 (-0.01, 0.11)	0.09
Peak knee flexion moment	0.82 (0.21)	0.73 (0.25)	0.08 (0.02, 0.14)	**0.02**
Peak knee adduction moment	0.50 (0.20)	0.56 (0.23)	0.06 (0.03, 0.08)	**< 0.01**
Knee joint contact force (BW)				
Peak lateral joint contact force	1.23 (0.37)	1.19 (0.38)	0.04 (-0.04, 0.11)	0.30
First peak medial joint contact force	2.74 (0.70)	2.65 (0.79)	0.09 (-0.01, 0.19)	0.07
Second peak medial joint contact force	1.83 (0.45)	1.76 (0.49)	0.06 (-0.12, 0.23)	0.50
First total peak joint contact force	3.53 (0.73)	3.36 (0.82)	0.17 (0.00, 0.35)	0.05
Second total peak joint contact force	2.47 (0.65)	2.65 (0.20)	−0.18 (-0.35, 0.00)	0.05

Values are mean (standard deviation); bold indicates significance $p < 0.05$; †Wilcoxon signed tank test; deg – degree, max – maximum; BW – bodyweight.

question their practical relevance. Indeed, the differences in tibiofemoral contact forces observed (~0.18BW) are lower than the minimum detectable change previously estimated to be 0.66BW [24]. Future investigations should explore immediate and long-term effects of larger brace-generated valgus torques.

Interestingly, we observed greater group-level muscle contributions to lateral compartment loading when walking with the brace compared to without (Fig. 4). This occurred because wearing the brace reduced relative contributions of net external loads to lateral compartment loading that was countered by concomitant increases in muscle contributions to lateral compartment loading. However, lateral tibiofemoral contact forces remained unchanged between brace and no-brace conditions at the group-level. The effect of the brace on external loading to the lateral compartment is counter-intuitive, but may have occurred due to subtle changes to the external gait biomechanics made by participants to accommodate the brace, e.g., slight increases in step width or adjusting foot placement, which combined result in increased laterally directed muscle activation patterns. Independent of this apparent subject-specific response to valgus bracing, knee pathology patients often have knee muscle impairments [10], and preservation of knee muscles remains a valuable clinical target as muscle retention is associated with reduced loss of medial tibiofemoral cartilage [29]. Increasing physical demands on lateral knee muscles when walking with a valgus brace may elicit favourable physiologic adaptations, such as increased muscle volume, and could reduce the need for valgus bracing in this population over the long-term. The clinical relevance of greater relative muscle contributions to lateral tibiofemoral contact loading warrants investigation.

Our muscle co-contraction analyses suggest that the brace lowered peak medial-to-lateral muscle co-activation. Previous investigations reported no effect of valgus bracing on co-activation measures of knee muscles in healthy individuals [26]. Higher medial knee muscle activation has been previously reported in response to rapid valgus perturbation, suggesting that medial knee muscles can indeed respond to support the knee in the frontal plane independent of their role as knee extensors or flexors [30]. Nevertheless, the pure valgus perturbation

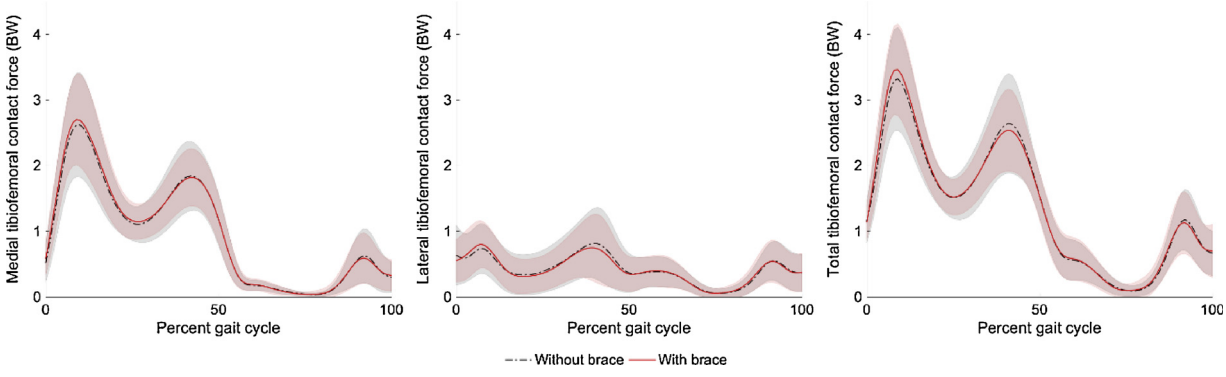

Fig. 1. Ensemble average (± standard deviation) medial compartment (left), lateral compartment (middle), and total (right) tibiofemoral contact forces normalized to body weight (BW) for brace (solid red) and no brace (dashed black) conditions over a gait cycle. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

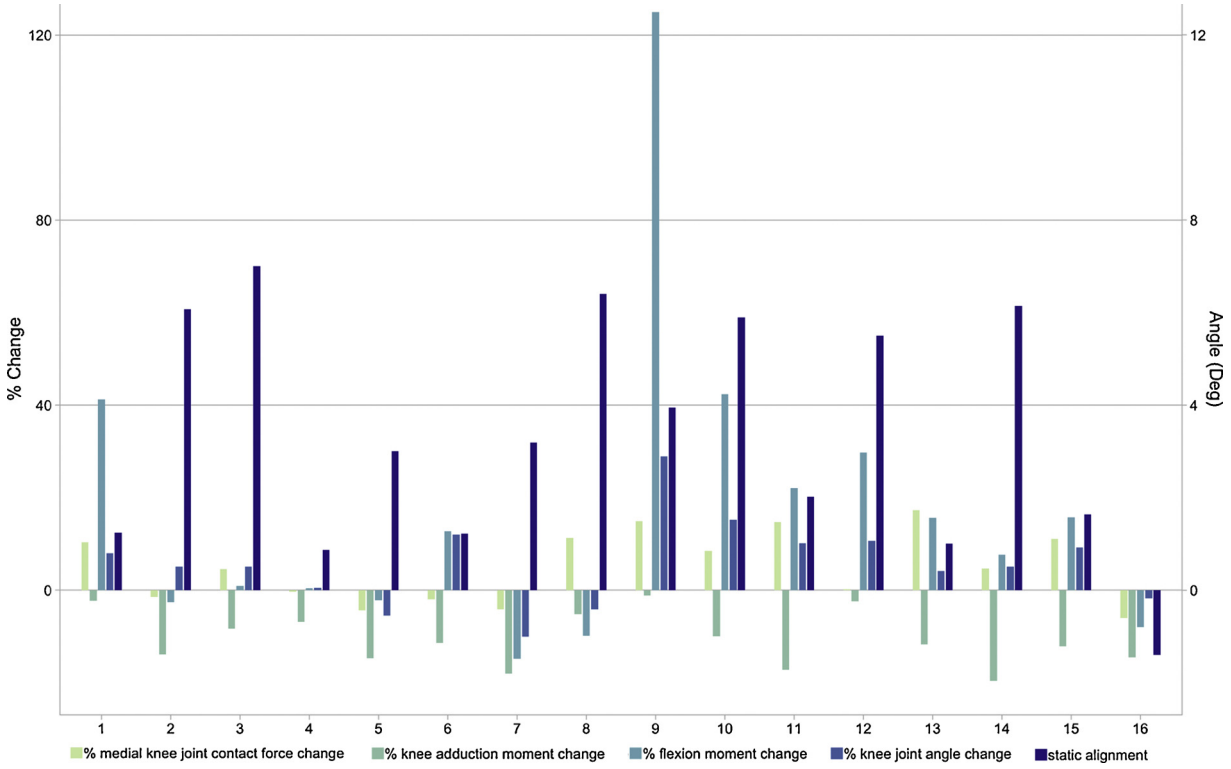

Fig. 2. Left hand axis: percentage change in first peak medial tibiofemoral joint contact force (light green), external knee flexion moment (green), external knee adduction moment (blue), and knee joint angle (dark blue). Right hand axis: surrogate measure of static frontal plane alignment (navy) at the time of first peak medial tibiofemoral joint contact force. Negative values indicate a lower peak medial tibiofemoral contact force (and other parameters) when walking wearing the brace compared to without. Positive values indicate a higher peak medial tibiofemoral contact forces (and other parameters) when walking wearing the brace compared to without. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

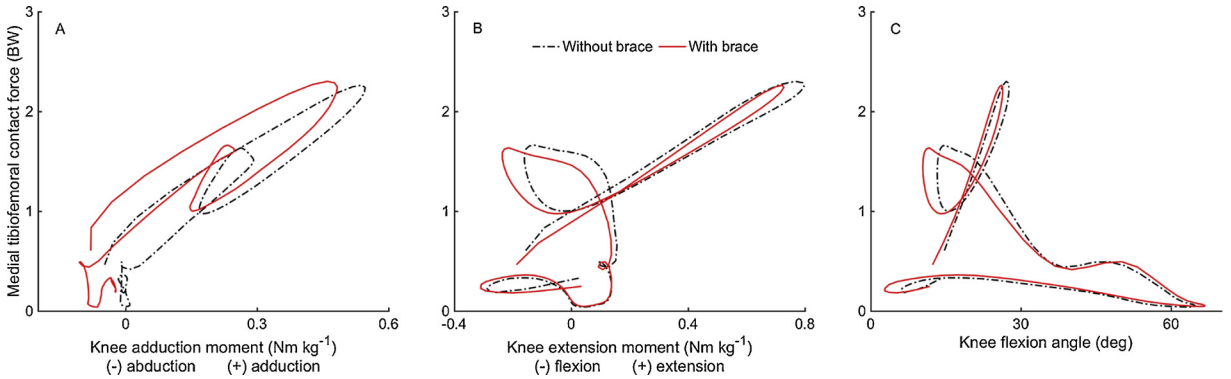

Fig. 3. Polar plots depicting the relationships between medial tibiofemoral contact force normalized to body weight (BW) and (A) external knee adduction moment ($Nm{\cdot}kg^{-1}$), (B) external knee extension moment ($Nm{\cdot}kg^{-1}$), and (C) knee flexion angle (deg) for brace (solid red) and no brace (dashed black) conditions. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

applied in this previous study used a seated open-chain mechanism under servomotor control, which inherently differs from the complex and dynamic knee loads produced during gait [30], and thus makes direct comparisons challenging. Although we observed lower medial-to-lateral muscle co-activation when walking wearing the brace compared to without, the effects on tibiofemoral contact forces were likely offset by higher extensor-to-flexor co-activation (Fig. 4).

The strengths of our study include a relatively homogenous sample of pain-free individuals, randomization of brace/no brace condition, and application of an EMG-driven NMS model to mechanistically examine immediate effects of valgus bracing on muscle activation and tibiofemoral contact forces. However, there are several limitations to consider. First, this was a preliminary study powered to detect moderate to large effect sizes. Second, assessment of the brace on healthy pain-free participants limits the external validity of the findings, and consequently its extensibility to people with knee pathology. We opted to assess pain-free individuals to reduce the influence of heterogeneous muscle activation patterns often reported in those with knee pathology [10], thus permitting a focus on the mechanisms by which bracing affects internal knee loading. However, the mechanism(s) by which bracing may affect tibiofemoral contact forces in those individuals with knee osteoarthritis or substantial varus alignment may differ from those determined in the present study. Second, brace fitting was conducted by several researchers (MH, DJS, GL, LED), and while one researcher (MH) was trained by the manufacturing company on brace-fitting, it is possible differences in brace-fitting occurred between researchers. Fourth, we assessed immediate effects of the valgus brace on measures of tibiofemoral contact forces and muscle activation patterns. It remains

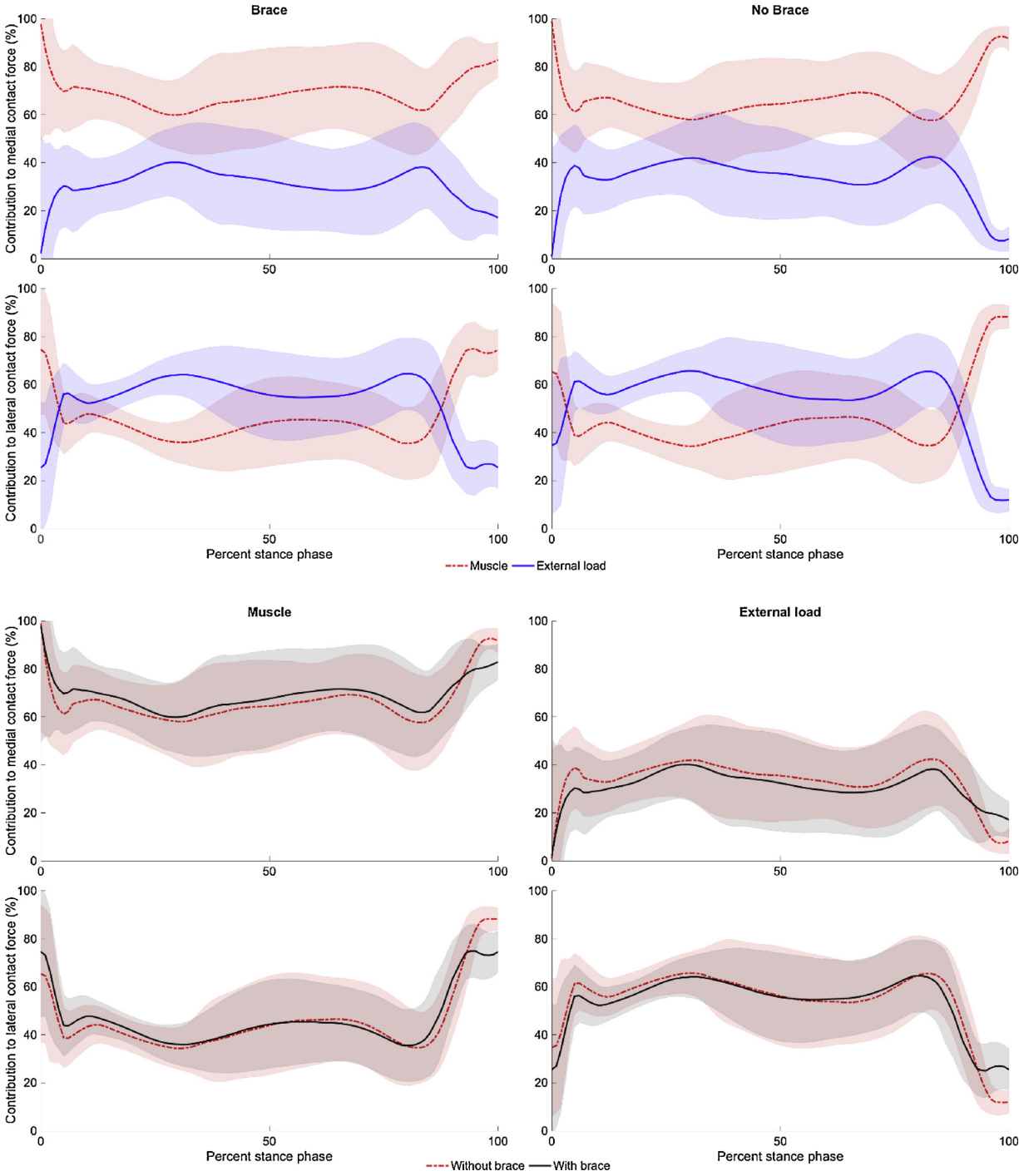

Fig. 4. Relative contributions (%) of muscle and external loads to medial (left) and lateral (right) compartment tibiofemoral contact force for brace and no brace conditions; $^{*}p < 0.05$.

unknown whether longer-term effects of wearing a valgus brace would alter muscle activation patterns, muscle forces, and tibiofemoral contact forces. Finally, validation of EMG-driven NMS models is hindered by the absence of methods to directly validate muscle force predictions and only limited data are available to validate tibiofemoral contact force predictions [22]. Nevertheless, EMG-driven NMS models are increasingly used to better understand knee pathologies [12] and have demonstrated excellent prediction of knee moments [11] and tibiofemoral contact forces measured using an instrumented knee [22].

5. Conclusion

Our preliminary findings suggest valgus knee bracing in pain-free individuals does not immediately alter tibiofemoral contact forces or relative contributions of knee muscles to those contact forces. Results suggest a greater relative contribution of muscle to medial compartment contact loading when wearing the brace compared to without. Further investigations are warranted to better understand longer-term effects of valgus knee bracing in people with medial knee osteoarthritis.

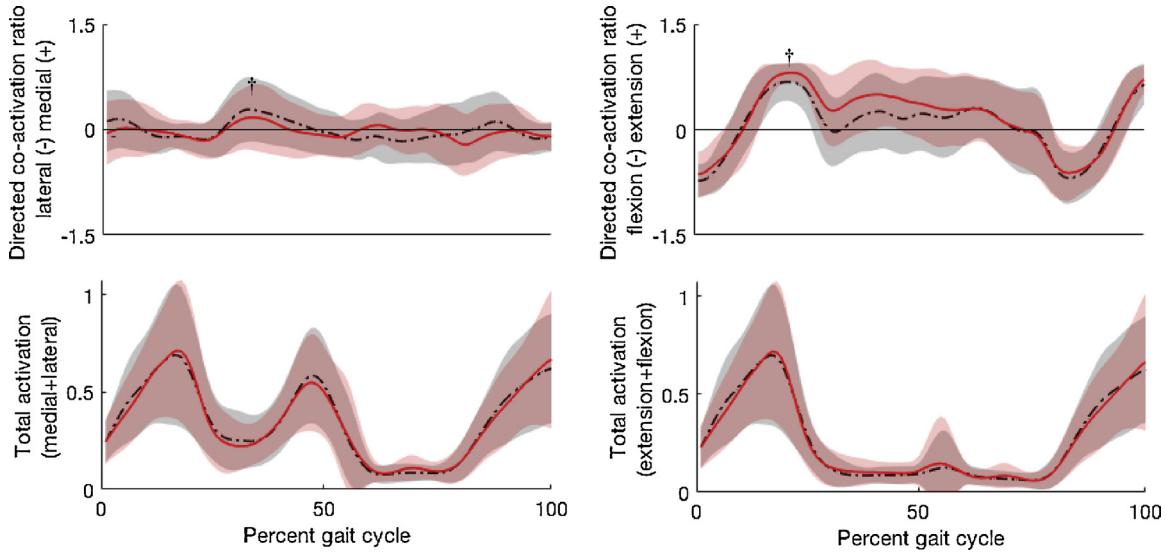

Fig. 5. Ensemble average (± standard deviation) directed co-activation (top) and total activation (bottom) patterns for brace (solid red) and no brace (dashed **black)** conditions over a gait cycle; †$p < 0.05$. Left: directed co-activation ratio = 0 indicates full co-activation between medial (vastus medialis, semimembranosus, medial gastrocnemius) and lateral (vastus lateralis, bicep femoris, lateral gastrocnemius) muscle groups. Right: directed co-activation ratio = 0 indicates full co-activation between extensor (vastus medialis, vastus lateralis, rectus femoris) and flexor (semimembranosus, bicep femoris) muscle groups. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

Author contributions

MH, LD, CP and DS conceived the idea for the paper. MH, LD, CP, GL and DS contributed to data acquisition and data analysis. MH, LD and DS wrote the first draft of the article. All authors provided scientific input and revised the paper. All authors approved the final version of the manuscript.

Conflict of interest

None to declare.

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